

T.R.
GEBZE TECHNICAL UNIVERSITY
GRADUATE SCHOOL

**MECHANICAL ANALYSIS AND PATIENT
EVALUATIONS OF NEW GENERATION HIP IMPLANTS**



DUNYA SENA YILDIZ

**A THESIS OF MASTER OF SCIENCE
DEPARTMENT OF MECHANICAL ENGINEERING**

ADVISOR: PROF. DR. ILYAS KANDEMIR

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*To my
Mother and
Husband*

ABSTRACT

It is becoming increasingly important to minimize wear and breakage in orthopedic implants over time and to make the right implant-patient combination. The main purpose is to examine how much the factors affect possible wear and the wear-reducing effects in line with those factors.

As a hip prosthesis, stem from Ti-6Al-4V alloy and a femoral head from Zirconia material were designed and analyzed with finite element analysis. For hip prosthesis, heads with different diameters of 28, 32, 36, 40 and 44 mm were designed to examine the effect of femoral head size on trunnionosis corrosion. To see the effect of the gap between the trunnion and the femoral head, which is one of the other potential factors in wear, modeling with -3, 0 and +3 offset gap was made and finite element analysis and static and dynamic analyzes were carried out. Within the scope of dynamic analysis, a person's walking moment was simulated for 5 seconds and modal and transient analyzes were performed by loading the forces acting on the hip and therefore the prosthesis.

The study was carried out in four stages to see the effects of head size, offset gap, offset angle and fixation on wear. Stress, deformation obtained from static analyses, and moment, vibration, frequency and sliding distance results obtained from dynamic analyzes were associated with wear factors and it was studied to determine which combinations could be associated in which situation. Modeling at all stages was modeled using Autodesk Fusion 360 and finite element analyzes were performed with ANSYS 19.2.

Keywords: Finite Element Analysis, Hip Prosthesis, Trunnionosis, Hip Implant, Wear Friction

ÖZET

Ortopedik implantlarda zamanla aşınma ve kırılmayı en aza indirmek ve doğru implant-hasta kombinasyonunu oluşturmak giderek önem kazanmaktadır. Temel amaç, faktörlerin olası aşınmayı ne kadar etkilediğini ve bu faktörlere uygun aşınma azaltıcı etkileri incelemektir.

Kalça protezi olarak Ti-6Al-4V alaşımından gövde ve Zirkonyum malzemeden femur başı tasarlanmış olup sonlu elemanlar analizi ile analiz edilmiştir. Kalça protezi için 28, 32, 36, 40 ve 44 mm'lik farklı çaplarda femur başları tasarlanarak femur baş boyutunun trunnionosis korozyonuna etkisi incelenmiştir. Aşınmadaki diğer potansiyel faktörlerden biri olan trunnion ile femoral baş arasındaki boşluğun etkisini görmek için -3, 0 ve +3 offset boşluğu ile modelleme yapılmış ve sonlu elemanlar analizi ile statik ve dinamik analizler gerçekleştirilmiştir. Dinamik analiz kapsamında bir kişinin yürüme anı 5 saniye boyunca simüle edilmiş olup kalça ve proteze etki eden kuvvetler yüklenerek modal ve transient analizler gerçekleştirilmiştir.

Bu çalışma, aşınma üzerinde baş boyutu, offset aralığı, offset açısı ve fiksasyonun etkilerini görmek amacıyla dört aşamada gerçekleştirilmiştir. Statik analizlerden elde edilen stres, deformasyon ve dinamik analizlerden elde edilen moment, titreşim, frekans ve kayma mesafesi sonuçları aşınma faktörleri ile ilişkilendirilmiş ve hangi durumda hangi kombinasyonların ilişkilendirilebileceği belirlenmeye çalışılmıştır. Tüm aşamalardaki modellemeler Autodesk Fusion 360 kullanılarak modellenmiş ve ANSYS 19.2 ile sonlu elemanlar analizleri gerçekleştirilmiştir.

Anahtar Kelimeler: Sonlu Elemanlar Analizi, Kalça Protezi, Trunnionosis, Kalça İmplantı, Aşınma Sürtünmesi

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LIST OF SYMBOLS AND ABBREVIATIONS

THA	: Total hip arthroplasty
THR	: Total hip replacement
PTFE	: Polytetrafluoroethylene
UHMWPE	: Ultra-high molecular weight polyethylene
Ti-6Al-4V	: Titanium-6Aluminum-4Vanadium
λ	: Film of thickness
a	: Radius of a hemispherical asperity shape
k_{cor}	: Corrosive wear coefficient
W_L	: Linear wear
K_w	: Wear coefficient
COF	: Friction coefficient
E	: Young modulus
Co-Cr-Mo	: Cobalt-Chrome-Molybdenum

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1. INTRODUCTION

1.1. Total Hip Arthroplasty

Total Hip Replacement (THR) or Total Hip Arthroplasty (THA) is a surgical procedure to replace the hip joint when it is severely damaged or diseased. During this procedure, damaged bone and cartilage are removed and replaced with an artificial hip prosthesis. Steps of the replacement operation are shown on **Figure 1.1**.

Total hip replacement is usually performed for patients who experience pain and limited mobility due to conditions such as osteoarthritis, rheumatoid arthritis, hip fracture, or avascular necrosis.

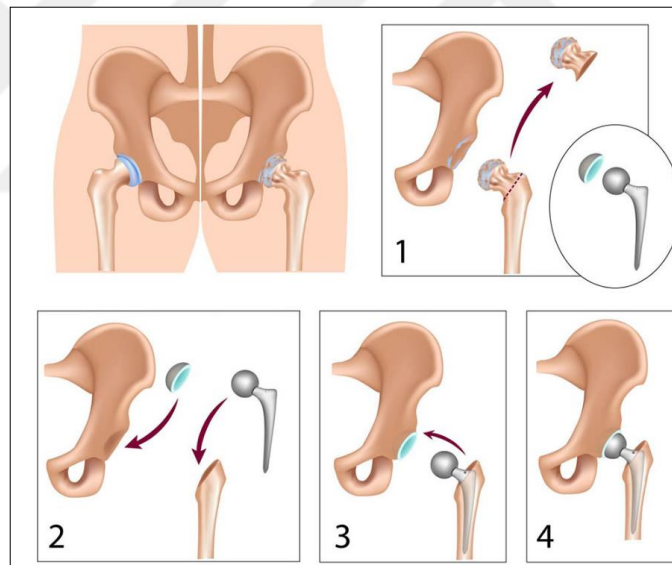


Figure 1.1: Total hip replacement steps [1].

Havelin et al., studied 17,444 hip prosthesis replacements in their study in 1993 [2]. In this ‘Three-dimensional stress analysis of the femur with prosthetic hip stem utilizing computed tomography scans of the femur for geometry and material properties’ study, of the surgical procedures examined, 2,350 were revision surgical procedures, and 69% of the patients were women. The data obtained are detailed in **Table 1.1** [3].

Table 1.1: Comparison of age versus revisions in total hip replacement (Norwegian Arthroplasty Register 9/87-12/90) [3].

Age	Revisions	Primary
<20	2	15
20-29	13	62
30-39	40	191
40-49	101	541
50-59	184	1543
60-69	675	4866
70-79	1012	6500
>80	322	1374
Total	2349	15092

When the operations performed and the ages of the patients are evaluated, primary operations were performed on many people from the age of 12 to 97 years, and on average, they were performed at the age of 70.

According to the data in **Figure 1.2**, revisions occur in several age groups, from 18 years old to 93 years old and the average age group is 71 years old. Based on the data obtained, the revision rate is 14.8 in women and 17.4 in men.

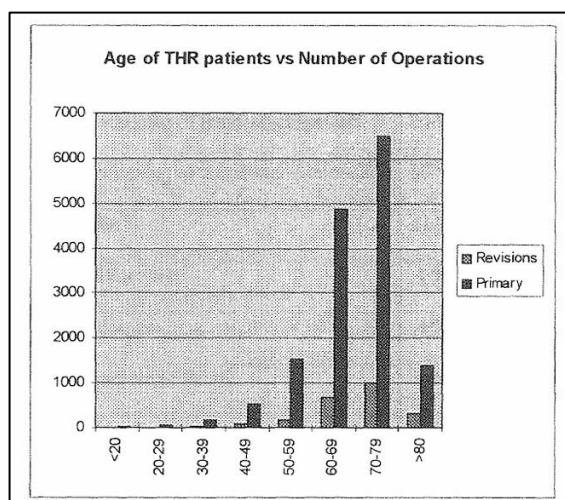


Figure 1.2: Number of operations versus age of patients (Norwegian Arthroplasty Register 9/87 – 12/90) [3].

Table 1.2: Causes of primary hip replacement operations [3].

Causes	Percentage
Primary arthritis	68%
Rheumatoid arthritis	4%
Sequelae after hip fracture	13%
Dysplasia with dislocation	2%
Other pediatric hip disease	1%
Ankylosing spondylitis	0.4%
Other	2%

The causes of the total hip replacements such as primary arthritis, dysplasia with dislocation and the reasons for THA replacement procedures and their occurrence percentages are given in **Table 1.2** and **Table 1.3**, respectively. The revision rate due to the acetabular component, which is one of the most common causes, is 56%. Loosening of the femoral head is the reason for 64% revision.

Table 1.3: Causes of revision operations [3].

Causes	Percentage
Loosening of femoral component	64%
Acetabular component loose	56%
Pain	11%
Dislocation	4%
Infection	4%
Fracture of femur	4%
Other	5%

To give long term successful results, the important point is choosing the correct implant with material and structure.

Additionally, biocompatibility of the material on the host tissue is the other factor of creating reliable effects.

In the first hip prosthesis used in the 1960s, polytetrafluoroethylene (PTFE) acetabular cup was used by combining a stainless-steel body. It has been concluded that this combination has low wear resistance due to wear and tear leading to infection [4].

Over time, rusting was replaced by steel and PTFE and studies were carried out with combinations of Co-Cr-Mo alloy and UHMWPE ultra-high molecular weight polyethylene materials. It was concluded that the wear resistance was at a good level.

Bobyn et al. studied on the effects of stem stiffness on femoral bone and analyzed the material properties and effects of titanium alloys and cobalt-chromium [5]. As a result of the study, it was determined that the resorption in the femoral bone was higher in the rigid stem.

1.2. Bone Tissue

Bone tissue is one of the building blocks of the body and is known as a very durable tissue. Bone tissue is composed of osteocytes, which are living cells in a matrix reinforced with minerals such as calcium and phosphorus and perform important functions such as mechanical support in various parts of the body, protection for organs and red blood cells (Figure 1.3). Bones have a rich blood supply, have a high ability to self-heal, and remain active throughout the individual's life [5].

The hip bone is a large bone that carries the weight of the body and is often subjected to various loads and stresses.

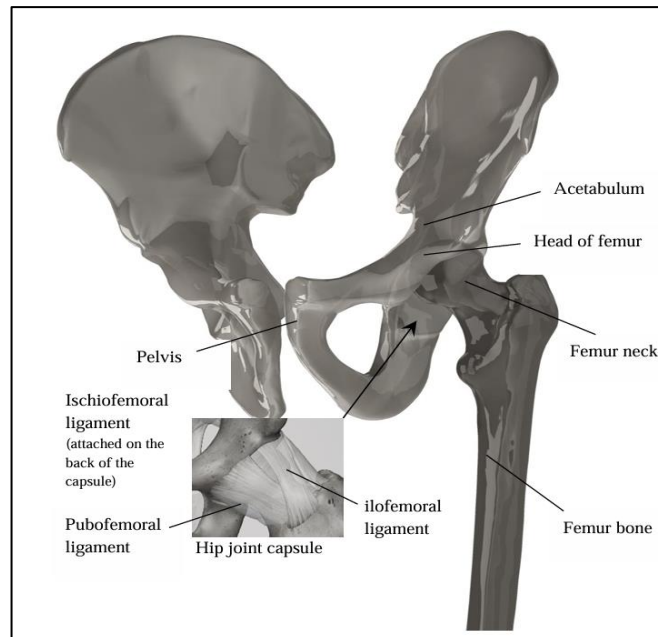


Figure 1.3: Bone Tissue [6].

Compared to other body parts, the hip bone has a lower risk of fracture.

One of the reasons for this is that the bones are protected by supporting structures such as muscles and ligaments around the hip joint.

Although the bone is strong, fractures occur in the hip bone due to age, sudden impacts, excessive force loading and possible cyst formation in the bone, while bone diseases such as osteoporosis can also reduce the load-bearing power of the bone [7].

The femur has a bone that extends from the hip joint to the knee joint and carries the weight of the body. The femur bones carry the weight of the body over and transfer it to the lower extremities when standing, walking, running or other physical activities. Fractures and damages occur in the hip due to falls, osteoporosis, sudden traumas, bone diseases and advancing age.

1.2.1. Biomechanics of Bone

The hip muscles have a significant impact on the skeletal system. The hip muscles are large, strong muscles that connect the upper body to the lower body and stabilize the joint between the pelvis (pelvic bone) and the femur (thigh bone).

Hip muscles enable various movements to be performed. These movements include daily activities such as walking, running, climbing stairs, bending and sitting. Additionally, the hip muscles help stabilize and support the body.

Kinetics studies the causes of motion, how objects in motion interact with forces, and the consequences of these interactions. Within the musculoskeletal system, the term kinetics is used to study the forces exerted by muscles and moments in joints. For example, when a person lifts weights, the kinetics of the forces exerted by the muscles behind this movement can be studied.

Biomechanics finds application in areas such as understanding human movement, improving sports performance, injury prevention and rehabilitation. Within the musculoskeletal system, the term biomechanics is used to study the forces, moments, and movements between muscles, bones, joint structures, and other tissues. Biomechanics studies are of great importance in fields such as optimizing sports performance or prosthetic design. The following items can be taken into consideration among the reasons for the importance of biomechanics studies on hip prostheses [8].

- **Improving Prosthetic Performance:** Biomechanics studies can be used to improve the performance of hip prostheses. Properties of the prosthesis such as durability, stability, range of motion and natural mobility can be optimized using biomechanical principles.
- **Ensuring Compatibility with the Body:** The compatibility of hip prostheses with the body is extremely important in terms of the patient's mobility, comfort and quality of life. Biomechanics studies evaluate factors such as prosthesis design, material selection and placement to improve the fit of prostheses to the body.
- **Ensuring Naturalness of Movement:** People wearing hip prosthesis will want the movements of the prosthesis to occur naturally. Biomechanics studies simulate the joint movements and muscle interactions of the prosthesis, optimizing the prosthesis design to provide natural mobility and comfort.
- **Reducing the Risk of Injury and Damage:** An improperly designed or applied hip replacement may increase the patient's risk of injury or tissue damage. Biomechanical studies can help minimize these risks by evaluating the effects of prosthesis on the body.

- **Supporting Rehabilitation:** The rehabilitation process of patients wearing hip prosthesis is critical in terms of correct use of the prosthesis and recovery of muscle strength. Biomechanics studies can guide the design of rehabilitation programs and the effective use of prosthesis.

The free body diagram (FBD) acting on the hip joint is a diagram that visually represents the external forces applied to the hip joint and the directions and magnitudes of these forces. The forces applied to the hip joint reflect the interactions on the bones that make up the joint (femur and pelvis) and the connections between these bones (muscles, ligaments, tendons). **Figure 1.4** shows the free body diagram acting on the hip joint and the force distribution under static conditions [9].

