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Biotribology of Artificial Hip Joints

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Abstract

Tribology is the science of interacting surfaces; when these surfaces are in a biological system, it is called as biotribology. With the increasing rate of joint replacement operations and need for artificial prosthesis, biotribology is becoming a very important and rapidly growing branch of tribology. Based on this fact, in this chapter, basic tribological concepts are presented in terms of friction, lubrication, and wear; then with these fundamental backgrounds the biotribological behavior of natural and artificial hip joints are discussed in detail. Moreover, material pairs that are used in artificial joint replacements and the application of surface modification for the enhancement of the tribological properties of these materials are handled. Furthermore, the determination of tribological behavior of joint materials such as wear, coefficient of friction, friction torque, and frictional heating by using conventional techniques and hip joint simulator are discussed. Finally, the measurement and analysis of wear in both retrieved prosthesis and experimental studies are discussed referring the latest research articles.

Keywords: hip joints, friction, wear, biotribology

1. Introduction

The term tribology, which is derived from Greek words *tribos* meaning rubbing and friction and *logos* meaning science, was first used in the Jost Report in 1966 [1]. It is the science of interacting surfaces and includes the subjects such as friction, lubrication, and wear. Tribology is multidisciplinary science that encompasses mechanical engineering, materials science, surface engineering (surface coating, surface modification, and surface topography analysis), lubricants, and additives chemistry [1]. *Nanotribology* is the area in which friction and wear are studied at the micro- and nanoscale levels. The term *biotribolgy* is used when the interacting surfaces are a part of a human body or an animal such as total hip and knee joints.

It was first used by Dowson in 1970 and was explained as tribology in biological systems such as skin, hair, eyes, synovial joints, spine, and oral tribology [2].

Artificial prosthesis is replaced with natural parts of the hip joints when these natural parts lose their functionalities because of osteoarthritic and traumatic situations. Different material pairs such as metal-metal, ceramic-ceramic, ceramic-polymer, and metal-polymer have been used for artificial femoral head and acetabular insert. Service life of these joint parts is restricted because of the tribological deficiency in the joint materials. Lubrication regimes, wear debris, friction coefficient, and frictional heating are the primary factors that affect the implant durability. Wear debris leads to adverse tissue reactions that cause aseptic loosening of the components and finally leads to implant loss [2]. Frictional heating may cause property changes in sliding materials, tissue damage, and changes in lubricant properties such as protein precipitation especially in the biotribological zone.

The measurement, characterization, and analysis of the biotribological properties of artificial hip joints are vitally important for developing long durable joint replacements. Radiographic, gravimetric, volumetric, and optical techniques are the current methods for measuring and evaluating wear in the total joint replacement components. For surface characterization, optical and laser profilometry, white light interferometry, and digital microscopy are the prominent methods that can be applied both macroscopically and microscopically for characterizing damage modes such as burnishing, abrasion, scratching, pitting, plastic deformation, fracture, fatigue damage, and embedded debris.

Although the researches have been conducted for determining the wear behavior of the total joint replacements in order to improve design, material and manufacturing quality, and the service life of these joints, the phenomena of how to design wear-resistant artificial joint parts and select ideal material pairs for these kinds of replacements are still unknown. Therefore, the examination of friction and wear characteristics of prosthesis material pairs, both for in vivo clinical applications and in vitro laboratory simulations, is still one of the most important topics for researchers [3–7]. This chapter describes how to handle biotribological behavior of natural and artificial hip joints in terms of friction, lubrication, wear, frictional heating, and wear evaluation referring the basic tribological rules and the latest research articles.

2. Basic concepts of tribology

Handbooks define tribology as the science and technology of interacting surfaces in relative motion with all practices. Basically, tribology deals with the study of friction, lubrication, and wear phenomenon that can be explained with a series of engineering subjects such as mechanics, solid mechanics, fluid mechanics, physics, applied mathematics, rheology, machine design lubricant chemistry, material science, thermodynamics, and heat transfer [8–12].

Tribology is vitally important for any national and global economy. It causes large amount of energy loss, material waste, time loss, labor loss, etc. [13]. For instance, early wear of an acetabular insert of artificial hip joint means revision surgery that takes time of surgeons and

causes waste of materials. Furthermore, new surgery means pain for patients who need rest after the surgery. Therefore, it reduces life quality and causes labor loss. Tribological problems, both in industrial and biological areas, cause enormous cost to global economy and waste of natural resources, as well as affect the social system. Therefore, understanding tribology in terms of friction, lubrication, and wear is crucial for the development of wear-resistant materials and designs.

In this section, the basic concepts of tribology such as friction, lubrication, and wear are summarized for better understanding of biotribology in hip joints.

2.1. Friction

Friction is generally explained as the resistance to relative motion between articulating surfaces [13, 14]. It is the main cause of wear and energy loss [13]. An energy input is provided for the motion of sliding surfaces and maintaining the motion. This energy is dissipated into the system, mainly as frictional heat that causes property changes in sliding materials [14], tissue damage, and changes in lubricant properties such as protein precipitation especially in the biotribological zone.

A deep understanding of friction and wear processes first requires the investigation of the influence of numerous effects accompanying the friction process, i.e., mechanical, electrical, hydroacoustic, physicochemical, and other effects, and their influence on physicochemical properties, structure of working surfaces, etc. [15]. Friction is generally classified as static friction, sliding friction, and rolling friction (see **Figure 1**).
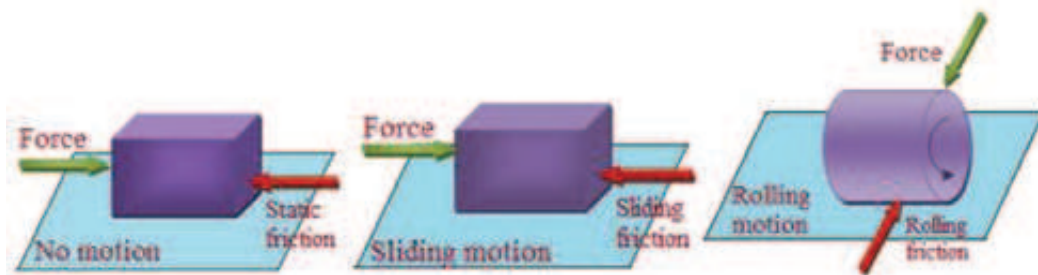

Figure 1. Scheme of friction force and motion.

In static friction, there is no motion [14, 16]. To slide a contacting body over another, a tangential force must be overcome which is named as *friction force* (*F*). It acts on the sliding surface plane and is usually proportional to the normal force (*N*) [14]:

$$F = \mu \cdot N \tag{1}$$

The proportionality constant termed as the coefficient of friction (*μ*) is used for quantifying sliding or kinetic friction, and it is defined as the ratio between the friction force *F* and the normal load *N*:

$$\mu = \frac{F}{N} \qquad (2)$$

The coefficient of friction generally ranges from 0.03 for well-lubricated bearing to 1 for dry sliding [14, 16]. These values change according to operating parameters, such as sliding speed, applied load or contact pressure, temperature, presence of lubricant, and the properties of materials in contact such as surface roughness of sliding pairs [17].

For comfortable walking, the coefficient of friction must be 0.2–0.3; on ice walking, the μ-value between shoe and ice pair nearly becomes 0.05. In a synovial joint, with the very efficient natural lubrication, the coefficient of friction is 0.02 [14].

Surfaces are not perfect at the microscopic level. Peaks, valleys, asperities, and depressions can be seen at high magnification even on the best polished surface (see **Figure 2**) [16]. When these two surfaces are brought together, they touch from the tips of the surface asperities. At that point, adhesion or cold welding, which generally refers to resistance to separating bodies from each other, may occur, and plastic deformation may take place on a very local scale. To start the sliding motion, these formations must be broken by the friction force. The main contribution to friction action is extended by adhesion and deformation, but additional contributions may occur, such as wear debris, presence of oxides, or adsorbed films [8, 14].
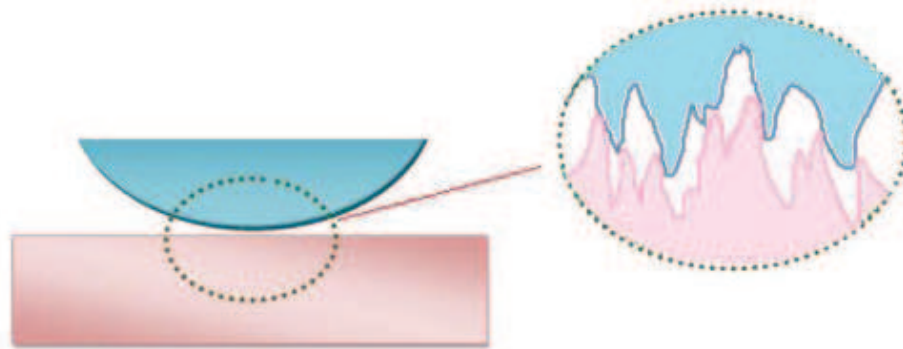

Figure 2. Microscopic detail of a real surface contact.

