Deformation and initial stability in hip arthroplasty

Effect of neck geometry and fixation – an experimental cadaver study

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1 Contents

2 Abbreviations and terms

BMI	Body mass index
BMD	Bone mineral density
BW	Bodyweight
BWm	Bodyweight meter
CCD	Collum-caput diaphysis (angle)
CoCr	Cobalt-chromium alloy
DXA	Dual-energy X-ray absorptiometry
E	Modulus of elasticity, Young`s modulus (in pascal or newton/m^2)
ε	(Principal) strain
FE	Finite element
F	Force
HA	Hydroxyapatite
LMM	Linear mixed model
LVDT	Linear variable displacement transducers
PMMA	Poly (methyl methacrylate) (in bone cement)
RSA	Radiostereometric analysis
σ	Stress
THA	Total hip arthroplasty
TPM	Total point motion (micromotion)

3 Scientific environment

This project was performed at the Department of Orthopaedic Surgery, Stavanger University Hospital, while I was working there as a resident. It was a position funded for 50% research and 50% clinical work from 2009 to 2013. During the work on my thesis, I received academic supervision at the Department of Clinical Medicine, and the Department of Clinical Dentistry – Biomaterials at the University of Bergen. I also received supervision from the staff of the Department of Neuroscience and Orthopaedic Research Centre at Trondheim University Hospital. The biomechanical testing was performed at the Orthopaedic Research Centre of Trondheim University Hospital. All illustrations in this thesis are created originally for this work, unless otherwise stated.

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4 Abstract

Introduction: The search for and development of the optimal joint implant include preclinical testing. Restoration of the individual and natural biomechanics in the hip joint is a central goal in hip arthroplasty, and can be achieved by varying neck length, version and angle. Modular necks are one way to achieve these adjustments despite a growing concern regarding their outcome. In hip arthroplasty, the implants can be attached to the bone with or without cement. Both methods have achieved good clinical results. In this thesis, the effect of varying the femoral neck angle and length was tested in an experimental setup simulating everyday activities. Further, a cemented and an uncemented femoral stem with similar geometrical shape were compared in a preclinical setup.

Methods: All implants were tested in human cadaver femurs by loading in a hip simulator in single leg stance and stair climbing activity. Changes in deformation pattern of the proximal femur were measured by strain gauges. Initial stability of the femoral stems was investigated using a micromotion jig. The effect on the deformation pattern and initial stability was studied when the neck version, angle and length were varied, due to either an eccentric femoral head or a modular neck. The deformation pattern and initial stability of a cemented and an uncemented stem of similar geometry were compared.

Results: Strain was reduced in the proximal femur for all implants tested, especially proximally on the medial side, compared to the intact femur. Increased offset combined with retroversion or reduced neck–shaft angle in an eccentric femoral head gave significantly increased strain values compared to the standard situation. All three eccentric femoral head configurations gave overall small micromotion of the femoral stem; up to 40 µm.

When testing the modular necks, the varus neck increased the micromotion up to 60 µm. Micromotion was significantly higher during stair climbing compared to single leg loading, and for distal level compared to proximal level in all modular necks.

The short neck had higher loss of strain in distal position on the lateral side, and the retroverted neck retained more strain proximal medially.

The cemented stem had slightly higher strains than the uncemented stem on the medial side, while uncemented stem had higher strains on the lateral side of the proximal femur. The differences were small, but statistically significant.

Conclusion: Varying the femoral neck version, angle and length by either an eccentric femoral head or a modular neck gave some variations in cortical strains in the proximal femur compared to a standard design. However, the differences might be too small to have any clinical significance. The initial stability was acceptable for the tested implants when varying the femoral neck angle and length.

The cemented stem was more stable than the uncemented stem, as expected. However, both stems had small micromotions at the bone-implant interface, and in a range, that is not expected to have a negative impact on long-term stability.

5 Acknowledgements

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6 List of publications

This thesis is based on the following papers, referred to by Roman numerals:

I: Wik TS, Enoksen C, Klaksvik J, Østbyhaug PO, Foss OA, Ludvigsen J, Aamodt A. In vitro testing of the deformation pattern and initial stability of a cementless stem coupled to an experimental femoral head, with increased offset and altered neck angles. Proc Inst Mech Eng, Part H: Engineering in medicine 2011; 225(8):797-808.

II: Enoksen CH, Gjerdet NR, Klaksvik J, Arthursson AJ, Schnell-Husby O, Wik TS. Initial stability of an uncemented femoral stem with modular necks. An experimental study in human cadaver femurs. Clin Biomech (Bristol, Avon). 2014; 29(3):330-5.

III: Enoksen CH, Gjerdet NR, Klaksvik J, Arthursson AJ, Schnell-Husby O, Wik TS. Deformation pattern and load transfer of an uncemented femoral stem with modular necks. An experimental study in human cadaver femurs. Clin Biomech (Bristol, Avon). 2016 Feb; 32:28-33.

IV: Enoksen CH*, Wik TS*, Gjerdet NR, Klaksvik J, Arthursson AJ, Schnell-Husby O. Load transfer in the proximal femur and primary stability of a cemented and uncemented femoral stem. An experimental study on cadaver femurs. (submitted to Proc Inst Mech Eng, Part H: Engineering in medicine, **joint first authors*)

7 Introduction

7.1 General background

Total hip arthroplasty (THA) has been used for several decades to treat destructive conditions of the hip joint. THAs are load-carrying constructions, dependent on proper fixation and primary stability to achieve long-time survival. The mean incidence of THA in industrialised countries is estimated to be 156.6 per 100 000 habitants, and the incidence is increasing, based on recent OECD and EU reports (1, 2). THA has been referred to as “the operation of the century” (3), and is believed to be a cost-effective treatment in patients suffering from osteoarthritis and degenerative conditions (4, 5). The number of primary THAs in Norway is over 8000, and additionally 1300 revisions are performed every year (6). The 15-year survival of THAs in the Nordic countries is reported from 84-88% (7). This warrant, the search for implants with improved survival and function.

7.1.1 Brief history of THA

Surgery for hip arthritis goes back to the 19th century. One of the first attempts to treat severe arthritis in the hip joint was made by John Rhea Barton, performing osteotomy in an ankylosed hip around 1826 (8). Later, in the middle of the 19th century, Léopold Ollier and John Benjamin Murphy combined osteotomy in the proximal femur with a soft tissue procedure, forming a new hip joint (9).

Themistocles Glück performed the first hip replacement in 1891, using an ivory ball and socket fixed to the bone with nickel-plated screws, and provided fixation through a mixture of plaster of Paris and powdered pumice (10).

In the mid 20th century the Norwegian-born orthopaedic surgeon Marius Smith-Petersen performed synthetic interpositional arthroplasty. The first implants had a breakage problem. Later he developed implants using cobalt-chromium alloy (CoCr) with more success (11, 12). Some of these implants showed good longevity, and patients with Smith-Petersen devices can still be seen in clinic, 50 years after insertion (13).

Devices more similar to present prostheses were developed by Harold R. Böhlman in 1939, using a CoCr ball fitted to a nail (14). In the late 1940s, Jean and Robert Judet used an acrylic endoprosthesis, which subsequently was made from CoCr (15). In 1940, Austin Moore was the first surgeon to replace hips with a metal prosthesis, and in 1952, Moore described an implant that allowed bone ingrowth. These implants were the first femoral stems to be commercialized (16).

Philip Wiles described the first ball and socket implants (THA) that were introduced in 1948, but they failed mechanically after a short time (17). This first THA was improved by Kenneth McKee (18), but still failed due to loosening and mechanical complications.

What could be termed a paradigm shift in the development of THA was initiated by Sir John Charnley, considered to be the founder of modern hip arthroplasty. He introduced low friction joints and acrylic cement fixation in 1958, and reported his first methodological experience two years later (19). The cement consisted of polymethylmethacrylate (PMMA). The cement and the Charnley hip system turned out to be a success worldwide, with acceptable long-term outcomes (20-25).

7.1.2 THA of today

The main objective of the THA operation is to achieve pain relief and optimal functioning of the hip joint. Today, there is a multitude of implants, commercially available from many manufacturers.

Prosthesis should fit anatomically and maintain mechanical fixation under dynamic loading. The implant should offer an acceptable range of motion in the joint and provide the required stability (26).

The modern THA typically consists of a femoral stem, a femoral head and a cup replacing the acetabulum (27) (Figure 1). The femoral neck is usually a fixed part of the femoral stem. The femoral stem may also be designed with a modular femoral neck, which allows a variety of angulations, lengths and offset of the femoral neck part as described in a mid-term follow-up (28).

The joint surfaces consist of the femoral head, typically made of metal alloy or ceramics, and the acetabular liner consisting of cross-linked polyethylene or ceramics (27) (Figure 1).

The femoral and the acetabular components depend on secure fixation to the femoral bone and acetabular socket, respectively. THA fixation methods are basically divided into two main groups: cemented, using a self-setting acrylic cement as fixation component, and uncemented, also termed cementless (27), where the implant by press-fit technique is adapted directly to the bone.

The femoral component is the main objective in this thesis (Figure 1).

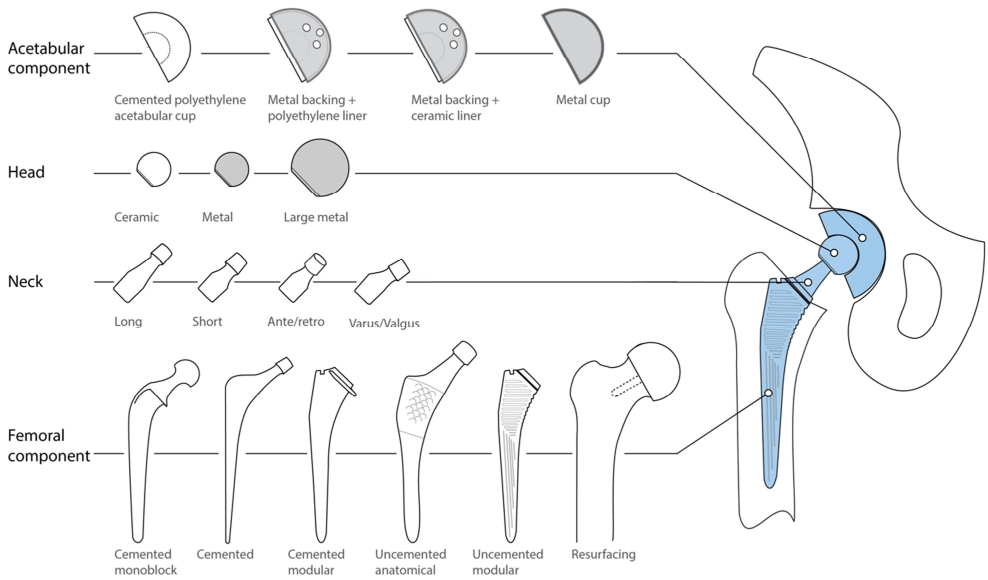

Figure 1. Different components in total hip arthroplasty.

7.2 Basic biomechanics

7.2.1 Bone

Bone is a living and dynamic tissue. Bone contains cells embedded in bone matrix (osteoid). The bone cells include osteoblasts (bone-forming cells), osteoclasts (bone-resorbing cells), osteocytes (bone- maintaining cells), and bone lining cells (29, 30). Osteoblasts produce organic matrix (osteoid) and regulate deposition of bone minerals, to form inorganic matrix (hydroxyapatite). Osteocytes are cells that differentiate from osteoblasts when trapped in bone matrix by secretion (30). These cells are assumed to be mechanosensory cells in bone, maybe together with the lining cells (31). The remodeling of bone is believed to occur under the action of a basic multicellular unit containing osteoblasts, osteocytes and osteoclasts together in process affecting a large number of regulatory actions (29).

The femur diaphysis has a surface of compact cortex, where the osteons are running mainly parallel to its long axis. The inner more porous core, referred to as spongiosa, consists of cancellous or trabecular bone (30). The characteristic trabecula found in the proximal femur is an example of bone-modelling reflecting the forces and loading of the hip. The orientation and density of the trabecular bone indicates the direction and magnitude of the forces acting on the proximal femur (30) (Figure 2).

There is a wide range of anatomical shapes of the proximal femur, and this has impact on the choice of stem design in THA surgery.

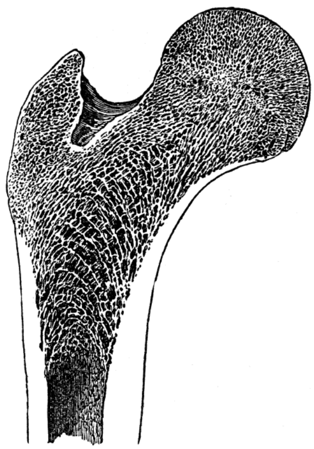

Figure 2. The trabecula in the proximal femur demonstrated in a CT scan. (License from Florida Center for Instructional Technology).

7.2.2 Hip joint

The hip joint consists of an articulation between the femur and pelvis. It is a ball and socket joint with three degrees of freedom. The hip is stable joint, due to an acetabulum, covering a sector of approximately 170° of the femoral caput (32). The femur is a long bone, exposed to high axial loading and muscle forces. In a two-leg stance situation, the pelvis is balanced over the hip joints with small contribution from the surrounding muscles (33). In single leg stance, the gravity axis shifts medially to the hip joint centre, due to the additional weight of the non-weight bearing leg. The large lever arm of the gravitational force generates a considerable moment about the hip joint, and the abductor muscles must compensate by increasing their action in order to maintain torque equilibrium (Figure 3). The resultant force in the hip joint is therefore higher in single leg stance than in two-leg stance. The main developing forces come from the abductors (34) and to some extent the tensor fasciae latae muscle and iliotibial band (ITB) (35) (Figure 3).

Telemetric studies have shown that the resultant force in the hip joint increases to 2-3 times bodyweight (BW) during walking (36, 37). The abductor forces in the hip have been measured at 1-2 BW (34, 38). During stair climbing, a torsional force is added related to flexion in the hip joint. Bergmann showed that this additional torque increased to a torsional moment of 2.24 % bodyweight-meter (BWm) (37). It has also been shown that a torsional force could affect the implants and subsequently lead to mechanical failures and loosening (39). These biomechanical considerations make it essential that uncemented implants are designed to achieve optimal primary stability immediately after insertion.

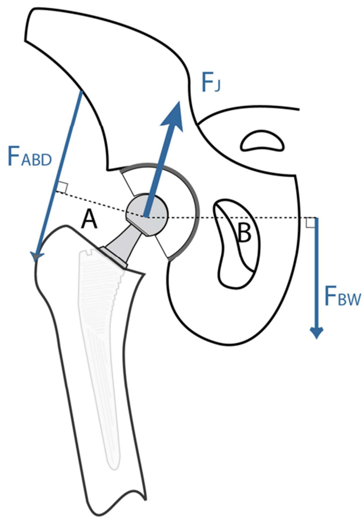

Figure 3. Typical forces in the hip joint, single leg stance. Arrows represent the abductor resultant force (F_{ABD}), bodyweight force (F_{BW}) and the hip joint reaction force (F_J), A represents the lever arm of the abductor forces. B represents the leverarm of the bodyweight.

7.2.3 Deformation of the proximal femur

Strain (ε) is the relative deformation of an object, expressed as relative change in dimension (30). Strain in human bones is an effect of force application. In the proximal femur, typically compression occurs on the medial side and extension on the lateral side (40). Strain is dimensionless, and can be presented as a percent value. Due to the magnitude in bone the strain is often given as microstrain, i.e. 10^{-6} m/m. Tensile strain is denominated as positive, and compressive strain as negative (30) (Figure 4).

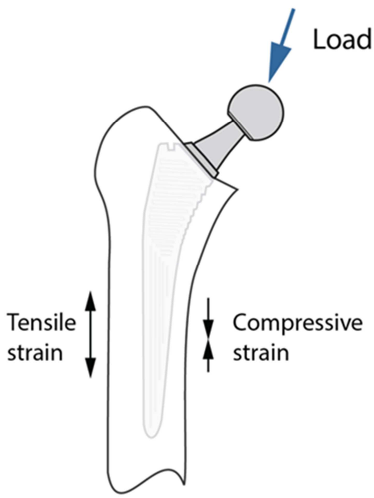

Figure 4. Loaded femur with compressive and tensile strain.

