Biotribology
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Edited by

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Tribology is the "science and technology of interacting surfaces in relative motion" and embraces the study of friction, wear, and lubrication. By extension, biotribology is usually defined as the tribological phenomena occurring in either the human body or in animals. Therefore it is possible to consider tribological processes that may occur after implantation of an artificial device in the human body and the tribological processes naturally occurring in or on the tissues and organs of animals. Animals, including humans, possess a wide variety of sliding and frictional interfaces.

The purpose of this book is to present a collection of examples illustrating the state-of-the-art and research developments in biotribology. Chapter 1 provides the biotribology of total hip replacement, the metal-on-metal articulation. Chapter 2 contains experimental wear studies of total joint replacements. Chapter 3 covers the influence of temperature on creep and deformation in UHMWPE under tribological loading in artificial joints. Chapter 4 contains information on large capacity wear testing. Finally, Chapter 5 is dedicated to biotribology of titanium alloys.

This book can be used as a textbook for a final undergraduate engineering course or as a reference on biotribology for postgraduate level. This book can also serve
as a useful reference for academics, materials and biomedical researchers, materials, mechanical and biomedical engineers, professionals in biotribology and medical tribology and related industries. The scientific relevance of this book is evident for many important research centers, laboratories and universities throughout the world. Therefore, it is hoped that this book will encourage and enthuse other research in this recent field of science and technology.

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1.1. Introduction

All structures of the human body undergo a natural process of aging over the course of life. This includes articular cartilage. Extreme wear of articular cartilage of the hip joint can be disabling and require treatment. The reconstruction of joint function by means of a hip joint replacement (see Figure 1.1) has been described as one of the most successful medical procedures [BER 05]. Most patients regain functionality of their affected joint within a brief period after a surgical intervention. Worldwide, about one million artificial hip joints are implanted annually.

The bearing surfaces of implants are subjected to friction and wear. Particles produced in the contact area between bearing partners are released into the tissue surrounding the implant. These particles can provoke inflammatory
tissue reactions, which can result in bone loss. Massive bone loss around an implant jeopardizes the fixation of the implant and results in loosening. This process, referred to as aseptic loosening, necessitates surgical revision of the implant. The frequency of revisions has increased dramatically in recent years.

Currently, one out of five artificial hips is subjected to revision within 15 years [KÄR 08]. Revision surgeries are complicated, risky and expensive, and revised implants are generally less successful than primary implants.

The recent rise in the use of joint replacements in young and active patients demands a drastic increase in implant durability. The selection of suitable materials plays an important role in the pursuit of this goal.

Currently, three main classes of materials are being used in hip replacements: ceramics, metals and polymers. The ceramics category includes aluminum oxide ceramics and zirconia-toughened alumina ceramics. The polymer category is dominated by ultra-high molecular weight polyethylene (PE), both in its conventional and cross-linked (X-PE) forms. Cobalt-chrome-molybdenum alloys (CoCrMo) make up the metal category.

These materials can be combined in several ways and are generally classified in hard-soft and hard-hard combinations. Hard-soft combinations consist of a metal or ceramic head paired with a PE or X-PE cup. Hard-hard combinations pair up two ceramic components (ceramic-on-ceramic) or two metal components (metal-on-metal). Hard-soft combinations are currently the most common. The prevalence of metal-on-metal bearings is however rising, in part due to their popularity among young and active patients.
1.2. Historical development of metal-on-metal bearings in total hip replacements

Total hip replacements took off around the world after the introduction of PE as a bearing material by Sir John Charnley in the 1960s [CHA 61]. While Charnley was developing PE on metal bearings, others were experimenting with metal-on-metal combinations. Most early developments took place in Great Britain. The McKee-Farrar prosthesis was implanted for the first time in 1950, followed by the Stanmore prosthesis in 1963 and the Ring prosthesis one year later [AMS 96]. The McKee-Farrar prosthesis (see Figure 1.1) was the most widely used metal-on-metal bearing in the 1960s and early 1970s [AMS 01]. The early success of these metal-on-metal bearings was, however, limited by a high and unacceptable failure rate [VIS 87]. Promising results of Charnley's PE bearing and the problems surrounding metal-on-metal bearings caused the latter to virtually disappear. McKee-Farrar ceased to implant his device after 1972 and PE bearings became the gold standard [AMS 96].

