J. Electrochem. Sci. Eng. **13(1)** (2023) 193-214; http://dx.doi.org/10.5599/jese.1636

Open Access : : ISSN 1847-9286
https://pub.iapchem.org/ojs/index.php/JESE

Review paper

Corrosion cracking in Mg alloys based bioimplants

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Received: December 10, 2022; Accepted: January 28, 2023; Published: February 19, 2023

Abstract

Recently, magnesium alloys have garnered a lot of interest as a potentially useful material for applications involving biodegradable implants. Cracking or fracture of metal-based implants under the combined action of corrosion and mechanical stresses, namely stress corrosion cracking (SCC) is an obviously critical criterion before any new material might be deployed as implants. Cracking or fracture of metal-based implants occurs under the simultaneous action of corrosion and mechanical stresses. This article gives a review of the existing literature on the SCC of magnesium alloys in corrosive environments, including simulated body fluid and the accompanying fracture process. It also indicates the knowledge gap that exists in this area of research. In addition, a high-level review of the preventative measures that may be taken to avoid potential corrosion fatigue failures in magnesium alloys is provided.

Keywords
Biomaterials; coatings; biocompatibility; implants; magnesium implants

Introduction

It is one of the most difficult tasks of our day to find solutions to the health problems faced by an ageing population. The usage of magnesium alloys to implant devices is emerging as an innovative method. The temporary implants are made out of classic materials that are still regularly used, such as stainless steel, Ti-alloys, Co-alloys, and other similar materials, it is usual practice to remove them by the use of a surgical procedure. This second operation results in more stress for patients and additional expenses, in addition to the potential for consequences such as patient morbidity and infection.

Mg boasts finest biocompatibilities with human body, making it one of the most desirable materials for use in metallurgical engineering. Additionally, magnesium alloys have the highest level of mechanical agreement with bones. Therefore, the compatibility of the alloys based on magnesium with human implants draws attention of researchers [1,2]. To be suitable for this application, the alloys must be resistant to cracking or fracture when subjected to both mechanical stresses as well as corrosive fluid. Magnesium alloys as a result of loading in the human body is an important study area, such as:

- When using implants made of conventional materials, one of the primary concerns [3] always involves the possibility of breakage caused by human bodily fluids. This kind of fracture of magnesium alloys used as bio-implants is a study field that is severely underexplored [4,5].
- Recent research [6-9] has shown that simulated-body-fluid aided stress corrosion cracking (SCC) may occur in magnesium alloys.
- In-vivo testing of a few different magnesium alloys designed specifically for use as body implants [10-12] revealed that these particular alloys had the unique property of harmlessly melting away as shattered bone heals (Figure 1). However, the resistance of alloys to human body fluid-assisted fracture is still a key problem, and it is a study area that has been woefully underexplored.

The main aim of this article is to facilitate a specific study of corrosion fatigue (CF) and fracture processes of magnesium alloys when exposed to corrosive.