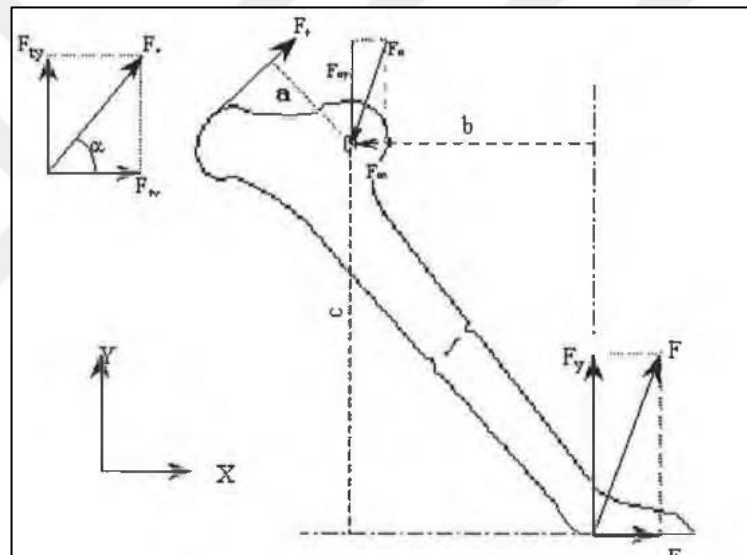


Figure 1.4: Free body diagram of foot with a, b, c moment arms [9].

The forces that can be shown on a free body diagram for a hip joint are generally as follows:

Weight Force (F_{grav}): The most basic force applied to the hip joint represents the person's body weight. This force is applied downwards to the femur (thigh bone) by gravity.

Muscle Forces (F_{muscle}): The muscles around the hip joint use their contractions to perform different movements. For example, the hip joint is controlled by movements

such as flexion, extension, abduction and adduction. These muscles apply force to the hip joint in various directions.

Reaction Force (F_{reaction}): The hip joint is in contact with other parts of the body, for example, placement in a sitting or lying position. At these contact points, there is a reaction force against the forces applied to the joint, which is shown in the free-body diagram.

Other External Forces: Other external forces may also act on the hip joint, especially depending on a particular activity or position. For example, when a person tries to lift a weight, the force exerted by the weight itself also acts on the joint [9].

These forces are represented as arrows on a free-body diagram. The direction of the arrows is determined by the direction in which the force is applied. This diagram helps analyze the forces acting on the hip joint and understand the balance in the joint.

1.2.2. Degree of Freedom of Hip Joint

The hip joint is one of the largest and most mobile joints in the human body. The hip joint is a ball-and-socket joint and can move around three main axes. This means that there are three degrees of freedom: flexion and extension, abduction and adduction, and internal and external rotation. In flexion, the leg moves forward and the hip flexes, while in extension, the leg moves backward and the hip bends back.

In the abduction and adduction movement, the leg moves sideways and away from the body with the abduction movement, while in adduction the leg moves sideways by approaching the body.

In internal and external rotation, internal represents the turning of the leg inward and external represents the outwards (**Figure 1.5**).

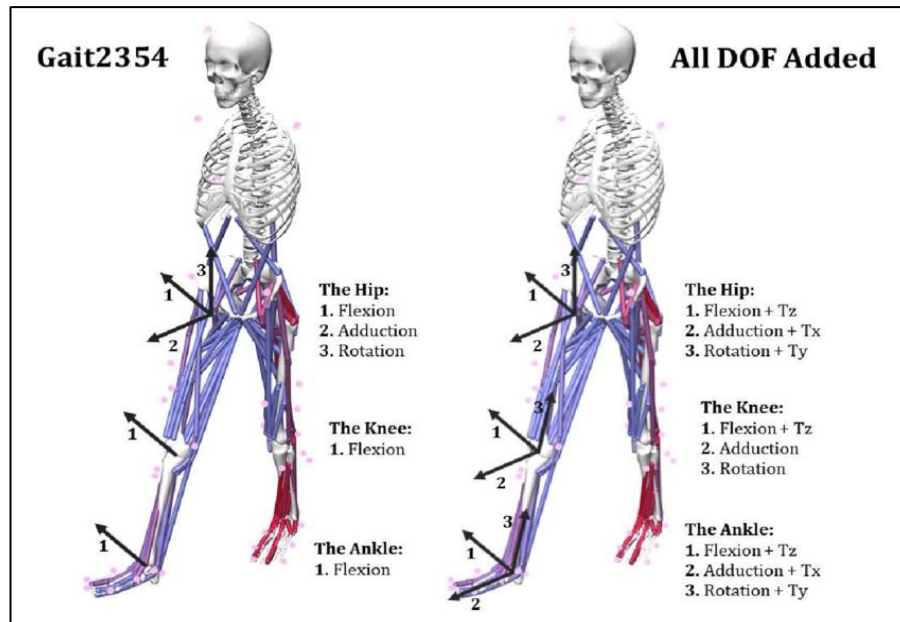


Figure 1.5: Degree of freedom of ankle, knee and hip joint [10].

The movement with the widest range of motion in the hip joint is the flexion-extension movement ranging from 20° in extension to 120° in flexion and occurs in the sagittal plane.

In the ranges from 45° in abduction and up to 30° in adduction, the abduction-adduction movement takes place in the frontal plane [11].

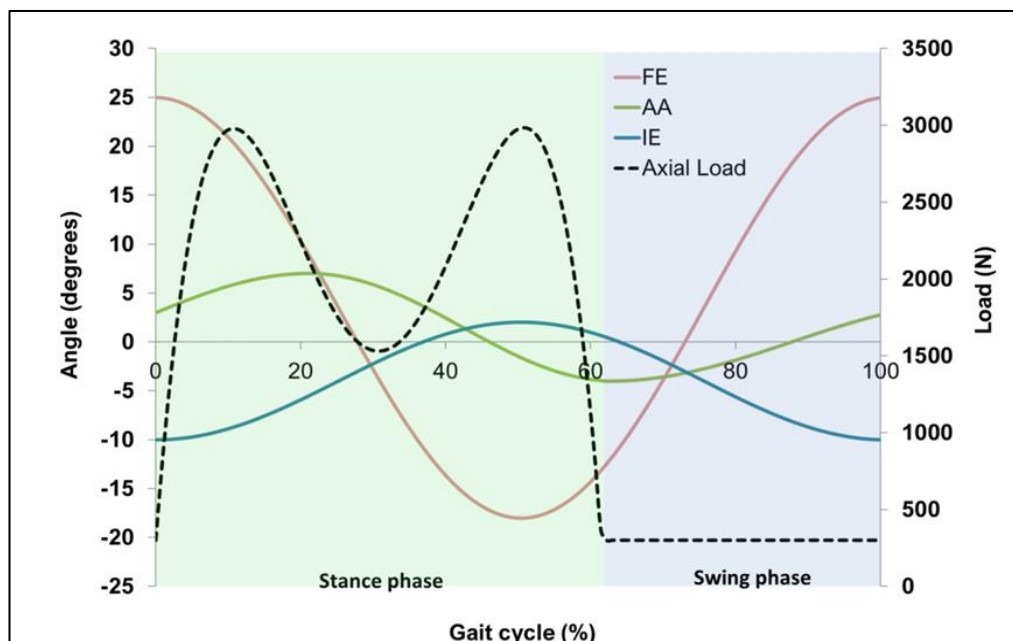


Figure 1.6: Gait cycle of degree of freedom motions [11].

Gait cycle describes a complete cycle of movements that a person makes while walking. This cycle is the time from the moment a foot first touches the ground to the moment the same foot touches the ground again. Gait cycle is divided into two phases indicated in **Figure 1.6** as stance phase and swing phase.

- Stance phase is the period when the foot touches the ground and carries body weight and constitutes approximately 60% of the gait cycle.
- The swing phase is the period when the foot lifts off the ground and swings forward and constitutes approximately 40% of the gait cycle.

In internal - external rotation, while the internal and external rotation changes by 40°, the movement occurs in the transverse plane.

1.3. Total Hip Prosthesis

Hip implant, also known as a hip prosthesis is a medical device used to replace a damaged or diseased hip joint. It consists of several components designed to mimic the function of the natural hip joint and restore mobility and relieve pain. Components typically included in a hip implant and important parameters are mentioned below.

Acetabular Component: This component replaces the damaged socket (acetabulum) of the pelvis. It is typically made of metal, ceramic, or plastic and may include a liner made of a high-density polyethylene material.

Femoral Component: This component replaces the upper part of the thigh bone (femur). It usually consists of a metal stem that is inserted into the femur and a ball-shaped head that replaces the natural femoral head. The femoral head can be made of metal, ceramic, or a combination of materials.

Articulating Surface: This is the interface between the acetabular component and the femoral head component. It may consist of a liner made of a wear-resistant material such as highly cross-linked polyethylene, ceramic, or metal.

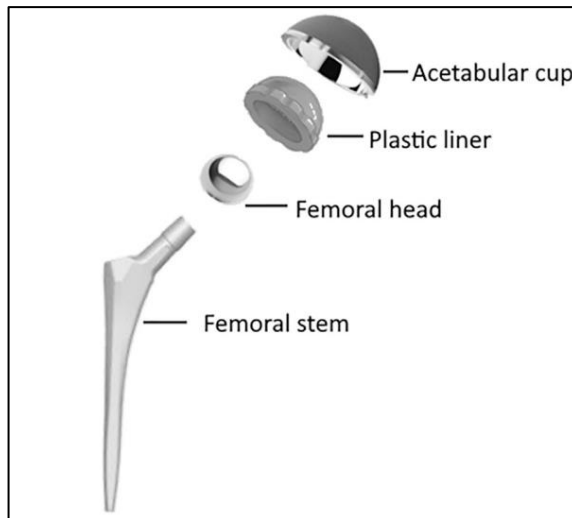


Figure 1.7: Components of the simple hip prosthesis [12].

Features that stand out and should be taken into consideration in hip prostheses are listed below.

- **Implant Size and Geometry:** The size and geometry of the hip implant components should be carefully chosen to match the anatomy of the patient's hip joint. Proper sizing and positioning of the components are essential for stability, range of motion, and longevity of the implant.
- **Material Selection:** The materials used in hip implants should be biocompatible, corrosion-resistant, and durable. Common materials include metals such as titanium, cobalt-chromium alloys, and stainless steel, as well as ceramics and highly cross-linked polyethylene.
- **Implant Design:** The design of the hip implant should promote stability, minimize wear, and optimize biomechanical function. Factors such as component shape, surface finish, and bearing type (e.g., metal-on-polyethylene, ceramic-on-ceramic) can affect the performance and longevity of the implant.
- **Surgical Technique:** The surgical technique used to implant the hip prosthesis is crucial for achieving optimal outcomes. Proper positioning and alignment of the components, preservation of bone stock, and soft tissue management are important considerations during surgery.
- **Patient Factors:** Patient-specific factors such as age, activity level, bone quality, and overall health should be taken into account when selecting and

implanting a hip prosthesis. These factors can influence implant selection, surgical approach, and postoperative rehabilitation.

Overall, the success of a hip implant depends on careful preoperative planning, precise surgical technique, and appropriate selection of implant components tailored to the individual patient's needs. Regular follow-up and monitoring are also important to assess the function and durability of the implant over time.

The femoral stem, one of the hip prosthesis components, ensures the transmission of loads to the supporting bone. Hip prostheses have been used together with replacement surgery for many years and there are various versions, and the existing stem models are listed in **Figure 1.8**.

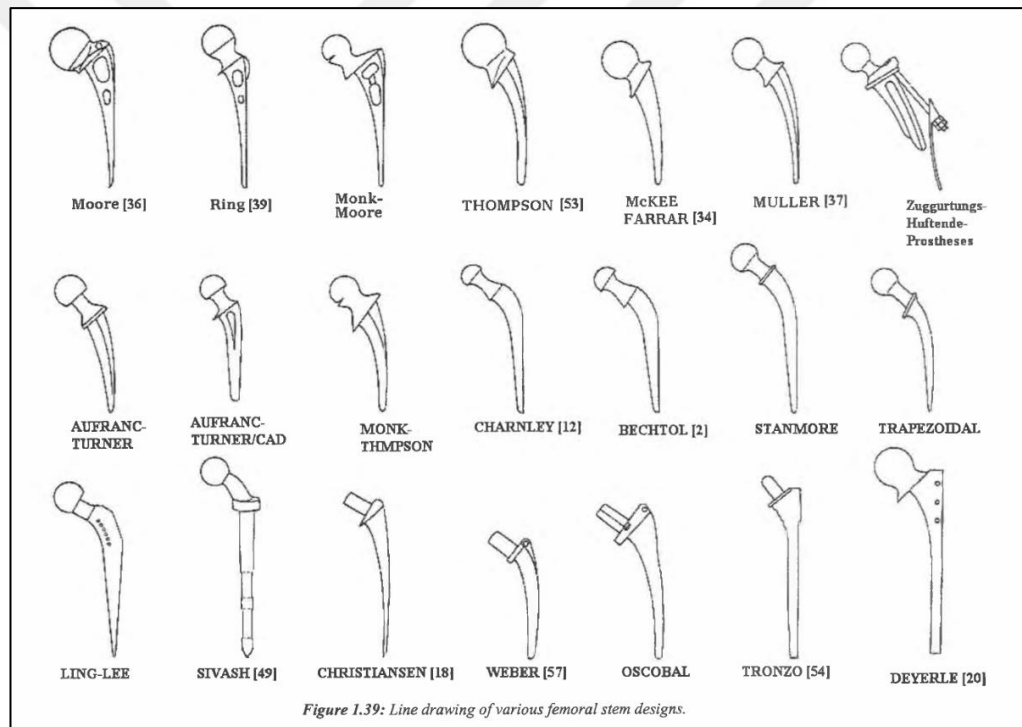


Figure 1.8: Several hip stem designs [9].

1.3.1. Material Properties of Hip Prosthesis

Elastic modulus is a measure of a material's elasticity or deformation. The elastic modulus of materials has a significant impact on the durability, deformation resistance and long-term performance of the prosthesis.

Materials with a lower elastic modulus tend to be more flexible and softer. This can help the prosthesis mimic joint movements naturally and provide a more comfortable experience. However, materials with a low elastic modulus may have less durability and long-term performance.

The mechanical properties of some materials used for hip prosthesis are listed in **Table 1.4**.

Table 1.4: Mechanical properties of materials which is used on hip stem [13].

Materials	Tensile strength (MPa)	Elastic modulus (GPa)
Alloy		
Co-Cr alloys	655-1896	210-253
Co-Cr-Mo	600-1795	200-230
Ti-6Al-4V	960-970	110
Stainless steel 316 L	465-950	200
Polymers		
UHMWPE	21	1
PTFE	28	0.4
Ceramics		
Zirconia	820	220
Alumina	300	380

According to the data in **Table 1.4**, while the elastic modulus of the Co-Cr-Mo alloy is at the level of 200-230 GPa, the stainless steel is 200 GPa, the titanium alloy has the lowest elastic modulus value with 110 GPa.

On the other hand, materials with a high elastic modulus tend to be harder and more durable. Such materials can increase the stability of the prosthesis and provide greater resistance for long-term use. However, harder materials can reduce comfort in some cases, restrict joint movements and increase the tendency for fractures.

In order to examine the femoral stem replacement made of titanium alloy, Eingartner et al. in their study titled 'Results of a cemented straight titanium alloy femoral stem after mean follow-up of 13 years', applied a cemented titanium-supported femoral stem to 239 patients and observed the change for 13 years [14].

For observation results, dysfunction was evaluated using the time from 0 to 100.

Patients with scores above 70 are considered to have good outcomes. When the results were obtained, the risk of aseptic loosening was eliminated by using the titanium body as cemented, and the general overall results were successful.

In order to examine stainless steels in hip prostheses, dos Santos et al.'s study conducted on the femoral root and microstructure made of austenitic stainless steel [15]. When the results are examined, chemically and microscopically small pieces emerge. However, since small differences in grain size can cause precipitation and changes, the distribution of the material is at a level that can affect its resistance and fatigue [15].

