

FACULDADE DE ENGENHARIA DA UNIVERSIDADE DO PORTO

Initial Stability of Non-Coated *vs* Pre-Coated Tibial Components Under Cyclical Load

The Effect of Fat Contamination

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Abstract

Introduction: The gradual degeneration of the cartilage in the joint is one of the most prevailing reasons for total knee replacement surgery. Despite this method being typically linked with satisfactory patient recoveries, revision surgery is nevertheless needed for at least 5% of the population. This subsequent procedure is associated with substantial pain, discomfort, probable joint instability, an increasing financial burden, and greater fatality rates. Avoiding or reducing the need for re-operation is thus a priority for the orthopedic community, which can only be accomplished by better understanding the failure mechanisms associated with the condition. Infections, instability, and aseptic loosening of the tibial component have all been linked to the failure of initial total knee replacements.

Goals of the project: The main aim of this project is to investigate the influence of implant design alteration as well as fat contamination on the fixation of the tibial baseplate.

Materials and methods: Two NexGen® (NexGen® Complete Knee Solution, Zimmer, Warsaw - Indiana, USA) tibial implants, one non-coated and one PMMA pre-coated, were implanted on SawBone® (Pacific Research Laboratories Inc., Vashon, WA) tibia bone models with PALACOS R+G® (Heraeus Medical GMBH, Germany) bone cement. Using a cyclical load to simulate a level-walking, three linear variable differential transformers (LVDT) recorded both the axial migration and micromotion suffered by the implant throughout the test. After this pre-conditioning state, a push-out test was done to access the adhesion strength for the implant and the bone-cement interface. Data were analyzed using Matlab® script. For statistical tests, two-factor ANOVA was used to explore the effects of the PMMA coating and the contamination on the groups using GraphPad Prism® (GraphPad Software, San Diego, CA, USA) software. Post-hoc power analysis was taken into consideration for the discussion of the results' strength.

Results: For stability metrics, neither the design alteration nor the contamination had any significant impact on migration or micromotion. Non-coated dry samples had mean maximum subsidence of 3.17 μm , and the same design when contaminated had mean subsidence of 3.58 μm . Pre-coated dry samples presented 4.08 μm for average maximum subsidence and their fat-contaminated counterpart, 10.46 μm . The highest micromotion was usually situated on the posterior-lateral site, except for pre-coated contaminated samples when this value was mostly recorded at the posterior-medial position and ranged from 20.50 μm to 22.45 μm for group means. In the case of push-out tests, the impact of the contamination was significant enough to be detected (ANOVA $p = 0.0268$). Non-coated groups had a mean failure load of 6090.85 N and 3316.67 N for dry and contaminated conditions (Tukey $p = 0.0407$) while pre-coated had 5265.00 N and 5475.15 N, respectively.

Discussion and conclusion: The differences in design were too minimal to cause a change in stability metrics. Pre-coated implants seem to cope better when implanted in fat-contaminated conditions than non-coated ones. For future setups, the use of cadaveric bones coupled with the design of a pull-out test rig would be advised to increase the clinical relevance of the project. Besides that, the incorporation of three-dimensional visual sensors would bring the measurements of the stability metrics closer to clinical RSA studies. Further investigations into the impact of load distribution on implant debonding might prove useful for the community.

Resumo

Introdução: Uma das razões mais prevalentes para a cirurgia de substituição total do joelho é a degeneração gradual da cartilagem na articulação. Apesar deste método estar tipicamente ligado a recuperações satisfatórias dos pacientes, a cirurgia de revisão é no entanto necessária para pelo menos 5% da população. Este procedimento subsequente está associado a dores substanciais, desconforto, provável instabilidade articular, uma carga financeira crescente, e maiores taxas de mortalidade. Evitar ou reduzir a necessidade da reoperação é assim uma prioridade para a comunidade ortopédica, o que só pode ser conseguido através de uma melhor compreensão dos mecanismos de falha associados à condição. Infecções, instabilidade e afrouxamento asséptico da componente tibial foram todos ligados à falha primária das próteses totais do joelho.

Objectivos: O principal objectivo deste projecto é investigar a influência da alteração do design do implante, bem como a contaminação por gordura na fixação do componente tibial.

Materiais e métodos: Dois implantes tibiais NexGen® (NexGen® Complete Knee Solution, Zimmer, Warsaw - Indiana, USA), um não revestido e um pré-revestido com PMMA, foram implantados em modelos ósseos da tibia SawBone® (Pacific Research Laboratories Inc., Vashon, WA) com o cimento ósseo PALACOS R+G® (Heraeus Medical GMBH, Germany). Utilizando uma carga cíclica para simular uma marcha em superfícies niveladas, três transformadores diferenciais lineares variáveis (LVDT) registraram tanto a migração axial como a micro-moção sofrida pela peça durante todo o ensaio. Após este estado de pré-condicionamento, foi feito um teste de push-out para aceder à resistência de fixação do implante na interface entre o osso e o cimento. Os dados foram analisados utilizando um script Matlab®, para testes estatísticos, foi utilizada ANOVA de dois factores para explorar os efeitos do revestimento PMMA e a contaminação nos grupos utilizando o software GraphPad Prism® (GraphPad Software, San Diego, CA, EUA). A análise de potência post-hoc foi levada em consideração para a discussão da solidez dos resultados.

Resultados: Para métricas de estabilidade, nem a alteração do design nem a contaminação tiveram qualquer impacto significativo na migração ou micro-moção. As amostras secas não revestidas tiveram uma descida média máxima de 3,17 µm, o mesmo design quando contaminado teve uma descida média de 3,58 µm. As amostras pré-revestidas apresentaram 4,08 µm para uma descida média máxima e a sua contraparte contaminada com gordura, 10,46 µm. A maior micro-moção situou-se normalmente no local posterior-lateral, com excepção das amostras pré-cobertas e contaminadas que tiveram-na situada no local posterior-medial. Tais valores variaram entre 20,50 µm a 22,45 µm. No caso de testes de push-out, o impacto da contaminação foi suficientemente significativo para ser detectada (ANOVA $p = 0,0268$). Os grupos não revestidos tiveram uma carga média de falha de 6090,85 N e 3316,67 N para condições secas e contaminadas (Tukey $p = 0,0407$), enquanto que os grupos pré-revestidos tiveram 5265,00 N e 5475,15 N, respectivamente.

Discussão e conclusão: As diferenças de design foram demasiado mínimas para causar uma mudança nas métricas de estabilidade. Os implantes pré revestidos parecem lidar melhor quando implantados em condições de contaminação do que os não revestidos. Para configurações futuras, a utilização de ossos cadavéricos juntamente com o desenho de um teste do tipo pull-out seria aconselhado, a fim de aumentar a relevância clínica do projecto. Além disso, a incorporação de sensores visuais tridimensionais aproximaria as medições das métricas de estabilidade dos estudos clínicos de RSA. Outras investigações sobre o impacto da distribuição da carga no descolamento dos implantes do joelho poderiam revelar-se úteis.

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Chapter 1

Introduction

1.1 Basic knee anatomy description

Having a normal knee would make life a lot easier.

(Andy Murray)

The understanding of the functionality and its relation with the form of the anatomic structures entangled with the development of biomechanics as a scientific field allowed scientists to continue improving human prosthetic and surgery methodology.

Fundamentally the knee is a type of diarthrodial "hinge joint" - such that motion is only allowed in one plane and to a certain degree of freedom - connecting the distal portion of the femur to the proximal one of the tibia. Its stability is guaranteed through the meniscus (medial and lateral) and the myriad ligaments, preventing valgus/varus positioning and anterior/posterior displacement of the whole tibia.

The description of the Knee joint can be segmented into its various components, as seen in the 2017 work of Enrico Vaienti *et al.* "Understanding the human knee and its relationship to total knee replacement" and in the 2022 book "Basics in Primary Knee Arthroplasty" by Becker, R. *et al.* and as summarized on figure 1.1. To unfold the joint's elements, some essential definitions are needed such as the elucidation of the term condyle. A condyle is characterized as an anatomical protuberance that supports the joint's hyaline cartilage, bearing most of the force passing through the articulation itself [1, 2].

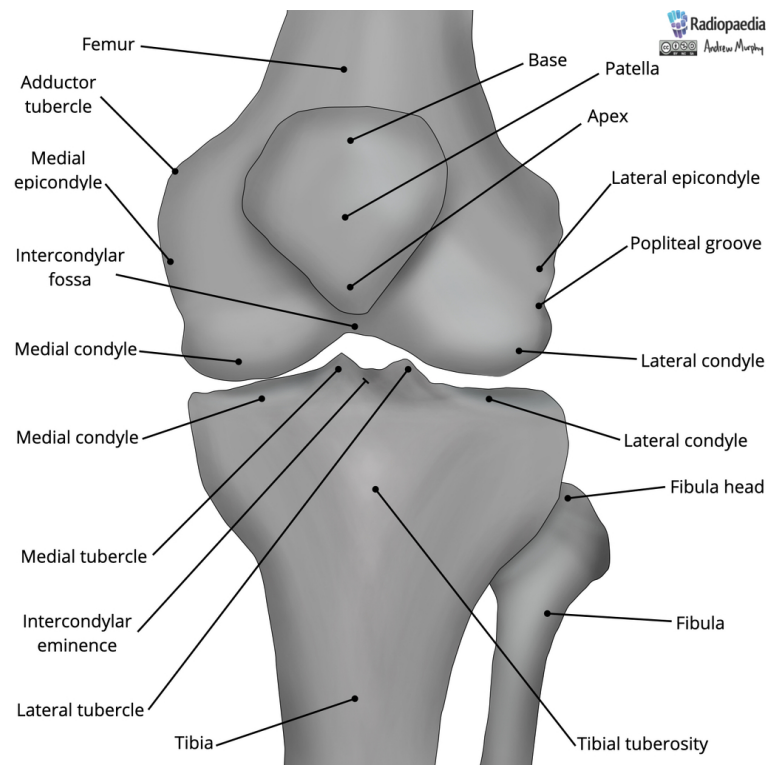


Figure 1.1: Diagram Illustration of The Knee and its Components [3].

1. Bone Components:

(a) The Distal Femur:

The femur is the human body's largest, strongest and heaviest bone and it articulates proximally to the hip joint and distantly to the knee. Its functions are mostly the resistance to bigger loads and the stability of the gait.

At its distant end, its "cuboid-shaped" base ends in two rounded condyles forming the upper part of the knee joint and delineated, posteriorly, by the intercondylar notch. Laterally, the epicondyles - bony structures near the condyles - present a binding site for the collateral ligaments while also serving as a guide for the professionals to assert the femur's rotation.

Finally, the anterior "groove" limited by medial and lateral bone ridges present on the Femur is called *femoral trochlea* and it articulates with the posterior face of the patella, allowing for its gliding movements during flexion and extension of the whole knee.

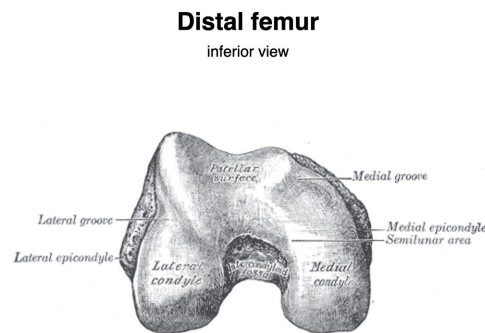


Figure 1.2: Gray's Illustration of the Distal Femur Anatomy [4].

(b) Proximal Tibia:

The tibia is the lower leg's main "weight-bearing" bone, being both "larger and stronger" than its counterpart. It is located medially from its "partner," the fibula, to which it is attached by a fibrous membrane, and participates in both the proximal and distal knee and ankle joints.

The tibial portion of the Knee Joint is comprised of two condyles (lateral and medial), two tibial plateaus - described as the "articular surface of each condyle" - and an intercondylar area - site of attachment for its various ligaments and segmented into anterior and posterior areas, as well as an intercondyloid eminence, also known as tibial spine. Mechanically, the main portion of the weight supported by this bone is done medially. When analyzed, the tibial axis is sloped posteriorly. This characteristic allows for knee flexion, with higher angles correlating to bigger flexion degrees.

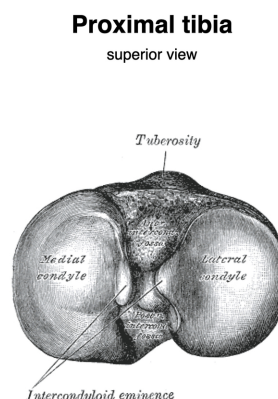


Figure 1.3: Gray's Illustration of the Proximal Tibia Anatomy [4].

2. The Menisci:

Connecting and revolving the femoral condyles to the tibial plateaus, there are two "C-shaped" fibrocartilaginous discs known as Menisci.

These structures function as a supporter for mechanical/kinematic aspects of the joint such as providing a better load transmission and distribution, absorbing the eventual shocks, and improving the diffusion of synovial fluid along the joint, therefore, improving the joint's lubrication.

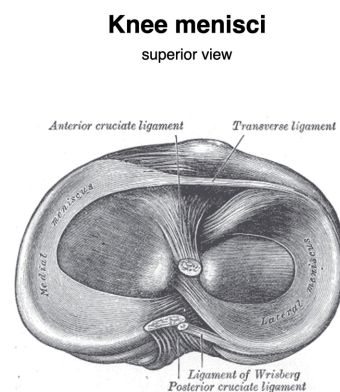


Figure 1.4: Gray's Anatomy Illustration of the Menisci surface at the Tibial Proximal End [4].

3. The Ligaments:

There are several ligaments related to the Knee Joint. In this anatomical description, however, only the intracapsular ones will be described as these are usually the main causes of concern regarding Total Knee Arthroplasty surgery.

(a) Cruciate Ligaments:

i. Anterior Cruciate Ligaments (ACL):

Acting as a constraint for joint motion, both angularly and in "drawer" movement, the ACLs are multifascicular structures based on several collagen bundles and physically shaped as an hourglass. The ligament is separated into two unique areas, anteromedial (AM) and posterolateral (PL), with various orientations at the insertion site to the tibia and varied behaviors depending on the joint condition. The ACLs attach to the tibia anteriorly on the tibial footprint (the region between the intercondylar eminences on the tibial plateau) and to the femur lateral to the intercondylar ridge.

