
2.1

The Historical Perspective of Hip Biomechanics

The biomechanics of the hip joint has been of great interest to researchers and clinicians since the early days of anatomical studies. Julius Wolff addressed the relation between the inner architecture of the bone and the functional loading already in the nineteenth century [31] and Friedrich Pauwels built the foundation for a mechanical approach to understand joint loading 65 years later [24]. Both researchers, despite dealing with very different questions (bone remodeling vs. fracture mechanics), are good examples for the spread of the biomechanics field. It can be defined as the science concerned with the internal and external forces acting on the human body and the effects produced by these forces [13]. Pauwels elaborated on the influence of valgus (steep) and varus (flat) anatomical position of the femoral neck, demonstrating for a given joint position, that a valgus neck is associated with a smaller lever arm of the abductor muscles and larger abductor muscle forces. This increases the magnitude of the resultant hip joint force and also changes its point of action in the pelvis to a more lateral position. His findings influenced the treatment of femoral neck fractures and femoral osteotomies in a major way. The biomechanical situation is more complicated when applied to Total Hip Replacement (THA) since all joint parameters are influenced by the operation: joint center, neck angle, offset, lever arms, and the range of motion until impingement. Range of motion and joint stability are decisive issues, especially in younger patients with high expectations on their quality of life after THA. The varus and valgus situation as well as the hip joint center are determined by the position of the implant in the pelvis and femur. This positioning also influences the local loading situation at the implant component – bone interface. For example, a slightly superior, posterior, and medial hip joint center after replacement can be associated with markedly higher joint forces (Fig. 2.1).

M.M. Morlock (✉), N. Bishop, and G. Huber
Biomechanics Section, TUHH Hamburg University of Technology,
Denickestrasse 15, Hamburg 21073, Germany
e-mail: morlock@tuhh.de

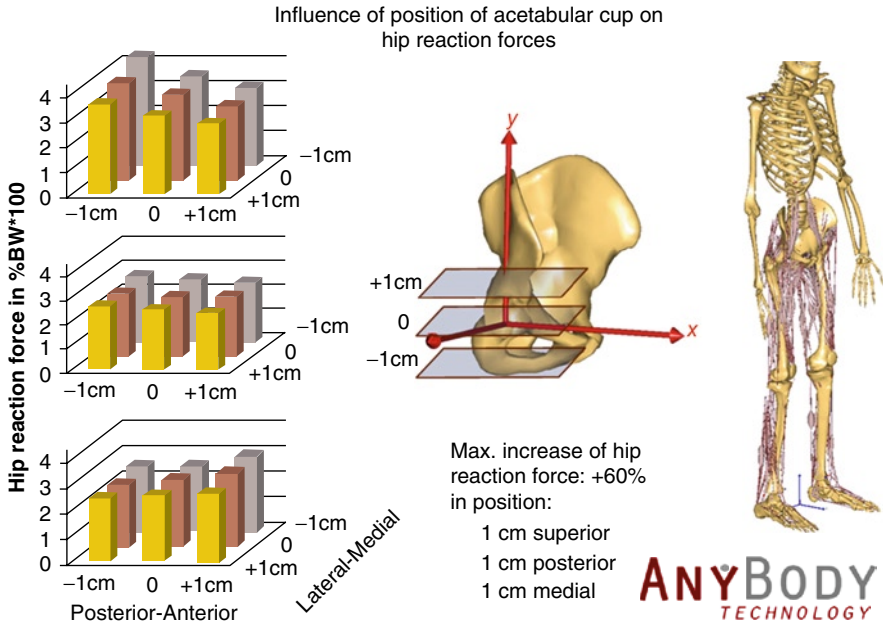


Fig. 2.1 Simulation of the influence of acetabular cup position on the resultant hip joint reaction force (Courtesy of Sebastian Dendorfer, Anybody technology)

In vivo measurements with an instrumented femoral stem showed that dynamic hip joint forces lie in a similar range to those calculated by Pauwels with his static approach. However, for high body weight (BW) and an unfavorable loading situation (25% higher than average), peak forces during walking ($3.9 \times \text{BW}$), stair climbing ($4.2 \times \text{BW}$), and stumbling ($11 \times \text{BW}$) are substantial [5]. These high forces can be a factor in the failure scenario of hip joint replacements, especially if implant specific characteristics (e.g., small surface area) additionally increase the stress at the interface between bone and prosthesis components. The same applies to the junctions of modular prostheses systems. Aggravating in this context is also the continuously increasing body weight in most of the industrial countries. Not only the forces, but also the motion of the joint can play a role in failure scenarios of THA. The range of motion (RoM) at the hip utilized during normal daily activities is already quite substantial: flexion/extension can reach up to 124° , abduction/adduction up to 28° , and internal/external rotation up to 33° [14]. The athletic activities being performed by some patients with THA spans from running, cycling, kick boxing, alpine skiing to free climbing – activities, for which the RoM is most certainly higher.