2.2. Lubrication

The main purpose of using lubricant is reducing the effect of normal and shear stresses on the solid surface contact [17]. Lubrication is one of the most effective ways of minimizing friction and delaying wear. Unfortunately, it is not the exact solution of the wear because wear occurs even with lubrication. Especially abrasive wear and delamination problem may occur under lubricated conditions [14].

Different types of lubrication regimes may arise between the sliding surfaces. In all lubricating modes, the surfaces are separated by a solid, a semisolid, a pressurized liquid, or the gaseous form of a lubricating film. *Dry-film (solid-film) lubrication* is a system in which a coating of solid-

state lubricant separates the sliding surfaces and the lubricant itself wears away. *Boundary lubrication* is the regime in which the interacting surfaces react with the lubricant components. Each surface is covered by a chemically bonded fluid or a semisolid film that may or may not separate opposing surfaces. In *thin-film lubrication*, the lubricant usually is not bonded to the surfaces and it does not separate sliding surfaces. Moreover, lubricant viscosity affects friction and wear. In *fluid-film lubrication*, the sliding surfaces are separated by a fluid film and the physical properties of the lubricant such as viscosity and pressure viscosity designate the performance of the lubricated surfaces. This lubrication regime can be divided into two subcategories such as hydrodynamic lubrication and elastohydrodynamic lubrication. *Hydrodynamic lubrication* is a regime in which the formation of the fluid film depends on the shape and relative motion of the sliding surfaces with sufficient pressure for separating the surfaces [8, 14]. For the mathematical explanation of hydrodynamic lubrication, an equation derived by Reynolds known as the "Reynolds equation" is used [13]. In elastohydrodynamic lubrication, friction and film thickness between the sliding surfaces are defined by the elastic properties of the contacting bodies. Although fluid-film lubrication is a desired regime, the boundary lubrication cannot be avoided. Boundary lubrication occurs during starting up and stopping stages of the motion [12]. In *mixed lubrication*, boundary and the fluid film lubricated regions are considered simultaneously [18].

To minimize the friction and wear of the sliding surfaces, understanding and determining lubrication mechanism are very important tools for the optimization of the bearing materials and geometries, both in engineering system and artificial joints. For the theoretical prediction of the lubrication regime, some classical engineering methods can be applied to artificial joints [11]. At that point, it is necessary to provide some basic information about geometrical and surface features of hip joint. Joints in a human body may be classified anatomically and physiologically such as plane, ball-in-socket, ellipsoid, hinge, condylar, pivot, and saddle [19]. Hip joint is considered as ball-in-socket geometry where contacting surfaces fit together. In theoretical calculations sometimes ball-on-plane equivalent configuration may be used for simplifying the geometry. In ball-in-socket types of geometries, contacting bodies have same diameters but with a clearance between the elements for suitable fit of the bodies and tribological reasons. R_{head}, R_{cup}, and $c = R_{cup} - R_{head}$ represent the femoral head radii, acetabular cup radii, and radial clearance, respectively. Another important parameter of sliding bodies is the average surface roughness R_a of the frictional surfaces. It is relevant for the determination of the lubrication regime [3], Schematic drawing of relation between the surface roughness and the film thickness can be seen in Figure (**Figure 3**).
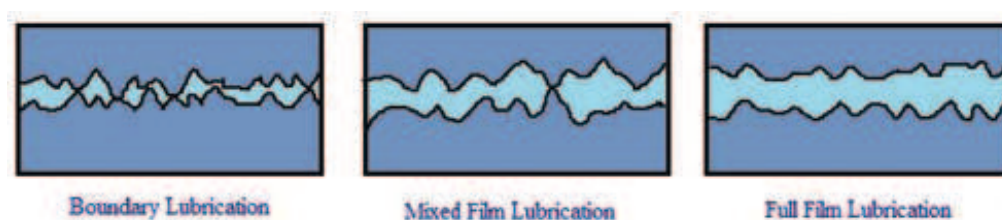

Boundary Lubrication   Mixed Film Lubrication   Full Film Lubrication

Figure 3. Schematic drawing of relation between the surface roughness and the film thickness.

The parameter λ, which is the ratio between the minimum film thickness $h_{\min}$ and the composite roughness of the sliding surfaces R_a, is generally used for determining the distance between the sliding surface asperities [16]:

$$\lambda = \frac{h_{\min}}{R_a} = \frac{h_{\min}}{\left[(R_{a\,\text{head}})^2 + (R_{a\,\text{cup}})^2 \right]^{1/2}} \qquad (3)$$

With the evaluation of λ, the lubrication regime can be identified in the following ranges [3]:

$0.1 < \lambda < 1$: boundary lubrication,

$1 \leq \lambda \leq 3$: mixed lubrication, and

$\lambda > 3$: full film lubrication.

The precision measurement of surface roughness for both femoral head and acetabular cup is important for the accurate determination of λ.

For the determination of film thickness, the following equation formulated for engineering can be used [11]:

$$\frac{h_{\min}}{R'} = 2.8 \left(\frac{\eta u}{E'R'} \right)^{0.65} \left(\frac{W}{E'R'^2} \right)^{-0.21} \qquad (4)$$

where R' is the equivalent radius that depends on the femoral head radius R_{head} and the radial clearance c.

$$\frac{1}{R'} = \frac{1}{R_{\text{head}}} - \frac{1}{R_{\text{cup}}} = \frac{c}{R_{\text{head}}(R_{\text{head}} + c)} \qquad (5)$$

In the ball-on-plane equivalent configuration, the entraining velocity (u) can be calculated from the angular velocity of the femoral head (ω):

$$u = \frac{\omega R_{\text{head}}}{2} \qquad (6)$$

The equivalent elastic modulus (E') can be determined by following equation:

$$\frac{1}{E'} = \frac{1}{2} \left(\frac{1 - v_{\text{head}}^2}{E_{\text{head}}} + \frac{1 - v_{\text{cup}}^2}{E_{\text{cup}}} \right) \qquad (7)$$

where E_{head}, v_{head} and E_{cup}, v_{cup} are the Young's modulus and Poisson ratio of the femoral head and acetabular cup material, respectively [3].

For the prediction of lubrication between the femoral head and acetabular cup bearing surfaces and its effect on friction, generated during articulation, the Stribeck diagram can be used (see **Figure 4**). In this diagram, the relation between the lubrication and friction is commonly illustrated [9, 11, 14, 20]

$$z = \frac{\eta u r}{W} \tag{8}$$

where z is the Sommerfeld number, η is the lubricant viscosity, u is the sliding speed, r is the radius, and W is the load.

Traditionally, the Stribeck curve is divided into three regions. Boundary lubrication is seen when the thickness of the lubricating film is less than or equal to the average surface roughness of the articulating surfaces. When the thickness of the lubrication film increases, a transition stage, called mixed lubrication, is generated. The articulating surfaces are separated from each other while in contact with some asperities, and the combination of the fluid film and boundary lubrication can be seen in this regime. Full fluid film lubrication occurs with the continuous decrease in the friction coefficient, and the articulating surfaces are completely separated [9, 16].
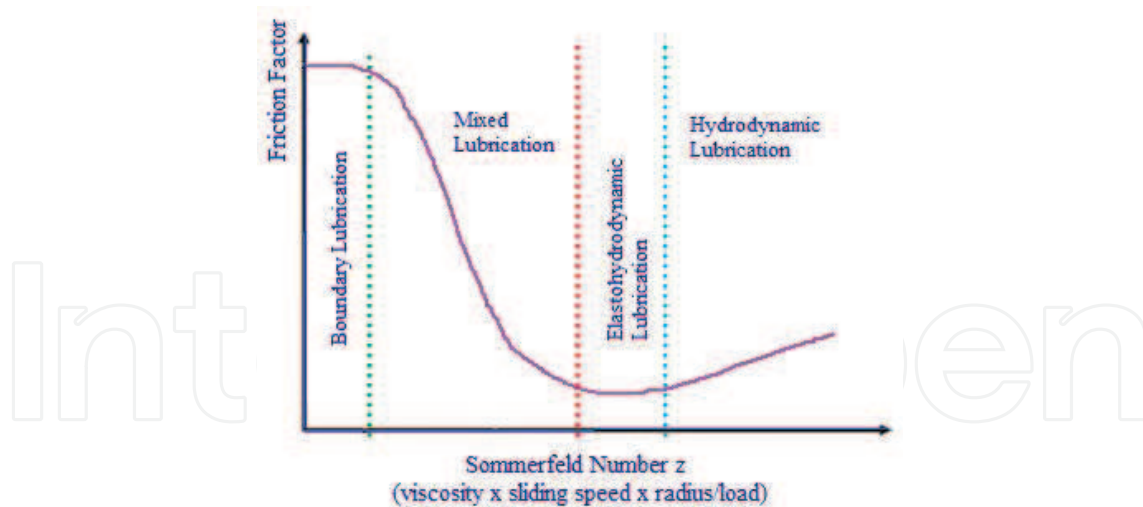

Figure 4. The Stribeck curve.