1.3.2. Titanium-6Aluminum-4Vanadium Material

Among biomedical materials, titanium alloys for hip prostheses show high compatibility in surgical procedures due to their resistance to corrosion, high strength and lightness when combined with body fluid [16].

Ti-6Al-4V alloy is Grade 5 titanium and consists of 4 wt% vanadium, 6 wt% aluminum and additionally mixed titanium.

Ti-6Al-4V alloy has an α - β composition. According to the phase diagram, the transformation from HCP (α hexagonal close packed) to BCC (body centric cubic) occurs at 980°C.

As shown in **Figure 1.9**, Ti-6Al-4V alloy has lower elastic modulus than pure titanium. The elastic modulus, which is lower than titanium, is reduced by the inclusion of vanadium, thus being in more effective contact with the modulus value of the bone tissue [17]. Detailed phase diagram demonstrated on **Figure 1.10**.

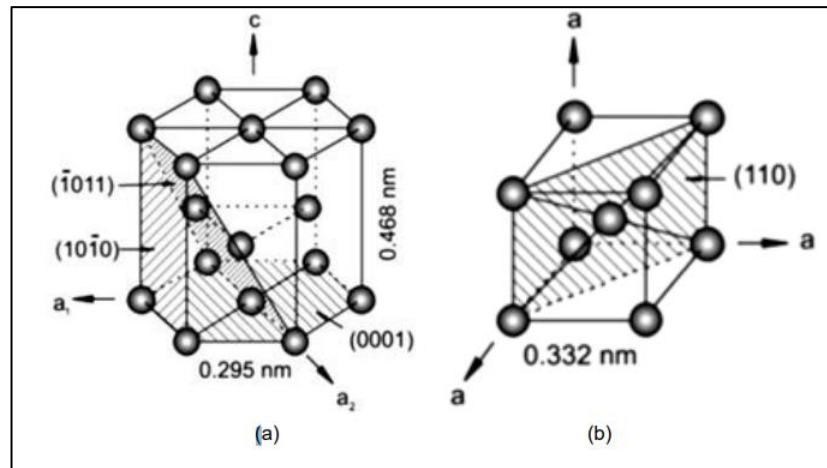


Figure 1.9: Structure of Ti-6Al-4V a) crystal α -HCP and b) crystal β -BCC [17].

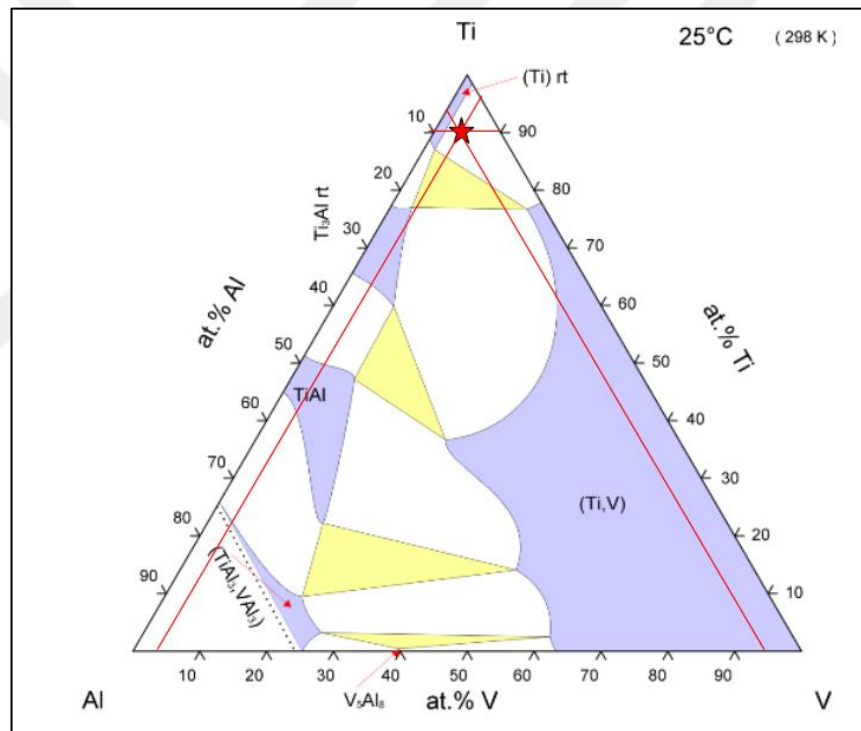


Figure 1.10: Phase diagram of titanium-aluminum-vanadium [18].

Another study (El-Sheikh, 2002, El-Sheikh et al., 2002) on the type of material, which is one of the most important factors in hip prostheses, showed that if titanium alloy is preferred over cobalt chrome alloy, it will have lower hardness and may cause an increase in the stress factor by transferring excess load to the femur [9].

1.3.3. Ceramics

Ceramic materials such as alumina and zirconia are also preferred as an alternative to metals when using femoral heads and acetabular cups in hip prostheses.

Previous studies have shown that the use of alumina as a ceramic on ultra high molecular weight polyethylene material causes 20 times lower corrosion compared to the use of metal [19].

Zirconia is one of the preferred biomedical materials with its breakage resistance and strength. Zirconia, one of the ceramic family, differs from other ceramics with its hardening feature coming from its own microstructure. When Zircon (ZrSiO₄) interacts chemically, pure Zirconia can be formed.

It can be preferred as a component in hip prosthesis, and in this case, 6% moles Y₂O₃ is added in addition to zirconia and it is partially stabilized.

Compared to alumina, it is known that its physical properties are lower than aluminum [20].

Al₂O₃ alumina ceramic is one of the first ceramics preferred in clinical applications due to its high corrosion resistance, biocompatibility and strength.

Noiri et al. applied the alumina-ceramic material on albino rabbits and examined its biocompatibility in the living body. Although there were no problems in the tissue accepting the implant at the beginning, tissue growth and fibroblast formation in the pores were observed at the end of the eight weeks [21].

1.3.4. Fixation Method of Total Hip Prosthesis

There are two types of fixation methods for hip prostheses as cemented and cementless. One of the differences between the two fixations is that cementless fixation has higher stress shielding.

Cemented and cementless fixation models are shown below in **Figure 1.11**.

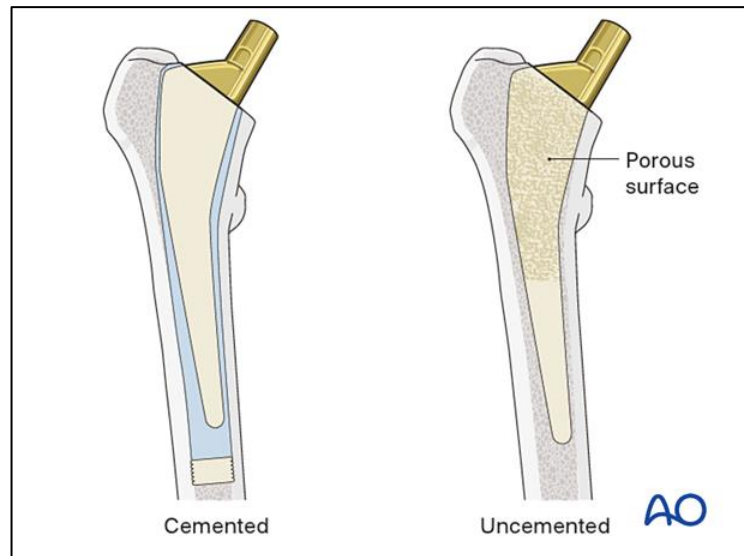


Figure 1.11: Cemented and cementless fixation [22].

While the fixation method is preferred in hip prostheses, it is applied by considering factors such as the patient's condition, age and bone health.

Cemented fixation is a fixation method using acrylic cement to place the implant into the bone. While high stability is achieved in the hip prosthesis with the cemented method, there may be an increase in the rate of wear and breakage due to hardness.

In uncemented fixation, the stem body is placed into the bone by screwing or applying pressure from above. In uncemented fixation, osseointegration occurs over time by the fusing of the bone and the implant. It can provide longer life since the risks of osseointegration and loosening and wear are less than cemented fixation.

1.3.5. Stem Neck Angle and Neck Length

In hip replacement surgery, the neck length and neck angle of the hip stem are crucial parameters that significantly influence the biomechanics, stability, and function of the hip joint.

The neck length of the hip stem is the distance from the center of the femoral head to the axis of the femoral stem. It essentially determines the length of the lever arm created by the femoral neck.

In his inquiry conducted in 2002, El-Shiekh emphasized that by reducing the stem neck diameter, the patient's range of motion increases and the long life of the prosthesis is

supported [9]. In the **Figure 1.12**, the range of motion is expressed when the diameter is long and short.

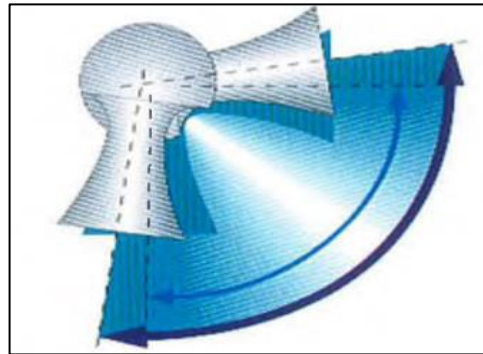


Figure 1.12: Range of neck movement [9].

In a 2021 study by Paul, Lazennec, Patel, Anjeria, Beaule and Schwarzkopf, in which they examined the effect of neck geometries (shown in **Figure 1.13**) in the femoral stem on the movement of the hip, it was defined that the range of motion of the hip prosthesis is the maximum range of motion of the femoral head that the stem neck can do without compression [23].

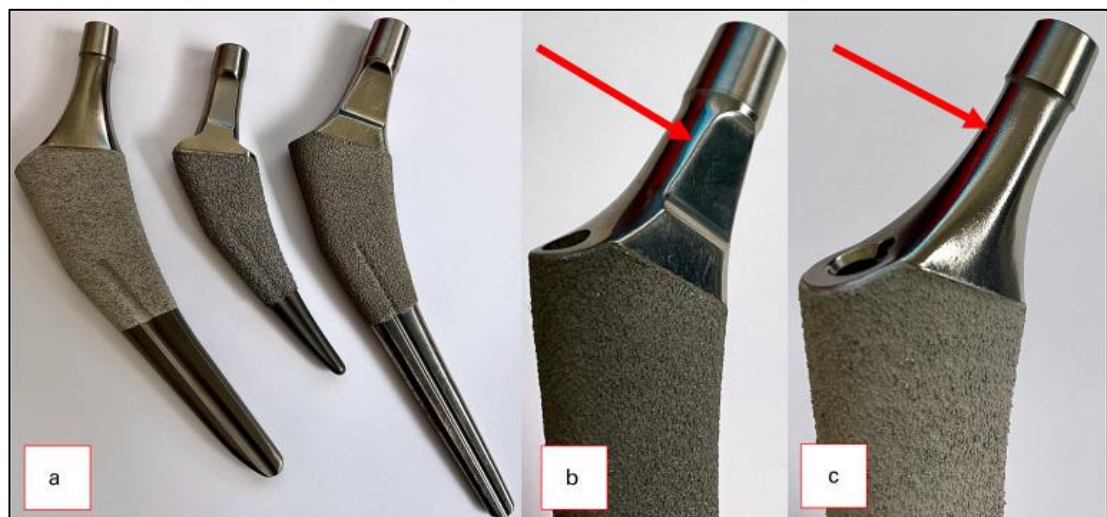


Figure 1.13: Several types of femoral stem a) with different necks, b) trapezoidal neck design and c) cylindrical neck design [23].

1.3.6. Diameter Effect of Femoral Head

Femoral head diameters used in hip prostheses are standardized within certain ranges. Heads generally vary between 22 mm and 44 mm. 22, 26, 28 mm represents small head, 32, 36 mm for medium head and 38, 40 and 44 mm for large head.

Diameter selection varies depending on the patient's age, gender, activity level and anatomical features.

Additionally, the material of the prosthesis used (metal, ceramic, polyethylene, etc.) and prosthesis design also affect this choice.

The 22 mm head diameter used in total hip arthroplasty surgeries in the 1960s grew to 28 mm in the 1990s and to 32 mm in the 2000s.

Usage of 36 mm heads has increased significantly since the mid-2000s. It seems that the most commonly used head diameter today is 32 mm [24].

Figure 1.14 lists the most commonly used head diameters by country in 2015.

The most preferred diameter size after 32mm is 36mm in countries such as Denmark, Germany, and 28mm in countries such as Netherlands and England. Factors in the selection of head diameter may be related to body size, weight and age distribution, depending on genetic factors in the country's population.

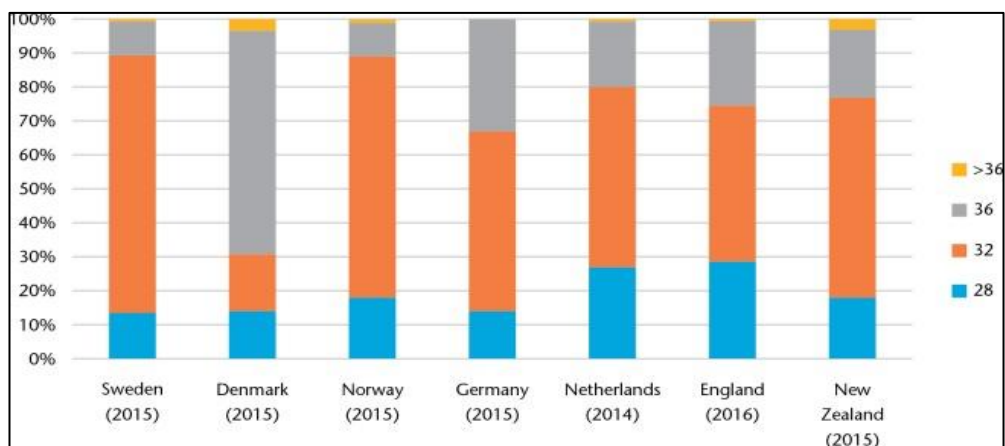


Figure 1.14: Distribution of femur head types based on diameter used in THAs.

Mistry et al. studied “Trunnionosis in total hip arthroplasty” conducted in 2016 and researched studies in the literature to see the effects of femoral head size on trunnionosis [25].

According to the study, the instability of the components decreases and the collision-free range of motion increases with the use of large diameters in the femoral head.

Besides that, as the diameter increases, the horizontal lever arm increases, which increases the torsion force for the head-neck junction [25].

In **Table 1.5**, created because of the research, the number of surgical implant procedures, the implant models used, and the results of the procedures are listed. Accordingly, one of the examples in their study by Tan et al. is that 44 implant operations were performed using a head diameter of 28 mm and corrosion was detected in the trunnion area ($P = 0.018$) [26].

In the study of Brock et al. proceeded with implants of different sizes as 12/14 and 11/13 sizes in the trunnion. According to the results of the research, material loss and corrosion was higher in the short trunnion than long one [27].

Table 1.5: Total hip replacement applications and outcomes [25].

Application	Number of Hip Prosthesis Applied	References	Outcome
28-mm heads with six taper designs	44	Tan et al.	Corrosion scores are increased for on trunnion ($P = 0.018$)
Short 12/14 trunnion versus long 11/13 trunnion	104 female tapers 11 stem trunnions	Brock et al.	Shorter trunnion has higher material loss than longer one (0.402 vs 0.123 mm ³ /year; $P = 0.03$)
Stems released between 1983 and 2012; 10 different taper designs; five metal alloys	85	Porter et al.	Correlation is negative between bending strength (-0.23 ; $P = 0.04$) and length of trunnion (-0.53 ; $P < 0.001$)
Taper diameters: type one (eight hips), 11/13 (six hips), 12/14 (26 hips)	40	Nassif et al.	Lower fretting scores in type one tapers compared with 11/13 ($P = 0.005$)

With the use of a large diameter head on the femoral head, jump distance and head-neck ratio also increase, and the impact range decreases in the range of motion [28].