In this chapter, an overview of the important factors influencing the function and longevity of THAs is attempted. For the identification of these factors, the analysis of the failure reasons for THA is helpful: The National Joint Replacement Registry of the Australian Orthopaedic Association lists “loosening/lysis of a prosthesis component” as the main (30%) and “dislocation” as the second most frequent indication (28%) for revision surgery [3]. Consequently, the emphasis is put on those biomechanical aspects related

to design and implantation procedure, which directly or indirectly influence the occurrence of loosening, lysis, or dislocation. These aspects (in no particular order) are:

- Range of motion
- Impingement implant fixation
- Tissue damage during implantation and tissue tension after THA
- Component orientation (stem, cup)
- Bearing material

The aspect “bearing material” is not addressed in this chapter since it is extensively covered in another chapter in this book.

2.2

Range of Motion

Range of motion is influenced heavily by prosthesis design. The number of different designs of THA prostheses used is very large and can only be roughly estimated at 10,000–100,000. Femoral components vary with regard to material, length, diameter, shape, surface structure, surface coating, fixation, and stem modularity (Fig. 2.2). All femoral components have in common that the ball head (either modular or monobloc) articulates with the acetabular component. A wide variety of head diameters ranging from 22.25 mm to approximately 60 mm is available (Fig. 2.2). Acetabular components vary mostly with respect to the fixation mechanism and the bearing material, whereas the shape of the



Fig. 2.2 Contemporary and historical femoral components documenting the wide variety of designs with head size diameters ranging from 62 to 22 mm (*left to right*: Zweymüller, Exeter, Corail, St. Georg, Silent & Resurfacing, CFP, Meta, Chamley)

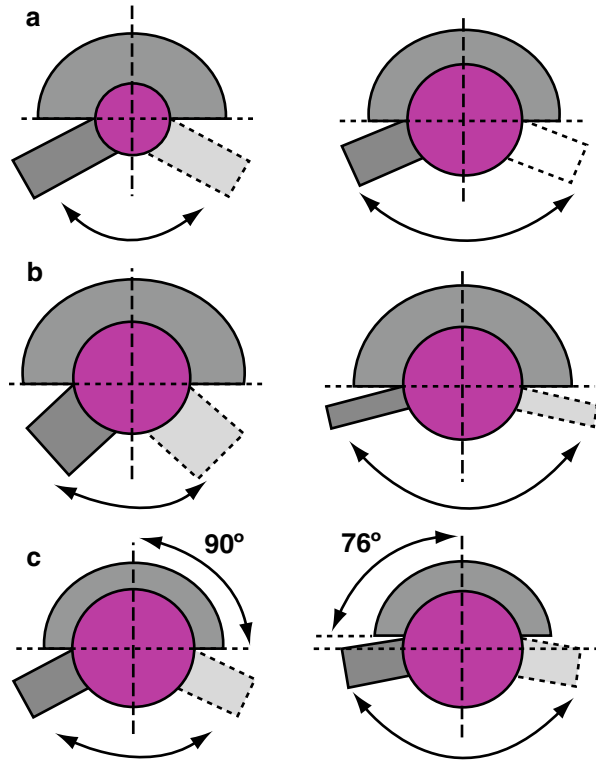


Fig. 2.3 View of the backside of new and revised acetabular components; from left to right (*columns*): press-fit uncemented cups, cemented cups, threaded (*bottom*: expansion) cups, monobloc cups

acetabular cups is quite invariable, being hemispherical (sub-hemispherical) or conical in nature (Fig. 2.3). The possible head size is restricted by the outer diameter of the cup (i.e., the anatomical situation) and the required combined thickness of the cup shell and the bearing insert. In monobloc cups, shell and bearing insert are one piece and are, consequently, made from the same material (typically Polyethylene in the cemented and Cobalt-Chromium alloy in the uncemented case).