Stress (σ) is the force per unit area, the ratio of a load is applied to cross section area. Engineering materials, and also bone, exhibit linear-elastic behavior (30). The classic load-deformation curve can be transformed to a stress-strain curve (30, 41) (Figure 5).

When strain and stress are low, the relationship is proportional and deformation is elastic. When stress increases, the material reaches a yield point where plastic deformation occurs. At even higher stresses the material fails (Figure 5). All physiological stresses are well within the elastic region (30). The stress in bone during loading can be measured through cortical deformation pattern. The most common method is by use of electrical resistance strain gauges (42).

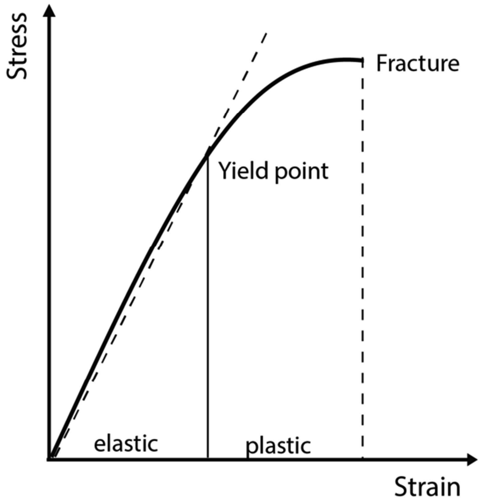

Figure 5. A schematic stress–strain curve of bone.

The relationship between stress and strain in the initial, nearly linear portion of the curve is termed, *modulus of elasticity* (E) (Young`s modulus) (30, 43). Materials have highly different elastic moduli. For example, titanium alloys have an E of 55–105 GPa and cobalt and iron based alloys approximately 200-230 GPa (44). Acrylic bone cement (PMMA) has a Young`s modulus around 2-3 GPa (45, 46).

The elastic modulus of human cortical bone can vary from 10 to 25 GPa, and human cancellous bone from 1 to 20 GPa, dependent on the localization in the cross section of the human bone measured (30, 44, 47, 48).

The *stiffness or rigidity* of a structure is its ability to resist deformation and is influenced by the elastic modulus (E) of the material involved and the geometry of the construct over which the force is acting (moment of inertia) (30).

7.2.4 Bone response to loading

The effect of mechanical factors on bone response is complex. Although the precise mechanisms of the cellular control still is partly unclear, the dynamic strain is considered important for the remodeling of bone (49, 50). This phenomenon is laid down in the so-called Wolff`s law, which deals with the relationship between a mechanical load and adaptive remodeling of the trabeculae within the bone (51). This relationship has later been investigated and discussed (29, 49, 52) (Figure 6).

Bone cells seem to use their functional strain surroundings directly and indirectly, in order to avoid fracture under deviant loading conditions (49). In daily life, high impact activities with versatile movements are more osteogenic than activities with conditions like swimming and cycling (53-55).

7.2.5 The concept of stress shielding

After implantation of a stiffer implant into a less stiff material such as bone, loads will be transferred through the stiffer object. In the proximal femur implanted with a femoral stem, forces will partly bypass the proximal bone and may be associated with progressive bone resorption in this area. The clinically observed bone resorption is also called the “stress shielding” phenomenon (56-61) (Figure 6). Ideally, new implant designs in THA should be designed to maintain a distribution of physiological loads in the proximal femur (60).

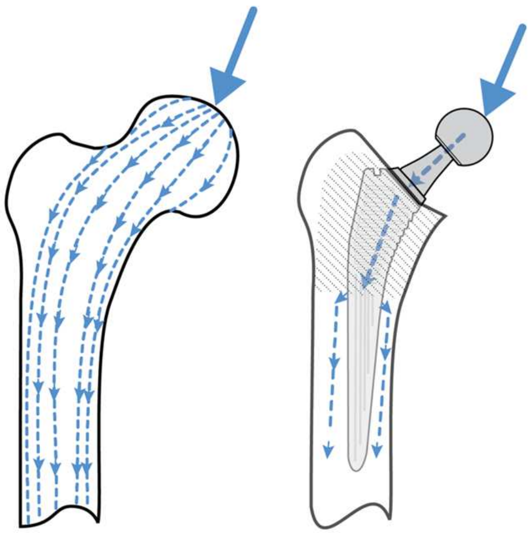

Figure 6. Load transfer in proximal intact femur (left). Load transfer in the implanted femur, where the proximal part is "bypassed" through the stiffer stem, leading to "stress-shielding" (right).

Adverse (unwanted) clinical consequences of proximal bone loss might be periprosthetic and trochanteric fractures, loss of adequate bone for revision of implants and increased exposure of the proximal femur to implant wear debris (57-59). Despite this observed bone resorption, it has been difficult to document increased risk of fractures and aseptic loosening in clinical series (59, 60). There is probably a multifactorial etiology, affecting the stress shielding in a bone-implant situation. Dual-energy x-ray absorptiometry (DXA) is often used to quantify adaptive remodeling around hip stems (62-64). Increasing stiffness of the implant, increases the stress shielding phenomenon (57, 65, 66). It is also known that the extent of coating influences the bone resorption as the resorption tends to appear in coated areas of the femoral stem (60, 67-69). Finally, it has been shown that preoperative BMD influences the extent of periprosthetic bone loss, where patients with low BMD have larger bone loss compared to those with higher BMD (62, 63, 70-74).

It seems important to preserve the bone stock and avoid a high level of stress shielding following THA to gain long-term stability (60).

7.3 Fixation of femoral stems

There are basically two fixation methods in THA, cemented and uncemented. Uncemented implants achieve long-time fixation by bony ingrowth and ongrowth to the surface layer of the implant, while the cement acts like a sealant to the bone in cemented fixations (27).

Uncemented implants are dependent on a substrate to secure the biologic fixation through bony ingrowth. Uncemented stems have a rough porous surface with or without a hydroxyapatite (HA) coating to stimulate bone growth. Implant design differ in coated areas: they may be fully or proximal coated, and the coating may be totally or partly circumferential. There exists a great variety of designs of the uncemented femoral stems, but they all have press-fit design to achieve optimal primary stability in the supportive bone (2, 43).

Cemented femoral stems may contain either a highly-polished surface, or a matte or grit-blasted surface. The bone cement acts as a filler between the implant and surrounding bone. The polished finish is designed to reduce friction between the cement and implant, and to reduce potential third body wear. The highly-polished stems are often collarless and are designed to allow controlled subsidence of the stem within the cement mantle.

The cemented and uncemented stems have unlike requirements regarding stiffness. Uncemented stems are frequently made of Ti-alloys with a lower E-modulus to reduce the stiffness and thereby the stress-shielding. While cemented stems often are made of CoCr alloy or stainless steel, featuring some higher E-module to reduce the stress transmitted to the cement avoiding micro-cracks (2).

It is common to combine a cemented and an uncemented component in THA, termed hybrid fixation. Both cemented and uncemented implants show excellent overall long-term survival (6, 75-77)

The choice of fixation method, in addition to published results on functional outcome and longevity, is often based on orthopaedic traditions and experience, and varies between countries and regions. Usually, both fixation principles coexist in clinical use for different subgroups of patients and indications.

Although there are many studies evaluating cemented and uncemented THA, the large number of implant design complicates comparison of the fixation methods. There are a few reviews comparing cemented and uncemented THA; these conclude that cemented stems perform better than uncemented stems (78-81).

7.4 Initial stability of uncemented stems

Uncemented femoral stems are dependent on initial primary stability in the first postoperative phase, to achieve bony fixation and long-term fixation (82-84). Uncemented prostheses achieve secondary fixation to the surface layer of the implant, by osseointegration. Excessive interface motion may inhibit bone ingrowth and in some cases lead to complications like early loosening of an implant (43).

The osseointegration process can be compared to primary fracture healing, and the ingrowth of bone at the implant surface occurs in three stages. The first phase is the initial inflammatory stadium. Second the reparative of woven bone takes place and in the third phase a remodeling of lamellar bone develops (85, 86). The theoretical basis of the ingrowth is shown in two studies and this process is assumed to last from four to twelve weeks and up to three years after implantation (87, 88).

The bone ingrowth and initial stability of an implant is dependent on factors related to both implant, the surgical procedure and the quality of the patient`s bone.

The implants are dependent on designs that secure the initial stability, so that rotational forces and initial sinking are avoided (89). However, the shape and geometry of the femoral stems vary to a great extent. A good apposition of the implant for osseous contact is therefore an important factor (90). Frequently the shape of the stems includes edges and grooves to mechanically improve the initial stability. The surface roughness of an implant affects the bone-implant contact and further the initial stability (2). The porous-coated stems have shown higher coefficients of friction than smooth stems (91). The optimal pore size of the porous coated surface is recommended to be between 50 and 400 µm (92-94). The coating of the femoral stem, often with a bioactive calcium phosphate such as hydroxyapatite (HA), is intended to facilitate the integration into the surrounding bone tissue and work as a chemical bonding.

The objective of the preoperative and surgical intervention is to achieve a press fit between the bone and implant for good primary stability (2). This requires adequate preoperative planning and operative technique including reaming and choice of implant (89).

Patient related factors are also essential for the initial stability of an implant and survival. Gender, age, BMI and activity level are contributes affecting the clinical outcome. The quality of bone in the proximal femur matters for the choice of uncemented implants and their initial stability (89).

Migration and micromotion are preclinical terms that express any movement of implant related to bone during physiological loading. Migration is used to describe permanent displacement of the femoral stem into the femoral canal, occurring during the first postoperative period (30, 95). Micromotion expresses a reversible motion at the bone-implant interface, and occurs while an implant is dynamically loaded (30).

Micromotion can be estimated by numerical analysis and by in vitro methods (30, 95-101). In the laboratory, micromotion of a femoral implant is usually tested in a cadaver or a synthetic femur (30). The implant-femoral movement is typically tested in a loading devices simulating a controlled hip load scenario (hip simulator). In vitro, the micromotion can be measured indirectly or directly, using extensometers or optoelectronic devices (30).

Experimental studies have shown that excessive micromotion can inhibit the biological integration of bone at the implant surface (82, 84, 102, 103). The exact range of micromotion that will allow osseointegration is not known and several studies have tried to approach this topic with various scientific methods (82, 84, 104, 105).

Cemented stems and initial stability are explained in chapter 7.3.

7.5 Modularity of femoral stems

Femoral stems with modular necks have been used in revision surgery for the last three decades (106), and have also been more recently applied in primary THA (107).

It is important to restore the natural biomechanics of the hip joint (108-111). Modular necks were introduced to primary THA with the intention to allow correction of leg length, offset and instability. Modularity in the femoral neck is achieved by an additional junction between the neck and the stem (Figure 7).

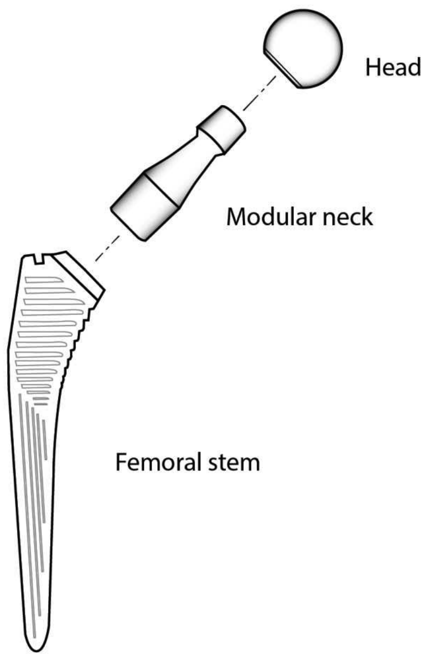

Figure 7. Modularity in a femoral stem.

There are some advantages favoring the modular necks. In preoperative planning, the different modular neck templates may help to restore variations in femoral anatomy, femoral neck length, shaft diameter and the collum-caput diaphysis angle (CCD) (112). The opportunity to adjust the offset and version plays a role in preventing impingement between the soft tissue and instrumental parts (108-111). One study has also reported that modular neck stems could improve the range of motion in the hip-joint (113).

There are some reports of good mid-term outcomes for modular necks (28, 113-115), but there is limited long-term documentation.

Experimental studies warn against fretting and corrosion regarding modular necks in THA (116, 117). Concerns were further raised regarding modular necks in primary THA in several case reports (118-122). Gill introduced the designation of pseudotumor formation as a result of corrosion at the neck stem junction leading to revision surgery (123). In 2010, the Australian Joint Registry (AOANJRR) addressed this issue (124). In the AOANJRR´s report from 2015, THAs with exchangeable femoral necks still have nearly twice the rate of revision compared to conventional THA after 10 years. Implant loosening and dislocation are the main reasons (107).

8 Study aims

8.1 General aims of the study

The overall hypothesis of the present study was to evaluate, in an in vitro cadaver model, whether a femoral stem with a certain modular femoral head and neck system, and also two different fixation methods, presents biomechanical advantages.

The main research questions were:

Will varying the geometry of modular necks or modular heads affect the strain pattern of the femoral bone and the initial stability?

Are there any differences in strain pattern and micromotion between an uncemented femoral stem and a cemented stem with similar geometric design?

8.2 Specific aims of the study

I: To study the changes in the proximal femoral strain and micromotion pattern of an uncemented femoral stem with a femoral head with increased offset, altered neck version and femoral neck-shaft angle.

II: To study the primary stability of an uncemented femoral stem with four different modular necks, varying version, length and neck-shaft angle.

III: To study the load transfer expressed by the cortical deformation pattern of an uncemented femoral stem with four different modular necks, varying neck-version, neck-length and neck-shaft angle.

IV: To study the initial stability and the cortical deformation pattern in the proximal femur between two stems with identical geometrical shape, but with different fixation.

9 Material and methods

9.1 General

In general, pilot studies were completed initially, to develop a reliable structure and algorithm in the test set-up. The implementation and use of methods was performed according to an established procedure (125-127). All implantations at the biomechanical laboratory were performed by an experienced orthopaedic surgeon according to the manufacturer's procedures (128). The testing and follow up were supervised by a skilled engineer at the laboratory.

9.2 Implants

9.2.1 Paper I

In Paper I, a straight, uncemented, collarless femoral stem in titanium alloy, with a slightly ribbed porous coating (Summit^{TM} high offset, DePuy International Ltd, Leeds, UK), combined with an experimental CoCr head of 47 mm (ASR^{TM}XL Anatomic Head System, DePuy International Ltd, Leeds, UK) was used. The experimental head consisted of an inner sleeve and an outer spherical part, allowing for eccentric displacement of the head on the entry of the femoral neck. Two positions of the experimental head were tested. Position 1 corresponded to 6° retroversion in the neck axis, where the taper was maximally displaced in anterior direction into the femoral head. Position 2 represented a reduction of the neck shaft-angle from 130° to 124°, where the taper was maximally displaced in superior direction (Figure 8). As a control, a standard 32 mm head in CoCr alloy was used. The experimental head had an increased neck length of 10.5 mm compared to the standard head, due to an extended inner sleeve. Three configurations were tested: standard, position 1 and position 2.

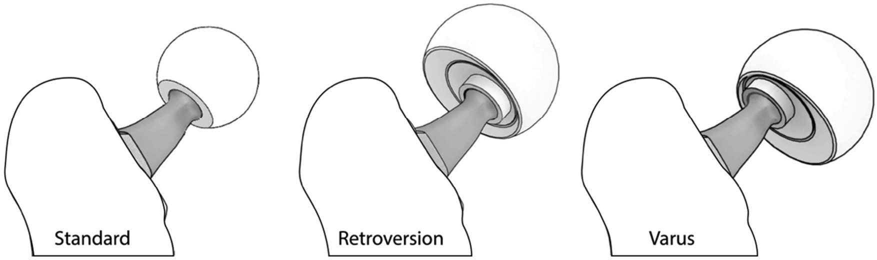

Figure 8. Illustration of the three configurations in Paper I: Standard, anterior displacement (position 1) and superior displacement (position 2) of the taper.

