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Press-fit stability of cementless femoral knee implants-
a biomechanical evaluation

Sanaz Berahmani
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Press-fit stability of cementless femoral knee implants—a biomechanical evaluation

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CHAPTER ONE

Introduction
1. Background

Total knee replacement (TKR) is the most accepted treatment in patients with advanced knee osteoarthritis (OA) or rheumatoid arthritis. Both diseases lead to degeneration of articular cartilage and subsequently the bony surfaces. These damages cause severe pain, loss of knee mobility, and remarkable reduction of quality of life [1]. The main goals of TKR are alleviation of pain and restoration of normal knee function and kinematics [2, 3]. During TKR the degenerated knee surfaces of the distal femur and proximal tibia are replaced by femoral and tibial prosthetic components (Figure 1.1). Knee articulation is restored by sliding of a metallic femoral knee component on the bearing surface of a plastic tibial insert. Patellar resurfacing using a patellar button in TKR is performed optionally depending on a surgeon’s preference.

Figure 1.1 A knee joint with total knee replacement. The different bony and prosthetic parts are identified.

A rather high survival rate of TKR ranging between 90 and 95% after ten years post-operatively has been reported [4-6]. This effectiveness is reflected in the growing number of operations being performed. The prevalence of TKR is projected to increase by 306% and 117% between 2012 and 2030 in the Unites States and England, respectively [7, 8]. Despite the promising results, revision TKR is still a major burden for patients, surgeons and the healthcare system mainly due to the large total number of failures [7-9]. In addition, revision is an undesirable surgery due to the complexity, the lower success rate, and high cost [10]. Recent studies have shown that the incidence of failure of TKR is significantly higher for younger patients [11, 12]. The last annual report of Australian registry showed that after three years of surgery, younger patients (<55 years old) have over four times higher revision
rates as compared to patients older than 75 years old [13]. Therefore, it is vital to understand, postpone or even avoid the revision TKR, especially for younger patients.

The etiology of TKR failure has been investigated extensively [14-17]. Aseptic loosening has been identified as one of the major indications for revision TKR in the literature [17, 18]. Different theories have been introduced about the causes of aseptic loosening [17] and it is almost certain that implant fixation method has an important role in the case of aseptic loosening of the implant [17, 19, 20]. The two widely known fixation methods are cemented and press-fit; the latter is also known as uncemented or cementless. Cemented fixation using polymethylmethacrylate (PMMA), or bone cement, is the most conventional and also the most popular technique [21-23]. However, to avoid foreign body reaction to cement particles [19], press-fit implants were developed with an eye towards younger and more active patients. This method of fixation relies on a biological bond by osseointegration at the bone-implant interface over a time span of a couple of months [24, 25]. The focus of the current thesis is on uncemented TKR and the following sections further describe this fixation method.

2. Uncemented TKR

Nowadays, uncemented TKR is not the first choice by most surgeons [23] and even went through a descending trend of popularity in the last decade [26]. Apart from higher costs of the uncemented implant, controversial reports on its survival rate are known as the main reason for its discredit. A meta-analysis by Gandhi et al. has shown a significant combined odds ratio of 4.2 (with a 95% confidence interval (CI) of 2.7- 6.5) for the failure of uncemented TKR including both randomized controlled trials and observational studies from 1950 to 2008 [27]. In the literature, the issue of high failure rates is mostly reported concerning the tibial component [27, 28], although this is not always specified clearly [29]. Chockalingam et al. reported 9.8% of loosening of uncemented versus 0.6% of cemented femoral knee implant at 6 years [30]. Hence, it seems that both femoral and tibial components are prone to early loosening.

Despite the discussions concerning the higher failure rates of uncemented TKR, many surgeons and researchers are focusing on this type of fixation as it is more appealing to offer a technique that generates a biological bond at the interface instead of fixation with a foreign material that can only degenerate in time. Cementless fixation, in principle, offers a direct attachment of the implant to the underlying bone via bone ingrowth in- and onto the internal surface of implant, and thereby should be able to provide a life-long fixation with no further risk of bone cement particle debris or fracturing [31].

It is widely accepted that adequate primary stability of uncemented implants is the key factor to initiate and achieve osseointegration [32-35]. Hence, to prevent aseptic loosening of the implant, it is necessary to focus on obtaining an adequate primary stability. The current
thesis, therefore, aimed to identify, understand and examine different design and patient-related factors which can affect the primary stability of an uncemented femoral knee implant. Accordingly, primary stability is first defined, after which effective factors are thoroughly explained.

2.1 Primary stability

During press-fit implantation, the primary (mechanical) stability is achieved by creation of frictional forces at the bone-implant interface, in which the frictional characteristics of the implant surface play an important role [36]. In addition, the bone is prepared to be oversized with respect to the inside of the implant. During implantation, compressive forces are generated at the interface which provide the primary stability of the implant. Primary stability can be expressed as the magnitude of motions, or micromotions, occurring at the bone-implant interface upon loading of the implant [37, 38]. In case of a good primary fixation, after a couple of weeks to months, bone will grow in- and onto the implant surface, providing secondary stability to the reconstruction [38, 39]. Several animal studies have shown that bone ingrowth occurs when micromotions at the interface remain below 40 to 50 μm [40, 41], while micromotions larger than 150 μm would result in the formation of fibrous connective tissue at the interface, compromising the fixation [38]. In vivo measurement of implant motion relative to the bone is possible using roentgen stereophotogrammetric analysis (RSA) [42], although it can only measure a larger scale, making it more suitable to measure migration than micromotions [43].

With every introduction of new implant designs, materials, or surface treatments, the primary stability of the implant is altered. It is therefore of extreme importance to extensively evaluate the effect of implant modification in the primary stability. For this purpose, several in vitro and in silico approaches have been developed to evaluate the micromotions at the bone-implant interface of press-fit implants [44-47]. However, due to the complexity of the bone-implant system in terms of materials and frictional contact properties, plus the fact that patient related factors (such as physical activities and bone quality) are highly variable, there are still many unknowns about the effects of design and patient-related factors on the primary stability of uncemented femoral knee implants, which are therefore challenging to incorporate in standardized pre-clinical tests.

2.2 Design-related factors

2.2.1 Surface morphology. The surface morphology of an orthopaedic implant plays a key role in the osseointegration process [48], the stress shielding effect [49], and shear-load bearing capacity [50]. A porous surface structure can potentially improve upon all of these parameters. Hence, different porous surface morphologies have been developed over the
last decades [51] (Figure. 1.2). It is known that pore size and volume fraction can promote bone formation by increasing contact surface and providing space for the perivascular tissues [39, 52, 53]. However, there is no consensus upon the optimum porosity; pore sizes ranging from 45 to 400 μm have been shown to provide an acceptable strength [53, 54]. Titanium and cobalt-chrome alloys are widely used due to their excellent osteoconductivity and biocompatibility [54]. The higher stiffness of solid metallic implants increases the risk of stress shielding, which can be prevented by applying a porous structure with a much lower stiffness [49]. Finally, the frictional capacity, which is defined by the roughness properties, can affect the primary stability of uncemented femoral knee implant, which will be discussed thoroughly in the current thesis (Chapters 2 and 3).

Figure. 1.2 A) Internal surface of femoral knee implant. B-D) Three different surface morphologies that can be applied to knee implants.

2.2.2 Press-fit interference. During uncemented femoral knee arthroplasty, a press-fit fixation is achieved by resecting the distal femur slightly larger than the inner dimensions of the implant. The oversize of the bone with respect to the implant is referred to as the interference fit. Press-fit interference is an important factor in the primary stability of uncemented implants, as it causes compressive forces at the interface. Applying an interference fit that is too large would endanger the implantation by increasing the risk of intra-operative bone fracture [55]; with an interference fit that is too small gaps will be created at the implant-bone interface, leading to inadequate stability. It is obvious that precise bone cuts are necessary to obtain an optimal interference fit. Accurate bone cuts are also important for optimal femoral knee implant alignment and positioning [56]. Computer navigation assisted TKR is aimed to improve the accuracy of bone resections and it is also used as a tool to measure cutting planes before and after bone resection [57]. However, the reported results, which are mainly
rotational deviations in coronal and sagittal planes, are only useful for the assessment of implant alignment, but provide little information on the appropriate range of interference fit. In chapter 6 of this thesis, different optical techniques were used to assess the geometry and to evaluate the press-fit interference of an uncemented femoral knee implant in vitro.

2.3 Patient-related factors

2.3.1 Bone. Bone response to an uncemented implant is not only limited to its osseointegration properties, but also the mechanical interaction of bone and implant during and after implantation is of great importance. Meneghini et al. found a significantly lower primary stability of uncemented tibial components in osteoporotic bone [58]. Aro et al. compared femoral stem subsidence three months postoperatively between patients with normal and osteoporotic bone, and showed a significantly higher subsidence for the latter group [59]. There are reports about a high prevalence of osteopenia and osteoporosis in patients with advanced OA who will receive a joint replacement [60], which emphasizes the necessity of investigating the effect of bone quality on the primary stability.

Bone tissue consists of the minerals calcium and phosphate, collagen, and water, and has anisotropic and visco-elasto-plastic material properties [61]. The bone elastic properties are essential to provide primary stability to knee implants, by storing strain energy. Several studies have demonstrated the variation of the materials stiffness over different anatomic locations [62, 63]. These elastic properties correlate with the bone mineral density (BMD), and can be represented through various linear and power relationships [64]. Hence, BMD serves as a measure for the local bone stiffness, which again underlines its effect on the primary stability. Accordingly, in almost all chapters of this thesis BMD was quantified and its effect on the primary stability was established.

Bone anisotropy, which originates from the orientation of trabeculae within the cancellous bone and collagen fiber orientation in cortical bone, indicates that the bone behavior depends on the direction of the applied load (Figure. 1.2). Bone has the maximum strength under loads parallel to the principal direction of its microstructure. Burgers et al. investigated mechanical properties of bone that are relevant to the press-fit situation in the distal femur around a cementless implant [65]. Their study showed a difference in stiffness and yield strength between more proximal and distal regions of the distal femur, which was related to bone anisotropy.

The viscoelastic properties of bone also affect the primary stability, as it reduces the pre-stresses created at the interface over time. Shultz et al. predicted that the stresses at the interface can relax as large as 50% after implantation [66]. Burgers et al. has shown that bone viscous behavior depends on the BMD [67]. Hence, several chapters of this thesis focused on the bone viscoelastic behaviour.
Several studies have demonstrated that during implantation of knee component bone stresses are caused that exceed the yield strength [68, 69]. Bishop et al. quantified the amount of plastic deformation by mimicking the clinical press-fit condition using cubes of bone [70]. They found a clear effect of press-fit interference and surface morphology on the permanent bone deformation. Therefore, chapters 6 and 7 of this thesis investigated this phenomenon in detail.

![Figure 1.2 The orientation of trabecula within cancellous structure of proximal femur.](image)

### 2.3.2 Activity and loading.

The knee joint experiences different loading profiles due to the variety of physical activities of the patient, such as walking (gait), kneeling, and stair ascending and descending. During each activity a different loading profile is applied to the femoral and tibial components. Kutzner et al. measured these *in vivo* forces and moments for different activities using an instrumented knee implant [71]. These data are extremely valuable when assessing the biomechanical conditions of knee implants. They also reported that measured forces were much higher than traditionally applied when testing knee implants [71].

The current generation of TKR patients is younger and more active than in the past, which entails higher demands on the TKR systems and their fixation. For instance, several publications have reported a higher incidence of early loosening in the high-flex femoral knee implants [72, 73]. It is therefore essential to evaluate the primary stability of TKR systems under a variety of severe loading conditions. Taylor et al. found the highest micromotions in a cementless tibial component with stair ascent and descent [47]. Chong et al. also reported a clear effect of loading conditions on the micromotion of tibial component [74]. As the current thesis focuses on the femoral component, the effect of the type of loading condition on the primary stability was investigated in the chapters 4, 5 and 6.
3. Overview

This thesis focuses on issues related to the primary stability of uncemented femoral knee implants. Various factors were investigated and their effects on the primary stability of press-fit implants were evaluated.

Chapter 2 aims to understand the biomechanical effects of press-fit interference, surface morphology, bone material properties, and their interaction on the primary fixation strength of the press-fit pegs.

Chapter 3 concerns mainly the effect of surface morphology on the primary fixation strength of an uncemented femoral knee implant. In addition, the required implantation force and the effect of BMD on the fixation strength are discussed. Secondary, the issue of femoral knee implant alignment was investigated due to the potential effect of surface morphology on improper implant seating.

In chapter 4, a cadaver-specific FE model of distal femur was developed to investigate the effect of activity on the micromotion of uncemented femoral knee implants. Implant-specific loading profiles of gait and deep knee bend (DKB) were provided by the Denver University. Additionally, FEA was used as a complementary guidance tool to explore the effect of simplifications in loading conditions during the experimental assessment of micromotion.

Chapter 5 concerns the micromotions of two uncemented femoral knee implants under two loading conditions resembling the peak tibiofemoral force of gait and deep knee bend (DKB). In addition, implant-specific loading profiles were adopted for both implants to ensure a fair comparison between the designs.

Chapter 6 concentrates on the issue of interference fit, cutting errors, and bone damage of an uncemented femoral knee implant using different state-of-the-art scanning techniques.

Finally, in chapter 7 an advanced bone material model, incorporating bone plastic behavior, was assigned to five cadaver-specific FE models. The experimental micromotion at the bone-implant interface was compared with the predicted micromotions by the advanced model and standard linear elastic material models. Furthermore, experimental and FE-predicted bone damage at the bone-implant interface was compared.

Chapter 8 discusses the major findings of current thesis and then, makes a link between the presented results and clinical practice. Weaknesses and strengths of testing methods used in this thesis are highlighted. Finally, future perspectives relative to the findings in this thesis are outlined.

Chapters 9 and 10 are the English and Dutch summaries of current thesis, respectively.
References

[14] Mulhall KJ, Ghomrawi HM, Seully S, Callaghan JJ, Saleh KJ. Current etiologies...
[29] Fehring TK, Odum S, Griffin WL, Mason JB, Nadaud M. Early failures in total knee
[42] Gao F, Henricson A, Nilsson KG. Cemented versus uncemented fixation of the femoral


CHAPTER 1


CHAPTER TWO

An experimental study to investigate biomechanical aspects of the initial stability of press-fit implants


J Mech Behav Biomed Mater, 42, 177-185
Abstract

Initial fixation of press-fit implants depends on interference fit, surface morphology, and bone material properties. To understand the biomechanical effect of each factor and their interactions, the pull-out strength of seven types of CoCrMo tapered implants, with four different interference fits, three different surface morphologies (low, medium and high roughness), and at two time points (0 and 30min) were tested in trabecular bone with varying density. The effect of interference fit on pull-out strength depended on the surface morphology and time. In contrast with our expectations, samples with a higher roughness had a lower pull-out strength. We found a similar magnitude of bone damage for the different surface morphologies, but the type of damage was different, with bone compaction versus bone abrasion for low and high frictional surfaces, respectively. This explains a reduced sensitivity of fixation strength to bone mineral density in the latter group. In addition, a reduction in fixation strength after a waiting period only occurred for the low frictional specimens. Our study demonstrates that it is essential to evaluate the interplay between different factors and emphasizes the importance of testing in natural bone in order to optimize the initial stability of press-fit implants.
1. Introduction

Press-fit implants have been introduced to orthopedics and dentistry as an alternative fixation for conventional techniques using bone-cement [1-3]. The immediate post-operative fixation of press-fit implants is provided by a mechanical interaction between bone stresses created during implantation and frictional properties at the implant-bone interface [4]. On the longer term, a complex biological process leads to osseointegration or bony ongrowth [5] providing so-called secondary fixation. It has been shown that ‘adequate initial stability’ is a prerequisite to achieve secondary fixation [2, 6]. The definition of ‘adequate initial stability’ is unclear, although it obviously is associated with the magnitude of cyclic motions at the bone-implant interface [7, 8] and the required force or torque to insert or remove the implant [9, 10]. Several studies have investigated parameters that affect the initial stability, such as surgical technique [11], implant design [9, 12], and patient-related factors [13].

From a surgical point of view, an important aspect is the size of the cavity relative to the size of the implant. By creating a cavity which is slightly smaller than the implant, the surgeon generates an interference fit, thereby stabilizing the implant [9, 10]. It is a common belief that a larger interference fit increases the fixation. However, when the interference fit is too large, the surgeon may damage the bone while introducing the implant, thereby jeopardizing the stability [14]. Hence, a sensible range of interference fit needs to be determined for each implant design.

From an implant design point of view, initial stability is affected by the shape [9, 15] and the surface treatment [10, 16]. Implant surface treatment has an important effect on both short- and long-term fixation strength. Initial stability depends on the frictional properties, defined by the roughness of the surface [17]. On the longer term, rougher and more open structures provide an additional surface area for osseointegration, which has been studied extensively [16, 18]. However, the reported effects of surface treatment on initial stability are contradictory. Shirazi Adl et al. [10] found a negative effect of a rough surface on the fixation, while Elias et al. [9] reported an improved initial stability for a rough dental implant design. Bone quality is a patient-related factor that cannot be controlled, but has to be coped with during the surgery. Inferior bone quality has been shown to be associated with a negative effect on the long-term outcome of the implantation [19, 20]. Meneghini et al. [13] found a reduced initial stability of orthopedic implants in osteoporotic bone. It has been shown that the bone quality also plays important role in the fixation of dental implants [9, 21]. Hence, it is essential to include the bone quality in the examination of press-fit implants, as a potential confounding factor.

Viscoelastic behavior of the bone [22-25] may further affect the initial stability. Relaxation of the bone may, in a relatively short time period, reduce the press-fit forces acting on the implant, jeopardizing the initial fixation before adequate bone ingrowth can occur. Norman
et al. [26] found a significant decline in fixation strength of a hip stem due to the bone stress relaxation. However, the effect of interaction between interference fit, surface finish, and viscoelastic response on the fixation strength is still not fully understood. Hence, to further optimize implant fixation, not only is it important to determine the role of each factor separately, but also to understand the interactions between these factors. Therefore, in the current laboratory study, we analyzed press-fit metal implants inserted into the distal femur, and assessed the effects of interference fit, surface morphology, bone material properties, and their interactions on the pull-out strength.

2. Materials and Methods

2.1 Study design

CoCrMo press-fit implants in the shape of cylindrical tapered pegs (taper angle: 1.72°), which is a standard design in femoral knee implants (DePuy Synthes Joint Reconstruction, Leeds, UK), were used for the current study. Pegs were used to maximize the control over the parameters due to their relatively simple geometry. To investigate the effect of each individual factor and the interaction between factors, a design of experiment (DOE) approach was chosen. The design included three main factors: interference fit (diameter), surface morphology (friction coefficient), and bone relaxation (time). In addition, bone mineral density (BMD) was included as a covariate.

2.1.1 Interference fit. Four diametrical interference fits were tested by using different peg diameters (diameter at the base: 6.2, 6.5, 7.3 and 7.6 mm), which were all implanted in holes with a 6 mm diameter, leading to diametrical interference fits of 0.2, 0.5, 1.3 and 1.6 mm at the base (Table 2.1).

2.1.2 Surface treatment. Three groups with a different surface treatment (Figure. 2.1) that had either a grit-blasted (one group) or rough porous surface (two groups) were included. The grit-blasted surface, referred to in this study as the “low friction” group (µ= 0.5), had a surface roughness of $R_s = 0.3 \, \mu\text{m}$. The first porous “rough” group with medium friction (µ= 0.95) had a so-called Porocoat® porous coating that is commercially available (DePuy Synthes Joint Reconstruction, Warsaw, IN, USA). Porocoat® is comprised of randomly arranged spherical beads. The third testing group was a novel experimental surface morphology with a very high friction coefficient (µ= 1.4). It is comprised of a random arrangement of highly irregular particles based on spherical beads, which form interconnected pores (DePuy Synthes Joint Reconstruction, Warsaw, IN, USA). The rough porous surface morphology groups were produced by sintering CoCrMo alloy powder. Further specifications of surface morphologies are given in Table 2.1.
To define an optimal DOE, in order to answer our research question, low and high frictional groups were selected and tested for the low (0.5 mm) and high (1.6 mm) interference fits. In addition, high frictional specimens with the novel experimental surface morphology were tested for interference fits of 0.2 and 1.3 mm to determine the relationship between interference fit and pull-out force. The medium frictional group, which had a design similar to a conventional peg of the cementless LCS® femoral knee implant (DePuy Synthes Joint Reconstruction, Leeds, UK) was used as a clinical benchmark. It was tested only for a high interference fit, and for a longer time period (Table 2.1). In total, thirteen combinations of conditions, using seven peg designs were defined (10 samples per design).

2.1.3 Bone material properties

2.1.3.1 Bone relaxation. The effect of bone relaxation was examined by performing the pull-out test in two different time groups. In time group T0 the pegs were pulled out immediately after insertion, while in group T30 the pull-out test was performed after a waiting period of thirty minutes (Table 2.1).
2.1.3.2 Bone mineral density: After drilling the holes, but prior to peg insertion, the bones were CT-scanned inside a water basin that was placed on top of a calibration phantom with known densities (0, 50, 100, 200 mg/ml calcium hydroxyapatite, Image Analysis, Columbia, KY). To avoid air artifacts on the CT-images, the specimens were placed under a vacuum while submerged in water before scanning. The following settings were used for the CT-scan (Siemens Somatom Sensation 64, Siemens AG, Germany): 120 kV, 200 mA and slice thickness of 0.6 mm with no overlap, which has been shown as an accurate technique to quantify the BMD [27]. The smallest possible field of view was assigned to increase in-plane resolution (circa 300 µm). In-house Dicom software was used to convert Hounsfield Units to calibrated BMD values [28]. The region of interest (ROI) to measure relevant BMD values was based on a finite element study of a pull-out test using similar peg designs while inserted in to PU foam with known material and frictional properties. It was observed that the Von Mises stresses as a result of the press-fit forces decayed about 2.5 mm away from the circumference of the hole. Hence, the BMD was measured over a volume with a 2.5 mm thickness surrounding the drill hole, along the length of the peg (Figure. 2.1). To distribute BMD uniformly amongst the thirteen conditions, a randomized block design (with BMD as a grouping factor) was used. Subsequently 10 BMD blocks corresponding to sample size were defined, after which the holes were randomly assigned to the implants. In this manner, poor, medium and high bone qualities were equally allocated to each design.

<table>
<thead>
<tr>
<th>Surface group</th>
<th>Specifications of surface</th>
<th>Experimental parameters</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Coefficient of friction (COF)</td>
<td>Porosity [%]</td>
</tr>
<tr>
<td>1</td>
<td>0.5* (Low)</td>
<td>0</td>
</tr>
<tr>
<td>2</td>
<td>0.95 (Medium)</td>
<td>44</td>
</tr>
<tr>
<td>3</td>
<td>1.4 (High)</td>
<td>60</td>
</tr>
</tbody>
</table>

* This value is obtained from Shirazi Adl et al. [17] and the remainder were measured by the manufacturer.

2.2 Specimen Preparation

Twenty-four human femurs (6 pairs, 7 male and 11 female, Age: 83± 5.7 years old (mean±
SD)) were obtained from the anatomy department of the Radboud university medical center. There was no data available about bone pathology. Visual inspection and CT scans revealed no signs of bone disease or damage that would interfere with our study.

The specimens were cut 150 mm above the distal end of the femur. The proximal end of the specimens was cemented using polymethyl-methacrylate in a metal cylinder. The bones were fixed during cementation by a clamp in order to align the specimens in an anatomical orientation (Figure. 2.1). Next, an electric diamond-blade cutting machine was used to make a transverse cut 10 mm from the distal end of the femur, in accordance with the manufacturer’s instructions, to reach the underlying trabecular bone. In each bone specimen, six holes were drilled 10 mm from each other, to have neither overlap in ROI of the BMD measurement, nor with the cortex (Figure. 2.1). The holes were longer than peg length to avoid any contact between the tip of the peg and the bottom of the hole. After drilling, calibrated high resolution images were taken of each hole to correct for any variance in the size of drilled holes. The drill size was then quantified using Image J (National Institutes of Health, USA) by drawing a contour around each hole and measuring the circumscribed area.

2.3 Test set-up

The bone specimens were fixed at the proximal end to a servo-hydraulic MTS machine table (Figure. 2.1). The peg was connected to a free-sliding jig (LWR 6150, SKF, Germany) by using a pin-clevis mechanism. The free-sliding jig was connected to 1 KN load-cell, which was attached to the MTS actuator (MTS Systems Corporation, Eden Prairie, Minnesota, USA). The jig ensured free movement in the horizontal plane, thereby avoiding artificial damage to the hole during the test. The pegs were leveled before each insertion test to avoid a load bearing contact area, after which it was inserted for a depth equal to its length at a displacement rate of 3 mm/sec.

2.4 Statistical analysis

A one-way ANOVA was used to check for uniform distributions of the BMD within each condition. A univariate ANOVA with pull-out force as the dependent variable, and interference fit, friction, and time as independent variables, and BMD block and area of the hole as a covariate, was performed. Regression analysis was used to evaluate the correlation between BMD and the pull-out force. A p-level of 0.05 was defined for statistical significance. A Shapiro-Wilk test of normality was used to check the normality of the distribution of BMD and the dependent variable. An interaction plot and a statistical prediction model were obtained to determine the significant interactions. We used a centered BMD in the prediction model, which was obtained by subtracting the average BMD from the BMD dataset. Hence, a zero value represented the average BMD. Other independent factors could be filled in
the model with a value of either 0 or 1, corresponding to low or high levels of friction, interference fit, and time.

3. Results

BMD was normally distributed within all thirteen conditions (178.66± 62.89 mg/cm³ (mean± SD)) and surface area was not a significant factor (28.17± 1.53 mm² (mean± SD)). In total, 5 pegs were lost to the analysis due to technical difficulties, of which four in the high friction group (6.2 mm; 2 pegs and 7.3 mm; 2 pegs) and one in the low friction group (7.6 mm peg). Pull-out force increased with increasing peg displacement until the peak pull-out force was reached, after which the force reduced slowly or rapidly depending on the type of frictional surface (Figure. 2.2).

![Figure 2.2](image)

**Figure 2.2** A typical example of the force-displacement relationship for the both frictional groups, with interference fit of 1.6 mm, shows how the pull-out force is affected by friction.

3.1 Interference fit

In group T0 (no waiting period), the pegs with a 1.6 mm interference fit had a significantly higher pull-out force than the pegs with a 0.5 mm interference fit, for both the low (P< 0.001) and high (P= 0.010) friction groups. However, after a waiting period of 30 minutes (group T30), only the difference for the high friction group was statistically significant (P= 0.006) (Figure. 2.3A).

In the high friction group, a significant effect of interference fit (increasing pull-out force...
with increasing interference fit) was found when compared with the other interference fit groups (Figure. 2.3B), except between 0.2 mm and 0.5 mm in the T30 group, and 1.3 mm and 1.6 mm in both the T0 and T30 groups. A non-linear variation of pull-out force and interference fit was observed in both time groups (Figure. 2.3B).

3.2 Surface morphology

In time group T0, both interference fit groups (0.5 and 1.6 mm) with the low frictional surface had a significantly higher pull-out force than the high friction groups (P< 0.001). However, after the thirty-minute resting period, only the low interference fit displayed a significant difference (P= 0.001). Comparison between the low (160.87±17.73N (average±SE)), medium (121.39±13.19N (average±SE)) and high (129.31±16.70N (average±SE)) friction groups showed that after the resting period, regardless of surface morphology, the pull-out force was not significantly different (Figure. 2.3A).