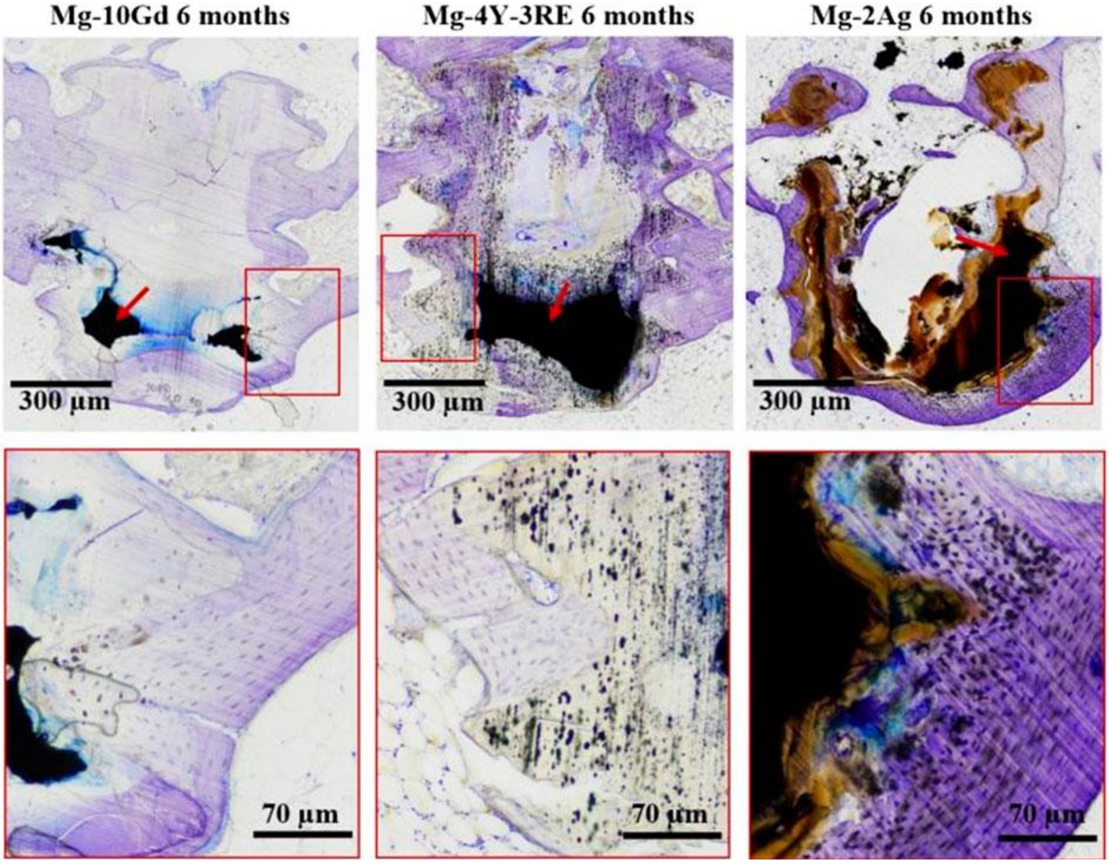

Figure 1. *After six months of implantation, 2D histology slides of Mg-10Gd, Mg-4Y-3RE, and Mg-2Ag that were stained with toluidine blue were examined. For every kind of screw, there is visible residual metal indicated by the red arrows in the black regions* [18]. *Permissions under CC NY 4.0 International Attribution*

Mg alloys as bio-degradable orthopedic implants

Bio-degradable orthopedic implants are medical devices that are designed to be absorbed by the body over time. They are typically made from polymers that are designed to break down into smaller particles and be eliminated from the body. These implants are used for a variety of orthopedic applications, such as joint replacement, fracture repair, and tendon repair. They offer several advantages over traditional implants, such as reduced inflammation and scarring, faster recovery time, and reduced risk of infection. Bio-degradable implants also reduce the need for revision surgery, as the implant will eventually break down and be eliminated from the body. Magnesium has lower elasticity modulus and specific density *i.e.* 42-45 GPa and 1.74-2.0 g/cm^3 respectively, both material properties of Mg resembles with human bones [13,14]. The use of magnesium alloys in the orthopedics implants rapidly increase the attention in research [13-15]. As a result, in contrast to the implant materials used in past, the Mg based biodegradable products are non-toxic to human body. In point of fact, it has been reported that the Mg^{2+} ions released due to degradation helps in the healing and growth of tissues [16]. However any excess Mg^{2+} is safely eliminated through the urine [15]. Additionally, the price of Mg alloys is significantly lower than the price of traditional implant alloys.

In spite of the fact that magnesium and its alloys possess a number of characteristics that are highly desirable in human implants; the use of these materials is extremely uncommon. Magnesium orthopedic implants are a type of implantable device used in orthopedic surgery to treat a variety of bone-related conditions. These implants are made from magnesium alloys, which are lightweight and biocompatible materials. They are designed to be used in place of traditional metal implants, such as titanium or stainless steel, due to their superior mechanical properties. Magnesium implants are often used to repair fractures or to replace missing or damaged bones. They can also be used to support artificial joints and to treat osteoporosis-related bone fractures. Additionally, they are being investigated for use in drug delivery and regenerative medicine applications. The most significant barrier is low corrosive nature of magnesium alloys in the physiological environment, which has a pH range from 7.4 to 7.6 [17]. Mg alloys may witness loss of their mechanical integrity in physiological environment because of body fluids in the degradation process. This might occur before the tissues have adequate time to restore. Despite the fact that these limitations have reduced use of Magnesium based alloys especially for the permanent implants. However, still for temporary implants Mg alloys are used due to their desirable characteristics. Therefore, it is possible to make use of non-toxic elements based Mg alloys in the human body. As a result, it is of the utmost importance to evaluate this type of fracture in the more recent magnesium alloys that have been developed specifically for use in body implants.

