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Comparison of rotational stability of short and straight stem prostheses in hip arthroplasty

An experimental study

Dissertation

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1. Introduction

The rate of total hip replacements is increasing which influenced the introduction of many prosthetic systems focusing to better the biomechanical aspects. Achieving the required biomechanical condition is a key factor that influences the patient's mobility, pain relief as well as implant longevity (Schmidutz et al. 2012a; Erceg 2009).The goal of any hip replacement is the restoration of a pain-free, functional joint that mimics the role of the normal anatomy as close as possible (Diehl et al. 2010).

1.1. Problem Statement

The shift of the patient clientele in total hip arthroplasty towards younger and more physically active patients, lead to changes in the demands on the prosthesis material. In doing so, the influence of the femoral neck bone and the torsional stability on the success of the treatment came increasingly into focus. Since the primary torsional stability is partly responsible for the adequate osteointegration of the prosthetic material, good treatment results can be achieved through conventional cementless prosthesis models (Kaminski et al. 2015; Nadorf et al. 2014). However, studies from the 1980s have already shown that the removal of stem material in the femoral neck region, as is necessary for conventional stem prostheses, influences the stability of the prosthesis. Yet, in the assessment of prosthetic design and treatment success, this relationship has received little attention for a long time (Nunn et al. 1989).

On the one hand to improve the anchoring and osteointegration of the hip joint replacement and on the other hand to meet the needs of the younger and active patient clientele, a so-called short stem prosthesis has been developed, which are increasingly used (Windhagen et al. 2015; van Oldenrijk et al. 2014). According to von Lewinski and Floerkemeier, about 30% of patients with hip replacement receive a short stem prosthesis (Lewinski and Floerkemeier 2015).

Numerous studies, most of which were *in vitro* studies, dealt with the biomechanics of these prosthetic forms, which differ in their design and fixation (Falez et al. 2015). There were better clinical results compared to straight stem prostheses (Kaminski et al. 2015). However, the data on short stem prostheses is insufficient. Which among other things, affects the long-term clinical outcome (Windhagen et al. 2015; van Oldenrijk et al. 2014; Nadorf et al. 2014).

In addition, there is no uniform classification of this type of prosthesis (Falez et al. 2015). To assess the efficiency of short stem prostheses, it is necessary to evaluate the biomechanical properties of the different prosthesis designs and to compare them to the results of conventional straight stem prostheses.

1.2. Aim of the investigation and research question

The study presented is part of an interdisciplinary *in vitro* project in the field of orthopedic endoprosthetics research. The aim of this study is to experimentally analyze and compare 5 currently available hip stem prostheses on the market, namely CLS®, EcoFit®, and TrendHip®, as well as the short stem prostheses Aida® and Metha®. The different models were biomechanically examined to investigate the following questions:

- What is the anchoring characteristic behavior of currently used femoral short stem and straight stem prostheses?
- How do different stem models differ in terms of anchoring behavior?

2. Theoretical Background

2.1. Anatomy of the Hip Joint

The hip joint (Articulatio coxae) connects the pelvis with the lower extremity and represents the second-largest joint in the human body. A hip joint is a special form of the ball and socket joint, it is referred to as a nut-joint (a nut in its shell). This design allows for a high degree of rotational freedom and allows the hip joint to move in all directions and levels. The range of motion for abduction and adduction is 45° / 0° / 35° according to the neutral-zero method. For internal and external rotation there is 40° / 0° / 25°. The range of motion of extension and flexion is 15° / 0° / 140° (Schmitz et al., 2017).

The acetabulum has a crescent shape and is partially covered by cartilage. The labrum is located at the margin of the acetabulum, which consists of fibrocartilage and connective tissue which engulfs the femoral head. The socket is formed by bony parts from the pubic bone, the ilium, and the ischium. The head of the femur represents the proximal end. Although the head is surrounded to some extent by the socket, only the cartilaginous facia lunata serves the articulation of both joint surfaces, where the force transmission takes place (Schiebler and Korf 2007).

In addition to the bony parts, the hip joint is surrounded by the joint capsule, ligaments, and muscles. The joint capsule extends from the edge of the acetabulum to the intertrochanteric line of the ventral femur and above the intertrochanteric crest on the dorsal femur. The ligaments include the pubofemoral ligament, the iliofemoral ligament, and the ischiofemoral ligament, which are the strongest ligaments in the human body. There are also the femoral head ligament and the transverse ligament. The hip muscles are divided into internal and external hip muscles, which are supplemented by adductors (Schmitz et al. 2017).

2.2. Surgical technique

There are many approaches to the hip joint. The most conservative approach is an incision that follows the course of the tensor fasciae latae muscle towards the head of the fibula, 1-2 cm infero-lateral to the anterior superior iliac spine. Following the incision, the subcutaneous tissue is separated from the fasciae of the tensor latae muscle, the fasciae is then opened and prepared between the rectus femoris muscle and the tensor fasciae latae muscle. Through blunt preparation into the capsule the cutaneus femoris lateralis nerve and the surrounding muscles should be preserved. Afterwards, the rectus femoris muscle is displaced medially, three Hohmann retractors are used to provide adequate exposure of the anterio capsule. Resection of the ventral joint capsule is carried out in this stage and osteotomy of the femoral neck at the planned level. In order to reveal the acetabulum two retractors must be placed laterally and medially to the acetabulum. (Gebel et al. 2012)

2.3. Indication of Hip joint replacement

Since there is no clear consensus for total hip replacement (THR) surgery, any symptomatic changes that affect the mobility, functionality and biomechanics, such as pain at rest or activity, difficulty in climbing stairs and/or wearing socks and shoes, affected walking distance and examined range of motion and amount of preserved joint space seen on radiographic examinations, may be reviewed for hip replacement surgery (Dreinhöfer et al. 2006). There are many prearthritic deformities, such as developmental hip dysplasia and Perthes' disease, that may lead to secondary osteoarthritis, these deformities affect the form and function of the joint (Hackenbroch 1998).

2.3.1. Osteoarthritis

The symptomatic hip osteoarthritis (OA) has an estimated lifetime risk of 25% for people living to 85 years of age (Murphy et al. 2010), and a 10% lifetime risk for receiving a total hip replacement (Culliford et al. 2012). The risk factors for hip OA can be divided into two categories, those at the joint level which are considered etiological for developing hip OA, and those at the whole person level which adds to developing OA indirectly through assisting the joint level risk factors. Joint level risk factors include developmental dysplasia of the hip (DDH), Femoroacetabular impingement (FAI) as well as joint injury and labral tears (Murphy et al. 2016). In DDH the shear forces are distributed from the anterosuperior part of the joint onto the acetabular rim, this is due to a shallow and misoriented socket leading to a lower contact surface area in the joint (Klaue et al. 1991). FAI consists of two types, cam and pincer FAI (Ganz et al. 2001). The abnormality in the cam variant is a thickened femoral head-neck junction. This results ultimately in parting and damaging the acetabular cartilage from the labrum (Beck et al. 2005). The Pincer variant exists of a deepened acetabulum resulting in over coverage of the femoral head. In flexion, the femoral neck comes in conflict with the labrum resulting in direct damage to the labrum and overtime the underlying cartilage (Ganz et al. 2003). It is accepted that the labrum provides a suction seal which affects stability of the joint, pressure distribution and maintaining the synovial fluid. When the labrum is torn these functions are affected negatively (Crawford et al. 2007). Labral tears are believed to show a close relationship to cartilage damage and bone marrow lesions which are two signs of OA (Neumann et al. 2007; McCarthy and Busconi 1995). Whole person level risk factors include age, sex, weight, genetics, ethnicity and occupation (Murphy et al. 2016). It is of interest to mention, for each five-units increase in body mass index correspondingly increases the risk of hip OA by 11% (Jiang et al. 2011). A 60% risk of OA can be attributed to genetics, labor intensive professions and high impact sports also increases the risk of OA (Harris and Coggon 2015; Amanatullah et al. 2015; Sulsky et al. 2012; MacGregor et al. 2000). The conservative treatment includes patient education, weight management and exercise (Nguyen et al. 2016). Diclofenac and Etoricoxib have shown strongest efficacy of all non-steroidal anti-inflammatory drugs (NSAIDs) for pain relief in hip and knee OA (da Costa et al. 2014). Hyaluronic acid is generally not recommended in hip or knee OA (Hochberg et al. 2012; NCGC 2011). Total hip arthroplasty (THA) can be considered for patients who have exhausted all other options. Over 1 million THA operations are carried out annually, more than 90% are the result of end-stage hip OA. The survival rate post-THA is more than 95% at 10 years and above 80% at 25 years (Pivec et al. 2012).

2.3.2. Necrosis

Avascular femoral head necrosis is most common between the ages of 30 and 50 (Kamal et al. 2012). There are many etiological risk factors, the more common factors include chronic alcohol abuse, nicotine abuse, ingestion of corticosteroids and the presence of trauma (Kamal et al. 2012). Other less common factors include human immunodeficiency virus (HIV), Sickel cell anemia, leukemia, pregnancy, chemo- and radiotherapy (Arbab and König DP 2016). Long term intake of corticosteroids has been proven to increase the risk of femoral head necrosis by 10% to 30% of cases analyzed (Koo et al. 2002; Wang et al. 2000). Patients being treated with 2 grams of prednisolone should be regarded as critical (Griffith et al.

2005). Excessive alcohol consumption of 320 grams ethanol per week increases the risk factor of femoral head necrosis by a factor of 2.8 (Matsuo et al. 1988).

As a conservative approach, Iloprost and alendronic acid (used for smaller lesions) are applicable as a curative therapy. According to ARCO classification, stage II should be treated with core decompression. More advanced lesions of Stage III and IV should be reviewed for indications of total hip replacement. Currently, the effectiveness of other treatment modalities such as bone marrow aspirate or stem cell transplantation cannot be specified (Roth et al. 2018).

2.3.3. Socioeconomic aspects of hip arthroplasty

From an economic point of view, the efficient procedure with the lowest possible complication rate is of high importance in hip arthroplasty (Lamo-Espinosa et al. 2015). The German endoprosthetic registry (EPRD), released the following summery in 2017 (Table 1) (Grimberg et al. 2017).

Table 1: Number of documented operations by ERPD in 2017.

Joint	Primary Implantations	Revision Procedures	All Procedure Types
Hip	140.871	16.453	157.324
Knee	112.734	12.880	125.614
Hip and Knee	253.605	29.333	282.938

The total documented operations by the EPRD in 2018 was 300.192. The number of cementless hip endoprostheses has increased from 2014 till 2018 by 4.2%. Although the increase in the implementation of short stem prostheses only registered an increase by 3.1% in the time period from 2015 to 2018 some hospitals reported higher short stem utilization than straight stems. The use of a head diameter of 36mm in total hip replacement has increased in 2018 by 6.5% since 2014 to a total of 37.9% (EPRD).