Head size directly influences the technical (theoretically possible) RoM. Increasing the head size from 28 mm to 36 mm yields an increase of 13° in the technical RoM (from 123° to 136°). This is derived for a hemispherical cup and a modern 12/14 mini taper completely embedded in the head (Fig. 2.4a) and a slender neck design (proximal diameter smaller than the distal diameter of the taper). The technical RoM is not directly related to the active or passive RoM achieved by the patient. This “true” RoM of the patient is heavily influenced by the orientation of the components, the muscular and soft tissue situation as well as the patient characteristics. Biomechanically most important is the position of the femur with respect to the pelvis, in which the end of the RoM is reached and the prosthesis neck “impinges” on the cup or impingement occurs somewhere else between femur and pelvis. Impingement can lead to sub-luxation or even dislocation of the hip joint. If impingement occurs repeatedly in positions inside the RoM required by the patient for either daily or athletic activities, dislocation is rather probable. In this situation, revision of the prosthesis (or at least one component of the prosthesis) is frequently required. The “jumping distance,” which is the distance the head has to “jump” before leaving the cup, amounts in hemispherical cups to 50% of the head diameter. In sub-hemispherical cups, the distance is respectively less.

Fig. 2.4 Design aspects with direct influence on the range of motion [22]. (a) head size (left: small, right: large) (b) taper diameter (large, small) (c) cup entrance plane (hemispherical, sub-hemispherical)



The advantage of larger heads with respect to the RoM and increased jumping distance is counter balanced by the higher friction moments, which have to be supported by the fixation of the bearing components. The friction increase with head diameter is pronounced for metal-on-metal articulations and worsened by the negative effect of resting periods on start-up friction with this material [8, 20]. The increased friction could be one of the important factors for the observed problems with cup loosening [17, 18] and taper corrosion [11]. A second disadvantage of larger heads is the higher separation of the joint that has to be achieved when relocating the head into the acetabulum. This separation corresponds to the jumping distance (in hemispherical cups). Consequently, the forces required to relocate the joint increase with larger heads. A weakening or damage of soft tissue structures can be the consequence.

Considering the advantages and disadvantages of large heads, the important question becomes: How large does it have to be? This question can only be answered by evaluating the clinical results. Nearly all publications document a decrease in the dislocation rate for an increase in head diameter. The absolute numbers, however, are quite different. For heads with a 28 mm diameter, they range from 0.6% [2] to 2.5% [25] or even 3.0% [6], 3.1% [1] and 3.6% [23]. For smaller head diameters the range is even wider: 3.8% [6] to 18.8% [23] for a 22 mm head. For larger head diameters, the rates are very low: for heads with 32 mm diameter only 0.5% [1], for 38 mm even 0.0% [25]. This indicates that the head diameter itself is only partly responsible for the dislocation rate. Implant position and

soft tissue tension achieved by the surgeon are probably equally or even more important: “The theoretical gain in stability obtained by using a large femoral head (above 36 mm) is negligible in cases where there is a high cup abduction angle [27].”

Considering the pros and cons of large and extra-large heads, it is proposed that the head diameter is limited to about 36 mm for primary hip arthroplasty; in the case of Polyethylene possibly even to 32 mm, since for hard-soft bearings wear increases with head diameter. An initiative supporting and spreading this proposal was founded in 2008 by Carsten Perka from the Charité in Berlin and the first author of this paper and called “the 36 and under club.”

The geometry of the stem taper is also an important aspect for the technical RoM: thinner tapers impinge later with the cup (Fig. 2.4b). The same applies to the neck geometry. Resurfacing of the femoral head is the extreme example for an extra-large diameter head with a thick neck. This explains why the technical RoM of resurfacing is about 31° – 48° below the RoM of a stemmed prosthesis with a 32 mm head [16]. Available taper sizes range from 8/10 to 14/16 – the numbers correspond roughly to the proximal and distal diameter of the taper in mm. Thinner tapers have the disadvantage that the torque required to loosen the head on the taper decreases, which can cause disadvantageous rotation of the head with respect to the taper in high friction situations. It should be emphasized that tapers are not standardized, which makes the replacement of the head in a revision situation challenging. This problem is enhanced by the multitude of available sizes (8/10, 9/11, 10/12, V40, 11/13, C-taper, 12/14, 14/16).