Corrosion of Mg alloys

Magnesium orthopedic implants are biodegradable and are becoming increasingly popular for orthopaedic and trauma surgeries. However, their use is limited due to the potential for corrosion. Magnesium is highly reactive and can corrode in the body, releasing toxic ions into the surrounding tissue. This can lead to local inflammation, infection, and tissue damage. Additionally, the corroded magnesium can accumulate in the body, leading to systemic toxicity. To prevent corrosion, magnesium implants should be coated with a protective layer such as titanium, hydroxyapatite, or polymers. Additionally, the implants should be designed to minimize the contact between the magnesium and the body fluids. Even in a medium that is only moderately corrosive, such as SBF, magnesium alloys corrode at an alarming rate. Magnesium corrosion always involves the release of a substantial amount of hydrogen. Hydrogen gas should be avoided in large quantities because it can cause subcutaneous gas bubbles to form, which can cause tissue layers to separate and potentially impede blood flow [19-21]. It has been revealed from the In-vivo and In-vitro analysis that due to HBF pitting is observed on the alloys mainly derived from Magnesium [22]. The existence of $Mg_6Zn_3Ca_2$ particle has also been reported that an alloy that was developed specifically for temporary implant applications, known as ZX50 [12]. This finding was made possible by the study of the alloy. When considering the use of magnesium alloy as a possible implant material, it is essential that the material decay slowly and uniformly rather than experiencing any localized deterioration. The pits results into the formation of brittle surface as atomic hydrogen enters into the Mg matrix due to bare and film free surface [5]. A newly developed ZX10 Mg alloy has showed remarkable resistance to pitting in addition to exhibiting the desirable characteristics of gradual and homogenous deterioration. This alloy does not produce any hydrogen bubbles and degrades in the intended manner. The Zn concentration was decreased, which resulted in the creation of the $(Mg, Zn)_2Ca$ phase. This phase is less noble than the Mg-matrix, which led to improved degrading performance [23].

Stress corrosion cracking and corrosion fatigue of Mg alloys

Stress corrosion cracking (SCC) is a form of corrosion that occurs when a material is subjected to both a tensile stress and a corrosive environment. It is a form of localized corrosion that is especially hazardous because it can occur without any external signs of corrosion, such as pitting or discoloration. SCC can occur in metals such as stainless steel, aluminum alloys, and copper alloys, and is often more rapid than general corrosion. It is usually caused by a combination of chemical and mechanical factors, such as a combination of high tensile stress and a corrosive environment. SCC is often encountered in industries such as oil and gas, petrochemical, and aerospace, and can lead to catastrophic failures of components and structures if not addressed properly. Implant devices are subjected to significant mechanical loadings when they are in the presence of HBF. As an example, a spine may be subjected to a load that is more than 3500 N, while a cardiovascular stent is continually subjected to cyclic loading as a result of the beating of the heart [5,7]. The presence of CF and SCC is a real possibility because to the dynamic stress that occurs inside the human body, in addition to the corrosive physiological environment. Failures of this kind often take place at stresses that are far lower than the design stresses for an environment that does not cause corrosion.

To prevent SCC in biomaterials, it is important to use corrosion resistant materials, properly clean and sterilize the material, use proper design considerations to reduce stresses on the material, and properly coat the material to protect it from corrosion. Additionally, material selection and design should be tailored to the specific environment the material will be exposed to, as different environments may have different levels of corrosive factors.