9.2.2 Papers II-III

Papers II and III, an uncemented collarless titanium alloy, fully coated with hydroxyapatite (HA) (Profemur® PRGLKITD Gladiator, Wright Medical Technology Inc., Arlington, TN USA 38002), was used combined with modular necks. Four modular titanium necks with different geometry and a 12/14 taper (Profemur® Modular Necks, Wright Medical Technology Inc., Arlington, TN USA 38002) were evaluated: 1. Straight long (PHAO 1204), 2. Straight short (PHAO 1202), 3. Retroversion short 15° (PHAO 1262) and 4. Varus short 15° (PHAO 1242) (Figure 9). The necks were connected with the oval end of the appropriate femoral neck implant into the femoral stem pocket. A standard 28 mm femoral head was used. The stems were randomly allocated to right or left femur before surgery.

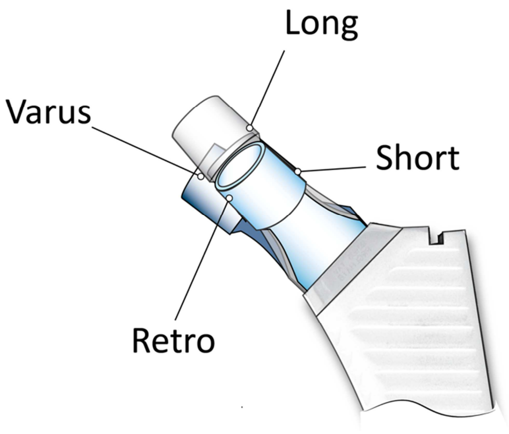

Figure 9. Modular necks in Papers II and III: long, short, retroverted and varus.

9.2.3 Paper IV

In Paper IV, the same uncemented stem as used in Papers II and III was compared to a cemented cobalt-chromium collared stem (Profemur® PRGLKITA Gladiator, Wright Medical Technology Inc., Arlington, TN USA 38002). These two stems (Figure 10) had similar geometry. The cemented stem had a light grit-blasted texture and a distal centralizer was used for cementation. A Methacrylate-based cement, with a mixing system (Cemex® Genta ID Green system 13A2420) (Tecres Medical, Verona, Italy), was used for implantation of the cemented stem. Both stems were tested with a short straight neck, and a 28mm caput (Gladiator, Wright Medical Technology Inc., Arlington, TN USA 38002).

Figure 10. The uncemented (left) and cemented (right) femoral stems.

9.3 Bone specimens

The femoral stems were implanted into Caucasian human cadaver bones. The femurs were collected from deceased patients who underwent planned medical post-mortem examinations. The femurs were collected within 24 hours at the departments of pathology of the university hospitals in Stavanger and Trondheim. Relatives had given consents before collection. Gender, height and bodyweight (BW) were obtained during autopsy, and an individual body mass index (BMI) was calculated for each subject.

The study was approved by the Regional Committee for Medical and Health Research Ethics (REK Vest 2009/359-CAG, Biobankregisteret Ref 2667).

Thirty-two pairs of femora were gathered during the period of collecting. Four subjects were excluded due to osteoporosis; one pair of femurs became damaged during preparation and five femurs failed during testing. Twenty-two single human cadaver femurs were included and tested. Mean donor age was 58 years (range 43–71 years) and sixteen male and six females (Table 1).

Table 1: Data of the subjects.

Paper	ID	Side	Gender	Age (years)	BMD (g/cm^2)	BodyWeight (kg)	Implant Size
I	1	L	M	61	1.029	.	7
	2	R	F	59	0.949	.	6
	3	R	F	65	0.849	.	5
	4	R	M	46	0.949	.	7
	5	R	M	64	0.853	.	8
	6	L	M	61	0.911	.	7
	7	L	M	60	1.080	.	6
	8	L	M	47	1.124	.	7
	9	L	M	71	1.002	.	7
	10	R	F	44	0.796	.	6
II-IV	3	R	M	59	0.943	60	4
	4	R	M	57	1.163	82	5
	7	L	M	66	0.963	90	7
	12	R	M	70	1.063	78	9
	14	L	F	53	0.959	55	3
	15	L	F	57	0.998	66	4
	18	L	F	62	0.896	58	5
	19	L	M	64	0.891	80	6
	20	R	M	53	0.894	71	9
	21	R	M	67	0.940	79	8
	22	L	M	47	0.962	79	7
	23	L	M	61	0.942	54	4

The femurs were handled and prepared according to a previously described and documented procedure (125-127). The femurs were wrapped in saline-soaked towels and stored at –80°C immediately after dissection. Standard radiographs (Philips Digital Diagnost) in two projections were used to estimate the size of the prosthesis and to exclude any skeletal pathologies. Dual-energy X-ray absorptiometry (DXA) (Paper I: Hologic Discovery A, Bedford, USA.) (Papers II-IV: GE Lunar Prodigy

Advance, USA) was obtained to indicate possible osteoporotic femurs. Bones with T-scores of the proximal femur below -2.5 were classified as osteoporotic and excluded.

The selection criteria of femurs included age <75 years in study I, later age <70 in Papers II - IV, no previous fracture in the femur and no current or previous malignancy in the femur. In Papers II-IV individual loading of the specimen was performed. Subjects with BMI in the range of 18–30 were accepted for the study (Table 1).

Frozen femora were thawed at room temperature and remaining soft tissue removed before testing. The frontal plane of the femur was first defined by placing the femur on a horizontal surface resting on the posterior condyles and the greater trochanter. Further, the anteversion of the femoral neck was measured and recorded for later orientation of the femur in the frontal and sagittal planes, before resecting the condyles. The femur was next fixed into a steel cylinder with an acrylic cement (Meliodent, Heraeus Kulzer GmbH, Hanau, Germany), where the central axis of the femur was preserved.

The proximal femur including the first 25 cm from the tip of the greater trochanter to the top of the cylinder was kept over the cylinder. A 40mm polyamide strap, attached to the greater trochanter with glue (X60, HBM GmbH, Darmstadt, Germany) and 6 screws (cortical 2.5mm) simulated the hip abductor muscles.

9.4 Biomechanical test setup

9.4.1 Hip jig – Paper I

In Paper I, the femurs were placed in a custom made hip jig, and loaded in a material testing machine (MTS 858 MiniBionix II, MTS System Corporation, Eden Prairie, Minnesota). (Figure 9). This first setup included a constant torsional moment and an iliotibial band (ITB). The femur was allowed to rotate freely around its longitudinal axis and to tilt freely in the medial/lateral plane, avoiding unphysiological bending moments. The femur was tilted and positioned 12˚ into valgus, corresponding to the physiological inclination during single leg stance (34). For all the experiments an acetabular cup with an inclination of 45˚ and 0˚ anteversion was used. A trochanter strap was fixed to the lever arm at an angle of 15˚ to the load axis (34); the femur was thus prevented from rotating by the acetabular component and the trochanter strap. This jig had a weight-and-pulley system acting on a transverse crossbar, so when the torsional load was applied to the femur, this pulley-system was connected to the metal cylinder. Attached to the femur, the ITB was simulated by a wire from the trochanter (Figure 10). When micromotion was measured, the ITB was excluded.

Two activities, single leg stance and stair climbing, were simulated during strain and micromotion testing. The vertical force was 5/6 bodyweight (BW), calculated to be 600 N (corresponding to 73 kg bodyweight). Stair climbing was simulated by adding a torque of 13.8 Nm. Torsional moment was calculated as 1.9 % bodyweight meter (BWm) when the trochanter band and ITB were included.

9.4.2 Hip jig – Papers II-IV

The testing in Papers II-IV was performed in new facilities in an upgraded hip simulator and loaded in a servohydraulic testing machine (MTS 858 MiniBionix II, MTS System Corporation, Eden Prairie, Minnesota USA) (Figure 11).

Two human activities were simulated in this setup; single leg stance and stair climbing. The femurs were loaded proportionally to their individual donors´ bodyweight (BW), accounting for the inter-femur variability and loading. The femurs were loaded with axial forces corresponding to 1.15 bodyweight, due to the calibration file used in the test setup (upgrade of the MTS). Each test consisted of 5 cycles. Stair climbing was simulated by adding a torque corresponding to 2.0% BWm. Torsional load was applied to the femoral head by pulleys and wire connected to a second actuator of the testing machine. An abductor strap attached to the greater trochanter was mounted, simulating the abductors.

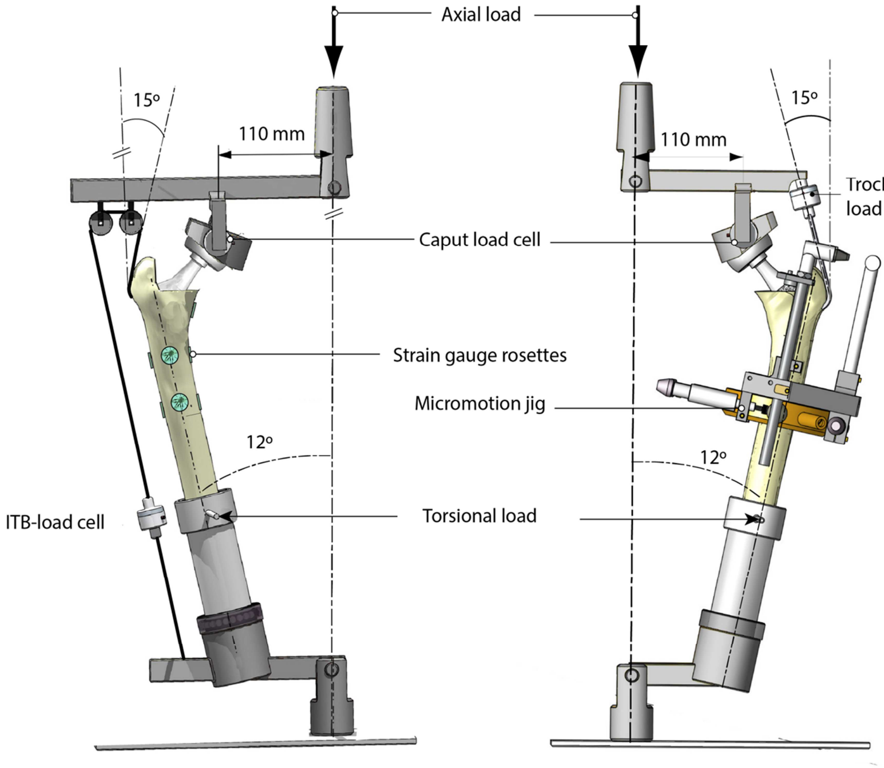

Figure 11. Hip simulators: with an ITB (left) and trochanter band and micromotion jig mounted (right).

9.4.3 Strain measurement

Prewired triaxial rosette strain gauges (FRA-3-23, Tokyo Sokki Kenkyujo) were used for strain measurements. Altogether seven rosettes were distributed on the anterior, medial and lateral sides of the proximal femur, at three predefined levels, 14, 34 and 64 mm inferior (table 2) to the lower boarder of the femoral head, corresponding to the Gruen zones around the proximal femur and previously used locations (125, 127, 129). The entire proximal femur was not covered for recording deformation in all areas, but the zones chosen were considered sufficient to address the issue of stress shielding.

Table 2. Predefined levels for the location of the strain gauges.

Level		Strain Gauge
A:	14 mm distally to the lower border of the femoral head	A_{med}, A_{ant}
B:	34 mm distally to the lower border of the femoral head	B_{med}, $B_{ant,}$ B_{lat}
C:	64 mm distally to the lower border of the femoral head	C_{med}, C_{lat}

The measurement of strain started on the intact femur in both loading conditions. Then the implanted femur was tested. Principal tensile strain was used for analysis of the deformation pattern on the lateral and the anterior aspects of the femur, whereas principal compressive strain was used for analysis of the medial aspect. The strain values are presented as percentage values relative to the strain values for the intact femur for each of the seven locations on both loading conditions.

The procedure including preparation and gluing the strain gauges to the femoral surface was based on a previously described method (125). The surface of the proximal femur was smoothened with sandpaper, and acetone and etchant (Scotchbond^{TM} Etchant, 3M ESPE, St. Paul, Minnesota) were applied and dried with N_2-gas. Then a primer (Scotchbond^{TM} Primer, 3M ESPE, St. Paul, Minnesota) was used. The next step was gluing the rosettes using epoxy glue (X60, HBM, Darmstadt, Germany), before finally covering the rosettes with waterproof sealing (Vitremer^{TM} Finishing Glass, 3M ESPE, St. Paul, Minnesota).

9.4.4 Micromotion measurement

The testing was implemented according to a previously described test setup evaluating primary hip stem stability in cadaver (126). The micromotion measurement device was based on two main components, a femoral ring attached to the femoral cortex, and a transducer frame attached to the implant. The femoral ring consisted of three 18 mm ceramic hemispheric ball probes fixed to a circular frame. The circular frame was locked to the bone with three screws that did not perforate the femoral cortex. The transducer frame was fixed to the implant through a yoke at the shoulder of the femoral stem, distal to the stem-neck junction (Figure 10). The junction between the stem and the modular neck was therefore not included in the measurement system. The frame could be moved freely along the femur in the superior/inferior direction, hence allowing micromotion measurements at any level along the prosthesis. Altogether, six Linear Variable Displacement Transducers (LVDTs) were used to obtain three-dimensional motion data (126). Three transducers (WA10, HBM, Darmstadt Germany) were positioned in parallel, and three transducers (W1T3, HBM, Darmstadt Germany) were positioned perpendicular to the longitudinal axis of the prosthesis. The outputs from the transducers were recorded by a measurement amplifier (UPM 100, HBM, Darmstadt, Germany). For each modular neck in both loading conditions (single leg and stair climbing), micromotion measurements were performed at a proximal and a distal level. The proximal level was defined as five mm distal to the proximal medial coating of the stem. The distal level was defined by the transition from horizontal to vertical grooves on the implant surface at the medial border (Figure 12).

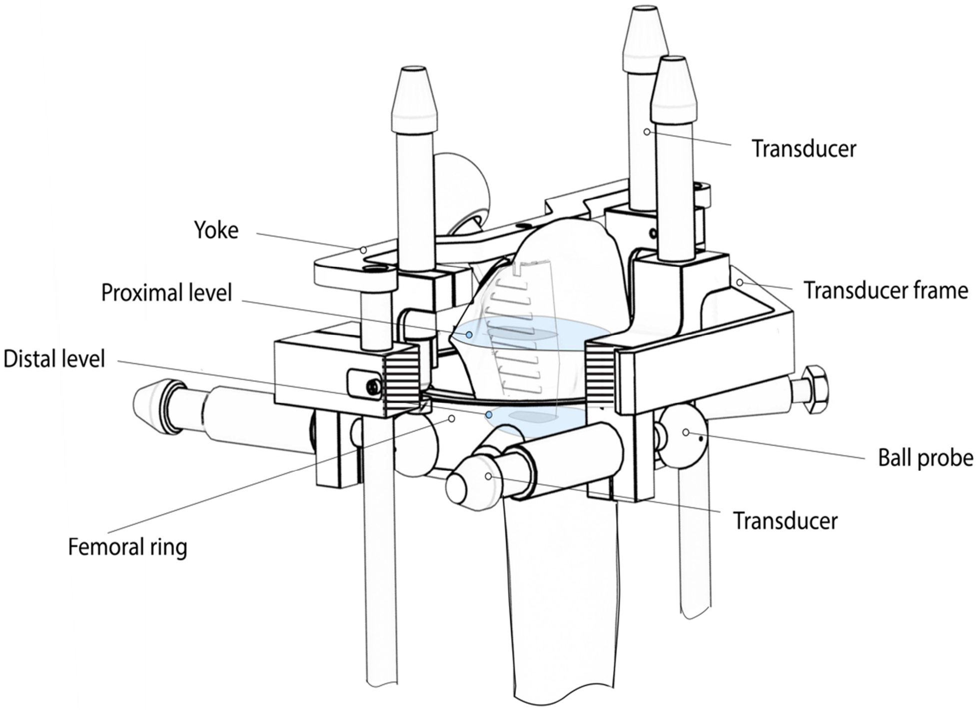

Figure 12. Micromotion jig in the hip simulator. Micromotion measurements at two predefined levels, one proximal and one distal level.