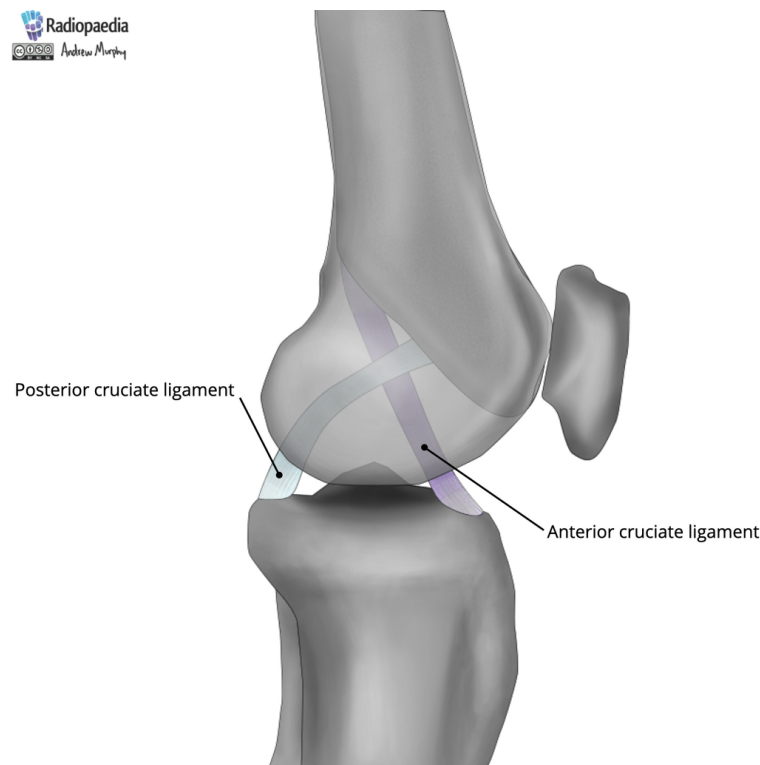


Figure 1.5: Diagram Illustration of the Ligaments and its insertion sites [3].

ii. Posterior Cruciate Ligaments (PCL):

PCLs are responsible to constrain movements of translation of the joint, the mobility in varus and valgus directions as well as some eventual external rotation. Similarly to ACLs, PCLs can be subdivided into two regions regarding insertion site, the anterolateral bundle (ALB) and the posteromedial bundle (PMD) that work in harmony to promote the stability of the joint. PCLs run from the anterior face of the medial femoral condyle to the posterior of the tibial sulcus.

1.2 Knee arthrosis

Described as a degenerative joint disease followed by a significant narrowing of the "joint space", arthrosis is, currently, the most recurrent of the so-called "Bone Conditions", accounting for the vast majority of the reasons for a Total Knee Arthroplasty Surgery. The "bone-to-bone" friction caused by the lack of sufficient cartilaginous tissue causes significant pain and swelling of the knee as well as instability of the joint. Many risk factors are associated with the disease, including age, weight, sex, and lifestyle. There is, currently, no available cure for osteoarthritis with treatments mostly tackling the patient's symptoms and aiming to reduce the disease evolution [5–7].

1.3 Knee implants - designs and function overview

Although the idea of using artificial materials to replace damaged parts of the human body is not new, it was only in the last centuries that technology made it a reality. By 1973 keystones were defined regarding the goal of a successful implant aiming to provide better overall joint stability, keeping the health of the implanted area, and providing a better experience in the case of a second surgery being needed at the site [8–10].

According to the Norwegian National Registry of Arthroplasty, in the period between 1994 until 2020, a total of 112.096 knee prostheses surgeries were done, counting for primary operations, re-operations and revisions. In the same period, the numbers associated with total knee replacement were 10x bigger than either shoulders or "others" joint replacement, i.e anything besides hips, knees, and shoulders replacement [11].

Based on the level of intervention during the replacement surgery as well as the patient's conditions two prosthetic models to tackle the failure of the knee joint were made. The following summary introduces them in descending order of "constrainability" [12].

1. Fixed Bearing

- (a) Constrained: Implants of this type provide the most stability to the joint and, therefore, are usually indicated for patients with moderate to serious bone loss combined with damaged ligaments. This positive side is counter-affected by the massive amount of bone resection needed to accommodate the larger implants when compared to other designs as well as the increased possibility of aseptic loosening due to abnormal constraints of the joint. There are two different types of constrained knee implant designs, varying on the level of integration between the different components that build the implant, namely the presence or absence of a rotating hinge connecting the femoral component to the tibial one.
- (b) Unconstrained: In a contrasting manner, the comforting prospects of these implants are the fact that it maintains most of the original ligaments of the joint, allowing the patient a bigger range of motion compared to the constrained one while providing stability with minimal bone resection. These are, consequently, more indicated for patients with some of the ligaments still intact as well as a less severe form of joint degeneration and can be segmented into:
 - i. Posterior-Cruciate Substituting (PS)
 - ii. Posterior-Cruciate Retaining (CR)

2. Mobile Bearing:

The mobile bearing design is much newer than the fixed ones, with the first commercially available device introduced in 1976. It is intended to reduce aseptic loosening due to motion of the implant during the patient's lifetime as well as improve its degrees of motion - allowing certain "axial rotation" that naturally occurs during flexion of the knee joint. Although promising, its benefits, such as the capacity to compensate for natural "goniometric mismatches" that may occur between the femur and the tibia, have not yet been verified by research to the point where its use is recommended over fixed ones previously stated [13].

Following multiple iterations, modern-day knee prosthesis can be found in either unicondylar, also known as Partial Knee Replacement, or bicondylar models. Though there aren't enough studies to justify one over the other - with the choice made solely by the surgeon - Total Knee Resurfacing is still linked to better overall long-term survival despite the former preserving all of the joint's cruciate ligaments [14, 15].

With that in mind, a basic Total Condylar Knee implant model can be segmented into three different components according to the Knee anatomical structures, as exemplified in figure 1.6 [8].

1. Femoral Component:

The upper part of the implant is the best-defined item in terms of design at the moment. It follows an anatomical shape, covering the lower part of the Femur, namely the patellar surface region of the bone, and creates a new condyle shape that matches its proprietary design for the tibial component.

2. Patellar Component:

Initially made of metal, the development in the polymer field made it possible to resurface the bone to diminish wear debris around the articulation using polyethylene and a single, central "fixation lug".

3. Tibial Component:

The tibial component is also divided into two aspects: the plastic spacer, which provides a smooth surface capable of undergoing articulation movement with little to no release of unwanted particles, and the baseplate, which is designed with overhanging "lips" to maintain joint stability during what is known as "asymmetrical loading". Design-wise, the baseplate commonly has a stem for better fixation to the bone; cement pockets to increase the contact area to the bone cement and variable roughness.

1.4 Factors associated with total knee arthroplasty failures

With a survival rate up to 95% - thanks to the advance in both implant design and surgery techniques - TKA (total knee arthroplasty) has a considerable high rate of satisfaction among its

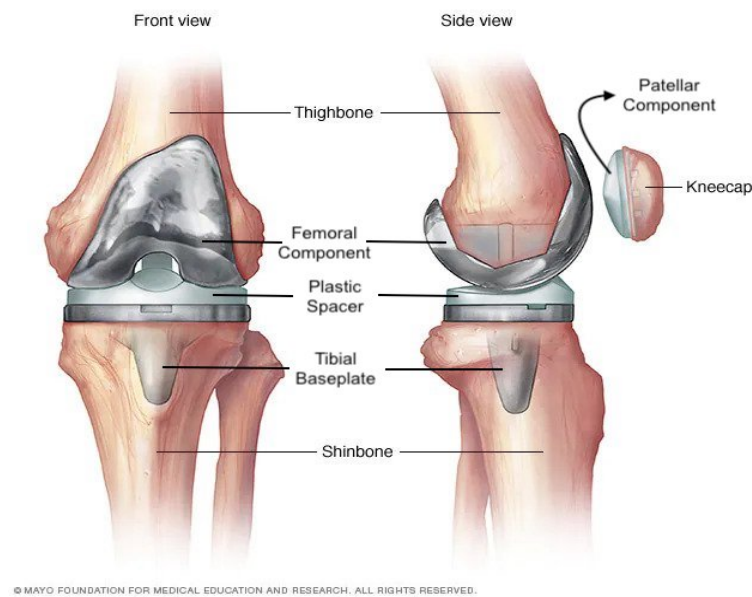


Figure 1.6: Basic Knee Implant Model Showing Anterior and Lateral view adapted from [16].

patients. Revision surgeries are still a challenge to both the medical community - with "not-so-defined" reasons and solutions - as well as to the patients themselves, that, as proven suffers from a lower "implant-survival" rate and quality of life.

Estimated to exceed 13 billion dollars by 2030, the American total economical cost related to TKA revisions has increased largely due to population growth, obesity, and a reduction in the age of the ultimate target patient. This burden can go up to approximately 13 million euros on the smaller Norwegian market. These values translate the relevancy of understanding the reasons and mechanisms of how the failure of the primary TKA happens [11, 17].

Various factors are associated with the failure of a TKA including instability of the knee joint, incorrect insertion of the implant, and aseptic loosening of the tibial component. These situations can be linked to surgical technique during the operative procedure which may cause variations of the cementing method or possible contamination of the interfaces with bone marrow fat [11].

Aseptic loosening - the 3rd largest reason for implant failure according to the Norwegian Arthroplasty Register - can be assessed through the analysis of the patient's x-ray via the presence of radiolucent areas indicating movement of the implant in question. Researches show that procedures with an average migration of 1mm in their first year have significantly higher chances of failure and are, consequently, prone to revision surgeries in the following 5 years [11, 18].

Many theories have been created over the year to explain the aforementioned effect on knee prosthesis, including high fluid pressure, particle disease, stress shielding, endotoxins, and, finally, micromotion as seen in figure 1.7. For the purposes of this project, the decision was made to concentrate on the stress shielding effect, which is heavily influenced by the implant's design and its relationship with the cementation conditions of the surgery (considering that bone marrow fat contamination of the tibial baseplate distal surface may be possible due to gaps in between the

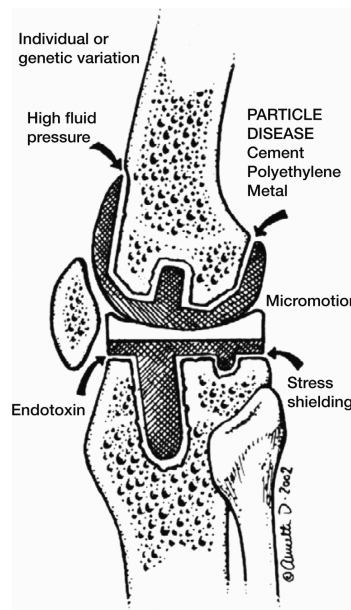


Figure 1.7: Illustration of the theorized reasons leading to aseptic loosening of a knee implant. Figure obtained from [19].

bone cement layer) as well as the resulting micromotion suffered by the baseplate, as bibliography sources relate [19].

Although the use of uncemented implants has increased in the past years, with the development of special implant coatings mechanisms such as the use of Hydroxyapatite (HAp) to induce osteointegration, their survival rate is still lower when compared to cemented ones. Still, the analysis of fixation technique vs implant success rate might clarify some of the gaps regarding the reasons for their failure. While pulse lavage of the implanted area prior to cementation has been shown to be one of the most important factors in avoiding aseptic loosening, it is theorized that migration of residual bone marrow fat to the implant(metal)-cement interface during early flexion of the joint may drastically reduce its bonding strength.

Besides that, it is important to dive into the relationship that different implant designs would have in balancing the aforementioned decrease in bonding strength. For the scope of this study, two implant designs - posterior stabilized, ligament retaining, and stemmed - will be investigated and their behavior when implanted in non-contaminated and fat-contaminated conditions, assessed through stability and adhesion tests.

Chapter 2

Aims of the study

The primary aim of this study is to investigate the fixation of the tibial component under different test conditions in an *in vitro* model, mainly associated with the implant design - non-coated vs pre-coated - and the effect of different environmental conditions on the cement interfaces - dry vs contaminated with fat.

The fixation quality will be assessed by stability testing - accounting for migration and micro-motion - and binding strength between the implant and bone cement.

We aim to compare:

- 1. Non-coated tibia component vs Pre-coated with poly(methyl methacrylate), a.k.a PMMA;**
- 2. The effect of lipid infiltration of the cement-implant interface.**

Finally, in accordance with the reference, and following the aims of this project there are three hypotheses:

H0. Null Hypothesis:

- (a.) Concerning the design, there will be no differences in stability or adhesion when comparing non-coated to pre-coated designs.
- (b.) In terms of fat contamination, the presence of fat will impact neither stability nor adhesion variables for both non-coated and pre-coated designs.

H1. Stability Related:

- (a.) Regarding the design effect, the PMMA pre-coated design will be more stable than the non-coated design, i.e, having smaller micromotion and migration values.
- (b.) For the contamination effect, the fat contamination will impact stability metrics.

H2. Adhesion Related:

- (a.) In terms of the design effect, the PMMA pre-coated design will have a significantly stronger bonding between the implant and the cement layer.

- (b.) Concerning the contaminating impact, the fat contamination will impact the implant's adhesion to bone cement.

A summary of the test and comparisons are can be seen in the scheme of figure 2.1:

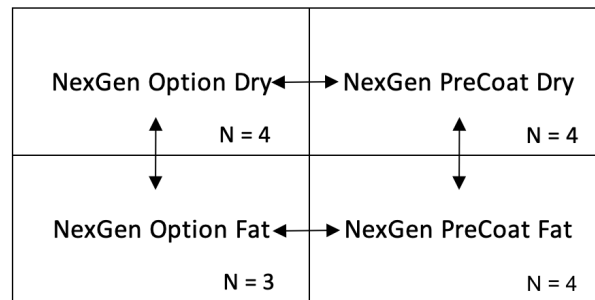


Figure 2.1: Scheme showing the study design with the relevant comparisons and the number of parallels for the different study groups.

Chapter 3

Materials

3.1 The Implants

NexGen® Implants account for approximately 50.3% of the total figures regarding Primary Total Prostheses during the period between 2014 and 2020 registered in Norway with 49.1% of those being from type NexGen Option(NexGen® Complete Knee Solution, Zimmer, Warsaw - Indiana, USA) and 37.2% having a PMMA coating on its surface (NexGen® Complete Knee Solution - Precoat, Zimmer, Warsaw - Indiana, USA). When implanted, both designs are coupled with a plastic articular surface of UHMWPE, with 14mm thickness in the case of this project.



Figure 3.1: Bottom View From a NexGen 3° Stem Tibial Plate (Option) Implant showing the Cement Pockets.



Figure 3.2: Bottom View From a NexGen 3° Stem Tibial Plate (PreCoat) Implant showing the Cement Pockets, the PMMA coating, and design changes.