Ashkanfar, Song Toh, English, Langton and Joyce analyzed the effects of different head diameters of 22 mm and 36 mm in their study of 'The impact of femoral head size on the wear evolution at contacting surfaces of total hip prostheses: A finite element analysis'. The wear rate on the polyethylene surface was examined for 22 mm

and 36 mm head diameters and an increase from 98.6 mm³ to 159.3 mm³ was observed [29].

In addition, there was an increase in the wear rate of the tapered surface of the femoral head from 4.18 mm³ to 4.95 mm³.

Over time, the tendency to use larger diameters in the femoral head in hip replacement surgeries has increased.

According to the results of the National Joint Registry of England, Wales and Northern Ireland, while the use of 36-mm heads in total hip arthroplasty surgeries performed in 2005 was 5%, an increase of 26% in 2009 and 35% in 2011 was observed [30].

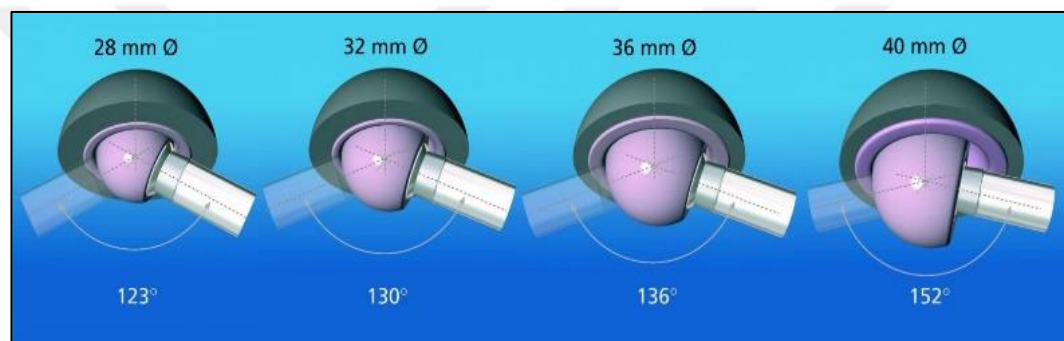


Figure 1.15: Ranges of motion of hip joint for different head diameters [31].

As it seen on **Figure 1.15**, larger femoral heads increase the range of motion.

1.3.7. Biocompatibility of Hip Prosthesis

Choosing of biocompatible material is crucial during and after use that the materials in the prostheses placed in the human body are not inflammatory or allergic.

However, primary biocompatibility depends on the biocompatibility of the materials as well as the reactions that will occur when human tissue and the implant come together [32]. Factors affecting biocompatibility in the implant are the host reaction of the material and degradation within the body. Various reactions may occur when the implant surface comes together with fluids and tissues in the human body [33].

2. LITERATURE VIEW

2.1. Effects of Frequency and Vibration on Hip Implants

In recent years, many studies have been conducted to determine the natural frequencies created by the femur bone. One of the studies carried out to determine the natural frequency and mode shapes of the femur bone was the study "Vibrational characteristics of the embalmed human femur" by Khalil et al [34].

In this study, while experimental data were analyzed with Fourier analysis, an analytical solution was made using 59 elements with the transfer matrix method. While it was estimated that the bone was vibrating freely at the beginning of the study, as a result of the study, the first 20 natural frequencies were examined for the free class condition in the femur bone and it was found that it varied between 250-7300 Hz [35].

Based on the evaluation of natural frequency, it was in a similar range.

Implicit FE coding is applied to determine the modal outputs of the femur bone by performing modal analysis with finite element analysis. The dynamic response equation (2.1) is as follows:

$$[M]\{\ddot{x}\}+[C]\{\dot{x}\}+[K]\{x\}=0 \quad (2.1)$$

where $[M]$ is global mass, $[C]$ is dampig matrice and $[K]$ is stiffness matrice of the model.

Nodal acceleration, velocity and displacement vectors are shown below, respectively.

$$\{\ddot{x}\}, \{\dot{x}\}, \{x\}$$

If $[C]=0$, undamped free vibration, solution of the equation is:

$$\{x\}=\{X\}e^{i\omega t} \quad (2.2)$$

where $\{x\}$ represents all masses amplitudes.

2.2. What is trunnionosis?

Trunnionosis is a wear that occurs in the part between the femoral head and stem neck due to factors such as loosening, friction and strain. In 1998, Cales and Stefani showed the effects of changes in trunnion design on trunnionosis and found that the design could have an effect on trunnionosis depending on the torsional forces between the trunnion and taper [36].

To analyze the parameters that cause trunnionosis, Lavernia et al. inspected the effects of femoral head diameter on wear in 2015. As a result of the finite element analysis, a regular increase in stress values occurred with the increase in head diameter [37].

In the study 'Finite Element Analysis on Vibration Modes of Femur Bone' conducted by Gupta and Tse in 2013, they examined the vibration acting on the femur bone and the resulting wear through experimental and finite element analysis [35].

As a result of the experimental test performed with the hip simulator, the appearance in the trunnion area is shown in **Figure 2.1**. As shown in a) and c), wear and mass loss occurred in the proximal superior and distal inferior.

Figure 2.2 shows the debris from wear inside the head component in contact with the trunnion.

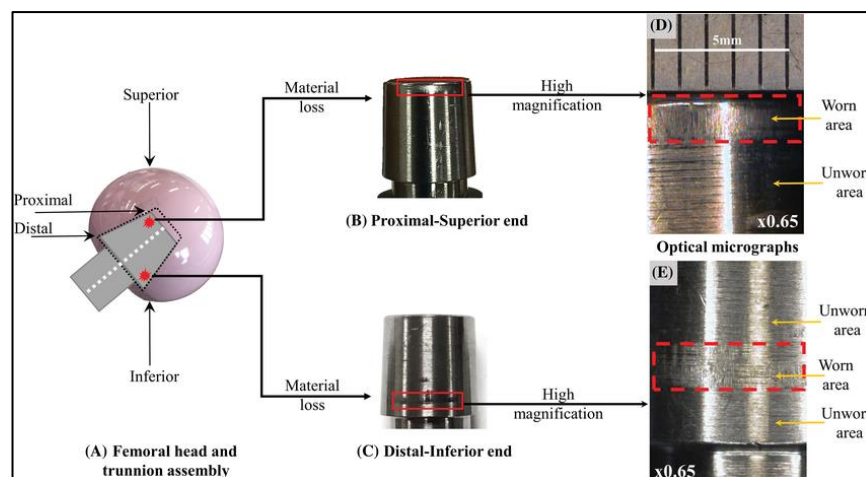


Figure 2.1: Trunnionosis on trunnion, showing with a) structure of neck into head, b) wear on proximal superior end and c) wear on distal inferior end plane.

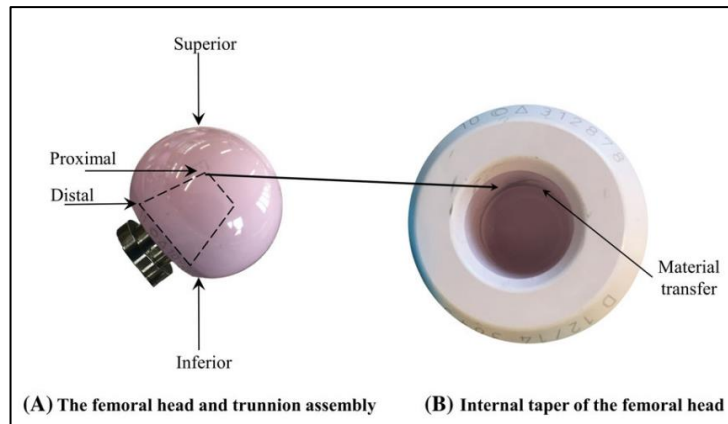


Figure 2.2: Trunnionosis-based material transfer.

2.2.1. Wear Classification

Surfaces are considered tribologically, all rough. While determining the deformation, the roughest protrusions on both contacting surfaces come into contact with each other and deformation occurs.

The actual contact area of the contacting surfaces is less than the total area of the two surfaces. Breakage may occur due to irregularities on the surfaces that provide actual contact. After breaking, particles come out of the material. If additional plastic deformation has occurred, the particles coming out between the contacting surfaces may be transferred to each other [38].

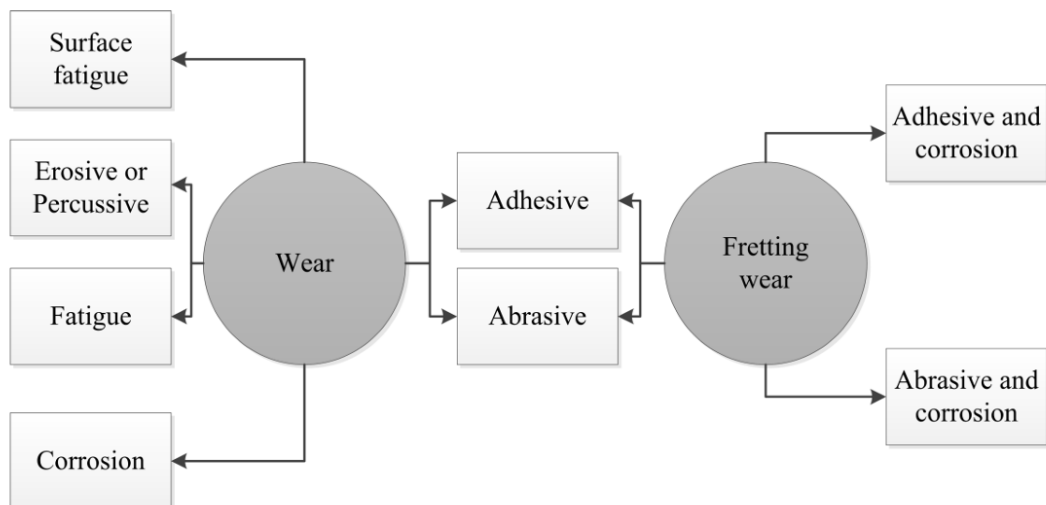


Figure 2.3: Wear classification and fretting wear [29].

The wear classifications and fretting wear connections in the **Figure 2.3** are explained with the following explanations and formulas.

In adhesive wear, when the material is worn, it becomes harder by welding and transferring the rough edges of the surfaces. After that, a fracture occurs and it moves away from the surface.

In corrosion wear, a film layer is formed while environmental reactions that cause the material to wear occur. Corrosion wear can be calculated using the following Equation 2.3 [25][39].

$$W_v = \frac{\lambda}{2a} n \pi a^2 L = \frac{\lambda}{2a} A_r L = \frac{k_{cor}}{H} F_n L \quad (2.3)$$

where λ is film of thickness, $\pi a^2 \lambda$ is volumetric wear loss, a is radius of a hemispherical asperity shape and k_{cor} is corrosive wear coefficient.

Abrasive wear occurs as a result of the deformation of the material by a material that is harder than itself (**Figure 2.4**). W_v volume loss can be calculated by using Equation 2.4.

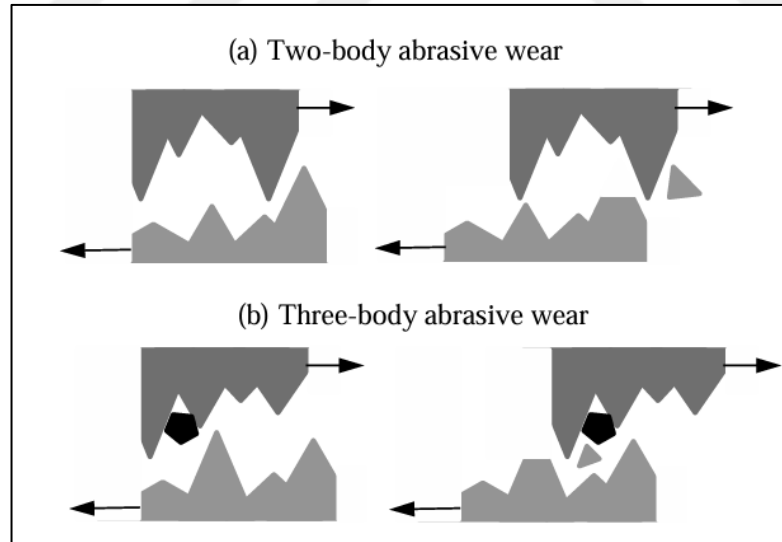


Figure 2.4: Abrasive wear which is a) two-body abrasive wear and b) three-body abrasive wear

$$W_v = \frac{k'}{H} \left(\frac{2 \overline{\tan(a)}}{\pi} \right) F_n L = \frac{k_{abr}}{H} F_n L \quad (2.4)$$

In erosive wear, particles hit the surface and cause wear. In brittle materials, particles are separated by breaking, while in ductile materials, they are separated by cutting.

A low rate of erosive wear occurs in the total impacts (ς) acting on the material. In Equation 2.5, U indicates the impact speed, m indicates the total mass, and (e_r) indicates the erosion rate.

$$e_r = \varsigma \frac{\rho U^2}{2H} \quad (2.5)$$

$$E_k = \frac{1}{2} m v^2 \quad (2.6)$$

Impacting energy can be calculated with Equation 2.7. It occurs as a result of impact energy wear.

$$E_i = \varsigma n E_k = \varsigma n \frac{1}{2} m v^2 \quad (2.7)$$

In fatigue wear, cracks may occur due to time or elasticity, and wear particles may form as they grow over time [29].

Surface fracture wear occurs when the stress level due to the forces acting on the material exceeds the ultimate stress level, and the particles break away from the surface.

While K_c expresses the fracture toughness, the volume loss per unit distance in the Equation 2.8 is based on the volume loss on N asperities rubbing on a surface with K_c toughness [40].

$$W_v = cN \left(\frac{(\frac{E}{H}) F_n^{9/8}}{K_c^{1/2} H^{5/8}} \right) F_n L \quad (2.8)$$

where ρ is density of material, $kabr$ is abrasive wear coefficient, k' is constant coefficient and Wv is volume loss.

In hip prosthesis, wear and tear may occur in the joint areas due to slipping. Calculation can be made by substituting the wear factors with Archard's wear equation and the result can be evaluated [41].

$$W_L = K_w P S \quad (2.9)$$

where P is contact stress, W_L is linear wear, K_w is wear coefficient, S is sliding distance.

Reginal et al. investigated the wear rates in the area between the femoral head and trunnion in their study. Wear rates are measured linearly and volumetrically in mm/year (Equation 2.10) and mm^3 /year (Equation 2.11), respectively.

The contact surface of the implant head and the trunnion is 14 mm, and the part above the trunnion is 12 mm.

In the study, wear rates for 1 million cycles were formulated as follows, allowing linear and volumetric wear calculations.

$$\text{Linear: } W_L = K_w P * S * 10^6 \text{ (mm/}10^6 \text{ cycles)} \quad (2.10)$$

$$\text{Volumetric: } W_L = K_L * A * 10^6 \text{ (}mm^3 \text{/}10^6 \text{ cycles)} \quad (2.11)$$

Friction coefficient (COF) and wear coefficient (K_w) values of Cobalt-Chrome, Titanium and UHMWPE material combinations are given in **Table 2.1** below.

Table 2.1: Wear coefficient and coefficient of friction of materials [42].