The femurs were preloaded before measurements started. Thereafter the loading was repeated 5 times, with relaxation intervals of 10 s between successive cycles. The mean of the measurements from the three last loadings was used for statistical comparisons.

The total point motion (TPM) was measured at two levels, proximal and distal, at the anterior, lateral and posterior aspects of the prosthesis in Paper I. In Papers II and IV, an average TPM was calculated for each of the two different levels (Figure 12).

9.5 Statistics

Power analysis of sample size was performed on strain data from previous laboratory studies; 10 subjects were included in paper I and 12 subjects included in Papers II - IV.

Deformation results from different locations and implant micromotion data from different levels are correlated. This requires a statistical model accounting for data dependency. The linear mixed model (LMM) was selected for statistical analyses of strain and micromotion in all four papers. This statistical method is considered to be robust when used in studies with factorial design and data dependency (130). The LMM accounts for the nature of the repeated measurements.

The literature search ended in May 2016.

In Paper I, the LMM was used to compare strain for three different eccentric femoral head designs. A separate analysis was conducted for each strain gauge with a significance level of $p<0.01$, due to multiple comparisons. The micromotion measurements were presented as mean calculated Total Point Motion (TPM) at the anterior, posterior and lateral side of the stem. Separate TPMs were calculated for the two loading conditions and at two levels. Normality of residuals was verified by Q-Q plots. Statistical analyses were performed using the software package IBM® SPSS Statistics version 16.

In papers II to IV, the statistical level of significance was set to 0.05 and p-values were Bonferroni corrected to adjust for multiple comparisons.

In Paper II, the initial stem stabilities for different modular necks were compared using LMM with the straight long neck as reference. An average TPM was estimated based on the TPM on the anterior, posterior and lateral side for each measurement level, and log- transformed values were used for the statistical analysis. Loading condition, measurement level and neck type served as fixed factors in the model. Normality of residuals was confirmed by histogram and Kolmogorov-Smirnov test. Statistical analyses were performed using the software package IBM® SPSS Statistics version 20.

In Paper III, LMM was used to analyze the strain pattern for four different modular necks. Strain results were expressed as percentage of intact strain values. Each of the seven strain gauge rosettes was analyzed separately. However, to account for the dependency between the strain gauge rosettes, percentage values from the other six strain gauge locations served as covariates in the LMM analysis. In addition, real strain values from the unoperated femur served as covariates. Statistically non-significant covariates were removed to define the most parsimonious model.

Neck type and loading condition were used as fixed factors in the model with the straight long neck as the reference neck. Normality of residuals was verified by histograms and Q-Q plots. Statistical analyses were performed using the software package IBM® SPSS Statistics version 21.

In Paper IV, LMM was used to analyse both the strain pattern and the initial stabilities of the two stems. Stem type, loading conditions and measurement level (micromotion analysis) served as fixed factors in the LMM analyses. Strain data was expressed as percentage of intact values from the unoperated femurs. Each strain gauge rosette was analyzed separately, considering that measurements from different rosettes were dependent. Real strain measurements from the unoperated intact femur and percentage strain values from the other six strain gauge locations were therefore included as covariates in the LMM analysis. Micromotion data was expressed as an average TPM for each measurement level and the log- transformed values were used for statistical analysis.

The residuals in strain and micromotion data were normally distributed, verified by Q-Q plots and histograms. Statistical analyses were performed using the software package IBM® SPSS Statistics version 21.

10 Summary of results

10.1 Paper I

Changes in deformation pattern and bone-implant micromotion in the proximal femur were explored after implantation of an uncemented stem coupled to a modular femoral head with increased offset, retroversion or reduced neck-shaft angle.

After insertion of a modular femur component, the strain was reduced, especially on the medial side. The strain was increased in Position 1 (increased offset and retroversion) and 2 (varus) compared to the standard femoral head, in medial and lateral location of the proximal femur (B_{med}, C_{med} and C_{lat}). The configuration with increased offset and altered neck angles gave a significant increase in strain, with a highest value 14.2%, compared to a standard femoral head on the distal anterior side (C_{ant}). The two loading conditions had statistically significant differences in all locations in the proximal femur, especially the anterior side with 86.3% (B_{ant}).

All three configurations with a femoral stem coupled to a modular femoral head had rather small TPM. At femoral head position 1 with increased offset and retroversion, the micromotion was measured at 40µm at the distal level.

The resultant forces in the hip joint were reduced in the test situation with the experimental heads, compared to the standard head.

10.2 Paper II

Micromotion and resultant hip joint forces were investigated in the proximal femur in an uncemented femoral stem coupled to different modular necks.

The modular varus neck showed the highest micromotions, 60µm, at stair climbing loading at the distal measurements level. The median micromotion for the reference neck was 38µm. Micromotion was significantly higher for the stair climbing activity compared to single leg loading, and for distal measurement level compared to the proximal level in all modular necks.

The resultant forces in the hip joint ranged from 2310N to 2500N, the highest values found with short and retroverted modular necks and the stair climbing activity.

10.3 Paper III

The deformation patterns in the proximal femur in a femoral stem coupled to modular necks with different geometry were evaluated.

All necks retained more strain than the reference neck at the lateral location (B_{lat}). The short neck had higher loss of strain at the distal lateral location (C_{lat}), and the retroverted neck retained more strain at the proximal medial location (A_{med}). The highest strain loss, compared to the unoperated femur, was observed in the proximal medial location (A_{med}), ranging from 13.6% (long) to 14.7% (retro) at single leg stance. Strain increased distally up to 66.3% of intact strain at distal medial location (C_{med}). The average strain values ranged from 76.9% to 77.9% on the lateral side. Anteriorly there was a difference between loading conditions, with an average of 130.1% (single leg) and 97.2% (stair climbing) at the proximal anterior location (A_{ant}). The corresponding values at the distal anterior location (B_{ant}) were 128.2% and 92.5% for corresponding values.

Median principal strain values ranged from -1733µm/m to 1672µm/m at the operated femurs. Strains were in general reduced on the medial and lateral side of femur, for all implants tested and in both loading conditions, compared to the intact femur.

10.4 Paper IV

Deformation patterns and initial stability after implantation of an uncemented and a cemented stem of identical geometrical shape were compared.

For the cemented stem, the strain measurements were higher than those of the uncemented stem on the medial side of the proximal femur. The differences were statistically significant in two out of three measurements sites on the medial side: 4.5 percentage points ($p<0.03$) at location B_{med} and 3.4 percentage points ($p<0.01$) at the C_{med} location, based on the overall model estimate. The uncemented stem had higher strain measurements on the lateral side. There was a significant statistical difference of 8.1 percentage points ($p<0.01$) at the distal level (C_{lat}).

For both implants, the cortical strains were reduced on the medial and lateral side of the proximal femur, compared to the unoperated femur. Strain increased distally along the stem and the strain measurements were more similar to the intact bone, for both the cemented and the uncemented fixation at the most distal measurement level. Strain values were in general more similar to physiological strain (intact values) on the anterior side of the femur.

The uncemented stem showed higher micromotions than the cemented stem in both loading conditions and both levels. The differences in TPM between the two implants was statistically significant, model estimates being 28.6µm versus 19.8µm ($p = 0.002$). In general, stair climbing was associated with higher micromotions than single leg stance, and the distal level showed higher micromotions than the proximal level.

11 General discussion

This thesis is based on a biomechanical in vitro model, using cadaver femoral bones. The outcome variables consisted of deformation, expressed as strain, and initial stability, expressed as micromotion. We tested a modular femoral head configuration and modular necks that could alter neck length and angulation. We also compared two different fixation methods.

Overall, the main findings in the study showed acceptable micromotion in all implants tested. The deformation patterns varied to a small degree between the implants and were probably too small to have clinical relevance. Despite this, a considerable loss of strain was observed in all operated femurs, compared to the intact values.

Preclinical studies are to some extent a simplification of a clinical setting. However, there is a range of variables that can be tested and evaluated, and different scenarios are easier to standardize in an experimental setup.

Introduction of new implants and methods in THA is time-consuming and research-intensive, and long observation time is needed to gain clinical acceptance. This is the background for the term “stepwise introduction”, coined by Malchau in 2000 (131). The first step in the innovation of new implants in orthopaedic surgery is the preclinical testing, which comprises laboratory investigations (132). Preclinical validation provides the opportunity to point out weaknesses of new designs, and avoid clinical introduction if the implant has too many failures. In vitro experiments serve as a basis for further testing and development of implants and for further clinical evaluation of the implant (42).

There are several ways to conduct preclinical testing. In vitro experiments driven by testing in a hip simulator have various setups, including investigating the implants´ primary stability with micromotion and migration, load transfer via strain gauges and structural strength using testing to failure.

An experimental setup allows paired testing, comparing two implants in one femur each or one implant and an unoperated situation. It is also possible to test modular components on a fixed femoral stem. The advantage is the opportunity to create comparable test groups, controlling the subjects´ individual variance. However, comparing the results can be difficult in the laboratory due to the testing conditions in different biomechanical setups (42). Different research groups often develop their own special patents, methods and experimental test setups that can make comparison of results challenging.

It is important to standardize the test setup in order to replicate and reproduce the results of in vitro studies. Further discussion will highlight the problems and limitations in the methodological considerations.

11.1 Methodological considerations

11.1.1 Implants

The experimental modular head evaluated in Paper I is not in clinical use. This modular head was based on the Metal-on-Metal (MoM) articulation, and this concept was recalled from the market in 2010 due to disturbingly higher failure rates (133). Data from several registries confirm poorer survivorship for MoM arthroplasties than for metal on polyethylene (6, 75, 107).

The modular neck concept and the femoral stems in Paper II-IV are clinically available (128). The modular neck implants have been evaluated by Omlor in 2010 on a mid-term basis, providing excellent clinical results (28). The implants evaluated in Papers II-IV were part of a clinical follow-up study on primary THA patients at Trondheim University Hospital. Preclinical setup was considered important to provide effective results on initial stability and deformation pattern of the femoral stems coupled to various necks. The concerns related to modular necks will be discussed in the modularity chapter (11.2).

11.1.2 Biomechanical testing

The hip simulator consists of a hip jig powered by a servohydraulic MTS MiniBionix II. The geometrical specifications were defined according to McLeish and Charnley (34), and the method has been standardized and evaluated at the Orthopaedic Research Centre of Trondheim University Hospital (125-127, 129).

During this project, the simulator has gained some improvements due to the moving of the biomechanical laboratory to new facilities. The design and base of the hip jig was the same, but a new supporting frame was added. The old mechanical torque device was replaced with a hydraulic actuator integrated in the hip simulator. A new controller was installed with updated controller software. The changes represented improvement in operational reliability, but the measurements were not affected. The test setup for the resultant muscle forces in the hip jig changed from including ITB in Study I to isolated abductor forces measured in the other studies.

There has been controversy regarding which muscles to include in the experimental setups (35, 42, 99, 134-136). For strain measurements, most laboratory studies include an abductor force keeping the bending moment in the femur (35, 42, 99, 135-138). Based on this evidence, we chose to simulate an abduction muscle. In Paper I, an ITB was added including the trochanter strap serving as an abductor force. The role of ITB in experimental studies has been discussed by many authors and according to Cristofolini there is diversity and little agreement on the experimental set up (42). Some conclude that the ITB has less resultant additive effect and seems unnecessary when creating a physiological loading condition simulating hip joint loading (35, 136). Based on these findings, the ITB was eliminated in the testing protocol for the results in Papers II-IV.

In the case of primary stability testing and involving muscles, the disagreement is significant (139). Some studies include only the abductors (83, 138, 140), some include multiple groups of muscles around the hip (99, 136) and some studies simulate a single hip contact force (95, 141, 142). The prevailing philosophy is to keep the test setup as simple and controllable as possible. On this basis, it was reasonable to keep only the abductors in Papers II-IV (126, 129, 143-145).

Fresh frozen human cadaver femurs were used in this experimental study. Many comparable in vitro studies use composite femurs (97, 102, 138, 146-149), and there are some advantages of synthetic bones. Composite bones are easier to store, easy to obtain, do not need preparation of soft tissue and have the same geometry, keeping variation between the bones to a minimum (42). The similarity of the synthetic bones with the lack of variation can increase the sensitivity in experimental studies (42), however, the synthetic bones will not have the natural variety between subjects. Cadaver femurs can be more difficult to provide, due to the ethical aspect and the reduction of post mortem autopsies. Using human bones for research requires approval from an ethical committee and consent from relatives. Despite the fact that this is time consuming, natural human bones often are preferred in experimental studies (150-154). Human cadaver bones are considered to be more clinically relevant, representing a natural group of subjects similar to the clinical scenario. Single femurs were used in this setup, randomized to left or right. The femurs served as their own control, in all four papers.

Many in vitro setup use a standardized loading force for all subjects. This is due to the preclinical experience of Cristofolini and his research group, and based on their recommendations, 600 N corresponding to 73 kg BW is the typical force applied for vertically loading the subjects (147). This loading setup was used in Paper I. In Papers II-IV, we used individual loading corresponding to the donor's BW, measured at autopsy. This is considered to be an advantage because it brings the setup closer to a clinical loading situation.

Subjects' specific loading gives a more correct picture of the absolute strain. In the present studies, strain measurements are presented as percentages of intact strain (relative strain). Individual loading is less significant for these data. Individual loading could also reduce the inter-femur variability of micromotion measurements. Fixed loading could give falsely elevated absolute strain or micromotion values. The theoretical advantages of individual loading could be challenged when the subjects are too heavy. We found that subjects with bodyweight above 90 kg yielded greater problems with failures during testing. This occurred despite the BMI limitation on the donors, and must be attributed to limitations of the technical machinery. Despite the BMI limitation, subjects with a BW exceeding 90 kilos did introduce challenges even with a normal BMI. For further research on this topic we will suggest a specific weight limit, rather than a BMI limit, because the specimens are fragile in the testing situation.

The single vertical loading was planned to be 0.83 of BW (5/6) at single leg stance. This loading was performed in Paper I. Due to a calibration file, the actual axial forces were 1.15 BW in Paper II-IV. This increased loading force probably led to some of the failures during testing. Despite this, the high loading is considered clinically relevant as the implants were tested in a conservative manner. Telemetric studies have showed a range of variation when testing daily activities (37, 155, 156). If the micromotion was increased because of this testing sequence, the identification of differences would be more likely. Absolute strain is dependent on the cortical thickness and BMD of the subjects, but as long as the strain values are presented relative to the intact results, the importance of individual loading is somewhat less important.

The hip and abductor forces and the bending moment on the femur could be changed due to changes in head position after insertion of the femoral stem. The altering of the magnitude and direction of the hip joint force, the abduction force and the resultant force (illustrated in Figure 2) would affect the bending moment of the femur and further an increase or decrease of the torsional moment (157). With an increased medial offset, the resultant force and abductor force are reduced. This leads to an increase in bending and torsional moment. In our setup, these concerns were controlled using a skilled orthopaedic surgeon for all implantations. During the testing procedure, the engineer checked that the medial offset was reproduced and that the angle in the trochanter strap was 15°, representing a biomechanical situation.

The cortical strain pattern was measured by strain gauges rosettes, considered a common technique measuring strain in bone (42). Each rosette consists of three strain gauges and seven areas of the proximal femur were covered and principal strain in these locations were measured (Table 2). The weakness is that the strain gauge only provides information from the local attachment site only, and do not cover the whole proximal femur. The strains in the trabecular bone cannot be evaluated by this method either. However, the advantage is the accuracy of the method showing a direct quantitative measure from each strain gauge position (42).