Aseptic loosening, which has been associated with PE particles, and the comparatively low wear of metal-on-metal bearings encouraged Müller and Weber in 1988 to bring a second generation of metal-on-metal bearings to the market (see Figure 1.1) [WEB 96]. The Swiss company Sulzer (now Zimmer) offers this bearing under the name Metasul® to a worldwide market. To date, more than 200,000 Metasul® devices have been implanted [TIP 05].

Since the introduction of the second generation, metal-on-metal bearings have been on the rise. Over the past few years, many developments have contributed to continuous improvement and third generation implants are now in use. An example of recent advances is metal-on-metal hip surface replacements (see Figure 1.1).
1.3. Design and materials

1.3.1. Implant geometry

Figure 1.2 is a schematic cut through a metal-on-metal surface replacement. Ideally, head and cup are two ball-shaped elements with slightly differing diameters. The small space between head and cup is referred to as radial clearance. No surface, no matter how well polished, is completely smooth and flat. The head and cup therefore deviate slightly from perfect spheres, resulting in deviations of roundness. The roughness of an implant surface is specified by the mean roughness, calculated as the arithmetic mean of absolute values. Geometric indicators used to characterize bearing partners include diameter, clearance, deviation of roundness and surface roughness.
1.3.2. Manufacturing methods and metallurgy

$\text{Co}_{28}\text{Cr}_{6}\text{Mo}$ is the most widely used alloy in metal-on-metal bearing material due to its good wear properties and high corrosion resistance. Traces of nickel, manganese or iron are also sometimes included. Two chemically virtually identical – but in their mechanical properties different – cobalt alloys are in use. Based on their manufacturing methods, these alloys are classified as wrought (ISO 5832-12, ASTM F1537) and cast (ISO 5832-4, ASTM F75) materials. In addition to the manufacturing method, carbon content critically influences metallurgical properties and microstructure. In this context, materials are classified as low carbon (lc, $< 0.15\%$) and high carbon (hc, $\geq 0.15\%$).
Figure 1.3. SEM Images of different CoCrMo Alloys: A) microstructure of a wrought CoCrMo alloy with high carbon content; B) microstructure of a cast CoCrMo alloy with high carbon content; C) microstructure of a cast CoCrMo alloy with high carbon content after heat treatment (HIP, SA)
1.3.2.1. CoCrMo wrought alloys (ISO 5832-12, ASTM F1537)

This alloy is used to manufacture hip replacement components through mechanical treatment or serves as a raw material for thermomechanical forming.

The microstructure is comprised of approximately 15-20 μm sized grains and at low carbon contents is monophasic and purely austenitic (face-centered cubic, fcc) [STR 96]. Increasing carbon content gives rise to chromium and molybdenum carbides, which results in a biphasic structure [LEE 06] (see Figure 1.3-A).

Carbides of the type M23C6 and M7C3 are about five times harder than an austenitic matrix [STR 96, LEE 06, SCH 96]. They are relatively small (approx. 3 μm) and regularly dispersed [STR 96, TIP 05]. These carbides cover about 5% of the surface area.

Under loading, a stress-induced face centered cubic (fcc) to hexagonal close-packed (hcp) martensitic transformation is possible. Martensite is harder and has impressive tribological properties [VAR 03, SCH 96]. Rising carbon content leads to higher tensile strength and ductility [STR 96, LEE 06]. The mechanical properties of wrought alloys are roughly twice as high as the properties of cast alloys (Table 1.1) [STR 96].