A further design aspect important for the technical RoM is the location of the entrance plane of the cup and the cup profile [22]. In a typical hemispherical cup design with a hemispherical bearing liner, the center of rotation lies in the middle of the entrance plane and impingement occurs, when either the taper or the neck of the prosthesis come into contact with the implant (Fig. 2.4c). In implants with an elevated liner, the center of rotation lies below the cup entrance plane, as such reducing the RoM. The opposite effect is achieved by sub-hemispherical cup designs. In these designs, the cup only spans about 152° – 166° instead of 180° [12]. In such designs, the center of rotation lies above (outside) the cup entrance plane [9]. The RoM is increased in sub-hemispherical cups, since impingement occurs later (Fig. 2.4c). The downside to this design variation is the decreased bearing surface, which is one of the factors made responsible for the increased wear in poorly functioning metal-on-metal prostheses.

2.3

Implant Fixation

THA implant components are fixed in the bone either by using cement or in an uncemented way with or without additional structures (e.g., screws). The choice of fixation method in THA varies greatly between countries: about 75% of all THAs in Sweden (2008), 51% in England & Wales (2003–2009) and 6.3% in Australia (2009), were implanted using cement for both components [3, 15, 21]. This indicates that the choice of fixation depends on many factors such as heritage of the surgeon, bone quality, and age among several others.

Table 2.1 Revision rates for the three commonly used fixation methods in THA

Revision rates by prosthesis type at one, three and five years for primary hip replacement procedures, undertaken between 1st April 2003 and 31st December 2009, which were linked to a HES/PEDW				
Prosthesis type	Number of patients	Revision rates (95% CI)		
		One year	Three years	Five years
Cemented	99,359	0.6% (0.6% to 0.7%)	1.4% (1.3% to 1.5%)	2.0% (1.8% to 2.1%)
Cementless	62,937	1.3% (1.2% to 1.4%)	2.5% (2.4% to 2.7%)	3.4% (3.2% to 3.7%)
Hybrid	31,662	0.9% (0.8% to 1.0%)	1.8% (1.6% to 1.9%)	2.7% (2.4% to 3.0%)

Adapted from [21]

The success of cementing relies heavily on the cementing technique, which has been continuously improved over the last two decades: vacuum-mixed cements, medullary canal plugs, centralizing elements, and the use of jet-lavage to clean the trabecular bone structures have been shown to effectively prolong the service life of prostheses [10]. Presently, cemented fixation still shows statistically the best results in terms of the whole THA population (Table 2.1). This changes when young and active patients are involved; in this patient collective, cemented prostheses do not perform as well as in the older population. This is the reason, why uncemented or hybrid fixation is common in this group.

The success of uncemented fixation depends on the ingrowth of bone. The ingrowth of bone is only possible, if the patient's activity-induced relative interface motion (micromotion) remains below a critical threshold in the early postoperative period (primary stability). Micromotion is induced by loading of the hip joint, which is almost impossible to avoid. In vivo measurements indicate that even static activities, such as lying in bed or working against resistance provided by the physiotherapist, create hip joint forces comparable to those occurring during unsupported walking [4]. The findings of Bergmann have challenged the advice of surgeons to patients to avoid full weight bearing and physical activities for the first few weeks after surgery. As a consequence, today only few surgeons still insist on partial weight bearing following total hip arthroplasty. It would appear that the quality of the initial fixation achieved by the surgeon, the characteristics of the implant surface, and the quality of the reamed or broached bony bed are more critical factors in achieving successful bone ingrowth than the influence of patient loads. The most frequently used method in uncemented implantations is "press-fitting." This method involves impaction of an implant into a cavity that is slightly smaller than the implant. The amount of implant oversizing is crucial in this context: too much oversizing makes seating of the implant difficult, requires high forces during implantation – which might cause fractures or fissures of the bone, and results in small contact areas between implant and bone [29]. Too little oversizing might result in a condition, in which the implant can move with respect to the bone, especially during situations with low compressive forces in the joint. Such a situation on the cup side is, for example, the reduction of the joint after implantation of the implant components, when the head is moved over the rim of the cup.