2.3. Wear

Wear is defined as the progressive loss of material from the surface of a body [18]. It is a complex phenomenon that involves multifarious events in a wildly unpredictable manner [10]. Numerous types of wear have already been defined in different studies related to tribology.

Although it is difficult to classify wear types without considering mating materials, for a general classification wear can be divided into five different groups, such as abrasive, adhesive, surface fatigue, erosive, and corrosive [11].

Abrasive wear is generally categorized according to contact types of the surfaces such as two-body and three-body wear. When the asperities of the harder surface abrade the softer one, it is called as two-body abrasion. If there are hard particles trapped between surfaces and abrade one or both of the surfaces, it is known as three-body abrasion [14]. *Adhesive wear* occurs due to high local friction that leads to tearing and fracture. This type of wear is generally defined as transfer of material from one surface to another during relative motion. The particles, broken away from one surface, may attach to another surface and act as an abrasive. This kind of wear may be seen between ceramic, polymers, and metallic material pairs or their combinations [9, 14, 17]. *Corrosive wear* is the wear that contains both mechanical wear and chemical reaction in which metal ions are released [9]. Besides the chemical and electrochemical reactions, environmental conditions govern the oxidative wear [11]. *Fatigue wear* is the displacement of the particles from the microscopic contact area of material surface by cyclic loading. It may lead to the generation of debris from the surface or cracks' propagation into the bulk material [11, 17]. *Erosive wear* is the loss of surface layer of the material caused by hard particles attacking to the surface. The attacking particles may be in solid, liquid, or slurry form. This type of wear may involve plastic deformation and brittle fracture. Erosive wear is similar to abrasive wear and generally be confused with it, but there is a definite distinction between erosive and abrasive wear. In erosive wear, the force is transferred to the surface by the particles due to their slowing down, while in abrasive wear, the force is externally applied. This kind of wear is not common in hip joints [8, 14].

3. Biotribology in hip joints

Biotribology is one of the newest and popular term dealing with friction, wear, and lubrication of interacting surfaces in a human or an animal body, while tribology is the study about these phenomena in engineering systems. Synovial joints, spine, skin, eyes, hair, and oral tribology are some examples of biotribological systems [21]. Because biotribology contains all tribological activities in these complex and natural biological systems, it is interdisciplinary like tribology. Moreover, besides the disciplines related to tribology, biotribology is also associated with biomechanics, biology, biochemistry, physiology, clinical medicine, and pathology. From the engineering point of view, for the development of long-lasting artificial interacting surface parts, it has to be well understood how these natural systems and artificial joints behave under biotribological conditions.

3.1. Natural synovial joints

There are over 300 joints in a human body. These joints have been categorized into two main groups such as synarthroses and diarthroses. Diarthroses joints are generally named as synovial joints because they have a cavity between the interfaces of the bones containing

synovial fluid. Moreover, joints may be classified according to their anatomical and physio-logical properties such as plane type, ball-in-socket, ellipsoid, hinge, condylar, pivot, and saddle-type joints. Hip joint is a ball-in-socket type configuration that has three degrees of freedom such as flexion-extension plane, abduction-adduction, and lateral-medial rotation [19]. Articular cartilage, bone, synovial fluid, ligaments, tendons, tissues, and soft tissue capsule are the main components of a natural synovial joint [8].

Natural hip joint is surrounded by a synovial membrane that builds synovial capsule. Synovial fluid separates the articulating bone surfaces by filling the joint cavity, and it performs under both fluid film and boundary lubrication regime. The articulating surfaces are coated by articular cartilage, which has very low coefficient of friction such as 0.02 [22]. Synovial membrane provides fresh synovial fluid into the cavity. The volume of this fluid is about 2 ml. Hyaluronic acid, lubricin, and globulin are the main biological components that provide unique lubricant properties to synovial fluid and make it the best lubricant for synovial joint [23, 24]. A healthy synovial fluid is a dialysate of blood plasma, which is comprised mainly of water (85%), hyaluronic acid, and protein [25]. All of these components make different contributions to the function of synovial fluid. For instance, hyaluronic acid serves to increase the viscosity of synovial fluid [26], whereas lubricin decreases the shear strength at the asperity contact interfaces [24, 27] and lipid layer on synovial membrane plays an effective role in the boundary lubrication mechanism of articular cartilage [28]. The synovial fluid composition changes from a healthy person to an osteoarthritis patient [29]. Therefore, the functionality and lubrication property of the synovial fluid may vary.
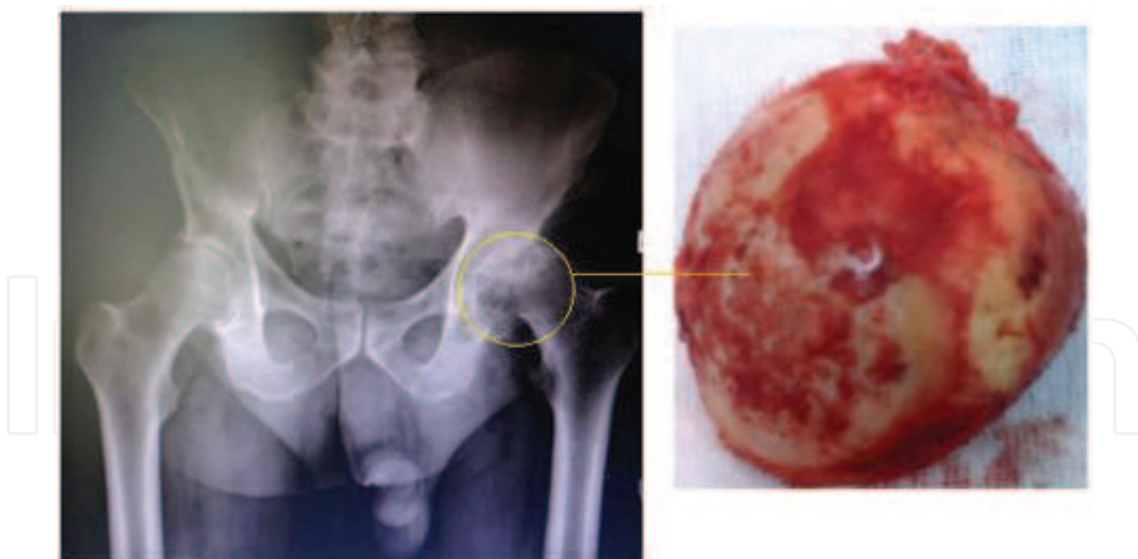

Figure 5. Radiographic image of diseased femoral head and its photography.

Ball-in-socket configuration, synovial fluid, and synovial capsule build a joint system that can transmit seven to eight times body weight during walking, climbing up stairs, jumping, or some other vigorous activities [3]. These systems provide approx. 70 years or more lifetime to the synovial joint with an overall wear factor (k) 10^{-9} mm^3/Nm [8, 22]. However, the lifetime of

this special system sometimes may end earlier than the expected time because of various clinical factors such as osteoarthritis, rheumatoid arthritis, necrosis, and trauma [22]. At that point, artificial joint elements are replaced with the fractured natural joint parts. For understanding the failure mechanism of the natural joint, biotribological studies have been conducted widely over the past 50 years. These researches are generally focused on the measurement of friction in synovial joints, the mechanisms of joint lubrication, measurement and analysis of cartilage wear and damage, and the lubrication properties of synovial fluid and its ingredients [8]. Studies in this area have made great contribution to understand normal joint function; however, the entire tribological system cannot be clarified yet because of the complexity of the natural synovial joint. A radiographic image of diseased femoral head before total joint replacement surgery and its photography after surgery can be seen in **Figure 5**.

3.2. Artificial hip joints

Artificial joint replacements have been one of the best solutions for patients affected by the clinical factors specified above. Although applications of total hip joint arthroplasty are very successful, the revision rate of these artificial joints is still unexpectedly high. For instance, in England 11 and 12% of all total hip replacements failed in 2011 and 2012, respectively [24]. The expected service life of these replacements is about 15–20 years. This duration is not enough for patients younger than 60 years old. Sometimes the joints may fail prematurely before the expected service life. This causes a lot of problems for both patients and surgeons because premature failure brings pain with revision surgery, need of extra money, and spending extra time. Biotribological behavior of the artificial joints is the primary failure factor that limits the service life of the prosthesis.
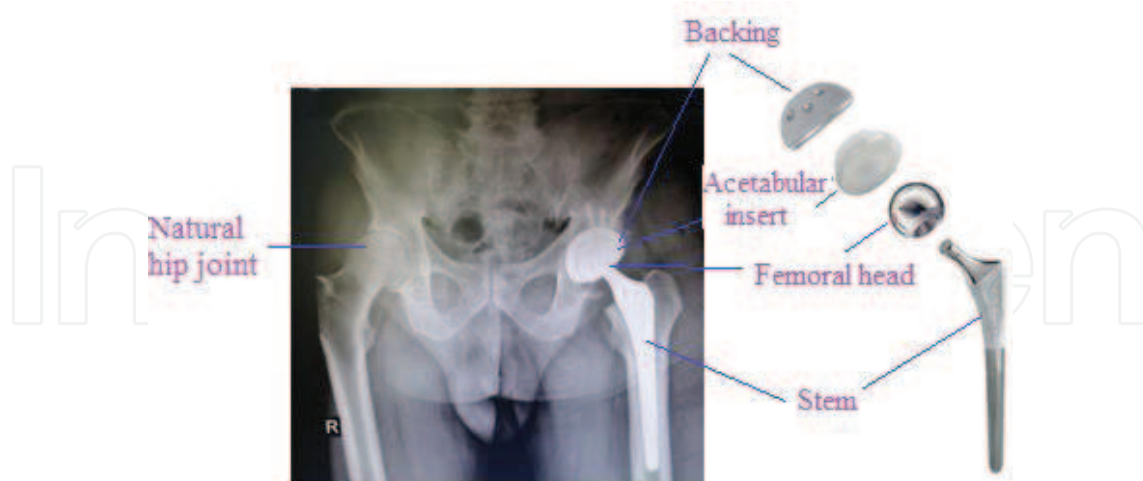

Figure 6. Components of artificial total hip joint.