![Figure 2.3](image)

**Figure. 2.3** A) Estimated average pull-out forces with standard error of mean. Each line between bars represents a significant difference as a result of interference fit (blue) or friction (green) or time (red). B) Relationship between pull-out force and interference fit in the high friction group (a line with two side tales shows significant difference for both time groups and blue font color indicates the same P-value for the both comparisons).

3.3 Bone material properties

3.3.1 Bone relaxation. Time was a significant factor only for the low friction group: (interference fit of 0.5 mm (P= 0.041) and interference fit of 1.6 mm (P< 0.001)) (Figure. 2.3A). Analyzing the time-dependent effect within the high friction group showed a non-significant lower pull-out force for all interference fits after the waiting period, except for the 0.2 mm sub-group, which was slightly higher (Figure. 2.3B).

3.3.2 BMD. The correlation between BMD and pull-out force showed a significant, moderate
to strong linear correlation, depending on the peg design (P< 0.05) (Figure 2.4). In general, the pull-out strength of the low friction group correlated strongly with BMD ($R^2 \geq 0.65$; P< 0.05). However, the high friction group had a poorer correlation ($R^2 \leq 0.74$; P< 0.05). The medium friction group also showed a strong correlation ($R^2 = 0.77$, P< 0.05).

![Figure 2.4](image)

**Figure 2.4** A) Linear correlation between pull-out force and bone mineral density (BMD) in both time groups, T0 and T30. Shaded area represents the low friction group.

### 3.4 Interaction of factors

#### 3.4.1 Interaction plot

Figure 2.5 depicts the interaction between the three variables included in this study: interference fit, friction, and time. It can be noted that the combination of friction (surface morphology) and time (bone relaxation) represented the only significant interaction, which can be noticed from the dependency lines not being parallel (P= 0.019). The interaction between time and interference fit was not statistically significant. However, after a resting period the pull-out force in the low friction group was reduced more in the higher interference fit pegs (38%) than in the lower interference fit group (29%).
3.4.2 Prediction model. To determine factors influencing the pull-out strength, a prediction model including interference fit (I), friction (F), time (T) and centered BMD (CB) and their interactions was adopted. Consequently, the following predictive model was obtained:

\[ \text{Pull-out force (N)} = 73.5 \text{ I}^* - 99.3 \text{ F}^* - 54.8 \text{ T}^* + 1.2 \text{ CB}^* - 0.7 \text{ F.CB}^* + 0.5 \text{ I.CB}^* - 0.5 \text{ T.CB}^* + 52.5 \text{ T.F}^* - 24.2 \text{ T.I} + 4.4 \text{ I.F} + 170.9 \]

* Determines statistically significant factors.

We did not include the interactions between more than two factors. The coefficient of each factor determines its weight on the prediction of pull-out strength. Hence, the model shows that the interference fit has a positive effect on the pull-out force, while the surface morphology has a strong negative effect, which seems counter-intuitive. However, the interaction between time and friction can overrule the negative effect of friction to a large extent. Therefore, a new prediction model was obtained using data from the time group T30 only, representing a long term condition, which clinically may be more relevant.

\[ \text{Pull-out force (N)} = 33.8 \text{ I}^* - 61.5 \text{ F}^* + 0.8 \text{ CB}^* - 0.6 \text{ F.CB}^* + 0.3 \text{ I.CB}^* + 34.7 \text{ I.F}^* + 123.8 \]

* Determines statistically significant factors.

It can be seen that the frictional properties can adversely affect the pull-out strength, while the interference fit has a positive effect. Table 2.2 provides two examples of predicted pull-out force in a longer time period using the prediction model leading to pull-out forces of 130.8 N and 267.8 N, respectively.
Table 2.2 Pull-out force is predicted for the two cases using the longer term prediction model.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Coefficient</th>
<th>First case Value</th>
<th>Definition</th>
<th>Second case Value</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>Interference fit (I)</td>
<td>33.8</td>
<td>1</td>
<td>High interference (1.6 mm)</td>
<td>0</td>
<td>Low interference (0.5 mm)</td>
</tr>
<tr>
<td>Friction (F)</td>
<td>-61.5</td>
<td>1</td>
<td>High friction (µ= 1.4)</td>
<td>0</td>
<td>Low friction (µ= 0.5)</td>
</tr>
<tr>
<td>Centered BMD (CB)</td>
<td>0.8</td>
<td>0</td>
<td>Average BMD (178 mg/cm³)</td>
<td>180</td>
<td>Maximum BMD (358 mg/cm³)</td>
</tr>
<tr>
<td>F × CB</td>
<td>-0.6</td>
<td>0</td>
<td>Interaction term</td>
<td>0</td>
<td>Interaction term</td>
</tr>
<tr>
<td>I × CB</td>
<td>0.3</td>
<td>0</td>
<td>Interaction term</td>
<td>0</td>
<td>Interaction term</td>
</tr>
<tr>
<td>I × F</td>
<td>34.7</td>
<td>1</td>
<td>Interaction term</td>
<td>0</td>
<td>Interaction term</td>
</tr>
<tr>
<td>Intercept</td>
<td>123.8</td>
<td></td>
<td></td>
<td>123.8</td>
<td></td>
</tr>
<tr>
<td>Pull-out force (N)</td>
<td>130.8</td>
<td></td>
<td></td>
<td>267.8</td>
<td></td>
</tr>
</tbody>
</table>

4. Discussion

In this study we aimed to investigate the effects of interference fit, surface morphology, bone material properties, and their interactions on the pull-out strength of press-fit implants. We believe that understanding these relationships does shed more light on the biomechanical behavior of press-fit implants and helps to understand the biological response of bone to biomaterial designs.

4.1 Effect of interference fit

Interference fit is a very important parameter for press-fit implants, because it is easy to control by changing the cavity or the implant size. Various studies have reported on the effect of interference fit on the mechanical performance of press-fit dental or orthopedic implants [9, 29, 30]. In general, the trend shows improvement of fixation by increasing the interference fit, but it can also lead to bone damage [14]. Therefore, it is essential to define a sensible range of interference fit, to prevent damaging of the bone or the implant. Our results confirmed that the interference fit can influence the fixation strength to a large extent, and its effect depends on the surface morphology and bone relaxation. After the 30-minute waiting period, we observed that a three times larger diametrical interference fit of a rough coated peg resulted in a two times higher pull-out force, whereas the effect was less obvious for the grit-blasted pegs, with only a 30% increase of pull-out force. Therefore, it is important to consider that a higher interference fit would improve the fixation in a non-linear manner, depending on
the surface morphology of the implant. Furthermore, we found that beyond a certain interference fit, the fixation strength does not increase any further. This is probably due to the creation of excessive stresses during insertion, causing damage of the bone, thereby limiting additional stability [29]. Based on our results, and for the current peg designs, a range of 650-800 µm of radial interference fit seems a sensible range, at which an optimal fixation strength can be achieved in cancellous bone. We would like to point out that we tested only one type of surface morphology at all four diameters, and the interaction of the implant to the interference fit may be different for implant designs with a different shape.

4.2 Influence of surface morphology

In contrast to what one might expect, rougher pegs had a lower fixation strength immediately after insertion, resembling results from Shirazi Adl et al. [10]. However, after the resting period, the pull-out force of the larger pegs was not significantly different between the low and high frictional groups. In addition, the medium friction group did not have a significantly different fixation strength relative to the two other groups. Shirazi Adl et al. [10] suggested that the reduction in fixation strength around rough coated implants could be explained by more bone damage caused during implant insertion. To investigate this hypothesis, we compared the measured area of each hole before and after testing to estimate the bone damage. Interestingly, bone damage was not statistically different between the frictional groups with the same peg diameter. However, we observed a large number of bone particles inside and around the holes with the rough coated pegs, while the low friction group produced a smooth surface inside and around the holes (Figure 2.6). In addition, although not quantified, after the pull-out test, more bone particles stuck to the rough surface coated pegs as compared to the low friction specimens.

Based on the damage assessment, the current study actually revealed two different modes of damage: ‘compaction’ and ‘abrasion’. During insertion, low friction pegs mainly compressed the bony wall of the hole, while the rough coated pegs mainly abraded the hole, which correspond to the results from Bishop et al. [31]. They distinguished permanent plastic and abrasive bone deformation for polished and rough surfaces, respectively, using micro-CT technique. However, they found a lower magnitude of damage for the polished surface, perhaps due to the more accurate nature of their technique. Consequently, the internal bone stresses may have been higher in the low friction group, as more bone material was present and being compressed, particularly for pegs with a larger interference fit. Therefore, in an immediate post-insertion condition, this causes higher frictional forces due to the higher radial stresses, even though the friction is lower. Due to the viscoelastic behavior of the bone, the pre-stresses dropped during the resting period, which
explains why after the resting period the fixation strength of the smooth implants were not different from the rougher implants anymore.

![Smooth contour of a hole from low friction group in comparison with rough contour of coated pegs.](image)

**Figure. 2.6** Smooth contour of a hole from low friction group in comparison with rough contour of coated pegs.

In the high friction group, bone relaxation was less pronounced due to lower radial forces, as a result of bone abrasion. Although not the focus of the current study, retained particles may promote early bone formation at the implant-bone interface, and potentially accelerate osteogenesis due to the release of biological factors [32]. Therefore, a distinction can be made between the use of smooth or rough implants. One could utilize a press-fit implant with a smooth surface structure when only direct post-operative stability is required (such as pegs in a femoral knee implant). However, if a long-term stability should be achieved, such as often the case with dental implants, a rough surface may be advantageous.

### 4.3 Effect of bone material properties

#### 4.3.1 Bone relaxation.

Based on our results, bone stress relaxation noticeably affects the fixation strength over time. Burgers et al. [33] also found a reduction of strain values on the surface of femoral knee implants after implantation, serving as an indirect measurement to quantify bone stress relaxation. Therefore, the viscoelastic behavior of bone over time is a factor that cannot be neglected in this type of analysis. This time aspect is related to the time required to stabilize the implant with bone ingrowth. Previous studies reported a wide range from 6 weeks to 9 months as a required period to achieve an effective bone ingrowth level [7, 34]. Therefore, further research is necessary to determine the residual stability in the longer time period before bone ingrowth occurs.

We found that a rough surface seems to be less sensitive to bone relaxation (with only a 13% reduction in fixation strength after 30 minutes) than smooth surface implants (with an average of 33% reduction in fixation strength). Therefore, from a viscoelastic point of view, rough surface structures may be more robust and beneficial to retain the primary stability over time, allowing bone ingrowth more effectively.
4.3.2 BMD. Our measurement technique was able to quantify the amount of bone mineral structure in a 3D volume around implants. We clearly demonstrated the effect of bone quality on the fixation strength of an implant, which emphasizes the role of bone quality on the potential outcome of press fit implants, which is in accordance with other studies [9, 13]. In addition, the correlation between BMD and pull-out force provides additional evidence for the difference in mode of fixation of the frictional groups. The pull-out force of the rough coated group had a moderate correlation with BMD, while there was a strong correlation for the low friction group. Hence, frictional forces of the rough coated implants are mainly dependent on the surface properties at interface, whereas smooth implants rely on the creation of pre-stresses in the surrounding bone which are dependent on the bone quality [33]. Therefore, BMD has less influence on the fixation strength of rough coated implants as compared to implant with a smooth surface finish.

4.4 The role of interaction terms

Bone material properties had a significant interaction with design related factors (interference fit and frictional properties). This mainly demonstrates the role of bone material properties on the biomechanical behavior of press-fit implants. We therefore strongly believe that, in spite of the inherent anatomical variability, real bone is preferred over synthetic bone materials if one is interested in the biomechanical behavior of existing or new implant designs or biomaterials. In addition, the significant interaction between interference fit and friction at a longer term indicates that changes in design parameters related to these factors should be assessed in the presence of both factors, and not independently.

4.5 Limitations

The main limitation of our study is that we considered only one design, implanted only in cancellous bone. Other design shapes and types of bone (such as the involvement of cortical bone around dental implants) may yield other results. Nevertheless, we believe that our findings are helpful to understand the general mechanical behavior of press-fit implants in a bony environment. Another limitation of our study was the loading regime, which was a simple push-in and pull-out test. It is clear that press-fit implants are subjected to shear and compressive forces in a physiological situation, which may further affect their mechanical behavior. However, the relatively simple loading condition allowed us to purely investigate fixation strength independently of the complicated loading regime that may act around an entire implant. Another limitation is the fact that we analyzed the viscoelastic effects for a period of only 30 minutes. A longer waiting period may have been of more clinical relevance. However, we noticed a fast reduction of the viscoelastic effect in the first minutes after insertion and therefore have captured most of its effect. Nevertheless, it would be interesting
to assess longer post-insertion times.

4.6 Conclusion

Our study demonstrates that both design and bone parameters can have a significant effect on the pull-out strength of press-fit implants, although the ultimate effect of each factor is dependent on other parameters due to their interactions. In general, pull-out strength increased with a larger interference fit and better bone quality, but it decreased when the surface morphology had a higher surface roughness. In addition, bone relaxation has a negative effect on the press-fit fixation strength. Based on the bone damage assessment and the weak correlation of pull-out strength with BMD in the rough group, we conclude that the fixation strength of a rough surface is more dependent on the frictional properties at the interface which are relatively insensitive to bone quality, whereas smooth specimens rely on compressive stresses within the bone and are therefore more sensitive to bone quality.

Acknowledgements

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References

6.
[28] Derikx LC, Vis R, Meinders T, Verdonschot N, Tanck E. Implementation of asymmetric
yielding in case-specific finite element models improves the prediction of femoral fractures.
Computer methods in biomechanics and biomedical engineering. 2011;14:183-93.
CHAPTER THREE

The effect of surface morphology on the primary fixation strength of uncemented femoral knee prosthesis: a cadaveric study


J Arthroplasty, 30(2), 300-307
CHAPTER 3

Abstract

We investigated the effect of surface morphology on the mechanical performance of uncemented femoral knee prosthesis. Eighteen implants were implanted on nine paired femurs and then pushed-off (left legs: a novel surface morphology; right legs: Porocoat as baseline). Bone mineral density (BMD) and anteroposterior dimension were measured, which both were not significantly different between groups. The insertion force was not significantly different, however, the loosening force was significantly higher in the novel group ($P=0.007$). BMD had a direct relationship with the insertion and loosening force ($P<0.001$). The effect of surface morphology on implant alignment was very small. We conclude that the surface properties create a higher frictional resistance, thereby providing a better inherent stability of implants featuring the novel surface morphology.
1. Introduction

Due to longer life expectancy and higher activity levels, young and active patients are at a greater risk of having revision total knee arthroplasty (TKA) [1, 2]. Recent long term follow-up studies have demonstrated that a hybrid fixation (uncemented femoral component with a cemented tibia) has a similar or even lower rate of revision compared to cemented fixation [3-5]. Therefore, further modifications of uncemented femoral components may provide additional benefit to improve the long-term clinical outcome especially for younger patients. Recent technologies in the field of metallurgy have introduced new opportunities for highly rough and porous metal surface morphologies [6]. A rough surface has an effect on the primary stability by increasing the shear-load bearing capacity at the bone-implant interface in the direct post-operative period [7, 8]. It has been shown that an adequate primary stability is a pre-requisite for a successful long-term fixation of uncemented implants [9, 10]. In addition, a porous surface plays a fundamental role in the mechanical interlock between bone and implant during secondary fixation [8, 11]. In this regard, a highly rough and porous surface structure from Cobalt-Chromium-molybdenum (CoCrMo) has been developed (DePuy Synthes Joint Reconstruction, Warsaw, IN, USA). However, using a highly rough surface may substantially affect the implantation procedure, as it may lead to higher insertion forces, thereby increasing the risk of intra-operative fracture [12] or bone damage. In addition, it might affect optimal alignment of the implant, which is a determining factor for the survivorship [13, 14], due to an incomplete seating of the implant on bone distal face.

This study focuses on the effects of surface morphology on the implantation and fixation strength of uncemented femoral prosthesis. We aimed to answer three research questions: (i) Does the novel surface morphology provide a better primary fixation strength? (ii) What is the effect of surface morphology on the insertion force? (iii) What is its effect on alignment of implant?

2. Materials and methods

2.1 Study design

Eighteen uncemented femoral knee implants were implanted on nine paired femurs (71-93 years [average: 86 years], 8 females). The Porocoat® porous surface morphology (DePuy Synthes Joint Reconstruction, Warsaw, IN), which is used in the conventional uncemented Sigma® and LCS® (DePuy Synthes Joint Reconstruction, Leeds, UK), was used as clinical benchmark. Left legs were allocated to the novel surface morphology group, and right legs to the Porocoat® group. The main comparison was performed between the pairs, to minimize the influence of anatomical variations. No randomization was performed between left and right leg, because only left implants with the novel surface morphology were available.
2.1.1 Surface morphology. The surface morphologies were applied to the conventional uncremented Sigma cruciate retaining (CR) femoral knee prosthesis, which is designed for the normal flexion range (DePuy Synthes Joint Reconstruction, Warsaw, IN). The novel surface morphology was an experimental surface coating with no FDA approval. The application process is proprietary for both surface morphology technologies, but in general terms, both surface morphologies are generated using a sintering technique. The novel surface morphology had an average volume porosity of 60%, an average pore size of 300 µm and an average thickness of 0.9 mm. The Porocoat had an average volume porosity of 44%, an average pore size of 250 µm and an average thickness of 0.75 mm. Scanning electron microscopy (SEM) images of the novel surface morphology show a random arrangement of highly irregular particles forming strongly interconnected pores (Figure. 3.1A), while Porocoat is comprised of randomly arranged spherical beads (Figure. 3.1B). No roughness data was available owing to the high porosity of the surface morphologies, which makes this measurement technique irrelevant. However, a friction coefficient was provided by the manufacturer, with 1.4 and 0.95 for the novel surface morphology and Porocoat, respectively.

![Figure 3.1](image)

**Figure 3.1** A) Scanning electron microscopy (SEM) of the novel surface morphology. B) SEM of Porocoat® surface morphology.

2.1.2 Femora preparation. Fresh frozen cadavers were obtained from the Anatomy department of the Radboud university medical center. Selection criteria were compatibility with implant size 3 using digital templates and no abnormalities judged by x-ray images. Bone specimens were thawed at room temperature and cleaned from soft tissues before the cutting session. All femurs had a rather healthy cartilage layer; hence they were osteotomized by an orthopedic surgeon with a 10 mm distal cut, corresponding to the normal surgical procedure. The distal femur was then cut off proximally at a length of about 10 cm. Finally the proximal end of the specimen was casted in a cylindrical pot using bone cement, which...
enabled rigid fixation during testing (Figure 3.3).

2.1.3 CT-scanning. The specimens were 3D scanned using computed tomography (CT) (Siemens Somatom Sensation 64, Siemens AG, Germany), along with a hydroxyapatite calibration Phantom (solid, 0, 50, 100, 200 mg/ml calcium hydroxyapatite, Image Analysis, Columbia, KY). The CT-images were used for two purposes:

2.1.3.1 Bone quality assessment. It is known that bone quality can affect the insertion force and the primary fixation strength [15, 16]. The Hounsfield units of the CT-images were converted to BMD values using in house software [17]. The CT-scan data was used to find a correlation between bone mineral density (BMD) and the outcome parameters. Regions of interests (ROIs) were rectangular cubes of 35×18×5 mm and 24×18×5 mm (width× height× depth) in the anterior region and in the posterior condyles, respectively. These dimensions were selected based on the nominal implant dimensions and an unpublished, finite element analysis of a similar implant design on a sawbones. This study showed that peak stresses resulting from press-fit forces disappeared at a depth of 5mm. The average BMD of each ROI of each specimen was used for further analysis. In addition, the BMD values were compared between groups.

2.1.3.2 Interference fit. Since variations of the bone cuts may affect the outcome parameters, we assessed its potential effect. In uncemented femoral knee implants, the press-fit forces are mainly provided by compaction of the bone in the anteroposterior (AP-) dimension. Hence, calibrated images of the medial and lateral views of the bone specimens after osteotomy were used to measure the average distance from the distal face to the anterior flange cut (distance ‘d’ in Figure. 3.2A). By using the slice thickness of the CT-scans and the distance d, the closest CT-slide to the beginning of the anterior flange cut was determined. Then, the AP-dimension was measured in the medial and lateral side (Figure. 3.2B). We used the average AP-dimension for our analyses.

![Figure 3.2](image)

Figure. 3.2 A) Calibrated images of medial and lateral views were used to measure the distance (d) between distal face and anterior flange cut to determine the appropriate CT-slide. B) A CT-slide with anteroposterior (AP) dimension measurements in the medial and lateral sides.
2.2 Strain gauge measurement

Because the initial fixation strength of the two surface morphologies may be directly related to the pre-stresses created during the implantation, we indirectly quantified the pre-stresses by applying strain gauges on the femoral component. Four unidirectional strain gauges (Y-FLA-5, Tokyo Sokki Kenkyujo Co. LTD, Japan) were attached to each implant at the medial and lateral sides of the anterior and posterior regions (Figure. 3.3). Strain gauge signals were recorded until the end of the experiment, and average strain values from the four strain gauges after the implantation and before push-off tests were compared between the two groups, to evaluate the implant deformation after the implantation and to compare the pre-stresses after two hours.

![Figure. 3.3 Four strain gauges were attached to each implant.](image)

2.3 Mechanical testing

2.3.1 Implantation. To measure the force required to insert the implant, a servo-hydraulic testing machine (MMED, MATCO, La Canada Flintridge, CA) was used. Hence, a compressive force (instead of manual impaction) was applied to the implant by using a Sigma femoral inserter/extractor, attached to a free-sliding jig (LWR 6150, SKF, Germany) (Figure. 3.4A). The free-sliding jig was used to avoid constraints in the AP direction during insertion. The displacement-controlled insertion test was stopped when the implant distally touched medial or lateral side of the bone, or when a sudden increase in the force signal occurred.
After the implantation phase, the specimens underwent a 30-minutes relaxation phase, while the signal of the strain gauges was recorded.

2.4.1 Assessment of implant alignment. Directly after implantation, calibrated images of the medial and lateral sides were taken to assess the flexion-extension (Sagittal angle) and varus-valgus (Coronal angle) orientations of the implants. Flexion and varus angles both were defined as positive values. An implant was defined as distally seated when it touched the distal face of the bone cut in both the medial and lateral views.

2.4.1.1 Sagittal angle: the sagittal angle was determined by measuring the angle between the distal faces of the implant and bone, using ImageJ (National Institutes of Health, USA).

2.4.1.2 Coronal angle: the coronal angle was quantified by using a calculation technique reported by Cooper et al. [18]. The formula estimates the deviation angle from the desired coronal alignment of 95° with respect to the anatomical axis, when the implant is incompletely seated on the distal bone cut, by using the distal gap size and the mediolateral (MD) size of the implant.

2.4.3 Dynamic loading. After the 30-minutes relaxation phase, the implants were subjected to a dynamic loading regime of one hour at 1 Hz at a flexion angle of 17° (Figure. 3.4B). This angle was based on the direction of the in-vivo peak load occurring during a cycle of gait [19]. The applied force interval was 250-1500 N, with a ratio of 70-30% on the medial and lateral sides, respectively. The upper limit of force was kept lower than reported in-
vivo contact forces, to reduce the risk of damage for bones with a lower quality [19]. We re-evaluated the implant alignment after the dynamic loading phase to assess whether the implant had re-aligned itself during the dynamic loading period. Specimens then underwent a second relaxation phase of thirty minutes before the push-off test.

2.4.4 Push-off test. The bone specimen was firmly fixed to the MTS table while simulating 150° of flexion, which was adopted as an extreme physiological loading condition (Figure. 3.4C). A transverse notch was made in the posterior condyles to facilitate the application of a push-off force by a custom-made load applicator, in accordance with a contact normal force vector of 150° of flexion, that was found previously in an FEA study [20]. Using this method, rather than using the tibial implant to apply the load, we prevented possible sliding between the femoral and tibial implant.

The loosening force was defined as the maximum recorded force during the push-off test. After the implant was removed, high-resolution images were taken of the inner surface of the implant to observe any bone remnants in or on the porous prosthetic surface.

2.5 Statistical tests

A univariate analysis of variance (ANOVA) with insertion force, loosening force, initial strain and final strain as dependent variables and surface morphology as an independent variable, with correction for the effect of BMD and AP-dimension, was performed. Paired Student’s t-tests were used to compare BMD and AP-dimension between left and right legs. Regression analysis was also used to investigate the correlation between different variables. A P-value of 0.05 was defined as significant.

3. Results

3.1 Bone mineral density measurement

A high degree of homogeneity was found between the bone pairs with 175 mg/cm³ as an average BMD for both groups, with a range of 94-307 mg/cm³ for the novel surface morphology group and 88-317 mg/cm³ for the Porocoat group. There was no significant difference between the groups (P= 0.999) (Table. 3.1).

3.2 Anteroposterior dimension measurement

The average AP-dimension was 45.58 mm (range, 44.38-47.11). Similar to the BMD results, the AP dimension was distributed homogenously over the groups [novel surface morphology: 45.44 mm (range, 44.38-46.65); Porocoat group: 45.73 mm (range, 44.49-47.11)]. No significant difference was found between the groups (P= 0.410) (Table. 3.2).
Table 3.1 General information, bone mineral density (BMD), insertion force, loosening force, and bone fracture site are given.

<table>
<thead>
<tr>
<th>Bone ID</th>
<th>Age-Gender</th>
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<th>BMD (mg/cm³)</th>
<th>Insertion force (N)</th>
<th>Loosening force (N)</th>
<th>Bone fracture site</th>
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</thead>
<tbody>
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<td>93-F</td>
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<td>Condyles</td>
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<tr>
<td></td>
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<td>Porocoat</td>
<td>88</td>
<td>1,760</td>
<td>239</td>
<td>No fracture</td>
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<tr>
<td>F</td>
<td>87-F</td>
<td>Novel</td>
<td>189</td>
<td>5,512</td>
<td>1,532</td>
<td>Condyles</td>
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<td></td>
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<td>G</td>
<td>90-F</td>
<td>Novel</td>
<td>221</td>
<td>4,845</td>
<td>2,085</td>
<td>Underneath of condyles</td>
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<tr>
<td></td>
<td></td>
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<td>214</td>
<td>5,045</td>
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<td>Condyles</td>
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<td>No fracture</td>
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<td>Novel</td>
<td>96</td>
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<td>Condyles</td>
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<td>Porocoat</td>
<td>88</td>
<td>2,569</td>
<td>409</td>
<td>No fracture</td>
</tr>
</tbody>
</table>

3.3 Insertion force

The mean of the insertion force in the novel surface morphology was 4262 N (95% CI: 3596-4927), and 3644 N (95% CI: 2978-4310) for the Porocoat® implants, but this difference was not statistically different (P= 0.186) (Table. 3.1; Figure. 3.5B). AP-dimension did not have a significant effect on the insertion force (P= 0.628).

BMD was a significant factor (P< 0.001) for the insertion force. A strong correlation between BMD and insertion force in the Porocoat group (R²= 0.88, P< 0.001) and a moderate correlation of the novel surface morphology group (R²= 0.65, P< 0.001) was found (Figure. 3.5A).

3.4 Loosening force

In 8 of the 18 cases, the implants were not loosened due to fracture of the bone before reaching the threshold of implant loosening, of which seven cases belonged to the novel surface
morphology group (Table 3.1). The novel surface morphology group had a significantly higher loosening force than the Porocoat group ($P = 0.007$) (Figure 3.5D). Similar to the insertion force result, BMD was a significant factor ($P < .001$), but not AP-dimension. A strong correlation was found between BMD and loosening force for the novel group ($R^2 = 0.84$, $P < 0.001$) and also for the Porocoat group ($R^2 = 0.88$, $P < 0.001$) (Figure 3.5C).