Material Combination	Coefficient of Friction (COF)	Wear Coefficient (K_w) (mm^3/Nm)
Cocr-Cocr	0.2	$1.68e^{-5}$
CoCr-Ti alloy	0.24	$1.31e^{-5}$
Cocr-UHMWPE	0.15	$3.5e^{-7}$

The basic wear measurement for trunnionosis is the volume of worn material proportional to the sliding distance [38]. It can be calculated using the Equation 2.12.

$$V = W_v / S \quad (2.12)$$

where V is volume of worn material.

2.3. Fatigue Life and Stress Calculation

Mean stress (σ_m) and alternating stress (σ_a) can be calculated by using the Goodman Mean Stress hypothesis [42].

$$\sigma_m = \frac{\sigma_{\max} + \sigma_{\min}}{2} \quad (2.13)$$

$$\sigma_a = \frac{\sigma_{\max} - \sigma_{\min}}{2} \quad (2.14)$$

In general, according to Goodman theory, the fatigue strength of a material is expressed as follows:

$$\frac{1}{N_f} = \frac{\sigma_a}{S_e} + \frac{\sigma_m}{S_u} \quad (2.15)$$

where N_f is safety factor, S_e is endurance limit and S_u is tensile strength.

In other words, the wear rate or wear coefficient (k) is expressed as the unit of volume worn per unit sliding distance of the surfaces in contact with each other [38].

$$k = \frac{V}{A_r} = \frac{W_v / L}{F_n / H} \quad (2.16)$$

where k is wear factor, V is wear volume, n is number of cycles, L is applied load and s is sliding distance.

Cross shear movement or multi-directional movement is one of the wear factors affecting the hip prosthesis [43].

Under normal conditions, strain hardening occurs when linear movements occur on the prosthesis. This increases the material's resistance to corrosion.

On the other hand, when multi-directional movement occurs, multi-directional tension and compression forces occur on the prosthesis.

Under these forces occur, it can create wear and corrosion vertically, while strain hardening occurs on one side [44][45].

In cases where there is wear on the material, particles released during wear may remain in the roughness area or fracture due to slip may occur [46].

In addition, Karmakar et al. in 1996, showed that subsurface fractures may occur when cyclic load occurs in ductile materials. In case of fractures, fatigue wear may persist on the contact surface [47].

Fatigue wear can be found with the Equation 2.17 [47].

$$W_v = k_{fat} A_r L = \frac{k_{fat}}{H} F_n L \quad (2.17)$$

k_{fat} specified in the equation can be related to the number of cycles or stress cycles until N_f failure occurs (Eq. 2.18).

$$N_f = \frac{1}{k_{fat}} \quad (2.18)$$

In addition to the stress on the implant due to movement and acting forces, micro movements may occur on the surfaces in contact and become permanent over time. In this case, fatigue strength decreases while wear and deformation increases.

Since wear-related slips are tangential to the surface, trunnionosis increases corrosion. Donaldson et al., in their 2014 study on micromotions that cause wear and corrosion on the implant, showed that there was an increase in micromotion with the increase in head offset (Figure 2.5) [48].

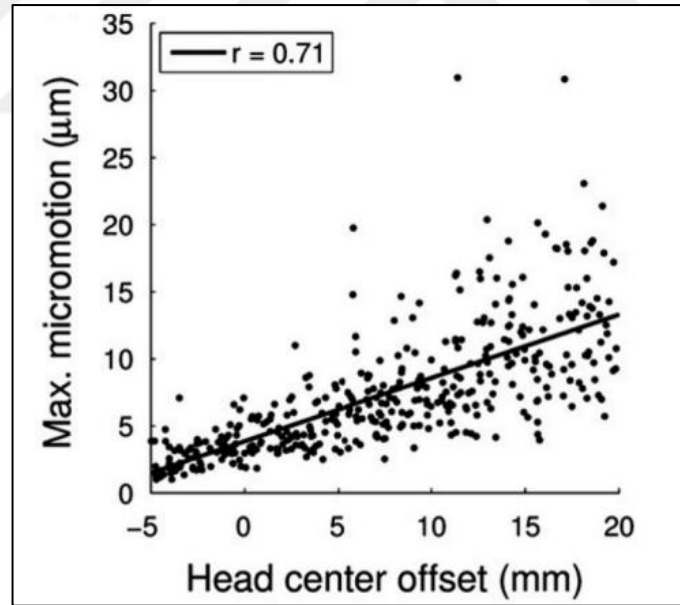


Figure 2.5: Head center offset versus micromotions [48].

Using finite element analysis, the safety factor and fatigue life of hip prostheses under different material types and forces acting during walking can be found by Goodman mean stress fatigue. The alternative stress-life curve S-N curve used in study by Belwanshi et al. is mentioned on **Figure 2.6** [49].

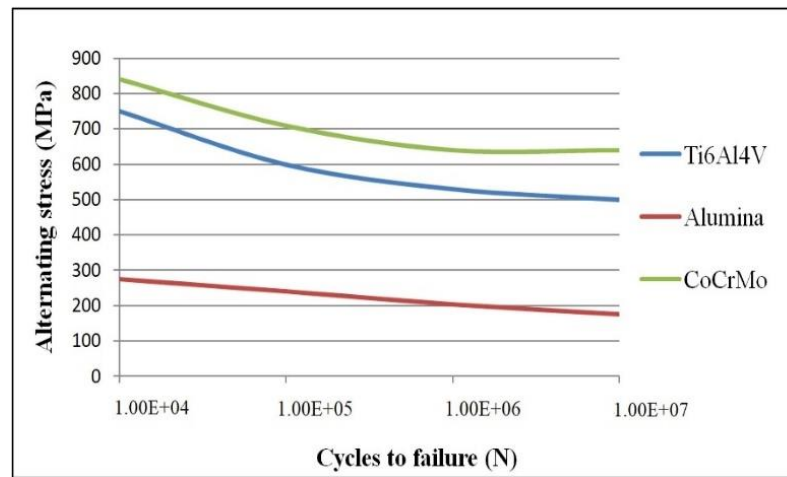


Figure 2.6: S-N curves of Titanium Alloy, Alumina and CoCrMo [42].

3. METHODS AND MATERIALS

3.1. Modelling of 3-D Hip Implant Design

Static and dynamic analyzes were performed with the finite element method in hip prosthesis. Models of different diameters and stem body were modeled using CAD program in order to see the effects of different head diameters and neck angle. Autodesk Fusion 360 program was used during modeling.

According to the analysis results for hip prosthesis, 2 stem models, 1 acetabular cup (Figure 3.1) and 5 different femoral heads were modeled for different diameter head sizes of 28, 32, 36, 40 and 44 mm in order to evaluate the wear factors. In these models, the offset gap within the femoral head was accepted as 0.

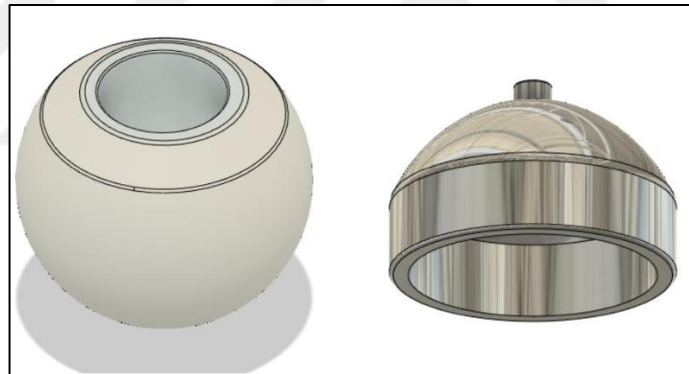


Figure 3.1: Modeling of femoral head and acetabular cup.

For the offset gap, which is another factor that may have an effect on the wear in the trunnion area, in addition to the fixed 0 size of the femoral head, a design was made with a -3 mm forward and +3 mm fully conical trunnion area located inside the head. A total of 10 different models were modeled in these designs for each head.

Figure 3.2 shows sample works for 32, 36 and 40 mm with -3, 0 and +3 mm offset gap.

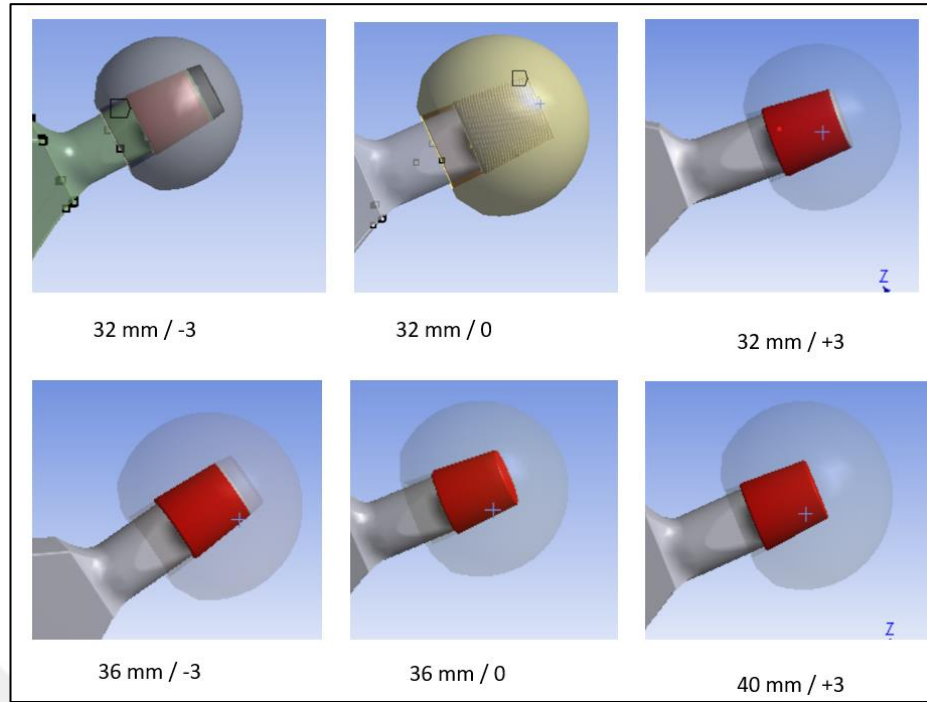


Figure 3.2: Different head diameter and offset gap configurations: 32, 36 and 40 mm head and -3, 0 and +3 mm offset gap.

The standard stem body was designed for the hip stem by giving the dimensions in **Figure 3.3**. The stem length from pointy end to trunnion is accepted as 140 mm and the angle from the trunnion to the curved starting point of the stem is accepted as 127°.

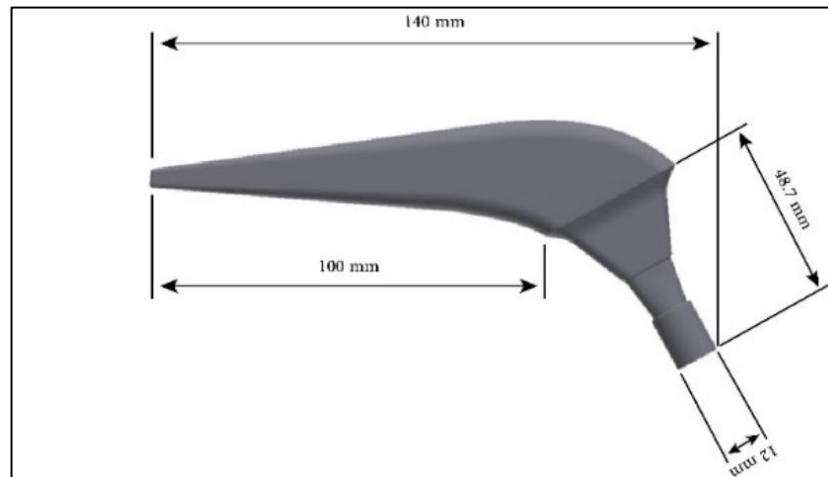


Figure 3.3: Stem component of hip implant dimensions used in this study.

In order to analyze the wear and stress factors that different offset angles between the trunnion and stem may create in the trunnion region, additional to the accepted

standard design with 127° offset angle between curve point to trunnion in this current study, a new 132° stem design was modeled.

In this design, accordance with change of angle, the offset length was reduced by -5 mm to see the effect of the reduced offset.

3.1.1. Material Properties of Modellings

The biocompatibility of the implant to be used in THR with the human body and the resistance of its material properties to breakage, wear and infection are crucial. Accordingly, titanium alloys and cobalt-chromium alloys are mostly preferred in THAs.

Titanium Alloy (Ti-6Al-4V) for the stem body and Zirconia (ZrO₂) for the femoral head were used in this study. The mechanical properties of the materials are given in **Table 3.1**.

Table 3.1: Material properties of hip implant used in the study [33].

Material	Common Name	Young Modulus (GPa)	Poisson Ratio
Ti-6Al-4V	Titanium Alloy	105	0,35
ZrO ₂	Zirconia	160	0,3

Titanium Alloy is considered linear elastic and isotropic.

One of the important parameters in hip implants is fatigue analysis. In fatigue analyses, the properties of the material are at the forefront. Based on these properties, S-N Curve was defined for Ti-6Al-4V shown in **Figure 3.4** and Zirconia shown in **Figure 3.5**, and static and dynamic analyzes were performed on ANSYS 19.2.

Fatigue properties of Titanium and Zirconia materials are shown with S-N curves and stress versus number of cycles are indicated.

In the analysis results, force reactions, vibration effects, stress deformation effects, and wear due to sliding were examined in order to examine the corrosion failures in the trunnion.

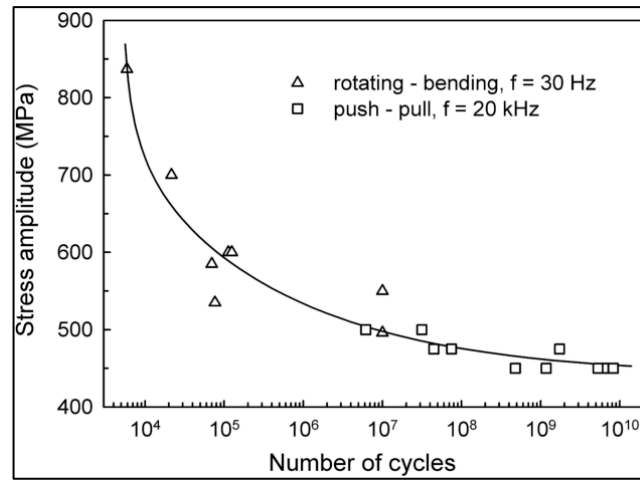


Figure 3.4: S-N curve of Ti-6Al-44V which is used for stem [50].

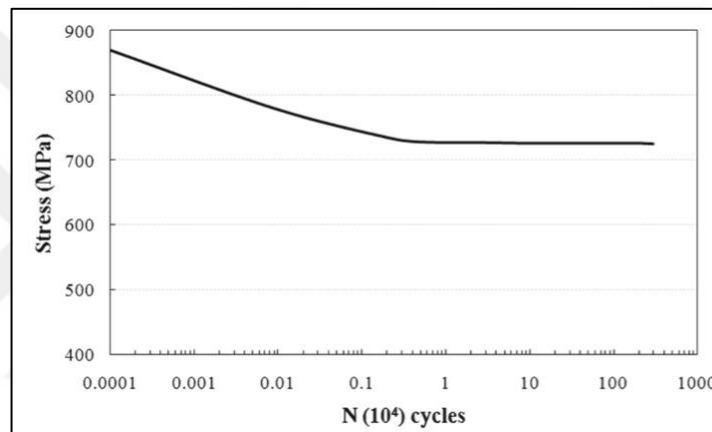


Figure 3.5: S-N curve of Zirconia (ZRO2) which is used for femoral head [51].