For measurement of primary stability, we used an indirect method. Primary stability measurement using a direct method would have required holes drilled through cortical bone. In our study, this was not an option for two reasons. Firstly, it would lead to a possible mechanical weakening and influence the values for strain measurements. Secondly, it would have led to problems during cementation, with leakage and damage to the areas for data registration.

In a retrieval study on dogs, Pilliar found that uncemented porous implants were stable with micromotion less than 28μm. The loose stems, with increased micromotion higher than 150μm, had a predominant ingrowth of fibrous tissue and lack of bone ingrowth. Micromotion less than 20μm was optimal for bone ingrowth, according to this author (82).

There is agreement that micromotion exceeding 150μm at the bone-implant interface prevents the osseointegration and favors the development of a fibrous tissue layer, which may lead to a lack of secondary stability in uncemented implants (82, 84, 103, 158). If a bioactive agent is added to the coating, such as hydroxyapatite (HA), the integration of bone is inducted, and this could allow for a higher threshold of micromotion. However, experimental studies indicate that 150μm is a reasonable limit (82, 84).

This in vitro model naturally does not simulate the in vivo biological bone ingrowth and ongrowth situation. This limitation must be considered in the interpretation of the results of stability. Only the initial postoperative stability can be evaluated and discussed more thorough further.

11.2 Is modularity needed in THA?

This thesis addresses two principles of modular implants. The first concept is an experimental modular head that can be coupled to the neck taper eccentrically. The other concept is modular necks with different directions. The modularity of both systems results in a variations of neck length, version and neck shaft angle.

The overall results regarding deformation patterns for the experimental head showed relatively small influence on the shift from a standard femoral head position to situations with increased offset and altered neck angle. In Paper III where strain distribution was compared in different modular necks implanted to an uncemented femoral stem, showing similar findings as the experimental head.

Similarly designed in vitro studies measuring deformation pattern in the proximal femur after implantation of uncemented stems with modular necks in synthetic bones reported a correlation between compressive strain on the side toward which the prosthetic neck was oriented and the extent of the neck version (148, 159). However, other neck combinations than those examined in the present studies were compared, and a direct link to our results cannot be established.

The micromotions were small for all three head positions in Study I. Also in Paper II the initial stability was within acceptable ranges. The varus and retroverted necks gave higher micromotion values than the straight neck, but these differences in micromotion would probably not affect secondary osseointegration.

Keeping the lever arm constant before and after implantation could be of importance for stem longevity and stability. A small femoral cemented stem, combined with high offset, could lead to increased risk of aseptic revisions (160). According to another study, increased offset could increase the micromotion of the stem (161). There was no statistically significant difference between the short and the long neck in our study.

When simulating stair climbing in our study, a constant individual torque was applied to the distal femur. The effect from increased femoral offset on the torsional moment of the implant was therefore not demonstrated. According to our results, increasing the offset and retroversion or reducing the neck shaft angle in vitro does not seem to have a clinical impact.

Two studies reported excellent survival in a 10 and 13year clinical follow-up after primary THA, with modular necks of similar type as used in the present studies (28, 115). This conclusion contrasts with reports from the Australian joint registry showing a significantly higher revision rate ten years after surgery for femoral stems with modular necks, compared to conventional THA (107). The most frequent reasons for revision in this report were loosening and osteolysis, and the question we asked for the present study (Paper II) was if the increased micromotion was a potential cause of loosening. The micromotion measured in the retro and varus modular necks in Paper II was significantly greater than the micromotion in the reference straight neck; however, the results in our study showed micromotion values at acceptable levels, and findings from the present study do not support the hypothesis that loosening of the implants can be attributed to increased micromotion.

One possible advantage of the modular necks is that only the head and the neck can be exchanged during revision, leaving the stem in the femoral canal. However, mechanical failure in the neck-stem junction first reported from 2010 disproves the benefit of this (118-122). In these reports, a long or varus neck was been pointed out as a risk for mechanical failure. The reports on mechanical failures were due to damage and cold welding in the junction (162, 163). Corrosion and fretting is a known phenomenon after inserting the modular necks, which predisposes for fatigue fractures. The origin of this process is believed to be a combination of fluid ingress and release of ions because of the mixed alloy interfaces and combinations. Modular junctions then become vulnerable to corrosion and fretting (116, 117, 119, 164).

The problem with metal ion release and tissue reactions around implants could lead to formation of pseudotumours, related to metal-on-metal bearings and resurfacing implants (165-167). The formation of pseudotumor has also been pointed out as a problem in the junctions of modular necks (123).

On the basis of national joint registry data from Australia in 2010-2011, a recall and safety notice was made in 2012 (168), due to a higher revision rate for modular neck implants (124). The joint registry in England and Wales performed a comparison between the modular neck system and fixed stems in uncemented implants, concluded that modular necks had a higher revision rate (75). Warnings from the UK Medicines and Healthcare Products Regulatory Agency were also issued in 2013 (169). The Norwegian arthroplasty registry also decided to warn the national orthopaedic society against the use of modular necks in primary THA. They pointed out this issue, concerning increased revision rates, in the introduction of the annual report since 2013 (170).

The concerns raised around modular components cannot be directly linked to the findings in our study, but underline the fact that these implants need additional follow-up over time (6, 75, 76, 107).

One of the problems with the data from the national registry studies is that all implants used in the clinical practice, with a variety of designs, concepts and materials, are included. We evaluated one type of modularity, similar to the concept Omlor studied (28). This specific modular neck system shows overall acceptable results, also according to the Australian joint replacement registry (107).

11.3 Effect of fixation method

Both cemented and uncemented femoral stems showed a reduction in strain in the lateral and medial parts of the femur compared to those of the unoperated femur, especially in the proximal part. The cemented stem had higher strains than the uncemented stem on the medial side and the uncemented stem had the highest strain measurements on the lateral side. The differences were small, but statistically significant. The results showed no consistency in the deformation pattern in the proximal femur, and did not favor any of the stems. These results are consistent with findings from previous studies (146, 147, 171).

Geometry and design features of implants are important factors in survival of prostheses. Registry studies and literature reviews have shown that cemented stems are slightly better than uncemented ones (6, 79-81, 172), but there is a tendency towards increased use of uncemented stems in clinical practice (6, 76, 81, 107, 172, 173).

Studies on hemiarthroplasty, comparing cemented and uncemented femoral stems in femoral neck fractures, conclude that there is better survival of cemented stems (174-179). Uncemented stems probably require better bone stock for longer survival of the prostheses (178, 179).

Periprosthetic bone remodeling around cemented and uncemented components, and different patterns of stress shielding related to type of fixation of the implant, have been evaluated by several authors. The conclusions are mainly that bone density is better preserved around uncemented stems (180-183). Grant, however, showed that an uncemented anatomical femoral stem had higher proximal bone loss than the cemented standard stem it was compared to, when measuring BMD in the proximal femur (184). He found small differences similar to our results. However, the two stems that were compared had quite different designs. The uncemented stem had an anatomical stem filling in the metaphysis while the cemented implant was much narrower, and therefore the stiffness of the two stems was quite different. These differences in design must be considered, and probably they affect the deformation and bone remodeling more than the fixation method.

Increased micromotion in cemented femoral stems may also lead to loosening of the implant (39, 185, 186). In this study, we found that an uncemented stem performed in an acceptable range of micromotions. The cemented stem showed micromotions that were significantly lower than the uncemented stem, as expected. Still, the difference between the cemented and the uncemented fixation was less than 10µm/m in the model estimates. Only a few in vitro studies have compared the micromotion of a cemented and an uncemented stem. Our findings correspond to the findings of

Cristofolini, which showed inducible motions of 16 to 34 µm for two cemented stems (146). Another study found the same magnitude for a cemented stem (26 µm) but higher movements for the uncemented stem (103µm) (140). There were two outliers in the cemented group, two subjects showed excessive micromotions, corresponding to the findings of Burke (140). This might be because the subjects in our study were loaded with 1.15 BW, so that a few microcracks in the cement mantle could have been induced.

The stiffness of the material used in implants is important with regard to bone remodeling around an implant (43, 60, 67, 187). The two stems in this study had similar geometry, but material and stiffness differed. The uncemented stem was made of titanium alloy with an elastic modulus (E) of about 110GPa, while the cemented stem consisted of a CoCr alloy, possessing an E of about 205GPa. The cortical bone had an E of about 15-20GPa. Thus, both stems were far stiffer than the bone, five to ten times, respectively. The difference in E could theoretically have contributed to the alteration of the deformation pattern. However, the differences found in our study cannot be explained by the stiffness of the stems, as the difference between the stems was small. It is uncertain whether the cemented stem and its higher stiffness affect the deformation pattern to a significant extent, compared to the uncemented stem. Another issue is the distance between the implant and cortical bone, comprising the cement in between. There is a chance that the stress pattern and deformation in the proximal femur could be affected by the cement, having a lower value of E.

The main finding is therefore a similar pattern for the two stems, where both femoral stems showed loss of deformation on the lateral and medial side of the proximal femur compared to an intact femur. The cemented stem was more stable in both loading conditions and levels compared to the uncemented stem, as expected. The uncemented stem showed micromotion up to 40µm, considered to be within an acceptable range for micromotion (62, 82, 84).

11.4 What are the implications of change in strain?

A femoral stem with increased stiffness can affect the bone and cause problems in the bone-implant interface (57, 65, 67). We observed a substantial decrease in strain proximally as a result of the altered load transfer, after insertion of a femoral stem, in all three papers (I, III and IV). This phenomenon is called stress shielding, as the load bypasses the proximal femur and is transferred to the bone more distally. Stress shielding is associated with proximal bone resorption, observed clinically after THA. Preservation of the proximal bone stock is considered important, but the optimal values of strain to maintain a physiological bone remodeling in the femur is not known. However, there is to some extent consensus on the relationship between mechanical load and adaptive bone remodeling (188, 189). A review from 2006 on femoral designs concludes that progressive bone loss through stress shielding has potentially critical consequences, and conservation of femoral bone stock is considered important (60). The development of new design features of hip stems is intended to reduce postoperative bone loss, addressing stress shielding as a problem. The implantation of a femoral stem affected the deformation pattern in the proximal femur more than the variations due to the modular necks.

11.5 What are the implications of micromotion?

Initial stability of the femoral stem is an important factor for long-term survival of the implant. Excessive micromotion between the implant and bone surface could lead to an inhibition of bone ingrowth in uncemented implants and hence to aseptic loosening (43, 70, 82, 84, 102, 103).

Since cemented stems are fixed with cement, they reach initial stability only hours after implantation. However, micromotion in cemented femoral stems may also lead to loosening of the implant (39, 185, 186). In Paper IV, the uncemented stem showed higher micromotion than the cemented stem in both loading conditions and levels. However, the cemented implant to some extent had micromotion above the level we expected. The same findings were experienced in a vitro study comparing cemented and uncemented femoral components in single leg stance and stair climbing (140).

In a comparative in vitro study on long-term performance, a stem with stem-cement debonding had micromovements over 173µm in the longitudinal direction in migration and 75µm in inducible motion. This was a study on synthetic bones loaded axially up to 1683N (146, 190). Both implants tested in our setup showed micromotion below 30µm. Still, there were some outliers in the cemented group showing excessive micromotion, and the question is whether this could be due to microcracks in the cement mantle. This problem was not investigated in this experimental study, and a retrieval study might be needed.

In a study on micromotion and migration, three concepts of implant-bone interface fixation were evaluated. A partially cemented stem was compared to a standard cemented and an uncemented press-fit stem. The hybrid and cemented stem were very stable initially, showing micromotion < 10µm in all types of loading conditions and localizations (96).

In a finite element model (FEM), micromotion was significantly improved for an uncemented stem due to reduced interfacial gaps, using a different broaching method (191). Manual broaching was done in our study to create a press fit for the uncemented implant. The micromotion result was measured at < 40µm, a level where osseointegration can occur. The results probably show that both stems would have achieved final acceptable stability in our model.

Stress shielding and micromotion are often pointed out as local factors in the bone formation process around the prosthesis, and explains why variations in bone ingrowth and ongrowth are found individually and among different implants (61). Due to these current factors related to the implant design, many types of designs of uncemented hip stems shows good long-term outcome (89).

12 Conclusions

Despite the fact that survival of many established hip prostheses is good, new implants need long observation time in order to evaluate gain in function or survival. This underlines the need for documentation of new implants, and experimental testing and preclinical studies constitute a scientific need to predict long-term clinical outcome (192). There is consensus in the orthopaedic research environment that new hip implant technology needs to be tested in vitro, going through a stepwise introduction before further clinical trials (132).

An uncemented femoral stem coupled to a modular femoral head configuration of retroversion or reduced neck shaft angle with an increase in medial offset showed significantly higher cortical strain in the proximal femur compared to a standard femoral head. The differences in strain are, however, considered too small to have a clinical impact (Paper I).

The initial stability of an uncemented stem with different modular head configurations was not affected under tested loading conditions, and showed acceptable micromotion in all head positions (Paper I).

An uncemented stem coupled to modular necks of different versions and lengths showed micromotion within an acceptable range, but subject-specific risk factors should be considered clinically (Paper II).

An uncemented stem coupled to modular necks of different versions and lengths showed only small, although statistically significant, variations in deformation pattern. One can-not expect any difference in bone remodeling in the proximal femur related to the use of modular necks with different geometry (Paper III).

An uncemented and cemented femoral stem showed acceptable initial stability. There were small differences, although statistically significant, between the stems in micromotion. The cemented stem was more stable in both loading conditions and levels. Two femoral stems with similar geometry, but different fixation showed loss of deformation of the proximal femur compared to the unoperated femur. There is no evidence that one of the stems had a deformation pattern that was clearly more similar to the intact femur (Paper IV).

Varying the femoral neck version, angle and length by either an eccentric femoral head or a modular neck gave some variations in cortical strains in the proximal femur compared to a standard design. However, the differences might be too small to have any clinical significance. The initial stability of tested implants showed acceptable micromotion.

The cemented stem showed higher initial stability than the uncemented stem, as expected. Both stems had small micromotions at the bone-implant interface, and in a range, that is not expected to have a negative impact on long-term stability.

13 Future directions

Preclinical studies are an important step in the development of new implants and methods. The goal for in vitro testing is to predict clinical outcomes and identify potential negative side-effects for new arthroplasty devices and methodological variants. In the short term, two implant systems can be effectively compared and evaluated. The challenge is the simplified, non-biological and short-term testing performed in the laboratory. Nevertheless, efforts should be made to develop more clinically relevant methods. Multi- point strain measurement, as used in the present studies, could be expanded to encompass more factors, such as simulation of bonding to bone and long-time migration of implants in dynamic biomechanical tests. Subject specific loading according to individual bodyweights is more relevant for simulation of true strain values and micromotion data. Individual pelvic size can be applied in the hip jig setup as well in an attempt to best reflect the original geometry and forces.

The laboratory data compiled could form the basis for e.g. finite element numerical models and compared with relevant clinical data. This could lead to a database that makes it possible to forecast clinical behavior with better precision than is possible today, providing patients with safe and appropriate arthroplasties.

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Initial stability of an uncemented femoral stem with modular necks. An experimental study in human cadaver femurs

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ABSTRACT

Background: Uncemented implants are dependent upon initial postoperative stability to gain bone ingrowth and secondary stability. The possibility to vary femoral offset and neck angles using modular necks in total hip arthroplasty increases the flexibility in the reconstruction of the geometry of the hip joint. The purpose of this study was to investigate and evaluate initial stability of an uncemented stem coupled to four different modular necks.
Methods: A cementless femoral stem was implanted in twelve human cadaver femurs and tested in a hip simulator with patient specific load for each patient corresponding to single leg stance and stair climbing activity. The stems were tested with four different modular necks; long, short, retro and varus. The long neck was used as reference in statistical comparisons. A micromotion jig was used to measure bone-implant movements, at two predefined levels.
Findings: A femoral stem coupled to a varus neck had the highest value of micromotion measured for stair climbing at the distal measurement level (60 μm). The micromotions measured with varus and retro necks were significantly larger than motions observed with the reference modular neck, $P < 0.001$.
Interpretation: The femoral stem evaluated in this study showed acceptable micromotion values for the investigated loading conditions when coupled to modular necks with different lengths, versions and neck-shaft angles.