Both designs allow, as previously mentioned, surgeries from type Cruciate-Retaining and present a triangular tibial stem of approximately 37.68 cm in length. Composition-wise, both models are made of 90% Titanium grade 5 standard alloys with 6% of Alumina and 4% of Vanadium (Tivanium® Ti-6Al-4V Alloy). The tibial baseplates present cement pockets on their tibial surface, allowing for a better cement adhesion as seen in figures 3.1 and 3.2. [20, 21].

These implants are currently available in 10 different sizes. A size 4 baseplate was chosen since it best matched the size of the SawBone model.

3.2 The Bone

To simulate the in vivo tibial structure while minimizing the variability seen in cadaver studies, bone replicates were used. The models from Sawbone® (Pacific Research Laboratories Inc., Vashon, WA) were made with rigid foams with 12.5 pounds per cubic foot (PCF) - for the cancellous bone simulation - and 40 PCF - for the cortical portion of the bone (product ref: 1522-912).

Both materials are composed of Polyurethane foam and the latter meets the ASTM F1839-08 material specification, more detailed description is given by the manufacturer and can be seen in tables 3.1 and 3.2.

Density			Cell		Compressive	Compressive
(PCF)	(g/cc)	Volume Fraction	Size (mm)	Count cells/cm	Strength (MPa)	Modulus (MPa)
12.5	0.20	0.17	0.5-1-5	6-8	3.9	47.5
Foam Block Tolerances:						
Foam Density				+/- 10%		
Block Dimensions				+/- 2mm		

Table 3.1: Physical Characteristics of the Cellular Foam according to the manufacturer's catalog.

Density			Compressive		Tensile		Shear	
(PCF)	(g/cc)	Volume Fraction	Strength (MPa)	Modulus (MPa)	Strength (MPa)	Modulus (MPa)	Strength (MPa)	Modulus (MPa)
ASTM D1622			ASTM D1621		ASTM D638		ASTM C273	
40	0.64	0.54	31	759	19	1000	11	130
Solid Foam Block Tolerances								
Foam Density				+/- 10%				
Block Dimensions				+/- 2mm				

Table 3.2: Physical Characteristics of the Solid Foam according to the manufacturer's catalog.

The model has a quadrangular base with a width of approximately 5.7 cm, a length of 7.6 cm, and a height of 7.6 cm. On the proximal end, the model is shaped in such a way as to match the resected surface of the tibia, ready for Implant preparation, and with dimensions of 46.2 x 75.3 mm.



Figure 3.3: Sawbone model prior to the preparation.

A summary of all implants and bone model weight can be seen in table B.1 contained in appendix B and table D.1 from appendix D, respectively.

3.3 Bone Preparation Instruments

To assert the repeatability of implant alignment, a plastic guide containing the nail's positioning holes seen in figure 3.5 was produced using 3D modeling. This guide was designed to be used before the metallic one and to provide enough cortical support on all three corners of the implant.

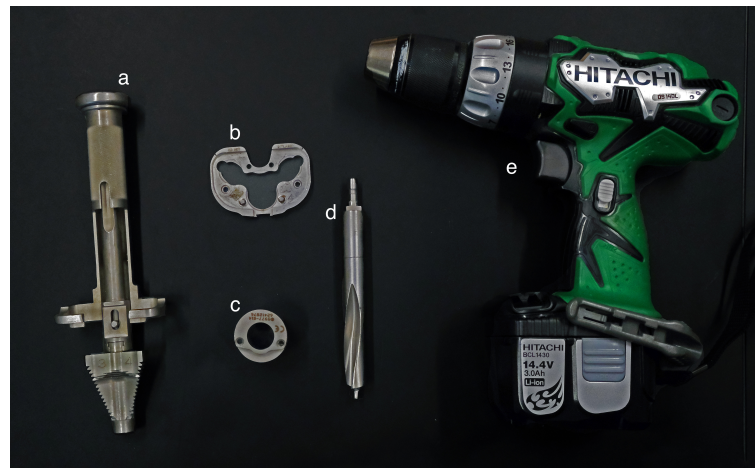


Figure 3.4: Instruments used for bone preparation: (a) Bone Impactor used to make space for the tibial implant stem, (b) The femoral finishing guide - fixed on the bone with metallic nails and used to drill the necessary holes on the bone on the correct positions, (c) Guide for drilling the stem hole in the correct angle, used in association with (b), (d) The drilling bur with approximately 16.7 mm of diameter, (e) Hitachi DS14DL Cordless Drill.

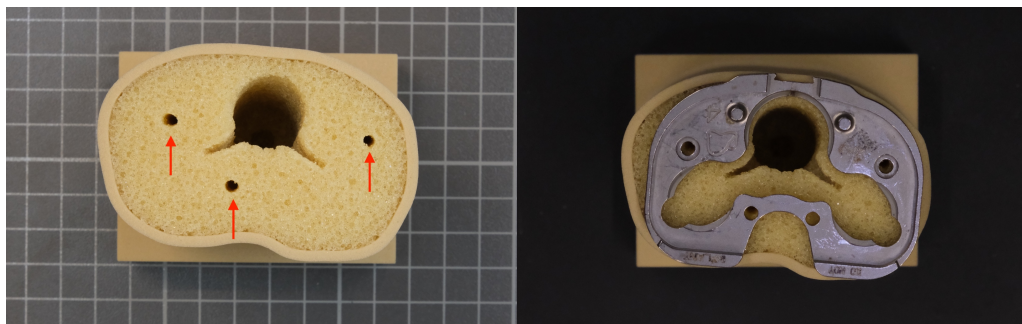


Figure 3.5: Bone Model surface showing the preparation patterns on the left side and the tibial sizing guide on the right side.

3.4 The Cement

Table E.1 contained in appendix E shows the amount of cement used in each test performed at the lab in grams.

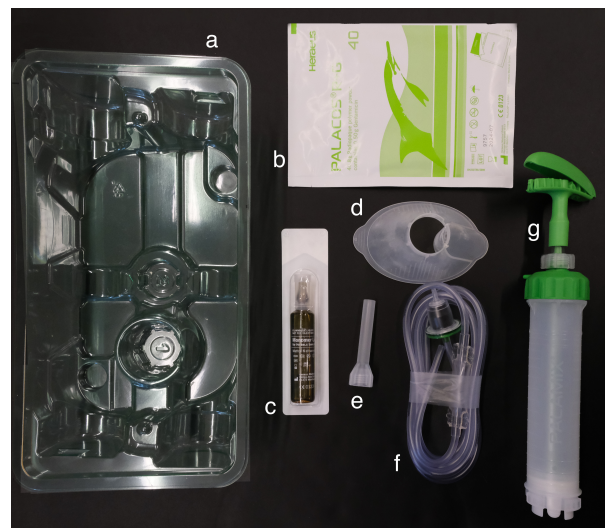


Figure 3.6: Instruments used for cement preparation: (a) Plastic Container for the other instruments, (b) Palacos® R+G Polymer Powder, (c) Palacos® R+G Monomer Liquid, (d) Funil used to mix the powder to the liquid components of the cement, (e) Plastic Tip used to apply the cement, (f) Vacuum Tube, (g) Palacos® Mixing System.

<i>Cement Powder Composition</i>	
<i>Component</i>	<i>Percentage</i>
<i>Poly(methyl acrylate, methyl methacrylate)</i>	81.9%
<i>Zirconium Dioxide</i>	15.0%
<i>Hydrous Benzoyl Peroxide</i>	1.0%
<i>Gentamicin Sulphate</i>	2.1%
<i>Monomer Liquid Composition</i>	
<i>Component</i>	<i>Percentage</i>
<i>Methyl Methacrylate</i>	98.0%
<i>N, N – Dimethyl – p – Toluidine</i>	2.0%

Table 3.3: Chemical Composition of the Palacos® R+G (Heraeus Medical GMBH, Germany) bone cement.

3.5 The Transducers

For the movement detection, three linear variable displacement transducers (LVDTs) from Solartron Metrology® (SOLARTRON Metrology Limited, Leicester - Leicestershire, United Kingdom) with a resolution of $1\mu\text{m}$ were used connected to a PC-based Analog-to-Digital converter Fylde micromANALOG 2(Fylde Electronic Laboratories, Fylde Road, Preston, Lancashire, UK) configured as to output the movement information in millimeters given the calibration results for each transducer.

Chapter 4

Methodology

The project's methodology can be separated into three phases such as bone preparation, cementation, stability testing, adhesion testing, and data analysis as summarized in the scheme from figure 4.8. For all the following sections, the results of the first comparison group (Option Dry) were previously acquired by another student in the context of another project in the same laboratory [22].

4.1 Bone preparation:

For bone preparation, surgical equipment matching the implant design is used. Initially, a self-made plastic guide with the same dimensions as the resurfaced proximal tibia is placed on the SawBone tibia surface by hammering metallic nails into its pre-designed holes, ensuring enough cortical support on the anterior side of the implant and both posterior condyles. This measure was done to decrease the variability in position. The official metallic guide is fixated on the SawBone® following the holes made previously with the plastic guide. This step provides a consistent positioning for the drilling of the stem hole (instruments (b) and (c) from figure 3.4). The drilling itself is carried out by ensuring that the bur (instrument (d) of figure 3.4) is inserted into the bone model until the predefined line. Finally, the bone impactor (instrument (a) from figure 3.4) is forced upon the drilled hole by hammering, creating enough space in the bone to host the implant stem and its wings. The plastic spacer is inserted after preparation. The masses of both the implant and the bone model are measured before and after cementation.

4.2 The cementation process:

For the cementation, full cementation (FC) with cement also into the stem hole approach was used. The bone cement used is from Palacos® R+G - the most common type of bone cement used in Norway - and its preparation follows the company's instruction with the help of a Palamix® vacuum mixing [11].

The procedure used to cement the implant onto the bone model is the same as used in the hospital in which this project is being done and follows the manufacturer's suggestion for these types of implants. A guideline for the methodology can be seen in table 4.1 with the time points given from the initial point of mixing the self-curing monomer and its liquid. Brochures regarding the manufacturer's instructions are included in figures A.1 and A.2 from Appendix A.

Its application is made with a Palamix® cement gun after a total of 2 min of curing first on the tibial surface of the implant and then into the surface of the tibial Sawbones.

The implant is then inserted into the bone model and partially fixated with a hammer immediately after cement application, any excess cement is removed quickly after assembly using a spatula.

A compressive force of 50N is applied to the specimen to remove additional excess cement until the bone cement has partially cured in a step equivalent to the "extension under compression" procedure performed in the operating theater, which takes approximately 13 minutes. The whole model is finally placed in a heating cabinet at physiological temperature (37°C) for 30 minutes until the total time has been reached.

4.2.1 Fat contamination:

The described methodology is common for both implant designs with an alteration for the situation of Fat Contamination. In this case, 120 μL of bone marrow fat collected from calf bone and heated to 70°C is applied on the far edges of the implant's tibial surface with the use of a pipette following the design of the cement pockets before the cementation itself. For a better understanding, please refer to figure 4.1 and 4.2.

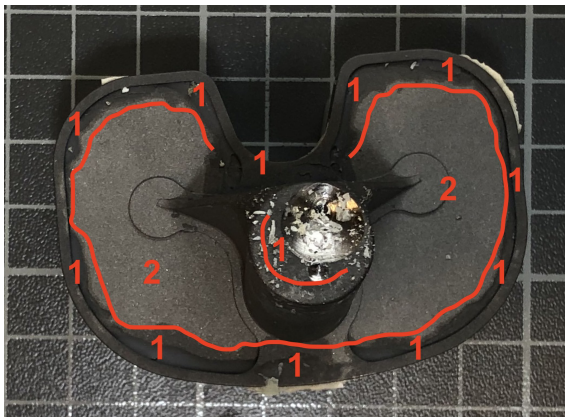


Figure 4.1: Figure showing the spreading of bone marrow fat on both the distal face of the non-coated tibial implant as well as at the stem, as highlighted in red. Areas with the number 1 indicated the contaminated area and number 2, the dry portion of the implant.

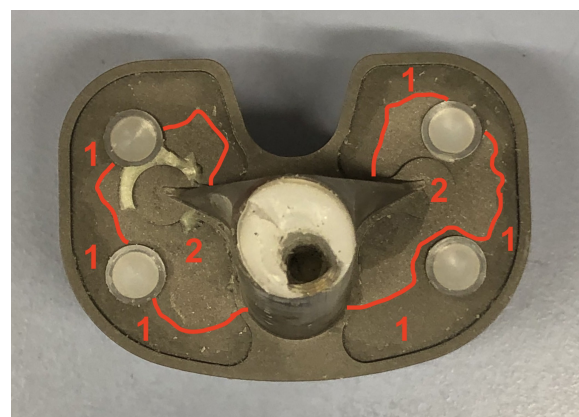


Figure 4.2: Figure showing the spreading of bone marrow fat on both the distal face of the pre-coated tibial implant as well as at the stem, as highlighted in red. Areas with the number 1 indicated the contaminated area and number 2, the dry portion of the implant.

Point in time (min:sec)	Duration	Action
00:00	00:10	Combining the liquid and powder phase
00:10	00:10	Having the liquid under vacuum without mixing
00:20	00:30	Mixing cement under Vacuum
00:50	01:10	Waiting for the cement to reach the sticky-phase and preparation of the cement gun
02:00	02:00	Cementing and Impacting the implant on the Bone Model
04:00	09:00	Applying a constant pressure of 50N on the Instron® System
13:00	30:00	Placing the specimen in the heating cabinet at 37°C

Table 4.1: Implant cementation guideline with specific time points

4.3 Stability testing:



Figure 4.3: Stability test ongoing. The figure shows LVDT's 3 and 4.

Following twenty-four hours from the implant's cementation, the model is put under cyclic load using a servo-hydraulic mechanical testing machine (MTS Model 858 Mini Bionix II®) running a program that simulates a person walking through twenty thousand (20.000) cycles during 6:18 hours. This simulation, seen in figure H.1 from Appendix H is done through a script for a "Level Walking" modeled by I. Kutzner et al and it was run by the MTS FlexTest (TM) 60 Station Manager software version 6.2B 712 in conjunction with Madaq (Fylde Electronic) software. [23].

As test preparation, three custom-made metallic LVDT holders are placed around the implant's lower edge, approximately 120° apart, for measuring the anterior (3), posterior lateral (4), and posterior medial (5) positions of the implant, for simplicity case, these will be named anterior, lateral and medial, respectively. A plateau design is then placed on the SawBone® itself, approximately 8mm under the lower edge of the base-plate, following the same distribution of the holders as shown in figure 4.4. Positioning of the transducers was chosen to measure axial displacement along the long

femoral axis of the joint since this is the direction responsible for recording larger movements in traditional knee implant stability studies [24].

The placement of the load is made following the bibliography's advice, in a slightly medial distribution as seen in the scheme of figure 4.5 and serves as a 'preconditioning' step for the future adhesion test.