3.2. Finite Element Analysis, FEA

Finite Element Analysis, FEA, is a numerical method used to analyze the behavior of complex structures in a static and dynamic sense, to perform simulations to see the effects of different conditions such as impact and load addition, and to test parameters to ensure optimization in designs. FEA works by breaking down the geometry of a structure or system into a finite number of smaller, simpler elements. These elements are analyzed using physical laws defined by differential equations.

The use of FEA in hip prostheses plays a critical role in the design and performance evaluation of the prosthesis. Some of the studies that can be analyzed and performed on hip replacement using FEA are listed below.

3.2.1. Design Optimization

FEA helps select the most suitable material for the prosthesis by simulating the mechanical properties of different materials.

FEA shows how the shape and structure of the prosthesis can be optimized. This is important to reduce the weight of the prosthesis, increase its durability and ensure biomechanical compatibility.

By analyzing the stresses and strains on the prosthesis with Stress and Strain Analysis, it determines the weak points of the prosthesis and possible fracture or wear areas. This allows necessary improvements to be made during the design process.

Motion and Function Analysis

It evaluates the compatibility of the prosthesis with natural joint movements by simulating the movements and functions in the joint where the prosthesis is placed. This is critical to increase the lifespan of the prosthesis and patient satisfaction.

Biomechanical Compatibility

FEA analyzes the adaptation of prosthesis to the biomechanical environment in the human body. This helps understand how the prosthesis interacts with bone, muscle and other soft tissues.

Safety and Durability

FEA evaluates the long-term durability and safety of the prosthesis. This helps determine how the prosthesis will perform over time and detect possible complications in advance.

3.2.2. Meshing, Contact and Boundary Conditions in Finite Element Analysis

In finite element analysis, meshing is the process of dividing the geometry to be analyzed into small and manageable subregions, that is, elements. These subregions are defined by nodes and elements.

The elements consist of simple geometric shapes (triangles, quadrilaterals, tetrahedrons, hexahedrons, etc.) and the behavior of the total system is analyzed by solving physical equations on each of these elements.

In this study, different mesh models were made several times with different numbers of elements to find the most accurate method. For the most optimal values, meshing was performed using approximately 90,000 elements and 130,000 nodes. For the mesh quality, Skewness results were evaluated and concluded as fine mesh.

Using ANSYS 19.2, degrees of freedom in x, y and z directions were defined using the finite element method. SOLID187 and CONTA174 contact element and TARGE170 target element were defined in these regions to examine the effects on the contact surfaces between the trunnion conical region at the stem tip and the femur head.

Since there is a tight and interconnected contact on the contacting surfaces under general usage conditions, the contact type was chosen as Bonded Contact to reflect reality.

For fixation, which is one of the supporting factors of ensuring stability within the body and longevity in hip prostheses, the intramedullary fixation method has been preferred and fixed support has been defined from the stem end. Fixed support areas are as shown in the **Figure 3.6**.

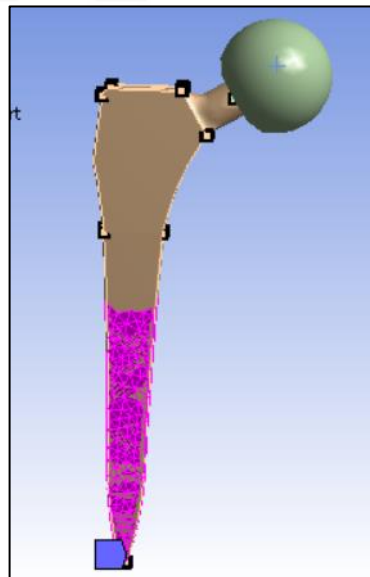


Figure 3.6: Fixation of hip implant.

In the fixation specified in Figure 3.6, the stem end is fixed to cover approximately half of the total stem length.

In addition to this fixation method, two additional different fixation models were made and analyzed in the study to see the effects of different fixation types on trunnionosis. Fixation 2 specified in Figure 3.7 covering more than half of the body and Fixation 3 covering the entire body.

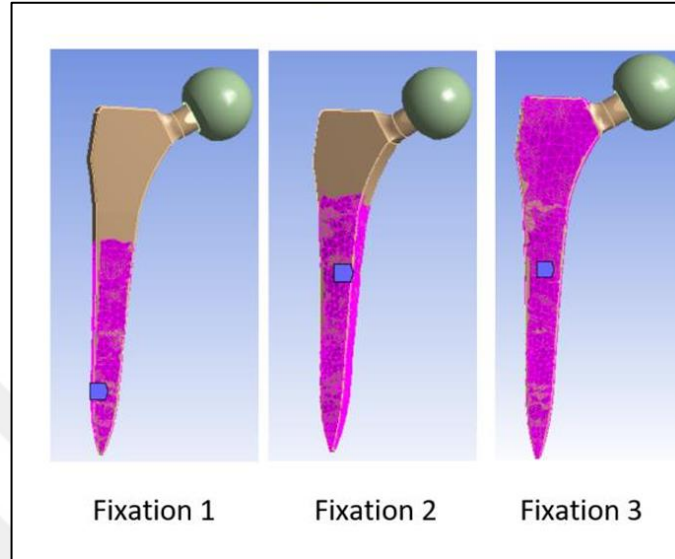


Figure 3.7: Different fixation types in study.

3.2.3. Loading Conditions of Bone Cement and Hip Prosthesis

Finite element analysis in hip prostheses is performed to simulate the forces and moments that the prosthesis will be exposed to in real life conditions and real loading conditions are used. Correct loading conditions play a critical role in evaluating the performance, durability and reliability of the prosthesis.

The prosthesis is subjected to various loads during different activities while inside the body. Accurate simulation of the forces applied to the prosthesis during daily activities such as walking, running, sitting and standing is supportive in understanding the real-world performance of the prosthesis.

Forces acting on the hip joint are shown on **Figure 3.8** and **3.9**.

In the study of 'Finite element modeling and analysis of a new cemented prosthesis' conducted by Kayabasi and Erzincanlı in 2005, they analyzed the forces acting on the hip prosthesis under static and dynamic loads [52].

It was stated that static analysis can be done based on body weight, while dynamic analysis can be done with a load around 10-20% more than body weight.

For a 70 kg patient, for 3 kN body weight, 1.25 kN abductor force and 250 N ilio tibial force were applied.

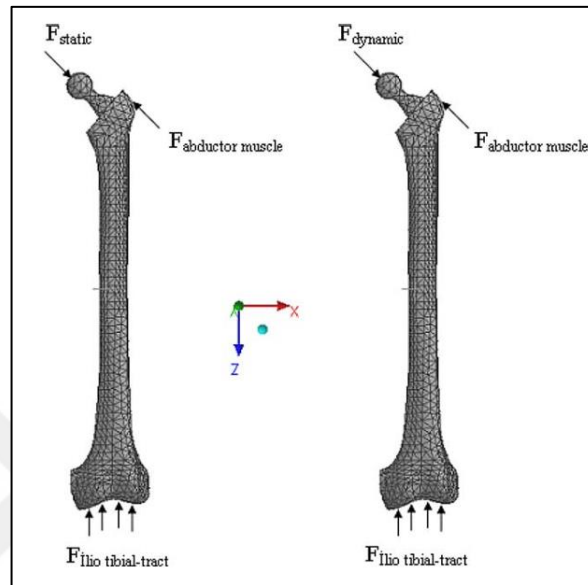


Figure 3.8: Forces such as F_{static} , $F_{abductor\ muscle}$ and $F_{ilio\ tibial-tract}$ acting on bone cement [52].

As it shown in Figure 3.9, hip joint force and abductor muscle force are applied in opposite directions at 20° angles.

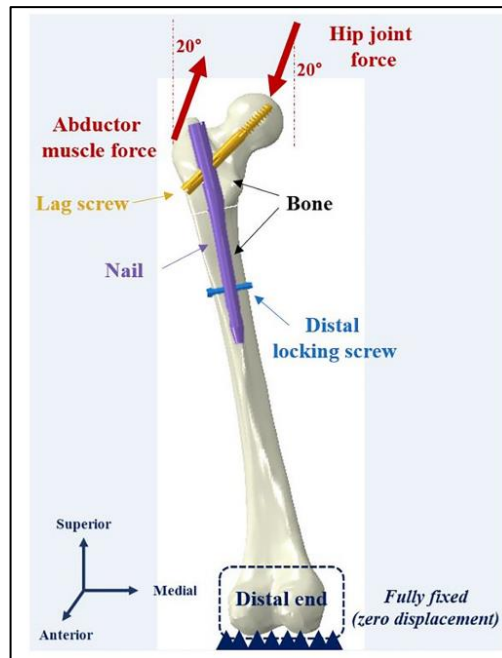


Figure 3.9: Loading conditions of hip joint.

Bergmann et al. investigated the real-time measurement of the direction, moment and magnitude of the forces applied by telemetering hip prosthesis in 1993.

In the study, implants were placed in the right and left hips of an 82-year-old patient with a body weight of 650 N. It is assumed that 30 months after THA, the patient can run at 8 km per hour and is healthy.

With the prosthesis placed on the left side of the patient, approximately 3 mm of restoration was achieved in the frontal plane, depending on the position of the existing femur, while with the implant on the right side, the femur became 1 cm more distal and 1,3 cm more lateral. Peak power was determined as 550% BW for a patient who could walk and run very fast using telemetering total hip prostheses.

For another patient walking at 3 km per hour, peak force was determined as 410% BW. Looking at the torsional moments, it was 40.3 Nm for the first active patient, while 24 Nm was measured for the second, less active patient.

As a result of the study, it was predicted that sudden forces may have an effect on the loosening of the implant.

3.2.4. Walking Dynamic Loading

5-second time-dependent simulation of the forces acting on the hip bone and therefore the hip prosthesis during walking is shown in **Figure 3.10**. In the study carried out by El'Sheikh et al. in 2003, the acting forces were stated as F_x , F_y and F_z [9].

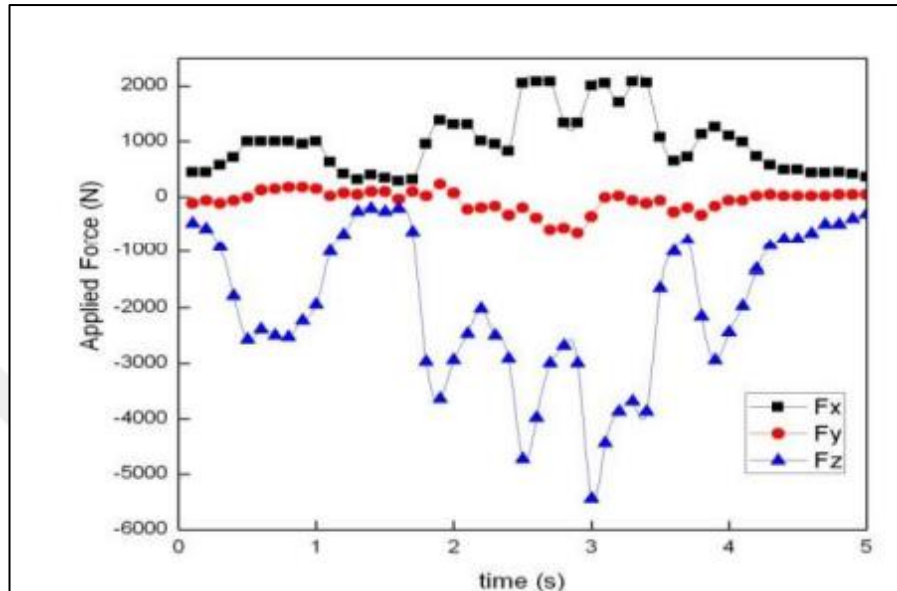


Figure 3.10: Dynamic loading conditions of hip bone during walking.

According to the Bergmann et al. study in 2001, hip rotations based on flexion-extension, internal-external and abduction-adduction are investigated (Figure 3.11) [53].

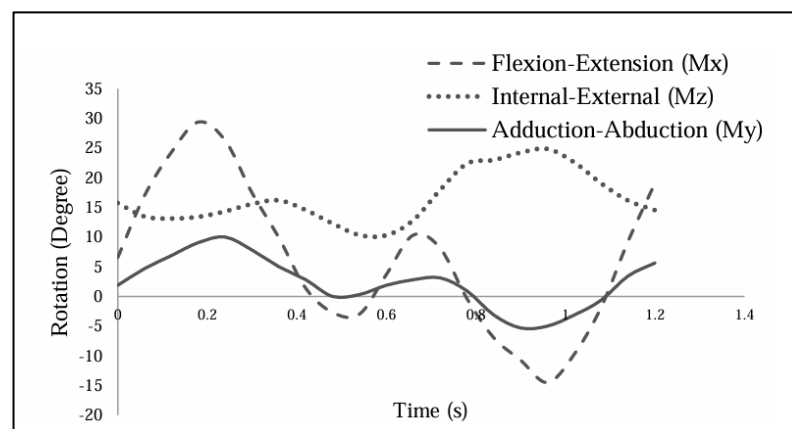


Figure 3.11: Hip joint rotations based on flexion-extension, internal-external and adduction-abduction during walking condition.

In the schematic representation of hip rotation (Figure 3.12), the flexion-extension movement in the medial-lateral plane, internal-external movement in superior-inferior plane and abduction-adduction movement occur in the anterior-posterior plane.

The outer surface of the femur head is positioned with the center of the head and is fixed in x, y and z planes, allowing only rotation.

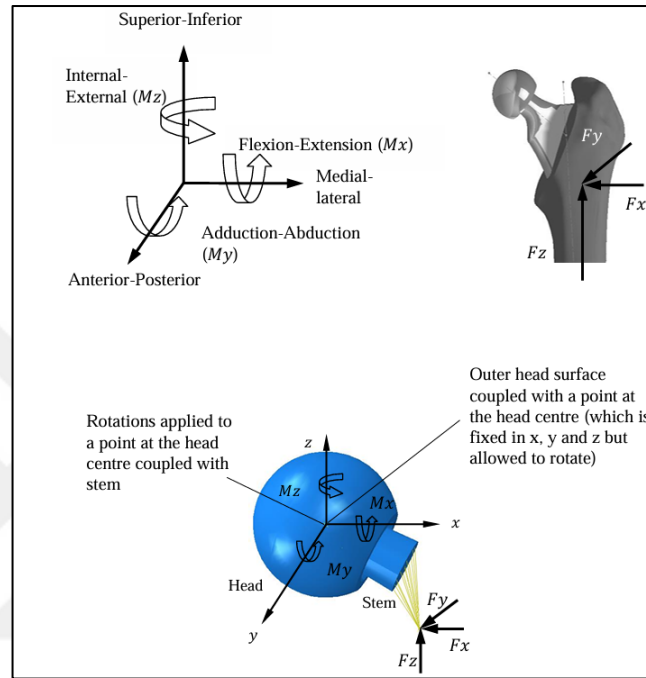


Figure 3.12: Loadings and boundary conditions of hip femoral head.

To simulate the static and dynamic analyses, study was carried out based on a 70 kg patient. While a 2506 N load (A) was applied to the femoral head at a 20° angle, 1250 N abductor muscle load (B) was applied to the stem as a remote force at a 20° angle.

In addition, a load of 250 N was applied to the stem tip as ilio-tibial load (C). The applied loads are shown in **Figure 3.13**.