3.5 Strain gauge measurement

In two pairs, strain gauges recorded too much noise during the second phase of relaxation (pairs H and I). These artifacts were probably caused by the relocation of the specimens for preparation of the push-off test set-up. These pairs were excluded from the strain assessments. In addition, the dynamic load was not applied to pair E, due to a risk of fracture; this pair was excluded from the analyses, as well.
Initial strain was significantly higher in the Porocoat group (P= 0.019), but the final strain values were not significantly different between the groups (P= 0.283) (Figure 3.6). BMD was a significant factor for the initial (P< 0.001) and final strains (P= 0.019), but AP-dimension was not a significant factor. In addition, initial strain was a significant factor for the final strain (P= 0.003).

**Figure 3.6** A) Comparison of average initial strain. B) Average final strain was compared between the two groups (Estimated mean ± 2 std. err.).

### 3.6 Assessment of implant alignment

Two implants out of eighteen were classified as distally seated, both of which belonged to the novel surface morphology group. BMD and insertion force were not significant factors for both sagittal and coronal angles.

#### 3.6.1 Sagittal angle

In the sagittal plane, all implants except one from the Porocoat group were in the flexed orientation directly after implantation. The sagittal angle was not significantly different between groups (P= 0.175), although the average angle was higher (1.61°± 0.25°) in the novel surface morphology group, in comparison with the Porocoat group (1.13°±0.21°) (Table 3.2). The largest angle was 2.87° of flexion for an implant with novel surface morphology.

#### 3.6.2 Coronal angle

Relatively small deviations in coronal alignment were found for both groups (Table 3.2). The novel group had a slightly lower angle, 0.34°± 0.06°, in comparison with the Porocoat, 0.46°± 0.14°, which was not significant (P= 0.154) (Table 3.2). The largest angle was 1.09° in a varus orientation from Porocoat group. Four out of the eighteen implants had a valgus orientation, which all belonged to the novel surface morphology group (Table 3.2).
After the dynamic loading phase, there was a slight change in the both sagittal and coronal angles, which was a statistically significant change only for the coronal angle of Porocoat (P= 0.008). Flexion-extension and varus-valgus orientations of the all implants remained the same (Table 3.2).

<table>
<thead>
<tr>
<th>Bone ID</th>
<th>Surface morphology</th>
<th>AP-dimension (mm)</th>
<th>Coronal angle (°)</th>
<th>Sagittal angle (°)</th>
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<td>A</td>
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<td>1.15</td>
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<td>0.47</td>
<td>2.56</td>
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<td>0.07</td>
<td>2.16</td>
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<td>1.10</td>
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</tr>
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<td>0.23</td>
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<td>Porocoat</td>
<td>47.11</td>
<td>0.2</td>
<td>2.11</td>
</tr>
</tbody>
</table>

Positive angles represent varus and flex orientation.
* Implant was distally seated.

4. Discussion

The aim of our study was to shed more light on the mechanical performance of uncemented femoral knee prosthesis in a simulated immediate post-operative condition, when adding a novel rough porous surface morphology. We used the conventional Porocoat surface morphology as a benchmark for the comparison [21].
4.1 Influence of the surface morphology on the fixation strength

Our results showed superior primary fixation strength of novel surface morphology relative to the Porocoat group with a 40% higher average loosening force. This difference is even more explicit when we consider that about 80% of implants from novel surface morphology group did not loosen, with fractures of the femurs at a higher force value than the loosening force of the contralateral component. Hence, in those cases not the bone-implant interface was the weakest link, but the bone itself. Interestingly, higher initial strains on the implant surface of the Porocoat group were measured, which indicates a larger outward deformation of the implant. Apparently, higher compressive pre-stresses were created at the bone-implant interface of the Porocoat group during implantation. This may be due to increased abrasion with the rougher novel coating, which would be consistent with observation of greater amount of bone retained on the implant surfaces at the end of the experiments (Figure 3.7A). In addition, the bone surfaces of the posterior condyles which were in contact with Porocoat implants were relatively flat and smooth as compared to the surfaces that were in contact with the novel surface morphology group, in which we even observed fracture of the posterior condyles in three cases (Figure 3.7B). However, the reduced abrasion and higher pre-stress in the Porocoat group did not lead to higher fixation strength. Conversely, after two hours, relaxation of the bone meant that there was no significant difference in pre-stresses between the two groups, and a higher fixation strength was achieved with the novel group. After stress relaxation within the bone, the higher coefficient of friction of the novel groups resulted in a superior primary fixation strength despite reduced initial pre-stresses.

Figure 3.7 A) Anterior and posterior regions of two implants after push-off test which implanted on a pair (top: the novel surface morphology; bottom: Porocoat). B) Fracture of posterior condyles was observed in three implanted specimens with the novel surface morphology (top) in comparison to the smooth surface of posterior condyles in Porocoat group (bottom).
Harrison et al. [7] reported that a surface morphology with a higher friction coefficient provided better primary stability in hip implants. That study showed a significantly lower micromotion of a hip stem with a rougher surface under cyclic loading. Gebert et al. [22] found a positive effect of friction coefficient on the fixation strength of an uncemented femoral resurfacing implant in a finite element study. Hence, improvement of fixation strength in the femoral knee component as a result of rougher surface morphology corresponds to previous studies, although in the current study a different implant design was analyzed. A higher initial fixation can decrease micromotions at the bone-implant interface, improving the circumstances for bone ingrowth [23].

Our results showed a clear effect of BMD on the loosening force. Hence, a better fixation strength is provided with bone of better quality. This is in accordance with Meneghini et al. [24] who measured a lower primary stability of tibial components implanted in osteoporotic bones. Thus novel surface morphology can be useful especially in a group of patient who suffers from osteoporosis or osteopenia to compensate for a lower bone quality.

4.2 Effect of novel surface morphology on the insertion force

We found 17% higher average insertion force in the novel surface morphology group, however, when pair H was excluded as an outlier, due to an unrealistic high insertion force, the difference was reduced to 8%. We can explain our findings by considering the frictional properties at the bone-implant surface. The friction force ($F_f$) at the implant-bone interface is defined by the equation: $F_f = \mu F_N$; in which $\mu$ is the friction coefficient and $F_N$ is the normal force acting on the implant interface. A higher friction coefficient was measured in the novel surface morphology, but, the strain gauge measurement implied to higher normal forces in the Porocoat group (Figure. 3.6). As these two features are compensating phenomena, the ultimate insertion forces are in a similar range in the two groups.

The average insertion force was a bit higher than reported by Burgers et al. [15] who measured about 2500N in their best bone quality case. We measured this range of insertion force only in the bones with BMD values lower than 150 mg/cm$^3$, which means low bone quality. Burgers et al. [15] suggested that a horizontal impaction in their set-up could have caused a lower impaction force in their study. Moreover, we measured our insertion force during a gradual insertion instead of dynamic impaction; the latter probably being associated with increased peak forces. Considering these two reasons, we would suggest that our measured values are in a similar range as the clinical impaction force. In general, our results showed that BMD is a determinant factor for the insertion force, which indicates that a higher insertion force is required in a bone with a higher quality.
4.3 Effect of novel surface morphology on the implant alignment

The optimal alignment of femoral knee implant is obtained when the implant is fully seated on the distal face of the bone. Bonner et al. [25] found a higher risk of implant loosening when the mechanical axis deviated ±3° in the varus/valgus direction. We could not quantify the deviation from the neutral mechanical axis, because the full femur was not available for some bone specimens; however, the deviation from the ideal coronal angle of 95° was not statistically different between the two groups. There was also no significant difference between groups in the deviation angle from the optimal sagittal alignment of zero. We found a flexed orientation in almost all implants, which is probably due to larger contact area of the anterior region as compared to the posterior condyles, pushing the implant into a flexed orientation. After the dynamic loading phase there was a significant change in the coronal angle of the Porocoat surface coated implants, but not in the novel surface morphology group, which may indicate a better fixation of the latter. We suggest that the low number of distally seated implants is a result of the insertion technique; we stopped the insertion procedure manually as soon as the implants touched the medial or lateral side of the distal bone cut, or when recording a high insertion force, to avoid any possible damage to the bone as a result of artificially high forces.

4.4 Limitations

As mentioned above, the implantation was performed with a machine with manual control of the implantation process. Evidently, an orthopaedic surgeon has more control over the implantation process from a seating and alignment point of view, as the final implant position can be manipulated easier by changing the force and orientation of the hammer blows. However, since in the current study fixation strength was our primary outcome measure, we used the MTS machine to standardize the implantation process as much as possible, to prevent any bone damage affecting the fixation strength. This likely affected the final implant seating as measured in the current study. Another limitation was that the AP-dimension measurement from the CT-slides was subject to errors, caused by the resolution of the CT-scan and the fact that we measured the cuts with only two distances. Taken these uncertainties into account, we did not observe any confounding effect of the AP-dimension measurement on the outcome parameters.

Finally, although we applied a high-flexion push-off force, we used a conventional Sigma CR implant design. Hence, we used a custom made load applicator to apply the appropriate force vector. We believe that the results found in this study are translatable to high-flex designs, since we assessed effects of the surface finish of which the findings are probably applicable for a variety of designs.
4.5 Conclusion

We found that the primary fixation strength of the uncemented femoral knee prosthesis is affected by the surface morphology of the implant interface and our data suggest that surface morphology can affect the implantation process. Increased roughness of the novel surface morphology caused greater abrasion and lower compaction of the bone, but higher frictional resistance, therefore resulting in a similar range of insertion force as the Porocoat. Our findings show that after bone relaxation, press-stresses equalized between coating types and the superior frictional properties of the novel surface morphology lead to a better primary stability. Furthermore, surface morphology had a small effect on the implant alignment. Further clinical and biomechanical investigation of the novel surface morphology is required to assess whether the benefits as found in this study are clinically relevant on the longer term.

Acknowledgments

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References


THE EFFECT OF SURFACE MORPHOLOGY ON THE FIXATION STRENGTH
CHAPTER FOUR

FE analysis of the effects of simplifications in experimental testing on micromotions of uncemented femoral knee implants


J Orthop Res, 34(5), 812-819
Abstract

Experimental testing of orthopaedic implants requires simplifications concerning load application and activities being analyzed. This computational study investigated how these simplifications affect micromotions at the bone-implant interface of an uncemented femoral knee implant. As a basis, validated in vivo loads of the stance phase of gait and a deep knee bend were adopted. Eventually, three configurations were considered: (i) simulation of the complete loading cycle, (ii) inclusion of only tibiofemoral loads (ignoring patellofemoral loads), and (iii) applying only a single peak tibiofemoral force. For all loading conditions the largest micromotions found at the proximal anterior flange. Without the patellofemoral force, peak micromotions increased 6% and 22% for gait and deep knee bend, respectively. By applying a single peak tibiofemoral force micromotions were overestimated. However, the peak micromotions corresponded to the maximum tibiofemoral force, and strong micromotion correlations were found between a complete loading cycle and a single peak load ($R^2= 0.73$ and $R^2= 0.89$ for gait and deep knee bend, respectively). Deep knee bend resulted in larger micromotions than gait. Our study suggests that a simplified peak force can be used to assess the stability of cementless femoral components. For more robust testing, implants should be subjected to different loading modes.
1. Introduction

Due to the aging population and an increasing number of younger patients, there is a stronger need for total knee replacements (TKR) with an improved endurance. Recent long-term follow up studies and registry reports have shown promising outcomes for uncemented femoral components in TKR[1-3]. However, a higher incidence of implant loosening has been reported for younger patients and some high-flex designs [4, 5], which necessitates the improvement of existing designs [6, 7].

Long-term stability at the bone-implant interface of cementless implants can be achieved by providing an adequate primary stability, which is mostly defined by the relative motion between bone and implant, also known as bone-implant micromotions[8-11]. Small micromotions (<40-50 µm) provide the necessary condition for secondary biological fixation, induced by bone growth in- and onto the (porous) surface of implant. Hence, direct post-operative evaluation of micromotions at the bone-implant interface of a cementless implant is useful to gain insight into long-term implant fixation.

Experimental studies are known as the gold standard to assess micromotions [12-14]. However, various limitations are apparent in experimental testing set-ups. One limitation lies in the loads that can be applied. Generally, only simplified loading configurations can be used, representing the peak loads occurring during activities of daily living (ADL) [15-17]. Most experimental set-ups to test the stability of the femoral knee implant are even more simplified, in which only the tibiofemoral (TF) contact forces are being applied based on a proposed load set-up from the International Organization for Standardization (ISO 14243-1 and ISO 14243-3) [18, 19]. However, the effect of neglecting the patellofemoral (PF) force and using a simplified peak TF force on the micromotion is unclear. Furthermore, a femoral knee implant experiences a wide range of activities, which have variable impact on primary stability [20]. Hence, it is important to assess the effects of different ADLs on the micromotions. However, it is quite complex to implement many different loading conditions in experimental set-ups [19].

Another limitation of experimental set-ups is the limited capacity to assess the micromotion patterns at the whole bone-implant interface. Usually a limited number of measurement sites (e.g. 1-4 sites) are included using uni-directional motion sensors (e.g. linear variable differential transducers) [16, 21]. However, prior to testing, it is not known at which locations these sensors should optimally be placed. This problem is more apparent for a femoral knee implant due to its complex geometry.

We propose that the application of Finite Element (FE) modeling can be used as a complementary guidance tool to address these limitations. Recently, a validated FE model of the Kansas knee simulator [22] has been developed, which successfully reproduced in-vivo forces measured by Kutzner et al.[23]. Such dynamic loading profiles can be utilized to predict...
micromotions under different complete physiological loading cycles, as well as simplified loading conditions. Hence, it provides insight into the differences in the micromotion due to variations in loading conditions and ADLs.

In FE analyses, it is essential to have a clear definition of micromotions. In an experimental set-up, the micromotion is defined as the difference in displacements of bone and implant, measured between an un-loaded and loaded state[18]. In an FE analysis, micromotions can be defined to correspond exactly to the experimental definition. Thus, when a complete loading cycle is being incrementally applied, the micromotion can be obtained at each incremental load with respect to the initial position. However, it is likely that a full trajectory of motion under a complete loading cycle results in a greater range of micromotion. This suggests a second definition of micromotion: to monitor the trajectory of motion and determine the largest distance between all increments[24]. Therefore, it is important to use the most adequate definition for addressing our research questions.

The objective of the current study, therefore, was to investigate how micromotions of an uncemented femoral knee implant subjected to different physiological loading conditions compared against those induced by simplified loading conditions. Using FE analysis, we aimed to answer following research questions: (i) What is the effect of omitting the PF forces on the micromotions, and do the largest micromotions occur during the peak TF force? (ii) What is the effect of two very different ADL’s (gait and deep knee bend) on the micromotion? (iii) What are the most relevant measurement sites in an experimental set-up based on the predicted micromotions at bone-implant interface? (iv) What is an adequate measure for micromotion to compare different loading conditions?

2. Materials and Methods

A 3D model of full femur was created from cadaveric CT-data using medical imaging software (Mimics 14, Materialise, Belgium). Bone cuts were made following the manufacturer instruction by using a cutting guide and only the distal part was kept for a length of about 10 cm. The CAD file of size 3 uncemented PFC® Sigma® Cruciate-retaining (CR) femoral knee implant was provided by the manufacturer (DePuy Synthes Joint Reconstruction, Warsaw, IN). Both models were imported into FE software Marc/Mentat 2007 (MSC. Software Corporation, Santa Ana, CA, USA) to create an FE mesh. The mesh density was based on a convergence study, in which the percentage of surface area less than 5 micron did not change by more than 2% and also the distribution of micromotion was similar for three different models with average edge length of 0.5 mm, 1 mm, and 2 mm. Therefore, solid linear four-noded tetrahedral elements with the average edge length of 2 mm was used for both implant and bone meshes (coarser elements were used for the inside of the bone), resulting in 43,476 and 33,406 of elements for bone and implant, respectively (Figure. 4.1).
The implant and the bone were modeled as an isotropic linear elastic material. The implant was assigned with material properties of cobalt-chrome alloy (Young’s modulus (E): 210 GPa, Poisson’s ratio (ν): 0.3). To assign realistic bone material properties to the FE model, a calibration phantom (solid, 0, 50, 100, 200 mg/ml calcium hydroxyapatite, Image Analysis, USA) was scanned along with the cadaver. Using in-house software, the Hounsfield Units were converted to equivalent calcium values and subsequently, these were related to the Young’s modulus (47.8 - 14,240 MPa) [25, 26].

2.1 Contact parameters

A single-sided touching contact algorithm was defined between bone and implant. We used a Coulomb bilinear (displacement) friction model with a coefficient of friction of 0.95, provided by the manufacturer for a Porocoat® surface coating (DePuy Synthes Joint Reconstruction, Leeds, UK). Around the femoral pegs, which had a grit-blast surface, a coefficient of friction of 0.4 was used [27]. To apply a sensible value of interference fit at the bone-implant interface, we simulated a pull-off test, which was performed by the DePuy Synthes Joint Reconstruction (Warsaw, IN), using the same implant design and a polyurethane foam block that had been hand cut by experienced surgeons. The results of this verification study showed that using the interference closure option of MSC.Marc, an interference fit in the anteroposterior (AP) direction of 180 to 300 µm provided a similar range of pull-out forces (925± 215N (mean ± SD)) as found in the experiment (886.1± 147.4N (mean ± SD)). In the current study, we applied two interference fits of 20 and 200 µm, as a worst case interference fit and a more realistic interference fit, respectively. In addition, no interference fit was applied around the pegs.
2.2 Loading conditions

The bone model was fully constrained proximally. To derive the implant-specific boundary conditions for the micromotions analysis, a previously published FE model of the lower limb [28], developed in Abaqus/Explicit, was utilized. Briefly, the model included the femur, tibia and patella, tibiofemoral ligaments, extensor mechanism, quadriceps and hamstrings muscles, and mechanical actuators (used to apply loads at the hip, ankle and muscles). Target TF joint loading profiles for stance phase of gait (SG) and deep knee bend (DKB) activities were adopted from published data from an instrumented telemetric tibial tray measuring in vivo knee joint loads [23, 29].

The lower limb model was initially implanted with the same TKR component design as the telemetric patients, and simulations were performed with a series of proportional-integral-derivative (PID) controllers, which applied the external loads (quadriceps and hamstrings muscle force, vertical hip force, ankle flexion-extension torque, internal-external torque and varus-valgus torque) required to reproduce the experimentally-measured loading conditions at the TF joint for each activity.

Subsequently, these external loading conditions were applied directly to the model which was virtually implanted with a rotating-platform (RP) Sigma® CR design, and SG and DKB simulations were performed. TF and PF force vectors acting on the femoral component as a result of contact with the tibial insert and patellar button were recorded and used as input for the current study. Subsequently, four other simplified loading conditions, mimicking typical loading profiles that would be used in an experimental set-up were also simulated. In the first variation, for both the SG and DKB activity, the PF contact force was excluded from the loading profile, applying only the TF contact force. In the second variation, only the single peak TF contact force of both activities was applied, omitting the remainder of the loading cycle. Hence, in total, six different loading cases were defined. All models were analyzed using linear analyses, in which the stance phase of gait and the deep knee bend were simulated in a quasi-static manner (MSC.Marc2007r1, (MSC. Software Corporation, Santa Ana, CA).

2.3 Micromotion

The relative displacement of the implant with respect to the bone was calculated using the same technique as Van der Ploeg et al. [24]. Briefly, the implant nodes in contact with the bone surface were projected onto the closest bone contact face at each time increment. The nodes at the tip of the pegs and those in the overhang area at the anterior flange, which had a large gap, were excluded from the micromotion analyses. The deformation of bone was accounted for in the computational algorithm and both the magnitude and direction of the micromotions were calculated by tracking the trajectory of the projections.
Two different measures of micromotions were defined in each increment: (i) the micromotion relative to the initial position of the implant (un-loaded phase), which is equivalent to the micromotions that would be measured in an experimental set-up (referred to as reference micromotion), and (ii) the largest possible distance between two points of all projected displacements, occurring during a discretized full cycle (referred to as resulting micromotion) (Figure. 4.2). The resulting micromotion provides more information on the largest possible displacement between bone and implant during a full physiological cycle of an activity.

To account for initial settling of the implant, for each activity, four consecutive cycles were run. Between each of the four cycles, an extra increment was added in which no force was applied. The micromotion results presented here were obtained in the fourth loading cycle, only.

For each simulation, the percentage of the implant surface area (excluding overhang area) with resulting micromotions less than a defined threshold was calculated to illustrate the effect on the ingrowth potential. In addition, the peak and average of both reference and resulting micromotions of implant nodes were obtained.

3. Results

The preliminary “settling” simulations revealed a migration pattern when the interference fit was 20 µm. However, for the higher interference fit, no settling of the implant was observed and the motion paths were virtually identical for the four loading cycles (Figure. 4.2).

**Figure. 4.2** (A) Trajectory of a node during four cycles of gait loading, showing settling of the implant at an interference fit of 20 µm. (B) Settling was absent for an interference fit of 200 µm. In addition, examples of reference and resulting micromotions are indicated in the micromotion patterns for both interference fits. The reference micromotion was calculated with respect to the initial position at every time point for each node; the red lines show one of the instances. The resulting micromotion (blue lines) was the largest distance between all projected displacements.
Both reference and resulting micromotions depended on the interference fit and type of activity. When looking at the incremental changes, the average of the reference micromotions clearly followed the pattern of the applied TF contact force, with a peak micromotion corresponding to the maximum TF force especially for the SG loading (Figure. 4.3). However, the peak micromotion during SG loading at the maximum TF force was reduced remarkably when the interference fit was increased (Figure. 4.3C). In addition, it appeared that omission of the PF force did not change the observed trend. Therefore, the first peak observed in the micromotions evolution was probably a result of the maximum flexion angle that was reached at that point in the loading curve (Figure. 4.3).

The maximum of the average reference micromotions was 25% to 50% less than the average resulting micromotions of the last incremental load (Figure. 4.3). Based on this comparison, the resulting micromotions were chosen as the main measure of micromotions to compare different loading conditions. In the following results, “micromotion” refers to the resulting micromotions, unless stated otherwise.

By increasing the interference fit, the peak micromotions under the DKB loading cycle decreased from 46 to 38 µm, and the surface area with micromotions lower than 5 µm increased from 56% to 76%. The effect of interference fit was less pronounced under SG loading, with an increase of the surface area with micromotions lower than 5 µm from 78% to 92%. However, at a threshold of 50 µm, no differences were seen between the two levels of interference fit.

All loading conditions resulted in a similar distribution of micromotions. The largest micromotions were present at the proximal part of the anterior flange, which appeared to be the most relevant measurement site (Figure. 4.4). The posterior condyles also showed relatively high micromotions, especially for the SG loading (Figure. 4.4), while the lowest values were found in the distal region.

The peak micromotion during the DKB was about 2.5 times higher than the SG loading condition (Figure. 4.4A and D). Although the distribution was similar, the trajectory of a node in the proximal part of the anterior flange was slightly different between SG (Figure. 4.4A) and DKB (Figure. 4.4D). The trajectory generally followed the same direction, first heading in the proximo-lateral direction, and subsequently returning back in the opposite direction forming a loop pattern.

In the absence of the PF force, the peak micromotions were larger (6% and 22% for the SG (Figure. 4.4B) and the DKB (Figure. 4.4E) activity, respectively). These differences slightly reduced when the interference fit was increased. In addition, the trajectory of micromotions in the proximal part of the anterior flange showed a larger motion towards the proximal direction in the absence of PF force (Figure. 4.4B and E). Applying a single peak TF contact
force resulted in a slight decrease of peak micromotion under the SG load (Figure. 4.4C), while it noticeably increased the peak micromotion of the DKB loading case (Figure. 4.4F). The correlation between the nodal micromotions at the implant interface under the complete physiological load (TF+PF) with the loading condition without the PF force was strong, with an $R^2 = 0.90$ and $R^2 = 0.93$ for SG and DKB loads, respectively (Figure. 4.5A and B).

**Interference fit: 20 µm**

![Diagram A](image1)

**Interference fit: 200 µm**

![Diagram C](image2)

**Figure. 4.3** The bars in each bar diagram show two measures of micromotions (reference and resulting) for the models with an interference fit of 20 µm (A and B) and 200 µm (C and D). The graph lines represent the tibiofemoral (TF) and patellofemoral (PF) contact forces for both the stance phase of gait (A and C) and the deep knee bend (B and D). The red line indicates the flexion angle, which is given for the major points. The diagrams show how the magnitude of micromotions is affected by the contact forces and flexion angle.
The micromotions of the loading condition with the single peak TF contact force were also correlated with the complete loading cycle, which showed slightly weaker correlations with $R^2 = 0.73$ and $R^2 = 0.89$ for SG and DKB loads, respectively (Figure. 4.5C and D). Despite these rather high correlation coefficients, the micromotions were generally overestimated when neglecting the PF contact force (Figure. 4.5A and B) or by considering only the peak TF force (Figure. 4.5C and D) as can be seen by the slope of the correlations.

**Figure. 4.4** Lines (top subfigures) show the trajectory of a single node from the proximal part of anterior flange (blue dot in the subfigures of implant) for six different loading conditions during the fourth cycle of simulation (interference fit of 20 µm). The gray square indicates the starting increment and the circle represents the last increment. The bottom subfigures show the distribution of micromotions for each loading condition; posterior condyles are shown on top-left side of each image. It should be noticed that the colour bar map has a different maximum for the stance phase of gait and the deep knee bend activities.
Figure 4.5 Strong correlations were found between nodal micromotions at the implant interface under the complete physiological loading and the loading condition without the patellofemoral (PF) forces for both gait (A) and deep knee bend (B) (interference fit: 20 µm). The same correlation with the single peak of tibiofemoral (TF) contact force of gait (C) and deep knee bend (D). Note that the slopes for all fitted correlations are below 1.0, indicating that micromotions under complete physiological loading are overestimated.
4. Discussion

We developed an FE model of an implanted distal femur to predict micromotions at the bone-implant interface of an uncemented femoral knee component under different physiological loading conditions as a complementary guidance tool for in-vitro tests. Using this model, we investigated the effect of simplification of loading condition, whether peak micromotions occur during the peak TF force, and the effect of activity. In addition, we aimed to determine the most relevant regions to measure micromotions in an experimental set up. However, first it was important to determine an adequate measure of micromotion before answering each specific research question. As far as we know, only Van der Ploeg et al. [24] have utilized the same definition for resulting micromotion in a simulation of the hip stem, which was also shown to result in a greater range of values compared to other measures of micromotion. In the current study, the comparison between two measures of micromotion showed that the reference micromotions underestimated the micromotions as predicted over a complete cycle (resulting). This indicates that the definition of micromotion affects the magnitude of the predicted values, and it is therefore important to explicitly explain the technique to obtain the micromotions. Based on our results we would recommend computing the micromotion from a full cycle rather than with respect to the initial position in a computational model.

There are only a few studies that have investigated the primary stability of femoral knee implants[18, 19, 21, 30, 31]. One possible reason for the limited amount of research is probably the excellent long-term outcome of femoral knee implants [3]. However, recent applications of new materials such as ceramic and PEEK [31, 32], or the application of new porous materials at the interface [33] necessitates development of new pre-clinical testing approaches. In addition, the distal femur is subjected to a complex loading regime during different ADLs, in which the position and direction of the contact loads change constantly, which requires the utilization of a sophisticated experimental set-up. As already mentioned, most experimental set-ups are designed for a limited flexion angle, mainly representing gait, and usually patellofemoral contact forces are omitted[18, 19]. Only Wackerhagen et al. [30] designed a custom-made set-up to apply both the TF and PF force. However, the effect of the PF force on the primary stability is yet unknown. Therefore, the current FE study aimed to assess the effect of the simplification of loading conditions. We found that ignoring the PF contact force slightly increased the magnitude of the micromotions, with the most pronounced effect during the deep knee bend load. The predicted micromotions under the most simplified loading condition, applying a single peak TF force, showed increased overestimation of micromotions in comparison with the loading condition omitting only the PF contact force. However, based on the good correlation, the similar motion pattern, and the fact that the maximum average reference micromotion was found at the maximum TF contact force, we suggest that a simplified loading condition consisting of the peak TF force is an
acceptable alternative for the complex loading regime during experimental measurement of micromotions. Nonetheless, this study clearly shows that by applying a complete physiological loading profile, using simulation techniques or a sophisticated experimental set-up, more comprehensive and accurate data about the primary stability of these implants can be obtained.