This is due to the ductile nature of the material due to which it undergoes elongation before fracture in corrosive environment. There have been multiple examples of failures due to fatigue in the human body [24-26], despite the fact that conventional implants have excellent fatigue strength. These unexpected breakdowns of an implant may have major repercussions, such as the arduous process of removing the defective device and the excruciating irritation or inflammation of the tissues around the implant [27]. When it comes to the use of metallic implants, this particular aspect of high cycle fatigue (HCF) is of the utmost importance. The fatigue strength of various implant materials is compared with that of native bone after 107 cycles in corrosive bodily fluid. In light of the fact that magnesium alloys have a lower resistance to fatigue in comparison to traditional implant materials it is absolutely necessary to carry out a comprehensive categorization of the fatigue behavior of Mg alloys before implantation.

CF and SCC will consider as a serious for implants made of Mg alloys due to the following reasons: (a) sharp shapes of temporary implant devices, and (b) Pitting of Mg alloys in chloride solutions [28] also in in HBF [5-9]. Implant devices made of Mg alloys will have a tendency to have sharp contours In point of fact, it has been documented that magnesium alloys are vulnerable to the occurrence of SCC in environments containing chlorides [29].

Even though research on the CF of magnesium alloys is very scarce, there is a respectable amount of published information on the SCC of these alloys in chloride solutions [29]. The mechanical and fractographic analysis depicts that investigations have shown that magnesium alloys are typically vulnerable to stress corrosion cracking (SCC). This is a pressing necessity (*i.e.* resistance to CF and SCC). It's possible that rare earth elements (RE) alloys with magnesium will be appealing. On the other hand, there is a paucity of information about the CF and SCC of alloys that include rare earth elements when exposed to chloride [29]. In addition, there is just one paper on the topic of CF investigations conducted on biocompatible magnesium alloys in bodily fluid [30]. The Fatigue limit of the natural bone, conventional implants materials and Mg alloys in physiological environments at 107 cycles were noticed. The fatigure strength ranges between 15-35 MPa for bones, 70-180 MPa for Mg alloys, 140-230 MPa for stainless steel, 250-280 MPa for Co-Cr alloys and 280-710 MPa for Ti-alloys, respectively [31-33].

Corrosion fatigue of Mg alloys

Corrosion fatigue is a type of fatigue failure in which a material suffers from both corrosion and fatigue. It occurs when repeated cyclic loading of a material leads to a combination of fatigue (mechanical) and corrosion (chemical) damage. In the case of biomaterials, corrosion fatigue is the process whereby repeated mechanical loading of a biomaterial leads to an increase in its susceptibility to corrosion in the presence of an electrolyte. Commonly used biomaterials such as stainless steel and titanium alloys are highly susceptible to corrosion fatigue. The damage caused can result in decreased strength and increased risk of failure through fracture or fatigue. The most common methods of mitigation against corrosion fatigue in biomaterials involve surface treatments such as passivation, electrochemical treatments, and coatings. Cracks caused by fatigue often begin at stress concentration sites, such as those that are developed during the manufacturing process [34]. A relatively recent study conducted testing on an alloy in air and a modified simulated bodily fluid hypothesized that inclusions and corrosion pits were the places where cracks first started to form (m-SBF) [35]. Because of this, a significant decrease in fatigue strength was detected when the alloy was evaluated using the m-SBF method (Figure 2). In addition, the nucleation and propagation of pits were found to be influenced by electrochemical circumstances as well as the amounts of applied stress. The researchers concluded that increasing the pitting resistance of Mg alloys in m-SBF would enhance the CF life of these alloys.