Mujica-Mota et al., calculated costs and outcomes of 154,470 AOK insured total hip replacement patients with osteoarthritis. The average costs for the primary implantations 2007-2009 were €7,221 per patient not including inpatient treatment costs. Revision costs averaged €12,573 per patient. Additional costs of the first 90 days after primary implantation were 2.4% for dislocation costing €3,697 per patient. Pulmonary emboli or deep vein thrombosis occurred in 0.45% of patients costing €3,141 per patient. Femur fractures occurred in 0.28% of cases costing €8,155 per patient. Other complications occurred in 3.84% and cost €9,106 on average per patient (Mujica-Mota et al. 2017).

2.4. Biomechanical Aspects of short Stem Prosthesis in Hip Arthroplasty

The restoration of the joint’s biomechanics is essential for the postoperative prognosis. However, only a few studies have compared both short and straight stems biomechanically. Amongst the few comparative studies was a retrospective study by Ries et al. on 100 patients, of whom 50 patients received a short stem endoprosthesis and the other 50 received a straight endoprosthesis due to hip osteoarthritis. The biomechanical parameters were the horizontal femoral offset, the femoral head center of rotation and limb length. In the results, both stem types showed an increased offset. The short stems had a 2 mm increase and the straight stems showed a 3,3 mm increase. The authors hypothesize that straight stems significantly improved offset. The offset is defined as the distance between the femur axis and the center of rotation of the femoral head. The femoral offset is involved in force and tension distribution, which influences the femur and stem. Concerning the center of rotation, both stems showed significant medialization being 4.2 mm for the straight endoprosthesis and 6.0 mm for the short stem endoprosthesis (Ries et al. 2015).

Jones et al. investigated rotational stability as an expression of biomechanics and the associated risk of periprosthetic fracture on uncemented short stem endoprostheses. In total, the authors tested 16 implants on artificial bones and two implants on cadaveric femora. Immediately after implantation, the implants were rotated rapidly until a fracture occurred. Thus, internal rotation was simulated, as can occur when stumbling around the fixed foot in practice. The results showed that short stem endoprostheses fractured significantly at higher rotation force (27.1 versus 24.2 Nm) and at a greater angle (30.3° versus 22.3°) than straight stem endoprostheses. This result was evident in both artificial and cadaveric bone. Based on these results, the authors suggested that the uncemented press-fit short-stemmed endoprosthesis allows greater rotational flexibility in the femur thus, reducing fracture risk on rotation (Jones et al. 2015).

2.5. Procedure specific complications in hip arthroplasty

Infection of the surgical site was reduced significantly since the adoption of laminar flow in operating theaters and prophylactic antibiotics to between 0.3 and 1.5% (Lindsay et al. 2011; Blom et al. 2003). Deep vein thrombosis was a common complication following total hip replacement. However, due to the implementation of modern thromboprophylaxis the rates have been reduced to 44% and symptomatic complications have been reduced to 1.3%. (Samama et al. 2007; Mantilla et al. 2002). Nerve injury is estimated to occur in approximately 0.6 - 3.7% following total hip arthroplasty. The most common nerve injured is the sciatic nerve (Hasija et al. 2018).

The occurrence of dislocation can vary between 0.2 and 10%. It is estimated that 2% of dislocations occur within the 1st year following surgery (Dargel et al. 2014). Most dislocations occurs posteriorly, therefore the posterior approach has been associated with a higher rate of dislocation 5.8% compared to the anterior approach 2.3% (Woo and Morrey 1982). Leg length discrepancy has been documented to occur in between 1 and 27% (Ranawat and Rodriguez 1997). Discrepancies measured have widely varied between 3 to 70mm (Sathappan et al. 2008). Limb lengthening has been associated with sciatic and peroneal palsies (Barrack 2004). Periprosthetic fracture tend to occur more intraoperatively in cementless fixations. It is estimated that for primary THA the rate of occurrence is 5.4% and higher for revision surgeries (Berry 1999). Within the complications after hip arthroplasty, aseptic loosening is one of the major complications leading to implant failure (Ovrenovits et al. 2015; Boot et al. 2015; Fernandez-Sampedro et al. 2015; Sadoghi et al. 2013). The prevalence of aseptic loosening is reported to be 2% to 18% in the literature (Adesanya et al. 2015; Moojen et al. 2010). A long-term study with an average follow-up of 36 years showed an incidence of 20% for stem loosening and 32% for loosening of the socket (Watts et al. 2016). It is difficult to diagnose septic loosening as the primary cause of failure without specific investigations such as blood work, direct aspirate cytological and microbiological analysis and PET scan (Love et al. 2009). Causes of secondary procedures were published by the EPRD, the most common cause was loosening (29.8%) followed by infection (15.2%), dislocation (11.7%), periprosthetic fracture (10.9%), and implant wear (8.1%) (Grimberg et al. 2017).

2.5.1. Periprosthetic pain

In some patients with hip arthroplasty, postoperative hip and limb pain occurs. These can occur both in the short and long term. About a quarter of patients report such hip and limb pain (Lungu et al. 2016).

Nevertheless, there are few studies in the literature dealing with chronic hip and limb pain after implantation of an endoprosthesis, such as the short stem endoprosthesis. According to Baert et al., extrinsic, and intrinsic factors can cause pain. The extrinsic causes include muscular changes or spinal disorders, while the aseptic prosthetic loosening or infection in the prosthetic area are considered as intrinsic factors. Moreover, the authors distinguish between central and peripheral causes. Central causes involve the central nervous system. Biomechanical and structural factors, such as micromotion and prosthetic misalignment, are considered peripheral factors (Baert et al. 2017).

In contrast, Amendola et al. evaluated whether chronic hip and limb pain could be empirically quantified after implantation of a short stem endoprosthesis. For this purpose, the authors examined 238 patients (261 implants) with a primary hip replacement using a cementless short stem endoprosthesis made of titanium. The average follow-up was three years. Among the patients, 66% did not complain of pain. Mild pain was described by 16% and 9% had moderate to severe pain in the thigh area. In one case, the pain symptoms led to revision surgery. Furthermore, the authors were able to demonstrate an indirect linear correlation between pain intensity and age of the patients. Due to a good growth rate and a marked increase in the Harris hip score in the pre- and postoperative comparison, the study found no correlation between postoperative chronic pain and prosthetic loosening (Amendola et al. 2017).

2.6. Stress shielding in hip joint replacement

Under physiological conditions, the bone structure and density change depending on the load. Thereby, an increase in bone density occurs under the influence of load on bone, vice versa a decrease in stress leads to a reduction in bone density. This has be described by the German anatomist Julius Wolff (Wolff 1986).

In hip arthroplasty, the relationship between load and bone structure becomes relevant, as force distribution changes due to the implant and is influenced by the implant design (Huang et al. 2019). The forces emanating from straight stems are largely transferred distally in the diaphysis which may lead to periosteal thickening in the affected area. The phenomenon is expressed by Wolff's law (Wolff 1986). Because the load distribution on the periprosthetic proximal bone is transferred distally, the proximal area shows bone loss. This is described as stress shielding (Sumner 2015). As a result, these changes can manifest clinically as aseptic prosthetic loosening (Kwon et al. 2013).

2.6.1. Primary and secondary stability

In hip arthroplasty, a distinction is made between primary and secondary stability. The primary stability defines immediate postoperative stability, which allows the immediate full load on the joint. Secondary stability describes the stability achieved by the osseointegration of the endoprosthesis (Ochsner 2013). Primary stability determines the long-term outcome. The stability can be improved by the choice of an appropriate endoprosthesis (Schmitz et al. 2017).

2.6.2. Benefits of short stem prostheses

An advantage of the short stems is found in the preservation of the femoral bone, especially in the area of the femoral neck of the metaphysis and diaphysis. Future replacement surgeries may benefit from the associated bone preservation with short stems (Kim et al. 2014; Jerosch et al. 2012). This advantage does not exist in case of revision of straight stems although they are established and clinically well proven (Kärrholm et al. 2017; Makela et al. 2014; Streit et al. 2012; Hailer et al. 2010). Regardless of the biomechanical results, short stem endoprosthesis is increasingly becoming the first choice (Kutzner et al. 2017; Ettinger et al. 2015; Gulow et al. 2007). Short stem endoprostheses were associated with a lower rate of hip pain and intraoperative fractures (Yu et al. 2016).

Moreover, multiple studies have reported high patient satisfaction, a dramatic decrease in pain, a high return to sports rate, low complication and revision rates with short stems (Donner et al. 2019; Schnurr et al. 2017; Yu et al. 2016; Schmidutz et al. 2012b)).

3. Methodical Approach

3.1. Study Design

The study was designed as an *in vitro* study on five different hip endoprostheses using artificial bone. The methodological approach was based on previous biomechanical analyses studies (Jahnke et al. 2016; Kinkel et al. 2015; Hamadouche et al. 2015; Nadorf et al. 2014; Kasten et al. 2012; Jakubowitz et al. 2011; Jakubowitz et al. 2008).

The target parameters defined through torsional torque were the associated micromotions on both bony and prosthetic surfaces. Each model had a total of 7 measurement points. Four measurement points were assigned to each bone (B1-B4) and 3 to each implant (P1-P3). The lesser trochanter was used as a reference point. Prosthesis measurement points were allocated at the stem's shoulder, metadiaphyseal transition, and the tip of the prosthesis. The bony measurement points were placed relative to the location of their prosthetic counterparts. The resulted relative micromotions (RM) reflected the degree of motion occurring at the designated measurement point and in term describing primary stability.

Torques without any retroactivity were applied in the ventro-dorsal direction into the hip stem prosthesis at a continuous interval of ± 7Nm. The torque interval is subdivided into 80 steps, with a series of three repetitions with two intervals each (= 480 measured values) being effected for each measuring point (Jahnke et al. 2016; Jakubowitz et al. 2008).