In the current study, we combined an advanced micromotion analysis with an implant-specific physiological loading configuration, based on dynamic FE analyses fed with in vivo patient measurements, and validated against experimental data. Our results showed that the DKB loading case would result in a higher range of micromotions. We also found that both the magnitude and the angle of the TF force vector had a clear effect on the micromotions (Fig. 3). Conlisk et al. [19] demonstrated that the micromotion of a femoral knee implant increases with a higher flexion angle. Kassi et al. [17] reported a higher range of micromotions during stair climbing in a hip stem. Chong et al. [16] also found higher micromotions for a cementless tibial implant with stair climbing. Similarly, our results show that for a femoral knee implant the deep knee bend activity has a more detrimental effect on the primary stability than the stance phase of walking. Expansion of the range of activities and incorporating these in the methodology presented here may aid in improving implant safety and may assist implant design optimization for cementless femoral TKR components.

Relative to micromotion values reported in the literature, which were in the order of 0 to 250 μm, we found smaller ranges of micromotion [18, 19, 30]. This is attributable to the fact that we did not model plasticity or visco-elastic behavior of the bone, which are important factors for the primary stability of cementless implants [34, 35]. To include the effect of bone plasticity and the visco-elastic effects on implant stability, future studies can focus on improving a more realistic continuum model of the mechanical behaviour of bone. In addition, experimental studies used transducers to quantify the displacements between the bone and implant, while our micromotions were calculated at the actual interface, which also affects the comparison [36]. Therefore, it is important to develop an accurate experimental approach to capture micromotions as adjacent as possible to the interface for validation purposes.

Furthermore, we found that the peak micromotions occurred at the proximal part of the anterior flange, regardless of the loading conditions. Zelle et al. [20] reported higher shear stresses beneath the proximal part of the anterior flange particularly during deep flexion, which is consistent with our findings. This suggests that the proximal area of the anterior flange is the most relevant site to estimate the maximum range of micromotions during in-vitro tests. In addition, distal regions can provide an estimate of a minimum range of micromotions, which can indicate whether the necessary condition for bone ingrowth is present or not [10]. Hence, these two regions are suggested as the most relevant sites to estimate the range of micromotion of a femoral knee implant. From clinical point of view,
there is a higher risk of inadequate bone ingrowth due to larger micromotions at the anterior region. However, it should be realized that our simulations only represent the situation immediately post-operatively, and did not include the effect of progressive bone ingrowth. In the long term, secondary fixation due to bone ingrowth in locations with a low micromotion may provide the additional stability required to decrease micromotions at the proximal side of anterior flange, which was not simulated in the current study.

Obviously, this study has a number of limitations. In the current study, implant-specific kinematic data was used and applied to a single femur. Hence, an alternative implant design or femur with a different bone quality may result in a different outcome. The interference fit values were extracted from experimental data using a PU foam block. We were unable to find suitable data in the literature on an appropriate interference fit for a femoral knee implant, likely due to the complex nature of the measurement. Abdul-Kadir et al. [15] studied the interference fit in a femoral hip stem and concluded that small values of interference fit (10-100 µm of diametrical interference fit) are adequate to stabilize the hip stem to a large extent. Gebert et al. [35] investigated the effect of interference fit on the primary stability of cementless femoral hip resurfacing implants. They found that a diametrical interference fit of about 140 µm would result in an acceptable primary stability. Hence, we applied an interference fit that was close to reported values in the literature for other implant designs next to a more drastic situation. In addition, we found a migration pattern with a low value of interference fit, which indicates that a certain amount of interference fit is necessary to achieve adequate primary stability.

4.1 Conclusion

The present study has shown that the micromotion of a full cycle of activity would be larger than micromotion of any single load points with respect to the un-loaded phase. Therefore, it is important to use an adequate measure of micromotion for prediction of micromotion. We have found an overestimation of micromotion while applying the most simplified loading condition using the single peak TF force of activity. However, the maximum micromotions of an uncemented femoral knee implant, do indeed, occur at the peak of TF contact force. Furthermore, strong correlation between nodal micromotions under a simplified loading condition with a complete physiological loading was found and also the micromotion distribution was quite similar between different loading conditions. Therefore, we suggest that the single peak TF force of an ADL would be adequate for in-vitro tests to assess the primary stability. In addition, due to the clear effect of activity on the micromotions, it is essential to assess micromotion under different type of activities. Finally, we would recommend measuring in-vitro micromotion in the proximal side of anterior flange and distal regions to detect regions of high and low micromotion, respectively, to estimate the range of micromotion of a femoral knee implant.
References


CHAPTER FIVE

Experimental pre-clinical assessment of the primary stability of two cementless femoral knee components

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Abstract

To achieve long-lasting fixation of cementless implants, an adequate primary stability is required. We aimed to compare primary stability of a new cementless femoral knee component (Attune®) against a conventional implant (LCS®) under different loading conditions. Six pairs of femora were prepared following the normal surgical procedure. Calibrated CT-scans and 3D-optical scans of the bones were obtained to measure bone mineral density (BMD) and cut accuracy, respectively. Micromotions were measured in nine regions of interest at the bone-implant interface using digital image correlation. The reconstructions were subjected to the implant-specific’s peak tibiofemoral load of gait and a deep knee bend loading profiles. Afterwards, the implants were pushed-off at a flexion angle of 150°. Micromotions of Attune were significantly lower than LCS under both loading conditions (P ≤ 0.001). Cut accuracy did not affect micromotions, and BMD was only a significant factor affecting the micromotions under simplified gait loading. No significant difference was found in high-flex push-off force, but Attune required a significantly higher load to generate excessive micromotions during push-off. Parallel anterior and posterior bone cuts in the LCS versus the tapered bone cuts of the Attune may explain the difference between the two designs. Additionally, the rims at the borders of the LCS likely reduced the area of contact with the bone for the LCS, which may have affected the initial fixation.
1. Introduction

Total knee replacement (TKR) with an uncemented femoral and cemented tibial implant has shown promising survival [1-3]. Implant registries, such as the Australian Registry, have shown a 95% survival rate after 10 years [1]. However, revision rates are still higher for younger patients (<55 years) than for older patients (>75) (14% vs. 3% at 12 years) [1]. Therefore, there is still a need for improvement of implant survival, as younger and more active patients require more years of service and more demanding activities. Furthermore, recent high-flex designs introduce a new prospect for a wider range of motion and alternative loading conditions which have to be accommodated by the uncemented implant [4]. There are some reports that suggest these high-flex activities may have a detrimental effect on the long-term success of some TKR systems [5, 6].

To achieve a long-lasting fixation, an adequate primary stability is required [7-9]. The primary stability of uncemented femoral knee components, which is often referred to as the inverse of ‘implant-bone micromotion’, is affected by several factors, such as the type of activity [10, 11], bone quality [12], surgical preparation [13], and implant characteristics. Loading profiles applied in experimental studies assessing micromotions have mainly been based on a simplified gait cycle according to ISO standards [14-17], while higher ranges of flexion (>120°) are usually omitted. However, it has previously been shown that it is important to test implants for various loading configurations representing various activities of daily living [10, 18]. Periprosthetic bone quality [19], and the accuracy of the bone cuts [13], may also have a significant effect on the initial press-fit stability of the implant. Finally, implant characteristics such as roughness [20], shape, and material [21] also influence primary stability. In light of all these variables, it is crucial to assess the primary stability of new TKR implants in a pre-clinical stage.

In the current study we evaluate the primary stability of the Attune® (DePuy Synthes Joint Reconstruction, Warsaw, IN) cementless femoral component, and compared it against a conventional implant as a successful clinical benchmark. Our main objective was to evaluate the micromotions of these two femoral implants under simplified gait and deep knee bend (DKB) loads, while accounting for bone quality and bone cut accuracy. After the experiments the implants were pushed off from the bone at a high flexion angle to assess the fixation strength. We aimed to answer the following research questions: (i) What is the difference in primary stability of the two implant designs under simplified gait and DKB loading conditions? (ii) Do bone quality and bone cut accuracy have an effect on the primary stability? (iii) Is there a difference in micromotion-force relationship between the two designs under simplified gait and DKB loading conditions? (iv) What is the difference in primary stability and fixation strength of the two implant designs at a high flexion loading angle?
2. Materials and Methods

2.1 Study design

Based on a power analysis to determine correct sample size, six samples were adequate when a mean micromotion difference of 50 µm and standard deviation of 30 µm were assumed. However, twelve pairs of fresh-frozen cadaveric femurs were obtained from the Anatomy department of the Radboud university medical center to select bone specimens with better quality based on their x-ray images and visual inspection. Subsequently, six pairs of femurs (82-93 years [average: 86 years], 4 males) were selected.

The prototype Attune implant (DePuy Synthes Joint Reconstruction, Warsaw, IN) was available for the left side only, restricting randomization between left and right femurs (Figure. 5.1B). Conventional uncemented LCS® (low contact stress) femoral knee implants (DePuy Synthes Joint Reconstruction, Leeds, UK), which have a long-lasting clinical history [22], were allocated to the right femurs (Figure. 5.1A). Both implants were coated with a Porocoat® porous surface morphology (DePuy Synthes Joint Reconstruction, Warsaw, IN).

2.1.1 Femoral preparation. Bone specimens were thawed at room temperature (22°C) for 24 hours before the cutting session. An experienced orthopedic surgeon (WR) performed the bone cutting session following the normal surgical procedure and also determined the implant size, which resulted in one size 6 and five size 7 Attune components and two standard+ and four large LCS components.

2.1.2 Bone quality assessment. The bone specimens were 3D-scanned using computed tomography (CT) (530 mA, 120 kV, pixel spacing of 0.352mm, and slice thickness of 0.6mm...
EXPERIMENTAL PRE-CLINICAL ASSESSMENT OF THE PRIMARY STABILITY

; Siemens Somatom Sensation 64, Siemens AG, Germany), along with a hydroxyapatite calibration phantom (solid, 0, 50, 100, 200 mg/ml calcium hydroxyapatite, Image Analysis, Columbia, KY) to compute bone mineral density (BMD). We measured BMD values using a previously protocolized technique [20]. Briefly, the press-fit fixation mainly depends on the bone quality in the anteroposterior (AP) direction. Hence, regions of interests (ROIs) were selected in the anterior region and in the posterior condyles based on the nominal implant dimensions. The average BMD of the three ROIs was used for the statistical analysis.

2.1.3 Cut accuracy assessment. To investigate the effect of bone cut variation, 3D-optical scans of the resected bone specimens were made (ATOS 3D-scanner, GOM mbH, Braunschweig, Germany) (Figure. 5.2A). Each scan was compared with the nominal bone cuts in all cutting planes and the deviation from the nominal cutting planes was defined for twelve different bony regions (Figure. 5.2B).

2.1.4 Implantation. Implantation was done by a second surgeon (SvdG), who received an extensive training session for implantation of both designs. After implantation, the distal femurs were cut off 100 mm proximally from the most distal end, after which the proximal side of the specimen was cast in a cylindrical pot using bone cement for the fixation purpose in later stages.

Figure. 5.2 A) 3D optical scan of bone specimens were obtained. B) Twelve bony regions were defined to find the deviation of bone cuts from nominal cuts. Blue means over-resected (i.e. the remaining bone is smaller than nominal) and red means under-resected bone (i.e. the remaining bone is larger than nominal).

2.2 Mechanical tests
2.2.1 Preconditioning. Before performing measurements, to settle down the implants each specimen was preconditioned by applying a dynamic loading regime using a servo-hydraulic testing machine (MMED, MATCO, La Canada Flintridge, CA) for 15 minutes at 1 Hz. The loading regime was the same as the applied load configuration during the micromotion measurement, which is explained in the next section. The preconditioning phase was followed by a minimum period of 15 minutes of relaxation.

2.2.2 Loading conditions. Due to the potential effect of the loading configuration on the micromotion, an implant-specific loading profile was applied [18]. In collaboration with Denver University, implant-specific loading profiles of the stance phase of gait and a deep knee bend (DKB) were derived from a validated Finite Element (FE) model of the Kansas knee simulator [18, 23], which successfully reproduced in vivo forces measured in the Orthoload database [24]. The resulting loading profiles were slightly different for the two implant designs. An important difference was that the tibiofemoral (TF) contact force of gait profile was distributed evenly over the condyles of Attune, while the LCS design had a higher medio-lateral load ratio.

For our experimental set-up, a simplified loading configuration was used, based on the implant-specific loading profiles. Our previous finite element (FE) study [18] indicated that micromotions under a single peak TF force are representative for a full cycle of activity. Therefore, the peak TF force of each activity (gait and DKB) was reproduced in the experimental set-up by applying a compressive force-controlled load at a rate of 100 N/sec (Figure. 5.3A and B; Table. 5.1).

However, for the DKB configuration a pilot test showed that the femur would fracture at the proximal fixation when applying the full TF force. Hence, only 40% or 50% of the peak DKB load was applied to small or large implant sizes, respectively, to avoid bone fractures. (Table. 5.1).

Table 5.1 Specification of loading profiles.

<table>
<thead>
<tr>
<th>Implant</th>
<th>Loading configuration</th>
<th>Peak tibiofemoral contact force (N)</th>
<th>Applied percentage of peak load (%)</th>
<th>Lateral-medial ratio (%)</th>
<th>Angle (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Attune</td>
<td>Gait</td>
<td>2,376</td>
<td>100</td>
<td>50-50</td>
<td>15</td>
</tr>
<tr>
<td></td>
<td>Deep knee bend</td>
<td>2,046</td>
<td>40 or 50*</td>
<td>42-58</td>
<td>90</td>
</tr>
<tr>
<td>LCS</td>
<td>Gait</td>
<td>2,223</td>
<td>100</td>
<td>35-65</td>
<td>16</td>
</tr>
<tr>
<td></td>
<td>Deep knee bend</td>
<td>2,156</td>
<td>40 or 50*</td>
<td>45-55</td>
<td>90</td>
</tr>
</tbody>
</table>

* The applied percentage of load was based on the implant size; Attune size 6 or 7 and LCS size standard+ or large were applied with 40 or 50% of the peak tibiofemoral load, respectively.

Matched size tibial inserts of the two designs were utilized to apply the load (Figure. 5.3A and B). The tibial insert was connected to a free-sliding jig (LWR 6150, SKF, Germany),
allowing it to freely find its position on the femoral implant.

2.2.3 High-flex push-off load. Since both implants were designed for a high-flex range of motion, we compared their fixation strength in a high-flex position using the same set-up as our previous study [20]. Shortly, a push-off load simulating 150° of flexion was applied to the posterior condyles by a custom-made load applicator (Figure. 5.3C). We applied a displacement-controlled load at a rate of 1 mm/minute. The force-displacement graph was recorded to find the maximum required load to loosen the implant.

![Image](image.png)

**Figure. 5.3** A) Experimental set-up under the peak load of gait loading condition. Two regions of interest (ROIs) were the posterior condyles and schematic representation of an image from these ROIs is shown with two black squares. B) Experimental set-up to apply peak load of deep knee bend (DKB) load. C) High-flex push-off test. D) A captured image (size: 11.8 mm × 8.9 mm) from distal region with two sampling squares of 255 × 255 pixels on the implant and adjacent bone, which shows the displacement distribution per sampling square.

2.3 Bone-implant interface micromotion measurement

2.3.1 Digital image correlation. Digital image correlation (DIC) is a relatively novel tool, which previously has mainly been applied to explore full-field displacements and strain fields [25, 26], but is also very useful to calculate implant-bone micromotions, very near to the actual interface [27]. It provides flexibility in the experimental set-up as it is a ‘no-contact technique’, which facilitated the application of different loading configurations. DIC software uses image correlation of speckle patterns applied to the studied object. To create a random speckle pattern for the DIC measurement, the specimens were thoroughly cleaned from soft tissue, especially near the interface. Then, a smooth layer of white paint (Sencys, Maxeda DIY Group, Amsterdam, Netherlands) was sprayed in three intervals, which was followed by a single black paint spray to apply a speckle pattern (Figure. 5.3). When the paint was dried, the specimens were kept in a blood analog solution for 12 hours to avoid potential negative effects of dehydration on bone mechanical properties [28].

We used a single SPOT™ Insight 2.0 Color digital camera (SPOT TM Imaging Solutions, Diagnostic Instruments, MI) with a 1” format CCD sensor (Kodak KAI-2020-CM). To achieve a higher magnification a convertor was also used, which resulted in the final resolution of 7.4 μm/pixel and field of view was 11.8 mm × 8.9 mm. Based on a pilot study, in which
using a micrometer known displacements was applied, at this resolution the root mean square error of the technique to measure micromotion was found to be 2.82 μm. The measurement surface was uniformly illuminated using a high intensity ring illuminator (Figure. 5.3A). The aperture was adjusted on f/11 and exposure time was 10 ms. The same focal depth on both the bone and implant surfaces was required to capture the sharpest possible speckle pattern. Therefore, we determined the locations with the lowest level difference between bone and implant in each region of interest (ROI). To compensate for potential pixel size variation caused by images being partially out of focus, an additional image of a calibration indicator was captured before each measurement. The post analysis was done using a free software Opticist 0.995 (Moiré Analysis Software) in sampling squares of 255 × 255 pixels (about 3 mm²) located at the implant and adjacent bone, as close as possible to the interface (the average distance between two sampling squares was about 1mm) (Figure. 5.3D). Micromotion at the sampling location was calculated by subtracting displacements of bone and implant. The provided displacements were in the image coordinate system. To determine pure shearing motions at the interface, the measured micromotions were transformed to the normal and shear directions based (Figure. 5.4). In this study we only present shear motions, as normal motions were small compared to the acceptable limit for bone ingrowth (<500 μm is acceptable [29]), and no differences were found between the two implant designs.

2.3.2 Measurements. A custom-made device was developed which allowed us to freely move the camera in three dimensions. In addition, specimens were fixed within a rotating clamp to easily re-position them. Hence, it was possible to define nine ROIs comprised the

![Figure 5.4](image_url)Micromotions were transformed to the shear and normal directions at the interface. From anterior and posterior view, only in-plane micromotion was measurable. From medial and lateral views, both components were measurable (black arrow shows the normal component of micromotion, while white arrows show the shear micromotion).
proximal tip of the anterior flange (ANT); the anterior, distal, and posterior regions from the medial and lateral views (MA, LA, MD, LD, MP, and PL); the posterior condyles from the posterior views (LP and MP) (Figure. 5.3A and 4). These ROIs enabled us to measure the displacements in mediolateral direction (from anterior and posterior views), distal-proximal direction (all views) and anteroposterior direction (medial and lateral views).

In each ROI, first a calibration image was taken, after which a second photo was taken representing the unloaded phase, then, the compressive load was applied. When the load reached its maximum, a third photo was taken, representing the loaded situation. Each measurement was repeated three times in the same order using the same load (peak TF force), resulting in 27 measurements per specimen (3 repeated measurements at 9 ROIs). We report the average of three measurements per ROI.

In addition, to find the relationship between the force and micromotion, a sequence of DIC-images (1 image per two seconds), synchronized with the applied load, was captured from the proximal tip of the anterior flange from the anterior view.

It was not possible to take photos of the condyles from the posterior views under the simplified DKB load due to obstructions by the experimental set-up (Figure. 5.3B), which resulted in seven ROIs. Again, a sequence of images (1 image per second) synchronized with the applied load was captured from the proximal tip of the anterior flange.

2.3.3 Push-off loading condition. To find the relationship between applied push-off load and the micromotion, a sequence of synchronized DIC-images with the applied load was obtained (Figure. 5.3C). The DIC-images were captured only from the posterior condyles, mainly from the medial view. The medial view was selected as a worst-case location because of the higher ratio of load on the medial side during the two previous loading configurations (gait and DKB), which could cause some bone damage (the lateral view was used in two cases due to better quality of the DIC-images). In addition, the ROI of the posterior condyle was chosen because the load was applied directly to the condyles.

Based on the synchronized recordings, the force required to reach a micromotion of 50 μm (threshold for bone formation [9]) and 150 μm (threshold beyond which fibrous tissue formation occurs [7]) was determined.

2.4 Statistical tests

Normality (Shapiro-Wilk) and homogeneity of variances (Levene’s test) were checked. A log-transformation was used in cases of non-normally distributed data or rejection of the equality of variances assumption. A univariate analysis of variance (ANOVA) with micromotion as the dependent variable and implant design as the independent variable, and the BMD and cut accuracy were covariates. A paired Student’s t-test was used to compare BMD and bone cut accuracy between left and right femurs. A P-value of 0.05 was defined as significant.
3. Results

3.1 BMD measurement

BMD was normally distributed over the femurs with a range of 68-201 mg/cm³. There was no significant difference between groups with an average density of 136 mg/cm³ and 126 mg/cm³ for right (LCS) and left (Attune) femurs, respectively.

3.2 Bone cut accuracy

Since the overall anterior-posterior size of the bone has the largest effect on the AP fit of the implant, the average of regions 3, 4, 11, and 12 (Figure. 5.2B) is presented here. The average deviation from the nominal bone cut in these regions was -0.17± 0.25 mm (mean± SD) and -0.08± 0.19 mm (mean± SD) for LCS and Attune, respectively. This means that the femurs were over-resected, but the difference between the two groups was not statistically significant.

3.3 Bone-implant interface micromotion measurement

3.3.1 Peak TF force of gait loading. Two measurement points were missed from the posterior view of lateral condyle for both the Attune and the LCS due to absence of bone at the edge of the posterior condyle. The micromotions were not normally distributed, and the equality of variances assumption was rejected, due to the large variation in the various ROIs. Therefore, for the statistical analysis the micromotions were transformed using a log transformation. Comparison between the nine ROIs of each implant design showed similar patterns, with the largest micromotions occurring in the medial view of the anterior flange (MA) region, followed by the tip of anterior flange (ANT) (Figure. 5.5). Distal and posterior regions did have a similar range of micromotion. We compared the micromotions of two designs, both per ROI and also as the average of all nine ROIs. ANOVA showed that BMD was a significant factor (P= 0.025), but the deviation from the nominal bone cut was not a significant factor (P= 0.443), indicating that the variation of bone-cut accuracy did not affect micromotions in this study. Therefore, the micromotions were corrected only for the effect of BMD. In five ROIs, significantly higher micromotions were found for LCS than for Attune (ANT, MA, MD, MP, and PM; Figure. 5.5). Overall, Attune had significantly lower micromotions than the LCS under the peak TF load of gait (Table. 5.2).
Table 5.2 Mean and 95% confidence interval (CI) of micromotions under simplified gait and deep knee bend loading conditions.

<table>
<thead>
<tr>
<th>Implant</th>
<th>Loading configuration</th>
<th>Mean micromotion (μm)</th>
<th>95% CI (μm)</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Attune</td>
<td>Peak TF load of Gait</td>
<td>32</td>
<td>25 - 41</td>
<td>*P&lt; 0.001</td>
</tr>
<tr>
<td>LCS</td>
<td>71</td>
<td>56 - 91</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Attune</td>
<td>Reduced peak TF load of DKB</td>
<td>55</td>
<td>46 - 66</td>
<td>*P= 0.001</td>
</tr>
<tr>
<td>LCS</td>
<td>83</td>
<td>70 - 99</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

![Figure 5.5](image)

**Figure 5.5** Non-log transformed mean of the regional micromotion of LCS and Attune under gait loading condition (ANT: tip of anterior flange; M: Medial view; L: Lateral view, D: Distal region, and P: posterior condyles; PM and PL: medial and lateral condyles from the posterior view). The error bars show standard deviation. The p-values are based on log-transformed data.

### 3.3.2 Simplified DKB loading

The micromotion distribution was not normal for the LCS and the test of equality of variances was rejected. Therefore, data was transformed using a logarithmic function. Neither BMD (P= 0.728) or bone cut deviation (P= 0.081) had a significant effect on the micromotions. The highest micromotions were found at the proximal tip of the flange from the anterior view (ANT) of both designs, followed by the lateral view of the flange (LA) for the LCS, and the medial view of the distal region (MD) for the Attune (Figure. 5.6). The lowest micromotions were measured at the posterior condyles from the medial and lateral views, for both implants. We found a significantly higher micromotion for LCS than Attune in four ROIs (ANT, LA, MD, and LD; Figure. 5.6). Similar to the micromotion results under peak TF force of gait, Attune had a significantly lower micromotion than LCS (Table. 5.2).
3.3.3 Push-off test. BMD was a significant factor for the push-off load (P< 0.001), but not bone cut deviation (P= 0.687). The average loosening force corrected for the effect of BMD was 1,597 ± 502 N and 1,579 ± 783 N (mean± SD) for Attune and LCS, respectively, which was not significantly different (P= 0.757). However, when analyzing the required force to reach a micromotion of 50 and 150 μm, it was found that a significantly higher force was demanded for Attune than for LCS (Table. 5.3). For this comparison, BMD was not a significant factor (P= 0.504 and 0.649).

Table 5.3 Mean and 95% confidence interval (CI) of required load to reach micromotion of 50 and 150 μm during high-flex push-off test.

<table>
<thead>
<tr>
<th>Implant</th>
<th>Micromotion threshold (μm)</th>
<th>Force (N)</th>
<th>95% CI (N)</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Attune</td>
<td>50</td>
<td>501</td>
<td>374 - 628</td>
<td>*P= 0.015</td>
</tr>
<tr>
<td>LCS</td>
<td></td>
<td>263</td>
<td>136 - 390</td>
<td></td>
</tr>
<tr>
<td>Attune</td>
<td>150</td>
<td>784</td>
<td>606 - 961</td>
<td>*P= 0.002</td>
</tr>
<tr>
<td>LCS</td>
<td></td>
<td>322</td>
<td>145 - 499</td>
<td></td>
</tr>
</tbody>
</table>

3.3.4 Relationship between micromotion and load. We found different relations between micromotion and applied load, depending on the loading configuration and implant design (Figure. 5.7). Generally, the relation was non-linear relationship under peak TF load of gait...
4. Discussion

This experimental study aimed to compare the primary stability of a new cementless femoral knee component with a successful conventional design. Our study showed a lower range of micromotion for the Attune design as compared to the LCS under both loading configurations, and also during the high-flex push-off test.

To explain the differences in micromotions, a closer analysis of the two designs is required. Both designs had an identical surface treatment, but the anterior and posterior cuts in the LCS were parallel, while the bone cuts of Attune were slightly tapered (5° anteriorly and 1° posteriorly). Possibly, the resulting tapered shape provided more resistance under the applied loads. The bone cuts may also explain the differences in the force-displacement curves during the push-off tests (Figure. 5.8). Although the first section of the two curves was quite similar, afterwards the LCS began to loosen at a lower force, which may have been a result of the parallel cuts. It should be highlighted that the LCS was completely loose before reaching the maximum push-off force, which was not the case for Attune.

Another difference between the two designs was the smaller surface area in contact with bone available for the LCS due to the presence of a rim at the borders of the implant, originating from the cement pocket of the cemented version (Figure. 5.1). This reduced contact area at the interface may have resulted in a sub-optimal press-fit fixation, which in turn may have affected the primary stability. Finally, since implant-specific loading profiles were used, a higher ratio of load was applied medially to the LCS, which was also where the highest
micromotions occurred. However, although the DKB loads were quite similar for both designs, micromotions still were significantly higher for LCS. We can therefore not attribute the differences in micromotions solely to the implant-specific loading profiles.

