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Wear and corrosion in medical applications

Mini-Review

Abstract: Biomaterials applied to replace or restore body functions are exposed to different mechanical forces and corrosion processes due to body fluids, which might result in the generation of corrosion products, wear debris and particles potentially leading to inflammation and inhibition or loss of function. This brief review will give a short overview about the processes of wear generation and corrosion, the occurrence of the respective wear products in different medical applications and their biological influences. Wear and corrosion are important factors that control and determine the long-term clinical performance of a biomaterial.

Keywords: medical devices, biomaterial, wear debris, corrosion, particles, inflammation, fibrosis

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1 Introduction

The specific function of degenerated or traumatized tissues/organs can be replaced or restored using a broad range of biomaterials applied in various forms of implants and medical devices in orthopaedic or dental surgery, as sutures or ligament replacement, in the cardiovascular field, skin wound healing or ophthalmologic and otologic applications. [1] Depending on the location and designated function (stress distribution, articulation, blood flow, light/ sound/ load transmission) the material has to fulfil specific requirements (morphology, porosity, mechanics, surface functionalization) and is exposed to various biological, chemical and mechanical influences. [1–3] According to the type of tissue metals, ceramics, polymers and appropriate composites are used. [4] Host factors (e.g. gender, age, medical/physiological constitution) play a crucial role for acceptance of the biomaterial just like the biocompatibility of the material. [5]

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1.1 Wear and corrosion

Biomaterials are exposed to different mechanical forces and corrosion processes due to body fluids, which might result in the generation of corrosion products, wear debris and particles potentially leading to inflammation and inhibition or loss of function.

"Wear can be defined as an […] progressive loss of material from one or both surfaces in relative motion between them." [6] Besides corrosion, abrasion, adhesion, fatigue, erosion can be classified as wear processes. [6] During the process of abrasion (two- or three-body-wear) a hard rough surface slides across a softer one. Two solid surfaces slide along each other during the process of adhesion and tend to plastic deformation of very small fragments, which is influenced by the properties of the lubricant. [6] Fatigue wear occurs during excessive application of cyclic loading to the material. [6] Hardness, wear resistance and fracture toughness determine the materials tendency to generate wear, which is further influenced by microstructural factors, lubricant rheology and geometry. [7,8] Particles can be generated by fretting induced by small-amplitude oscillatory micromotions. [6] Two-component implants can exhibit microgaps potentially invaded by body fluids (e. g. saliva) or microorganisms producing a biofilm to act as lubricant, which leads to friction. [9]

Corrosion can be described as a gradual degradation of materials due to the electrochemical environment of body fluids (e. g. blood, plasma, saliva) containing anions and cations and dissolved oxygen. [3] Blood contains high levels of electrolytes and is therefore provoking accelerated material corrosion due to high ionic conductivity and anodic and cationic reactions. Thereby, metallic materials react in different and even galvanic manner in the corrosion process. [3,10] The process of corrosion is influenced by present bacteria, pH, which is decreased around the implant as well as thermodynamic forces and the kinetic barrier. [3] About 2.5×10^{-4} mm/year are mentioned as a tolerable corrosion rate for metallic implant systems. [3] Moreover, the presence of wear accelerates the process of corrosion or occurs simultaneously (tribocorrosion), even synergistic interactions are described. [3,9] The loss of structural integrity and function of the implanted biomaterial ensue. The rapid formation of a stable oxide layer at the surface or appropriate coatings might

inhibit or at least delay corrosion, since corrosion is also associated with delayed wound healing. [3] Although, wear debris, ion release and aseptic implant loosening are mainly associated with joint prosthesis in orthopaedics [11] there are other applications of tribology in the biomedical field, too.

1.1.1 Orthopaedics

Corrosion has also been identified in stem-cement interfaces of hip-implants, but also in bone plates and screws at the bone-stem and permanent implants of toe, finger, spinal or shoulder. [3] Stainless steel (SS) and commercial pure titanium used for internal fixation showed an 86 %-rate to corrode after one year of implantation with significantly increased corrosion grades for SS [12] and distinct inflammation and tissue reactions in the surrounding soft tissue with high amounts of particles [13]. Spinal fixation devices made of SS were examined regarding the generation of particles in vivo and showed up to 22.3 x 10^9 particles per gram in some specimen. [14] Within 57 retrieved thoracolumbar spine implants wear occurred in 75% and corrosion in 39% of the implants after at least one year. Thereby, 58 % of SS implants were affected compared to titanium. Corrosion was mainly present at the interfaces between the single implant components. [15]

Thomson et al. [16] examined the biocompatibility of particles from ligament prosthesis in vitro and in vivo revealing no cytotoxicity, but an inflammatory potential of high concentrations at least in vitro.