Different material combinations such as ceramic-on-ceramic (CoC), metal-on-metal (MoM), ceramic-on-metal (CoM), metal-on-polymer (MoP), and ceramic-on-polymer (CoP) are used in total hip replacements. The most common combination is metal femoral head ultrahigh-

molecular-weight polyethylene (UHMWPE) cup [30, 31]. UHMWPE has been the most preferred acetabular cup material for the past four decades with excellent biocompatibility, chemical stability, impact load damping properties, and low friction coefficient [32, 33]. However, the wear debris of UHMWPE induces adverse tissue reactions and third-body wear damages that cause aseptic loosening and implant failure. Therefore, in order to extend the implant life, especially for young and more active patients, improvement in the UHMWPE properties such as low friction coefficient, third-body wear resistance, generation of small amounts of wear debris, and low cellular reactions to such wear debris became a need [31, 34, 35]. Components of the total hip joint can be seen in **Figure 6**.

Radiation-induced cross-linking has been a very effective way for the modification of UHMWPE microstructure, so the first-generation cross-linked UHMWPE has been developed [35–37]. This cross-linked UHMWPE has shown higher wear resistance than the conventional UHMWPE, but mechanical properties, oxidation, and delamination resistance have been decreased because of the postheat treatment operation [33, 34, 38]. With the aim of reducing the restriction of cross-linked UHMWPE, the second-generation UHMWPE has been developed by the addition of α-tocopherol or vitamin E as a natural antioxidant [36]. Tribological studies reported that the addition of vitamin E enhanced the oxidation and delamination resistance of the conventional and cross-linked UHMWPE while maintaining the mechanical properties by stabilizing the residual-free radical without postirradiation melting [31, 39–41].

As an alternative to the hard-on polymer implant pair, hard-on-hard artificial joint materials such as MoM, CoC, or CoM have been used for increasing wear resistance and so the service life of the artificial joints. These types of bearing pairs have shown better wear resistance and given less wear debris than the conventional MoP pairs. CoC bearings are recognized as the most wear-resistant pairs because of their very hard, smooth surface, and effective lubrication, but squeaking is a restrictive problem that causes distress to the patients as reported in clinical studies [42–44]. A photographic image of the surgical application of ceramic femoral head can be seen in **Figure 7**.
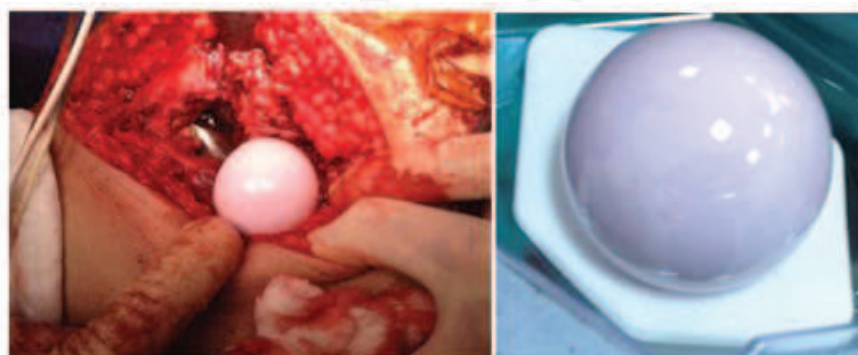

Figure 7. Surgical application of ceramic femoral head.

MoM joints have been reported as low volumetric wear rate bearing besides high stability with large heads [45]. Metal ion release, metallic wear debris, and tribochemical reactions are the

main problems for these artificial joints [46, 47]. Moreover, during the gait cycle MoM total hip joints are exposed to higher impacts, which may cause severe pain in patients than hard-on-polymer joints. For combined advantages of CoC and MoM pairs, CoM artificial hip joints improved recently. Simulator studies showed a reduced wear of CoM joints than MoM pairs. In addition, limited chromium release has been reported for CoM total hip joints in short-term clinical results [48].

For understanding the tribological behavior such as friction, lubrication, and wear of the artificial joints, advanced in vitro simulator tests have been performed for simulating an artificial joint with real articulating components under dynamic loading, multidirectional sliding, and lubrication conditions. Extremely useful outcomes are obtained from these studies and the results have been used for the improvement of design and properties of artificial joint materials. However, tribological behavior of the artificial joints does not only depend on implant material and design but also on patient-related factors such as lifestyle, body weight, age, gender, and synovial fluid that are poorly understood and cannot be simulated adequately in vitro test studies [24]. Biotribological behaviors of different artificial joint materials have been discussed in the following section referring the literature studies.

3.3. Friction, lubrication, wear

As mentioned above, friction forces are caused by the adhesion and cohesion forces. The proper definition and understanding of friction in synovial joints serve to predict and determine quantity of wear in artificial hip joints [49]. The coefficient of friction (COF) in artificial hip joint surfaces varies according to joint material pairs, geometry of joint parts, lubrication condition, and loading [16]. In order to make tribological assessments such as the determination of wear behavior, friction factor or COF of artificial materials, conventional tribological methods such as pin-on-disc, ball-on disk, and pin-on-plate can be used (see **Figure 8**). Furthermore, for the simulation of real joint conditions such as working with real joint parts and applying multidirectional loads, more complicated joint simulators have been used. The pressure distribution between the contact areas of the hip prosthesis cannot be known precisely. Because of this, it is difficult to determine COF accurately. At that point, a dimensionless parameter called friction factor (f) is used instead of COF. In experimental studies, the frictional torque (T) is measured first and then the friction factor is calculated by using the following equation:

$$f = \frac{T}{RL} \tag{9}$$

where R is the radius of the femoral head and L is the normal load [11, 20, 50, 51].

For simulator studies, the test conditions are standardized in ISO 14242-1. In these studies, the tests have been performed with the anatomical position of the components, e.g., in flexion-extension, adduction-abduction, and internal-external rotation planes with 1 Hz frequency under the dynamic loading condition and 25% bovine calf serum lubrication. In some friction

simulator studies, this configuration was simplified, for example, the prosthesis was inverted with respect to anatomical position and the tests were performed just in the flexion-extension plane ± 25° [11, 50, 52, 53]. These kinds of friction simulator friction factors are used in the lubrication of bovine serum, for CoC bearing it is reported as between 0.04 and 0.07, for MoP bearing 0.06–0.08, for CoP 0.05–0.08, and for MoM as 0.12–0.27 [11, 16, 50].
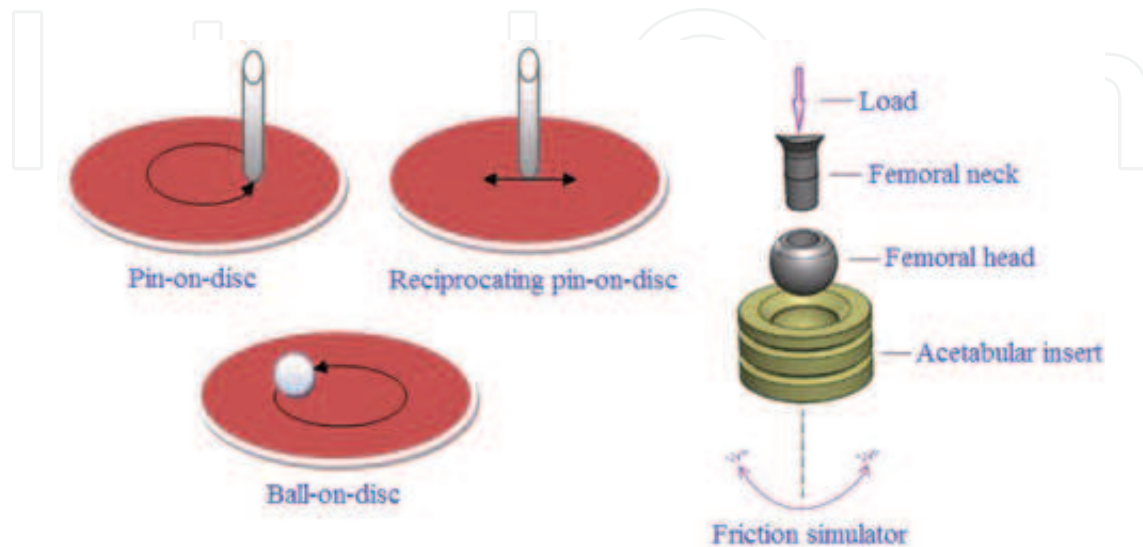

Figure 8. Schematic drawings of the conventional wear testing devices (pin-on-disk, reciprocating pin-on-disk, and ball-on-disk) and friction simulator.