2.4

Tissue Damage and Joint Tensioning

The amount of soft tissue damage during surgery and the tension in the remaining soft tissue after THA implantation are also important factors for the stability of the joint. It has been shown that the surgical approach influences the dislocation rate [6, 19]. Other studies, however, investigated the function of the hip joint after different approaches and did not show a functional difference [26]. The controversy might be due to the complex situation: The surgical approach by itself determines which muscle groups have to be cut or detached or split by the surgeon. The surgeon, however, determines the extent of the involvement of the soft tissue structures. Furthermore, the positioning of the implant components, in order to reproduce the anatomical joint centre and offset, heavily influences the biomechanical situation in the joint during loading. This might explain why two different surgeons can achieve different results with respect to dislocation rate with the same surgical approach.

Hard tissue damage can also occur during implantation. This can either result in direct complete fractures or fissures of the bone, or in micro-fractures of trabecular bone, which can develop to a complete fracture later on (Fig. 2.5).

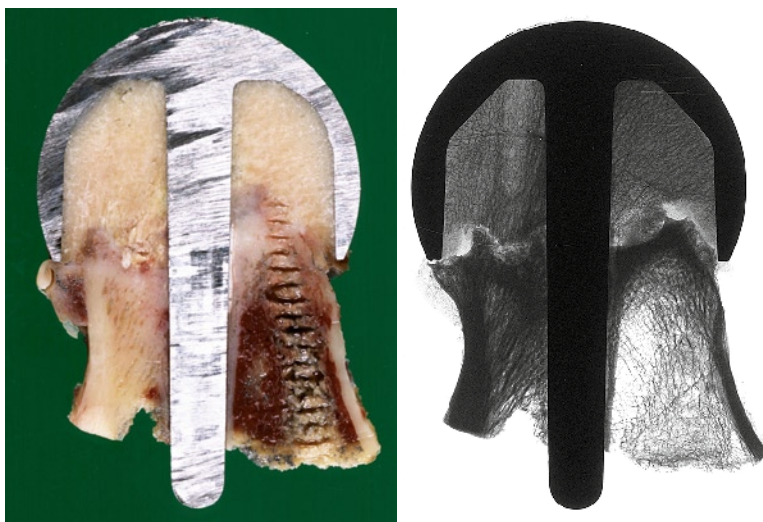


Fig. 2.5 Histological section and contact X-ray of a revised resurfacing prosthesis (1,374 days after implantation). The fracture gap across the femur-head junction demonstrated pseudarthrotic tissue. The fracture might have been initiated by high impaction forces during surgery, which were required to seat the implant due to the massive amount of cement used (preparation and section courtesy of Drs. Joseph Zustin and Michael Hahn, University Hospital Hamburg)

2.5 Component Orientation

Component orientation and position is probably the most important biomechanical aspect for the tribological and functional success of a THA procedure. The material and manufacturing issues, which have been the limiting factor for the success of the procedure in the past, have been successfully addressed in the last two or three decades.

On the acetabular side, poor component position directly influences friction, wear, and the risk of dislocation due to the reduced effective jumping distance [18, 27, 28]. In metal-on-metal articulations, run-away wear (Fig. 2.6a) with all the possible biological implications such as metallosis or pseudo-tumors can be the consequence. Malpositioning in large metal-on-metal bearings usually involves cup inclinations above 50° and/or anteversion above 15° (Table 2.2). In Ceramic-on-ceramic bearings, rim loading can cause stripe wear,