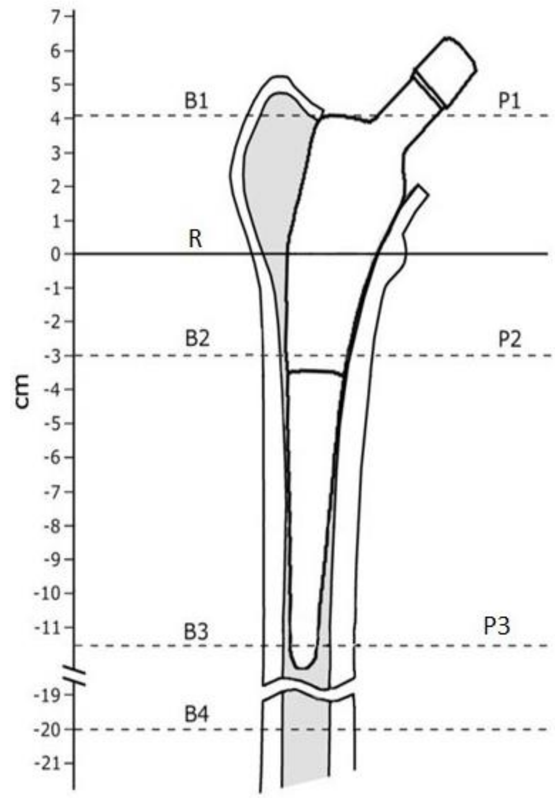

Figure 1: Illustration of all measurement points (R) reference point related to the middle of the lesser trochanter. (P1-P3) Prosthesis measurement points. (B1-B4) Bony measurement points.

3.2. Endoprostheses

The hip endoprostheses used were short and straight stem endoprostheses. Both types of prostheses were monoblocks and had similar material properties.

The following figure (Figure 2) shows all the prosthetic system that will be discussed in the following section.

Figure 2: EcoFit®, CLS®, Trendhip®, Aida® and Metha® prostheses (from left to right).

3.2.1. Straight stem prostheses

The straight stem endoprostheses included the CLS®, EcoFit®, and TrendHip® prosthetic systems.

Cementless Spotorno System CLS® (Zimmer Biomet Holdings, Inc., Warsaw, Indiana, USA)

The cementless endoprosthesis CLS® is a titanium-based endoprosthesis with proximal ribs that increase the contact surface area which improves stability and osteointegration. Its three-dimensional tapered design allows for better proximal force transference and the distal side of the implant has a narrowed portion to avoid contact with the cortical bone. (straight stem size 9.0 was utilized in this study).

EcoFit® (Implantcast GmbH, Buxtehude, Germany)

EcoFit® consists of the titanium material implatan® TiAl6V4 and the surface area is treated with implaFix® (commercially pure titanium) or a hydroxyapatite coating. The cemented EcoFit® stems are composed of Implavit® (chromium-molybdenum alloy) with a polished or matt surface and are treated with a titanium nitride coating which is suitable for patients with metal sensitivity (straight stem size 8.75 was utilized in this study).

TrendHip® (B. Braun Medical Inc., Bethlehem, Pennsylvania, USA)

The prosthesis is available in both cementless and cemented designs and is coated with hydroxyapatite. The cementless rectangular self-locking stem, medial and horizontal grooves, allow for increased contact to the bone and adequate primary stability. (straight stem lateralized size 3 was utilized in this study).

3.2.2. Short stem prostheses

The short stem endoprostheses examined included Aida® and Metha®.

Aida® (Implantcast GmbH, Buxtehude, Germany)

Aida is a metaphyseal cementless prosthesis. The implant has a tapered shape with a trapezoidal cross-section and is made from TiAl6V4. The distal stem end is polished, the proximal portion of this prosthesis has a microporous coating consisting of hydroxyapatite and commercially pure titanium. This microporous coating serves to improve secondary osseointegration. Moreover, the trapezoidal cross-section design of the stem increases rotational stability (lateralized size 1 was utilized in this study).

Metha® (B. Braun Medical Inc., Bethlehem, Pennsylvania, USA)

The implant is made from titanium and its upper two-thirds are coated with Plasmapore®, a combination of microporous Ti-plasma rough coating and a 20µm layer of dicalcium-phosphate-dihydrate on top. This layer increases the bone-implant contact surface area and thus, improves osseointegration. Primary stability and force transference are supported by the implant's conical shape, medially angulated stem and through the metaphyseal anchoring of the implant at the femoral neck (monoblock size 1 was utilized in this study).

3.3. Bone material

The artificial bone used was composite bone 4th generation, size M, # 3403 (Figure 3) (Sawbones® Europe, Malmo, Sweden). An advantage of using artificial bone is eliminating biological factors such as bone density and structure. This allowed for a better mechanical evaluation. Moreover, Cristofolini et al. demonstrated, that mechanical examinations carried out on synthetic bone were sufficiently comparable to physiological bone (Cristofolini et al. 1996).

Figure 3: Sawbones® artificial femur 4th generation. SKU #3404.

3.4. Preparation and Implantation of the Prostheses

The osteotomies were carried out by an experienced surgeon. The stems were drilled (drill: BE 1100, Metabo, Nürtingen) creating a hole for attaching the measurement pins. To assure the stability of the specimens, they were implanted basally in a plaster block. To standardize and to create an equivalent situation for all implants, the implantations were done by using a pressing machine (ZugDruck-Universalprüfmaschine Inspekt table blue 20 kN, Hegewald & Peschke, Nossen) (Figure 4). In order to produce the anatomical 6° of adduction, a 6° wedge was inserted under the blocks. A cyclic load of 25 cycles of 2kN and 25 cycles with 4kN for each stem was done. After that, the specimens were radiologically examined to determine proper placement of the prosthesis.

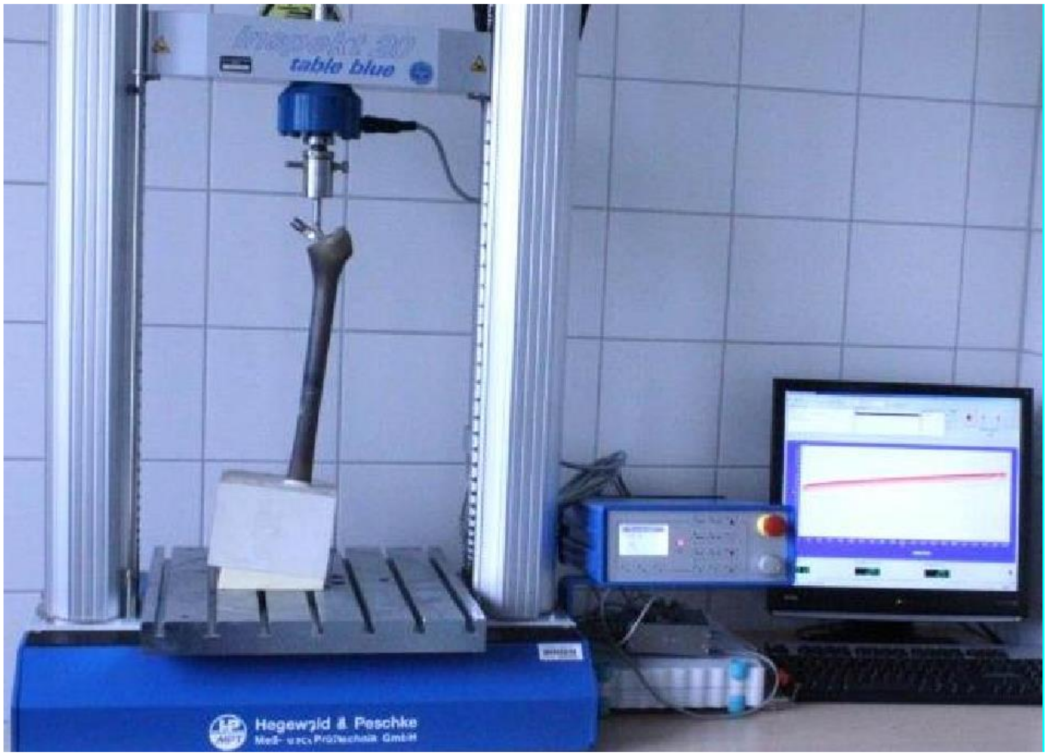

Figure 4: Press machine.

3.5. Measuring device and test procedure (measurement protocol)

After implantation, the model was placed in a stability measurement device (Figure 5). A pin was inserted to a measurement point. Followed by the attachment of the sensory cube to the pin. A multi-rotational jointed hinge was fixated on the lesser trochanter. Attached to the hinge was a frame with openings as insertions for the sensory rods. The hinge allowed free placement of the sensory rods according to the situation which was 6 linear variable differential transducers (LVDT's) (Typ P2010, Mahr GmbH, Göttingen) arrranged in 3-2-1 setup with a resolution of 0.1μm for a 3-plane assessment, which has already been established in previous publications studying the following prostheses Metha®, Aida®, MiniHip™ (Corin Group, Cirencester, Gloucestershire, UK), AMIS-K-stem® (Medacta©, Castel San Pietro, Switzerland) (Jahnke et al. 2018; Jahnke et al. 2016; Hamadouche et al. 2015).

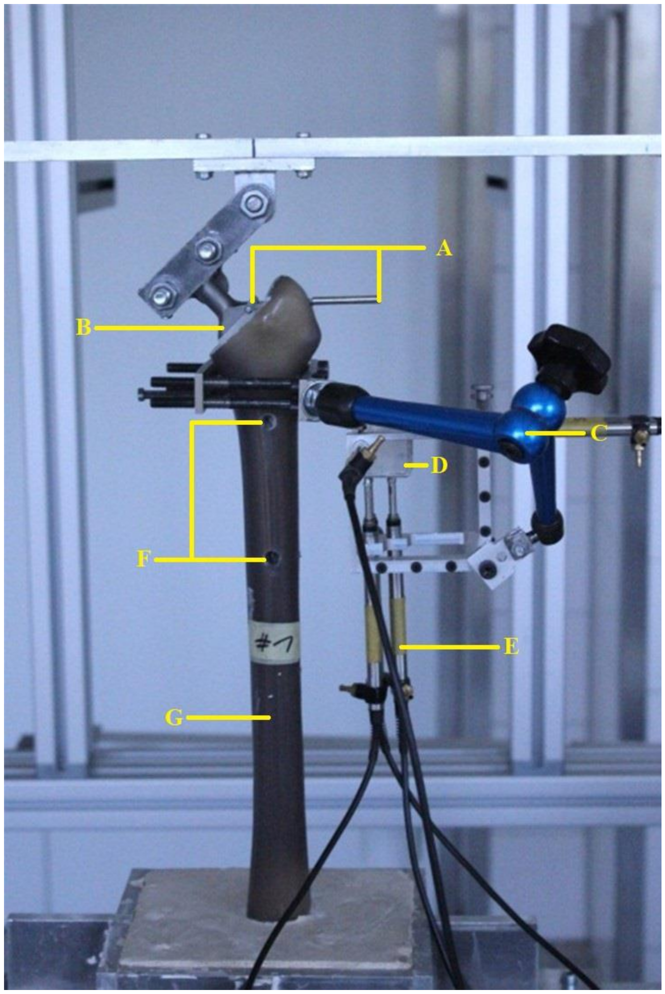

Figure 5: Device used for stability measurement. A) Measurement pins attached at P1(left) and B1 (right). B) Implanted prosthesis. C) Multi jointed hinge. D) Sensory cube. E) Sensory rods. F) Drilled positions for measurement of P2 and P3. G) Synthetic bone (sawbone).