![Figure 5.8](image)

**Figure 5.8** A typical force-displacement graph of two implants during push-off test belongs to the same pair. The implants behaved similar until the first section (indicated in the image), but afterwards different trends were observed.

A limited number of studies investigated the primary stability of femoral knee implants [14-17, 30], mainly using uni-directional motion sensors. Wackerhagen et al. [30] found micromotions in a range of 0 to 162 µm for uncemented implants under an axial tibial load of 700 N. Micromotions were measured at the tip of the anterior flange, only in the anteroposterior direction, which we defined as the normal component of micromotion, rather than shear micromotion reported here. Nonetheless, the normal micromotions in our study, recorded from the medial and lateral view of the anterior flange during the reduced peak TF load of DKB (MA= 161 ± 61 µm (Attune) (mean± SD) and 112 ± 99 µm (LCS); LA= 67 ± 39 µm (Attune) and 61 ± 33 µm (LCS)), showed a similar range of values. Cristofolini et al. [16] measured micromotions of a cemented implant under a cyclic load of 168-2600 N, simulating a gait cycle. Their measurements were done at the anterior flange and posterior condyles, in the proximal-distal direction. In contrast to our findings, they found the lowest micromotions at the anterior flange (<20 µm) and the highest values posteriorly (average of 100 µm). The difference between their results and our findings is likely caused by the difference in implant fixation (cemented vs. non-cemented). However, in general, the ranges of micromotion were quite similar. Conlisk et al. [17] quantified several directions of motion in uncemented and cemented designs, in a mechanical set-up with a maximum compressive load of 1643 N. Their study also showed the difference in directions of micromotion, with the
maximum micromotion in the proximal-distal direction, matching our measurement during the gait loading. In addition, the range of loosening force found in this study was close to our previous findings [20].

Although LCS has a good clinical record [22] we found relatively high micromotions in some regions for the LCS, which even exceeded the threshold reported for bone ingrowth [7]. Evidently, our current findings are representative only for the immediate post-operative situation. This suggests that in a clinical setting, osseointegration of the implant at locations with low micromotions may have a secondary stabilizing effect, thereby reducing micromotions in the regions that displayed high micromotions in our experimental set-up. In addition, although the Attune had a lower range of micromotion, further clinical evaluation is necessary to assess its performance on the longer term.

Several studies have demonstrated the effect of bone quality on the primary stability [31, 32]. Interestingly, we found that BMD was a significant factor only under gait loading. These observations indicate that bone quality has an interaction with the applied load. Possibly, these differences are due to bone anisotropy, as the two loading profiles acted in different directions (flexion angles of 15° and 90° for gait and DKB loads, respectively), loading the bone under different angles. In addition, we quantified the bone quality only in terms of BMD, which does not provide all information on its local mechanical properties. Furthermore, similar to our previous results [20], we found a significant correlation between BMD and push-off force ($R^2= 0.85; P< 0.001$). However, there was no correlation between BMD and required force to reach a micromotion threshold of 50 or 150 µm. Based on these results we suggest that the effect of BMD on micromotions is less obvious than on fixation strength.

We did not see a significant effect of bone cut variation on the micromotion, probably due to the relatively small amount of variation in cuts. In addition, the cut accuracy only provided information about the initial interference fit, which may be affected during implantation due to abrasion, visco-elasticity, and plastic deformation of the bone, leading to a reduction of the ultimate effect of bone cut imprecision. Additional studies are required to evaluate the effect of interference fit and the interaction of bone and implant during implantation in more detail.

### 4.1 Limitations

Obviously there is a number of limitations that may have influenced the current study. The applied loading conditions consisted of only the peak TF force of each activity, based on implant-specific kinematics. Therefore, we lacked a full loading cycle for both activities as the gold standard [18], and omitted the effect of the PF force. It is likely that the PF force would have a stabilizing effect, and therefore the current loading configuration may represent a worst-case loading scenario [18]. Due to limitations with the DIC set-up it is currently not possible to analyze a full loading configuration. Another limitation was application of
a reduced portion of the peak load of DKB, to reduce the risk of bone fracture. In vivo the muscles and soft tissue structures surrounding the knee joint may prevent the fractures that we experienced in our experimental set-up. However, as our goal was to compare the two designs, it was necessary to lower the loads to be able to evaluate the effect of TF loads in a DKB position.

A limitation to the measurement approach is that the DIC system does not differentiate between actual micromotions and possible deformations occurring in the bone. Bone deformations within the sampling region are partially accounted for in the image correlation procedure, while the bone deformation between the two sampling regions may have affected the micromotions. However, by choosing sampling points as close to the actual interface as possible, the effect of bone deformation on the micromotion measurements was limited to a minimum. In addition, the measurements in the different ROIs were not performed simultaneously, as our system consisted of a single camera only. Consequently, the loading profiles needed to be repeated for measurements in each ROI. Although we tried to minimize variations by preconditioning the reconstructions prior to performing the actual measurements, there may have been slight differences between the measurements in the nine ROIs.

Finally, another limitation to our study is the age of the femora used in the experiments (82-93 years). Although the typical TKR patient usually is younger (under 70 years), unfortunately the availability of femurs of this age category is very limited at our institution. It is therefore very likely that the bone quality in our specimens was lower than in the typical TKA patient. As a result, the micromotions presented here may represent a worst-case scenario in terms of patient bone quality.

4.2 Conclusions

The primary stability of two implant designs were tested under simplified gait (peak TF load) and DKB loading profiles, which showed lower range of micromotion for Attune compared to LCS under these specific loading conditions. In addition, the mechanism of implant loosening under a high-flex push-off load was investigated, which showed no difference in terms of fixation strength between the two designs, but the mechanism of loosening was different. A significantly higher load was required for Attune to reach the micromotion threshold of 50 and 150 μm as compared to LCS. Bone quality had a minor effect on micromotions, while no effect of cut accuracy was found. We observed different non-linear and linear relationships between micromotions and applied load, depending on the type of loading, and implant design.
Acknowledgments: This study was supported by a research grant from DePuy Synthes Joint Reconstruction (Leeds, UK). We would like to thank Dr. W. Reng and Dr.ing. S. van de Groes for their support to cut the bone specimens and implantation session. We also gratefully acknowledge M. Miller, I. Romme, and D. Strijker for their assistance with the DIC set-up.
References


[12] Aro HT, Alm JJ, Moritz N, Makinen TJ, Lankinen P. Low BMD affects initial stability and delays stem osseointegration in cementless total hip arthroplasty in women: a 2-year


[25] Dickinson AS, Taylor AC, Ozturk H, Browne M. Experimental validation of a finite element model of the proximal femur using digital image correlation and a composite bone


EXPERIMENTAL PRE-CLINICAL ASSESSMENT OF THE PRIMARY STABILITY
CHAPTER SIX

Evaluation of interference fit and bone damage of an uncemented femoral knee implant

Berahmani, S., Hendriks, M., de Jong, JJ., van den Bergh JP., Maal, T., Janssen, D., & Verdonschot, N.

Under review (Clinical Biomechanics)
Abstract

During implantation of an uncemented femoral knee implant, press-fit interference fit provides the primary stability. It is assumed that during implantation a combination of elastic and plastic deformation and abrasion of the bone will occur, but little is known about what happens at the bone-implant interface and how much press-fit interference fit is eventually achieved. Five cadaveric distal femora were prepared and femoral knee components were implanted by an experienced surgeon. Micro-CT- and conventional CT-scans were obtained pre- and post-implantation for geometrical measurements and to measure bone mineral density. Additionally, the position of the implant with respect to the bone was determined by optical scanning of the reconstructions. By measuring the differences in surface geometry, assessments were made of the cutting error, the actual interference fit, the amount of bone damage, and the effective interference fit. Our analysis showed an average cutting error of 0.67±0.17 mm, which pointed mostly towards bone under-resections. We found an average actual AP interference fit of 1.48±0.27 mm, which was close to the nominal value of 1.5 mm. We observed combinations of bone damage and elastic deformation in all bone specimens, which showed a trend to be related with bone density. Higher bone density tended to lead to lower bone damage and higher elastic deformation. The results of the current study indicate different factors that interact while implanting an uncemented femoral knee component. This knowledge can be used to fine-tune design criteria of femoral components to achieve adequate primary stability for all patients.
1. Introduction

The primary fixation of an uncemented femoral total knee replacement (TKR) component is achieved by a press-fit placement of the implant. The objective is to restrict micromotions at the bone-implant interface, which is a prerequisite for successful bone-ingrowth and stable biological fixation on the long term [1-3]. The press-fit fixation is obtained due to the fact that the inside dimensions of the implant are slightly undersized with respect to the bone cuts created intra-operatively. This relative difference in dimensions is commonly known as the interference fit.

The actual interference fit is often different from the nominal interference fit, as it depends on the final implant position (which does not necessarily line up with the pre-planned position), and on the accuracy of the prepared bone cuts. In addition, during the implantation process compressive stresses are being built up inside the bone and at the implant-bone interface. The combination of the compressive stresses, the shearing motion at the implant-bone interface and the rough implant surface will cause plastic deformation and abrasion of the bone, effectively altering the actual interference fit [4]. However, still a considerable amount of elastic energy is stored in the bone, responsible for the final effective interference fit and subsequent stability of the implant [5].

The amount of interference fit designed in the implant system can be influenced by careful tuning of the intrinsic accuracy of the instruments that are used to prepare the bone cuts [6] and position the implant [7, 8], which in turn may allow for compensation of inalterable patient-related factors, such as low bone density to optimize stability [9, 10]. Surprisingly little is known about the actual and effective interference fit, or on how much bone damage actually occurs during implantation of the femoral component. Therefore, interference fit is usually ignored or simplified when computational models based on finite element analysis (FEA) are implemented to pre-clinically test implants [11-13], limiting the FEA capability to entirely evaluate new implant designs. Therefore, the main goal of the current study was to assess the actual and effective interference fit and amount of bone damage occurring during implantation of an uncemented femoral knee component. For this purpose, several optical and imaging techniques were used to measure geometrical changes occurring to the bone during the procedure.

2. Materials and methods

In short, components were implanted on cadaver femurs, while before, during, and after implantation a series of scans were made that allowed for monitoring the changes that took place in the underlying bone. In the following sections, the materials and methods used for these analyses will be described, as well as all the sequential steps that were taken.
2.1 Implants

The cementless Sigma® cruciate retaining femoral knee implant (DePuy Synthes Joint Reconstruction, Leeds, UK) was used. This implant has a porous surface coating, Porocoat® (DePuy Synthes Joint Reconstruction, Warsaw, IN, USA), which is comprised of randomly arranged spherical beads. The implant system has a nominal interference fit of 0.75 mm at the anterior, distal, and posterior cut planes, which is equal to the average thickness of surface coating.

2.1.1 Assessment of implant deformation. To capture the deformation of the implant during implantation, four strain gauges were connected to the outer surface of each implant medially and laterally, in the anterior and posterior region (Figure 6.1A). Strain gauges were placed at locations nearest to where peak strains were expected to occur (based on previous Finite Element (FE) analyses [11]). Extra care was taken not to position them at locations where the implantation instruments contacted the implant. Strain was recorded using QuickDAQ (Data Translation GmbH, Germany) at 100 Hz, a few seconds before implantation until 10 seconds after implantation. The average strain of the last 10 seconds was used for the analyses.

2.2 Bone specimens and implantation procedure

Two pairs and one single fresh-frozen cadaveric femora (85± 3 years old; 1 pair was male) were selected from ten donated femurs to exclude highly osteoporotic bones by assessing x-ray images (provided by the Anatomy Department of the Radboud university medical center). The bone specimens were thawed at room temperature and an experienced orthopaedic surgeon made the bone cuts following the normal surgical procedure using standard intramedullar instrumentation. A femoral cutting block was fixed with two threaded pins to first resect about 9 mm of distal femur from the most prominent distal part of condyle, followed by determination of the required implant size, which resulted in four size 5 implants and one size 3. All cuts were made using an oscillating saw with a blade thickness of 1.47 mm (DePuy Synthes, Leeds, UK) and two holes were drilled for the femoral pegs. The implants were subsequently placed by the same surgeon.

2.3 Scanning procedures

2.3.1 CT-scanning. To evaluate the bone mineral density (BMD), after the cutting session (but before implantation), the bone specimens were 3D-scanned using computed tomography (CT) with in-plane resolution of 0.351 mm and slice thickness of 0.6 mm (530mA; 120Kv; Siemens Somatom Sensation 64, Siemens AG, Germany). To convert the Hounsfield Units to BMD, a hydroxyapatite calibration phantom (solid, 0, 50, 100, 200 mg/ml calcium hydroxyapatite, Image Analysis, Columbia, KY) was scanned along with the cadavers.
BMD was measured in three volumes of interest in the anterior region (ANT) and posterior condyles (lateral (PL) and medial (PM)) using a previously protocolized technique [10]. The anterior and posterior regions were selected due to their main role in providing initial fixation in the anteroposterior (AP) direction.

### 2.3.2 Micro-CT scanning

To capture the bone damage occurring during implantation as accurately as possible, after the cutting session, micro-CT scans of the resected bone specimens were made (XtremeCT II, SCANCO Medical AG, Bruettisellen, Switzerland; peak voltage of 68kVp, tube current of 1470 µA, 200 ms integration time). All images were reconstructed using an isotropic voxel size of 60.7 µm. These scans served as the baseline for the damage assessment. Next, the femoral components were placed on the femurs as described earlier. After implantation and performing a series of measurements on the specimens, the implants were split through the condyles using an electric diamond-blade cutting machine, which allowed us to gently remove the components without causing additional bone damage (Figure 6.1B). Subsequently, the micro-CT scans of the distal femurs were repeated, to evaluate the amount of damage. Using the micro-CT data sets, surface meshes of the outer surface of the bone specimens (pre- and post-implantation) were created using medical imaging software (Mimics 14 and 18 & 3-matic®, Materialise, Leuven, Belgium) for further analyses.

![Figure 6.1](image)

**Figure 6.1** A) Four strain gauges were connected to each implant to record implant’s deformation. B) Implants were sawed using a diamond blade to access bone surface. C) The regions of interest are demonstrated (green area) in the anterior flange and posterior condyles (ROI is only shown in one condyle).

### 2.3.3 Optical scanning

To determine the relative position of the implant with respect to the bone, after implantation (Figure 6.2A), optical scans were made of the reconstructed distal femurs (TRIOS Color-P13, 3Shape, Copenhagen, Denmark) (Figure 6.2B). The accuracy of
the system as provided by the manufacturer was 100 µm.

Figure 6.2 A) After implantation, an optical scan of bone and implant was made (B), which was utilized to determine the position of the implant with respect to the bone (C).

2.4 Outcome parameters

2.4.1 Cutting error. The nominal cutting planes were provided by the manufacturer, which were used to assess cutting errors. For this purpose, the nominal distal cutting planes were superimposed onto the distal face of pre-implantation bone. As during surgery the distal cut is the first cutting plane, this surface was taken as the reference plane. Since the main press-fit is provided in the AP direction, deviations in the anterior and posterior regions were of main interest. Therefore, three regions of interest (ROIs) corresponding to the shape of anterior flange and posterior condyles were defined on the bone surface: anterior (ANT), the medial (PM) and lateral (PL) posterior condyle (Figure. 6.1C). Using a Matlab script (Matlab 7.12.0 (R2011a), Mathworks, MA, USA), the AP distance between each point from the cutting planes and the bony surfaces was calculated per ROI by finding two closest points from two surfaces and then extracting the AP distance between them. In this way, the maximum and average AP distance was obtained per ROI.

2.4.2 Actual interference fit. To evaluate the actual interference fit, CAD files of the implants were provided by the manufacturer. Using an iterative closest point (ICP) algorithm, the surface mesh of the implant (based on the CAD file) and bone (based on the pre-implantation micro-CT) were superimposed on the optical scan of each reconstruction (Figure. 6.2C). The average AP distance between the surface of the pre-implantation femur and the internal surface of implant was determined for the defined ROIs. This measure was taken as the actual interference fit after implantation.

2.4.3 Bone damage – outer surface. To evaluate the amount of damage caused during the implantation process, the surface meshes of the pre- and post-implantation micro-CT scans were registered on top of each other using the ICP algorithm. Again, the average AP distance
between the surface of the pre- and post-implantation femurs was determined for the defined ROIs.

2.4.4 Bone damage – internal. In addition to the surface analyses, visual inspection of the actual pre- and post-implantation micro-CT scans was performed to evaluate the bone damage at the micro-level. For this purpose, the micro-CT scans were visually assessed to investigate changes in trabecular structure, abrasion, and possible micro-fractures. Since during the repetitive bone scans the specimens were not placed in the exact same orientation, micro-CT slices from the same distance from the most distal slice were selected and compared.

2.4.5 Effective interference fit. To measure the elastic deformation of the bone which is basically a reflection of the effective interference fit, the AP distance between the surface mesh of post-implantation scan and the correspondent node from internal surface of implant was obtained.

2.5 Data analysis

Outcome parameters (cutting error, actual interference fit, effective interference fit, and bone damage) per ROI were demonstrated in two manners: a frequency plot with a surface fraction on the vertical axis, and a distribution map on the surface mesh of the pre-implantation micro-CT. It should be noted that the areas with prosthetic overhang in the flange area was excluded in the calculation of the average values in the anterior ROI. In addition, the average of each of the outcome parameters in the AP direction was obtained by first averaging the average of the posterior regions and then summed it with the average of anterior region; hence, these values represent each sample instead of each ROI. Linear regression analysis was used to evaluate the correlation between each outcome parameter and BMD.

3. Results

3.1 Bone density

The BMD in the three volumes of interest are given with the average value in Table.6.1. The distribution of BMD showed a wide range from low to high bone density. The bones within the two pairs had similar values.

3.2 Cutting error

None of the specimens had a perfect resection, with a tendency towards more under- than over- resected bone, which occurred randomly in both anterior and posterior regions (Figure.6.3A). By comparing all ROIs, the PL region of specimen B-Right had the maximum deviation from
the nominal cutting planes (under-resection) with 0.72± 0.25 mm (mean± standard deviation) was found in. The largest average deviation in the AP direction was found in specimen B-Left (0.85± 0.24 mm). No correlation was found between BMD and cutting error (Figure.6.4A).

Table 6.1. Bone mineral density (BMD) for the three volumes of interest (VOI) is given. In addition, average BMD per specimen is also stated.

<table>
<thead>
<tr>
<th>Specimen-side (age, sex)</th>
<th>BMD per VOI (mg/cm³)</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Anterior</td>
<td>Posterior-medial</td>
<td>Posterior-lateral</td>
<td>Average</td>
</tr>
<tr>
<td>A-Left (83, F)</td>
<td>170.84</td>
<td>176.2</td>
<td>221.79</td>
<td>189.61</td>
</tr>
<tr>
<td>B-Left (89, F)</td>
<td>121.44</td>
<td>96.8</td>
<td>143.1</td>
<td>120.45</td>
</tr>
<tr>
<td>B-Right (89, F)</td>
<td>139.33</td>
<td>100.62</td>
<td>184.14</td>
<td>141.36</td>
</tr>
<tr>
<td>C-Left (82, M)</td>
<td>256.83</td>
<td>233.94</td>
<td>314.27</td>
<td>268.35</td>
</tr>
<tr>
<td>C-Right (82, M)</td>
<td>285.08</td>
<td>284.89</td>
<td>384.65</td>
<td>318.21</td>
</tr>
</tbody>
</table>

3.3 Actual interference fit

A more homogenous distribution of the actual interference fit in the posterior and anterior regions was found in comparison to the cutting error, which can be seen from both the frequency graphs with the three lines overlaying each other and the distribution maps, with a more homogenous colour in the three ROIs (Figure.6.3B). The highest interference fit in the AP-direction was found for specimen C-Right (1.69± 0.36 mm).

3.4 Bone damage assessment

3.4.1 Outer surface. Generally, damage in the anterior region was less evident by visual inspection of the bone specimens, and it seemed that the bone surface was polished (Figure. 6.5A), however, bone was clearly deformed and crushed in the posterior region (Figure. 6.5B). At the medial and lateral condyles per specimen a similar pattern of damage was observed, which can be seen clearly in the frequency graphs (Figure. 6.6A). However, the inter-specimen comparison showed noticeable differences between specimens, with higher damage associated with lower bone density (R²= 0.94; P= 0.006) (Figure. 6.4B). In one case (specimen B-Right), bone was even pushed outwards (extruded) in the distal region of the condyles (Figure. 6.6A (B-R)). This specimen also had the largest nodal damage in the proximal region of condyles (1.75 and 1.70 mm laterally and medially, respectively).
Figure 6.3 A) Frequency graph and distribution map of cutting error and B) actual interference fit are demonstrated for the five specimens. For the cutting error, values lower than zero indicates that bone is over-resected and higher than zero indicates under-resected bone. For the actual interference fit, negative values show a gap between implant and bone surface. Positive values means that implant is penetrated into the bone for the given value.
3.4.2 Internal. Several visible phenomena occurred in the post-implantation micro-CT. In one case, a deep fracture was seen in the proximal region of the lateral condyle (specimen A). In addition, depending on the location in the anterior and posterior regions, bone densifications in the form of a thin layer of a white line at the interface (specimen B-Left in figure. 6.7) and/or clear bone abrasion were seen, probably accompanied by permanent deformation of the bone (specimen B-Right in figure. 6.7).

3.5 Effective interference fit

A large elastic deformation in the specimens of good bone quality was found with $1.12 \pm 0.18$ mm and $1.38 \pm 0.32$ mm in specimen C-L and C-R, respectively (Figure.6.6B). Other specimens had a smaller effective interference fit and the correlation with BMD was statically significant ($R^2 = 0.94; P= 0.006$) (Figure. 6.4C).

3.6 Implant deformation

Three strain gauges, which belonged to three different specimens, were lost during implantation. No clear inter-specimen trend was found in terms of strain gauge location, but generally the largest strain ($933.25 \pm 771.05$ $\mu$ε (mean± SD)) was recorded in the PL strain gauge (Table. 6.2). In addition, the correlation between BMD and strain showed a significant strong correlation ($P= 0.007$) (Figure. 6.4D).

Figure 6.4 A) The relationship between cutting error, B) damage, C) effective interference fit, and D) strain with BMD is demonstrated.

Figure 6.5 A) A typical surface mesh of post-implantation micro-CT (specimen B-L), which shows polished surface of anterior flange (A) versus clear crushed surface of posterior condyles (B).
Table 6.2 Recorded strain of four strain gauges and their average per location and per specimen are given.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Strain per location (µε)</th>
<th>Average strain (µε)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>AL</td>
<td>PL</td>
</tr>
<tr>
<td>A-Left</td>
<td>481</td>
<td>703</td>
</tr>
<tr>
<td>B-Left</td>
<td>-</td>
<td>507</td>
</tr>
<tr>
<td>B-Right</td>
<td>269</td>
<td>445</td>
</tr>
<tr>
<td>C-Left</td>
<td>855</td>
<td>-</td>
</tr>
<tr>
<td>C-Right</td>
<td>1172</td>
<td>2078</td>
</tr>
</tbody>
</table>
| Average ± SD | 694.25± |933.25± |772.00± |507.25± 
|           | 400.17 | 771.05 | 634.52 | 416.97 |

4. Discussion

The current study focused on the evaluation of interference fit and bone damage occurring during implantation of femoral knee implant using image analysis techniques; because both are affected by cutting errors and bone density, we included these parameters in our investigation as well.

To our knowledge, this is the first study investigating bone damage that occurs during implantation of a femoral knee component using micro-CT, which enabled us to demonstrate different phenomena that can happen to trabeculae in a press-fit situation underneath the femoral component. It was out of our scope to perform quantitative analysis of micro-CT scans, but our observations indicated that bone can be damaged beyond its interface with the implant. In addition, bone particles can penetrate into the inter-trabecular spaces, which can be considered as a negative observation for initial fixation, but on the longer term these particles may stimulate bone ingrowth[14]. According to our expectation, different phenomena occurred to the bone specimens during implantation: bone was deformed both permanently (inelastic) and reversibly (elastic) (Figure. 6.7). In addition, abrasion occurred creating bone particles in three locations: 1- at the surface coating of implant (Figure. 6.8A and B), 2- within the porosity of bone, seen as a white line at the interface of the post-implantation micro-CT scans (Figure. 6.7, specimen B-Left), 3- and bone particles being scraped out, which was evident only in the low bone density specimens (Figure. 6.8C). However, the degree of each of these phenomena was largely dependent on the bone quality. Bishop et al. [4] measured the press-fit bone damage using blocks of bone and titanium platens with different surface coatings, and one of these (beaded) having very similar properties to the surface coating of the implant in our study. Similar to our results, they found a combination of plastic deformation and abrasion, which was quantified mainly as abrasive when using a rough surface coating. More recently Damm et al. [15] also demonstrated the local bone densification due to implantation.
Figure 6.6 A) Frequency graph and distribution map of bone damage and B) elastic interference fit are demonstrated for the five specimens. Negative values of figure A demonstrate how far the surface of post-implantation bone is relative to the pre-implantation situation. Positive values indicate that post-implantation bone is projected with respect to the pre-implantation, which occurred only in one case (B-R). In the figure B, positive values show the degree of elastic deformation, which is the distance between surface mesh of post-implantation bone and implant surface.
Figure 6.7 Two slides from (A) pre- and (B) post-implantation micro-CT scan of each specimen are shown to illustrate different phenomena that occurred during implantation. Slides are selected from the same distance from the most distal slide.
The amount of interference fit designed into the system by the manufacturer was 0.75 mm in the anterior and posterior regions, resulting in 1.5 mm of AP interference fit in total. Similarly, we found an average actual interference fit of 1.48± 0.27 for all specimens. However, the actual interference fit was not distributed homogenously, between posterior and anterior regions and also per ROI, due to combination of cutting errors and imperfect implant seating. The work of Otani et al.[16] is one of the few investigations into the range of the cutting error, and the sources of this error. They reported a range of cutting error between 100 to 1000 µm dependent on the type of saw blade and the use of a fixed cutting guide. In addition, they showed a higher deviation from nominal cuts in the more proximal region of both anterior and posterior cut planes. Our measured range was similar to that study and also in the current study the same increasing trend towards the proximal region was observed in most of the cases. Hence, the cutting error might be due to toggling/bending of the saw blade and displacement of the cutting guide, as shown by Otani et al. [16]. In addition, thirteen out of the fifteen ROIs currently analyzed were under-resected (less bone removed than planned), which means in most regions the primary fixation was not jeopardized by introducing a gap between bone and implant. No trend was seen between cut accuracy and BMD, which is consistent with our previous study [10]. This suggests that other factors such as surgical technique and surgeon experience may be more important.

![Image](image-url)  
**Figure 6.8** A) Internal surface of posterior condyles and B) anterior flange of the implant after splitting the implant, which show remnant bone particles (specimen C is depicted). C) Bone particles were scraped out after implantation which can be seen as a layer around the interface of the posterior condyles (inside the boxes).

The actual and effective interference fits that were determined in the current study may actually have been affected by deformation of the implant (which was assumed to be completely rigid during calculation procedure), as indicated by the strain measurements with the femoral components. To quantify the amount of implant opening, and thus its effect on the interference fit, we performed a simple finite element simulation in which the posterior condyles were fixed, while applying an external load to the anterior flange. Our analysis showed that an average opening of 100 to 500 µm in the AP direction would result to the same range of strain as found in the experiment. Consequently, the average elastic deformation would be 820±480 µm (mean± SD) minus the computed range of implant opening, which
would result in a range of effective interference fit between 30 µm to 880 µm. Damm et al. [15] also found a wide range of effective interference fit between 30 to 80% of nominal interference fit mainly depends on the nominal interference fit. Unlike our findings, the effect of bone quality on the amount of bone permanent deformation was not evident in their study, but resulted in a lower stability. This can highlight the necessity to evaluate bone samples with a realistic dimensions to capture the entire bone behaviour.