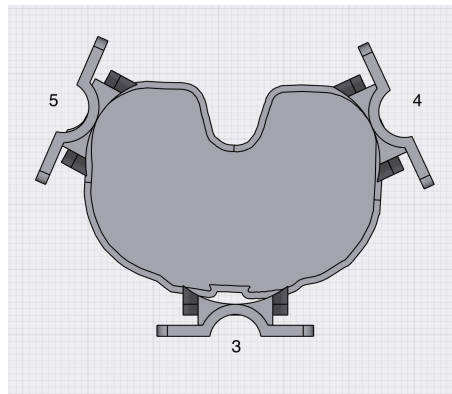


Figure 4.4: LVDT Holders positioning with numeric identification, (3) anterior, (4) lateral and (5) medial.

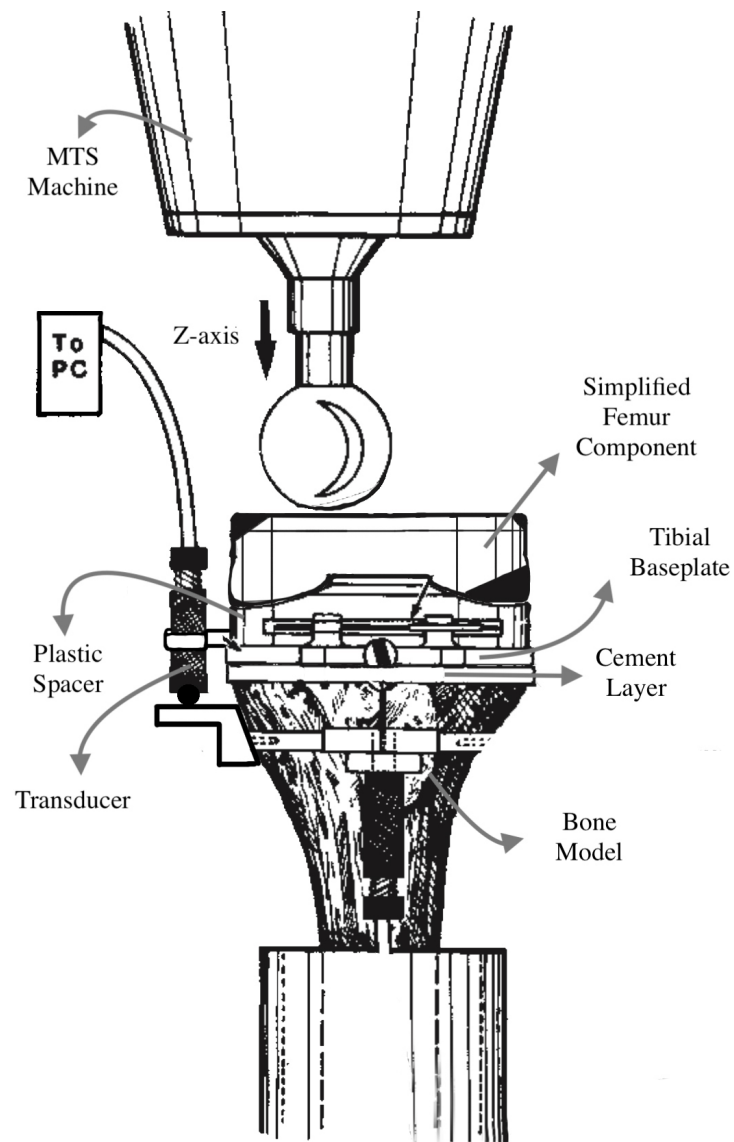


Figure 4.5: Scheme showing the components and their positioning during the stability test. Attention to the slightly medial loading of the implant. Image adapted from [25].

4.4 Adhesion testing:

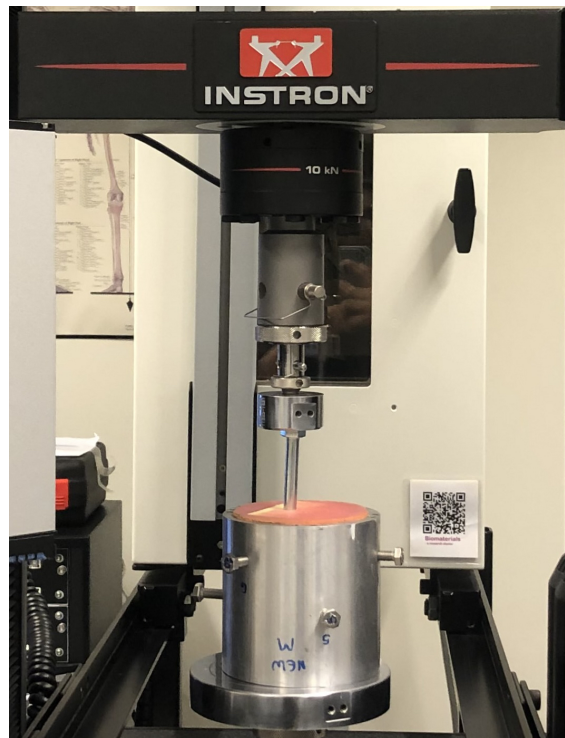


Figure 4.6: Push-out test being done in one of the samples embedded in resin.

Axial push-out strengths were measured using a 10.000 N uniaxial load cell on an Instron 5966 universal testing machine operated by an Instron digital controller and Bluehill 3 (version 3.76) software. Using a vertical drill and a leveling tool, a hole was drilled on the stiff base of the tibia SawBones up to about 3.5 cm below the implant baseplate, attempting to obtain the lower tip of the implant's stem. The hole has a dimension big enough to fit the rod used to push out the implant itself, as seen in figure 4.6. The construct is placed on a 11.7 (height) x 9.96 (diameter) cm metallic cylinder and secured on four small columns. The model and the push-out rod were positioned to be aligned in the direction of the axial push-out force. The tests used a standard configuration for push-out tests, with 0.05 mm/s displacement control. The maximum load endured before failure and their respective failure model were recorded for each sample. The types of failure were segmented into 4 groups - Survival, Bone Failure, Implant Debonding, and failure between the bone and cement.

4.4.1 Polymer embedding:

A part of the final samples was fully embedded in self-polymerization denture resin (Meliodent Heat Cure - Rapid Repair®, Kulzer - Mitsui Chemicals Group, France). The model was placed on a 11.7 (height) x 9.96 (diameter) cm cylinder and secured by 6 screws ensuring a central and level positioning. Approximately 450ml of the resin was prepared and poured into the metal. This was

a method of strengthening the model and preventing fractures of the SawBone® material during the push-out tests. A regular push-out test was performed past 24 hours from the embedding.

4.5 Data analysis

The raw data output by a DAC box (Digital to Analog Converter) Fylde micromANALOG 2(Fylde Electronic Laboratories, Fylde Road, Preston, Lancashire, UK) has a specific template previously defined with 5 columns containing the time-stamps - in seconds - the force returned by the MTS machine - in kN - and the three LVDT data - in millimeters - recorded in a frequency of 100 samples/seconds.

This data is processed using a Matlab® environment to extract the motion curves, the maximum implant migration as well as implant's maximum micromotion points. For the full MatLab code see Appendix C figures C.1 to C.3.

For the zeroing of the LVDTs, the hundredth sample is taken as a point in which the cyclical load hasn't started yet, it is important to note, however, that due to limitations of the MTS the samples are always subjected to a load, usually of 100N, while the data is being recorded.

Based on a previous recent paper about the influence of fat contamination on pull-out strength of several others implant designs, the groups were defined to have at least 3 parallels for all test combinations [26]. Using GraphPad Prism v9.4.1 (GraphPad Software, San Diego, CA), the data was initially analyzed using two-way ANOVA to assess for an impact of implant type, presence of bone marrow fat, and any crossed effect on push-out loads. Tukey's multiple comparisons test was used as post-hoc test to compare all four test groups. Tests with $p \leq 0.05$ were considered statistically different

4.5.1 Migration and micromotion variables:

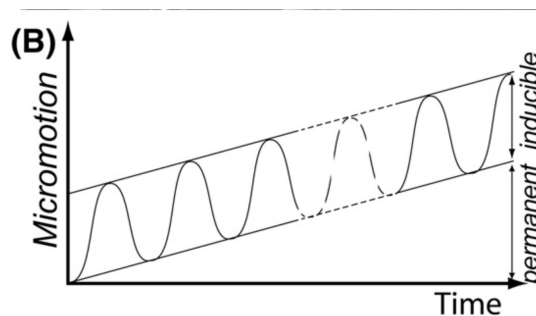


Figure 4.7: Scheme of the two types of movement analyzed taken from [24]

Under the cyclical load, the model presents mainly two different types of movements: inducible or reversible displacement referred to as **micromotion**, and permanent displacement, called **migration**. A scheme of the movements can be seen in figure 4.7.

For permanent displacement, the axial position of the model at its unloaded points was stored and analyzed. Migration results were further segmented into values of subsidence and lift-off to better indicate the trajectory of movement suffered by the tibial baseplate. The location with the highest value throughout the whole test was used as an indicator for maximum subsidence, while for maximum lift-Off, the position with the highest negative value was chosen. It is important to mention that positive values of movement indicate downward movement along the Z axis of figure 4.5, also known as subsidence.

Inducible displacement, on the other hand, denotes the total stability of the construct under the load and, when in vivo is strongly correlated with osteointegration. According to references, micromotion is caused primarily by changes in primary implant stability - predominantly by the difficulty of creating a perfect match between the implant and the bone's "host site" which in turn translates into micro gaps between the implant and the cement layer - although also impacted by the variances in the bone elastic modulus and the implant's material [27]. For this project, micromotion - as of the end of the 20.000th cycle - is presented at its maximum position.

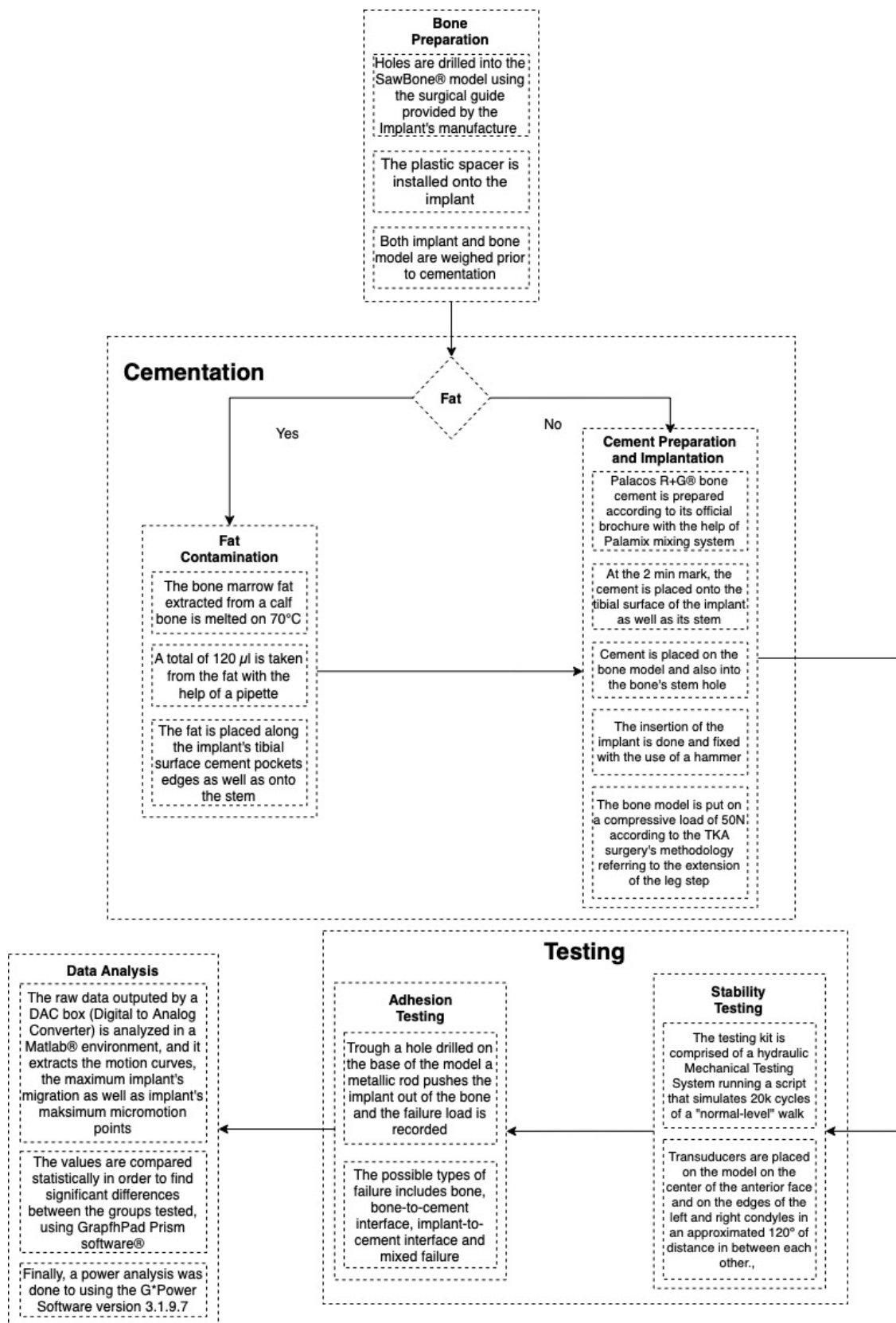


Figure 4.8: Flowchart showing the steps for a full test, (1) Bone preparation; (2) Cementation of the implant; (3) Stability and adhesion tests; (4) Data analysis with statistical comparisons.

Chapter 5

Results

5.1 Cement thickness

After implantation, the thickness of cement was measured under anterior, lateral, and posterior positions for every sample. Individual results are shown in table 5.1.

<i>Cement Thickness on all positions</i>				<i>Cement Thickness on all positions</i>			
	<i>Anterior (mm)</i>	<i>Lateral (mm)</i>	<i>Medial (mm)</i>		<i>Anterior (mm)</i>	<i>Lateral (mm)</i>	<i>Medial (mm)</i>
<i>Non-Coated dry</i>				<i>Pre-Coated dry</i>			
13	2.38	1.57	0.85	01	0.85	1.50	1.12
19	2.61	1.22	1.05	02	1.13	0.65	1.83
20	2.19	1.05	0.91	03	1.40	1.03	1.83
21	2.93	1.06	1.34	07	5.12	3.70	3.66
<i>Mean :</i>	2.53	1.23	1.04	<i>Mean :</i>	2.13	1.72	2.11
<i>Total Mean :</i>		1.60		<i>Total Mean :</i>		1.98	
<i>Non-Coated contaminated</i>				<i>Pre-Coated contaminated</i>			
30	1.91	1.20	1.31	04	1.67	1.97	1.85
31	1.18	1.42	1.39	05	1.73	1.90	1.65
32	2.74	1.38	1.18	06	3.56	3.01	3.16
<i>Mean :</i>	1.94	1.33	1.29	08	3.25	2.75	3.03
<i>Total Mean :</i>		1.52		<i>Mean :</i>	2.55	2.41	2.42
				<i>Total Mean :</i>		2.46	

Table 5.1: Cement Thickness recorded on all test samples presented in the three different positions.