The proximal part of the stem was attached to a lever arm. A rope system was applied to the lever arm, it carried 2 linear actuators which applied forces to the lever arm. Thus, transferring the forces to the stem. The actuators were placed plane and moved in a counter direction to its counterpart.

As a result, very small but measurable micromotions were generated without affecting the anchoring stability of the prostheses. In order to determine the primary stability, the relative micromotion of the bones and the stems were recorded at four different measuring point levels (ventral prosthetic measurement points: P1-P3; lateral bone measurement points: B1-

B4) and the relative micromotions (RM1: proximal measuring point; RM2: metaphysis/diaphysis transition; RM3: prosthesis tip (short stems). Relative micromotions RM1 to RM3 result from the difference between the normalized angles of rotation (NAR), a_Z/T_Z of the bone and the prostheses described as $\Delta\ a_Z/T_Z$.

3.6. Statistical evaluation

Statistical analysis was carried out with the statistical program SPSS® (Version 24). After a descriptive evaluation of the target parameters (rotation at the bone and prosthesis measurement points, relative micromotions) and calculation of mean values, minimum and maximum values as well as the standard deviation, the target parameters were determined in a second step Depending on the type of prosthesis compared. A two-factorial univariate variance analysis is used to detect differences between the prostheses as a function of the different measuring point levels. For the pair-wise comparisons, an LSD post-hoc test is used. Multiple comparisons are adjusted using the Bonferroni correction. A P-value <0.05 is considered statistically significant.

4. Results

4.1. Explorative Analysis

Every stem (straight endoprosthesis: CLS®, EcoFit®, TrendHip®, short stem prostheses: Aida®, Metha®) was examined on five bones individually. The mean, minimum and maximum values, as well as the standard deviations for the torsions at the measuring points of bone and prosthesis, were studied. The micromotions were observed for the whole collective, for the short and straight stem endoprostheses. The results of the individual types of endoprostheses are discussed separately.

4.1.1. Results of individual stem types

This section reviews each stem type individually. Each examined implant contains three tables describing the absolute measurement values of prosthesis, bone and relative micromotions.

Short-stem Aida®

Table 2: Prosthetic results for Aida®

	P1		P2		P3	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	4.31	16.35	-1.30	4.32	-5.67	2.50
# 02	4.03	13.98	-1.41	4.40	-6.04	2.62
# 03	4.02	12.27	-1.59	1.96	-6.03	1.82
# 04	3.88	10.62	-1.44	1.91	-6.63	1.39
# 05	4.19	15.20	-1.25	5.42	-5.61	3.21
Ø	**4.09**	**13.68**	**-1.40**	**3.60**	**-6.00**	**2.31**
SD	**0.17**	**2.28**	**0.13**	**1.58**	**0.41**	**0.71**

Table 3: Synthetic bone results for Aida®

	B1		B2		B3	
Test	Height [cm]	NAR αz/Tz [mdeg/Nm]	Height [cm]	NAR αz/Tz [mdeg/Nm]	Height [cm]	NAR αz/Tz [mdeg/Nm]
# 01	4.31	-1.37	-1.30	0.41	-5.67	-7.76
# 02	4.03	-0.19	-1.41	0.07	-6.04	-9.70
# 03	4.02	-0.76	-1.59	0.30	-6.03	-13.25
# 04	3.88	-0.35	-1.44	0.13	-6.63	-15.47
# 05	4.19	0.00	-1.25	0.00	-5.61	-7.20
Ø	**4.09**	**-0.53**	**-1.40**	**0.18**	**-6.00**	**-10.68**
SD	**0.17**	**0.54**	**0.13**	**0.17**	**0.41**	**3.57**

Table 4: Relative micromotion results for Aida®

	RM1		RM2		RM3	
Test	Height [cm]	RM Δαz/Tz [mdeg/Nm]	Height [cm]	RM Δαz/Tz [mdeg/Nm]	Height [cm]	RM Δαz/Tz [mdeg/Nm]
# 01	4.31	17.71	-1.30	3.91	-5.67	10.26
# 02	4.03	14.17	-1.41	4.33	-6.04	12.32
# 03	4.02	13.03	-1.59	1.66	-6.03	15.07
# 04	3.88	10.97	-1.44	1.78	-6.63	16.86
# 05	4.19	15.20	-1.25	5.42	-5.61	10.41
Ø	**4.12**	**14.22**	**-1.43**	**3.42**	**-5.91**	**12.98**
SD	**0.16**	**2.51**	**0.15**	**1.65**	**0.21**	**2.91**

The value difference between P1 and B1 was relatively high, which suggested presence of micromotions. The prosthetic value increased negatively at P2 while the bony measurement point B2 showed minimal value difference compared to B1. The measuring point P2 came to close adherence to its boney counterpart B2, which conveyed the highest stability at the metadiaphyseal area. P3 recorded a comparable value to P2. However, its bony counterpart B3 increased negatively resulting in an increased relative micromotion at the stem's tip. The highest stability of the prosthesis was recorded at the middle measuring point RM2 followed by the proximal measuring point RM1 and lastly the distal measuring point RM3.

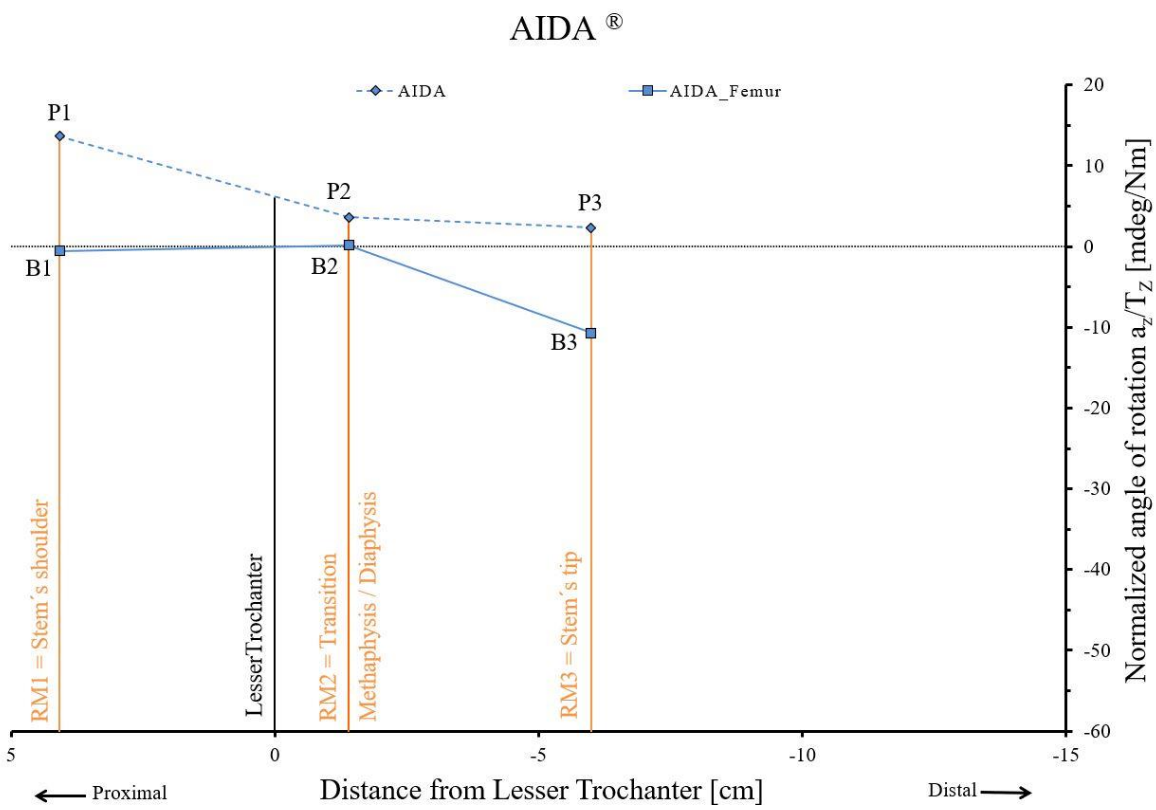

Figure 6: Summary of the results for the short stem endoprosthesis Aida®.

Short- stem Metha®

Table 5: Prosthetic results for Metha®

	P1		P2		P3	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	4.13	17.64	-1.28	5.93	-4.03	6.88
# 02	3.99	16.24	-1.40	7.09	-4.20	2.32
# 03	4.09	16.76	-1.46	7.93	-4.31	5.44
# 04	3.89	14.97	-1.48	4.45	-4.27	5.64
# 05	4.13	18.52	-1.36	3.74	-4.13	6.61
Ø	**4.05**	**16.83**	**-1.40**	**5.83**	**-4.19**	**5.35**
SD	**0.10**	**1.35**	**0.08**	**1.75**	**0.11**	**1.96**

Table 6: Synthetic bone results for Metha®

	B1		B2		B3	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	4.13	-0.09	-1.28	0.03	-4.03	-8.12
# 02	3.99	-0.63	-1.40	0.22	-4.20	-7.77
# 03	4.09	-0.61	-1.46	0.22	-4.31	-9.02
# 04	3.89	-0.28	-1.48	0.11	-4.27	-10.27
# 05	4.13	-0.65	-1.36	0.21	-4.13	-10.14
Ø	**4.05**	**-0.34**	**-1.40**	**0.11**	**-4.19**	**-4.91**
SD	**0.10**	**0.42**	**0.08**	**0.15**	**0.11**	**3.21**

Table 7: Relative micromotion results for Metha®

	RM1		RM2		RM3	
Test	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]
# 01	4.13	17.73	-1.28	5.90	-4.03	15.00
# 02	3.99	16.87	-1.40	6.87	-4.20	10.09
# 03	4.09	17.37	-1.46	7.71	-4.31	14.46
# 04	3.89	15.25	-1.48	4.34	-4.27	15.92
# 05	4.13	19.17	-1.36	3.53	-4.13	16.76
Ø	**4.07**	**17.28**	**-1.38**	**5.67**	**-4.18**	**14.44**
SD	**0.07**	**1.42**	**0.09**	**1.73**	**0.14**	**2.59**

The greatest value difference was observed at P1 and B1. The prosthetic value recorded a decrease from P1 to P2 leading to a close adherence between P2 and B2 which recorded the lowest relative micromotion value. The prosthetic value recoded minimal difference between P2 and P3. However, the Bony value began to increase negatively resulting in an increased relative micromotion at the prosthetic tip. This prosthesis showed the highest stability at the middle measurement point RM2 followed by RM3 and lastly RM1.