Burger et al. [17] performed similar strain gauge measurements which showed a range of strain close to our worst bone quality. In our previous study[10], we also found a wide range of strain dependent on the bone density, which had a maximum of 700 µε for the best bone quality. Therefore, current results are similar to the previous findings, but probably the reason for a higher range of strain in the specimens with the same BMD as our previous study is the larger size of implant.

4.1 Limitations

We have tested only one type of implant, performed the implantation by one surgeon, and had a limited sample size. However, we believe that the current study provides a detailed insight into the bone damage and interference fit of cementless femoral knee implants. Due to providing fixation of implant mainly by clamping power of AP-interference fit, measurements were only in the anterior and posterior regions. Though, it is possible that a more complete analysis of the fully resected bone may provide a deeper understanding of implantation procedure. Our observation, however, showed that the complete contact in the distal surface and chamfers was not achieved in all cases. Therefore, we believe that bone damage and interference fit would be minimal in these regions and would not affect our results.

4.2 Conclusion

This study indicated that several factors are involved in the press-fit implantation of uncemented femoral knee implants. We found a complex interaction between cutting error, implant positioning, and bone density. It was demonstrated that on average the nominal interference fit was achieved, but it was not a homogenous distribution over all surfaces. In addition, bone damage and effective interference fit are dependent on the bone density, which emphasizes the significance of good patient selection. Finally, the interference fit determined in the current study can be used in computational analyses to evaluate primary fixation of femoral components, and can be used to fine-tune the design process when developing cementless femoral TKR components.

Acknowledgments: This work was supported by a research grant from DePuy Synthes Joint Reconstruction (Leeds, UK). The authors would like to thank orthopaedic surgeon, Dr. S. van de Groes, for performing implantation and Mr. F. Baan for his help with the optical scanning.
References

CHAPTER SEVEN
Experimental and computational analysis of micromotions of an uncemented femoral knee implant

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Abstract

It is essential to evaluate the micromotions at the bone-implant interface of an uncemented femoral total knee replacement (TKR) using a reliable tool. In the current study, experimental measurements of micromotions were compared with predicted micromotions by Finite Element Analysis (FEA) using two bone material models: linear elastic and post-yield material behavior, while an actual range of interference fit was simulated. The primary aim was to investigate whether a plasticity model is essential in order to calculate realistic micromotions. Additionally, experimental bone damage at the interface was compared with the FEA simulated range. TKR surgical cuts were applied to five cadaveric femora and micro- and clinical CT-scans of these un-implanted specimens were made to extract geometrical and material properties, respectively. Micromotions at the interface were measured using digital image correlation technique. Cadaver-specific FEA models were created resembling the experimental situation. The average experimental micromotion was 53.1 ± 42.3 µm (mean ± standard deviation (SD)), which was significantly higher than the micromotions predicted by both models, using either the plastic or elastic material model (26.5 ± 23.9 µm and 10.1 ± 10.1 µm, respectively; p-value < 0.001 for both material models). The difference between the two material models was also significant (p-value < 0.001). The predicted damage had a magnitude and distribution which was comparable to the experimental bone damage. We conclude that, although the plastic model could not fully predict the micro motions, it is more suitable for pre-clinical assessment of a press-fit TKR implant than using an elastic bone model.
1. Introduction

Long-term stability of press-fit joint replacement implants can only be achieved when the micromotions at the bone-implant interface remain small postoperatively [1, 2]. Various experimental and numerical testing techniques have therefore been developed to evaluate micromotions at the bone-implant interface [3-5]. Computational models based on Finite Element Analysis (FEA) allow the investigation of more complex loading conditions [6, 7], although, it is still difficult to capture the complex interactions of material and contact properties of the bone-implant system. Therefore, it is challenging to predict realistic ranges of micromotions with FEA and further improvement of these models is required.

Simulations of bone-implant micromotions usually incorporate a linear elastic material bone model [6-9], though, bone is recognized as a plastic material [10]. Hence, the press-fit fixation which is obtained intraoperatively by introducing a difference in sizing between the implant and host bone, cannot be simulated correctly, as in such analyses the contact forces always increase with increasing interference fit [4, 11, 12]. Consequently, bone stresses exceeding the yield limit are calculated, and the stability of the component is overestimated. Subsequently, in such studies the reported values of interference fit typically are far smaller than actual clinical nominal interference fit as determined by the surgical cuts and implant dimensions [4, 11].

In addition, during an actual implantation of press-fit implants bone stresses are generated that exceed the yield strength of bone [13]. As a result, substantial amounts of permanent deformation or bone damage are observed at the interface after implantation [14]. Hence, plastic deformation and damage occurring at the interface during implant insertion may have a significant effect on the interference fit and the resulting interface micromotions.

The aim of this study was therefore to investigate whether a plastic bone material model is essential to calculate a realistic range of micromotions at the bone-implant interface of an uncemented femoral knee replacement, while applying a realistic nominal interference fit. To achieve this goal, in vitro measurements of micromotions were compared with cadaver-specific FEA models, assigned with either elastic or plastic material models.

2. Materials and Methods

2.1 Experimental measurements

2.1.1 Study design. The experimental component of current study has been partially described before [14]. In summary, five fresh-frozen cadaveric femurs (two pairs; 85± 4 years old (average± std)) were received from the Anatomy Department of the Radboud university medical center, and implanted with uncemented Sigma® cruciate retaining (CR) femoral knee implants (DePuy Synthes Joint Reconstruction, Leeds, UK). An experienced
orthopedic surgeon determined implant sizes (one size 3 and four size 5) and performed the bone cuts according to the manufacturer’s instructions. Next, the bone specimens were both CT- (Siemens Somatom Sensation 64, Siemens AG, Germany) and micro-CT scanned (XtremeCT II, SCANCO Medical AG, Bruettisellen, Switzerland) to extract material properties and detailed bone geometries, respectively (Table. 7.1). The bone mineral density (BMD) of each bone specimen was measured in a previously described protocolized manner [15]. In the results section, the specimens were ranked from lowest to highest BMD (i.e. specimen 1 had the lowest and specimen 5 the highest BMD).

After scanning, the femoral components were placed by the surgeon, after which the position of the implant relative to the bone was registered by acquiring optical scans of the specimens (TRIOS Color-P13, 3Shape, Copenhagen, Denmark). The accuracy of the system as provided by the manufacturer was 100 µm.

<table>
<thead>
<tr>
<th>Technique of scanning</th>
<th>Voltage</th>
<th>Tube current</th>
<th>Integration time</th>
<th>Resolution (µm)</th>
<th>Slice thickness (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>CT</td>
<td>120 kV</td>
<td>500 mA</td>
<td>-</td>
<td>351</td>
<td>600</td>
</tr>
<tr>
<td>Micro-CT</td>
<td>68kVp</td>
<td>1470 µA</td>
<td>200 ms</td>
<td>60.7</td>
<td>60.7</td>
</tr>
</tbody>
</table>

2.1.2 Micromotion measurement. Digital image correlation (DIC) technique was chosen to measure micromotion as close as possible to the bone-implant interface. The DIC set-up comprised a single SPOT™ Insight 2.0 color digital camera (SPOT TM Imaging Solutions, Diagnostic Instruments, MI, USA) with a 1” format CCD sensor (Kodak KAI-2020-CM) and a magnifier to increase the resolution to 7.4 µm/pixel (Figure. 7.1). A black-and-white speckle pattern was applied to each specimen to perform image correlation measurements between the unloaded and loaded images. In total, seven regions of interest (ROIs) were selected: at the anterior flange from the coronal plane view and at the anterior, distal, and posterior regions from sagittal plane view in both medial and lateral sides of the studied objects (Figure. 7.2). In each ROI, the measurement was repeated three times; the average of the three measurements was used later for the comparisons. Micromotion measurements were performed under four different static loading conditions, which were extracted from physiological loading profiles of gait and deep knee bend (DKB) activities [16, 17] (Table. 7.2). The loading profiles at an angle of 14° and 90° represented the peak tibiofemoral (TF) force of gait and DKB, respectively. Only 60% of the maximum load was applied under 90° of flexion, to prevent fracture of bone at the distal bone fixation. For each position, before performing measurements, the specimens were pre-conditioned for 15 minutes at 1 Hz with the position-dependent loading profile to allow initial settlement of implant. All loads were applied using a servo-hydraulic testing machine by a force-control by applying a compressive...
force-controlled load at a rate of 100 N/sec (MMED, MATCO, La Canada Flintridge, CA). Opticist 0.995 (Moiré Analysis Software) was used for post-analysis. Sampling squares of $255 \times 255$ pixels (about 3 mm$^2$) were defined at the implant and adjacent bone, at the closest possible distance to the interface. Micromotions at the sampling locations were calculated by subtracting the bone displacements from the implant displacements. The horizontal and vertical components were then transformed to the normal and shear directions, based on the orientation of the bone-implant interface.

**Figure. 7.1** High-resolution camera with telecentric lens and light sources to capture micromotion at the bone-implant interface at the anterior flange under a loading condition with a flexion angle of $30^\circ$. Load was applied using a matched size tibial implant.

<table>
<thead>
<tr>
<th>Activity</th>
<th>Flexion angle ($^\circ$)</th>
<th>Load (N)</th>
<th>Applied percentage of load (%)</th>
<th>Medial-lateral ratio (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gait</td>
<td>14</td>
<td>2,387</td>
<td>100</td>
<td>50 – 50</td>
</tr>
<tr>
<td></td>
<td>30</td>
<td>1,782</td>
<td>100</td>
<td>40 – 60</td>
</tr>
<tr>
<td>DKB</td>
<td>60</td>
<td>1,974</td>
<td>100</td>
<td>64 – 36</td>
</tr>
<tr>
<td></td>
<td>90</td>
<td>2,250</td>
<td>60</td>
<td>63 – 37</td>
</tr>
</tbody>
</table>

**2.1.3. Damage assessment.** After the DIC measurements the implants were cut through the condyles using an electric diamond-blade cutting machine. A second series of micro-CT scan was obtained to evaluate the damage at the bone interface. Accordingly, the surface geometry of post-implantation micro-CT scan was registered onto the pre-implantation micro-CT scan using the iterative closest point (ICP) algorithm. The anteroposterior (AP) distance between the two scans at three ROIs, anterior region and posterior condyles, was taken as a measure
for the bone damage (see also section 2.2.3).

2.2 Finite element model

For each distal femur, surface geometries based on the clinical CT, pre-implantation micro-CT, and post-implantation micro-CT scans were retrieved. Semi-automatic segmented volumes were obtained using medical imaging software (Mimics 14® and 3-Matic®, Materialise, Leuven, Belgium).

The optical scans were processed using Rhinoceros 5.0 (McNeel, WA, USA) to obtain detached surface meshes of the implant and bone. First, the optical scan of the implant was registered onto the 3-D model of the implant, which was created from the CAD file supplied by the manufacturer (DePuy Synthes Joint Reconstruction, Warsaw, IN). The optical scan of the bone was then transformed using the same transformation matrix. Next, the surface mesh of the pre-implantation micro-CT scan was registered onto the transformed optical scan of the femur. Consequently, a 3D-model of the bone and implant was obtained, with the same orientation as the implanted experimental specimen (Figure. 7.2).

The surface models were then imported into the FEA software Marc/Mentat 2007 (MSC. Software Corporation, Santa Ana, CA) to create FEA meshes. An average edge length of 1 mm was selected based on a previous mesh convergence study [16]. Solid models of the bone and implant were created using linear four-noded tetrahedral elements (Table. 7.3).

![Figure 7.2](image)

**Figure. 7.2** A) Medial, posterior, lateral, and anterior views of an experimental specimen. B) The correspondent finite element model. The schematic regions of interest (ROIs) are demonstrated by a rectangle.

2.2.1 Material properties. The implant was simulated with isotropic linear elastic material properties of cobalt chrome (Young’s modulus: 210 GPa, Poisson’s ratio: 0.3). Bone material
properties were assigned by mapping calcium equivalent values to the elements, by first converting Hounsfield Units (HUs) to $\rho_{\text{QCT}}$ values based on a calibration phantom that was scanned along with the cadavers (solid, 0, 50, 100, 200 mg/ml calcium hydroxyapatite, Image Analysis) as developed previously using an in house software [18]. Element-specific Ca values were related to ash densities ($\rho_{\text{ash}}$) and subsequently to the Young’s modulus using relationships from Keyak et al., which resulted in the given range of Young’s modulus in the table 7.3 [19, 20]. An isotropic linear elastic material model and a non-linear isotropic post-yield material behavior were adopted according to Keyak et al. [19, 20] using the Von Mises yield criterion, in which $\sigma_y$ (yield stress) equals $102 \times \rho_{\text{ash}}^{1.8}$ for both trabecular and cortical bones. A Poisson’s ratio of 0.3 was assigned in both material models.

### Table 7.3
Number of nodes and elements are given for the implant and bone models. In addition, the range of Young’s modulus (E) per bone specimen is given.

<table>
<thead>
<tr>
<th>Solid model</th>
<th>Implant size 3</th>
<th>Implant size 5</th>
<th>Bone 1</th>
<th>Bone 2</th>
<th>Bone 3</th>
<th>Bone 4</th>
<th>Bone 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Number of elements</td>
<td>254,911</td>
<td>434,889</td>
<td>302,337</td>
<td>282,715</td>
<td>227,346</td>
<td>266,465</td>
<td>220,278</td>
</tr>
<tr>
<td>Number of nodes</td>
<td>49,227</td>
<td>81,978</td>
<td>62,282</td>
<td>58,638</td>
<td>46,727</td>
<td>54,792</td>
<td>47,964</td>
</tr>
<tr>
<td>E (GPa)</td>
<td>210</td>
<td>210</td>
<td>0-29.76</td>
<td>0-20.90</td>
<td>0-27.75</td>
<td>0-25.14</td>
<td>0-25.64</td>
</tr>
</tbody>
</table>

### 2.2.2 Boundary conditions, contact definition, and micromotion.
The FEA simulation was divided into three distinct phases. First, the femoral component was virtually implanted, while in the second phase the reconstruction was subjected to the experimental loads. In the final phase, the contact between bone and implant was eliminated to quantify the permanent bone deformation. We did not simulate the actual implantation by forcing the implant into the bone. Instead, the surface nodes of the bone that initially penetrated the implant were incrementally pushed towards the implant surface using the interference fit option of the FE software. During this simulation, the bone was gradually compressed until it touched the implant surface. The femur was fixed proximally in all directions, while during the simulated implantation phase the implant was only fixed in the proximal-distal direction to adequately constrain the model for convergence purposes. A frictional touching contact algorithm based on a Coulomb bilinear (displacement) friction model with a coefficient of friction of 0.95 was defined between bone and implant. Next, during the simulation of the experimental tests, the experimental loading conditions were applied to the implant. Hence, for each of the four loading conditions, one increment was unloaded, while the next one was loaded (Table 7.2). The first loading condition was repeated twice to allow initial settling of implant [16]. Similar to the experimental measurement, micromotions at the seven ROIs were calculated.
by subtracting the nodal displacement between bone and implant. All simulations were performed using MSC.MARC (MSC.MARC2007 r1, MSC Software Corporation, Santa Ana, CA, USA).

2.2.3 Damage assessment. After calculating micromotions under all loading conditions, the contact between the bone and implant was removed to quantify the amount of permanent deformation of the bone. For this goal, a series of nodes of bone surface underneath the anterior flange and posterior condyles, were selected (the same ROIs as experimentally measured; see section 2.1.2). The total permanent displacement of these nodes in the AP direction, after virtually removing the implant, was defined as damage to allow for comparison with the experimental results.

2.3 Data analysis

The micromotions of all ROIs per specimen for the four loading conditions were compared between experimental and two FEA predicted micromotion using a t-test. Linear regression analysis was used twice to correlate experimental and FEA micromotions. Once to compare the overall trend between experimental and FEA results using the average micromotion of all ROIs per specimen and per loading condition. Furthermore, a specimen-specific regression model incorporating all ROIs and the four loading conditions was obtained to assess the predictability of FEA per specimen. In addition, linear regression analysis was used to assess relationship between experimental and FEA bone damage.

3. Results

3.1 Effect of bone material model on the micromotion

The average experimental micromotion was 53.1± 42.3 µm (mean± standard deviation (SD)), which was significantly higher than the micromotions predicted by using either the plastic or elastic material model (26.5± 23.9 µm and 10.1± 10.1 µm, respectively; p-value< 0.001 for both material models) (Figure. 7.3A). The difference between the two material models was also significant (p-value< 0.001) (Figure. 7.3A). In addition, the plastic bone model captured the effect of bone quality on the micromotion with higher micromotions for the bone samples with lower BMD, which was not found for the elastic bone model (Figure. 7.3A).

The correlation of the average micromotions per specimen (from the seven ROIs) for the four loading conditions showed that there was a moderate correlation (p-value < 0.001) between experimental and predicted micromotion by the plastic material model and no correlation was found for the elastic bone model (Figure. 7.3B and C).
A) Experimental micromotions were significantly higher than predicted micromotions using both material models. Also micromotions calculated by the plastic material model were significantly higher than the elastic ones. It should be highlighted that specimens are ordered from low to high bone quality. B) The correlation between experimental and predicted micromotion by elastic bone model including four loading conditions. C) The same correlation with the predicted micromotion by the plastic bone material model.

### 3.2 Specimen-specific regression model

A weak to moderate correlation was found between experimental and FEA results per specimen, which was significant for four out of five specimens in both material models (Table. 7.4). Generally, the slopes and correlation coefficients were smaller for the elastic material model than for the plastic bone model.

#### Table. 7.4 The specimen-specific regression models.

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.55</td>
<td>0.16</td>
<td>5.98</td>
<td>1.57</td>
<td>0.61</td>
<td>0.63</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>2</td>
<td>0.17</td>
<td>0.056</td>
<td>14.99</td>
<td>4.1</td>
<td>0.15</td>
<td>0.25</td>
<td>0.041</td>
<td>0.007</td>
</tr>
<tr>
<td>3</td>
<td>0.21</td>
<td>0.038</td>
<td>15.6</td>
<td>9.24</td>
<td>0.11</td>
<td>0.018</td>
<td>0.087</td>
<td>0.5</td>
</tr>
<tr>
<td>4</td>
<td>0.38</td>
<td>0.13</td>
<td>0.03</td>
<td>4.4</td>
<td>0.6</td>
<td>0.2</td>
<td>&lt;0.001</td>
<td>0.027</td>
</tr>
<tr>
<td>5</td>
<td>0.2</td>
<td>0.06</td>
<td>9.66</td>
<td>3.22</td>
<td>0.26</td>
<td>0.19</td>
<td>0.006</td>
<td>0.025</td>
</tr>
</tbody>
</table>

### 3.3. Damage assessment

Figure 7.4 shows the distribution of the damage measured in the experimental specimens and
simulated in the FEA simulations incorporating the plasticity model. The damage predicted by the plastic FEA simulations had a magnitude and distribution which was comparable to the experimental bone damage. However, particularly in the anterior region, smaller areas of bone damage were predicted by the FEA simulations. The slope of correlation graphs suggested that generally the FEA under-predicted the bone damage (Figure 7.4B). Interestingly, damaged seems to be simulated more accurately in specimens with lower BMD in terms of slope and correlation coefficient.

Figure 7.4 A) Experimental bone damages are demonstrated, it should be highlighted that in the specimen no. 2, bone was deformed outwards in the proximal region of posterior condyle (red region); B) Correlation between experimental and FEA damage, it should be noted that experimental results with positive damage values are treated as outlier and negative damage sign is changed to positive for the demonstration purposes; C) FEA bone damage. Blue means that bone is removed and/or permanently deformed.
4. Discussion

The main goal of the current study was to compare the computational prediction of micromotions using both a linear elastic and a plastic bone material model with experimental data. In addition, the experimental bone damage patterns were compared with the simulated permanent deformation of bone. We found that a plastic model significantly improved the prediction of micromotions at the bone-implant interface of an uncemented femoral knee implant in comparison with a linear elastic model. In the current study, the resected bone before the implantation process was used as the basis model, which means the bone dimensions were remarkably larger than the inside dimensions of the implant in the AP direction. In case of the elastic bone model this led to high contact forces at a limited number of nodes, while the plastic bone model resulted in a more homogenous distribution of contact forces at the interface. Consequently, adding plasticity resulted in simulated bone damage that resembled the experimental cases, and resulted in realistic values for the actual interference fit present at the bone-implant interface, which to our knowledge has not been reported before.

A limited number of studies have incorporated a non-linear plastic model in the FE analysis of press-fit implants [21-23]. Rothstock et al. [22] evaluated also the effect of the material model on the primary stability of press-fit implant. Unlike our results, they found a similar range of micromotions for the elastic and plastic material models, and explained that low interference fit, with a maximum value which was about one fifth of the applied interference fit in our study, was probably the reason for this. Hence, interference fit is an important interfacial parameter that cannot be ignored.

Due to the lack of experimental data on the real interference fit, most computational studies assume either no interference fit (bone and implant are modelled “line-to-line” at the interface) [6, 8], or utilize the interference fit option of FE software, in which implant nodes virtually penetrates the nodes of bone [4, 16]. In practice, press-fit interference is achieved by introducing a dimensional difference between bone and implant, impacting the implant onto the oversized bone. However, a computational simulation of this situation leads to excessive stresses when a nominal interference fit is used (1-2 mm), and non-converging simulations. Hence, we eventually could assign a realistic range of interference fit by the virtual implantation method as explained before.

Our results showed that the plasticity model also captured the effect of bone quality on the micromotion, with a higher range of micromotions found for specimens with lower bone quality, which was not seen in the elastic bone model. This can be explained by a larger range of bone damage in the bone with inferior quality, which led to lower press-fit forces at the bone-implant interface and subsequently a lower resistance to shear loads. In case of the elastic bone model, applying the nominal values of interference fit resulted in high bone stresses that apparently diminished the effect of inferior bone quality on the shear resistance
of the bone samples.

Despite the improved predictive ability of the plastic model, the micromotions were still underestimated. One possible explanation for this is the fact that no viscoelastic behavior was incorporated in our bone models. In a previous experimental study using cadaveric femora we found that the fixation strength of press-fit implants can reduce over time [24]. Norman et al. [25] also demonstrated that push-out load of press-fit stems were significantly reduced after a waiting period. A second factor that might affect the predicted micromotions is bone abrasion. We observed bone debris at the edge of the posterior condyles after the implantation. Also remnant particles were seen at the implant surface for all specimens. Damage results also indicated that the plasticity model underestimated the amount of experimental damage, as the bone abrasion is also partially attributed to damage, which was not simulated. A more detailed analysis of the pre- and post-operative micro-CT scans could perhaps provide more information on the amount of plastic deformation and bone abrasion, as previously demonstrated by Bishop et al. [26]. Furthermore, a coefficient of friction can also affect the shear capacity of bone-implant system. We applied only one coefficient which was provided by the manufacturer (DePuy Synthes, Warsaw, USA) for this type of surface coating using sawbones, but bone is a fatty tissue, which may decrease the frictional capacity using cadaveric bones.

Micromotions at the bone-implant interface of uncemented and cemented femoral knee implants were investigated in a limited number of studies [3, 27, 28]. Cristofolini et al.[27] reported micromotions in a wide range from 10 to 200 μm for a cemented femoral knee implant and under loading conditions resembling gait. Conlisk et al.[3] also measured a wide range for different cementless and cemented femoral knee implants under three different loading conditions (full extension, 10°, and 20°). In general, our experimental micromotions for the loading conditions resembling DKB (at an angle of 60° and 90°) was closer to previously reported results and micromotions under loading conditions at an angle of 14° and 30° had a smaller range in our study, which could be due to the type of implant, fixation, or measurement technique.

It was not the focus of the current paper to investigate in detail the effect of loading conditions on the micromotions. However, interestingly, we found significantly higher micromotions for loading conditions at a larger flexion angle, even though the applied load was lower (1,974N and 1,350N for 60° and 90° versus 2,387N and 1,782N for 14° and 30°). The concern about the higher incidence of femoral loosening under high-flex activities [29, 30], could be associated with the higher range of micromotions that we found at higher flexion angles. It must be mentioned, however, that patellofemoral forces were not included in the current loading configuration, which probably would have a stabilizing effect, particularly at higher flexion angles[16].

The specimen-specific linear regression models showed that the predictive ability of FEA
model was dependent on the specimen. Hence, further improvements of the FEA model are required to achieve stronger correlations in some cases. This may be due to confounding factors such as the relationship between viscoelasticity and bone quality, or abrasion as explained before. Therefore, it is necessary to further improve the current material model to determine the source of the discrepancies.

Although current experiments and simulations were performed with only a single implant design, with its specific interference fit and surface characteristics, the basic principles of the approach presented here could be extended to other implant designs, thereby possibly increasing the robustness of the computational analyses. Using a similar approach, it is also possible to test other damage material models, such as the crushable foam model. Kelly et al. [23] found that such a model may be preferred over a Von Mises formulation when simulating the plastic behavior of synthetic bones. In such a set-up, the simulated trabecular bone is confined by a stiff cortical bone, causing confined compression condition. It remains to be seen whether the crushable foam model would further improve the material model.

4.2. Conclusion

Substantial non-linear material behavior (plasticity) was observed experimentally during implantation of uncemented femoral TKR, which emphasizes the necessity to include non-linear material behavior, while studying press-fit implant using computational technique. This leads to an improved prediction of micromotions at the bone-implant interface. However, inclusion of plasticity does not lead to perfect predictions, and therefore, the material models need to be further improved in order to entirely predict initial stability of uncemented femoral TKR components.

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CHAPTER 7

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CHAPTER EIGHT
General discussion and future perspective
CHAPTER 8

Abstract

This thesis investigated the biomechanical aspects of press-fit fixation of uncemented femoral knee implant employing in vitro testing and finite element analysis (FEA). Our main findings characterized the factors that play an important role in achieving adequate primary stability. Although this research does shed more light on current understanding of press-fit aspects, micromotion, and how the primary stability can be improved, it is still important to make a link between the presented fundamental research and clinical practice. Furthermore, various strengths and weaknesses about the pre-clinical testing techniques utilized in this thesis (in vitro testing and FEA) will be discussed. In the last section, future perspectives relative to the findings in this thesis are outlined.
1. Main findings

**Surface morphology.** The effect of three different surface morphologies with a low, medium and high roughness on the fixation strength (pull-out force) of femoral pegs was investigated in chapter 2. Unlike our expectation, we found a significantly higher fixation strength of smooth pegs compared to the pegs with the highest roughness in an immediate post-insertion phase. This was contradictory to our findings in chapter 3, which demonstrated a 40% higher push-out force for an uncemented femoral knee implant with a rougher surface morphology. To explain this contradiction, we should consider the post-implantation time. We found a noticeable interaction between the surface morphology and waiting period due to bone stress relaxation in chapter 2, which demonstrated that after a waiting period the press-fit forces would decline. Consequently, the higher frictional capacity of the rougher surface morphology to provide the fixation strength became more prominent. Therefore, in chapter 3 (with relatively long waiting periods), the rough surface morphology did provide a higher shear resistance. Generally, we concluded that a rough surface morphology can improve the primary stability of implant, especially after a longer time span.