Out of the four comparisons group draw - non-coated dry (1) / contaminated (2) and pre-coated dry(3) / contaminated (4) - the first three presented relatively similar total mean thickness, ranging from 1.52mm to 1.98mm. Non-coated (design 1) and dry samples had a higher difference between all positions, with more than 2.0x the amount of cement on the anterior side when compared to the lateral or medial. The pre-coated contaminated group had the highest total thickness mean when compared to others, although their distribution between the 3 positions does not show large

differences. Significant differences were found between the groups (ANOVA $p = 0.0496$) yet Tukey's did not detect any multiple comparison difference.

After measuring the sagittal length of the implant - constant for both designs - the angle of implantation was calculated based on the following equations. This information is important for future discussions about the alignment of the baseplate. Samples with implantation angles bigger than 90° are classified as having an anterior tibial slope, otherwise, with a posterior tibial slope.

$$L = 44\text{mm} \quad (5.1)$$

$$Ts = \arccos\left(\frac{\text{Anterior Height} - \text{Posterior Height}}{L}\right) \quad (5.2)$$

Measured sagittal length of the implants.

Equation for implantation angle.

Implantation Angle on all samples			Implantation Angle on all samples		
	Angle($^\circ$)	Tibial Slope Classification		Angle($^\circ$)	Tibial Slope Classification
<i>Non-Coated dry</i>			<i>Pre-Coated dry</i>		
13	88.5	posterior	01	90.6	anterior
19	88.1	posterior	02	90.1	anterior
20	88.4	posterior	03	90.0	neutral
21	87.8	posterior	07	88.1	posterior
Mean :	88.2	posterior	Mean :	89.7	posterior
<i>Non-Coated contaminated</i>			<i>Pre-Coated contaminated</i>		
30	89.1	posterior	04	90.3	anterior
31	90.3	anterior	05	90.1	anterior
32	88.1	posterior	06	89.4	posterior
Mean :	89.2	posterior	08	89.5	posterior
			Mean :	89.8	posterior

Table 5.2: Implantation angle calculated for each sample.

In nine of the fifteen studies, the posterior tibial slope was present, predominantly in the non-coated groups. The bulk of the implantation in pre-coated samples was on an anterior slope, yet the mean implantation angle still supports a posterior placing. It is important to emphasize that the variation from neutral implantation ($= 90^\circ$), while present, is minor, with the largest variance equaling 2.2° in sample 21.

5.2 Stability Tests

No loosening or debonding of the tibial baseplate was observed during the stability tests.

5.2.1 Migration

An overview of all results is seen in table 5.3. It is important to highlight the sign convention chosen in chapter 4, with positive values indicating downwards movement on the Z axis or subsidence of the tibial baseplate ($\downarrow$).

<i>Maximum Subsidence</i>				<i>Maximum Lift – Off</i>				<i>Maximum Subsidence</i>				<i>Maximum Lift – Off</i>			
		<i>Value (μm)</i>	<i>Position</i>			<i>Value (μm)</i>	<i>Position</i>			<i>Value (μm)</i>	<i>Position</i>			<i>Value (μm)</i>	<i>Position</i>
<i>Non-Coated dry</i>								<i>Pre-Coated dry</i>							
	13	6.73	Lateral		-0.35	Anterior			01	7.11	Lateral		-0.06	Medial	
	19	3.13	Medial		-1.01	Anterior			02	2.29	Lateral		-8.40	Lateral	
	20	1.25	Medial		-2.86	Lateral			03	0.79	Medial		-4.13	Lateral	
	21	1.56	Lateral		-1.87	Anterior			07	6.13	Medial		-0.81	Anterior	
<i>Non-Coated contaminated</i>								<i>Pre-Coated contaminated</i>							
	30	3.80	Medial		-0.83	Lateral			04	16.39	Medial		-1.81	Anterior	
	31	4.16	Anterior		-0.54	Lateral			05	1.84	Medial		-2.75	Lateral	
	32	2.79	Medial		-6.83	Lateral			06	19.06	Medial		-4.07	Anterior	
									08	4.56	Medial		-0.68	Anterior	

Table 5.3: Summary of Migration Results for all samples (subsidence and lift-off).

5.2.1.1 Maximum Subsidence

Since all samples displayed subsidence behavior at least once throughout the test, their highest subsidence values were considered. The mean maximum subsidence of both designs either with or without bone marrow fat is displayed in figure 5.1, individual numbers are listed in table 5.3. For spatial distribution analysis, 11 out of 15 cases had their maximum subsidence medially, 3 laterally and 1 anteriorly.

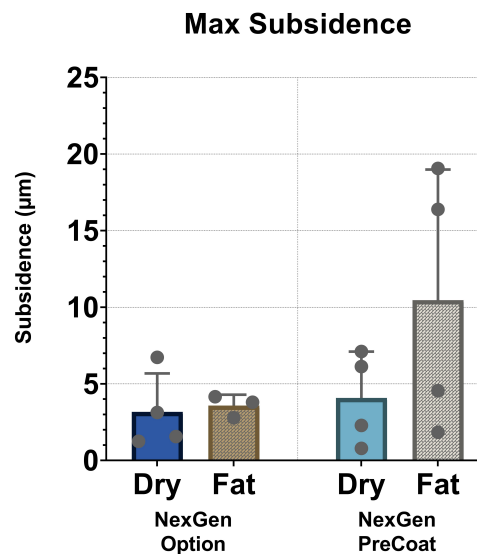


Figure 5.1: Bar graph of the maximum subsidence throughout the whole test for all groups studied represented in mean and standard deviation.

1. The design effect: The two-way ANOVA test revealed that there was no evidence of the contribution of the design factor to maximum subsidence values ($p = 0.1559$).
 - (i). Dry conditions: Tukey's multiple comparisons demonstrated that the PMMA pre-coated design did not subside significantly more at its maximum (mean non-coated

dry: 3.17 μm and mean pre-coated dry: 4.08 μm) than the non-coated version of the implant ($p > 0.99$).

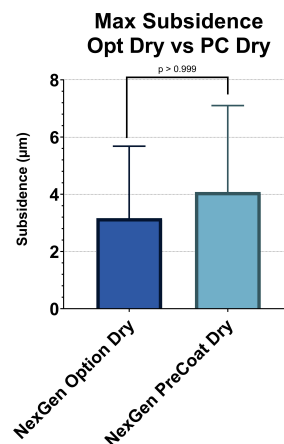


Figure 5.2: Pair-wise comparisons of max. subsidence between the non-coated (Option) and pre-coated designs (PreCoat) in dry conditions represented in mean and standard deviation.

2. The contamination effect: Similarly, there was no indication of the contamination element influencing maximum subsidence values ($p = 0.2107$).
 - (i). Non-coated, dry vs fat: The mean for the non-coated dry group (3.17 μm) and the mean for the non-coated contaminated one (3.58 μm) was very similar. No statistical difference was found ($p > 0.99$).

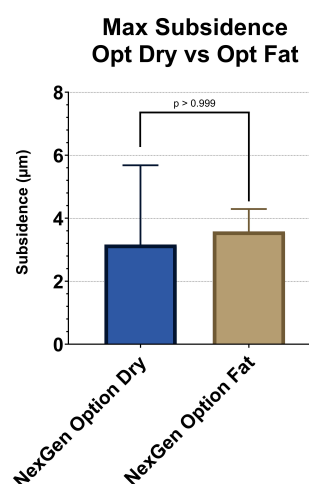


Figure 5.3: Pair-wise comparisons of max. subsidence between the non-coated (Option) implanted in dry and fat contaminated conditions (Option) expressed in means and standard deviations.

- (ii). Pre-coated, dry vs fat: The fat seems to have greatly enlarged the average values - approximately by two times - when implanted in contaminated environments (mean

pre-coated dry: 4.08 μm and mean pre-coated fat: 10.46 μm), yet Tukey's multiple comparisons indicated no difference ($p = 0.3080$).

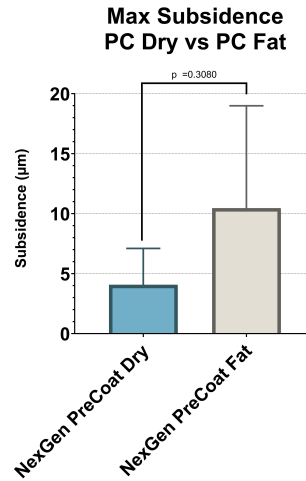


Figure 5.4: Pair-wise comparisons of max. subsidence between the pre-coated (PreCoat) implanted in dry and fat contaminated conditions (PreCoat), mean and standard deviation.

- (iii). Non-coated (fat) vs Pre-coated (fat): The coating appeared to have enhanced the influence of fat contamination on the point of the sample's greatest subsidence (mean non-coated fat: 3.58 μm and mean pre-coated fat: 10.46 μm). Again, no significant difference was noticed ($p = 0.3096$).

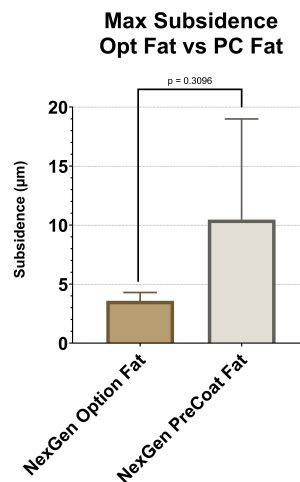


Figure 5.5: Pair-wise comparisons of max. subsidence between the non-coated (Option) and pre-coated designs (PreCoat) in fat-contaminated conditions.

3. Crossed-Effect: No sign of interaction between design and implant conditions on maximum subsidence was noted either ($p = 0.2680$).

5.2.1.2 Cement thickness correlation

When comparing the cement thickness of the point with the maximum subsidence for each sample, there is a tendency for a positive correlation ($r = 0.4991$ and $p = 0.0602$), as seen in figure 5.6.

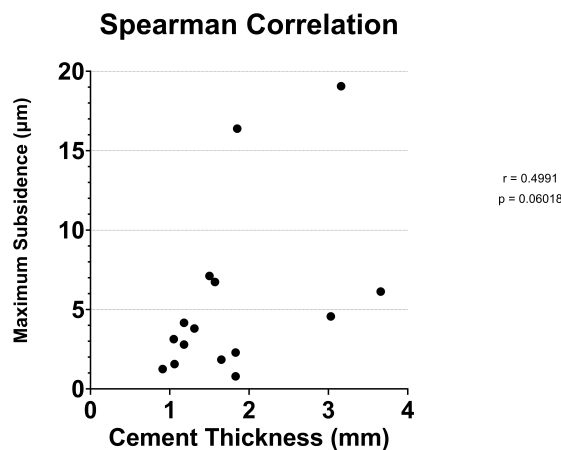


Figure 5.6: Graph of the Spearman correlation between cement thickness and maximum subsidence for individual samples.

5.2.1.3 Maximum lift-off

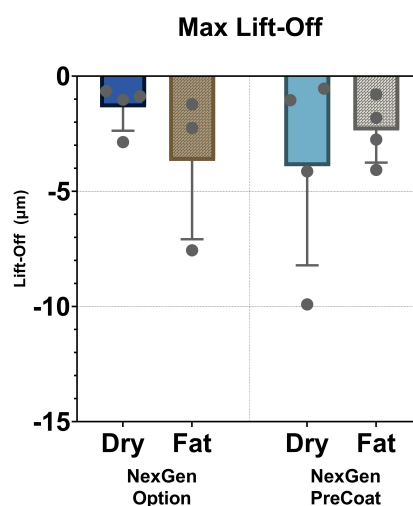


Figure 5.7: Bar graph of the maximum lift-off throughout the whole test for all groups studied represented, means and standard deviation.

Maximum lift-off results occurred mostly anteriorly and laterally - with 7 out of 15 cases for both positions - and only once medially. Neither the contamination factor nor the design alteration had significant impacts in the lift-off values ($p = 0.7965$ and $p = 0.6829$). The cross interaction of these factors also did not significantly impact the final results ($p = 0.2141$).

5.2.2 Micromotion

Micromotion Results			Micromotion Results		
	Maximum (μm)	Sum of all Values (μm)		Maximum (μm)	Sum of all Values (μm)
<i>Non-Coated dry</i>			<i>Pre-Coated dry</i>		
13	⁽⁴⁾ 24.20	47.00	01	⁽⁴⁾ 16.80	35.30
19	⁽⁴⁾ 20.20	41.30	02	⁽⁴⁾ 22.80	44.80
20	⁽⁴⁾ 17.30	33.20	03	⁽⁴⁾ 13.10	26.70
21	⁽⁴⁾ 25.50	45.40	07	⁽⁵⁾ 29.30	52.00
<i>Non-Coated contaminated</i>			<i>Pre-Coated contaminated</i>		
30	⁽⁴⁾ 20.90	46.00	04	⁽⁴⁾ 27.40	49.67
31	⁽⁵⁾ 18.10	35.70	05	⁽⁵⁾ 15.40	27.80
32	⁽⁴⁾ 22.80	41.20	06	⁽⁵⁾ 29.80	48.45
			08	⁽⁵⁾ 17.20	32.32

Table 5.4: Summary of micromotion results for all samples.

⁽ⁿ⁾Indicates the positioning in which the highest micromotion occurred, $n = 4$ indicating the lateral side, $n = 5$, medial.

Similarly to previous metrics, all the groups had comparable micromotion values and no differences were found between them ($p > 0.90$ for all Tukey's multiple comparisons). There was no indication of variation caused by the contamination factor or by the changes in the design ($p = 0.9021$ and $p = 0.9281$) nor by the cross-effect of these factors ($p = 0.6075$). Grouped results can be seen in figure 5.8 and individual ones in table 5.4. Position-wise, the highest micromotion was on the lateral side for 10 out of 15 cases and on the medial side for 5 cases.