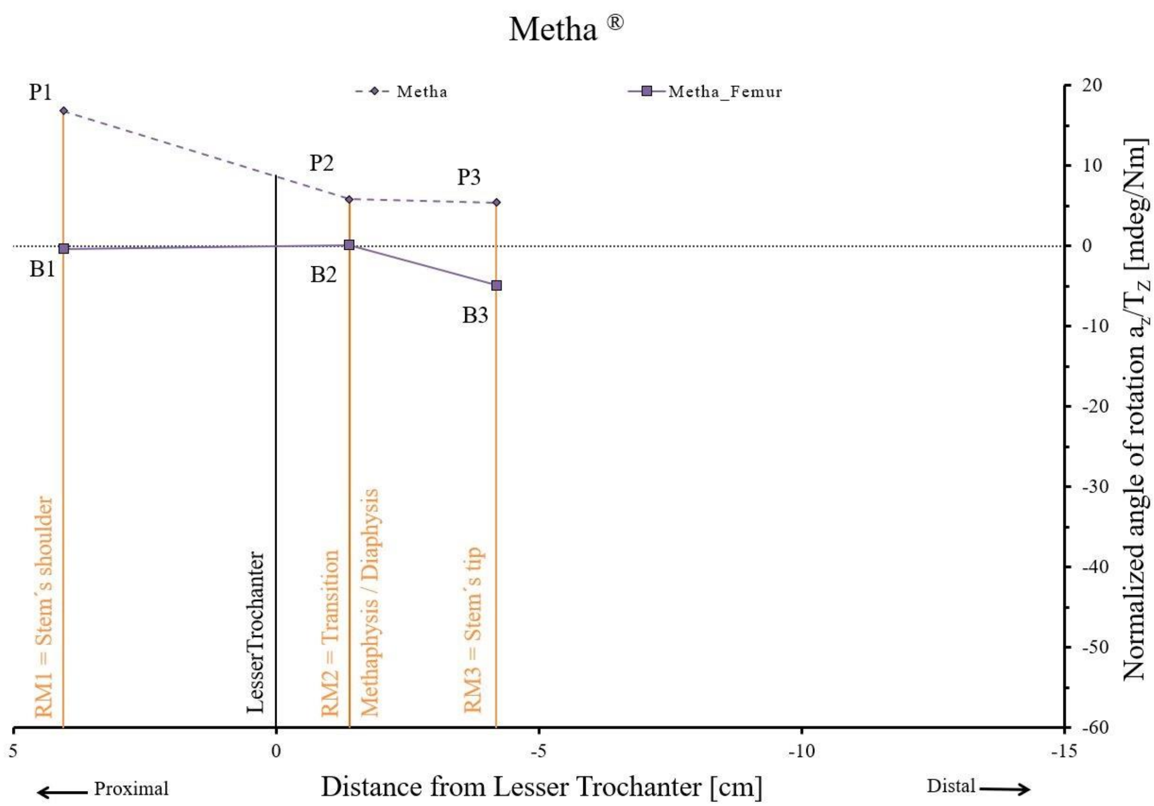

Figure 7: Summary of the results for the short stem endoprosthesis Metha®.

Straight stem TrendHip®

Table 8: Prosthetic results for TrendHip®

	P1		P2		P3	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	4.41	16.20	-1.87	7.27	-8.83	6.03
# 02	4.60	16.75	-1.50	7.73	-8.43	3.68
# 03	4.18	12.50	-1.93	2.13	-8.88	1.30
# 04	3.83	11.81	-2.00	1.40	-8.93	0.74
# 05	4.60	15.71	-1.50	5.42	-8.43	3.95
Ø	**4.32**	**14.60**	**-1.76**	**4.79**	**-8.70**	**3.14**
SD	**0.33**	**2.27**	**0.24**	**2.90**	**0.25**	**2.15**

Table 9: Synthetic bone results for TrendHip®

	B1		B2		B3		B4	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	4.41	5.53	-1.87	-2.34	-8.83	-24.98	-14.59	-38.98
# 02	4.60	-0.31	-1.50	0.10	-8.43	-24.52	-14.50	-44.75
# 03	4.18	-0.13	-1.93	0.06	-8.88	-23.16	-14.50	-46.44
# 04	3.83	0.20	-2.00	-0.10	-8.93	-25.31	-14.67	-44.48
# 05	4.60	-0.23	-1.50	0.07	-8.43	-20.48	-14.50	-43.92
Ø	**4.32**	**1.01**	**-1.76**	**-0.44**	**-8.70**	**-23.69**	**-14.55**	**-43.71**
SD	**0.33**	**2.53**	**0.24**	**1.07**	**0.25**	**1.97**	**0.08**	**2.81**

Table 10: Relative micromotion results for TrendHip®

	RM1		RM2		RM3	
Test	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]
# 01	4.41	10.68	-1.87	9.61	-8.83	31.01
# 02	4.60	17.07	-1.50	7.63	-8.43	28.20
# 03	4.18	12.63	-1.93	2.08	-8.88	24.45
# 04	3.83	11.61	-2.00	1.51	-8.93	26.05
# 05	4.60	15.94	-1.50	5.35	-8.43	24.43
Ø	**4.40**	**13.59**	**-1.77**	**5.23**	**-8.71**	**26.83**
SD	**0.21**	**2.78**	**0.23**	**3.49**	**0.25**	**2.80**

The statistical results of both bone measuring points, B1 and B2 showed a near identical value with a very minimal difference. B3 dissipated greatly when compared to those proximally located measuring points, and B4 showed a similar pattern. The prosthetic proximal measuring point P1 recorded a higher value than both distal prosthetic measuring points. The recorded value at P2 was lower and showed a shift towards its bony counterpart B2, indicating a higher anchoring characteristic. P3 also showed a comparable behavior to P2 with minimal value difference. However, it recorded the greatest value difference to its bony counterpart B3 when compared to the other measuring points. The relative micromotions recorded the highest stability at RM2 followed by RM1 and lastly RM3.

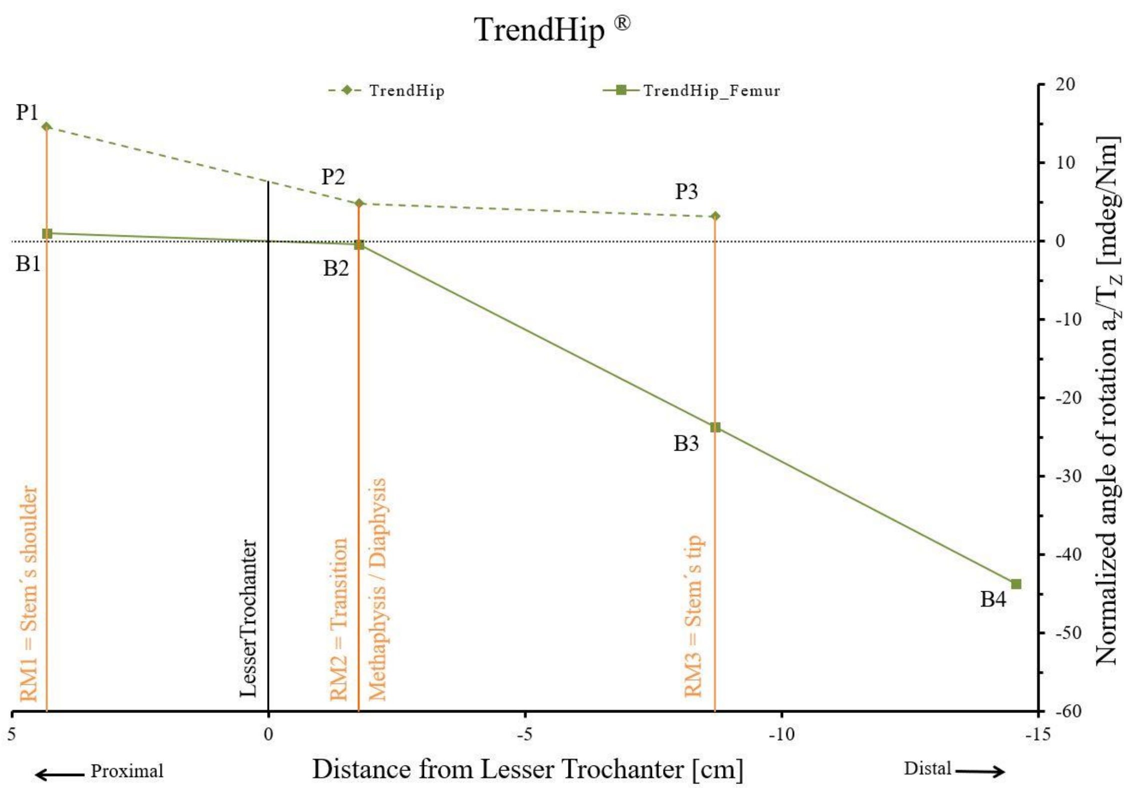

Figure 8: Summary of the results for the straight stem endoprosthesis TrendHip®.

Straight stem CLS®

Table 11: Prosthetic results for CLS®

	P1		P2		P3	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	3.69	10.73	-1.52	1.75	-10.54	1.13
# 02	3.39	9.99	-1.85	3.44	-10.74	2.50
# 03	3.43	10.33	-1.61	3.53	-10.87	-0.14
# 04	3.59	8.01	-1.48	2.26	-10.66	0.22
# 05	3.61	9.06	-1.69	3.32	-10.61	3.53
Ø	**3.54**	**9.62**	**-1.63**	**2.86**	**-10.68**	**1.45**
SD	**0.13**	**1.09**	**0.15**	**0.80**	**0.13**	**1.55**

Table 12: Synthetic bone results for CLS®

	B1		B2		B3		B4	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	3.69	0.33	-1.52	-0.14	-10.54	-21.73	-14.50	-39.51
# 02	3.39	2.00	-1.85	-1.09	-10.74	-31.60	-14.50	-45.91
# 03	3.43	1.55	-1.61	-0.73	-10.87	-21.40	-14.50	-40.77
# 04	3.59	1.30	-1.48	-0.53	-10.66	-22.84	-14.50	-40.63
# 05	3.61	1.13	-1.69	-0.53	-10.61	-25.94	-14.50	-42.69
Ø	**3.54**	**1.26**	**-1.63**	**-0.60**	**-10.68**	**-24.70**	**-14.50**	**-41.90**
SD	**0.13**	**0.62**	**0.15**	**0.35**	**0.13**	**4.25**	**0.00**	**2.52**

Table 13: Relative micromotion results for CLS®

	RM1		RM2		RM3	
Test	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]
# 01	3.69	10.40	-1.52	1.88	-10.54	22.86
# 02	3.39	7.99	-1.85	4.53	-10.74	34.09
# 03	3.43	8.77	-1.61	4.26	-10.87	21.26
# 04	3.59	6.71	-1.48	2.80	-10.66	23.06
# 05	3.61	7.93	-1.69	3.85	-10.61	29.47
Ø	**3.50**	**8.36**	**-1.66**	**3.46**	**-10.72**	**26.15**
SD	**0.16**	**1.36**	**0.17**	**1.10**	**0.17**	**5.44**

The highest recorded value at the prosthetic measurement points was at P1 compared to those distally measured. P2 recorded a low value which adhered closely to its bony counterpart B2 and thus implying good force transmission at the metadiaphyseal area. The prosthetic value P3 recoded half the value of P2. At the bony measurement point B1 and B2 showed a comparative value. However, B3 recorded a higher value and B4 recorded a little less than double the value of B3. The relative micromotions recorded the highest stability at RM2 followed by RM1 and lastly RM3.