The ISO and ASTM standards differ only marginally in their mechanical demands. The chemical alloy definitions are the same in the two standards. However, the ASTM standard deviates slightly from the literature [TIP 05] in its classification of carbon content and defines low carbon in the range of 0-0.14% and high carbon in the range of 0.15-0.35%.
Table 1.1. Comparison of cobalt alloys according to [STR 96]
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<table>
<thead>
<tr>
<th></th>
<th>Cast high carbon</th>
<th>Wrought high carbon</th>
<th>Wrought low carbon</th>
</tr>
</thead>
<tbody>
<tr>
<td>Carbon content in %</td>
<td>&gt;0.20</td>
<td>&gt;0.20</td>
<td>&lt;0.08</td>
</tr>
<tr>
<td>Grain size (average) in μm</td>
<td>2000</td>
<td>20</td>
<td>15</td>
</tr>
<tr>
<td>Carbide size (average) in μm</td>
<td>30</td>
<td>3</td>
<td>-</td>
</tr>
<tr>
<td>Hardness HV</td>
<td>420</td>
<td>470</td>
<td>430</td>
</tr>
<tr>
<td>0.2% - Yield strength in MPa</td>
<td>450-600</td>
<td>830-1,200</td>
<td>500-950</td>
</tr>
<tr>
<td>Tensile strength in MPa</td>
<td>650-1,000</td>
<td>1,170-1,600</td>
<td>1,000-1,374</td>
</tr>
<tr>
<td>Bending fatigue strength in MPa</td>
<td>190-400</td>
<td>500-870</td>
<td>450-830</td>
</tr>
</tbody>
</table>

1.3.2.2. CoCrMo cast alloys (ISO 5832-4, ASTM F75)

Cast implants are manufactured using the investment casting method, also referred to as lost wax casting. This process involves the creation of a master pattern from wax, which is subsequently coated with a ceramic material. The ceramic layer can be used as a mould upon melting and removal of the wax. This method makes the production of complex geometries such as knee and hip implants possible [MET 04].

As is the case for wrought alloys, the ISO and ASTM standards for cast alloys are only marginally different with
regard to the chemical composition of the alloys. The mechanical requirements are the same in both standards.

In cast alloys the number and size of carbides rises with increasing carbon content. With a grain size of 2,000 μm, the microstructure is much coarser than the structure of a wrought alloy. The size of dendritic blocky carbides in fcc cobalt-rich matrices is about one order of magnitude larger than the carbides in wrought alloys [STR 96, SHI 93].

The structure of cast components often exhibits pores or holes, resulting in reduced mechanical properties [CAW 03]. Some manufacturers use hot isostatic pressing (HIP) and solution annealing (SA) in order to close these porosities and to achieve a homogeneous microstructure [NEV 04].

Non-heat treated, cast implant components are referred to by the term “as cast”. The HIP procedure exposes components to a temperature of 1,200°C and a pressure of 103 MPa in an argon atmosphere for about four hours. The primary goal of the HIP procedure is the closure of internal porosities [MET 04, NEV 04].

The SA procedure exposes components to a temperature of 1,200°C in a vacuum, followed by cooling in nitrogen. This process improves ductility and leads to a homogeneous microstructure [DOB 83, NEV 04]. Heat treatment causes blocky M7C3 type carbides to dissolve (see Figure 1.3-B) without a phase change in the matrix, leading to lamellar and insular M23C6 type carbides (see Figure 1.3-C) [CAU 02, VAR 03]. The carbide content is markedly reduced by heat treatment [CAW 03, MET 04].

In some surface replacements the cup is manufactured with a porous coating, which is supposed to facilitate bone ingrowth. This coating is applied in a sintering process similar to the HIP and SA heat treatments.
The main manufacturing- and material-dependent criteria influencing the mechanical and metallurgical properties of hip implant bearings include: carbon content of the alloy, primary manufacturing method (cast or wrought) and heat treatment.

1.4. Tribology of metal-on-metal bearings in total hip replacement

1.4.1. Wear and types of friction

Lubrication is a key factor defining wear and frictional properties of a hip implant. Friction and wear can be greatly reduced by a lubricating film separating the bearing partners. The mode of friction varies depending on implant geometry, surface roughness, material elasticity, load on and relative velocity of the bearing partners, as well as viscosity of the lubricant.

The Stribeck curve (see Figure 1.4) shows different modes of friction. Coefficient of friction ($\mu$) and film thickness ($h$) are shown as a function of the Sommerfeld number, which is unitless and based on viscosity ($\eta$), velocity ($v$) and pressure ($p$).

Given a narrowing gap, the sequence of modes of friction and increasing film thickness is as follows:

- boundary lubrication;
- mixed film lubrication;
- elastohydrodynamic lubrication;
- hydrodynamic lubrication.