Measuring, modeling, and predicting of friction on the total hip prosthesis have played an important role in developing artificial hip joints material and geometry. Moreover, the determination of friction can be used for the estimation of the lubrication regime between articulating surface of artificial hip joints as mentioned in the previous section.

Lubrication is one of the most complex and important factors that affect the implant friction and wear [31, 38]. However, the lubrication mechanism of natural joint cannot be fully understood, it is definitely known that there is a perfect lubrication system in a healthy natural joint. Different lubrication mechanisms, such as boundary lubrication, hydrodynamic lubrication [54] elastohydrodynamic lubrication, fluid film lubrication, and mixed lubrication, have been reported for lubrication in natural joints [16, 55]. However, in artificial joints the alternatives are limited because of joint material properties, surface qualities, and implant geometries. As a result of this, boundary lubrication and mixed lubrication regimes are leading lubrication mechanisms in artificial hip joints [19].

In MoP artificial joints, boundary lubrication regime is a dominant mechanism with λ values in the range 0.1–1. The surface asperity contact cannot be avoided because of the soft and rough surface of UHMWPE and thereby boundary lubrication occurs. In MoM bearing, the mixed lubrication regime can be seen generally with λ in the range of 0.6–2.9. CoC bearing surfaces are so hard and they have superior manufacturing tolerances and minimum surface roughness as compared with metal or polyethylene. The fluid film lubrication regime is predicted for these pairs with λ values between 5.7 and 28.3 [16, 55–57].

For improving surface quality and enhancing the lubrication properties of artificial joints, surface modification techniques have been an effective solution for artificial joint materials. Ion implantation, electron beam or gamma radiation, plasma surface treatment, and surface texturing are the most common surface modification techniques that are applied to improve the wear resistance of articulating surfaces without changing bulk material [58].

Surface texturing has been a well-known way for many years especially in machine bearings but it has started to be excessively popular in the last decade for artificial joints [5, 31]. In some previous studies, surface patterning was studied on artificial joint materials to reduce friction and wear. Young et al. [59] measured 43% lower friction coefficient of surface patterned samples than unpatterned disk samples by using pin-on-disk testing apparatus. Ito et al. [60] formed concave dimples on the metal femoral head surface, and they reported that the dimples served in reducing polyethylene abrasive wear by providing better lubrication and capturing wear particles. In reference [61], it is reported that lower friction coefficient and wear factor were obtained with surface-dimpled disk samples (see in **Figure 9**). Besides reducing wear rate and friction coefficient, surface dimples also serve in decreasing frictional heating of acetabular insert and femoral head [31, 62–65]. By acting as a reservoir of lubricant and capturing wear debris and bone cement particles, surface dimples provide better lubrication and thereby better tribological behavior [66].
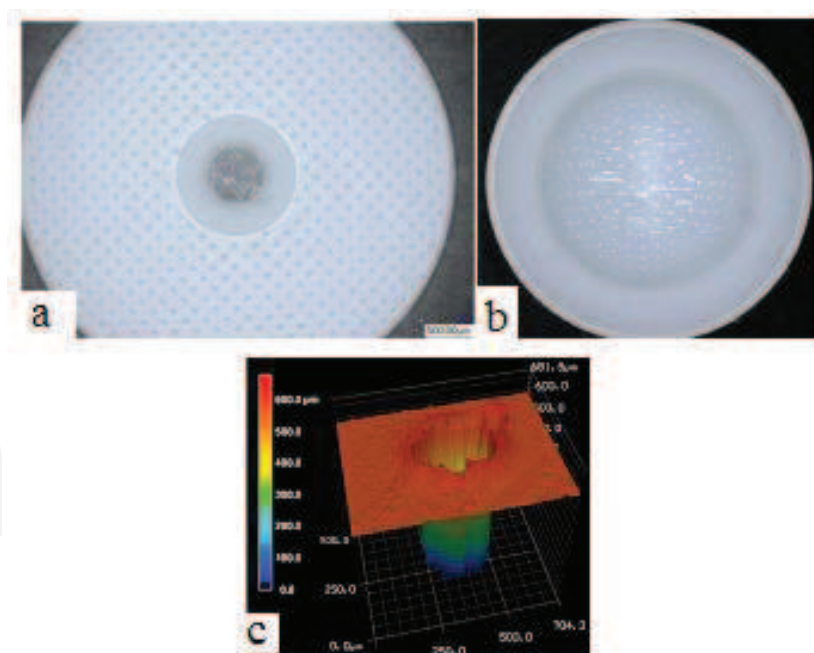

Figure 9. Microscopic images of surface dimples. (a) Dimples on UHMWPE disk sample before wear test, (b) dimples on the inner surface of UHMWPE acetabular insert after wear test with PMMA third-body particles, and (c) laser scanner microscope image of a dimple for characterization of its dimensions [31, 61].

There are many factors that affect the lifetime of a total hip prosthesis. Short-term failures are generally of biological origin, whereas long-term failures are related to material properties

with a biological response [19]. Wear has been considered as the primary factor that limits service life of hip implants. The wear debris generated during articulation of joint materials such as UHMWPE, metallic, or ceramic counter face could cause adverse tissue reactions, aseptic loosening, osteolysis, and finally implant loss [22, 31, 67, 68]. Microscopic image of worn surfaces of retrieved prosthesis can be seen in **Figure 10**.
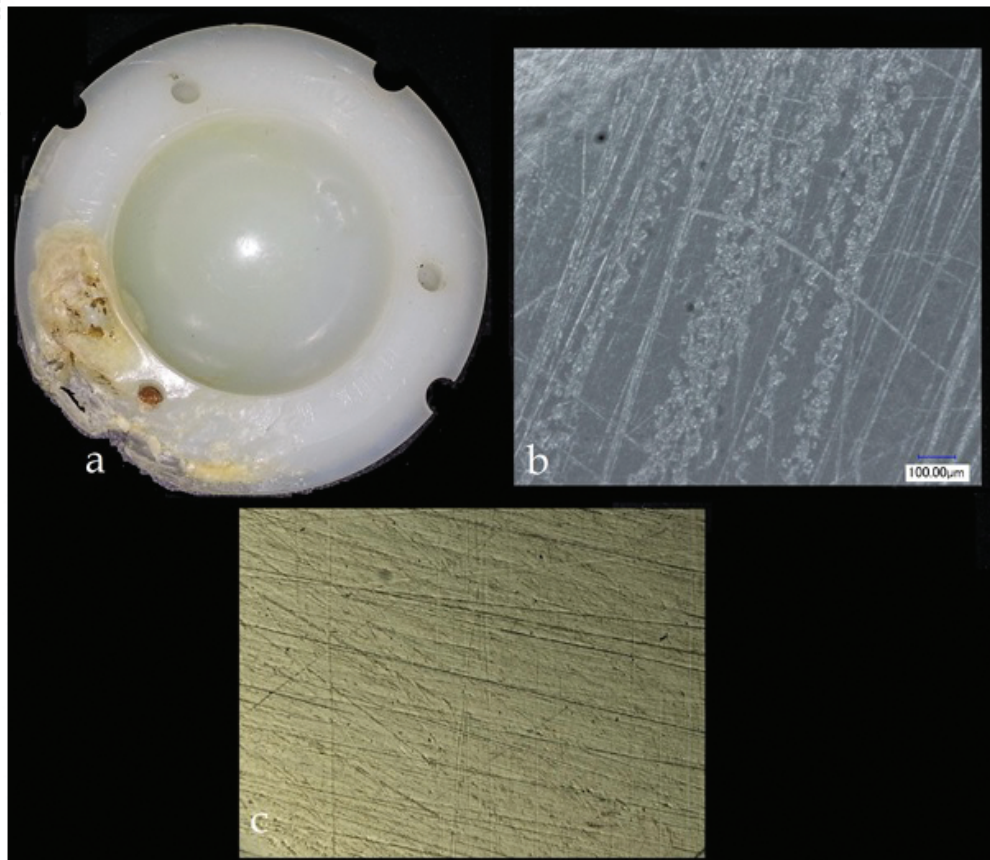

Figure 10. Worn surfaces of retrieved hip prosthesis. (a and b) UHMWPE acetabular insert and (c) CoCrMo femoral head.