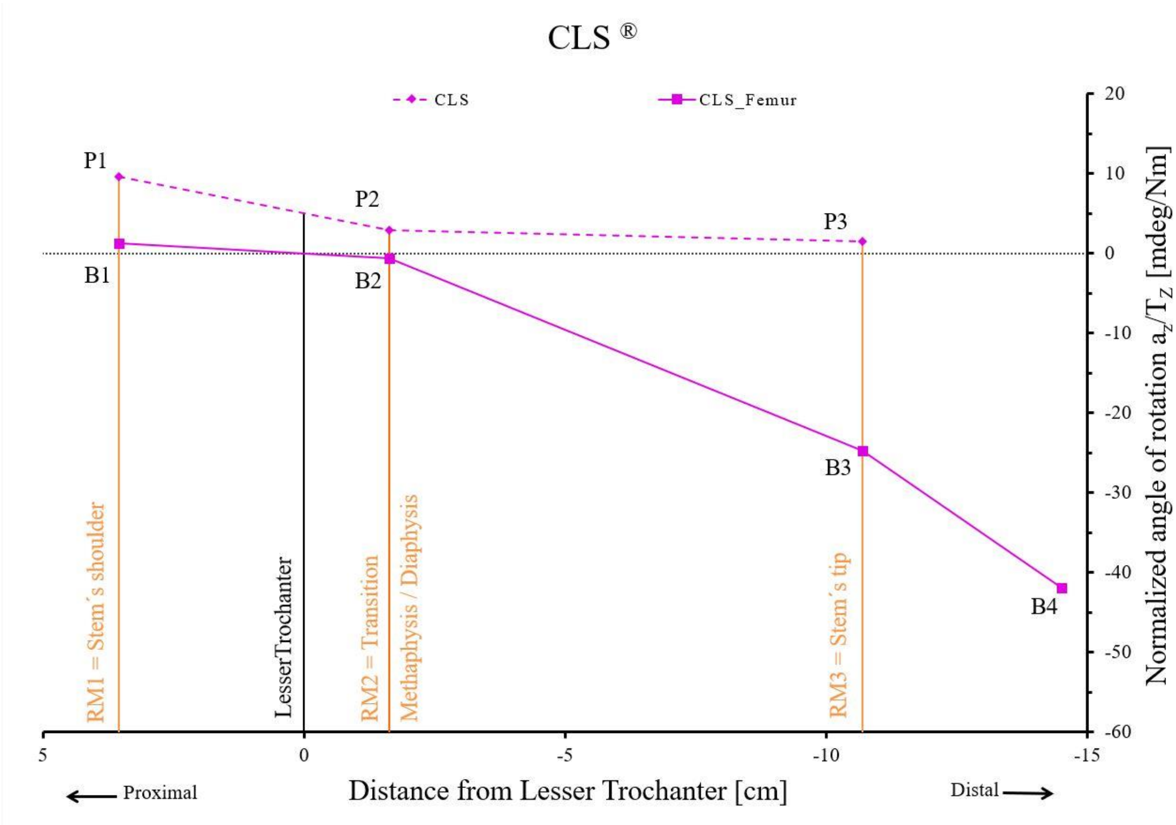

Figure 9: Summary of the results for the straight stem endoprosthesis CLS®.

Straight stem EcoFit®

Table 14: Prosthetic results for EcoFit®

	P1		P2		P3	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	4.44	16.04	-1.51	3.24	-8.59	0.85
# 02	4.60	14.56	-1.50	3.81	-8.43	1.16
# 03	4.18	12.82	-1.93	1.67	-8.88	1.71
# 04	4.39	15.40	-1.69	2.04	-8.67	1.91
# 05	4.18	14.03	-1.93	3.46	-8.88	3.42
Ø	**4.36**	**14.57**	**-1.71**	**2.85**	**-8.69**	**1.81**
SD	**0.18**	**1.25**	**0.21**	**0.93**	**0.19**	**1.00**

Table 15: Prosthetic results for EcoFit®

	B1		B2		B3		B4	
Test	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]	Height [cm]	NAR α_z/T_z [mdeg/Nm]
# 01	4.44	1.18	-1.51	-0.40	-8.59	-16.66	-14.50	-44.52
# 02	4.60	0.82	-1.50	-0.27	-8.43	-22.68	-14.50	-44.08
# 03	4.18	0.66	-1.93	-0.31	-8.88	-24.85	-14.50	-45.43
# 04	4.39	1.04	-1.69	-0.40	-8.67	-19.88	-14.50	-43.73
# 05	4.18	1.07	-1.93	-0.49	-8.88	-19.50	-14.50	-42.15
Ø	**4.36**	**0.95**	**-1.71**	**-0.37**	**-8.69**	**-20.71**	**-14.50**	**-43.98**
SD	**0.18**	**0.21**	**0.21**	**0.09**	**0.19**	**3.15**	**0.00**	**1.21**

Table 16: Relative micromotion results for EcoFit®

	RM1		RM2		RM3	
Test	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]	Height [cm]	RM $\Delta\alpha z/Tz$ [mdeg/Nm]
# 01	4.44	14.86	-1.51	3.64	-8.59	17.51
# 02	4.60	13.74	-1.50	4.08	-8.43	23.84
# 03	4.18	12.16	-1.93	1.98	-8.88	26.56
# 04	4.39	14.36	-1.69	2.44	-8.67	21.78
# 05	4.18	12.96	-1.93	3.96	-8.88	22.92
Ø	**4.41**	**13.62**	**-1.65**	**3.22**	**-8.63**	**22.52**
SD	**0.21**	**1.08**	**0.25**	**0.95**	**0.23**	**3.31**

P1 registered the highest measured value followed by P2 and P3. There was minimal difference between the measurements of P2 and P3 as well as the measured values of B1 and B2. However, B3 recorded a higher negative value. P2 showed the closest relationship to its bony counterpart B2 and thus the highest stability. The relative micromotion calculated was lowest at RM2 followed by RM1 and RM3 respectively.

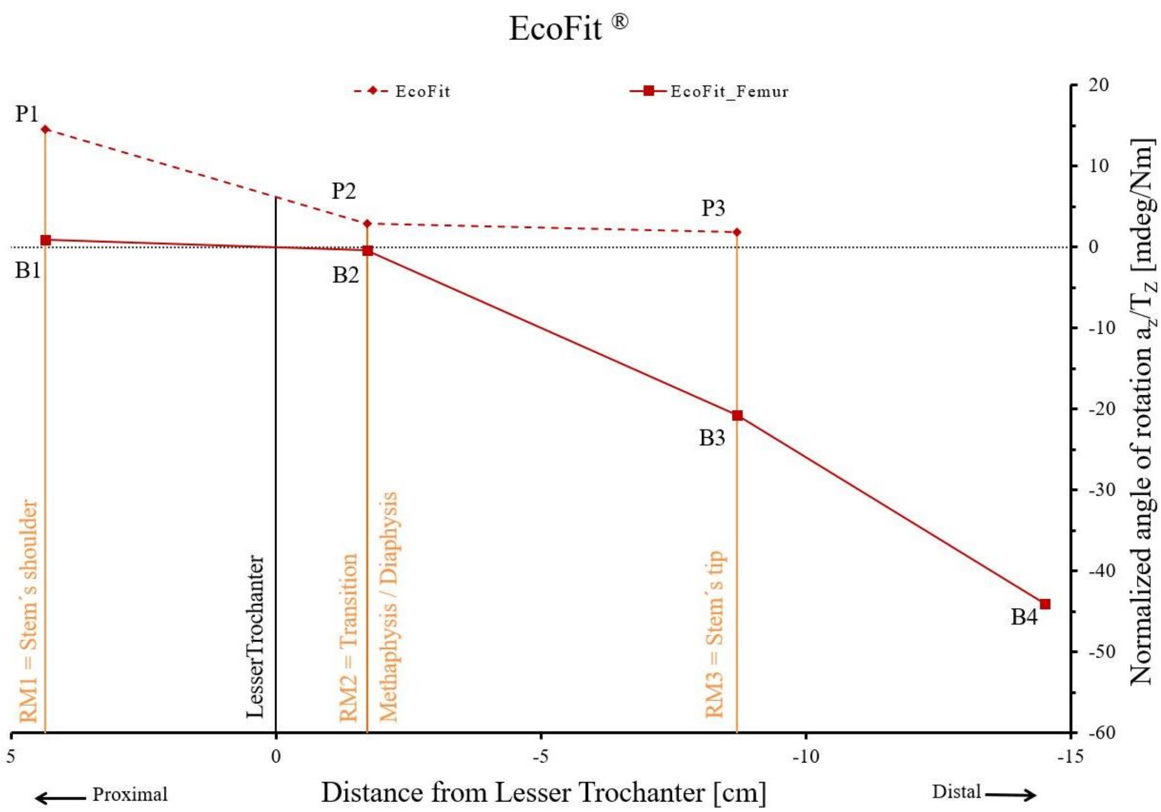

Figure 10: Summary of the results for the straight stem endoprosthesis EcoFit®.

4.2. Collective Results:

Due to different stem lengths, RM3 is calculated by interpolation depending on the distal measuring point level of the shorter Metha® stem in order to achieve a comparison between all stems. The relative micromotions RM4 of the distal stem tips of the variable straight stem lengths are also calculated by interpolation depending on the stem length of the comparatively short ECOFIT® stem, thus allowing statistical comparison between the straight stem models.

Table 17: Relative micromotions RM1-RM4 in mdeg/Nm as mean values and SD values of the tested prosthesis models.