1.1.2 Dentistry

(Artificial) tooth wear is affiliated to the processes of attrition (tooth-tooth contact), erosion (tissue dissolution by acidic substances) and abrasion (interaction between teeth and other materials) and also delamination and fatigue. [17,18] Litonjua et al. [19] name a material loss of 50 to 68 µm/year of enamel in natural teeth indicating dependence from the studied population and age. The same forces have to be assumed affecting artificial dentures and dental composites. Turssi et al. [20] showed a volumetric loss between 0.4 to 1.6 mm^3 when applying 80 N at a frequency of 1.9 Hz for 105 cycles to five different dental resin composites, while the upper limit of the physiological chewing frequency are 2 Hz. Arsecularatne et al. [21] measured an average coefficient of friction values in the range 0.03 - 0.09 after comparing different resin materials under 2 - 10 N with 66 cycles/min and artificial saliva lubricant. Just to name a few studies. Oral implants are made of a metal (abutment) and a ceramic component (abutment, crown), whose mismatch of mechanical properties can be a reason for mechanical failure. [9,22] Investigating the influence of titanium particles from dental implants to mesenchymal stem cells (MSCs) and fibroblasts in the process of peri-implantitis, Bressan et al. [23] stated a chain of events outgoing from the increased production of reactive oxygen species. Thereby, neutrophil cell recruitment resulted in ECM degradation due to higher levels of matrix metalloproteinases, which led to an altered differentiation of the MSCs and activation of osteolytic processes. Titanium particles in the range from 9 - 54 nm diameters have been detected in peri-implantitis biopsies. [9] The corrosion of endosseous implants is influenced by the acidic pH of local body fluids, temperature, plaque and food properties. Galvanic corrosion is a common problem. [3] Metal ions have been detected in the implant surrounding gingiva and bone. [9]

1.1.3 Cardiovascular implants

Artificial heart valves or ventricular assist devices (VAD; e. g. pumps) involve moving components and are therefore able to produce mechanical wear and friction. Further, a blood caused fluid-friction is generated at the surface of cardiovascular devices, also a friction between device and soft tissue. [24] Mechanical heart valves (MHVs) and bioprosthetic heart valves (BHV) underlie different loading forces being higher in MHVs; which are associated with impact wear and friction wear. [24] The bearing wear rate measured in a VAD was less than 1.46 µm per year, but the applied measurement procedure was strongly restricted. [24] Nitinol as shape-memory alloy is used for minimal-invasive applications in heart valve therapy. Nitinol wires have been tested regarding their wear resistance performing an accelerated wear test (up to 20 Hz, 200 million cycles) to imitate the effects of long-term wear (5 years) that led to an increase of local corrosion rates. [25]

The biomaterial properties of stents might cause restenosis, e.g. in case of drug-eluting stents and peripheral vessels. [26] Generation of metallic debris and alteration of mechanical properties due to in vivo stress corrosion can lead to material fatigue and stent fracture as shown in retrieval analysis of explanted stents made of nitinol. [3,26] Especially, in the overlapping area of stents fretting wear is observed. [26] Wear has been detected in an explanted vascular stent-graft for endovascular aneurysm repair, which might lead to blood leakage. [24] Nickel ions from shape memory alloys are further associated with allergic reactions. [3] The usage of biodegradable magnesium materials could elude the problem of corrosion-induced cracking and fatigue. [3]

Wear studies regarding heart valves, cardiovascular advices or stents are barely published. Nevertheless, presence of wear should not be excluded since tribological processes occur.

1.2 Consequence of wear and corrosion

Cellular effects depend on the material, but also on particle size as well as morphology, chemistry and number, which differ between materials and applications. [9] Fibrils and specific morphologies of particles are associated with an increased cellular reaction [27], as are smaller particles. [9] Particles in the range of 0.24 - 7.2 µm are considered to be notably reactive and inflammatory. [28]

The peri-prosthetic environment hosts different tissue-specific cell types. Proteins attach to the surface of particles and ions allowing their internalization into cells. [9] Particle contact, phagocytosis or pinocytosis of particles (size 150 nm - 10 µm) cause the release of various mediators (interleukins, growth factors, chemokines). [28] Monocytes and macrophages play a major role in recognition of particles causing recruitment, proliferation, as well as differentiation and maturation of precursor cells via cytokines and pro-inflammatory mediators. [29] Within this particle-mediated cascade present cells mutually inhibit or activate each other provoking inflammation, fibrosis or tissue degradation due to e. g. matrix metalloproteinases. [9,28,30] Particles linked to bacterial lipopolysaccharides trigger increased monocyte migration in gingival and bone tissue. [9] Also, cell properties, such as the cellular maturation state, have shown to influence the reaction to particulate wear debris. [31]

Concentration of titanium in serum and urine has shown to be increased after surgery. Depending on particle size the distribution via gastrointestinal tract, blood and lymph is probable, even via attachment to major transport proteins. [3,9,26] Enrichment of particles in erythrocytes leading to toxicity and immunologic effects has been proven. Titanium oxide particles of 25 - 80 nm and 155 nm have been detected in several organs. [9]

Elements released in the course of corrosion cause inflammatory reactions and local immune response by affecting the local environment of the tissue, but also discoloration of it and a foreign body reaction. [9] Titanium ions stimulate interleukin secretion by macrophages resulting in increased bone resorption. [9] Nickel and CoCr cause carcinogenicity, while vanadium is proved to be cytotoxic and aluminium is suspected to cause Alzheimer disease. [3] Even respiratory and cardiovascular diseases have been associated with wear particles. [9]

2 Conclusion

Biocompatibility studies of biomaterials are mainly conducted using bulk materials or scaffolds, although it has been shown that particles and ions of the same material cause cytotoxicity and inflammation in comparison [32], therefore wear and corrosion are important factors that control and determine the long-term clinical performance of a biomaterial. [33] Although, wear and corrosion levels might seem trivial in particular applications, there is a necessity for the usage and establishment of tissue and application specific methods and a combination of physical, chemical and biological ones to measure and value even low concentrations.

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