1. Introduction

Total hip arthroplasty (THA) is considered to be a successful treatment for destructive diseases of the hip joint. The number of implant designs and the use of uncemented implants have increased (The Norwegian Arthroplasty Registry, 2013). Uncemented prostheses are dependent on adequate primary stability to achieve bony fixation and long term stability of the implant (Callaghan et al., 1992; Pilliar et al., 1986; Soballe et al., 1992b). The uncemented implants gain long-time fixation by osseointegration to the surface layer of the implant. Excessive interface motion reduces or inhibits bone ingrowth and may lead to loosening of the prosthesis (Cross and Spycher, 2008).

Initial stability of a femoral stem is dependent on a number of factors such as implant design, surface roughness, surgical technique and patient related factors like quality of bone (Khanuja et al., 2011). The movement at the bone-implant interface can be expressed as migration and micromotion. Migration is an irreversible movement of the stem into the femoral canal, typically occurring during the first postoperative period (Buhler et al., 1997). Micromotion is a reversible movement at the bone-implant interface that occurs under dynamic loading. Micromotion can be estimated by numerical analyses or by a multitude of methods, involving in vitro measurements (Baleani et al., 2000; Buhler et al., 1997; Gortchacow et al., 2011; Gortz et al., 2002; Kassi et al., 2005; Nogler et al., 2004; Tarala et al., 2011). Experimental studies have found that excessive micromotion can compromise or inhibit the biological integration of bone at the implant surface (Engh et al., 1999; Jasty et al., 1997; McKellop et al., 1991; Pilliar et al., 1986; Soballe et al., 1992b), however the exact range of motion that will allow osseointegration is not known.

Modular neck in THA is a concept allowing variations in neck-shaft angles, neck version and neck length. These necks have been introduced to improve accuracy when reconstructing the anatomy and hip joint biomechanics.

The use of modular necks in primary THA has increased in recent years. There are some reports of good mid-term outcomes (Matsushita et al., 2010; Omlor et al., 2010), but long-term documentation is limited. A few case reports and studies raise concerns of corrosion, mechanical failure and pseudotumour formation related to

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the concept of modularity (Dangles and Altstetter, 2010; Gill et al., 2012; Skendzel et al., 2011; Sotereanos et al., 2013; Viceconti et al., 1996, 1997). The Australian Joint Registry reports that THA with exchangeable femoral necks has twice the rate of revision compared to conventional THA after 7 years. The primary reasons for revision are implant loosening and dislocation (Australian Orthopaedic Association National Joint Replacement Registry, 2012).

The purpose of this study is to evaluate the primary stability of an uncemented femoral stem with four different modular necks varying version, length and neck-shaft angle.

2. Methods

2.1. General

This study was approved by the regional medical research ethics committee. Pilot studies were completed to develop a satisfactory testing sequence and structure, and the testing was performed according to a well-established procedure (Ostbyhaug et al., 2010; Wik et al., 2011).

2.2. Implant system

A collarless cementless titanium alloy stem fully coated with hydroxyapatite (HA) (Profemur® PRGLKITD Gladiator, Wright Medical Technology Inc., Arlington, TN 38002, USA)(Fig. 3) was implanted into 12 human cadaver femoral bones and randomly allocated to right or left sides. All implantations were done by an experienced orthopaedic surgeon according to the manufacturer's procedure (Wright Medical Technology, 2013).

Four different modular titanium necks with a 12/14 taper (Profemur® Modular Necks, Wright Medical Technology Inc., Arlington, TN 38002, USA) were evaluated: 1. straight long (PHAO 1204), 2. straight short (PHAO 1202), 3. retroversion short 15° (PHAO 1262) and 4. varus short 15° (PHAO 1242) modular components (Fig. 1). The necks were connected with the oval end of the appropriate femoral neck implant into the femoral stem pocket. A standard 28 mm femoral head was used.

2.3. Human cadaver femurs

The femoral stems were implanted into Caucasian human cadaver femurs. The femurs were collected from deceased patients that underwent medical post-mortem examinations within 24 h. Consents from the relatives were collected before interfering. Twelve human femurs completed the testing, mean donor age was 58 years (range 43–70 years), nine male and three female (Table 1).

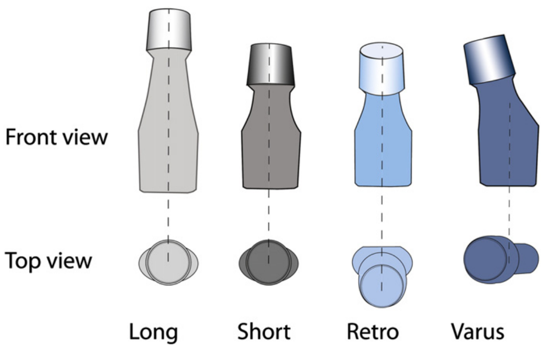

Fig. 1. Profemur® modular necks: Straight long, straight short, retroversion short 15° and varus short 15° in front view and top view.

Table 1
Data of the subjects; N = 1–12, gender, age, body mass index (BMI) and bone mineral density (BMD).

ID	Gender	Age	BMI	BMD
1	M	59	24	0.943
2	M	57	28	1.163
3	M	66	27	0.963
4	M	70	23	1.063
5	F	53	20	0.959
6	F	57	25	0.998
7	F	62	20	0.896
8	M	64	29	0.891
9	M	53	22	0.894
10	M	67	25	0.94
11	M	47	24	0.962
12	M	61	18	0.944

The femurs were handled and prepared according to an earlier described and well documented procedure (Aamodt et al., 1997; Wik et al., 2010). The femurs were wrapped in saline-soaked towels and stored at −80 °C immediately after dissection.

Standard radiographs in two projections were used to estimate the size of the prosthesis and to exclude any skeletal pathologies. Dual-energy X-ray absorptiometry (DXA) was obtained to point out possible osteoporotic femurs (Table 1). Bones with T-scores of the proximal femur below −2.5 were classified as osteoporotic and excluded.

Criteria for selection of femurs included age <70 years, body mass index (BMI) between 18–30, no previous fracture in the femur and no current or previous malignancy in the femur. Twenty-one pairs of femora were collected. Three patients were excluded due to osteoporosis and one pair was destroyed during preparation. Five femora failed during testing (three due to periprosthetic fractures and two due to trochanter band failure).

Before testing, the femurs were thawed at room temperature and remaining soft tissue removed. First the frontal plane of femur was defined by placing the femur on a horizontal surface resting on the posterior condyles and the greater trochanter. Second the anteversion of the femoral neck was measured and recorded for later orientation of the femur in the frontal and sagittal planes, before resecting the condyles. The femur was then fixed into a steel cylinder with an acrylic cement (Meliodent, Heraeus Kulzer GmbH, Hanau, Germany), where the centre axis of femur coincided with the centre axis of the cylinder. The femur was kept humid by a saline-soaked towel during preparation.

The distance from the tip of the greater trochanter to the top of the cylinder was 25 cm for all specimens. To simulate the hip abductor muscle a 40 mm polyamide strap was attached to the greater trochanter using glue (X60, HBM GmbH, Darmstadt, Germany) and 6 screws (cortical 2.5 mm) (Fig. 2).

2.4. Hip simulator

The implanted femurs were mounted into a hip jig and loaded in a servohydraulic testing machine (MTS 858 MiniBionix II, MTS System Corporation, Eden Prairie, Minnesota, USA). This constituted the hip simulator (Fig. 2). The femur was allowed to rotate freely around its longitudinal axis and to tilt freely in the medial/lateral plane, to avoid unphysiological bending moments.

The femur was tilted 12° into valgus, corresponding to physiological inclination during single leg stance (McLeish and Charnley, 1970). For the experiments an acetabular cup with an inclination of 45° and 0° anteversion was used. A trochanter strap was fixed to the lever arm at an angle of 15° to the load axis (McLeish and Charnley, 1970). The femur was prevented from rotating by the acetabular component and the trochanter strap.

Two human activities were tested; single leg stance and stair climbing. The femurs were loaded proportional to their individual donor body weight (BW). A single vertical force, originally planned to

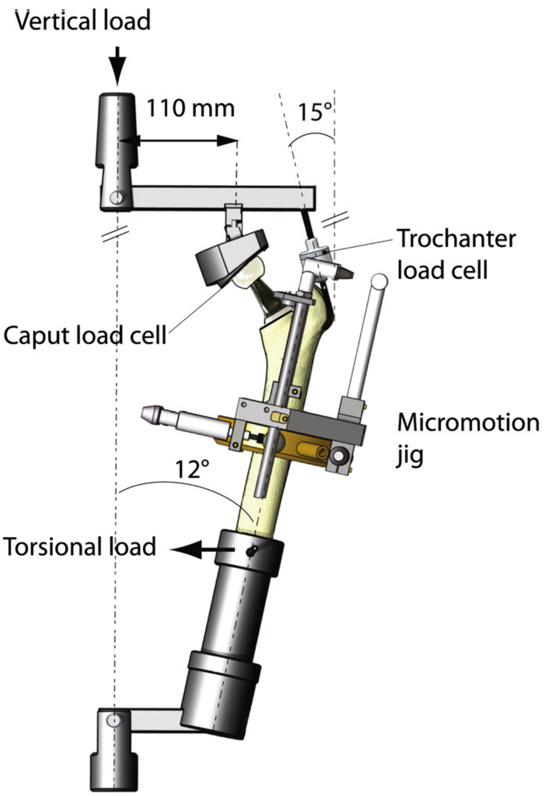

Fig. 2. Schematic illustration of the hip simulator.

be 0.83 of body weight, was applied through the actuator of the MTS to the femur simulating the single leg stance (Fig. 2). The femurs were actually loaded with axial forces corresponding to 1.15 body weight, due to the calibration file used in the test setup. Each test consisted of 5 cycles with a consistent axial load. Stair climbing was simulated by adding a torque corresponding to 2.0% body weight meter. Torsional load was applied to the femoral head by pulleys and wire connected to a second actuator of the testing machine.

2.5. Micromotion measurement

The testing was accomplished according to previously described test setup used to measure primary hip stem stability in cadaver studies (Ostbyhaug et al., 2010; Wik et al., 2011). The micromotion measurement device was based on two main components, a femoral ring attached to the femoral cortex, and a transducer frame attached to the implant. The femoral ring consisted of three 18 mm ceramic hemispheric ball probes fixed to a circular frame. The circular frame was locked to the bone with three screws that did not perforate the femoral cortex. The transducer frame was fixed to the implant through a yoke at the shoulder of the femoral stem distal to the stem–neck conjunction. The conjunction between the stem and the modular neck was therefore not included in the measurement system. The frame could be moved freely along the femur in the superior/inferior direction allowing micromotion measurements at any level along the prosthesis. Altogether six Linear Variable Displacement Transducers (LVDTs) were used to obtain three-dimensional motion data (Ostbyhaug et al., 2010). Three transducers (WA10, HBM, Darmstadt Germany) were positioned parallel and three transducers (W1T3, HBM, Darmstadt, Germany) were positioned perpendicular to the longitudinal axis of the prosthesis. The outputs from the transducers were recorded by a measurement amplifier (UPM 100, HBM, Darmstadt, Germany). For each modular neck in both loading conditions (single leg and stair climbing) measurements were performed at a proximal and a distal level. The proximal level was defined 5 mm distal to the proximal medial coating of the stem. The distal level was defined by the transition from horizontal to vertical grooves on the implant surface at the medial border (Fig. 3).

The femurs were preloaded and thereafter the loading was repeated 5 times, with relaxation interval of 10 s between successive cycles. The mean of the measurements from the three last loadings was used for statistical comparisons. The micromotion measurements were described by three translations and three rotations of the stem at each measurement level.

We also calculated the total point motion (TPM) at the anterior, lateral and posterior aspects of the prosthesis at each measurement level. All measurements are presented according to the coordinate system representing a left stem.

2.6. Statistics

The average TPM was calculated for each of the two measurement levels, and the log-transformed values were used for statistical analysis. A linear mixed model (LMM) was used to analyze the differences in micromotion and resultant forces between the necks. The LMM accounts for repeated measurements and data dependency (Gueorguieva and Krystal, 2004).

The model was designed with three fixed factors; loading condition (single leg/stair climbing), measurement level of the femoral stem (distal/proximal) and four different necks (long, short, varus and retroversion). The residuals of the log-transformed data were normally distributed confirmed by histogram and Kolmogorov–Smirnov test.

The interaction between the necks, measurement level and loading conditions was evaluated to assess whether the activity or measurement level influenced the differences found between the necks.

In the statistical comparisons the straight long neck served as the reference. Level of significance was set to 0.05. All statistical analyses were performed using the software package IBM® SPSS Statistics version 20.

The resultant hip joint forces are presented as mean with confidence intervals, as the data were normally distributed.

3. Results

3.1. Micromotion

Micromotions of the femoral stem with retro and varus necks were larger than with the reference neck ($P < =0.001$, Table 2). The highest median TPM was demonstrated by the femoral stem with the varus neck (60 μm) at the distal level for stair climbing activity (Fig. 3). The corresponding median value for the reference neck was 38 μm. The varus neck had the largest median value for lateral translation and varus rotation (Table 3).

There was no difference between the straight short and the straight long neck. Overall, micromotions measured for stair climbing were larger than for the single leg loading ($P < 0.001$) and micromotions at the distal level were also larger compared to the proximal level ($P = 0.01$).

There were no significant interactions between the loading condition or the measurement level and the femoral necks.

3.2. Resultant hip joint forces

The resultant hip joint forces in this experimental setup ranged from 2310 to 2500 N (Table 4). The forces measured with the short neck and the retro neck were significantly larger compared to the long neck,

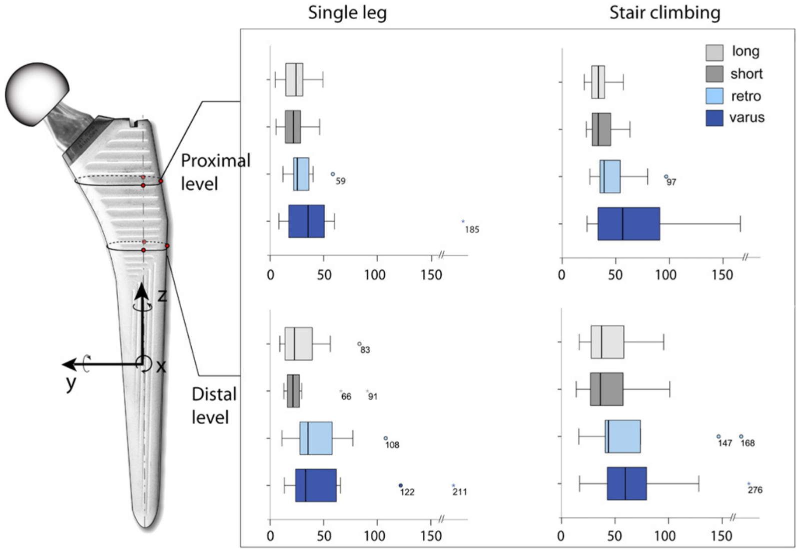

Fig. 3. The box-plot shows median total point of motion (TPM) and quartiles for the four different necks in single leg stance and stair climbing at the proximal and distal measurement levels. The points are outliers, and the stars are extreme outliers. N = 12.

$P = 0.001$ and $P < 0.001$ respectively. A larger resultant hip joint force was also observed for stair climbing activity than for single leg stance ($P = 0.002$).

4. Discussion

In this cadaveric study we evaluated the primary stability of an uncemented femoral stem in human cadaver femurs varying the version and length of a modular neck. The stems coupled to retroverted and varus necks exhibited larger micromotions compared to the reference neck. Overall the median total point motion values were less than 60 μm in all testing configurations (Fig. 3).