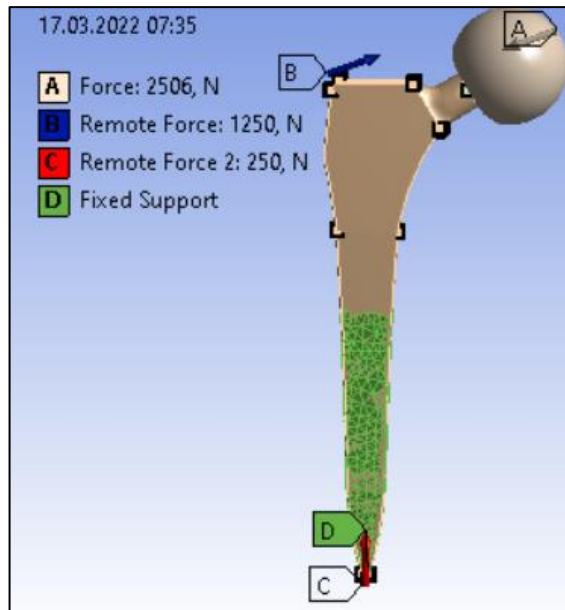


Figure 3.13: Loadings and boundary conditions used in study.

During the entire study, 175 simulations were carried out with finite element simulation and were made using ANSYS 19.2 solver 16-bit Intel Core i7 operating system.

For the dynamic and static analyzes of the 16 simulations, 32 CPU hours were given for some of them and 12 hours for some of them.

4. RESULTS

The analysis results performed with FEA were evaluated separately depending on the changing parameters.

4.1. Frequency – Vibration Effects

Modal Analysis, a type of dynamic analysis, determines the natural frequencies at which a structure will resonate.

By performing modal analysis, resonance frequencies were determined and vibration mode shapes were obtained. During the analysis, the frequency range of 0-10,000 Hz was used based on previous studies (Darwich et al., 2020) and literature information [42].

As a result of the analysis, 10 resonance frequencies in the range of 0 - 10 kHz were obtained. The results obtained were evaluated depending on head diameter, offset opening, offset angle and fixation.

4.1.1. Head Size and Offset Gap Configurations

The frequency values obtained for offset gaps were compared, taking the same head diameter of 36 mm as reference. While the highest frequency increase occurred at 3m, this increase was the least at +3 mm (Figure 4.1).

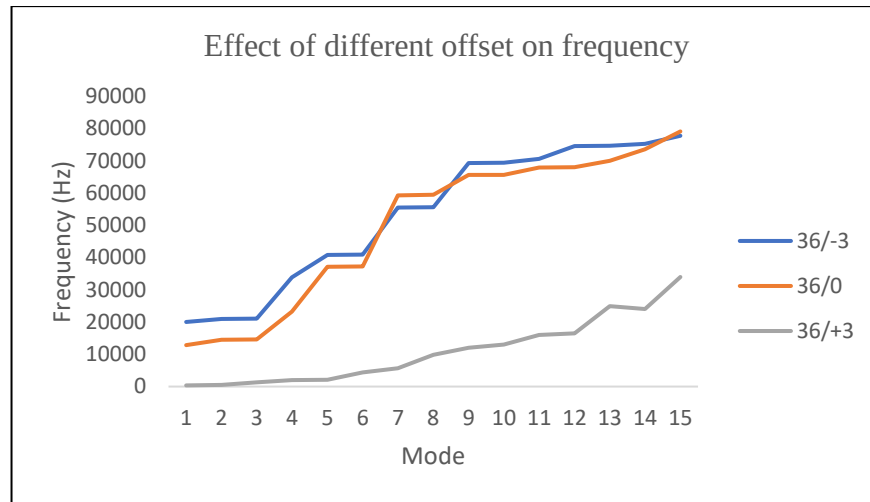


Figure 4.1: Effect of different offset dimensions on generated frequency of modal analysis.

Effect of different head diameters were evaluated with the same offset gap, the highest frequency occurred in the smallest diameter (Figure 4.2). The frequency value gradually increased in each mode. A similar increase rate occurred for all head diameters.

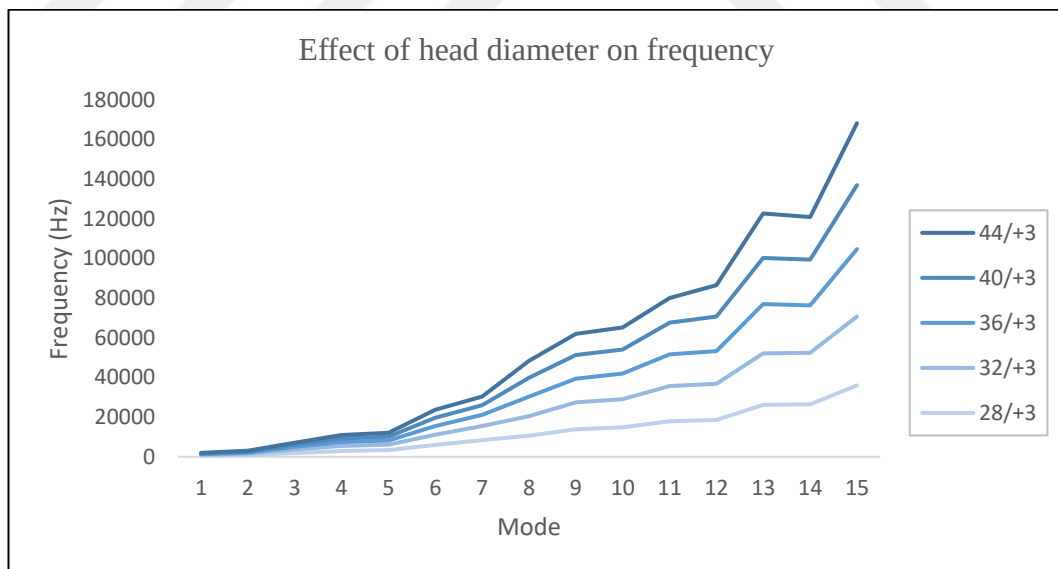


Figure 4.2: Effect of head size on generated modal frequency.

4.1.2. Effect of Offset Angle of 127° and 132°

In 127° and 132° offset angle designs, 32 mm head that has +3 mm offset gap was used and the frequency values were analyzed for 15 modes. Looking at the frequency results on Figure 4.3, obtained frequencies were parallel.

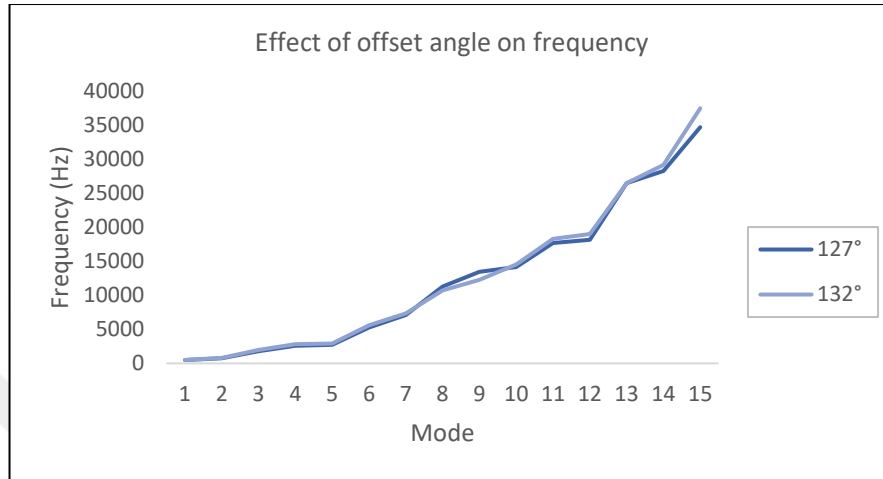


Figure 4.3: Modal analysis results for different hip implant offset angle: 127° and 132°.

4.1.3. Fixation Types

As shown in Figure 4.4, the fixation area from the stem end was increased and modal analyzes were performed to see the effects on vibration and frequency with increased fixation. Based on the results, as expected, the frequency values increased as the fixed surface increased.

As indicated in the graph, the frequency increase in Fixation 1 and 2 increased parallel to each other. With Fixation 3, the tightness increases with the complete implantation of the stem, which also increases the rate of increase in frequency compared to other fixations.

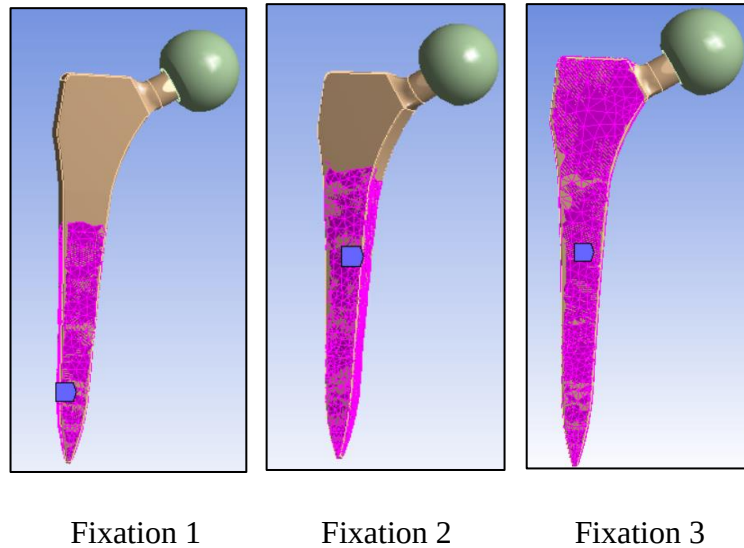


Figure 4.4: Different fixation types.

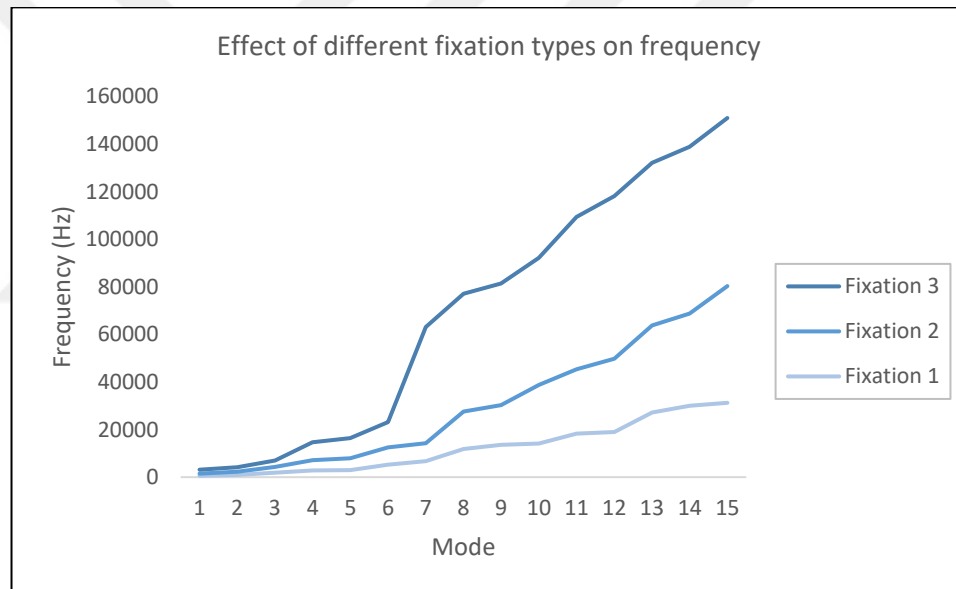


Figure 4.5: Effect of different fixation methods on modal frequency.

4.2. Moment Reactions

In order to evaluate the wear factors that may occur due to the material strength on the materials, a simulation was carried out based on the forces acting on the hip prosthesis during walking. By simulating a 5-second walking moment with transient finite element analysis, the effects of different head diameters, different offset measurements and fixation at the time of walking were examined.

By adapting Goodman's Theory, the safety factor (N_f) value can be calculated according to the alternating stress, mean stress, endurance limit (S_e) and ultimate tensile strength (S_u) values of the material, as shown in the Equation 2.19.

$$\frac{1}{N_f} = \frac{\sigma_a}{S_e} + \frac{\sigma_m}{S_u} \quad (4.1)$$

According to the stress results, the maximum equivalent stress measured for optimum design should not exceed the endurance limit of the material. The design made in this way is considered safe [54].

4.2.1. Head Size and Offset Gap Configuration

Using the 28 mm head diameter as a reference, 5-second walking moment simulation was performed on implants using heads with -3, 0 and +3 mm offset gap sizes.

The moment reactions that occur due to the forces acting on the implant during walking are shown in Figure 4.6.

According to the results, there was an increase in the moment when moving from -3 mm to +3 mm.

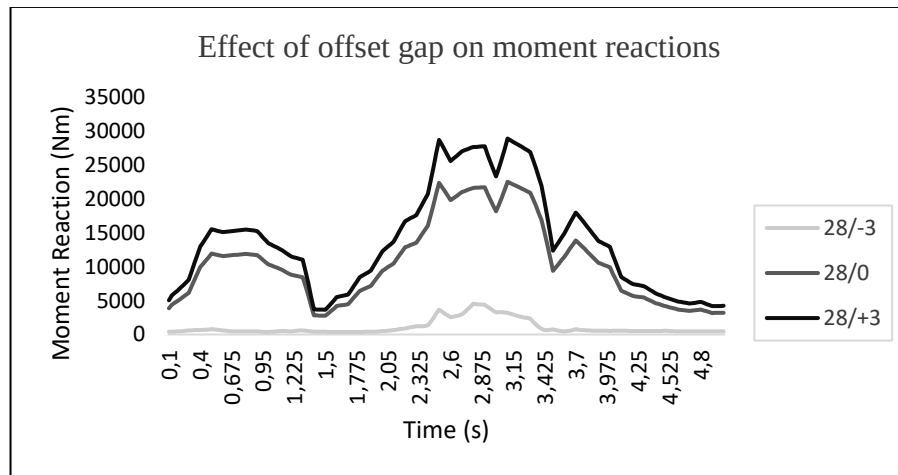


Figure 4.6: Moment reactions during walking with used different offset gap sizes.

As well as moment reactions resulted from increasing head diameter were analyzed using the same offset gap (+3 mm) and the moment increased as the head grew from 28 mm to 44 mm (Figure 4.7).

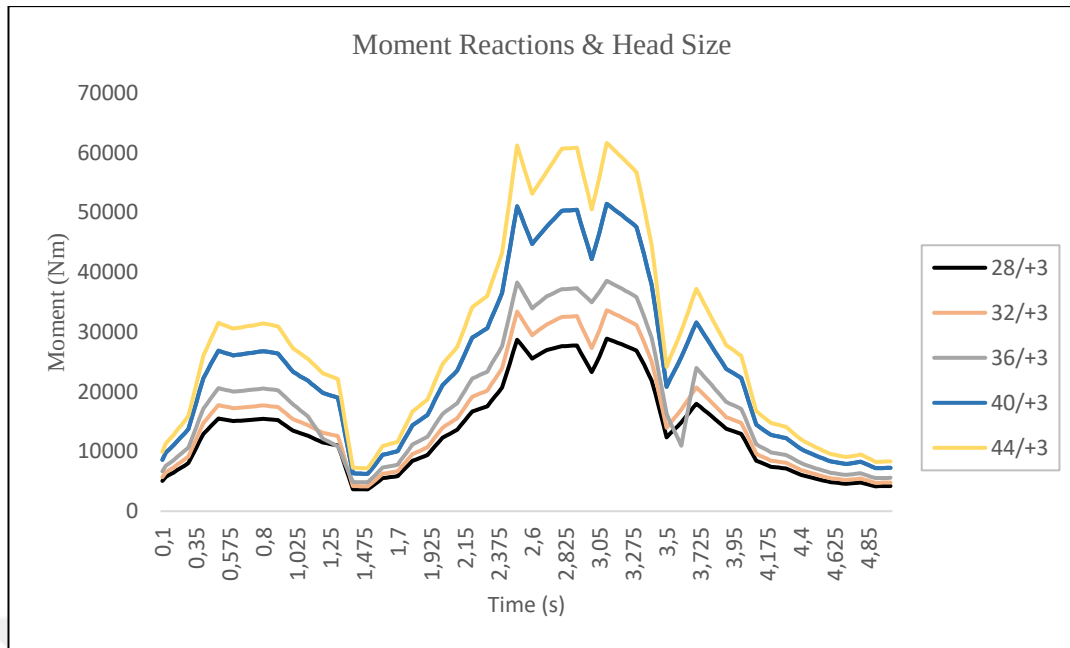


Figure 4.7: Effect of different head diameter on moment reactions at walking condition.

