Factors Responsible for Secondary Surgery Due to Implant Failure in Tissue Replacement

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***Abstract*—Despite the fact that there is abundant availability of metallic materials, it might be quite difficult to match the expectations of young, energetic individuals after surgery. Revision surgery is necessary when an implant fails for a number of reasons. Aseptic loosening, metallosis, surgical or operational failure due to human or implanting machinery error, infectious failure in the periprosthetic joint due to inadequate hygiene maintenance during or after surgery, poor bone integration, and other mechanical mismatches can all lead to orthopedic implant failure. Young's and bulk modulus mismatches accounted for 18% of implant failures brought on by aseptic loosening. It causes significant bone loss or osteolysis, gradual wear and tear of the high-load bearing joint, stress-shielding effects, debris release that triggers negative cellular reactions, osteoporosis illness, and implant failure. Nearly 20% of implant failures are caused by infection. It causes septic loosening, which impairs implant function and causes discomfort and redness.**

***Index Terms*—Secondary surgery, Implant failure, Natural Tissue, Replacement.**

1. INTRODUCTION

Even if there is a plentiful supply of metallic materials on the market, achieving the expectations of young and energetic individuals after surgery is quite difficult. Revision surgery is required when an implant fails for a variety of reasons. As shown in Figure 1, failure of an orthopedic implant can be caused by a mechanical mismatch such as aseptic loosening, metallosis, surgical or operational failure caused by an error from human or implanting machinery, infectious failure in the periprosthetic joint caused by improper hygiene maintenance during or after surgery, poor bone integration, and others. Aseptic loosening created a mismatch in mechanical parameters such as Young's and bulk modulus, accounting for 18% of implant failures. It causes stress-shielding effects, gradual wear and tear of high-load bearing joints, the production of debris that triggers undesirable cellular responses, significant bone loss or osteolysis, and eventually osteoporosis illness and implant failure. Infection causes roughly 20% of implant failures. It causes septic loosening, which causes discomfort, redness, and poor implant function (Heimann 2017; Quinn et al. 2020; Siddiqui et al. 2018; Kim et al. 2012; Lahiri et al. 2011b; Roy 2020).

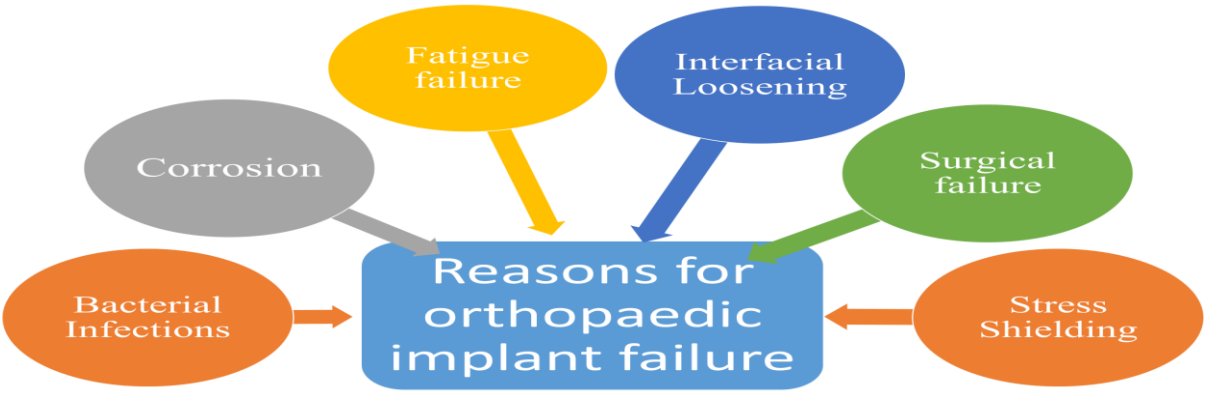

Figure 1 The factors responsible for secondary surgery due to implant failure.

Titanium alloys are the preferred material for many critical applications due to their high toughness-to-mass ratio and corrosion resistance. Titanium alloys are employed in a variety of demanding applications, including static and rotating gas turbine engine components and oil refinery heat exchangers. Because of their high corrosion resistance, titanium alloys are utilized in chemical processing, desalination, valve and pump components, and maritime equipment. Furthermore, the high toughness-to-mass ratio reduces the weight of components made from this alloy. The most common titanium alloy is Ti-6A1-4V. However, Ti-6A1-4V alloy has poor surface characteristics for wear and high temperatures. However, following implantation, a link with live bone often does not form, and the implant's integration into bone tissue typically takes many months. As a result, there is considerable interest in expediting the process of osseointegration and therefore lowering surgical constraints.