Sliding surfaces in hip joint are conformal surfaces that fit together [16]. This makes geometry, size, and manufacturing tolerance of the joint components vitally important because initial wear occurs if there is any mismatch during bedding-in of prosthesis. This is a problem especially for MoM bearings because MoP pairs compensate this problem by polyethylene creep [56]. Wear of CoC pairs is negligible along normal walking because the wear volume is too low [69]. A well-functioning polyethylene acetabular insert exposes clinical wear such as 50–100 μm of penetration per year that means nearly 80 mm^3 volume loss per year [11, 16, 70]. In experimental hip joint simulator studies, MoP bearing wear rate was 40 mm^3/million cycle, whereas CoP bearing wear rate was 25 mm^3/million cycle. For MoM bearing, the wear rate was reported to be 1.0 mm^3/million cycle and for CoC artificial pairs the wear rate was recorded to be 0.1 mm^3/million cycle [11].

Three main wear mechanisms have been reported for UHMWPE such as fatigue, adhesion, and abrasion wear. *Fatigue wear* occurs when the surface of the material is weakened by cyclic loading. *Adhesive wear* is generated by the transfer of material from one surface to another when these two surfaces are articulating against each other under load. The transferred particles could break off and fuse together and then may act as third-body particles causing abrasive wear. Adhesion and fatigue wear generally work together. *Abrasive wear* occurs when hard asperities on the sliding surfaces or third-body particles are trapped between these surfaces. These asperities cause loss of the material from softer surface of the articulating pairs [10]. These wear mechanisms primarily occurs at the microscopic scale, whereas fatigue wear may occur at the macroscopic scale in the form of delamination [56]. The oxidation of polyethylene causes degradation and nearly 80% decrease in fatigue strength [71]. It is clear that the wear resistance of polymers is directly related to its mechanical properties and physical morphology [68].

Literature works about retrieved prosthesis show that third-body abrasive wear is a very important parameter affecting the service life of artificial joints [72]. By scratching the metal femoral head, third-body particles promote the wear rate of UHMWPE acetabular cup. PMMA bone cement particles are believed to be the main cause of third-body particles [73]. In addition, bone particles, metal beads or fibers from porous coatings and hydroxyapatite coatings, and corrosion products from the metal tapers and metal fragments from other fixation devices may be the source of third-body particles [74–77]. Microscopic images of PMMA particles can be seen in **Figure 11**.
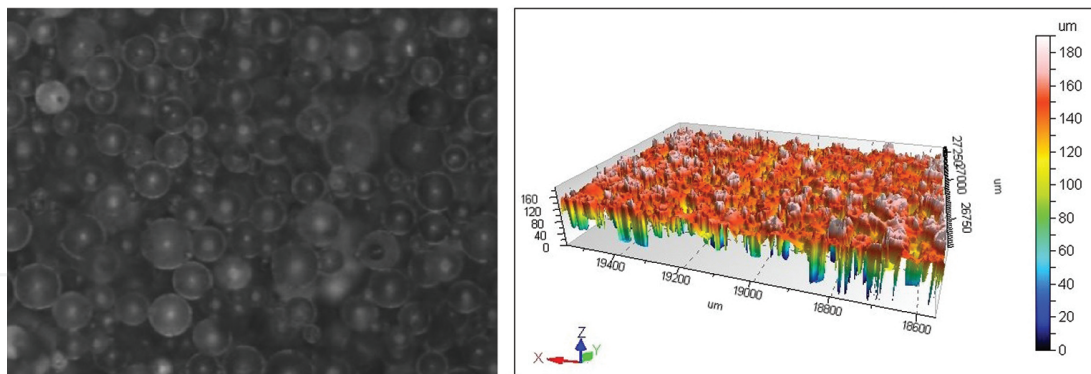

Figure 11. (a) Microscopic image and (b) surface texture of the PMMA particles used as bone cement for fixation of the artificial joints [78].

Bragdon et al. [72] reported three possible interacting mechanisms of third-body abrasive particles with acetabular cup and femoral head sliding surfaces after being trapped at the interface. In the first mechanism, particles may embed in polyethylene surface that causes to reduce the contact area between the head and the cup. In the second mechanism, third-body particles may adhere to the femoral head under pressure and finally some particles may roll freely between the surfaces. In a pin-on-disk configuration, the embedded particles may cause pitting, and the free particles that roll between surfaces may cause scratches on the UHMWPE disk surface as can be seen in **Figure 12**.
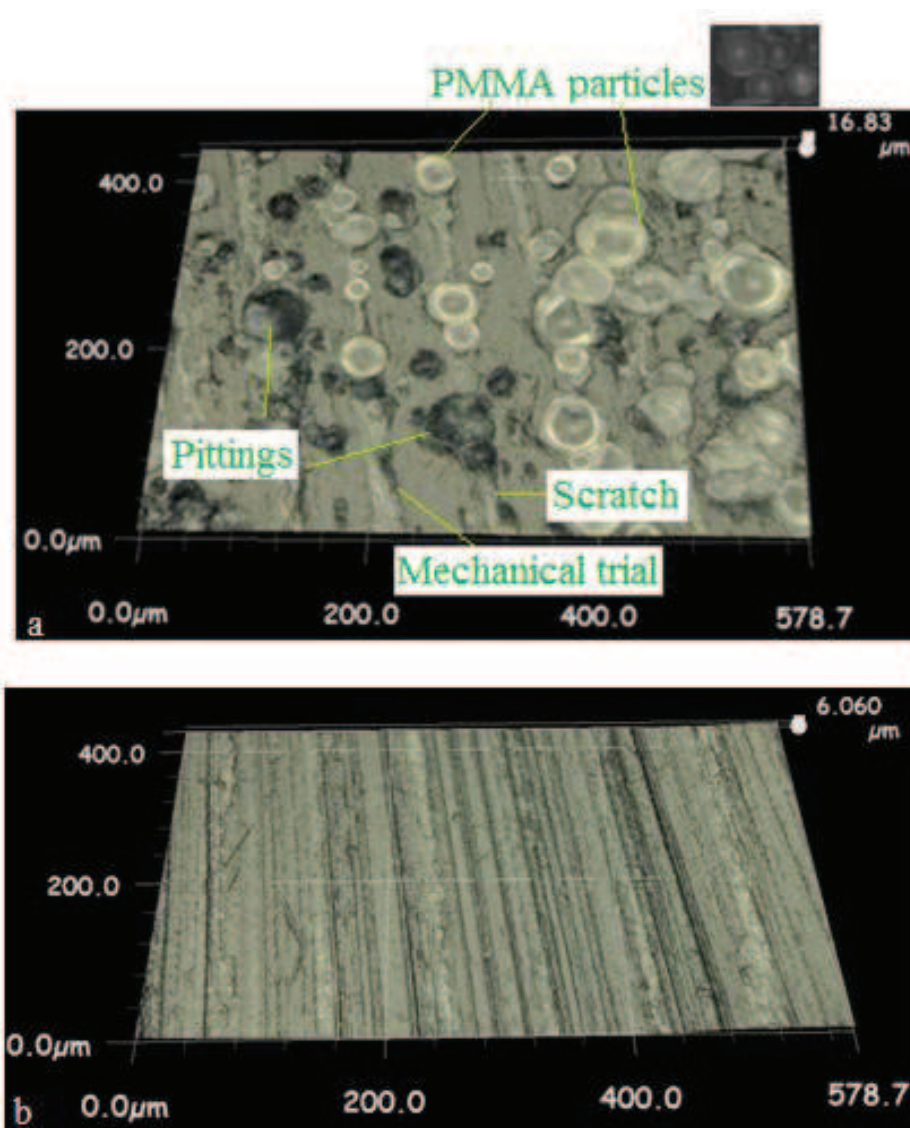
Figure 12. Worn surface of UHMWPE (a) with PMMA third-body particles, (b) without-third-body particles in 25% bovine calf serum lubrication [78].