4.2.2. Effect of Offset Angle of 127° and 132°

Changing offset angle can influence torsional force and related wear. To see the results, transient analysis was performed with the forces during walking for both angle modeling and the results are shown in **Figure 4.8**. Accordingly, a greater increase in moment occurred in the 132° design.

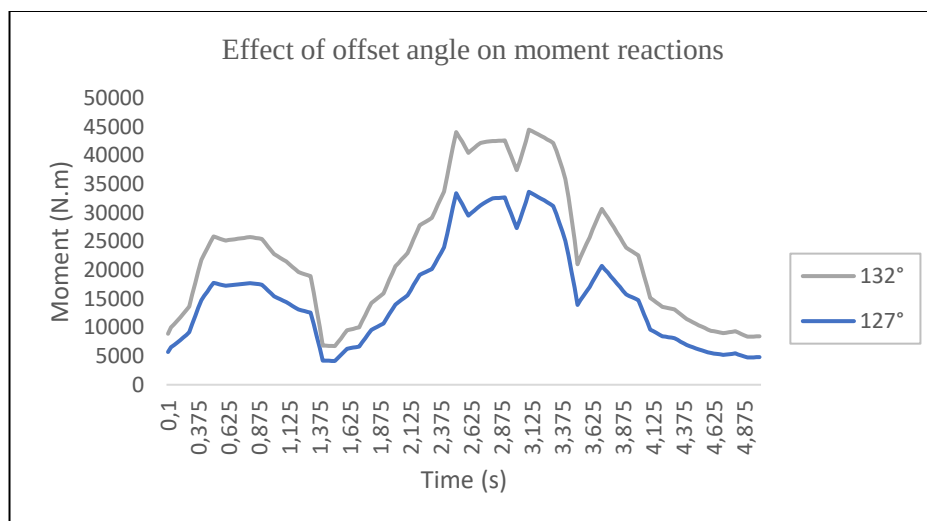


Figure 4.8: 127° and 132° offset angle effects on moment reactions.

4.2.3. Effect of Fixation Types on Moment Reactions

Walking moment simulation was generated based on 3 different fixation models and the highest moment occurred in the 3rd fixation where the stem was completely fixed.

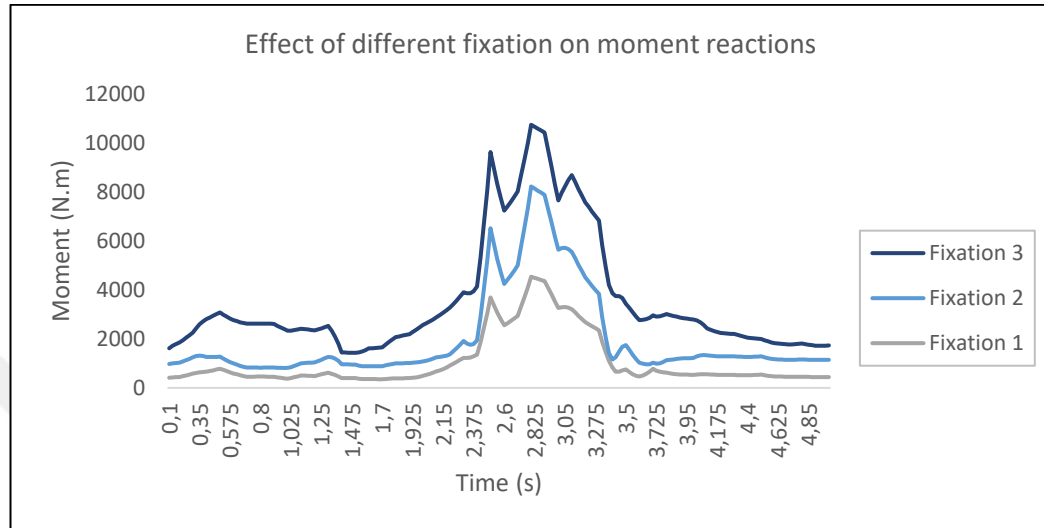


Figure 4.9: Effect of fixations on moment reactions: Fixation 1, 2 and 3.

4.3. Contact Tool Analysis (Sliding Distance)

4.3.1. Head Size and Offset Gap Configurations

The pressure values in the conical part of the implant body were evaluated, and the contact length for -3 mm, 0 and +3 mm gap offset measurements was 20, 17 and 14 mm, respectively. According with contact analysis, there was a decrease in pressure as the contact length increased. In other words, -3 mm gap offset creates less pressure on the conical surface than +3 mm gap offset.

The effect of increasing the head diameter on sliding distance was examined. As a reference, +3 mm increased offset was taken to evaluate similar dimensions.

Sliding distance occurring in the trunnion region was evaluated and there was a micro-level increase in the sliding with increasing head diameter.

When the same head diameter was evaluated for different offset gap sizes, a clear comparison and interpretation could not be made.

Table 4.1: Contact tool analysis for different head sizes and sliding distance results.

Head Size	Sliding Distance on Trunnion (mm)
28/+3	0,00053427
32/+3	0,00054853
36/+3	0,00056154
40/+3	0,00061863
44/+3	0,00080265

4.3.2. Effect of Offset Angle on Sliding Distance

A similar single reference head size was used to evaluate the sliding distance for offset angles. While there were deviations in the results, in general, the amount of slip and penetration was less in the 127° design. However, based on the results obtained, no clear interpretation could be made of its effect on wear (Table 4.2).

Table 4.2: Offset angle effects on sliding distance and penetration analysis results.

Factors	127°	132°
Sliding Distance (mm)	0,000539	0,00078
Penetration (mm)	9,88E-06	4,23E-05

5. DISCUSSION

For analyze hip prostheses with finite element analysis and realize it with correct results, it was critical to measure the components of the implant with the correct design and define the material properties correctly. In the main body of the prosthesis, the titanium alloy and its properties were determined to reflect reality, and dynamic analyzes were carried out for modal and walking simulation to show the static and vibration-related effects for stress and deformation effects, and the causes of wear in the trunnion area and its consequences were evaluated under four headings.

5.1. Stress – Deformation Outcomes

As a result of the static analysis, stress and deformation values were examined depending on different head diameters, offset angles and fixation types.

5.1.1. Head Size and Offset Gap Configurations

The results of static and dynamic analyzes of offset combinations of -3, 0, +3 mm for 5 head sizes of 28, 32, 36, 40 and 44 mm were examined. The amount of stress increased with increasing head diameter.

However, since variable results were obtained, a clear interpretation could not be made about the effects of changing offset measurements as well as changing head diameter.

For deformation, the largest head showed the highest deformation value.

As a result, it has been said that the use of large head diameter may increase the wear in the trunnion area.

5.1.2. Effect of Offset Angle

Stress and deformation effects were compared, while more stress occurred in the 127° design compared to 132°, less deformation occurred.

Accordingly, in hip implant which has 127° offset angle, the wear rate in the trunnion area can be further reduced compared to 132°.

5.1.3. Fixation Types

Stress and deformation values were considered for three fixation models, and while there was an increase from 1 to 3, there was a decrease in deformation (Table 5.1).

When fixation models are evaluated in terms of stress and deformation, the reason for the highest stress in the 3rd fixation is that increased stability increases the natural frequency of the stem and triggers the formation of stress.

On the other hand, the reason of less deformation is seen in Fixation 3 is the effect of increased stability on the wear factor.

In general, the use of Fixation 3 can be a model that reduces the risk of wear in the trunnion area in cases where stability is more important than moment reactions.

Table 5.1: Stress and deformation results for Fixation 1, 2 and 3.

Factors	Fixation 1	Fixation 2	Fixation 3
Stress (MPa)	64,254	69,938	72,22
Deformation (mm)	0,13532	0,057788	0,043513

5.2. Frequency – Vibration Effects

5.2.1. Head Size and Offset Gap Configurations

One of the important parameters for total hip replacement (THR) is its stability. A certain force is applied while placing the implant into the bone, and this force affects stability. As the stability increases, the resonance frequencies applied by the system also increase.

As a result of modal analysis, the frequencies obtained for 15 modes were evaluated for the trunnion region. The reason for the decrease in frequency when moving from 28 mm to 44 mm is the effect of head size on stability.

The use of a small head on the implant increases stability, and with increased stability, frequency values also increase. As a result, increased stability reduces the effect of trunnionosis in the trunnion area.

Additionally, the same head diameter was simulated for 3 different offset gaps and the highest frequency increase occurred at -3 mm offset. Accordingly, while the use of -3 mm head is more effective on stability, the amount of wear in the trunnion area decreases with increasing stability.

5.2.2. Offset Angle of 127° and 132°

The results of 127° and 132° offset angle designs were taken into consideration and when switching from 127° to 132°, there was a -5 mm decrease in offset size. Increasing the offset size of implants is effective in increasing the range of motion and reducing instability.

At the same time, increasing the offset is effective in reducing the force required for abductor movement and thus gives positive results on the mechanical behavior. Charnley (1970) and Muller (1970) clarified that increasing the femoral offset and medializing the acetabular component provides biomechanical benefits in the implant in their studies.

At the end of the modal analysis, both designs were parallel to each other with minor deviations based on frequency. Overall, a clear interpretation could not be made based on the frequency values due to the change in offset.

5.2.3. Fixation Types

According to the study 'Vibration analysis on partially cemented custom hip stem a per operative study' conducted by Jaecques et al., stability increases by increasing the contact surface between the bone and the implant during fixation and the resulting development of contact pressure. As the stability increases, the resonance frequency also increases, and the graph shifts to the right.

If two consecutive FRFs (frequency response functions) are the same (correlation 0.99), the contact conditions are considered to have reached optimal stability.

The results of 3 fixation models were examined and the reason for obtaining the highest frequency for Fixation 3 was associated with the maximum stability of this model in intramedullary fixation of the stem and the increased stability increasing the resonance frequency.

5.3. Moment Reactions

5.3.1. Head Size and Offset Gap Configurations

Low friction torque is required to minimize the torsional force, which is one of the reasons for loosening of the implant and causing wear in the trunnion area. As indicated in the results, an increase in torque occurred with the increase in head diameter. Therefore, the lowest torque levels are obtained in small heads. In this way, while the force is constant, the offset distance is reduced and trunnionosis decreases as the buckling force decreases with low torque levels. (Frankel, Balderston and Rothman, 1990).

Looking at the results, the moment increase from 28 to 32 mm is less than that of 40- and 44-mm heads. As specified, since the head size provides range of motion, in cases where range of motion is required, the 36 mm head is the one that gives the closest reaction to the 28 mm head as a large head.

36 mm gave the closest results compared to 28 mm and did not cause as much increase as the difference between 40- and 44-mm heads.

For offset gap sizes, the least moment reaction occurred at -3 mm, while more torque occurred at 0 and +3. -3 mm should be preferred to reduce wear corrosion in the trunnion area.

If the moment reactions are correlated, the 28 mm head and -3 mm offset gap configuration is the most effective combination in reducing wear.

When it is necessary to use a large head, it can be interpreted that the use of a 36 mm head with a -3 mm offset gap will be safer than the 40- and 44-mm head in terms of wear factor.

5.3.2. Offset Angle of 127° and 132°

For offset angles, more moment occurred at 132° in the moment occurring during walking. The 127° design is safer compared to the 132°, as the increased moment causes buckling force, increasing the risk of trunnionosis.

Although a clear interpretation cannot be made with the frequency values, the obtained moment results clearly show that the use of the 127° design has a much lower effect on wear compared to the 132° design. The results support the positive effect of the offset increase in the theory on stability.

The preferred configuration to minimize wear in the trunnion area will be achieved by using a small head with a -3 mm offset and choosing a 127° design.

It can be commented that when the 132° design is preferred, using a small head and -3 mm offset gap may cause less risk of trunnionosis compared to using a large head.

5.3.3. Fixation Types

As stated in the previous sections, the 3rd model causes the highest frequency and least deformation in fixation. On the other hand, the moment created by the system also increases with increasing fixation. In operations where the effects of vibration and moment are ignored, the stability of the system can be kept at maximum by using Fixation 3.

5.4. Contact Tool (Sliding Distance)

5.4.1. Offset Angle of 127° and 132°

In this study using titanium material, penetration, which is one of the effects causing trunnionosis, was evaluated and assuming that the area multiplied by penetration is equal to the wear volume, it can be said that the head size is effective in penetration. Wear also occurs due to penetration.

As to the simulation of offset angles during walking, which affects the wear on the contact surface, it can be said that the 127-degree design poses less risk for trunnionosis, as the amount of slippage in the 127-degree design is less than the 132-degree design.

As well as, the penetration value, which is another factor in wear, is as low as expected at 127 degrees, which will minimize the wear rate in the trunnion area compared to other designs.

6. CONCLUSION

Using finite element analysis, the factors causing trunnionosis on the implant were evaluated by performing 105 analyses on 22 models. The finite element method is not a method that gives precise results with its numbers, but rather a method that enables convergence and realistic predictions.

As a result of vibration analysis evaluations, +3 mm offset was the configuration that caused the least vibration. Accordingly, in studies where stability rather than vibration is required, -3 mm and 0 mm can be preferred as corrosion reducing models, respectively.

Among the head diameters, the smallest head is the model with the most frequency and the most stability. Small head and -3 mm offset may be effective to minimize the risk of trunnion corrosion. In cases where range of motion is required, a large head is preferred, while the corrosion rate can be reduced with a minimum 36 mm head. Since the 44 mm head causes the highest frequency, in cases where vibration is desired to be kept to a minimum, a less vibration configuration can be achieved by offsetting it by +3 mm. In another cases where stability is at the forefront and a large head must be used, a 44 mm head with a -3 mm offset gap can be preferred. The disadvantage of this combination may be that it causes more resonance frequencies in the system.

Since moment is one of the main factors causing trunnionosis, the effect of -3 offset found in the frequency analysis in reducing the risk of wear supports the results of moment reactions.

In cases where maximum range of motion is desired, the head with a -3 mm offset can be preferred and the wear rate can be reduced by reducing the moment ratio.

The best combination to minimize the buckling force was determined to be a small head -3 mm offset.

Besides that, for the offset angle, the 127° design is more effective in reducing the trunnionosis than the 132° design.

According to the rate of sliding during walking simulations on the trunnion contact surface between the head and the stem, the sliding rate increases with increasing head

diameter. The increase in the amount of slip increases the risk of trunnionosis because it triggers micromotion.

Therefore, the positive results of the use of small heads for trunnionosis were also supported by analyzes on the contact surface. If it is desired to apply wear-reducing combinations in the trunnion area due to fixation, in cases where stability is desired to be increased, Fixation 3 is preferred. If moment balance is important in the procedure performed, moment interaction can be minimized and the risk of trunnionosis can be reduced by choosing a small head. Since the moment is highest in Fixation 3, the use of large heads may further increase the moment in the system and increase the torsional force in the trunnion region.

By taking all the study parameters and the measurements obtained as reference, a study can be conducted with the data obtained from past surgeries.

By gathering the results and surgery data together, AI studies can be conducted using them as reference values in artificial intelligence-supported applications, and as a result of the study, predictions can be made about the patients.

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BIOGRAPHY

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PUBLICATIONS AND PRESENTATIONS FROM THE THESIS

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