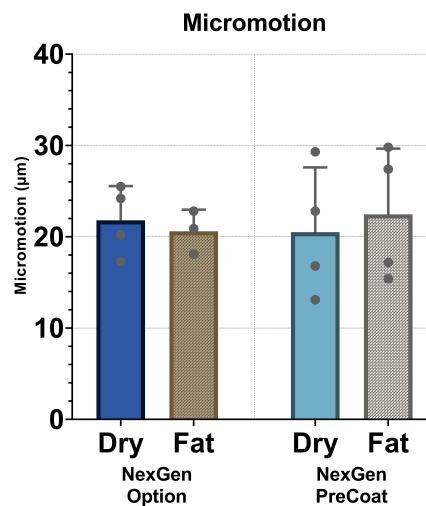


Figure 5.8: Bar graph of representative micromotion values at the end of the 20,000th cycle for all groups studied represented in means and standard deviation.

5.3 Adhesion Tests

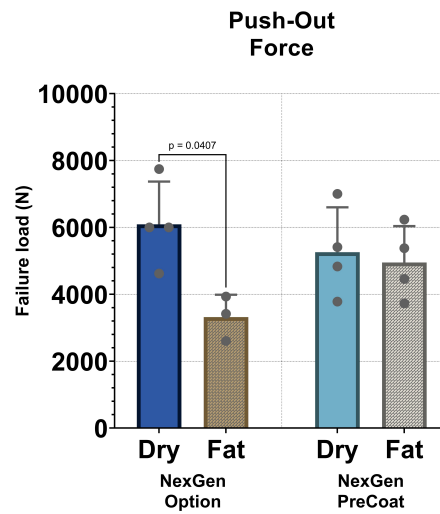


Figure 5.9: Bar graph of push-out forces for all groups represented in means and standard deviations.

Individual results and descriptions of the type of failure are shown in table 5.5 and failure modes are exemplified in figures 5.14 to 5.19.

<i>Push – Out Results</i>			<i>Push – Out Results</i>		
	<i>Force(N)</i>	<i>Type of failure</i>		<i>Force(N)</i>	<i>Type of failure</i>
<i>Non-Coated dry</i>			<i>Pre-Coated dry</i>		
13	6000.0	Survival	01	4832.2	Bone Failure
19	6000.0	Survival	02	5410.0	Bone-Cement Failure
20	7743.3	Survival	03	7000.0	Survival
21	4620.1	Bone Failure	07	3781.8	Bone-Cement Failure
<i>Non-Coated contaminated</i>			<i>Pre-Coated contaminated</i>		
30	2602.5	Implant Debonding	04	3727.6	Bone Failure
31	3414.6	Implant Debonding	05	6230.9	Bone Failure
32	3932.9	Bone Failure	06	4457.0	Bone Failure
			08	5378.5	Survival

Table 5.5: Raw Push-Out data for all samples and their failure modes.

Only the contamination factor proved to be significant on the impact of push-out forces ($p = 0.0268$). The same cannot be said for solely the design alterations ($p = 0.5228$) nor for the cross effect ($p = 0.0657$).

5.3.1 Dry conditions

Non-coated samples had a mean push-out force of 6091 (± 1279) N vs 5256 (± 1344) N for the pre-coated ones. For the first group, 3 out of the 4 samples did not present implant debonding

surviving intact while reaching the maximum load limitation, and 1 suffered from a failure on the bone itself. For the pre-coated design, 2 of the 4 samples suffered from failure on the interface between bone and cement, 1 had a bone failure and 1 survived intact the whole test.

1. The design effect:

- (i.) Dry conditions: Tukey's multiple comparison test indicated no statistical differences between the non-coated group and the pre-coated when in ideal, dry conditions ($p = 0.7425$).

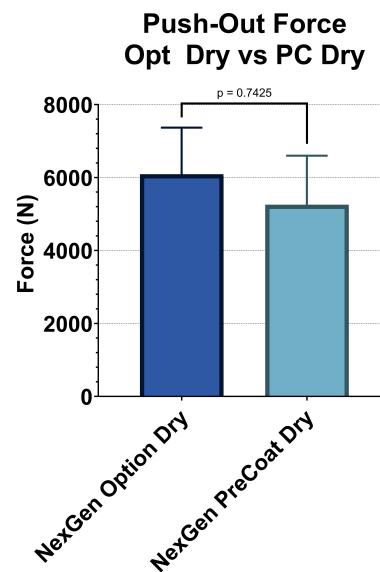


Figure 5.10: Pair-wise comparisons of push-out forces between the non-coated (Option) and the pre-coated design (PreCoat) in dry conditions.

5.3.2 Contaminated conditions

Push-out results show a mean force of 3317 (± 670) N for the non-coated group, with 2 of the 3 samples suffering from debonding of the implant from the bone-cement layer and 1 bone failure. For the second design, a mean force of 5475 (± 1267) N was obtained with 3 of the 4 samples suffering from bone failure and 1 reaching the setup limitations without suffering from any kind of failure.

2. The contamination effect:

- (i.) Non-coated, dry vs fat and Pre-coated, dry vs fat: The presence of fat significantly decreases the bonding strength for the non-coated design ($p = 0.0407$) while not affecting the pre-coated one ($p = 0.1859$).

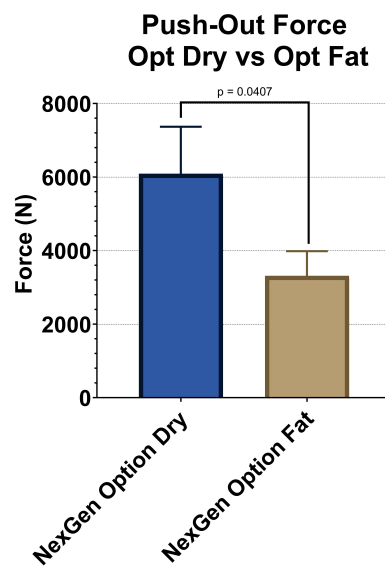


Figure 5.11: Pair-wise comparisons of push-out forces between the non-coated (Option) design in both dry and fat contaminated conditions.

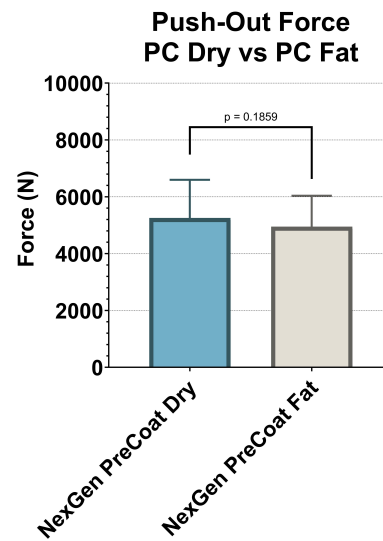


Figure 5.12: Pair-wise comparisons of push-out forces between the pre-coated (PreCoat) design in both dry and fat-contaminated conditions.

- (ii.) Non-coated (fat) vs Pre-coated (fat): Although the mean failure load recorded for the pre-coated group is approximately 1.65x bigger than the non-coated, the post-hoc Tukey's test did not indicate a significant change between them ($p = 0.3057$).

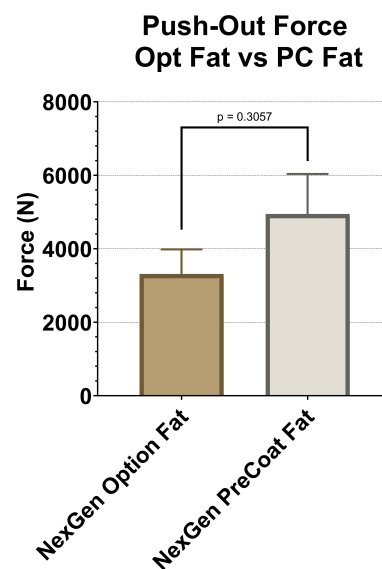


Figure 5.13: Pair-wise comparisons of push-out forces between the non-coated (Option) and the pre-coated design (PreCoat) in fat-contaminated conditions.



Figure 5.14: Top view from a failure between the bone and the bone-cement with the crack starting at the cement interface and spreading to the bone (sample n°32).

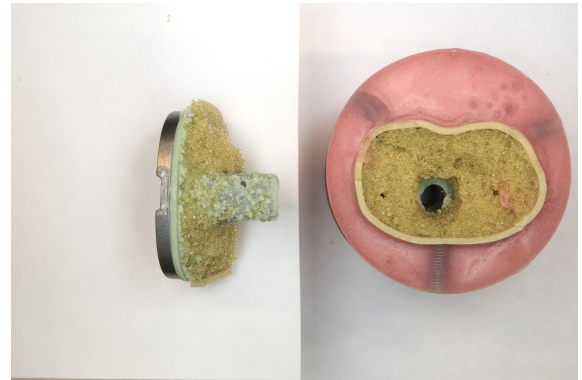


Figure 5.15: Lateral view from a failure between the bone and the bone-cement with the rest of the model embedded into resin (sample n°32).



Figure 5.16: Top view from a bone failure (sample n°05).

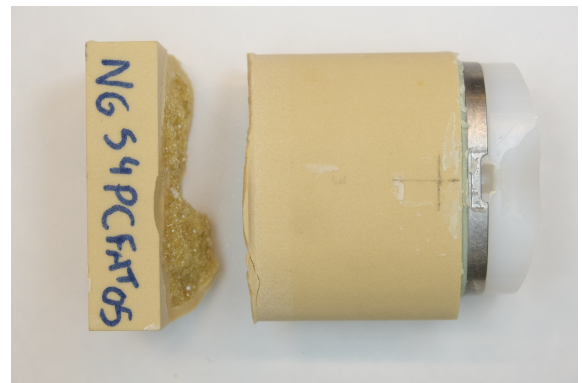


Figure 5.17: Lateral view from a bone failure (sample n°05).



Figure 5.18: Top view from a common implant detachment failure from the implant point of view (sample n°31).

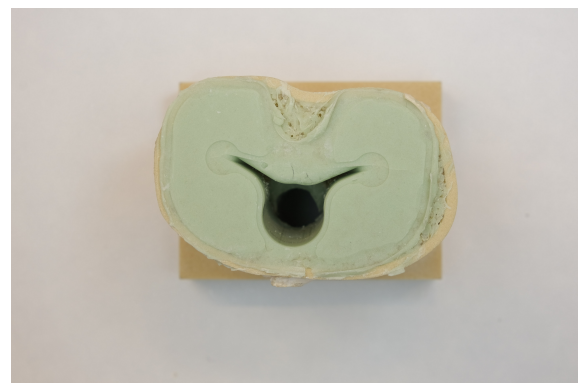


Figure 5.19: Top view from a common implant detachment failure from the bone point of view (sample n°31).

Chapter 6

Discussion

Total knee arthroplasty (TKA) or total knee replacement (TKR) is an orthopedic surgical technique that involves replacing the articular surfaces of the knee joint. Currently, the degradation of the natural cartilaginous tissue and the subsequent degeneration of the bones account for approximately 90% of the reasons why this procedure is done. With the progressive aging of the population, TKA occurrence is expected to double by 2030. Although considered highly successful, both implant decay over time and external factors, such as aseptic loosening of the tibial tray, could call for a TKA revision surgery. This secondary operation is not only considered extremely painful but also very expensive. In Norway alone, the economic burden associated with these revision surgeries was estimated to be approximately 110 million Norwegian kroner (NOK) in 2016 which would translate to 13.24 million euros in 2022. In bigger markets, such as America, this number can exceed the billion euros mark.

Implants with higher early mobility have been established to be more likely to fail. Most of the research links inducible and permanent displacement with implant stability. The purpose of this study is to examine the impact of alteration of the implant design on its stability, namely the presence or absence of a PMMA coating on the metallic layer of the tibial baseplate and its response related to adhesion metrics when these designs are implanted under non-optimal conditions, i.e., when contaminated with bone marrow fat.

6.1 The cementation process

It was assumed that the type of bone model used did not allow for a considerable depth of penetration of the cement due to the lack of interconnected porous on the plastic foam, translating into the thickness of the exterior cement. Because the cementation process was not performed by a professional - although, with the assistance of one -, differences in cement thickness were measured in the anterior, posterior lateral, and posterior medial positions to classify the samples according to their initial alignment.

Malalignment of the tibial baseplate is known to be strongly linked to initial TKR failure and consequent implant debonding, often appearing shortly after surgery. The optimal positioning of

the component, on the other hand, is still being debated, and more scientific research is needed for a complete understanding [28].

An optimal *in vivo* implantation should achieve cement depth penetration of 3 to 4 mm, with thinner layers linked with early implantation failure and thicker layers with "thermal-induced" osteonecrosis [25, 29, 30]. On the other hand, the exterior cement thickness of successful TKAs when studied in real bone is measured to be around 1mm. The experimental figures sit, on average, right in between the optimal values for internal and external cement thickness but are, nonetheless, lower than what the bibliography suggests if compared only to cement depth penetration.

6.2 Implant stability

The migration numbers on pre-coated contaminated baseplates indicate a tendency (*ns*) for bigger movements. This effect might be related to the overall thicker cement mantle in this group as seen in table 5.1 and on the positive correlation presented in subsection 5.2.1.2 and figure 5.6. Lift-off values, on the other hand, are similar between the four different groups.

The micromotion was consistent between all groups tested, indicating uniformity in the implantation procedure, as well as good stability of the baseplate itself and their minimal design alteration. Neither their design differences nor the possible contamination affected their stability. Values in the 20 μ m range such as those discovered in this research are comparable with similar *in vitro* studies [25, 31, 32].

The system used in this project is an alternative simplified method used to measure the migration and micromotion results of the tibial components which did not allow for rotation measures nor translation on any other axis than Y. While analyzing only axial displacement gives us an idea of the implant's behavior, a more robust analysis that considers the whole implant as a solid object would provide stronger results. Either using visual three-dimensional methods or by adding more transducers on different axis, the results of our study would be more comparable to those from traditional RSA (radiostereometric) and their MTPM (maximum total point motion).

Assuming that 20.000 cycles would indicate about three months following implantation, it is crucial to examine the substantial differences in the migration results of this study when compared to a normal RSA analysis for a comparable time point.

The migration results in micrometers, such as those obtained in this *in vitro* study (-6.83 μ m to 19.06 μ m), appear to be lower compared to the millimeter scale on the Y-axis (transverse axis) obtained in standard RSA investigations (0.2 mm to 2.1 mm) [18, 33, 34]. That being said, RSA's resolution is limited to 0.2 mm, so no smaller numbers can be obtained. Besides, RSAs are usually reported at the 2-year mark, instead of the 3 months assumed for this project. Assuming a linear progression (approximately 6.36 μ m per month) for the biggest migration value in our case (19.06 μ m) would result in a 2-year motion of 0.15 mm, which is well into the acceptable range of motion for tibial baseplates. Nonetheless, the small values can also be explained considering the relation between compressive modulus and its strain - equations 6.1 and 6.2 - in which a smaller modulus value would result in higher strain values (maximum of one), translating into a material

that deforms less when suffering a compressive force. This variable is important since the implant rests on the cortical shell in all three measured positions (anterior, lateral, and medial).

$$E = \frac{\sigma}{\epsilon} \quad (6.1)$$

Equation for Compressive Modulus.

$$\epsilon = \frac{\text{Compressed Length}}{\text{Original Length}} \quad (6.2)$$

Equation for Strain.