3.4. Frictional heating of articulating surfaces

Most portion of frictional work between articulating surfaces is converted into heat during the wear process. This heat causes temperature rise in artificial joint parts, and the temperature rise may influence the properties of lubricant, the rate of wear, fatigue, creep, and oxidative degradation of bearing materials [14, 79–81]. Moreover, temperature rise may contribute cup loosening by causing bone necrosis and surrounding tissue damages [52]. In literature studies, frictional temperature rise has been measured with different ways by various researchers. These are experimental methods such as in vitro and in vivo measurements, theoretical calculations, and computer simulations [52, 79–84]. An example of experimental setup for the measurement of frictional heating between UHMWPE acetabular insert and CoCrMo femoral head can be seen in **Figure 13**.
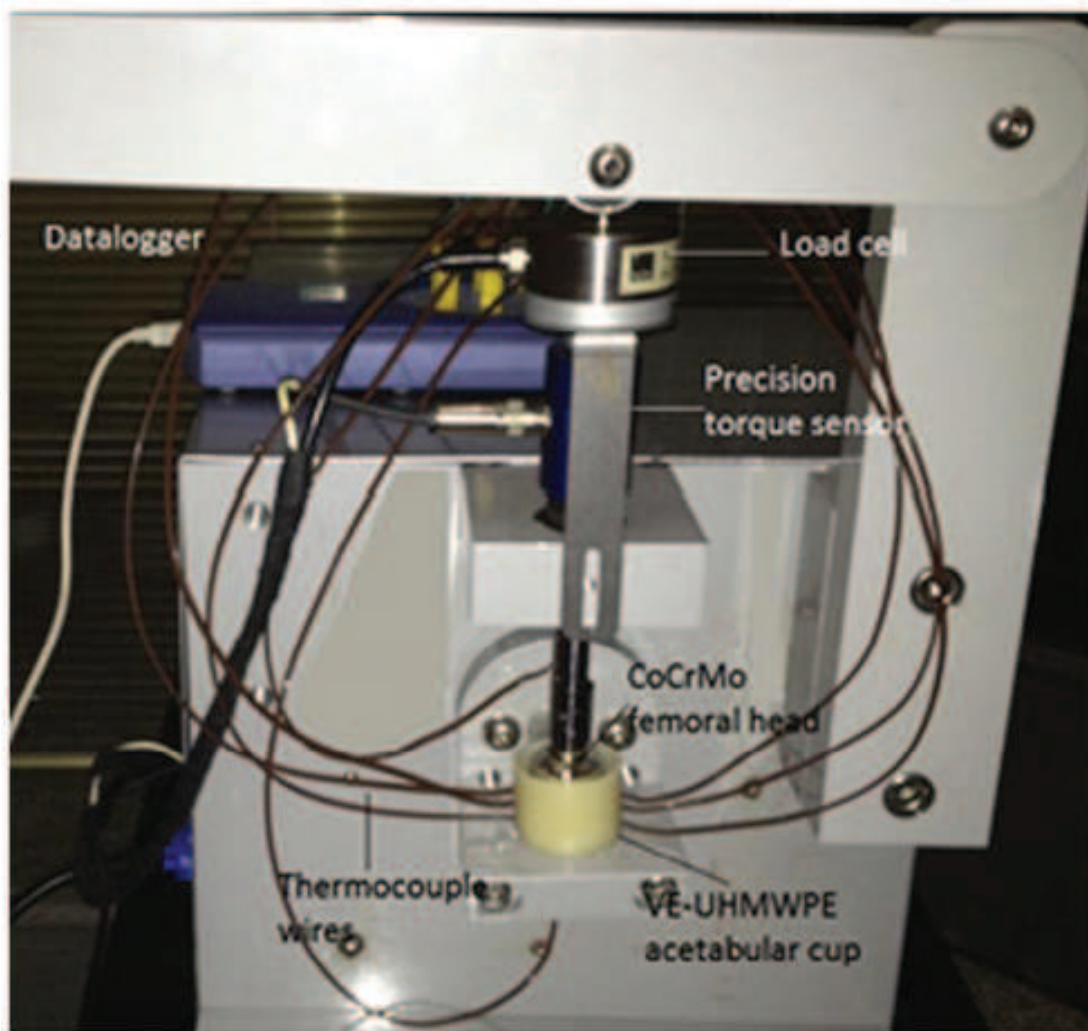
Figure 13. Friction simulator for measurement frictional heating between artificial hip joint components [31].

In an in vitro study, Lu and McKellop [52] reported that frictional heating may promote the protein precipitation from the lubricant and as a result property change of the lubricant. Exhaustion of proteins may cause incomplete boundary lubrication mechanism and so these changes may accelerate adhesive wear rate. In another study, Bergmann et al. [82] measured the temperature rise of hip prosthesis in patients' bodies after 1 hour walking, and they reported that the maximum temperature value was 43.1°C in vivo. In a computer simulation study [83], synovial fluid's temperature was found as 46°C by 2D and 3D finite element analysis. By using a thermomechanical finite element model of the ball-cup interaction, the peak temperature of the contact surface was predicted as 51°C [85]. In an experimental study [86], frictional temperature between zirconia femoral head/UHMWPE measured as 39°C at the contact point of femoral head whereas it was measured as 36°C between zirconia femoral head/ vitamin E blended UHMWPE.

Frictional temperature rise is related with lubrication condition and material properties such as thermal conductivity and elastic modulus. For example, while two materials with higher

elastic modulus used as frictional pair, the contact area would be smaller and a larger area would be exposed to the lubricant during the cycle. Therefore, with better lubrication and cooling, frictional temperature would be lower [31, 52]. In reference [31], the surface dimples' effects on frictional temperature rise of artificial joints was studied. It is reported that frictional temperature rise (ΔT) of surface-dimpled UHMWPE/CoCrMo sample was 8.39°C, whereas undimpled samples temperature rise was recorded as 11.22°C. Surface dimples acted as a reservoir for lubricant and provided continuous lubrication for cooling the surfaces, so by increasing surface lubrication quality, lower temperature values were recorded [31, 59, 87, 88] The biological defects occur at about 40, and 6°C temperature rise between articulating surfaces may cause fibrous tissue formation and possibly prosthetic loosening. Therefore, it is clear that service life of artificial joints can be negatively affected by frictional heating [80, 83].

3.5. Measurement of wear in artificial joints

The measurement, evaluation, and analysis of wear are vitally important for understanding the wear mechanism of artificial joints and improvement of new materials with new designs. Wear measurement of artificial joint materials may be applied in vivo and in vitro conditions with different measurement techniques. In vivo evaluation of the implant provides in situ monitoring of the patient more closely and for considering revision surgery by determining wear amount. Moreover, the wear rate of the implants and histological changes can be determined with periodical measurements. In vitro measurement studies include the examination of retrieved prosthesis, which provides data about wear mechanisms and wear rate of the materials, and the evaluation of lubricant, wear debris and debris distribution. The objectives of all these evaluations are to determine the wear rate and lifetime of implant and to understand wear mechanisms, tissue reactions, and other related events that occur in artificial joint system. Therefore, it would be possible to develop new materials and designs [19, 89].

Radiographic, gravimetric, volumetric, and optical techniques are current methods for measuring and evaluating the wear in the total joint replacement components [90]. X-ray techniques, magnetic resonance imaging (MRI), microcomputed tomography, and biochemical markers are methods for the measurement of in vivo cartilage wear. It is possible to see bone deterioration on joint by an X-ray technique where the MRI scans provide a very detailed view of the damaged tissue [25].

Two- or three-dimensional techniques, radiostereometric analysis, and manual or computer-assisted plain radiography techniques are the best known radiographic methods used for measuring polyethylene wear. Radiographic techniques allow estimation about femoral head migration into the cup [89–92].

The most common used techniques to evaluate wear are the measurement of weight change called as gravimetric method and the measurement of dimensional changes known as volumetric wear [8]. Gravimetric method is the standardized method with ASTM F2025-06, F1714-96, and ISO 14242-2 used for the measurement of wear in the total joint prosthesis. In this method, the weight difference between the initial weight and the weight after wear test were determined. According to ISO 14242-2:2000, a balance with the accuracy of 0.1 mg must

be used for weight measurements. Because of the fluid absorbing property of polymer component, the weight difference may be undetectable or the final weight may sometimes be higher than its initial value. For reducing the error due to fluid absorption, polyethylene specimens are soaked into the lubricant until they reach saturation that takes days or sometimes weeks. For reference purposes, a loaded but not articulating control specimen is used for determining fluid sorption of a specimen [90, 93].

Gravimetric wear can be calculated as follows:

$$W_n = W_{an} + S_n \tag{10}$$

where W_n is the net mass loss after n cycles of loading, W_{an} is the average uncorrected mass loss, and S_n is the average increase in mass of the control specimen over the same period.

By using the least-squares linear fit relationship between W_n and the number of loading cycles n, the average wear rate a_G can be calculated as follows:

$$W_n = a_G.n + b_n \tag{11}$$

where W_n is the net loss in mass after n cycles and b is a constant.

Although this method is effective to determine experimental wear amount in simulated conditions, it is not applicable to evaluate the clinical wear of retrieved prosthesis for which there is no prewear data available. Gravimetric method does not provide information about wear mechanisms, surface property changes, and plastic deformation of the component. Moreover, material transfer from the metal component or bone cement that is attached into the UHMWPE can cause a significant error while determining the weight loss both for in vivo and in vitro applications. These are the main limitations of this method [90, 94].

Volumetric methods such as the coordinate measuring machines (CMMs) and micro computed tomography (micro-CT) have recently been alternative methods to gravimetric method [42]. The use of CMM is standardized by ISO 14242-2:2000. This standard requires a CMM "with maximum axial-position error of measurement D":

$$D = 4 + 4L \times 10^{-6} \tag{12}$$

where D is in μm and L is the numerical value of the dimension in meters.