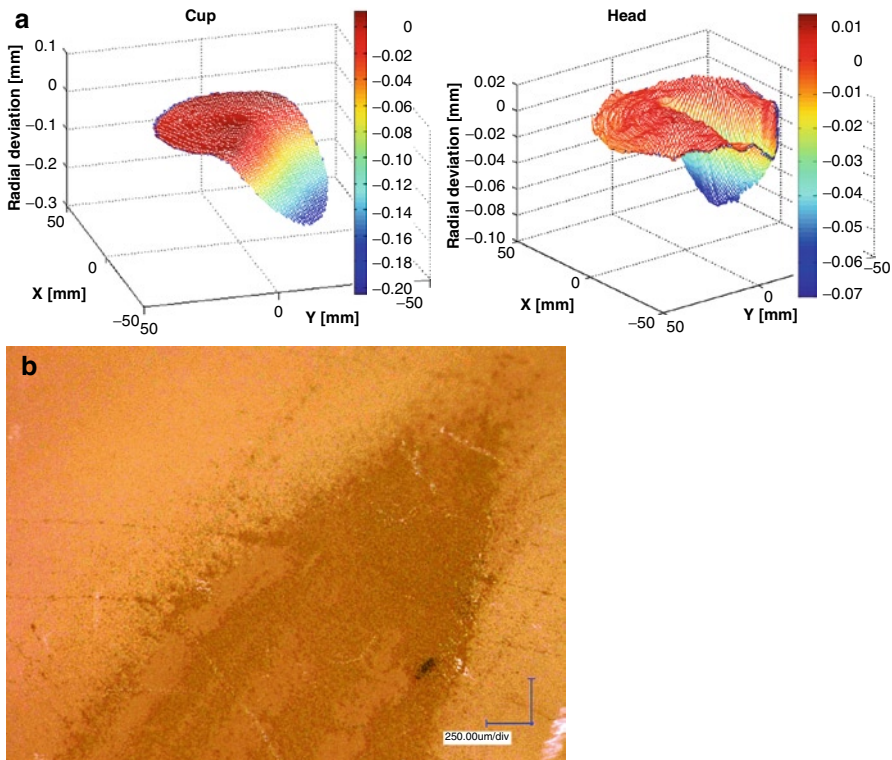


Fig. 2.6 (a) Example for the deviation from spherical for a rim-loaded metal-on-metal implant couple. The cup shows a distinct rim-loading pattern with wear dominantly at the edge of the cup. The head shows a deep scar at the location, where it was articulating with the rim (1,344 days in situ; Cup inclination: 57°, cup anteversion: 27°; total wear: head 34.4 mm³, cup: 68.9 mm³; revision reason: metallosis). (b) SEM picture of the stripe wear on a retrieved Al₂O₃ ceramic ball head (dark area)

Table 2.2 Time in situ and wear rate for revised hip resurfacing prostheses and XL-metal heads and cups. Cup inclination and anteversion were determined from ap-X rays using IMATRI public domain software (www.imatri.net). Classification into the rim loaded group was based on the wear pattern. Parameters are reported for implants exhibiting rim load pattern and implants not exhibiting this pattern (*Oneway Analysis of Variance*). For couples (*head and cup*), the wear rate of the head is reported

Rim loading	Time in situ (days)			Cup inclination (°)			Cup anteversion (°)			Wear rate (mm ³ /year)		
	n	Mean	SD	n	Mean	SD	n	Mean	SD	n	Mean	SD
No	145	243	327	114	48.1	8.2	106	10.2	8.0	138	0.40	0.80
Yes	28	690	406	23	54.8	12.4	23	21.5	12.2	27	7.42	5.17
	$p<0.001$			$p=0.002$			$p<0.001$			$p<0.001$		

resulting in increased friction (Fig. 2.6b). Stripe wear is a local damage of the ceramic surface due to break out of ceramic grains and consequent roughening. Increased friction can result in the excitation of vibrations of the prosthesis components, which can lead to audible noise phenomena, if the friction is large enough [30]. This phenomenon is mostly observed for hard-on-hard bearings, since hard-on-soft bearings cannot reach the required high friction coefficient. Increased friction also causes high moments at the cup-bone and head-prosthesis-bone interface, which can lead to problems with cup and stem fixation or also to problems in the fixation of the head on the stem [11, 17]. Hard-on-hard bearings are more sensitive to positioning than hard-on-soft bearings since their superior tribological characteristics rely on fluid film lubrication. If the fluid film breaks down, the tribological characteristics deteriorate rapidly.

On the femoral side, component position is becoming a more and more important issue since shorter prostheses are becoming more popular due to their bone conserving philosophy. It has to be realized that shorter prostheses also have a shorter lever arm to resist the moments introduced by the joint force: the shorter the stem, the higher the loading of the interface between the stem and the bone [7]. Since shorter stems typically also have less surface interface with the bone, the local bone stresses are even higher. This stress rising consequence of short stem designs mainly plays a role in uncemented stems during the ingrowth phase. If the lever arm of the joint force exceeds the load capacity of the anchoring bone, the prosthesis can rotate into varus (Fig. 2.7). In the