2. LITERATURE REVIEW

Steffi et al. (2018) proposed manipulating osteoclast interactions with orthopedic biomaterials to balance osteoclast resorption and osteoblast deposition for the best orthopedic surgery implantation. It investigated the effects of implant surfaces, bioceramics, and polymers on osteoclast activity, taking into account topography, chemical composition, and surface modifications. According to studies, coarser implant surfaces stimulate osteoclast activity, while smooth surfaces inhibit differentiation. Surface alterations caused by anti-osteoporotic medicine may improve implant integration by reducing osteoclast activity. In vitro studies revealed that implant surface properties influence osteoclastogenesis, osteoclast activity, and bone remodeling. The study identified research gaps, such as osteoblast activity studies without osteoclast differentiation. The authors proposed investigating implant surface topography, chemical compositions, and physiochemical impacts on osteoclast behavior. Future study may show that pharmaceuticals that regulate osteoclast activity improve osteointegration. The results may improve bone integration in orthopedic implants over time (Steffi et al., 2018).

Cadar et al. (2017) investigated nanostructured and multisubstituted hydroxyapatite (HAp) including Mg, Zn, Sr, and Si as orthopedic and dental bone replacement materials, as well as metallic implant coatings. Biomaterials were created and described, with Ca, P, Mg, Sr, and Si release in water and SBF monitored for 1-90 days. XRD and FTIR validated the biomaterial structure and water-SBF interactions. The time-dependent element release was evaluated using ICP-OES. Multisubstituted HAp materials produced physiological components at controlled rates, suggesting that they might be employed for bone regeneration and as coatings to enhance metallic implant biocompatibility and osteointegration. Replacement limit inconsistencies and the inability to reliably incorporate replacement components throughout material manufacture and characterization were not addressed in the study. The paper concentrated on orthopedic and dental applications; hence it did not include other medical applications or innovative biomaterials for future research. Some studies reported 2.46 wt% magnesium substitution in hydroxyapatite, while others proposed higher amounts. Silicon phosphorus replacement has a theoretical limit of 5.8 wt%, while real substitution is often between 3-5 wt%, highlighting the need for more study to better understand these restrictions and their implications on material characteristics and performance (Cadar et al. 2017).

Instead of biocompatible and biodegradable bone wax, hydroxyapatite (HA) was investigated as an orthopedic biomaterial owing to its hemostatic and bone-regenerating capabilities. HA beat $CaSiO_3$, calcium-attapulgite, and calcium tripolyphosphate in blood clotting activity, particularly when corrected for surface area and activity. Ca2+ ions Synthetic HA increased blood clotting response, which is required for bone repair and integration, hence its effects on biological tissues were investigated. The hydrothermal production of HA utilizing $Ca(OH)_2$ and Na_2HPO_4 allowed for precise control of properties for optimum tissue interaction. Comparison of hemostatic polymers with chitosan. More research is required to better understand how HA production influences biological interactions and hemostatic efficacy. Yang et al. (2017) propose investigating HA's in vivo interactions with biological systems in order to identify its biodegradable and biocompatible hemostatic bone healing properties.

Hendrik et al. (2016) employed precise force field and pH-resolved surface models to simulate the chemical bonding, structural, surface, interfacial, and mechanical properties of hydroxyapatite (HA) based on experimental data. This force field was used by AMBER, CHARMM, GROMACS, and others to model apatite-biological systems of various compositions and ions. pH-resolved surface models provide better approximations of apatite surface interactions, especially at different pH values. It discusses how HA affects bone and tooth mineralisation. Quantitative monitoring of inorganic-biological assembly at 1–100 nm aids understanding of complex biological–mineral interactions. Previous models incorrectly predicted HA surface chemistry, interfacial interactions, hydration, and protonation. Work addresses these concerns. These gaps impede bone and tooth mineralization research. The study also demonstrates that current models underestimate high OH-ion concentrations at the HA surface during hydration and protonation, as well as physiologically uncommon pH values above 14. This limitation is required for biological system solution simulation. To better recreate the habitats of living organisms, the authors recommend demonstrating protonation effects on phosphate ions at various pH levels and using more realistic solution conditions.