Stem/ relative motion	AIDA® Δ αz/Tz mean (SD) [mdeg/Nm]	Metha® Δ αz/Tz mean (SD) [mdeg/Nm]	TrendHip® Δ αz/Tz mean (SD) [mdeg/Nm]	CLS® Δ αz/Tz mean (SD) [mdeg/Nm]	EcoFit® Δ αz/Tz mean (SD) [mdeg/Nm]	P-values
RM1	14.22 (2.51)[a]	17.16 (1.63)[b]	13.59 (2.78)[c]	8.36 (1.35)[a,b,c,d]	13.62 (1.08)[d]	a=0.009; b<0.001; c=0.027; d=0.026
RM2	3.42 (1.65)	5.71 (1.69)	5.23 (3.49)	3.46 (1.10)	3.22 (0.95)	-
RM3	10.01 (3.57)	14.44 (2.59)	14.36 (1.29)	11.76 (2.41)	14.44 (2.32)	-
RM4	-	-	28.36 (2.61)[a]	23.08 (5.05)[a]	25.35 (4.84)	a=0.007

* Small superscript letters indicate the p-values in pairwise comparison between the prostheses for example RM1 in Aida® was significant to RM1 in CLS®.

At the proximal prosthesis measuring point (RM1), significantly smaller relative micromotions of the CLS® prosthesis could be detected compared to all other stem models ($P < 0.05$). In all stem models, the smallest relative micromotions were found at the metaphyseal/diaphyseal measuring point (RM2), which in turn signals maximum press-fit in this area. There were no statistically significant differences between the various stem types in terms of their primary stability at the metaphyseal/diaphyseal transition (RM2) and at the level of the distal tip of the Metha® prosthesis at the measuring point (RM3). A statistically significantly lower relative micromotion could only be detected at the measuring point of the distal stem tips of the straight stems (RM4) of the CLS® stem compared to the Trendhip® stem ($P < 0.01$) (Table 17).

By means of the measuring points on the prostheses (P1-P4) and the synthetic femora (B1-B4) as well as the respective relative motions, the following motion graph was established, which represents the average anchorage characteristics of all five stems (Figure 11).

Comparing the anchoring characteristics of the individual stems with each other, it is of special interest here that all stems have their main anchoring zone in the area of the metaphyseal/diaphyseal transition at the level of the lesser trochanter. In all prostheses, the proximal stem region shows the second largest relative micromotions. All stems have the largest relative motions at the distal prosthesis tips. On the whole, all stems have a similar anchoring principle and they have an almost identical influence on the deformation of the femur (Figure 11).

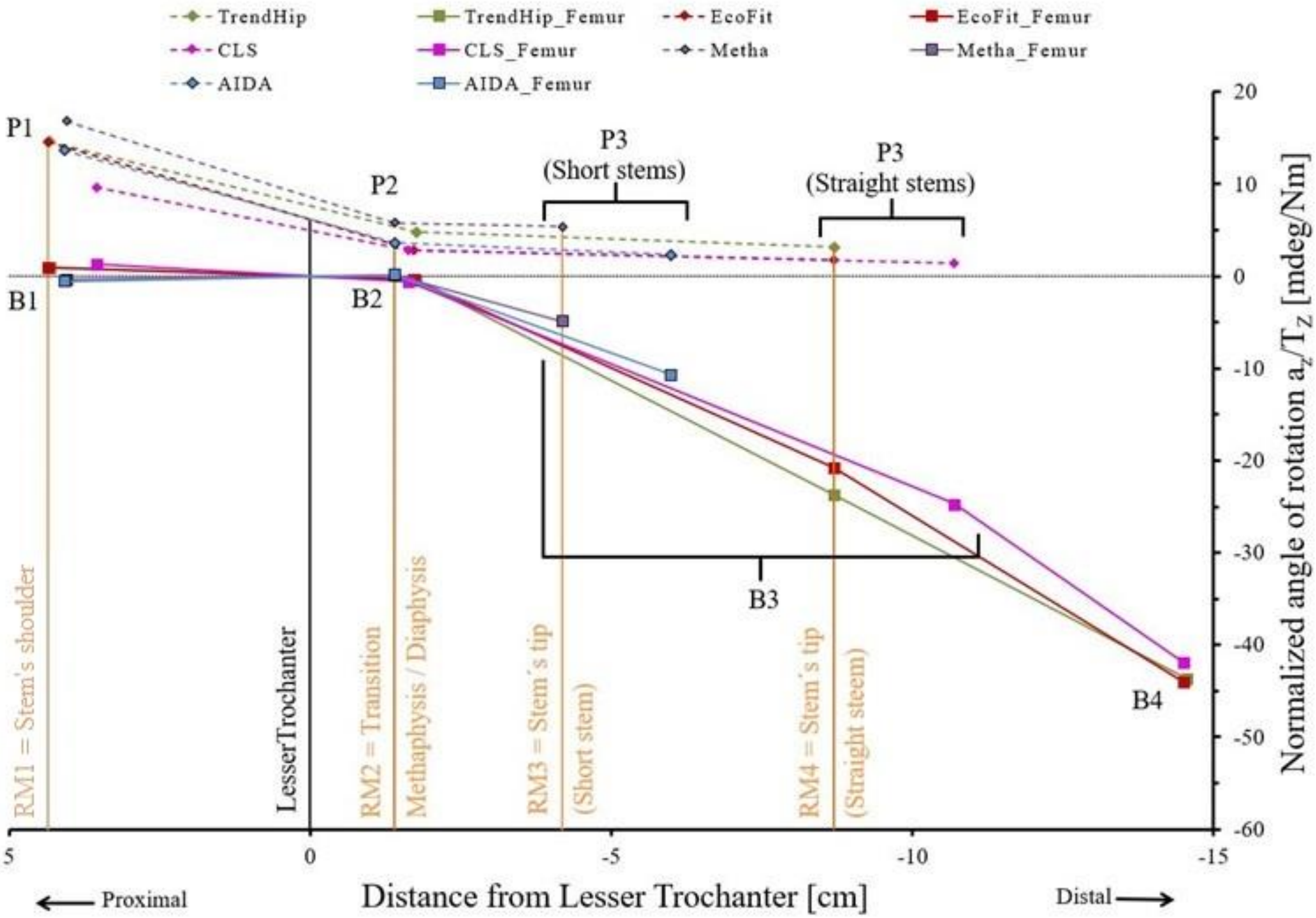

Figure 11: Summary of the results for all examined endoprosthesis.

5. Discussion

5.1. Discussion of the results

During the last decade cementless short stems have been developed and are becoming increasingly used in everyday clinical practice in order to mimic the physiological load application into the femur and to minimize the resulting bone remodeling processes. Many studies have also been carried out to investigate the advantages and disadvantages of these short stems (Jahnke et al. 2018; Augustin et al. 2018; Boller et al. 2018; Jahnke et al. 2015; Pepke et al. 2014; Jahnke et al. 2014; Floerkemeier et al. 2013; Bieger et al. 2013; Pozowski et al. 2013; Bieger et al. 2012; Fottner et al. 2009; Röhrl et al. 2006). These studies presented that the less distally anchoring short stems resulted in a more physiological and largely proximal force application compared to straight stems and counteract the reactive bone remodeling processes in the proximal femur. A direct dual-energy x-ray absorptiometry (DXA) comparison of a clinical follow-up of periprosthetic bone remodeling processes of a short stem prosthesis Fitmore® (Zimmer Biomet Holdings, Inc., Warsaw, Indiana, USA) and an established straight stem CLS® (Spotorno, Zimmer Biomet Holdings, Inc., Warsaw, Indiana, USA) also confirms that the short stem causes less reactive bone remodeling compared to the straight stem (Meyer et al. 2019).

Even with cemented prosthetic stems, there exists already an approach of shortening relatively long stems and maintaining the physiological elasticity of the femur (Hamadouche et al. 2015). Considering the results of corresponding studies, the trend obviously goes to short stems. For this reason, this biomechanical study was carried out to test and to compare established cementless straight stems with cementless short stems and to discover any similarities or serious differences between the prosthetic systems.

There are few studies that have compared straight stems with short stems. Bieger et al. investigated in an *in vitro* study the influence of a short stem prosthesis Fitmore® on proximal stress shielding compared with an already established short stem Mayo® (Zimmer Biomet Holdings, Inc., Warsaw, Indiana, USA) and a straight stem CLS®. They found that while all examined stems reduce cortical longitudinal stresses in the proximal femur, short stems tend to show better primary torsional stability (Bieger et al. 2012). In the study by Westphal et al., a biomechanical *in vitro* comparison of straight stems IPS™ and SUMMIT™ (DePuy Synthes, leeds, UK) was also performed with a short stem PROXIMA™ (DePuy Synthes, leeds, UK) in human femora. They were able to show that the short stem prosthesis is not only torsionally more stable than the straight stems but also reduces the systems' elasticity, which accounts for a more physiological load application of the short stem prosthesis. They concluded that the short stem prosthesis should minimize stress shielding due to its anchorage characteristic and to its more elastic behavior (Westphal et al. 2006). In terms of biomechanics, the cementless short stems as well as the straight stems have a predominantly metaphyseal/diaphyseal anchorage at the transition point in the area of the lesser trochanter with an acceptable mean total motion, which in turn promotes physiological metaphyseal force transmission and reduced proximal stress shielding. Kim et al. reviewed the long-term results of the short stem IPS® (DePuy Synthes, leeds, UK) in 500 patients under the age of 65. They observed no thigh pain in any patient and little stress shielding (Kim et al. 2014). The authors findings support the hypothesis that a short, tapered and polished distal stem minimizes stress shielding related proximal femoral resorption and reduces symptomatic manifestations by avoiding a diaphyseal fixation.

The examined two short stems AIDA® and Metha® show no signs of anchorage and should thus reduce proximal stress shielding. This behavior is also evident in the straight stems, among which the distally uncoated ECOFIT® straight stem must be highlighted once more since thanks to its polished stem tip it is unlikely to osseointegrate distally.

In another study, the same group examined the CBC® (Mathys AG, Bettlach, Switzerland) straight stem and compared it *in vitro* with the Optimys® (Mathys AG, Bettlach, Switzerland) short stem for primary stability in the human femur. They were able to show that the shortened stem shows a primary stability comparable to the straight stem. Moreover, the examined short stem caused cortical stresses in the femur which were more physiological than those of the straight stem (Bieger et al. 2013).

Nadorf et al. studied the new GTS™ (Zimmer Biomet Holdings, Inc., Warsaw, Indiana, USA) short stem compared to the established CLS® straight stem *in vitro* in terms of primary torsional and tilting stability. They were able to demonstrate that despite the shorter stem length the GTS™ showed an anchoring and deformation behavior nearly identical to the CLS® straight stem and thus they could not express clear preference for a particular prosthetic stem design (Nadorf et al. 2014).