High micromotions between bone and implant can lead to formation of fibrous tissue leading to loosening of the prosthesis (Pilliar et al., 1986; Soballe et al., 1992a). Although both experimental and clinical studies have been performed to evaluate the tolerance of interface micromotions, the threshold has not been established. Data from a retrieval animal study showed that stable implants had micromovements less than 28 μm whereas loose implants had values higher than 150 μm (Pilliar et al., 1986). Later publications have suggested a higher tolerance of micromovements when the implant is coated with a bioactive calcium phosphate, such as hydroxyapatite (Engh et al., 1999; Rahbek et al., 2005; Soballe et al., 1992b). In this study we have measured differences in stem micromotions between four modular necks with a standard long neck as a baseline for comparisons. Although the median values for the varus and retro necks showed a statistically significant increase in micromotion values, the clinical relevance of this finding needs to be more elucidated.

In one study it has been shown that the particular modular neck concept evaluated in this experiment, had excellent survival in a 10 year clinical follow-up (Omlor et al., 2010). However, the Australian Joint Registry has reported a significant higher revision rate seven years after surgery for femoral stems with modular necks, compared to conventional THA (Australian Orthopaedic Association National Joint Replacement Registry, 2012) The most frequent reasons for revision according to the registry were loosening and osteolysis.

There are case reports on failure of the modular components (Skendzel et al., 2011; Sotereanos et al., 2013) and one case series reporting three patients with pseudotumours after surgery with THA and modular necks (Gill et al., 2012). These publications raise concerns of possible fretting and corrosion due to micromovements at the stem–neck junction of the modular neck. These aspects are not directly addressed in this study.

Table 2
P-values of the linear mixed model (LMM) analysis; comparing the varus, retro and short necks to the long neck; comparing stair climbing to single leg stance; comparing proximal level to the distal level. N = 12.

Comparisons	*P*-value
Long neck–varus neck	<0.001
Long neck–retro neck	0.001
Long neck–short neck	0.832
Single leg–stair climbing	<0.001
Distal level–proximal level	0.010

Table 3
Micromotion data for three translations and three rotations for the four modular necks showing median values and quartiles. Positive and negative values indicate the direction of the movements. The values represent the distal level in stair climbing activity. N = 12 for every comparison.

Movement	Short	25% 75%	Retro	25% 75%	Varus	25% 75%	Long	25% 75%
Posterior translation (μm)	7	−2 10	−8	−27 −3	6	1 25	1	−4 6
Lateral translation (μm)	26	8 56	38	11 70	49	20 78	30	6 62
Inferior translation (μm)	18	10 25	13	−23 30	16	7 24	14	11 23
Varus rotation (deg)	0.09	0.05 0.14	0.09	0.04 0.19	0.13	0.08 0.20	0.10	0.06 0.15
Posterior tilt (deg)	0.03	0.00 0.04	0.05	0.02 0.09	0.02	0.00 0.04	0.03	0.01 0.05
Retroversion (deg)	0.08	0.05 0.12	0.11	0.05 0.16	0.12	0.06 0.18	0.12	0.06 0.17

Table 4
Mean resultant hip joint forces with 95% confidence interval. N = 12.

Loading	Intact	(CI)	Short	(CI)	Varus	(CI)	Retro	(CI)	Long	(CI)
Single leg	2310	(2129, 2491)	2404	(2180, 2629)	2310	(2081, 2538)	2433	(2202, 2663)	2323	(2098, 2549)
Stair climbing	2366	(2165, 2567)	2473	(2239, 2706)	2365	(2132, 2597)	2500	(2259, 2741)	2360	(2132, 2588)

The lever arm can be of importance regarding the stem longevity and stability. For example it is indicated that a small size femoral cemented stem, combined with high offset could lead to increased risk of aseptic revisions (Thien and Karrholm, 2010). Moreover increased offset could increase the micromotion of the stem from tests with loads simulating stair climbing (Doehring et al., 1999). The effects of larger femoral offset were reflected in both an increased bending movement and increased torsional moment about the axis of the implant (Doehring et al., 1999). In the present study there was no statistically significant difference between the short and the long neck. It should be noted that when simulating stair climbing in our study, a constant individual torque was applied to the distal femur. The effect from increased femoral offset on the implant torsional moment was therefore not demonstrated.

In a previous study on a different stem it was found that neither increased femoral offset, reduced the neck shaft angle nor increased retroversion influenced the micromotions as compared to a standard version and neck shaft angle (Wik et al., 2010). This is in contrast to the present findings that show that the varus and retroverted necks gave higher micromotion values than the straight ones.

Our study showed that micromovement values for all necks are comparable with previous studies on femoral stems with good long term clinical outcomes (Ostbyhaug et al., 2010; Wik et al., 2010).

Ex vivo studies represent simplifications of the in vivo situation, and the results of such studies should therefore be interpreted with caution. The resultant forces given in Table 4 are approximately 40% higher than those recorded in previous studies from the same research group (Ostbyhaug et al., 2010; Wik et al., 2010, 2011) and forces from telemetric in vivo studies (Bergmann et al., 2001) but are within the range reported in earlier studies (Davey et al., 1993; Doehring et al., 1999). Excessive loading is rarely used in cadaver studies because of the increased risk of failures, and can probably explain the higher number of failures/fractures in this study compared to previous studies performed with the same setup. However, testing with insufficient loads can underestimate the micromovements of implants. In the present study the tested femoral stem showed adequate primary stability in spite of the high resultant force.

The mean resultant hip forces registered with the short and the retro necks were larger than with the long neck in this study, a finding which is in line with findings in previous studies and could be explained by the relation between increased offset and reduced abductor force (Davey et al., 1993). There is, however, no apparent correlation between alternations in resultant forces and recorded micromotion values for the various neck types. Thus implant stability seems to be more dependent on actual moment generated about the implant axis rather than the magnitude of the hip resultant force (Table 4).

Measuring micromotion is challenging as invasive measuring procedures could affect the micromechanical environment. The fact that the transducers (LVDTs) measure the displacement between two fixation points rather than the direct and local micromotion, may lead to an overestimation of the micromovements at the bone-implant junction (Gheduzzi and Miles, 2007). However, our protocol is previously thoroughly validated in this regard (Ostbyhaug et al., 2010). A direct measure of micromotion requires holes to be drilled through the cortical bone, which would induce a mechanical weakening. Also, assuming that the stiffness of the femoral stem is high compared to the bone, the measurement error is likely to be negligible.

The two measurement levels chosen in this study were both located at the proximal third of the stem, where one would expect press-fit and subsequent bone growth to the stem. The stems are fully coated, and a complementary measurement level could have been included. The femurs served as their own control thus errors related to differences between individual femurs are then minimised.

The total point movement of the femoral stem with the varus neck was 43% to 65% larger than the reference neck. The data varied between subjects, some exhibited micromovements multiple times the median values for the reference neck (outliers, Fig. 3). A multitude of clinically related factors such as osteoporosis, body mass index, surgical technique, implant design and geometry will affect the primary stability of an implant. Consequently there is a risk that the increased micromotions of the varus and retro necks in combination with the mentioned risk factors, could lead to micromovements that inhibit osseointegration and secondary stability.

5. Conclusion

The uncemented stem evaluated in this study showed adequate primary stability when coupled to modular necks with different lengths and versions. Even though the varus and retro necks showed higher micromotions than the long neck, the micromotion values are within currently accepted range, but risk factors of the individual should be considered.

Conflicts of interest and source of funding

There are no conflicts of interest.

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Deformation pattern and load transfer of an uncemented femoral stem with modular necks. An experimental study in human cadaver femurs

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ABSTRACT

Background: Modular necks in hip arthroplasty allow variations in neck-shaft angles, neck version and neck lengths and have been introduced to improve accuracy when reconstructing the anatomy and hip joint biomechanics. Periprosthetic bone resorption may be a consequence of stress shielding in the proximal femur after implantation of a femoral stem. The purpose of this study was to investigate the deformation pattern and load transfer of an uncemented femoral stem coupled to different modular necks in human cadaver femurs.
Methods: A cementless femoral stem was implanted in twelve human cadaver femurs and tested in a hip simulator corresponding to single leg stance and stair climbing activity with patient-specific loading. The stems were tested with four different modular necks; long, short, retro and varus. The long neck was used as reference in statistical comparisons, as it can be considered the “standard” neck. The deformation of bone during loading was measured by strain gauge rosettes at three levels of the proximal femur on the medial, lateral and anterior side.
Findings: The cortical strains were overall reduced on the medial and lateral side of femur, for all implants tested, and in both loading conditions compared to the unoperated femur. Although there were statistical significant differences between the necks, the results did not show a consistent pattern considering which neck retained or lost most strain. In general the differences were small, with the highest significant difference between the necks of 3.23 percentage points.
Interpretation: The small differences of strain between the modular necks tested in this study are not expected to influence bone remodeling in the proximal femur.

1. Introduction

Modularity is a well-known concept in revision arthroplasty, and the use of modular components in primary total hip arthroplasty (THA) has increased in recent years (Australian Orthopaedic Association National Joint Replacement Registry, 2014; National Joint Registry for England and Wales, 2014, The New Zealand Joint Registry, 2013)

Reconstruction of hip joint geometry is one of the goals in arthroplasty, but can be challenging, especially in cases of hip joint deformity. Leg length discrepancy or inadequate femoral offset, may lead to poorer clinical outcome for the patients (Kotwal et al., 2009; Lecerf et al., 2009).

The concept of modular necks allows for variations in neck-shaft angles, neck version and neck lengths in THA and can improve the anatomical relation and hip joint biomechanics (Krishnan et al., 2013).

There is limited long-term documentation on modular necks in primary THAs. There are reports of good mid-term results (Matsushita et al., 2010; Omlor et al., 2010), however, according to the Australian Joint Registry the revision rate of THA with exchangeable femoral necks is twice the revision rate of conventional THA 8 years after surgery, implant loosening being one of the primary reasons (Australian Orthopaedic Association National Joint Replacement Registry, 2014). In addition several case series and case reports have shed light over problems with modular necks, due to fretting, corrosion and pseudotumor formation (Dangles and Altstetter, 2010; Gill et al., 2012; Skendzel et al., 2011; Sotereanos et al., 2013; Viceconti et al., 1996, 1997) Pastides et al., 2013.

The human bone remodeling is a complex process, where the mechanical stimulus of the bone cells is an important factor (Engh et al., 2003; Glassman et al., 2006). The clinical observation of bone remodeling, usually referred to as Wolff's law, is that bone density increases when load increases, and decreases when load decreases. Periprosthetic

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bone resorption in the proximal femur is a well-known phenomenon after THA, and is commonly explained by adaptive bone remodeling due to stress-bypassing in the proximal femur. The phenomena is termed stress shielding, referring to that after implanting a stiffer femoral stem, the proximal femur is shielded or protected from loading (Glassman et al., 2006).

Stress shielding seems to be influenced by the fixation techniques, material properties and stem design, as well as patient-related factors. An alteration of the biomechanical environment and hence adaptive bone remodeling may lead to compromised support of the femoral stem and subsequent loosening of the prosthesis and complications during revision surgery.

There are a few experimental studies of deformation patterns and modular femoral necks in synthetic bones. These studies have used different angle, version and length in modular necks, looking at the pattern of load transfer in proximal synthetic femur after insertion of the implants (Politis et al., 2013; Umeda et al., 2003).

Human cadaver femurs have some advantages over synthetic bones in experimental set-ups, as they provide an expected natural variation in both geometry and material and are therefore more clinically relevant. However, they are not easy to obtain and must be handled with special care. To our knowledge there are no studies on modular necks recording cortical deformation in human cadaver femurs.

The purpose of this study was to evaluate the load transfer expressed by the cortical deformation pattern of an uncemented femoral stem with four different modular necks varying neck-version, neck-length and neck-shaft angle in human cadaver femurs.

2. Material and methods

2.1. Implant system

Four modular titanium necks with a 12/14 taper (Profemur® Modular Necks, Wright Medical Technology Inc, Arlington, TN USA 38002) were evaluated: 1. Straight long (PHAO 1204), 2. Straight short (PHAO 1202), 3. Retroverted short 15° (PHAO 1262) and 4. Varus short 15° (PHAO 1242) modular component (Fig. 1). The necks were connected into the femoral stem pocket through the oval end. A 28 mm femoral head was used for articulation. Cementless titanium alloy collarless stems fully coated with hydroxyapatite (HA) (Profemur® PRGLKITD Gladiator, Wright Medical Technology Inc, Arlington, TN USA 38002) were implanted, randomly allocated to right or left side. The implantations were performed by the same experienced orthopedic surgeon and according to the manufacturer's procedure (Wright Medical Technology, 2013).

Fig. 1. Profemur® modular necks: varus short 15°, retroversion short 15°, straight short and straight long (reference neck) in front view.

2.3. Human cadaver femurs

Caucasian human cadaver femurs were collected from deceased patients that underwent medical post-mortem examinations within 24 h. Consents from the relatives were obtained before interfering. The Regional Committee for Medical Research Ethics, Western Norway, approved the project. Twelve human femurs were tested, mean donor age was 58 years (range 43–70 years), nine males and three females. The same set of subjects was also used in a previous study (Enoksen et al., 2014).

The femurs were handled and prepared according to an earlier described and well documented procedure (Aamodt et al., 2001).

Two projections X-ray were used to estimate the size of the prosthesis and to exclude any skeletal pathology. Dual-energy X-ray absorptiometry (Lunar Prodigy Advance, General Electric Healthcare, California, USA) were obtained to diagnose any osteoporotic femurs. Bones with T-scores in the proximal femur below −2.5 were classified as osteoporotic and excluded.

The inclusion criteria of femurs were age ≤70 years in accordance with clinical practice at our department for uncemented stems. A body mass index ranging from 18 to 30 representing normal weight and to comply with the hip simulator, designed for normal size femurs and normal loading. Exclusion criteria were no previous fracture in the femur and no current or previous malignancy in the femur. A collection of twenty-one pairs of femurs was available. Five subjects failed during testing, three subjects were excluded due to osteoporosis and one pair was destroyed during preparation. Single femurs from twelve donors were therefore eligible for testing.

Before testing, the frontal plane of femur was defined by placing the femur on a horizontal surface resting on the posterior condyles and the greater trochanter. The anteversion of the femoral neck was measured and recorded for later orientation of the femur in the frontal and sagittal planes. The condyles were then resected and the femur was fixed into a steel cylinder with acrylic bone cement (Meliodent, Heraeus Kulzer GmbH, Hanau, Germany), aligning the center axis of femur with the center axis of the cylinder. The distance from the tip of the greater trochanter to the top of the cylinder was 25 cm for all specimens. Hip abductor muscles were simulated with a 40 mm polyamide strap attached to the greater trochanter using methacrylate glue (X 60, HBM GmbH, Darmstadt, Germany) and 6 cortical screws (Fig. 2).

2.4. Hip simulator

The hip simulator used in this study is well documented (Aamodt et al., 2001; Enoksen et al., 2014; Ostbyhaug et al., 2009; Wik et al., 2011). The operated femurs were mounted into a hip jig and loaded in a servohydraulic testing machine (MTS 858 MiniBionix II, MTS System Corporation, Eden Prairie, Minnesota, USA) (Fig. 2). The femur could rotate freely around its longitudinal axis and tilt freely in the medial/lateral plane, to avoid unphysiological bending moments.

The femur was tilted 12° into valgus, corresponding to physiological inclination during single leg stance (McLeish and Charnley, 1970). The femoral angle was kept the same for every test situation by adjusting the lower end of the cylinder, holding the femur. An acetabular cup with an inclination of 45° and 0° anteversion was used in this test set up. A trochanter strap was fixed to the lever arm to simulate the abductor muscles. The attachment of the strap to the lever arm was adjusted to achieve an angle of 15° to the load axis (McLeish and Charnley, 1970) in every test situation.