6.3 Adhesion

Before the experiments, design 2 was hypothesized to have a stronger binding due to the PMMA coating. The findings did not support this expectation, with pre-coated samples having similar ranges of push-out force compared to non-coated samples. The bone marrow fat infiltration drastically decreased the bonding strength between the implant and the cement for non-coated samples, with a mean push-out strength of almost half of their dry counterparts. The same cannot be said for pre-coated tests, in which contaminated and dry implants had very similar bonding forces. Pre-coating the tibial baseplate with PMMA seems to reduce the effect of eventual fat contamination, maintaining its bond to the bone cement.

J. Ryan, Martin *et. al.* tested the adhesion strength for 4 different implant design in cadaveric bones comparing their results when the implant had been pre-conditioned, being exposed to motion and stability testing after cementation, vs no conditioning at all. For comparisons, table 6.1 displays the pull-out results of those different designs when subjected to motion compared to the push-out value of the two designs studied in our project. Under dry conditions, non-coated and pre-coated adherence are comparable to the Attune and Persona implants. Regardless of the scenario, both NexGen designs evaluated on this project outperformed the Vanguard competitor [35].

	<i>Pull Out Results</i>				<i>Median Push – Out Results</i>	
	<i>DePuy Synthes © Attune</i>	<i>Zimmer Biomet © Persona</i>	<i>Stryker © Triathlon</i>	<i>Zimmer Biomet © Vanguard</i>	<i>NexGen © Option</i>	<i>NexGen © PreCoated</i>
Strength (N)	6146.00	5496.00	4054.00	2196.00	* 6000.00 ** 3414.60	* 5121.10 ** 5804.70

Table 6.1: Comparison Table for adhesion tests vs reference values from [35]

* *Samples in dry conditions*

** *Samples in contaminated conditions*

Kelly, Blane C *et. al.* used a similar setup to explore the impact of fat-contamination on a first and second-generation Attune (DePuy Synthes, Warsaw, IN) tibial baseplate, and the PFC Sigma (DePuy Synthes, Warsaw, IN) implant. When compared to results obtained in this project, both non-coated and pre-coated NexGen designs performed better than the 1st gen Attune and Sigma, when implanted in dry conditions. The NexGen contaminated non-coated group had values similar to those of the contaminated Attune [26].

In Appendix F, group-wise graphs of the load curves applied to the samples are presented from figures F.1 to F.4.

6.4 Weaknesses and limitations

6.4.1 The implant's alignment and the force direction

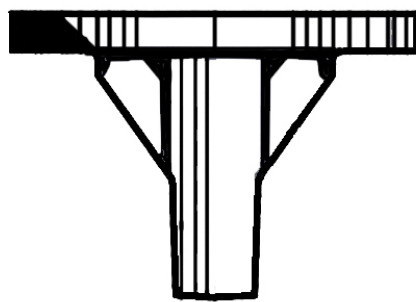


Figure 6.1: Scheme of an anterior view of the tibial baseplate showcasing the coronal plane, adapted from [36].

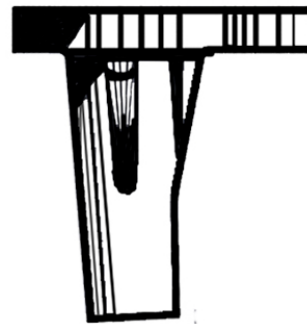


Figure 6.2: Scheme of a right sagittal view of the tibial baseplate, adapted from [36].

Although the optimal alignment of the tibial baseplate component is still debated in the clinical community, an anterior tibial slope on the sagittal plane is said to increase the risks of implant instability in patients [28,37].

Load distribution in biomechanical in vitro studies of the knee joint can significantly change the findings obtained. That being said, totally accurate reproduction of its orientation was not always possible mostly because the rig built for our experiments could not automatically position itself and due to the previously mentioned differences in cement thickness between the specimens.

While non-coated samples had a posterior tibial slope for both dry (88.1°) and fat-contaminated situations (89.2°), the pre-coated dry group had 2 of its 4 samples implanted in an anterior tibial slope (tests n°1 and 2). Those differences might impact the distribution of the load through the baseplate, such as the variation of the positioning for highest subsidence and lift-off presented in table 5.3 or even cause tilting of the femoral component during the stability test [28,37].

Lastly, the tilting of the tibial baseplate is observed in virtually all samples, either in the sagittal or coronal plane and its cause can be assigned to this load variation.

6.4.2 Bone quality and its effects

Figure 5.14 to 5.19 contains 3 of the 4 different failures encountered during the push-out testing phase of the project. It is important to note the high recurrence of bone failure in this experiment, proving itself a great weakness for the correct estimation of the adhesion strength of those samples. The push-out force for tests that presented a bone failure is, therefore, assumed to be higher than the value recorded.

6.4.3 Power analysis

A post-hoc power analysis was performed to calculate the power achieved by the studied parameters. A probability error (α) of 0.05 was defined the software used was G*Power Software version 3.1.9.7.

The ideal sample sizes to achieve a 0.95 power would be $N > 7$ samples on the previous comparisons. Only the adhesion tests were used for demonstration as they are the most relevant outcome variable to this project.

(a) Adhesion test for non-coated design (dry vs fat)

Comparing a mean value for the dry samples of 6090.85 N and its standard deviation of 1279.35, to the fat contaminated with 3316.67 N and 670.57 for standard deviation, the effect size d is calculated to be 2.72. The two tails power of this comparison is evaluated to be **0.81**. This comparison stands out as it was the only significant difference found on the project.

The values presented in the previous point indicate a relatively moderate probability that in case these tests would be repeated in a future project, the same differences regarding the effect of eventual fat contamination for both designs would be observed.

Achieving the sufficient sample size for 0.95 power in all four groups would require an additional 17 weeks since on average a single test takes 3 days not considering the rig availability.

Chapter 7

Conclusion and future prospects

This chapter will summarize the important research findings in connection to the research goals and questions discussed in chapter 2, as well as their worth and contribution to the orthopedic community. It will also re-evaluate the study's limitations further explored in section 6.4 and provide recommendations for further research.

The main goal of this project was to compare two different designs of the tibial baseplate and their performance when implanted in fat-contaminated settings, namely their stability under a cyclical load and their adhesion to the bone model following the pre-conditioning with periodic forces.

When comparing only stability metrics, the findings show that both designs performed remarkably similarly. The contamination with bone marrow fat, on the amount used, had little effect on the early implant's stability. When exposed to fat on the interface between the baseplate and the bone-cement, non-coated samples saw a significantly greater reduction in push-out force than pre-coated samples, indicating an implant design that is less challenging for the operating doctor in terms of potential contamination of the implanted site.

Based on the previously discussed points, the transition from non-coated to pre-coated tibial baseplate as the most commonly used design in the local Haukeland student hospital appears to provide both the professional and the patient with a safer starting point for TKA's, despite its relatively higher cost. Furthermore, there have been limited in vitro studies comparing these specific designs, which could give a small benefit to this project.

With this in mind, it is critical to assess the project's limits and to approach its outcomes with caution. First and foremost, the author's lack of expertise in the cementation procedure may have influenced certain findings, particularly at the start of the project. As observed in the power analysis, time restrictions and the small sample size have an impact on the final results. Considering the high recurrence of bone failures, there might have been a mistake in choosing push-out tests as the rig for assessing adhesion results instead of pull-out ones.



Figure 7.1: Summary of the proposed hypothesis and their final results at the end of the project. Analyzing the proposed hypothesis in chapter 2. For stability metrics, neither design nor contamination altered the the final means of the groups measured on this project. When looking at failure loads from the adhesion test, the presence of fat was the most significant factor altering the results, with some degree of cross-effect as well albeit not statistically significant.

Further research might use visual three-dimensional markers and software to measure the implant's migration and micromotion, bringing the final results closer to traditional RSA studies. The difference in motion recorded between anterior and posterior points is also a possible point for additional investigations. Future investigations may also focus on the comparisons between the pre-coated design and other common baseplates from other companies, as well as the 3D understanding of the biomechanics happening during a cyclical level walking simulation.

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Appendix A

PREPARING THE BONE CEMENT

THE IMPORTANCE OF VACUUM MIXING

The ability of bone cement to transmit mechanical forces between the implant and the bone plays a vital role in cemented arthroplasty.²¹ Since the 1990s, modern cementing technique has included vacuum mixing to achieve better mechanical properties of the bone cement. Pores at the cement-implant interface can give rise to cracks, which propagate through the cement mantle.^{22,23} Additional porosity within the cement mantle will speed the growth of the cracks. Limiting the propagation of these cracks is tantamount, as they are predictive of early aseptic failure in total knee arthroplasty.⁷ Vacuum mixing reduces air inclusions at the surface and in the cement matrix itself, giving rise to a stronger cement (Figure 2).²⁴ This improves the survival rate of the implant.^{25,26}

COMPARISON OF CEMENT POROSITY
PALACOS® mixed in open bowl vs. vacuum cartridge

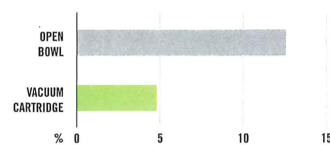


Figure 2: Adapted from 24. Geiger MH, et al. 2001.

MIXING TECHNIQUE

The bone cement should be mixed in a vacuum cartridge mixing system (like PALAMIX® or PALACOS® pro) with steady up- and downward movements of the mixing rod over the cartridge's entire length at one stroke per second until a homogenous dough develops. Multiple studies have shown that cartridge mixing systems improve the density, bending modulus, porosity, and durability (as measured by the fatigue strength) of PALACOS® R bone cement^{27,28} and may avoid the additional cement sampling step in open bowl systems.

PALACOS® is ready to use when a gloved "finger-test" demonstrates the cement is no longer sticky. In an early application phase, the doughy cement should be used to better penetrate the trabecula and improve the cement-implant bond strength.^{8,25,29}

The polymerisation process may be negatively impacted by mixing at speeds faster than one stroke per second, mixing for longer than 30 seconds, or introducing heat during mixing. Any of these may alter the timepoint at which the surgeons should apply the cement for optimal bone interdigitation and tibial tray bone strength.³⁰

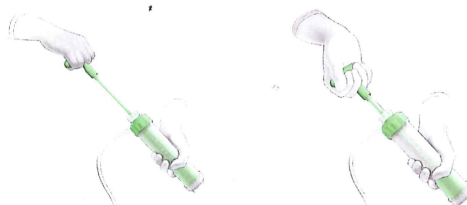


Figure A.1: Page from PALACOS's manual for Cement Preparation

CEMENT APPLICATION

TIBIAL APPLICATION

To minimise the risk of blood or fat being introduced to the cement mixture, a change of gloves is recommended before beginning application. PALACOS® should be applied to both the tray and stem of the tibial component. Applying cement only to the underside of the tray (so-called "surface cementing") reduces the push-out force required to separate the tray from the cement mantle and has been found to increase the rate of early aseptic failure.^{8,31-33}

Additionally, PALACOS® should be applied down the tibial keel hole with a knee pressuriser to increase the cement's interdigitation into the tibial plateau. In total knee arthroplasty, 3–5 mm is considered an ideal intrusion depth.³⁴ Interdigitation below 3 mm is less suitable to provide adequate interlock over the life of the prosthesis due to its inability to hold up to the shear forces generated during use.^{35,36} Insufficient interdigitation is also associated with greater risk of early aseptic loosening.⁷ On the other hand, interdigitation beyond 5 mm may increase the risk of heat necrosis-related negative effects on osteocyte viability.³⁷ Using a lesser quantity of cement during application and applying cement to only one of the surfaces (implant or bone) are both associated with suboptimal interdigitation (Figure 3).^{29,34}

The implant should be inserted to the bone by driving its components into position with an appropriate impactor. The bone-cement bond can be optimised by minimising the time between applying cement to the surfaces of the

MEAN PENETRATION OF CEMENT MANTLE BY FIXATION METHOD
For cementing of the tibia in TKA - $P = 0.007$

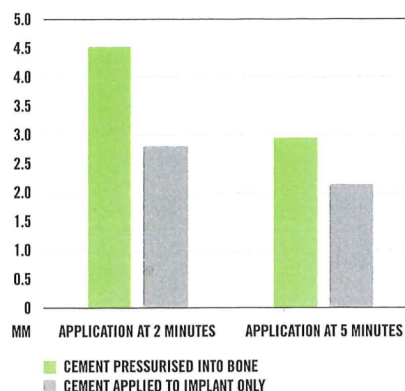


Figure 3: Adapted from 29. Pérez-Mañanes R, et al. 2012.

tibia (bone and implant) and joining the two cement surfaces together. Prolonged time between application and joining can lead to the formation of a skin on the bone cement surface, which can reduce the cement-cement bond strength.³⁸ Excess bone cement should be removed carefully while it remains in the working phase.

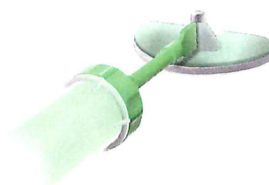


Figure A.2: Page from PALACOS's manual for Cement Application on Tibial End

Appendix B

<i>Implant weight in each sample</i>			
	<i>Test number</i>	<i>Implant ref.</i>	<i>Weight (g)</i>
<i>NexGen pre-coated</i>	1	5980-37-02	88.7
	2	5980-37-02	88.9
	3	5980-37-02	88.5
	4	5980-37-02	88.5
	5	5980-37-02	88.7
	6	5980-37-02	88.3
<i>NexGen Option</i>	13	5986-37-02	102.6
	19	5986-37-02	103.0
	20	5986-37-02	102.9
	21	5986-37-02	102.7
	30	5986-37-02	102.5
	31	5986-37-02	102.2
	32	5986-37-02	103.5

Table B.1: Weight in grams for all the implants used on the project after their preparation.