3. MATHEMATICAL MODEL FOR SURFACE ENERGY MEASUREMENT

The crystalline lattice parameters of samples planned analyse using X-ray diffraction, Panalytical X'Pert, Cu Ka radiation. The diffracted signal was collected over a 20 range of 60° with a step size of 2° and a fine slit(Saber-Samandari et al. 2013; Tercero et al. 2009). Ca/P ratio, cruystallinity, and calcination loss of HA phase present in the coatings was calculated using ISO standards.

$$D = \frac{K\lambda}{\beta_{\frac{1}{2}} \cos\theta} \qquad (1)$$

Where, 'K' is the broadening constant (0.9), '$\boldsymbol{\lambda}$' is the wavelength ($\boldsymbol{\lambda}$ = 1.542A˚) of CuKa radiation, $\beta_{1/2}$ is FWHM for the diffraction in radians, 'θ' is the diffraction angle.

Crystallinity Index (CI) has been calculated by considering the intensity of (3 0 0) diffraction peaks (I_{300}) and the intensity of the valley between (1 1 2) and (3 0 0) diffraction peaks ($V_{112/300}$) using Equation (2)(Hussain et al. 2023)

$$CI = \left(1 - \frac{V_{112/300}}{I_{300}}\right) \times 100 \qquad (2)$$

The Owens-Wendt-Rabel-Kaelble (OWRK) method is a widely used approach for calculating the surface energy of solids, as well as its polar and dispersion components. This method involves using a geometric mean equation to relate the surface tension of a liquid and the surface energy of a solid. The OWRK method is particularly useful in surface chemistry and adhesion studies, as it helps to determine the interfacial tension and wettability between different materials.

The total surface energy (σ_s) of a solid is the sum of its dispersion and polar components:

$$\sigma_s = \sigma_s^d + \sigma_s^p \qquad (3)$$

Similarly, the total surface tension of the liquid (σ_1) is the sum of its dispersion and polar components:

$$\sigma_1 = \sigma_1^d + \sigma_1^p \qquad (4)$$

The interfacial tension between the solid and the liquid (σ_{s1}) is calculated using a geometric mean equation that combines the dispersion and polar components of both the solid and the liquid:

$$\sigma_{s1} = \sqrt{(\sigma_s^d \cdot \sigma_1^d + \sigma_s^p \cdot \sigma_1^p)} \qquad (5)$$

The contact angle (θ) between the solid and the liquid is an important parameter used in this method. It is related to the surface energy components of the solid and the liquid through the following equation:

$$\cos(\theta) = (\sigma_{s1} - \sigma_{s1}) / \sigma_1 \qquad (6)$$

The OWRK method is a critical tool in surface science for determining the adhesive properties between solid and liquid surfaces. The method uses the surface energy components of both the solid and liquid phases, and the contact angle to calculate the interfacial tension, which provides important information about the wettability and adhesion characteristics of materials.

The total surface energy of a solid (σ_s) is made up of its dispersion and polar components (σ_s^d and σ_s^p).

Similarly, the liquid's total surface tension (σ_1) is composed of its dispersion (σ_1^d) and polar (σ_1^p) components.

The interfacial tension between the solid and liquid, σ_{s1}, is a key factor in determining the interaction between the two surfaces and can be derived using the OWRK geometric mean equation.

By analyzing the contact angle (θ) and the surface energy components of the solid and liquid, one can gain insight into the adhesive strength and wettability of different materials. This method is widely applied in various industries, including coatings, adhesives, and surface treatments.

The Owens-Wendt-Rabel-Kaelble (OWRK) method provides a robust and reliable framework for calculating the surface energy components of solids and liquids. By using geometric mean equations, this method enables the calculation of interfacial tension and contact angles, both of which are essential in understanding the interactions between materials. This approach is indispensable in surface science, where the properties of interfaces play a significant role in determining the performance and behavior of materials in real-world applications.

4. RESULTS AND DISCUSSION

Materials introduced into the body must be handled carefully since they directly replace bodily components and bio-functions. As a result, there is no possibility of any infection from the biomaterial causing implant failure.

4.1. Mechanical Factors

- Fatigue Failure: Repetitive stress over time can lead to cracks or breakage in implants.
- Wear and Tear: Continuous articulation in joint replacements (e.g., hip or knee) can lead to debris generation and mechanical degradation.
- Loosening or Dislocation: Inadequate fixation, poor alignment, or patient activity can cause implants to loosen or migrate from the original site.
- Stress Shielding: Implants taking on too much load can lead to bone resorption due to reduced mechanical stimulation of surrounding bone tissue.