**Press-fit interference.** This factor was investigated differently in chapters 2 and 6. In chapter 2, several diametrical interference fits were evaluated by insertion of femoral pegs into bone holes with a smaller diameter. Highly controlled interference fits were achieved by controlling the size of the hole and the peg diameters. This study showed a significant positive effect of interference fit on the pull-out strength. Interestingly, on the longer term, the effect of interference fit was more visible for the pegs with a higher roughness. The interaction between waiting period and interference fit again emphasizes the role of stress relaxation on the fixation strength. In an immediate post-insertion phase, a large interference fit created high pre-stresses, which declined in time. Therefore, the initial effect of interference fit is less evident after a waiting period. Furthermore, it was found that beyond a certain interference fit, no further improvement of fixation strength is achieved. A range of 650-800 μm was suggested to be a sensible range of radial interference fit for the tested design.

In chapter 6, the press-fit interference of an uncemented femoral knee implant was investigated. Here, we introduced a distinction between the nominal, actual and effective interference fit. The average actual interference fit, which was basically the combination of the nominal interference fit, cutting error and implant seating, was found to be 1.48± 0.27 mm, close to the nominal interference fit of 1.5 mm. However, during implantation the bone would abrade and deform plastically, altering the actual interference fit. The effective interference fit would therefore represent the primary stability. We found a wide range of effective interference fit depending on the bone quality, ranging from 130-1384 μm. Our analyses also indicated that the implant cannot be assumed to behave as a rigid body, as anteroposterior
(AP) opening of the femoral component of as much as 500 μm was measured, which further reduces the effective interference fit. In chapter 7, we reported micromotions measured in the implanted bone specimens from chapter 6. Interestingly, we found an acceptable range of micromotions in all samples, despite the wide range of effective interference fit. Hence, it can be concluded that small amounts of effective interference fit can adequately provide the primary stability, although it may have a more important role under more extreme loading, such as during a pull-out test, as was found in chapter 2.

**Bone.** In all in vitro studies, bone mineral density (BMD) was measured following a standardized protocol using calibrated CT-scans. The effect of BMD on the fixation strength was evaluated in chapters 2 and 3 and its effect on the primary stability was the focus of chapters 5 and 7. The effect of bone stress relaxation on the created pre-stresses during implantation and also its effect on the fixation strength was also investigated in chapters 2 and 3. In chapter 2, the effect of bone stress relaxation on the fixation strength was assessed by comparing the fixation strength in an immediate post-insertion phase with the fixation strength after a waiting period. Overall, BMD significantly affected both the fixation strength (pull-out force) and primary stability (micromotion). We found different significant relationships between BMD and the fixation strength in chapters 2 and 3; which was the most noticeable between the fixation strength of the uncemented femoral knee implant and BMD in chapter 3, with a strong correlation coefficient (R²) of almost 0.9. A moderate to strong linear correlations, depending mainly on the surface coating was found in chapter 2. Generally, lower roughness and higher interference fit increases the correlation coefficient. In chapter 5, we found that the loading configuration can be a confounding factor and BMD was only a significant factor when its effect on the micromotion was evaluated under the simplified gait loading and no significant effect was found under the DKB loading configuration. We explained that bone anisotropy may lead to different relationship between micromotion and BMD depends on the direction of applied load.

As explained previously, the effect of bone stress relaxation on the fixation strength of femoral pegs (chapter 2) was dependent on the type of surface morphology, with a much smaller effect on the fixation strength of pegs with a rougher surface. Strain measurements on the external surface of the implant were used as an indirect measurement to evaluate bone stress relaxation in chapter 3. We found a decline in post-implantation strain over two hours, which was considered as a sign of stress relaxation. Bone stress relaxation may negatively affect the primary stability in a longer time period, but we did not further investigate this phenomenon in this thesis.

**Activity and loading.** In chapters 4, 5, and 7, we assessed the primary stability of three
uncemented femoral knee implants, Sigma®, LCS® and Attune® (DePuy Synthes, Warsaw, IN, USA). The micromotion was predicted using FEA in Chapter 4, while it was measured in vitro in chapter 5. Chapter 7 was a validation study with a combination of FEA and in vitro measurements. The primary stability of implants was evaluated under two loading conditions resembling the peak tibiofemoral (TF) force during gait and a deep knee bend (DKB), which were adopted from full loading profiles. These loading regimes were obtained from the validated FE model of the Kansas Knee simulator, which previously reproduced in vivo forces presented by the Orthoload database. In addition, in chapter 7 the Sigma design was evaluated under two extra loading conditions with different loading angles (30° and 60°) for a more comprehensive assessment of the micromotion.

Our analyses showed that a more demanding activity, such as the DKB, resulted in a higher range of micromotions, which could be attributed to the loading angle rather than to magnitude of the force. In fact, the peak force was much lower for the DKB (1.35 KN versus 2.4 KN for gait), but produced more than three times higher micromotions (86 µm for DKB versus 25 µm for gait).

Although, we applied a simplified loading profile with only a peak TF load, ignoring patellofemoral (PF) forces in chapters 5 and 7, the FEA study of chapter 4 demonstrated that micromotions at the bone-implant interface were higher when applying a complete DKB loading profile compared to gait, regardless the presence of PF force. As found in chapter 4, the PF force has a stabilizing effect on micromotions, but still a higher flexion angle with a larger TF loading angle caused higher micromotions, and the PF forces could not help to substantially reduce the micromotions. Finally, it is worth noting that average micromotions in all three designs were lower than 150 µm, which is known as a hazardous threshold.

2. Translation to the clinical practice

Providing adequate shear resistance and consequently primary stability is one of the key objectives for the design of surface morphologies. Previously, other design parameters that are important to achieve optimal osseointegration of implants, such as pore size and volume fraction have been investigated extensively [1-3]. It is known that the complex healing process that occurs between the bone and implant depends on the combination of biological, biomaterial, and biomechanical factors [4]. A porous surface structure promotes osseointegration by means of providing a larger contact area, which allows for better migration and adhesion of the cellular bone matrix into the implant [2]. However, several studies have shown that inadequate shear resistance at the bone-implant interface can endanger the healing process by excessive micromotions at the interface [5, 6] that can damage the newly formed tissues. Therefore, it is important that the shear (i.e. frictional) resistance of a novel surface morphology is assessed at an early stage in the design process. However, despite
highly sophisticated pre-clinical biomechanical assessments (experimental testing, FEA and animal studies), it is obvious that the efficacy of novel surface morphologies can only be appropriately proven in a clinical study.

Relative to the importance of bone quality on fixation of cementless implants, it is important to realize that uncemented implants commonly are applied to patients of younger age. However, clinical studies occasionally have reported a high rate of osteoporosis in patients recommended for uncemented joint replacement [7], which highlights that bone quality of patients is not commonly evaluated. In addition, an osteoporotic condition can also negatively affect the process of bone formation [8]. Aro et al. [9] reported that patients with low BMD (osteopenia or osteoporosis) showed higher migration of a cementless femoral stem during the first three months post-operatively due to the lack of osseointegration. Interestingly, BMD has not been analyzed as a factor in failure in long-term clinical studies; however, it is suggested that due to the effect of BMD on the primary stability, a more accurate measurement of BMD may be necessary, especially for female patients with a higher risk of osteoporosis [10].

In this thesis, bone quality was defined in terms of BMD, and clinical CT-scanning was used to measure BMD. However, our BMD measurement protocol was based on a non-clinical approach. In addition, it is not common to CT-scan TKR patients, to prevent exposure to high radiation doses. Dual-energy X-ray absorptiometry (DXA) is the most common clinical tool to diagnose osteoporosis [11], which is based on an estimation of BMD values in terms of a T-score (the number of standard deviations from the normal young adults BMD mean). Due to the two-dimensional nature of this technique, the projection of cortical and trabecular bone is combined and it is represented in terms of an areal BMD (g/cm²). Based on our findings by considering the correlation graphs between micromotion and BMD, a cut-off value of 0.15 g/cm² is found to provide acceptable range of micromotions in cadaveric bones. More research is needed to define the equivalent cut-off value in patients. In addition, the exact relationship between areal and volumetric measure of BMD is not known, though, strong correlations between these two outcomes were found for different anatomical sites [12]. Hence, it is difficult to translate the current cut-off value directly to a clinical measure, although it is strongly recommended to evaluate patient bone quality pre-operatively using available clinical tools.

Another concern is the type of loading that current TKA implants are exposed to. Conventional TKR designs allow flexion angles of up to 120°, and more recent high-flex designs aim for flexion angle as large as 150° [13-16]. Hence, a wide range of activities of daily living (ADL), such as squatting and kneeling, can be expected. However, an association between range of micromotions and flexion angle was proposed in the current thesis. The time frame between surgery and secondary fixation by means of osseointegration has not been established. Jasty et al. [17] studied the torsional stability of cylindrical porous-coated implants in the distal femur
of canines and observed signs of adequate growth in the presence of limited micromotions after 6 weeks. Hofmann et al. [18] reported that signs of developing bone ingrowth into porous-coated titanium implants inserted in distal femur of TKR patients was present until 9 months post-operatively. To prevent the adverse effect of excessive micromotion due to high demand activities such as squatting, it is sensible to recommend patients to avoid these types of activities during a period of around one year post-implantation. In addition, individual rehabilitation plans can be considered for this group of patients in clinical practice.

It is obvious that quantitative assessment of primary stability is not feasible intraoperatively. However, chapter 3 of this thesis showed that implantation force had a strong correlation with bone density, and a higher impaction force was associated with a better fixation. In addition, it is possible to record the impaction force using a surgical mallet instrumented with an impact load cell [19], or, less accurately but more practically, surgeons could try to sense the impaction and build up experience to assess prosthetic stability. In the case of a low or limited impaction force, there is a risk of inadequate primary stability and another type of fixation is preferable.

3. In vitro testing and computer simulation as complementary tools

The principal goal of the current thesis was to improve our understanding of primary stability of press-fit femoral knee implants. Hence, not only was it essential to perceive the complex concept of primary stability, but also a comprehensive evaluation of uncemented knee implants under a set of relevant physiological loading cases was necessary. Therefore, studies using both in vitro testing and computer simulation based on FEA were established to employ their synergistic effect for a better understanding, and also to conquer limitations of either technique [20]. In addition, due to the multi-factorial nature of the problem, it is not possible to investigate all factors using only one of the methodologies. Although not included in this thesis, before performing the first series of in vitro tests (chapter 2 and 3), FEA sensitivity studies were performed in which the effect of bone stiffness, coefficient of friction, shape of femoral pegs, and interference fit on the fixation strength was evaluated. These preliminary sensitivity studies provided valuable data to specify which factors were influential and needed to be investigated more extensively. Using this information, the effects of the identified factors were examined in vitro. Interestingly, it was observed that mainly due to simplifications in the bone material model, such as the absence of viscoelastic behavior, we could not entirely predict our findings in chapter 2 using FEA.

When we shifted our focus towards primary stability (micromotion) in chapter 5, several discussion points concerning the most suitable loading condition were raised as explained in the preceding paragraph. The adequacy of a simplified TF contact force, neglecting the PF force based on a proposed load set-up from the International Organization for Standardization
(ISO 14243-1 and ISO 14243-3) for assessment of micromotion, was questionable [21-23]. In addition, a femoral knee implant experiences more diverse loading regimes during ADLs, which could be more critical than normal walking, based on previous studies for other implant types [24-26]. Finally, our objective was to evaluate multiple femoral knee implant designs (Attune®, Sigma®, and LCS®) with different kinematics and subsequently different loading profiles. It is almost impossible to address all these issues using in vitro testing, and FEA is a powerful tool to easier assess the effects of design and loading condition on micromotions. Therefore, implant-specific loading regimes for different complete physiological loading cycles were provided using a validated finite element model, which had previously reproduced in vivo measured forces [27, 28]. Accordingly, the main objective of chapter 4 was to compare the difference between a simplified loading profile consisting of only a peak TF force and a more complete physiological loading cycle in terms of micromotion. By comparing the micromotion patterns with FEA we could define an appropriate (simplified) loading set up for the experimental measurement in chapter 5. Furthermore, the developed technique in chapter 4 equipped us with a unique pre-clinical testing tool to assess the effect of variations in femoral knee implant designs, such as material properties and shape, on the micromotion in a relatively reliable and quick manner. In our view, chapter 4 and 5 of this thesis have shown that in vitro testing and FEA can be used as complementary tools to reinforce each other.

Nonetheless, it was crucial to validate our finite element model to determine its predictive capacity [29], particularly due to existing differences between predicted and measured micromotions in chapter 4 and 5. Hence, chapter 7 was a validation study using cadaver-specific FE models of implanted distal femoral, which had undergone different loading conditions. A more realistic bone material model entailing post-yield behavior improved the predictive accuracy of the finite element model. However, most likely due to existing simplifications in the current bone material model, which neglects viscoelasticity and abrasion, we did not achieve an entirely validated finite element model in the current thesis, which can be the focus of future studies.

4. Future perspective

At first glance, the very high survival rates of TKR in different national joint registries indicates that developments in the field of knee arthroplasty in the last decades were in the right direction [30]. But at the same time, larger numbers of patients [31] increase the burden of early implant loosening, which generally occurs in a small group. Achieving adequate primary stability is a known factor to prevent early failure. Hence, it is critical to continuously improve our understanding of fixation for both the short and the long term, which is only possible by performing multidisciplinary research in which in vitro testing, FEA modeling,
animal models, clinical trials, follow-up studies, and retrieval studies are combined. In addition, it is essential to constantly verify and develop the conventional set-ups and models of each of these disciplines. In terms of assessment of primary stability, we reached the phase that an advanced bone model was utilized and physiological implant-specific loading profiles were applied. However, no link was made between micromotions and bone ingrowth. Hence, future studies can focus on developing an FEA based tissue adaptation and ingrowth model to relate bone ingrowth to micromotions [32, 33]. Such a model would be useful to judge a new design in a span of multiple years rather than an immediate post-operative period. Furthermore, animal models can be used to verify and investigate whether the predicted bone ingrowth can be found in living tissue.

TKR is also becoming popular in different population groups, such as Asian countries [15], and there are also various surgical techniques, for example posterior versus anterior referencing in TKR or femoral sizing based on soft tissue balancing versus conventional measured resection. These surgical variations can change the kinematics of TKR. In addition, it was already explained that different loading profiles need to be incorporated when assessing primary stability to incorporate different activities. Implant designs should therefore be robust to patients, surgical, and loading variability. However, it is impossible to adequately cover these sources of variation using preclinical test methods as described here. More recently, statistical shape and intensity models (SSIM) or active appearance model were developed, which introduces a new perspective to test thousands of models [34, 35]. In this approach a very small training set is used as a baseline to generate thousands of models. An automatic implantation process can substantially augment this tool for assessing the implant designs much more broadly. It is therefore suggested to develop SSIMs of uncemented femoral knee implants and evaluate the effect of different variables on the primary stability on a larger scale. This aids to evaluate new designs with more confidence, and to identify outliers, which are typically missed when conventional preclinical test methods are employed.

5. Concluding remarks

Comprehensive evaluation of the effect of different design- and patient related factors on the primary stability of uncemented femoral knee implant, employing \emph{in vitro} and \emph{in silico} preclinical testing methods, were performed. These developed methods can be extended to other implant types and designs with some modifications. Current results illustrated the complex interaction between these factors to achieve adequate primary stability in the femoral knee implant. It became obvious that a robust evaluation of implant designs is essential, which can be achieved by including a range of bone quality, loading conditions, and design parameters. In addition, multidisciplinary integrated research combining different biomechanical subfields including implant’s fixation and kinematic, image analysis, and
in-detail evaluation of bone structure, will be important to help manufacturers to improve current implant designs and also to guide surgeons to decide upon the appropriate type of implant and/or fixation method for a specific patient.
References


[15] Ha CW, Park YB, Song YS, Kim JH, Park YG. Increased Range of Motion Is Important


CHAPTER NINE

Summary
Introduction: Chapter 1

Uncemented Total knee replacement (TKR) is often recommended to younger patients (<55 years old) with advanced knee osteoarthritis. There are conflicting reports about the survival rate of TKR’s with this type of fixation, although generally cemented fixation has a better reputation [1, 2]. The most common reason for failure of uncemented TKR is aseptic loosening, which is not well understood [2], but it is known that an inadequate primary stability can lead to early implant loosening. Aseptic loosening of both femoral and tibial knee components has previously been reported [3, 4], and it is therefore essential to understand the underlying mechanism to achieve adequate primary stability in both TKR implants. The main goal of this thesis, therefore, was to investigate the press-fit mechanism and to scrutinize the factors involved in the primary stability of the femoral knee implants. To achieve these goals, in vitro testing using cadaveric bones and in silico methods based on Finite Element Analysis (FEA) were utilized.

In chapter 1, the concepts of primary stability and design- and patient related factors were presented. In addition, it was explained why robust examination of the primary stability of new implant designs using different pre-clinical testing methods is essential.

Press-fit mechanism: Chapters 2 and 3

Several past studies have indicated that interference fit, surface morphology, and bone material properties play a key role in the primary stability of press-fit implants [5-7]. Interference fit is the main component for creating press-fit forces at the bone-implant interface and its interaction with frictional properties of surface morphology leads to generation of shear resistance at the interface. Low bone mineral density (BMD) is known to have an adverse effect on the primary stability [8]. However, the inter-relationship between these factors has previously not been studied thoroughly. In addition, the common assumption that a larger interference fit or a rougher surface would result in a better fixation was occasionally rejected [5, 6]. Understanding the underlying mechanism of press-fit fixation and the effective factors was therefore addressed in the first two chapters of this thesis.

In chapter 2, the effect of interference fit, surface properties, bone material properties, and also their interactions on the fixation strength of press-fit pegs was assessed. Seven cylindrical tapered peg designs of uncemented femoral knee implants with four different diameters, three surface morphologies (low, medium, and high roughness), and at two time points (relaxation time: 0 and 30 min) were inserted into 6 mm holes, which were drilled in distal human femora. BMD was measured using calibrated CT-scans around each hole, and using a randomized block design with BMD as a grouping factor was randomly assigned to samples. A design of experiment (DOE) approach was used to study the interaction between the factors. Pegs were inserted and extracted using an MTS machine, while recording force...
and displacement. A significant positive correlation between BMD and pull-out force was found, which reinforces the need to account for BMD in biomechanical studies and clinical practice. This study also highlighted the importance of relaxation time in studying bone interactions, with surface morphology and interference showing different interaction effects with relaxation time. Although pegs with low roughness (smooth) initially had a higher pull-out force, this effect reduced over time whereas the pull-out force for rough pegs was maintained. Smooth pegs also showed time sensitivity in relation to interference and the benefit of increased interference reduced over time, whereas it was maintained in rough pegs. These findings were explained by different damage mechanisms, compressive versus abrasive, associated with different surface treatment. It was hypothesized that a rougher surface can shift the main component of shear resistance at the interface from normal forces due to bone compression to frictional capacity of surface coating. Additionally, the pull-out force increased with interference fit in a non-linear manner, which emphasizes that a sensible range of interference fit needs to be defined for each press-fit implant. This study indicated that both design- and patient related factors can affect the pull-out strength, although the ultimate effect of each factor was dependent on other parameters due to their interactions.

In chapter 3, an in-vitro experiment was performed to evaluate the fixation strength, implantation force and alignment of an uncemented femoral knee prosthesis with two different surface morphologies applied to the implant surface. The Porocoat® surface coating was used as a clinical benchmark for the comparison with an innovative advanced rough Cobalt-Chromium-molybdenum porous surface coating. Implantation force was included due to its potential effect on intraoperative bone damage [9, 10] and the ease of implantation, also implant alignment was monitored as it is a determining factor for implant survivorship [11]. Nine pairs of femora were used for the implantations using an MTS machine with the uncemented Sigma® femoral knee implant, which had either a Porocoat surface coating (friction coefficient: 0.95) or the advanced surface coating (friction coefficient: 1.4) (DePuy Synthes Joint Reconstruction, Warsaw, USA). Phantom calibrated CT-scans of the bones were obtained to measure BMD, and to measure the anteroposterior (AP) dimension. Additionally, strain gauges were bonded to the external surface of each implant to monitor the reduction of induced bone stresses. After implantation, each specimen underwent two relaxation phases of thirty minutes, separated by one hour of dynamic loading. Force and displacement was recorded during the push-off performed at 150° of flexion, which was adopted as an extreme physiological loading case.

No significant difference was found in the insertion force nor implant alignment between the two groups. Push-off force was dependent on the surface coating with a significantly higher load recorded for the advanced surface coating (P= 0.007). We also observed eight periprosthetic fractures during push-off testing, seven of which were from the advanced
surface coating group, which indicated that the true push-off force in this group would have been even greater. BMD was a significant factor for both insertion and push-off forces (P<0.001), but not the AP-dimension. Strain gauge measurements showed significantly higher initial strain values in the Porocoat group, while the final strain values were similar. The fact that we found that the advanced surface coating did not affect the insertion force, but leads to a 40% increase in push-off force without generating higher strains tells us that the advanced surface coating may abrade, rather than compress, the bone during implantation resulting in lower initial stresses. Despite these lower contact stresses, the frictional properties of the advanced surface were high enough to ensure a higher push-off force and thereby a superior stability relative to the Porocoat design.

**Micromotions at the bone-implant interface: Chapters 4 and 5**

A durable biological fixation between implant and bone depends largely on the micromotions at the bone-implant interface [12, 13]. It is therefore essential to evaluate the primary stability of uncemented femoral knee implant in terms of micromotions. However, it has been shown that the loading condition can significantly affect the micromotions [14, 15]. Additionally, there are reports about a higher incidence of early implant loosening in some high-flex designs [16, 17]. Hence, it is also crucial to evaluate the effect of loading and (high-flex) activity on the micromotion. The primary goal of chapters 4 and 5 was therefore to develop *in silico* and *in vitro* techniques, respectively, to quantify the micromotions. Additionally, the effect of different loading conditions and activities was investigated in these two chapters. In chapter 4, using FEA, we simulated implant-bone interface micromotions during normal gait and deep knee bend (DKB) loading profiles. In addition, two simplified loading conditions consisting of only tibiofemoral (TF) loads, ignoring patellofemoral (PF) loads and applying only a single peak TF force, were included. The aim was to study how these simplifications, which are often incorporated in experimental testing, affect the micromotions. An FE model of a distal femur was generated based on calibrated CT-scans, after which a Sigma cruciate-retaining (CR) Porocoat component (DePuy Synthes Joint Reconstruction, Leeds, UK) was virtually implanted based on the manufacturer’s instructions. Using a frictional contact algorithm (µ=0.95), an initial press-fit fixation was simulated for two interference fits (20 and 200 µm), which was previously verified against experimental data. The micromotions were calculated by tracking the projection of implant nodes on the bone surface. The applied loading patterns were based on discretized simulations, providing incremental loads for each activity based on implant-specific kinematics, derived from the Orthoload database using inverse dynamics [18]. This provided the opportunity to calculate incremental micromotions, but also the *resulting* micromotions, which was the largest possible distance between two points of all projected displacements. In addition, the micromotion (*reference*) relative to the
initial position of the implant (un-loaded phase), which is equivalent to the micromotions that would be measured in an experimental set-up was calculated. The percentage of implant surface area with micromotions less than a defined threshold was calculated. In addition, the peak and average of both reference and resulting micromotions of implant nodes were obtained.

The largest micromotions occurred at the proximal anterior flange, regardless of type of loading condition or activity. The percentage of surface area was increased when the interference fit changed from 20 to 200 µm, particularly for DKB. Tracking nodes over multiple cycles showed implant migration with an interference fit of 20µm. Without the PF force, peak micromotions increased 6% and 22% for gait and DKB, respectively. By applying a single peak TF force micromotions were overestimated. However, the peak reference micromotions corresponded to the maximum TF force, and strong correlations were found for micromotions when comparing a complete loading cycle and a single peak load ($R^2 = 0.73$ and $R^2 = 0.89$ for gait and DKB, respectively). DKB resulted in larger micromotions than gait. This study suggested that a simplified peak force can be used to globally assess the stability of uncemented femoral knee components, which was used later in chapter 5. In addition, we also found that implants should be subjected to different loading modes for more robust testing.

In Chapter 5, the micromotions at the bone-implant interface of two uncemented femoral knee components were investigated while the reconstructions were subjected to the simplified peak TF force of gait and an adapted DKB load. Six pairs of femora were prepared following the normal surgical procedure. Calibrated CT-scans and 3D-optical scans of the bones were obtained to measure BMD and bone cut accuracy, respectively. After implantation of the appropriately sized implants (Left legs: Attune®; right: LCS®), a black-and-white speckle pattern was applied to each reconstruction for measuring relative displacement between bone and implant using digital image correlation (DIC). The micromotion measurement was repeated three times in nine regions of interest (ROIs) close to the interface. Afterwards, implants were pushed-off simulating 150° of flexion, while force and displacement were recorded.

BMD and bone cut accuracy were not significantly different between the groups. Under both loading conditions, Attune had a significantly lower micromotion. Cut accuracy was not a significant factor, and BMD was only significant for the comparison under gait loading (not under DKB conditions). High-flex push-off force was not significantly different, although Attune required a significantly higher load to reach a micromotion of 50 or 150 µm during the push-off test. Potential factors to explain the higher micromotion of LCS were parallel anterior and posterior bone cuts in the LCS versus the tapered bone cuts of the Attune. In addition, LCS has less surface area in contact with bone due to the presence of a rim at the borders of the implant. Taking into account the promising clinical outcome of LCS and also
the lower range of micromotion of Attune, we suggested that the Attune has the potential to be at least as successful as the LCS system from a bone fixation point of view. However, further clinical evaluation of the Attune is necessary to assess its performance on the longer term.

**Validated FE model: Chapters 6 and 7**

Chapters 2 and 3 highlighted the effect of bone material properties on the press-fit fixation. In addition, we observed a complex bone damage mechanism with a combination of bone compression and abrasion during implantation of press-fit pegs in chapter 2. However, we used a linear elastic bone material model in chapter 4 ignoring plastic bone deformation or damage, which forced us to assign a far smaller interference fit to limit simulated pre-stresses at the interface. All these simplifications were the main motivation to evaluate the interference fit and bone damage during implantation in more detail in chapter 6. In addition, an advanced bone material model including non-linear post-yield behavior was assigned to cadaver-specific FE models, which were validated against experimental data in chapter 7.

The goal of chapter 6 was to assess the actual and effective interference fit and the amount of bone damage during implantation of an uncemented femoral knee component. Five cadaveric distal femora were resected and Sigma CR femoral knee components were implanted by an experienced surgeon. Micro-CT scans and conventional CT-scans were obtained pre- and post-implantation for geometrical measurements and to measure BMD. In addition, the position of the implant with respect to the bone was determined by optical scanning of the reconstructions. By measuring the differences in surface geometry, assessments were made of the cutting error, the actual interference fit, the amount of bone damage, and the effective interference fit.

An average cutting error of 0.67± 0.17 mm was found, which pointed mostly towards bone under-resections. We found an average actual AP interference fit of 1.48± 0.27 mm, which was close to the nominal value of 1.5 mm. We observed combinations of bone damage and elastic deformation in all bone specimens, which showed a trend to be related with BMD. Higher BMD tended to lead to less bone damage and more elastic deformation.

In chapter 7, cadaver-specific FEA models were created resembling the experimental situation. Two different material models with either only linear elastic material behavior or including bone plasticity were assigned, while an actual range of interference fit was also simulated. The primary aim was to investigate whether a plasticity model is essential in order to calculate realistic micromotions. Additionally, experimental bone damage at the interface was compared with the FEA simulated range. Micromotions at the interface were measured using DIC technique.