Appendix C

```
%% Input the File

total = load("C:\Users\mayah\OneDrive\Ambiente de Trabalho\TUDO\Plotar o Triangulo no
Ambiente 3d\Teste Maya 8 NexGen PreCoated Fat 8\exported_NexGenno8_2.txt");

%% Separating the File
time = total(:,1);
force = total(:,2);

lvdt3 = total(:,3); %Anterior
lvdt4 = total(:,4); %Lateral
lvdt5 = total(:,5); %Medial

%% Zeroing
lvdt3_zeroed = lvdt3 - lvdt3(100,1);
lvdt4_zeroed = lvdt4 - lvdt4(100,1);
lvdt5_zeroed = lvdt5 - lvdt5(100,1);
%% Lower Peaks (migration)
[lvdt3_down,~] = findpeaks(-lvdt3_zeroed,100,'MinPeakDistance',1);
lvdt3_down = -lvdt3_down;
[lvdt4_down,~] = findpeaks(-lvdt4_zeroed,100,'MinPeakDistance',1);
lvdt4_down = -lvdt4_down;
[lvdt5_down,~] = findpeaks(-lvdt5_zeroed,100,'MinPeakDistance',1);
lvdt5_down = -lvdt5_down;

%% Upper Peaks
[lvdt3_upp,~] = findpeaks(lvdt3_zeroed,100,'MinPeakDistance',1);
[lvdt4_upp,~] = findpeaks(lvdt4_zeroed,100,'MinPeakDistance',1);
[lvdt5_upp,~] = findpeaks(lvdt5_zeroed,100,'MinPeakDistance',1);

%% Maximum Lift-Off
[maxLiftOff3,Index3] = min(lvdt3_down);
[maxLiftOff4,Index4] = min(lvdt4_down);
[maxLiftOff5,Index5] = min(lvdt5_down);

fprintf("Maximum Lift Off-----\n")
fprintf("LVDT3 = %4.2f microM at [%d]\n",maxLiftOff3*10^3,Index3)
fprintf("LVDT4 = %4.2f microM at [%d]\n",maxLiftOff4*10^3,Index4)
fprintf("LVDT5 = %4.2f microM at [%d]\n",maxLiftOff5*10^3,Index5)

%% Maximum Subsidence
[maxSubs3,Index3Subs] = max(lvdt3_down);
[maxSubs4,Index4Subs] = max(lvdt4_down);
[maxSubs5,Index5Subs] = max(lvdt5_down);

fprintf("Maximum Subsidence-----\n")
fprintf("LVDT3 = %4.2f microM at [%d]\n",maxSubs3*10^3,Index3Subs)
fprintf("LVDT4 = %4.2f microM at [%d]\n",maxSubs4*10^3,Index4Subs)
fprintf("LVDT5 = %4.2f microM at [%d]\n",maxSubs5*10^3,Index5Subs)

%% Manually Correct the sizes of the detected upper peaks and detected lower peaks

%% Micromotion
micromo3 = (lvdt3_upp - lvdt3_down);micromo3 = micromo3(end,1);
micromo4 = (lvdt4_upp - lvdt4_down);micromo4 = micromo4(end,1);
```

Figure C.1: Matlab code for data analysis

```

micromo5 = (lvdt5_upp - lvdt5_down);micromo5 = micromo5(end,1);

fprintf("Final Micromotion-----\n")
fprintf("LVDT3 = %4.2f microM at [20.000]\n",micromo3*10^3)
fprintf("LVDT4 = %4.2f microM at [20.000]\n",micromo4*10^3)
fprintf("LVDT5 = %4.2f microM at [20.000]\n",micromo5*10^3)

```

Figure C.2: Matlab code for data analysis

```

function [new_name,fileCreated_Bool] = convertingFile(file_name,path_name)
%This function is created to turn the file straight out of the Madaq
% to one that we can analyze in Matlab. This is a necessary step
% to the well functioning of the App.

full = fullfile(path_name,file_name);
S = fileread(full);

% Header Delimitation.
ini = strfind(S,'Channel : 1');
fin = strfind(S,'Ch6(Volts)');
if (length(fin) > 1)
    fin = fin(2);
end

if isempty(ini) || isempty(fin)
    fileCreated_Bool = 0;
    error('The file was not the correct one')
end

% Substituting the commas with dots on the file.
[~,name_of_file_without_ext,ext] = fileparts(file_name);
new_name = [name_of_file_without_ext '_Converted' ext];
fid = fopen(fullfile(path_name,new_name),'wt');
fprintf(fid, strrep(S(fin+13:end), ',', '.'));
fclose(fid);

% Separate only the important columns that we need.
file_loaded = load(fullfile(path_name,new_name));
final_file_converted = table(file_loaded(:,1),file_loaded(:,2),file_loaded(:,4),
file_loaded(:,5),file_loaded(:,6));
writetable(final_file_converted,fullfile(path_name,
new_name),'Delimiter','\t','WriteRowNames',false,'WriteVariableNames', 0);

fileCreated_Bool = 1;
end

```

Figure C.3: Matlab code for converting the data outputed by the MADAQ box

Appendix D

<i>Bone model weight in each sample</i>		
	<i>Test number</i>	<i>Weight (g)</i>
<i>NexGen pre-coated</i>	1	105.3
	2	106.0
	3	107.1
	4	106.5
	5	107.6
	6	111.3
<i>NexGen Option</i>	13	106.5
	19	106.4
	20	108.3
	21	105.9
	30	105.5
	31	107.7
	32	106.9

Table D.1: Weight in grams for all the bone models used on the project after the preparation

Appendix E

<i>Cement weight in each sample</i>		
	<i>Test number</i>	<i>Weight (g)</i>
<i>NexGen pre-coated</i>	1	13.8
	2	10.9
	3	14.6
	4	14.8
	5	16.8
	6	17.2
<i>NexGen Option</i>	13	18.9
	19	19.7
	20	18.9
	21	20.6
	30	15.6
	31	13.7
	32	14.5

Table E.1: Weight, in grams, of the cement used in all the experiments.

Appendix F

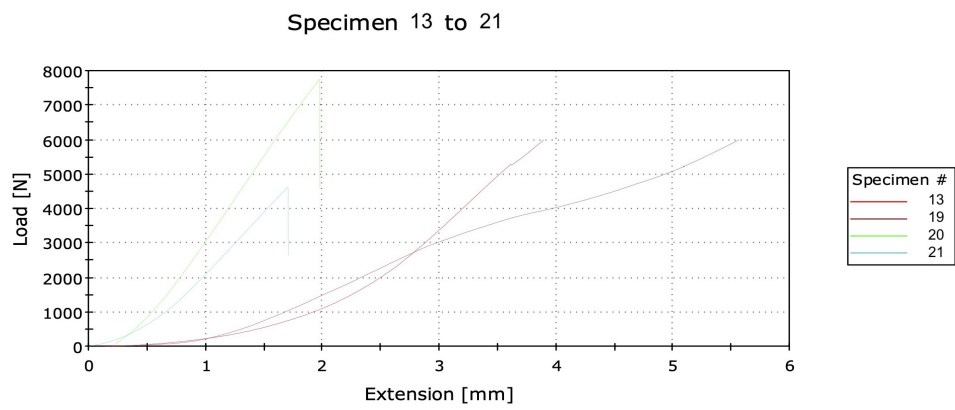


Figure F.1: Push-out test graph for non-coated design samples in dry conditions

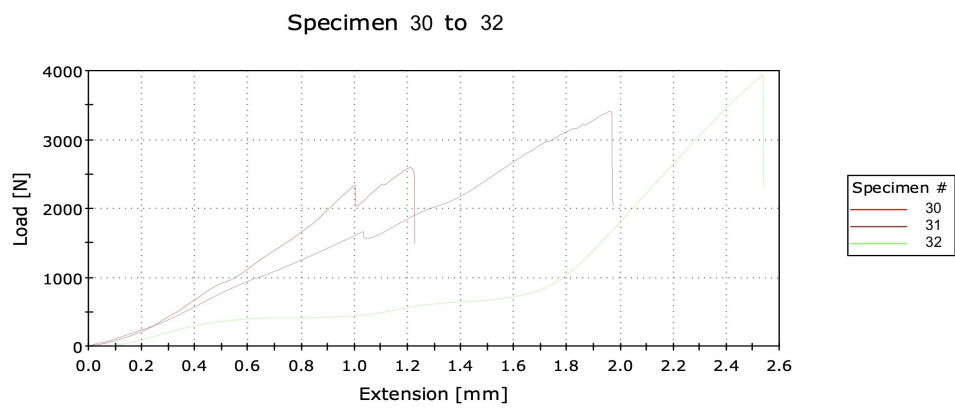


Figure F.2: Push-out test graph for non-coated design samples in fat contaminated conditions

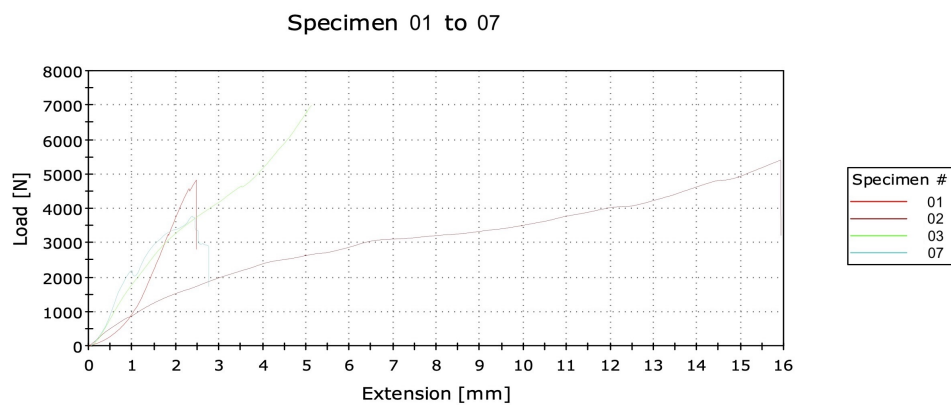


Figure F.3: Push-out test graph for pre-coated design samples in dry conditions

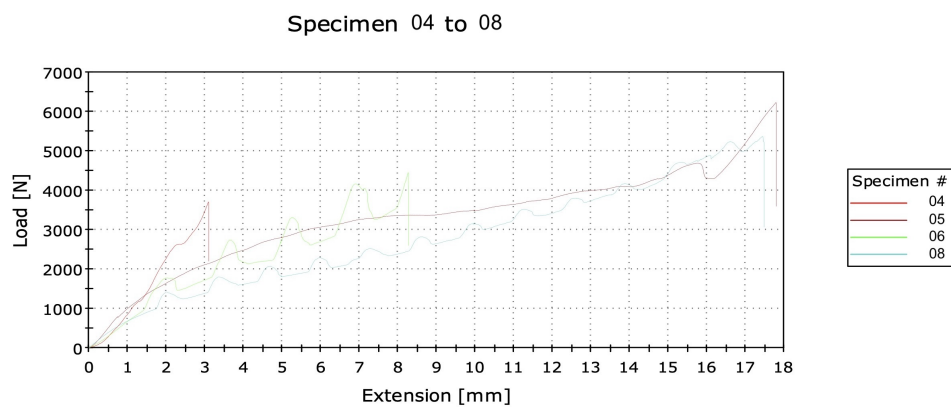


Figure F.4: Push-out test graph for pre-coated design samples in contaminated conditions

Appendix G

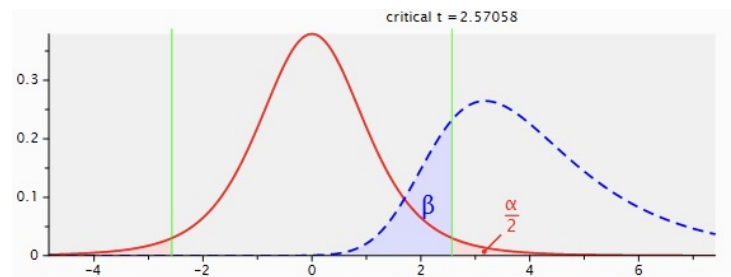


Figure G.1: Graph created by the power analysis for non-coated dry vs fat conditions

Test family		Statistical test	
t tests		Means: Difference between two independent means (two groups)	
Type of power analysis			
Post hoc: Compute achieved power – given α , sample size, and effect size			
Input Parameters			
Determine =>	Tail(s)	Two	
	Effect size d	2.7161267	
	α err prob	0.05	
	Sample size group 1	4	
	Sample size group 2	3	
Output Parameters			
Noncentrality parameter δ		3.5562446	
Critical t		2.5705818	
Df		5	
Power (1- β err prob)		0.8089502	

Figure G.2: Parameters set as input and output for the post-hoc analysis of non-coated dry vs fat conditions

☒ $n1 = n2$

Mean group 1	6090.85
Mean group 2	3316.673
SD σ group 1	1279.349
SD σ group 2	670.574

Calculate Effect size d 2.716127

Calculate and transfer to main window

Close

Figure G.3: Parameters set as input for the post-hoc analysis of non-coated dry vs fat conditions

Appendix H

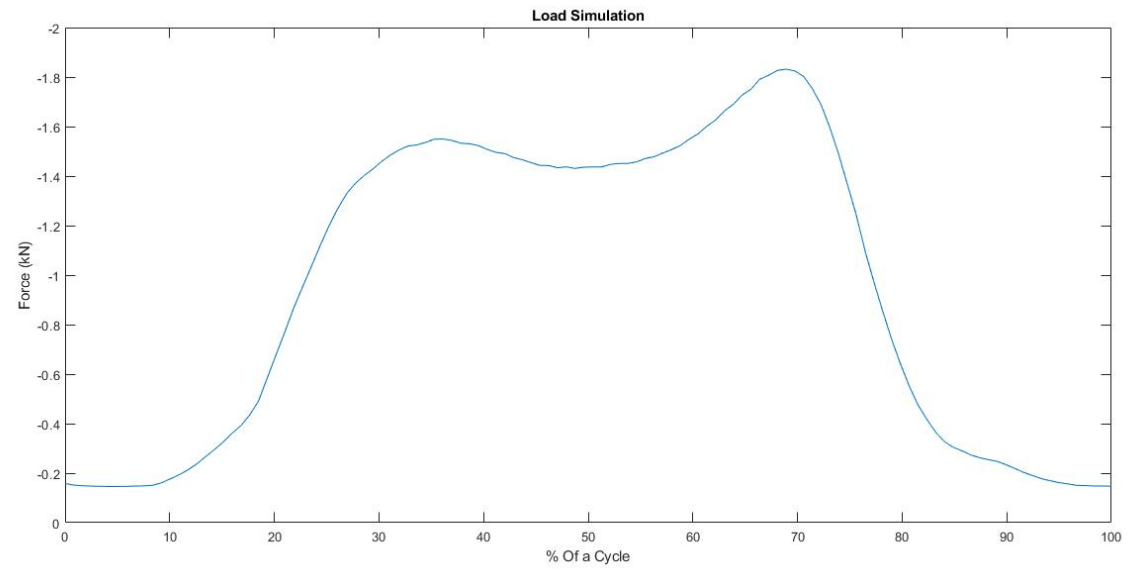


Figure H.1: Graph of the Load simulation for a level walking in the percentage of a cycle (from 0 to 100).