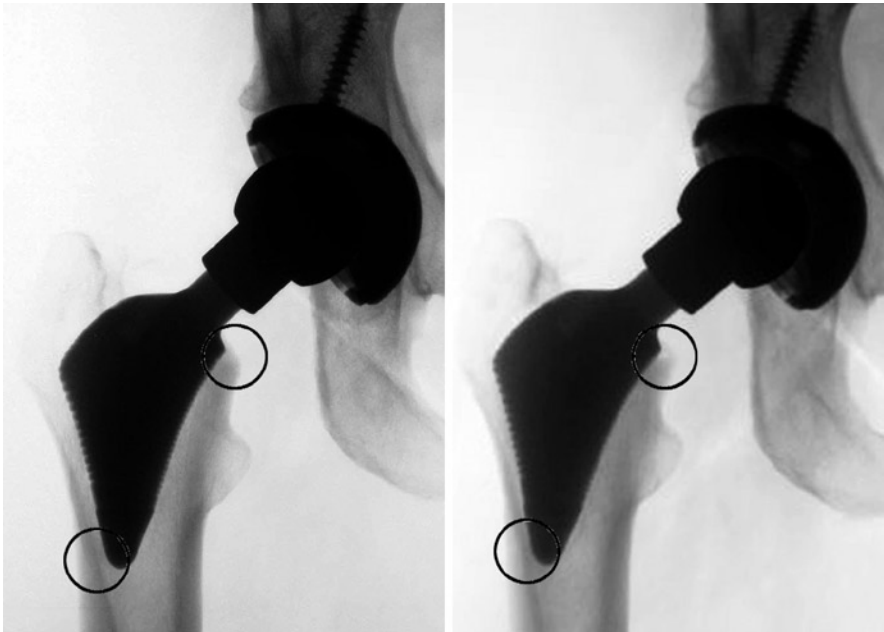


Fig. 2.7 A short hip endoprosthesis direct post operatively (*left*) and 6 month post op (*right*). Due to the very large head length and the small prosthesis size the implant rotated slightly (3°) into varus (*circles* indicate areas, where the migration can be seen)

worst case, fracture of the calcar can be the result. Once the bone has ingrown, this problem is greatly reduced since the ingrown bone-implant interface can also transfer tension.

2.6

Final Remarks

Some surgeons call total hip arthroplasty the most successful surgery in the history of orthopedics. This certainly seems justified considering the growing number of surgeries performed every year and the success rates in the registries. From a biomechanical point of view, the problem of THA is under control, as long as the patient and surgeon act carefully and responsibly. Established implants and bearing materials have clinically been shown to be successful in a vast majority of patients over periods in excess of 15 years. However, certain developments have to be watched carefully in order to maintain this success story:

- The continuous development of shorter prostheses (increasing the risk of bone failure)
- Lower wearing bearing materials (developed in the simulator under optimal conditions)
- New smaller surgical approaches (making the positioning of the implants more difficult)
- Surgeons doing only a few THA cases per year

There is a continuing need to improve implants and utilize newly available materials, but this process has to carefully consider the risks and side effects of new developments and not purely focus on the benefits. Continuous surgeon education and training for new implants and procedures is an essential requirement for the introduction of any new developments into clinical use. It has to be realized that successful pre-clinical testing does not guarantee clinical success but rather comprises a minimal requirement. The international standards should be extended to include testing of adverse implant conditions instead of only considering an optimal situation. The story of the extra-large heads for metal-on-metal articulations are a good example: Larger heads do reduce the risk for dislocation and the amount of wear debris generation (in the simulator). This was the reason for their introduction. The advantage of extra-large heads with respect to dislocation remains, to a certain extent, even if the implants are positioned poorly. In contrast, the advantages with respect to wear and friction are not just reduced, but are reversed if the implants are positioned poorly. This was not realized at the time when they were introduced, and resulted in massive problems in their clinical application. High friction situations in the patient were contributing to cup loosening and taper problems, as well as run-away bearing wear, causing metallosis and pseudo-tumors. The patients, who had high expectations for these state of the art prostheses, nourished by industry and surgeons, were greatly disappointed when they had to realize the unexpected problems. A THA system, which has been optimized from a biomechanical and tribological point of view, but requires an accuracy of positioning and assembly, which cannot be routinely achieved by an experienced user, is probably not the best option for the patient.

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