The average experimental micromotion was 53.1± 42.3 μm (mean± standard deviation (SD)),

\[\text{Average experimental micromotion} = 53.1 \pm 42.3 \text{ μm} \]
which was significantly higher than the micromotions predicted by both FE models, using either the plastic or elastic material model (26.5± 23.9 µm and 10.1± 10.1 µm, respectively; p-value< 0.001 for both material models). The difference between the two material models was also significant (p-value< 0.001). The predicted damage had a magnitude and distribution which was comparable to the experimental bone damage. It was concluded that, although the plastic model could not fully predict the micro motions, it is more suitable for pre-clinical assessment of a press-fit TKR implant than using an elastic bone model. Further improvement of the FE model, such as incorporating bone viscoelastic response, adjustment of frictional properties between bone and implant, and implementation of another damage material model could lead to a better predictive ability of the FE model.
References


CHAPTER TEN

Samenvatting
Inleiding: *Hoofdstuk 1*

Jonge patiënten (<55 jaar) met knie-artrose krijgen vaak een ongecementeerde totale knieprothese (TKP). Er zijn tegenstrijdige rapporten over de overlevingskansen van TKP’s met deze vorm van fixatie, hoewel in het algemeen gecementeerde fixatie een betere reputatie heeft [1, 2]. De meest voorkomende vorm van falen van de ongecementeerde TKP is aseptische loslating, maar het verloop van dit proces is nog niet volledig bekend [2]. Wel is bekend dat de kans op vroegtijdige loslating van de knie prothese hoger is bij onvoldoende primaire stabiliteit. Aseptische loslating van zowel femorale en tibiale knie prothesen zijn gerapporteerd [3, 4], en daarom is het essentieel om de onderliggende mechanismen te begrijpen, om zodoende een goede primaire stabiliteit voor beide implantaten te bereiken. Het primaire doel van dit proefschrift was om de press-fit fixatie te onderzoeken en de factoren die betrokken zijn in de primaire stabiliteit van nieuwe prothesen met het gebruik van verschillende pre-klinische testmethoden essentieel is.

**Press-fit mechanism: Hoofdstukken 2 en 3**


In hoofdstuk 2 is het effect van de press-fit, oppervlaktemorfologie, botkwaliteit en de interacties hiertussen op de sterkte van fixatie van press-fit pinnen bestudeerd. Zeven
cilindrische conische pinnen met vier verschillende diameters, drie oppervlaktemorfologieën (laag, midden en hoog ruwheid), en op twee tijdstippen (ontspanning tijd: 0 en 30 min) werden geïmplanteerd in distale femora. BMD werd gemeten rond elke gat met gekalibreerd CT-scans. Een design of experiment benadering werd gebruikt om de interactie tussen de factoren te bestuderen. Pinnen werden geplaatst en verwijderd met behulp van een MTS machine, terwijl kracht en verplaatsing werden opgenomen.

Een significante positieve correlatie tussen BMD en pull-out kracht werd gevonden waaruit bleek dat het belangrijk is om rekening te houden met BMD in zowel biomechanische studies als in de klinische praktijk. Deze studie benadrukt het belang van relaxatie van bot bij het bestuderen van interacties van oppervlaktemorfologie en press-fit. Hoewel gladde pinnen eerst een hogere pull-out kracht hadden was dit minder na een langere wachttijd, terwijl de fixatiekracht voor ruwe pinnen intact bleef. De verschillen tussen de implantaten kunnen worden verklaard door verschillende schademechanismen die optreden in het bot. Een gladder oppervlak zorgt voor meer compressie op de interface, waardoor het bot meer ingedrukt wordt, terwijl een ruwer oppervlak het bot afschraapt. Een ruwer oppervlak zorgt daardoor voor minder compressie, maar heeft wel een hogere wrijvingscoëfficiënt, waardoor de fixatie ook op langere termijn behouden blijft. Deze studie geeft aan dat zowel het ontwerp als de botkwaliteit de fixatie kan beïnvloeden.

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In hoofdstuk 3, werd een in vitro experiment uitgevoerd om de sterkte van fixatie, implantatie kracht en de opstelling van een ongecementeerde femorale knieprothese met twee verschillende oppervlaktemorfologieën te evalueren. De Porocoat® oppervlaktemorfologie werd gebruikt als een referentie voor de vergelijking met een nieuwe ruwe kobalt-chroom-molybdeen poreuze coating. De implantatie kracht werd opgenomen vanwege het potentiële effect op de intra-operatieve bot schade [9, 10] en het gemak van de implantatie. Ook werd de positionering van de implantaten gemeten [11]. Negen paren femora werden geïmplanteerd met de ongecementeerde Sigma® prothese, met een Porocoat oppervlaktemorfologie (wrijvingscoëfficiënt: 0.95), of met de geavanceerde oppervlaktemorfologie (wrijvingscoëfficiënt: 1.4) (DePuy Synthes Joint Reconstruction, Warsaw, USA). Op gekalibreerde CT-scans werden de BMD en de anteroposterieure (AP) afmetingen gemeten. Bovendien werden rekstrookjes op de prothesen geplakt om de afname van botspanningen te bestuderen. Na implantatie had elk specimen twee fasen van ontspanning van dertig minuten, afgewisseld met een uur van dynamische belasting. Kracht en verplaatsing werden gemeten tijdens de push-off test in 150° van flexie, representatief voor een extreme fysiologische belasting.
Er werd geen significant verschil gevonden in implantatie kracht of implantaatpositie tussen de twee groepen. Push-off kracht was afhankelijk van de oppervlaktemorfologie, met een significant hogere belasting voor de geavanceerde oppervlaktemorfologie ($P = 0.007$). Acht breuken werden geobserveerd tijdens de push-off testen, waarvan zeven afkomstig waren uit de geavanceerde oppervlaktemorfologie groep, wat erop wijst dat de push-off kracht in deze groep nog groter zou zijn geweest. BMD was een belangrijke factor voor zowel implantatie en push-off krachten ($P < 0.001$), maar niet voor de AP dimensie. Rekstroommetingen toonden een significant hogere initiële rek in de Porocoat groep aan, terwijl de rekken op langere termijn vergelijkbaar waren. Het feit dat de geavanceerde coating geen invloed had op de implantatie kracht maar leidde tot een toename van 40% in push-off kracht, zonder het genereren van hogere spanningen, toont aan dat de geavanceerde oppervlaktemorfologie tijdens de implantatie het bot afschraapt, wat leidt tot een lagere initiële spanningen. Ondanks deze lagere contact spanningen, waren de wrijvingseigenschappen van de geavanceerde oppervlak hoog genoeg om te zorgen voor een hogere push-off kracht en daardoor een betere stabiliteit ten opzichte van de Porocoat.

**Microbewegingen op het bot-implantaat interface: Hoofdstukken 4 en 5**

Een duurzame biologische fixatie tussen implantaat en bot hangt grotendeels af van de microbewegingen op het bot-implantaat interface [12, 13]. Daarom is het essentieel om de primaire stabiliteit van ongecementeerde femorale knie prothesen te evalueren. De opgelegde belastingen hebben echter een significante invloed op de microbewegingen [14, 15]. Daarnaast zijn er studies over een hogere incidentie van vroege implantaat loslating in sommige high-flex ontwerpen [16, 17]. Daarom is het belangrijk om het effect van opgelegde belastingen en (high-flex) activiteit op de microbeweging te bestuderen. Het primaire doel van de hoofdstukken 4 en 5 was het ontwikkelen van *in silico* en *in vitro* technieken om de microbewegingen te kwantificeren. Bovendien werd het effect van verschillende opgelegde belastingen en activiteiten onderzocht in deze twee hoofdstukken.

In hoofdstuk 4 zijn met behulp van FEA de microbewegingen op de bot-implantaat interface gesimuleerd tijdens lopen en diepe kniebuiging (DKB) belastingprofielen. Daarnaast zijn twee vereenvoudigde belastingprofielen getest om te bestuderen hoe deze vereenvoudigingen, die vaak noodzakelijk zijn in experimentele opstellingen, de microbewegingen beïnvloeden. Een FEA model van een distaal femur werd gebaseerd op gekalibreerde CT-scans, waarna een Sigma CR Porocoat prothese (DePuy Synthes Joint Reconstruction, Warsaw, IN, USA) werd geplaatst. Met behulp van een wrijvend contactalgoritme ($\mu = 0.95$), werd een press-fit fixatie
gesimuleerd met twee verschillende press-fits (20 en 200 micron). De toegepaste belastingen waren gebaseerd op simulaties van implantaat-specifieke kinematica, afgeleid van de Orthoload database [18]. Het percentage van het prothese-oppervlak met microbewegingen minder dan een bepaalde drempelwaarde werd berekend.

De grootste microbewegingen vonden plaats aan het proximale deel van de anterieure flange, ongeacht de belasting of activiteit. De microbewegingen werden lager wanneer de press-fit werd verhoogd van 20 naar 200 µm, vooral voor de DKB belasting. Onder een vereenvoudigde belasting namen de microbewegingen toe met respectievelijk 6% en 22%, voor lopen en de DKB. Het toepassen van alleen de maximale tibiofemorale kracht leidde tot een overschatting van de microbewegingen. Er werd echter een sterke correlatie gevonden voor micro bewegingen tussen een volledig belastingprofiel en een enkele piekbelasting ($R^2= 0.73$ en $R^2= 0.89$ voor lopen en DKB). De DKB belasting resulteerde in grotere microbewegingen dan onder een loopbelasting. Deze resultaten suggereren dat een vereenvoudigde piekkracht kan worden gebruikt om de stabiliteit van ongecementeerde femorale knie prothesen te evalueren (zie hoofdstuk 5). Daarnaast blijkt uit deze studie dat het belangrijk is om reconstructies te testen onder verschillende belastingprofielen.

In hoofdstuk 5 werden de microbewegingen op de bot-implantaat interface van twee ongecementeerde femorale knie prothesen gemeten, terwijl de proefstukken werden belast met de vereenvoudigde piekkracht tijdens lopen en met een aangepast DKB belasting. Zes paren femora werden voorbereid volgens de standaard chirurgische procedure. Gekalibreerde CT-scans en 3D-optische scans van de botten werden verkregen om BMD en nauwkeurigheid van de botresectie te meten. Na implantatie van de juiste maat prothese (links: Attune®; rechts: LCS®), werd een zwart-wit stippen patroon aangebracht op iedere reconstructie om de microbewegingen te meten met behulp van digital image correlation (DIC). De metingen werden drie keer herhaald in negen regio’s nabij de interface. Daarna werden prothesen van de femora geduwd in 150° flexie, terwijl de kracht en de verplaatsing werden gemeten.

BMD en nauwkeurigheid van botresectie waren niet significant verschillend tussen de groepen. Onder beide opgelegde belastingen had Attune significant lagere microbewegingen. Nauwkeurigheid van botresectie was geen significante factor en BMD was slechts significant voor de vergelijking met de loopbelasting (niet onder DKB). High-flex push-off krachten waren niet significant verschillend, hoewel voor Attune een significant hogere belasting nodig was om microbewegingen van 50 of 150 µm te bereiken. De verschillen tussen de twee implantaten worden mogelijk veroorzaakt door verschillen in de vorm van de botresecties.
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(parallel voor LCS en tapered voor Attune) en door de grootte van het contactoppervlak tussen bot en implantaat, wat kleiner is voor LCS. Hoewel de Attune lagere microbewegingen had dan de klinische succesvolle LCS, is klinische evaluatie van de Attune noodzakelijk om de lange termijn resultaten te beoordelen.

Validated FE model: Hoofdstukken 6 en 7


Het doel van hoofdstuk 6 was om de effectieve press-fit en de hoeveelheid van botschade te beoordelen tijdens de implantatie van een ongecementeerde femorale knie prothese. Vijf humane kadaverfemora werden geïmplanteerd met een Sigma CR prothese. Micro-CT scans en conventionele CT-scans werden pre- en post-implantatie gemaakt om de afmetingen te bepalen en de BMD te meten. Bovendien werd de plaatsing van het implantaat ten opzichte van het bot bepaald door middel van optische scans van reconstructies. Door het meten van het verschil in oppervlak geometrie werden de botresectie fout, de initiële press-fit, de botschade, en de effectieve press-fit gemeten.

Een gemiddelde botresectie fout van 0.67± 0.17 mm werd gevonden, die meestal duidde op onder-resectie van het bot. We vonden een gemiddelde actueel press-fit van 1.48± 0.27 mm, die in de buurt van de nominale waarde van 1.5 mm was. We zagen combinaties van botschade en elastische vervorming in alle bot proefstukken, wat correleerde met BMD. Hogere BMD leidde tot minder botschade en meer elastische vervorming.

In hoofdstuk 7 werden vergelijkbare kadaver-specifieke FEA modellen gemaakt. Materialmodellen met lineair elastisch materiaal gedrag of met botplasticiteit werden gebruikt, terwijl de werkelijke press-fit werd gesimuleerd. Het primaire doel was om te onderzoeken of een plasticiteitsmodel van belang is om realistische microbewegingen te
berekenen. Bovendien werd de experimentele botschade op de interface vergeleken met de gesimuleerde botschade. Microbewegingen op de interface werden gemeten met digital image correlation.

De gemiddelde experimentele microbeweging was 53.1±42.3 micron (gemiddelde ± standaarddeviatie), die significant hoger was dan de FE microbewegingen voorspeld door de twee materiaalmodellen (plastisch model (26.5± 23.9 micron) of elastisch model (10.1± 10.1 micron), p <0,001 voor beide materiaal modellen). Het verschil tussen de twee materiaalmodellen was ook significant (p-waarde <0,001). De gesimuleerde botschade hadden een grootte en distributie die vergelijkbaar was met de experimentele botschade. Er werd geconcludeerd dat, hoewel het plasticiteit model de microbewegingen niet volledig kon voorspellen, het geschikter is voor het simuleren van een press-fit TKP prothese dan een elastisch botmodel. Verdere verbetering van het FEA model, zoals inclusie van viscoelastisch gedrag, aanpassing van de wrijvingseigenschappen tussen bot en prothese, en gebruik van andere damage modellen zou kunnen leiden tot een nog betere voorspelling van de microbewegingen.
References


[13] Jasty M, Bragdon C, Burke D, O’Connor D, Lowenstein J, Harris WH. In vivo skeletal responses to porous-surfaced implants subjected to small induced motions. The Journal of


CHAPTER ELEVEN
Acknowledgments
PhD portfolio
List of publications
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Acknowledgments

I still clearly remember the first summit that I climbed, Mount Tochal (3933 m). It was very challenging and asked a lot of persistence especially for a 15 years old teenager. At that time, I felt very proud, but it was just years later after climbing many more summits that I realized, climbing a summit is not all about you. A trustful leader (s) and companionable team members are essential for a successful and enjoyable climbing experience. Six years ago when I moved to the Netherlands, I expected that there would be no opportunity for climbing in a lowland country. But this was a wrong assumption, as obtaining a Ph.D. degree was the most challenging summit ever that I achieved. And of course it was not possible without a trustful leader (s) and companionable team members, whom I would like to express my deep gratitude here.

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you did as well. Once more many thanks for your enthusiasm and support.

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Dear Loes, Anne, Priyanka, and Karlijn, dear Dames, I find myself very lucky to share such a ‘gezellig kiphoekje’ with you during all these years. As an expat with a small circle of friends, you were always beyond colleagues for me and it was more a friendship that even last after some of you left the ORL. We have shared so many up and down moments from preparing personal gifts for other colleagues to have long discussions about our personal challenges as a Ph.D. candidate. You were great team members in my journey to the summit and I would like to thank you from the bottom of my heart for every delightful moment.

A very unique experience of my journey was its dynamicity that I could meet so many wonderful people; some are still at the ORL and some already left the ORL. I shared good memories with all of them in different occasions such as lunch breaks, lab BBQ, conferences or ORL events and meetings. The list is of course very long, but still I would like to recall all names. Dear Esther Tanck, Tony, Jorret, Maria, Hendi, Astrid, Pawel, Rene, Pieter Holtman, Pooya, Joke, Lennert, Thom, Marco, Wouter, Erwin, Kai, Hamid, Florieke, Aga, Veronica, Wojtek, Jan, Josa, Esther, Branco, Thomas, and Benjamin, I would like to thank you for all good memories. I also met many amazing people at Flinders University, dear Solmaz and Amin, Vida and Amir, Elli and Amin, Azin and Ali, Javad, Maged, Rami, Hamed, Rowan, Alberts, Mark, Laura, Dhara, Isabelle, Christina, Nathan, Dermot, and Bryant, you made my time at Flinders much more memorable.

I received help from several interns to accomplish couple of chapters of my thesis. Dear Sal, Maartje, Mariska, Ellemiek, and Dieuwke, your enthusiasm and input made my journey more
enjoyable, many thanks for that!

It was my privilege to be the secretary of Radboud Institute for Health Sciences (RIHS) Ph.D. Council for two years. I met so many great people and together we arranged different scientific and social activities. Also, Prof. Bart Kiemeney and Dr. Marieke de Visser supported the Ph.D. council with organizing those activities. I would like to thanks them all for great time and memories.

Fantastic friends and family members also accompanied me during this journey whom I shared great moments with and I would like to extend my sincerest gratitude for their friendship and support.

So many amazing friends made my time more pleasant in Nijmegen, some of them even I got to know before I moved to Nijmegen. Dear Shahla and Mehran, Parisa and Ali, Roya and Ehsan, Bahare and Alireza, Parisa, Marzieh, Elnaz and Yoolla, Cristiane and Roger, Elisa, Sevda and Ali, many thanks for being great friends, all good memories and your help whenever I asked for.

It is difficult to imagine that friends can be like a family, but having you, dear Zohre, Maral and Nikos, Bita, Niloofar and Shahin, Milad, Lili, and Bahareh, in good and difficult moments proved me that fantastic friends can be the same as family. I am very proud to have you as my friends and hope our friendship last forever. Thanks for all enjoyable moments, dances, and great foods.

Dear Freidoon, I would like to express my deepest gratitude to you for support with my party.

Dear Mario and Annemiek, our friendship is not so old but sometimes they are friends who you don’t need to know for ages to feel very close with them. I find myself very lucky to have you as my friends and hope to have so many more K3 gatherings in the upcoming years. Thanks for being so great and kind friends!

Dear Asal, my oldest friend, though we were far from each other, our contact was always in place. It was great that last year you came to the Netherlands and it was very enjoyable to catch up and to try different exciting winter sports together. I am happy to have you as my friend and thanks for being a great friend.

Dear Nasim, we shared so many special moments when we were so young and full of dreams. I am happy that so many of our dreams came true and I am quite sure that other dreams to travel South America and Africa together will also happen soon. Our hours of walk and talk on the streets of Tehran, Venice, Amsterdam, Antwerp and Melbourne are so memorable and
I would like to thank you from bottom of my heart for every second of our friendship.

Dear Afsaneh, Andre and Aryan, my journey to the Netherlands was somehow the very first start point of my Ph.D. journey and you were very much involved in this journey from the moment that Hamid and I arrived to Schiphol. Your hospitality especially on the first months of our stay is unforgettable and I would like to thank you for your incredible support and kindness.

My dearest Hamid, my dream was always to climb as many as possible summits next to a person whom I love. And how lucky I was to meet you so young and we could share the best years of our life to achieve so many summits shoulder to shoulder. We even met each other while climbing a summit and built our relationship while climbing other ones. I know that there were moments that I was not a pleasant companion, especially when I was finishing a project or finalizing a publication, but you were always so patient and your love was so strong that returned me back to the track. My gratitude for all fantastic moments cannot be expressed by words and I hope to climb many more summits next to you as this is all I want.

Sanaz Berahmani, January 2017
## PhD Portfolio

Name PhD student: **S. Berahmani, MSc**  
PhD period: **01-05-2012 / 27-03-2017**  
Department: **Orthopaedic Research Lab**  
Promotor(s): **Prof. Dr. ir. N. Verdonschot**  
Co-promotor(s): **Dr. ir. D. Janssen**  
Graduate School: **Radboud Institute for Health Sciences**  

### TRAINING ACTIVITIES

#### a) Courses & Workshops

<table>
<thead>
<tr>
<th>Course</th>
<th>Year(s)</th>
<th>ECTS</th>
</tr>
</thead>
<tbody>
<tr>
<td>RIHS Introduction Course, RIHS, Radboudumc, Nijmegen</td>
<td>2014</td>
<td>1.4</td>
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<tr>
<td>Careers Workshop PhD, Radboudumc, Nijmegen</td>
<td>2015</td>
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<tr>
<td>Presentation Skills, Radboud into language, Nijmegen</td>
<td>2014</td>
<td>0.5</td>
</tr>
<tr>
<td>Scientific Integrity, Radboudumc, Nijmegen</td>
<td>2014</td>
<td>1.4</td>
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#### b) Seminars & lectures

<table>
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<tr>
<th>Seminar/Workshop</th>
<th>Year(s)</th>
<th>ECTS</th>
</tr>
</thead>
<tbody>
<tr>
<td>RIHS PhD council- Workshop on Networking for PhD students</td>
<td>2013</td>
<td>0.1</td>
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<tr>
<td>(Technical Forum)</td>
<td></td>
<td></td>
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<tr>
<td>RUMC PhD councils wide workshop-How to convince the editor</td>
<td>2013</td>
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<tr>
<td>(Technical Forum)</td>
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<tr>
<td>RIHS PhD council workshop- How to present your poster</td>
<td>2014</td>
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<tr>
<td>(Technical Forum)</td>
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<tr>
<td>PON workshop- How to print your thesis</td>
<td>2015</td>
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<tr>
<td>(Technical Forum)</td>
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<tr>
<td>UMC-wide workshop on Social Media and Disruptive Technology</td>
<td>2015</td>
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<td>(Technical Forum)</td>
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#### c) Symposia & congresses: (OP/PP indicates oral or poster presentation)

<table>
<thead>
<tr>
<th>Event</th>
<th>Year(s)</th>
<th>ECTS</th>
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<tbody>
<tr>
<td>Nederland Orthopedische Veriniging, Amsterdam, NL. (OP)</td>
<td>2013</td>
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<tr>
<td>Dutch Bio-Medical Engineering Conference, Egmond aan Zee, NL. (OP)</td>
<td>2013</td>
<td>0.75</td>
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<tr>
<td>3rd NCEBP PhD Council Retreat, Wageningen, NL. (PP)</td>
<td>2013</td>
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<tr>
<td>Annual congress of International society of arthroplasty (ISTA),</td>
<td>2013</td>
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<tr>
<td>Florida, USA. (OP+ 2 × PP)</td>
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<tr>
<td>Knee Surgery And Rehabilitation- Institution of mechanical engineers</td>
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<tr>
<td>(ImechE), London, UK. (OP)</td>
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<td>World Congress of Biomechanics (WCB), Boston, USA. (PP)</td>
<td>2014</td>
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<tr>
<td>NCEBP Science day, Nijmegen, NL.</td>
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<tr>
<td>International Symposium on Computer Methods in Biomechanics and</td>
<td>2014</td>
<td>1.75</td>
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<tr>
<td>Biomedical Engineering, Amsterdam, Netherlands. (2 × OP)</td>
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</table>
Annual congress of International society of arthroplasty (ISTA), Kyoto, Japan. (OP+PP) 2014 1.75
Symposium on Statistical Shape Models and Applications, Delémont, Switzerland 2014 0.75
21st congress of the European Society of Biomechanics, Prague, Czech Republic. (OP) 2015 1.25
Annual congress of International society of arthroplasty (ISTA), Vienna, Austria. (OP) 2015 1.25
22nd Annual Scientific Meeting of Australian and New zealand Orthopaedic Research Society, Melbourne, Australia. (OP) 2016 1
6th Dutch Bio-medical Engineering Conference, Egmond aan Zee, NL. (PP) 2017 0.75
Nederland Orthopedische Verinniging, Den bosch, NL. (OP) 2017 0.5

**d) Other**

- Journal club- Orthopaedic Research Lab 2013-2015 3
- Lablunches- Orthopaedic Research Lab 2013-2015 3
- Secretary of the PhD Council of Radboud Institute for Health Sciences 2013-2014 2

**TEACHING ACTIVITIES**

**e) Lecturing**

<table>
<thead>
<tr>
<th>Event</th>
<th>Year(s)</th>
<th>ECTS</th>
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<tbody>
<tr>
<td>MSc course- 5HM02 - Tissue: Biomechanics and Engineering</td>
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<tr>
<td>BSc course- Mechanical loading device- Introduction to clinical human movement sciences</td>
<td>2013-2015</td>
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<td>BSc course- Orientation day</td>
<td>2013-2014</td>
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<tr>
<td>BSc course- Journal club</td>
<td>2015</td>
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**f) Supervision of internships / other**

<table>
<thead>
<tr>
<th>Internship</th>
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<th>ECTS</th>
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<tbody>
<tr>
<td>BSc internship- S.N. van Kessel</td>
<td>2012</td>
<td>1</td>
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<tr>
<td>BSc internship- M. Adriaansen</td>
<td>2012</td>
<td>1</td>
</tr>
<tr>
<td>BSc internship- E.E. Koenders</td>
<td>2013</td>
<td>1</td>
</tr>
<tr>
<td>MSc internship- M. Hendriks</td>
<td>2015</td>
<td>3</td>
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</table>

**TOTAL** 39.5
LIST OF PUBLICATIONS

JOURNAL PAPERS


Berahmani S, Janssen D, Verdonschot N. Experimental and computational analysis of micromotions of an uncemented femoral knee implant. Journal of Biomechanics (Submitted)


PODIUM CONFERENCE PRESENTATIONS


and bone damage of an uncemented femoral knee implant”. 29th congress of the International Society for Technology in Arthroplasty, 5- 8 October, Boston, USA.


component”. Nederland Orthopaedische Vereniging (NOV) Jaarvergadering, 7- 8 February, Amsterdam, Netherlands


**OTHER CONFERENCE PRESENTATIONS**


Curriculum Vitae

Sanaz Berahmani was born in Tehran, Iran on April 6th, 1986. She obtained her high school diploma in Mathematics and Physics in 2004 from Paknezhad High school in Tehran. Immediately afterwards, she started her bachelor’s study in Biomedical Engineering-Biomechanics at the Amirkabir University of Technology in Tehran. After her graduation in 2009 she worked as a sales engineer, followed by a position as commercial manager at AMANCO INTL. She then moved to the Netherlands in September 2010 to pursue her master study in Biomedical Sciences at the Radboud University in Nijmegen. After finalizing her master’s internship at the Orthopaedic Research Lab (Radboud university medical center, Nijmegen) under supervision of Dr. Daan Waanders and Prof. Nico Verdonschot, she was asked to work on her master thesis, which actually was the starting point for the current thesis. Hence, in April 2012 after her graduation, she started working on her PhD project under supervision of Dr. Dennis Janssen and Prof. Nico Verdonschot at the Orthopaedic Research Lab, under a research grant from DePuy Synthes Joint Reconstruction.

During her studies, she published several publications and presented her work at several national and international conferences. One of her studies was selected as the best scientific presentation at the annual meeting of the Dutch Orthopaedic Association (2013), and another manuscript was voted the best biomechanics paper in the field of arthroplasty by the International Society for Technology in Arthroplasty (2015). In May 2016 she moved to Adelaide, Australia, for six months to perform a research fellowship at Flinders University, under supervision of Prof. Mark Taylor. Her visiting project entitled ‘A population based study to evaluate the effect of rotational alignment on the primary stability of an uncemented femoral knee implant’ became possible by her receiving several awards: the ‘Frye Stipendia’ from Radboud University, the ‘Mobility award’ from the European Society of Biomechanics, and a travel grant from the Anna Fonds. Next to her research activities, Sanaz has been the secretary of the PhD Council of the Radboud Institute for Health Sciences for two years, which started in September 2012.

Currently she is employed as a postdoctoral researcher at the Radboud university medical center and is involved in the development and medical regulation of osseointegration implants for patients with a lower limb amputation.