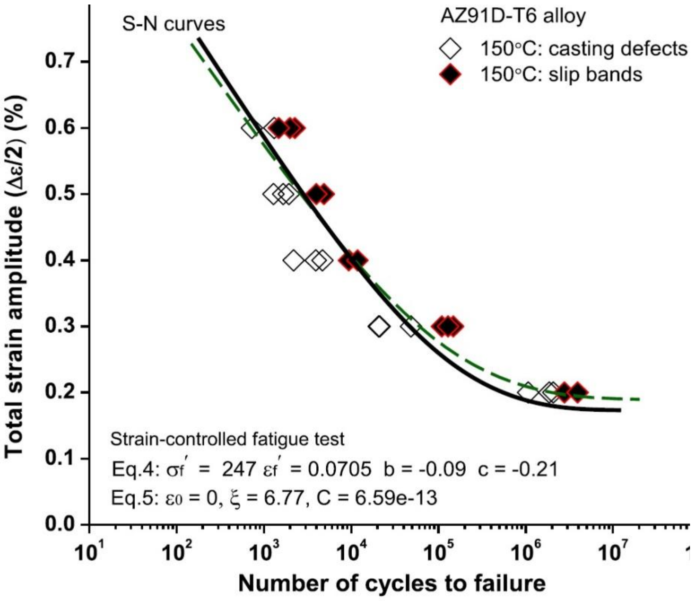

Figure 2. *S-N curves for AZ91-T6 Mg alloy under corrosion fatigue at 150 °C [36].*
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Slip bands and twin limits are two more possible locations for the initiation of cracks [37,38]. Because magnesium and alloys of Mg undergoes twinning [38]. In corrosive settings, a sharp tip of the corrosion pit provides high stress intensity, and so acts as the starting location for fatigue cracks (Figure 3). Pits play a significant part in the acceleration of fatigue fracture propagation because they may strike the necessary balance among the electrochemical dissolution and cyclic stress [30] came to the conclusion that fatigue fractures developed from micropores when the alloys were tested in air. When the alloys were put through fatigue cracking tests in a corrosive environment, corrosion pits were found to be the cause.

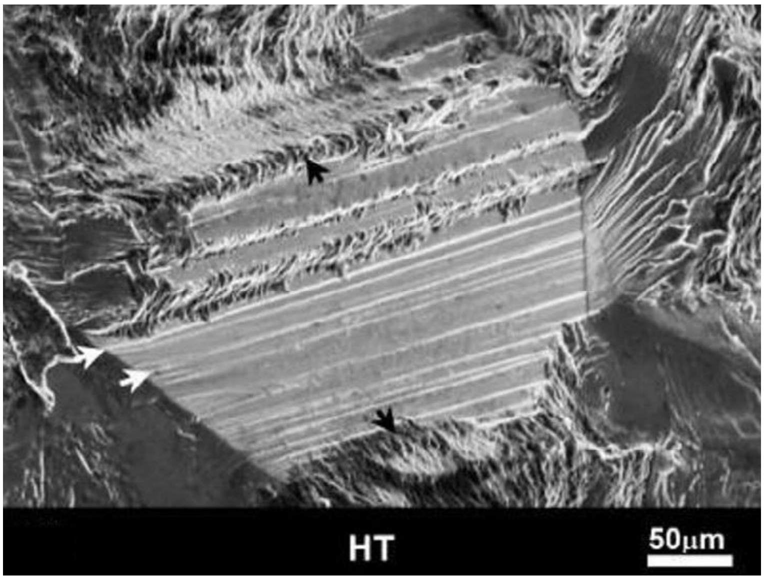

Figure 3. *SEM images of fatigue crack initiation sites in AZ91-T6 Mg alloy at 150 °C [36]. Permissions under CC NY 4.0 International Attribution*

The fatigue strength of a few different magnesium alloys is analyzed and compared in both air and many different corrosive conditions in Figure 4.