In addition, "Mesh spacing" must not be greater than 1 mm and "relocation of the test specimen" must not affect the measured volume more than 0.05%. For determining the wear amount of the prosthesis first of all initial, unworn geometry of the sample is measured and after wear test the measurement is repeated. Therefore, by comparing the initial and final geometries of the sample, the amount of volumetric wear is calculated [32]. These evaluation steps are applicable for simulator studies but in clinical application there is no initial geomet-

rical data about retrieved prosthesis. At that point, measurements are taken from unworn regions of the retrieved prosthesis and by using collected data unworn geometry of the sample is predicted. A coordinate measuring machine, with 2 μm minimum accuracy, should be used for three-dimensional wear analysis [90, 95, 96]. An example of the CMM measurement of UHMWPE acetabular insert can be seen in **Figure 14**.
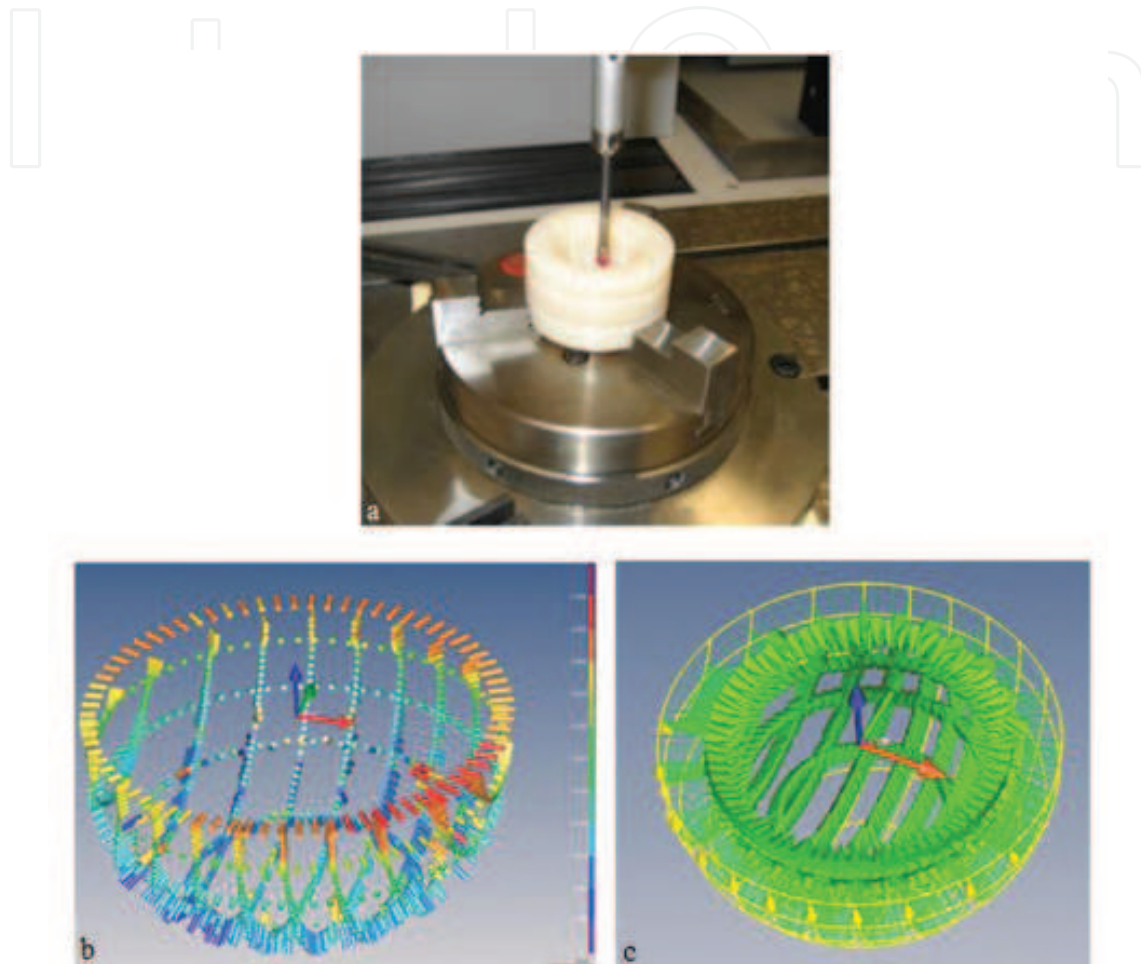

Figure 14. (a) CMM measurement of a UHMWPE acetabular insert, (b) defining measurement paths, and (c) inner surface of the prosthesis saturated with measurement points.

It is possible to analyze wear behavior of both hard-on-hard and hard-on-polymer retrieval prosthesis by using CMM. For evaluating the actual wear amount of polyethylene prosthesis, it is important to determine creep deformation. The CMM method provides advantages about defining wear volume, wear scars distribution, and creep deformation, but it is time consuming and shows uncertainty about the evolution of wear measurement [90, 97, 98]. Different surface characterization techniques such as mechanical, optical and laser profilometry, white light interferometry, and digital microscopy (see **Figure 15**) can be used for analyzing the wear behavior of the implant surfaces. These techniques may be applied both macroscopically and microscopically for characterizing damage modes such as burnishing, abrasion, scratching, pitting, plastic deformation, fracture, fatigue damage, and embedded debris [93, 97].
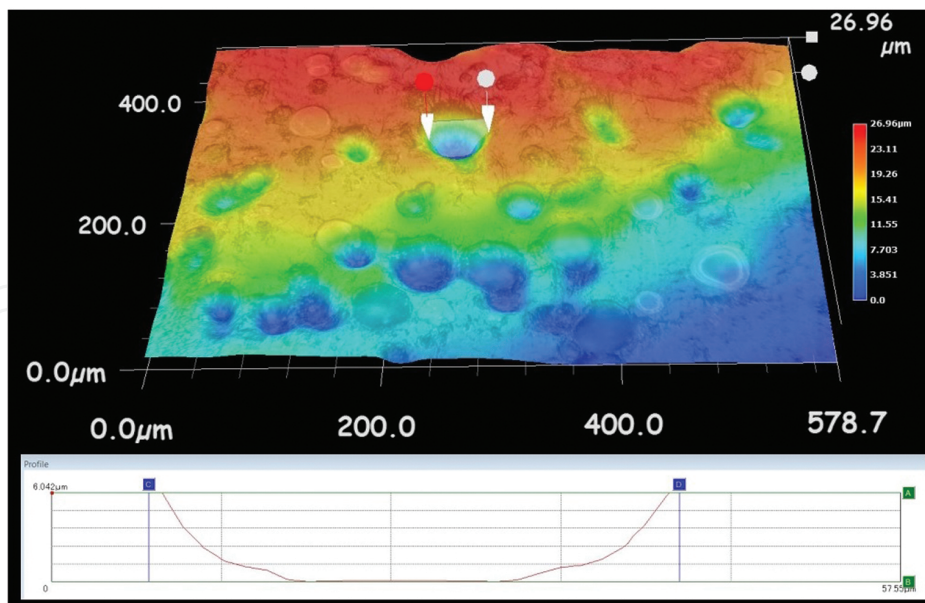
Figure 15. Determination of local wear damage on polyethylene sample by digital microscopy.

Surface profilometry is one of the most preferable techniques especially for the determination of wear in primitive wear tests such as pin-on-disk, pin-on-plate, and ball-on-disk (see **Figure 16**).
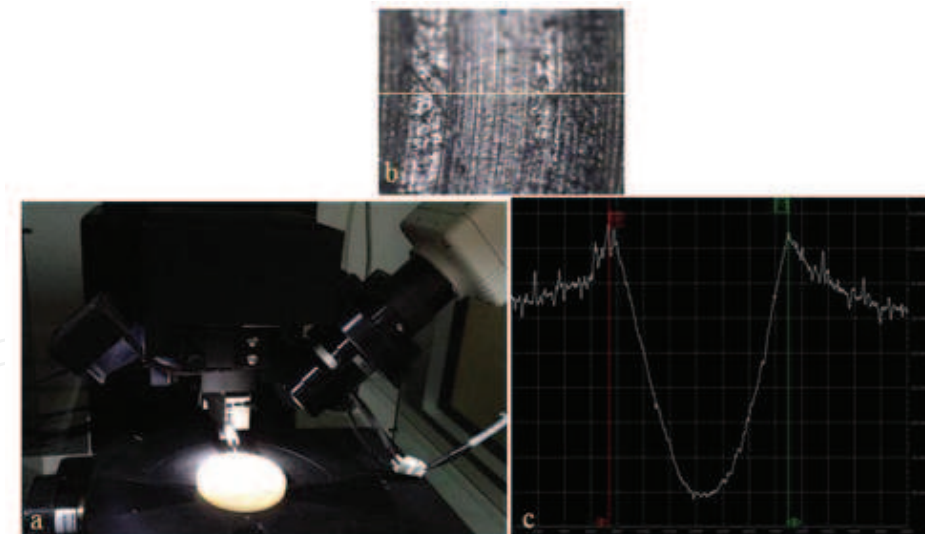

Figure 16. (a) Measurement of wear by mechanical surface profilometer, (b) wear track, and (c) wear track profile [78].

By using the cross-sectional area of wear track and its length the wear volume is calculated. Then, the wear factor (*k*) of the sample can be determined by using the following equation:

$$k = V \,/\, NS \tag{13}$$

where k is the wear factor (mm^3/(N m)), V is the wear volume (mm^3), N is the applied load (N), and S is the friction distance (m) [61, 99, 100].

For the determination of the wear amount of very low wearing metal-on-metal pairs, lubricant samples obtained during a wear test can be analyzed by inductively coupled plasma mass spectrometry (ICP-MS) to measure the metal ion concentration [93].

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