Single leg stance and stair climbing activities were tested. The femurs were loaded in the vertical axis proportionally to their individual

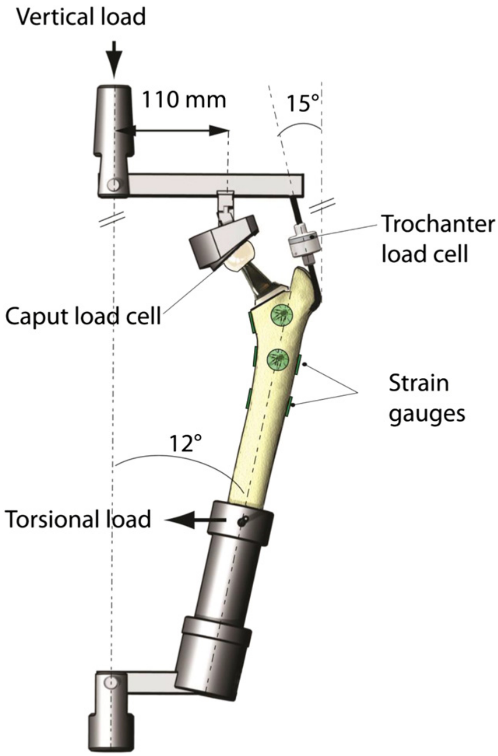

Fig. 2. Schematic illustration of the hip simulator.

donor bodyweights (BW) corresponding to 1.15 bodyweight. Each test consisted of 5 cycles with a consistent axial load. Adding a torque corresponding to 2.0% bodyweight meter simulated stair climbing (Bergmann et al., 2001). Torsional load was applied to the femoral head by pulleys and wire connected to a second actuator of the testing machine. Strain and micromotion measurements were obtained simultaneously throughout the testing sequence, and the micromotion results are previously published (Enoksen et al., 2014).

2.5. Strain measurement

Prewired rosettes with three strain gauges, mounted at 45° angles (FRA-3-23, Tokyo Sokki Kenkyujo, Tokyo, Japan), were used. The strain gauge outputs were recorded by a measurement amplifier (UPM 100, HBM, Darmstadt, Germany). Seven rosettes were distributed on the medial, anterior and lateral aspect of proximal femur, at three predefined levels, 14, 34 and 64 mm inferior to the femoral head (Fig. 2). These seven chosen positions correspond to the Gruen zones around proximal femur and the positioning and attachment of the strain gauges rosettes were performed according to an established procedure (Aamodt et al., 2001). In this study principal tensile strain was used for analysis of the deformation pattern on the lateral and the anterior aspect of the femur, whereas principal compressive strain was used for analysis of the medial aspect.

2.6. Statistics

Strain data was expressed as percentage of intact values. Each strain gauge rosette was analyzed separately, but measurements from different rosettes were considered to be dependent. A linear mixed model (LMM) was used to analyze the differences in strain measurements and accounted for repeated measurements and data dependency (Gueorguieva and Krystal, 2004).

Loading condition was used as a covariate in the analysis. Strain values from the unoperated intact femur and percentage strain values from the other six strain gauge locations were initially included as covariates. Statistically non-significant covariates were removed to define the most parsimonious model.

The four different necks (long, short, varus and retroversion) were modeled as fixed factors. The long neck was chosen as reference in the statistical analysis, as it is more used than the short neck (Omlor et al., 2010) and recommended by the manufacturer in order to avert skirted heads. The residuals were normally distributed, verified by Q–Q plots and histogram. The *P*-values were Bonferroni corrected to adjust for multiple comparisons by the statistical software. Level of significance was set to 0.05. All statistical analyses were performed using the software package IBM® SPSS Statistics version 21.

3. Results

The measured strain values at the seven strain gauge positions showed similar pattern for all the four necks tested (Figs. 3 and 4), however, there were some small differences. The median principal strain values ranged from − 1733 μm/m to 1672 μm/m among the operated femurs (Table 1). The strain values after implantation of an implant were related to the strain values of the unoperated femur and presented as percentage of intact values. We thus found that all necks retained more strain than the reference neck at the Blat position. The short neck showed higher loss of strain at strain gauge position Clat, and finally the retroverted neck retained more strain at Amed. The largest difference was 3.23 percentage points (Table 2).

The highest loss of strain compared to the unoperated femur was observed at strain gauge position Amed proximally on the medial side, with strain values ranging from 13.6% (long) to 14.7% (retro) at single leg stance of the corresponding unoperated strain values. The strain gradually increased distally up to an average of 66.3% of intact strain at Cmed, for both loading conditions (Figs. 3 and 4). Laterally, the average strain values were quite similar, regardless of level and loading condition, ranging from 76.9% to 77.9%. On the anterior side there was a pronounced difference between loading conditions, with an average of 130.1% (single leg) and 97.2% (stair climbing) at Aant. Corresponding values for the most distal strain gauge Bant were 128,2% and 92.5%.

The cortical strains were overall reduced on the medial and lateral side of femur, for all implants tested and in both loading conditions compared to the unoperated femur (Figs. 3 and 4).

4. Discussion

In this experimental study we assessed the strain distribution to evaluate the load transfer in the proximal femur following insertion of a cementless femoral stem coupled to modular necks of different version, angle and length.

Periprosthetic bone resorption around femoral stems is a known phenomenon, and can be related to an alteration of deformation pattern in the cortical bone as this represents the load pattern of the femur. In this study absolute cortical strains on the medial and lateral side were reduced in the proximal femur for all combinations tested compared with the unoperated femur. The largest decrease was found in the calcar region for all necks. The differences between the neck combinations tested were overall small, with the highest difference of 3.23 percentage points between the short and the reference neck at Blat position. The effects from the small strain differences are probably not clinically relevant, especially when considering the magnitude of strain loss observed from intact to operated femur. Decreased strain in proximal femur following hip surgery is a result from altered load transfer, also called stress shielding. Clinically, this stress shielding effect is associated

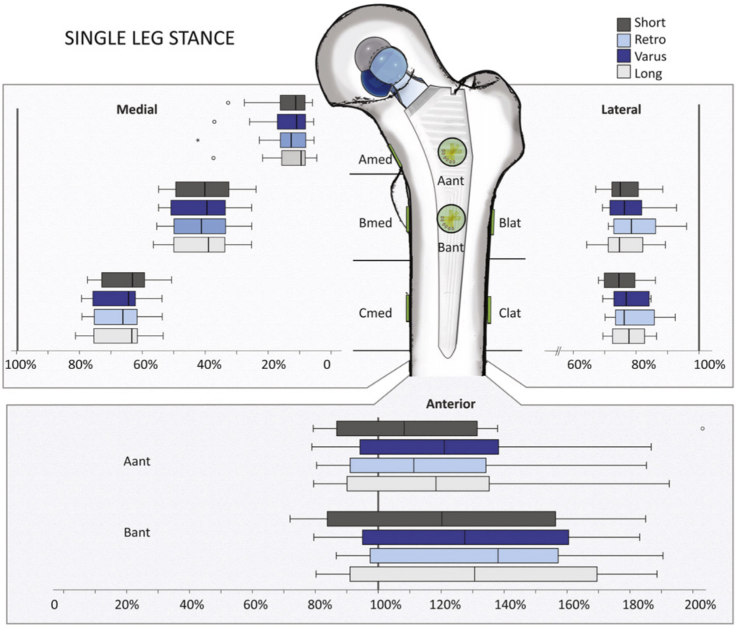

Fig. 3. The box-plot shows descriptive strain values in percentage of unoperated femur in four different modular necks in single leg stance at seven locations in proximal femur (n = 12). The boxes represent interquartile ranges with the median as a vertical line. The whiskers extend to the minimum or maximum value within 1.5 times the width of the boxes.

with proximal bone resorption. Preservation of the proximal bone stock is considered important, but the magnitude of strain required to preserve sufficient bone remodeling in the femur is not known. There is however an agreement that there exists a relationship between mechanical load and adaptive bone remodeling (Lecerf et al., 2009; Sibonga et al., 2007). A review concerning femoral designs and stress

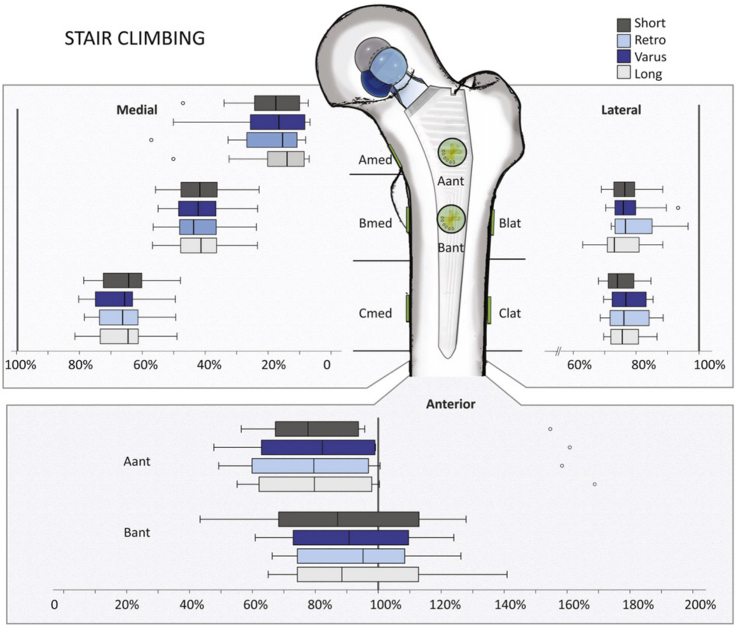

Fig. 4. The box-plot shows descriptive strain values in percentage of unoperated femur in four different modular necks in stair climbing at seven locations in proximal femur. The boxes represent interquartile ranges with the median as a vertical line. The whiskers extend to the minimum or maximum value within 1.5 times the width of the boxes.

Table 1
Median two loading activities, seven different strain gauge positions, and four modular necks.

	Single leg stance					Stair climbing				
Location	Intact	Short	Varus	Retro	Long	Intact	Short	Varus	Retro	Long
Amed	−1962	−214	−203	−222	−201	−1952	−325	−320	−312	−277
Bmed	−2269	−937	−955	−1013	−966	−2154	−963	−967	−978	−944
Cmed	−2460	−1668	−1703	−1712	−1675	−2469	−1708	−1733	−1720	−1699
Aant	713	772	811	788	738	959	671	719	711	655
Bant	548	589	631	653	641	836	739	769	763	760
Blat	2065	1608	1643	1672	1619	2046	1617	1625	1662	1585
Clat	2036	1485	1546	1590	1529	2007	1525	1578	1599	1548

shielding states that progressive bone loss through stress shielding has potentially critical consequences, and that preservation of the femoral bone stock is important (Glassman et al., 2006).

In addition, new design features of the femoral stem intend to reduce postoperative bone loss, thus addressing the stress shielding as a problem (Falez et al., 2015; van Oldenrijk et al., 2014).

A similar designed study (Politis et al., 2013), measured patterns of stress in proximal femur after implantation of cementless stems with modular necks in three synthetic bones. Anteverted neck combinations showed higher stress at the anterior surface, retroverted had increased stress on the posterior side of proximal femur and the varus neck showed increased compressive stress in the calcar region (Politis et al., 2013). A comparable study (Umeda et al., 2003), also reported a correlation between compressive strain on the side toward which the prosthetic neck was oriented and the extent of neck version (Umeda et al., 2003). The results from these studies are not directly comparable to ours, as we did not compare the same neck combinations.

We found statistically significant differences between the varus neck and the reference neck at one location, and at two locations for each of the two other necks. Overall, the differences were small, but the strain pattern we registered was similar to findings in a previous in vitro study (Wik et al., 2011) on modular experimental heads where the femoral neck angle and length could be adjusted.

Measurement side and measurement level were the most dominant predictors of strain loss, whereas the type of neck had a limited influence on the magnitude of strain conservation.

It can be discussed which areas of the femurs that are of most interest regarding strain. We attached strain gauges at previously chosen locations (Aamodt et al., 2001; Ostbyhaug et al., 2009; Wik et al., 2011) which were based on the definition of the seven Gruen zones.

In a previous study on the same modular necks we showed that the resultant forces and the average micromotions were within normal range (Enoksen et al., 2014) but the varus and retroverted necks showed small but statistically significant larger micromotion values than the long neck.

The choice of human cadaver femora as a test model represents a strength in experimental research, where the natural variety in anatomical geometry and bone quality is reflected among the test specimens. However, the use of cadaver bones in preference to synthetic bones may lead to more frequent testing failures. To reduce this problem, we developed strict exclusion criteria before testing, and none of the tested femurs had bone-related pathology.

The bones were loaded in two different human activities modes with subject-specific loading. The majority of laboratory studies use a standardized loading condition of the femurs. To our knowledge there are no other studies describing patient specific loading and we consider this loading to be more relevant when testing realistic deformation patterns in experimental studies.

Different hip simulator designs, loading conditions and choice of test specimens complicate direct comparisons between studies. It is shown that simplified set-ups can provide similar hip resultant forces (Basso et al., 2014). However, simulation of major muscle forces influences cortical deformation significantly (Duda et al., 1998). The strength of our study is the set up, including that cadaver femurs were tested with physiologically relevant forces defined by individual bodyweights. We also simulated both single leg stance and stair climbing with combined axial and torsional forces.

Ex vivo studies represent simplifications of the in vivo situation, and the results of such studies should therefore be interpreted with caution. The main limitation of an experimental study such as the present is that bone remodeling is an ongoing process that continue after bone ingrowth. The situation simulated in this study reflects only the immediate post-operative condition. Another limitation is that strain is measured at only the predefined points, whereas strain is continuously distributed along the proximal femur. The experimental set up used in this study is, however, well documented in previous studies (Aamodt et al., 2001; Ostbyhaug et al., 2009; Ostbyhaug et al., 2013; Wik et al., 2010, 2011). The set-up in these studies is basically similar, and they have all showed realistic resultant hip joint forces.

Clinical results on modular necks are not consistent (Australian Orthopaedic Association National Joint Replacement Registry, 2014, Omlor et al., 2010). There are several case report reporting failures, among others fatigue fractures of the necks due to corrosion and fretting (Dangles and Altstetter, 2010; Gill et al., 2012; Pastides et al., 2013; Skendzel et al., 2011; Sotereanos et al., 2013; Viceconti et al., 1996, 1997). The question asked in these studies as to why these failures occur is not addressed by the present study.

Nevertheless, the results of this study showed only small differences between modular necks varying length, version and necks shaft angle. These results indicate that one should not expect difference in the bone remodeling in the proximal femur related to the use of different modular necks.

Table 2
Differences in percentage of intact strain between the different necks with corresponding *P*-values. The values represent overall model estimates for the loading conditions adjusted for covariates.

	Short-long		Varus-long		Retro-long	
Location	Diff	*P*-value	Diff	*P*-value	Diff	*P*-value
Amed	−0.09	1.000	0.02	1.000	2.17	0.022
Bmed	0.50	0.829	−0.28	1.000	0.80	0.206
Cmed	−0.27	1.000	0.36	0.527	−0.52	0.187
Aant	9.54	0.580	2.95	1.000	0.63	1.000
Bant	−0.60	1.000	−1.25	1.000	5.22	0.617
Blat	3.23	<0.001	1.39	0.027	2.51	<0.001
Clat	−1.43	<0.001	−0.44	0.334	−0.21	1.000

Conflicts of interest and source of funding

There are no conflicts of interest.

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Erratum

Erratum

In Material and methods:

Page 32, chapter 9.3, line 7: “(Table 1)” – “(Table 1)” should be deleted

Page 32, chapter 9.4.1, line 3: “(Figure 9)” – should be “(Figure 11)”

Page 33, chapter 9.4.1, line 9: “(Figure 10)” – should be “(Figure 11)”

Page 34, chapter 9.4.3, line 4: “(table 2)” – “(table 2)” should be deleted

Page 34, chapter 9.4.3, line 5: “(125, 127, 129)” – should be “(125, 127, 129) (Table 2)”

Page 36, chapter 9.4.4, line 2: “(Figure 10)” – should be “(Figure 12)”