4.2. Biological Factors

- Infection (Peri-implantitis or Biofilm Formation): Bacterial colonization of the implant surface can trigger chronic inflammation, leading to tissue damage and implant failure.
- Immune Response or Allergic Reactions: Host rejection or allergic response (e.g., to metal ions like nickel or cobalt) can compromise integration and function.
- Poor Osseointegration: Inadequate bone-implant bonding, especially in dental or orthopedic implants, leads to instability.

4.3. Material-Related Issues

- Material Degradation: Corrosion or oxidation of metallic implants, or degradation of polymeric materials, can compromise mechanical integrity and biocompatibility.
- Inappropriate Material Selection: Use of non-biocompatible or suboptimal materials can result in toxicity, immune responses, or mechanical failure.

4.4. Design and Manufacturing Defects

- Improper Design Geometry: May not accommodate stress distribution or biological integration effectively.
- Manufacturing Defects: Microcracks, surface irregularities, or poor finishing can initiate early failure.

Table 1 Multifunctional roles expected from biomaterial

Characteristics	Description
Augmentable	Ability to enhance or supplement the natural function of body parts
Anti-fungal	Resists or not serve as a sink for fungal development
Anti-fouling	Materials surface property to prevent bacterial attachment
Anti-corrosive	The material should resist corrosion in the biological environment
Bactericidal	An ability of a material to act and directly kill the pathogen
Bioactivity	Supportive response of the material to perform the biological functions of the tissues

Characteristics	Description
	surrounded by it
Bioactive fixation	In the permanent implantation, a firm chemical and biological bond formation at the junction of material surface and tissue interface.
Bio-compatibility	Must not elicit an adverse response or reaction to surrounding tissue, compatible to blood, and be integral to the body environment
Biostability	The material ability to maintain its properties for a longer duration *in vivo*
Interfacial stability	The ability of the material to prevent surface mechanical failures under high load-bearing conditions
Non-carcinogenic	Must not illicit carcinogenic material to form an inflammation or cancer
Nontoxic	Must not release toxic substances which harm the body environment
Nonpyrogenic	Materials should not elicit heat or fever when inserted into the body
Osseointegration	The ability of material in the enhancement of bone cell growth, which increases the interaction between the implant and surrounding tissue
Osteoconductive	The ability of a material to facilitate bone ingrowth through the surface of a biomaterial
Osteoinductive/ osteogenetic	Stimulating ability of a material to develop bone-forming cell lineage into the material the process also referred to as osteogenesis
Resorpability	Support gradual degradation over time to be replaced by the natural tissue
Sterilisable	Capable of undergoing sterilization to kill microbe if present by a different technique like dry heat, autoclave, ethylene oxidation, radiation, without losing its original property
Therapeutic capable agent	Supportive for drug delivery and growth factors at required times
Wettability	The material tendency of adherence/repulsion with the water molecules.
Wear-resistant	In bone joint replacement, the material should resist friction and not elicit wear particle into the body

Despite the therapeutic effectiveness of HA coatings, their brittleness results in worse mechanical and tribological qualities. Particularly poor surface mechanical qualities, such as low fracture toughness, bending strength, bonding strength, tensile strength, and wear resistance. Furthermore, HA-modified biomaterial surfaces are prone to bacterial colonization. As a result, researchers in this field will confront several problems and possibilities when creating HA composite coatings in order to achieve required qualities and overcome HA coating shortcomings.

5. CONCLUSIONS

A combination of surgical, biological, mechanical, and patient-specific variables often leads to secondary procedures. Careful material selection, surgical planning, patient screening, and postoperative care are necessary to reduce implant failure. These issues are being lessened by emerging technologies such as bioactive coatings, smart implants with sensors, and personalized 3D-printed implants. These biomaterials are chosen for their properties, which include biocompatibility,

degradability, mechanical strength, and bioactivity. Natural polymers mimic the native extracellular matrix to support cell adhesion and signaling; synthetic polymers allow precise tuning of degradation and mechanics; ceramics and bioactive glasses promote bone ingrowth; metals and alloys provide load bearing durability; composites synergize multiple properties; hydrogels recreate soft tissue environments and enable drug/cell delivery; and decellularized matrices retain native architecture and biological cues to guide regeneration.

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