In view of our results, it can be seen that the established straight stems as well as the short stems have almost identical anchoring behavior and that our results are in line with those indicated in the literature (Hamadouche et al. 2015; Nadorf et al. 2014; Bieger et al. 2013; Bieger et al. 2012; Westphal et al. 2006). The rotational stability of the Metha® as well as the AIDA® short stem shows a pronounced press-fit at the metaphyseal/diaphyseal transition point in the area of the lesser trochanter. At the prosthetic shoulder in the proximal region, there are slightly more relative micromotions, but they don't exceed an acceptable range. In case of the Metha® as well as the AIDA® short stem, the prosthesis tip in the distal region has the greatest relative motion in the bone/prosthesis interface. However, due to the uncoated and polished stem tip of both short stems this cannot be interpreted as criterion of instability, nevertheless it can even be a desirable criterion, so that in the diaphysis no osseointegration of the implants can occur. The rotational stability between short and straight stem endoprostheses in hip arthroplasty was comparable. Therefore, no relevant mechanical advantages of either type of prosthesis could be demonstrated.

Comparing the anchoring characteristics of the short stems with the anchoring characteristics of the straight stems, it is obvious that the anchoring characteristics of the short stems are remarkably similar to that of the straight stems. All prostheses have a clear anchorage at the metaphyseal/diaphyseal transition point around the trochanter minor. This anchoring characteristic is in turn illustrated by the relatively steep negative slope of the straight line connecting P1 and P2 and the relatively flat negative slope of the straight line between P2 and P3 resulting from the torsion of the stems.

5.2. Implications for the Practice

Since short and straight stem endoprostheses did not show clinically relevant differences in rotational stability, short stem endoprosthesis should be given priority in practice, unless contraindicated. Choi and Kim similarly showed good follow-up results (average follow-up time: 4.6 years) after implantation of short stem Proxima™ endoprosthesis. The authors examined 47 patients (56 implants, male: n = 20, average age: 54 years). Neither prosthetic loosening nor osteolysis could be demonstrated in the follow-up (Choi and Kim 2016). In a study by Kim et al. of 500 patients (630 implants, average age 52.7 years, mean follow-up: 15.8 years), showed that a stable fixation of the prosthesis can also be achieved with the short stem IPS™ endoprosthesis. The 15-year survival rate in the collective was 98.7% for the acetabular component and 99.4% for the femoral component of the prosthesis. Osseointegration was demonstrated in all examined prostheses. The infection rate was 0.6%. None of the patients examined complained of thigh aches. Based on these results, Kim et al. assessed that acceptable long-term outcomes could be achieved with the short-stemmed endoprosthesis, especially in patients under the age of 65 years with good bone quality (Kim et al. 2014). Similar results were reported by Cinotti et al., who were unable to demonstrate deterioration of the implants stability in the 9-year follow-up for the short stem prosthesis IPS™ and recommended this prosthesis primarily for patients with good bone structure (Cinotti et al. 2013). Yu et al. demonstrated that stable fixation of the short stem prosthesis can be achieved even in elderly patients (mean age: 75 ± 10.4 years). Schmidutz et al. compared the biomechanical properties of straight stem prostheses and short stem endoprostheses for a total of 100 patients. Included in the clinical investigation were patients with first interventions. Exclusion criteria included hip dysplasia, infection, trauma, and malignant pelvic disease. The short stem was Metha® as in the study presented here. The femoral offset was higher in the short stem as compared to the straight stem prosthesis, but the values were in tolerance range. The authors investigated whether short stem endoprosthesis can be used to achieve adequate reconstruction of the biomechanics because implantation of the short stem endoprosthesis is determined by the height of the resection level on the femur. Furthermore, short stems had a clinically relevant influence on leg length due to the higher resection level on the femur (Schmidutz et al. 2012a).

5.3. Limitations of the study

Synthetic femora are quite different from human femoral preparations. However, using these synthetic femora resulted in a pure mechanical representation which allowed a high standardization to be achieved. Therefore, a high reproducibility of the measured values were generated, which in turn support the validity of our study. However, statements about periprosthetic remodeling processes due to osseointegration cannot be made. Our results should therefore be regarded as indicative and are to be validated based on clinical studies. Nevertheless, the low number of examined samples (n = 5) repeated measures must be mentioned as such a small sample size is expected to distort the statistical results.

However, the same measuring method was applied in previous biomechanical *in vitro* studies, which supports the validity of our measuring method as well as our measuring results (Jahnke et al. 2016; Hamadouche et al. 2015; Kinkel et al. 2015).

In practice, the rotational stability of the short and straight stem prostheses on the patient or the biological bone must be re-evaluated. Only in this way can a more comprehensive statement be made about the stability of the individual prostheses in the clinical setting.

6. Conclusion

All the investigated stems generally show a comparable anchoring pattern and an almost physiological, metaphyseal force application onto the area of the metaphyseal/diaphyseal transition in the region of the trochanter minor. The area of the diaphysis registered the highest relative micromotions in all stem models, which would counteract undesirable osseointegration. Since the comparatively straight stems demonstrated a nearly identical anchoring pattern to that of the examined short stems, shortening of the established straight stems could be taken into consideration. This could offer the advantages of minimally invasive surgery and bone-saving resection as well as preservation of cancellous bone in the metaphyseal and diaphyseal areas of the femur in case a revision is necessary.

However, the question remains open why established straight stems are not shortened and adapted in order to achieve the mentioned advantages of short stems.

Further *in vitro* studies should be undertaken and validated in clinical studies to examine how much the straight stems could be shortened and if the shortened straight stems will retain adequate primary stability and anchoring pattern.

7. Abstract

7.1. Summery

Background: The rate of hip joint osteoarthritis and associated endoprosthetic replacement is increasing. This is due to the increased life expectancy in modern nations. Additionally, younger patients requiring prosthetic surgery are becoming more prevalent. Straight stem prostheses are well established in the literature and showed good survival rates. However, because of the clinical advantages of short stem prostheses, this study aims to evaluate the biomechanics of the mentioned systems *in vitro* and to compare the anchoring pattern of straight versus short stems.

Methods: A total of 5 systems were examined, Two short and three straight stems prostheses. The examined prostheses included straight stem systems CLS®, EcoFit®, and TrendHip® as well as the short stem prostheses Aida® and Metha®. Each system was implanted in five synthetic femora. Afterward, torque was applied in the ventrodorsal direction continuously with an interval of ± 7Nm. Six inductive extensometers on four measurement levels were utilized to acquire a statement about micromotion between the endoprostheses and synthetic bone.

Results: The results showed that all endoprostheses regardless of length exhibited similar anchoring pattern. However, CLS® showed significantly less micromotion ($P<0.05$) at the proximal measuring point compared to the other systems. All systems registered the lowest relative micromotion at the metaphyseal/diaphyseal measuring point. However, none of the prostheses exhibited significant and quantitatively significant differences in most of the measurements. Therefore, neither the short nor the straight stem endoprostheses had an advantage in rotational stability. The deformation properties between short and straight endoprostheses were comparable.

Conclusion: Due to the insignificant differences in anchoring patterns between short and straight stem endoprostheses. The short stem endoprostheses offer a less invasive procedure, require less bone resection and are advantageous in case a revision is needed. These benefits can also be attained by shortening straight stem prostheses.

Keywords: straight stem systems, CLS®, EcoFit®, TrendHip®, short stem prostheses, Aida®, Metha®, rotational stability, biomechanics.

7.2. Zussamenfassung

Hintergrund: Die Rate der Hüftgelenkarthrose und des damit verbundenen endoprothetischen Ersatzes nimmt zu. Dies ist auf die gestiegene Lebenserwartung in modernen Nationen zurückzuführen. Darüber hinaus werden jüngere Patienten, die eine prothetische Operation benötigen, immer häufiger. Geradschaftprothesen sind in der Literatur gut etabliert und zeigten gute Überlebensraten. Aufgrund der klinischen Vorteile von Kurzsschaftprothesen zielt diese Studie jedoch darauf ab, die Biomechanik der genannten Systeme *in vitro* zu bewerten und das Verankerungsmuster von geraden und kurzen Shäften zu vergleichen.

Methode: Es wurden insgesamt 5 Systeme, zwei kurz-und drei gerad- Shaftprothesen untersucht. Zu den untersuchten Prothesen gehörten die geraden Shaftsysteme CLS®, EcoFit® und TrendHip® sowie die Kurzshaftprothesen Aida® und Metha®. Jedes System wurde in fünf synthetische Femora implantiert. Danach wurde das Drehmoment in ventrodorsaler Richtung kontinuierlich mit einem Intervall von ± 7Nm aufgebracht. Sechs induktive Extensometer auf vier Messebenen wurden eingesetzt, um eine Aussage über die Mikrobewegung zwischen Endoprothesen und synthetischem Knochen zu erhalten.

Ergebnisse: Die Ergebnisse zeigten, dass alle Endoprothesen unabhängig von der Länge ein ähnliches Verankerungsmuster aufwiesen. CLS® zeigte jedoch am proximalen Messpunkt deutlich weniger Mikrobewegung ($P<0,05$) als die anderen Systeme. Alle Systeme registrierten die niedrigste relative Mikrobewegung am metaphysealen/diaphysären Messpunkt. Keine der Prothesen wies jedoch signifikante und quantitativ signifikante Unterschiede in den meisten Messungen auf. Daher hatten weder die kurzen noch die geraden Shaftendoprothesen einen Vorteil in der Rotationsstabilität. Die Verformungseigenschaften zwischen kurzen und geraden Endoprothesen waren vergleichbar

Schlußfolgerung: Aufgrund der geringeren Unterschiede in den Verankerungsmustern zwischen kurzen und geraden Shaftendoprothesen. Die Kurzsschaftndoprothesen bieten ein weniger invasives Verfahren, erfordern weniger Knochenresektion und sind vorteilhaft für den Fall, dass eine Revision erforderlich ist. Diese Vorteile können auch durch die Verkürzung gerader Stielprothesen erreicht werden.

Schlüsselwörter: Geradschaftprothese, CLS®, EcoFit®, TrendHip®, Kurzschaftprothese, Aida®, Metha®, Rotationsstabilität, biomechanich.

8. List of Abbreviations

B1-B4	Bone measurement points 1- 4
BVO	Berufsverband der Ärzte für Orthopädie (professional association of physicians for orthopedics)
DDH	developmental dysplasia of the hip
DXA	Dual-energy X-ray absorptiometry
EPRD	Endoprothesenregister Deutschland (The German endoprosthetic registry)
FAI	Femuroacetabular impengment
HIV	Human immunodeficiency virus
kN	Kilonewton
LVDT	lineare variable differential transformator
max	Maximum
md	Micromotion at the distal implant
min	Minimum
mp	Micromotion at the proximal implant
N	Newton

NAR	Normalized angle of rotation
Nm	Newtonmeter
NSAIDs	Non-steroidal anti-inflammatory drugs
OA	Osteoarthritis
P	Significance
P1-P3	prosthetic measurment point 1 - 3
RM	Relative micromotion
SD	Standard deviation
THA	Total hip arthroplasty
THR	Total hip replacement
TLR4	Toll-like receptor 4

9. List of Figures

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11. References

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List of publications:

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Erklärung zur Dissertation

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