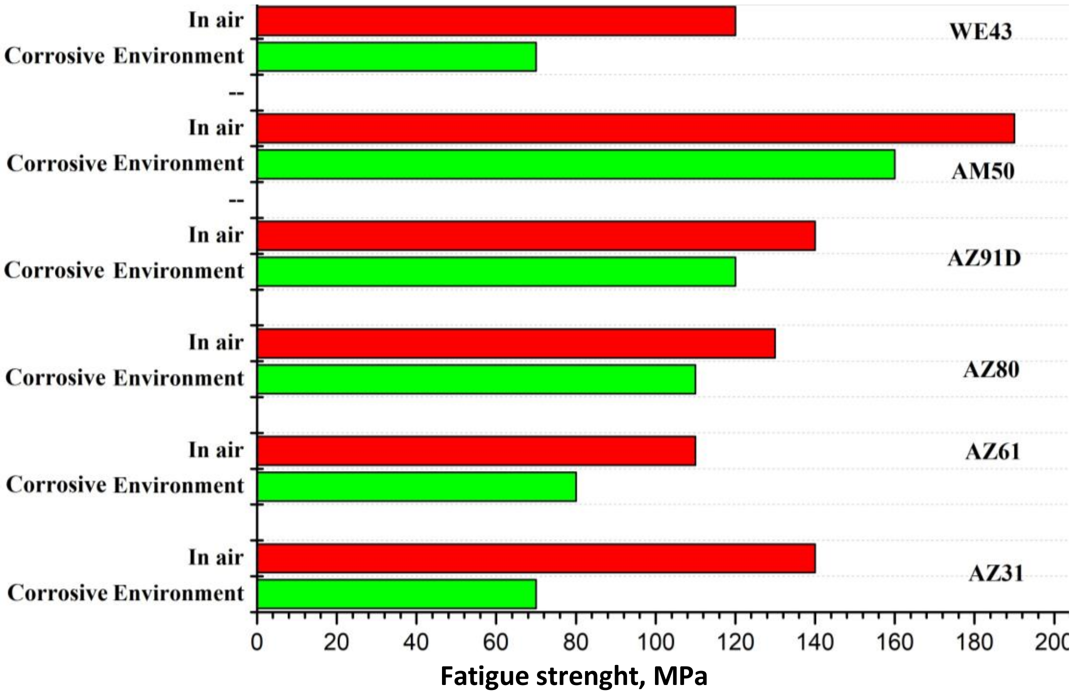

Figure 4. *Fatigue strength comparison of different Mg alloys tested for 107 cycles in air and corrosive environments* [30,42-46].

The most damaging effects are caused by aqueous solutions that include ions of chloride, phosphate, and nitrate. This is because these ions speed up the process of dissolving the corrosion layer of ($Mg(OH)_2$) [34]. For fatigue strengths in a select number of settings, however, the hydroxide layer's inherent instability—even under benign conditions is readily apparent. This is shown by the fact that certain environments have higher fatigue strengths than others (Figure 4). Corrosion has been shown to have a part in the beginning stages of fatigue cracking as well as their progression, which means that surroundings that are corrosive tend to have a negative impact on the fatigue life of a material.

The negative difference effect, magnesium can produce hydrogen even at anodic potentials [39]. In film surface may be caused by the regional growth of a less embedded integrated due to the undesirable microstructure of the underpinning alloy that cause anodic reaction (pitting), and/or the practices will help of the physical loading. Both of these factors can contribute to the deterioration of the exterior film's integrity [40,41]. It is commonly believed that a breach in the surface corrosion film provides a pathway for hydrogen to enter the matrix, and that a concentration of hydrogen in a specific area then accelerates the spread of the crack. The general public now agrees with this theory.

They further believe that anodic dissolution plays a secondary role in the accelerated production of fatigue fractures, with hydrogen embrittlement playing the primary role [47-49]. Figure 3 further demonstrates that fatigue crack occurring in AZ91D Mg alloy is due to mechano-chemical phenomenon. This is shown by the fact that fatigue cracking occurs at a faster rate. Under anodic charging conditions, there is a larger propensity for cracking, which may be explained by the evolution of hydrogen and the contemporaneous pitting. On the other hand, when the material was subjected to cathodic charging conditions, the fatigue life was significantly extended. This was owing to the fact that pit depths under cathode circumstances were insufficient to cause cracking at low stress amplitudes.

Ripples and striations emerge on the cracked surface as a result of crack propagation during the second stage of fatigue. Figure 5(a) displays a striated pattern that serves as an example. Each striation is the result of a single cycle of stress, and it indicates the location of a fracture front that is progressing [48,49].

A comparison of fracture propagation rates in various conditions may be accomplished by measuring the inter-striation space between cracks. Even though the production of striations is a telltale sign of stress, it is possible for some materials to fail due to fatigue even in the absence of striations [48,50]. Striation may be difficult to see if the surface that was cracked has corroded, and if there is debris left behind from the products of corrosion. Also, before attributing striations to CF, one should exercise care since comparable characteristics may be formed by SCC of magnesium alloys, as can be shown in Figure 5(a). This can be seen in Figure 5(b). The striation is the result of crack-tip sharpening and blunting that occurs repeatedly under fatigue loading. This is in contrast to the SCC characteristic of magnesium alloys, which can be entirely attributable to the hydrogen mechanism. It is possible that using fractography alone to differentiate between CF and SCC is not the most straightforward course of action to take.

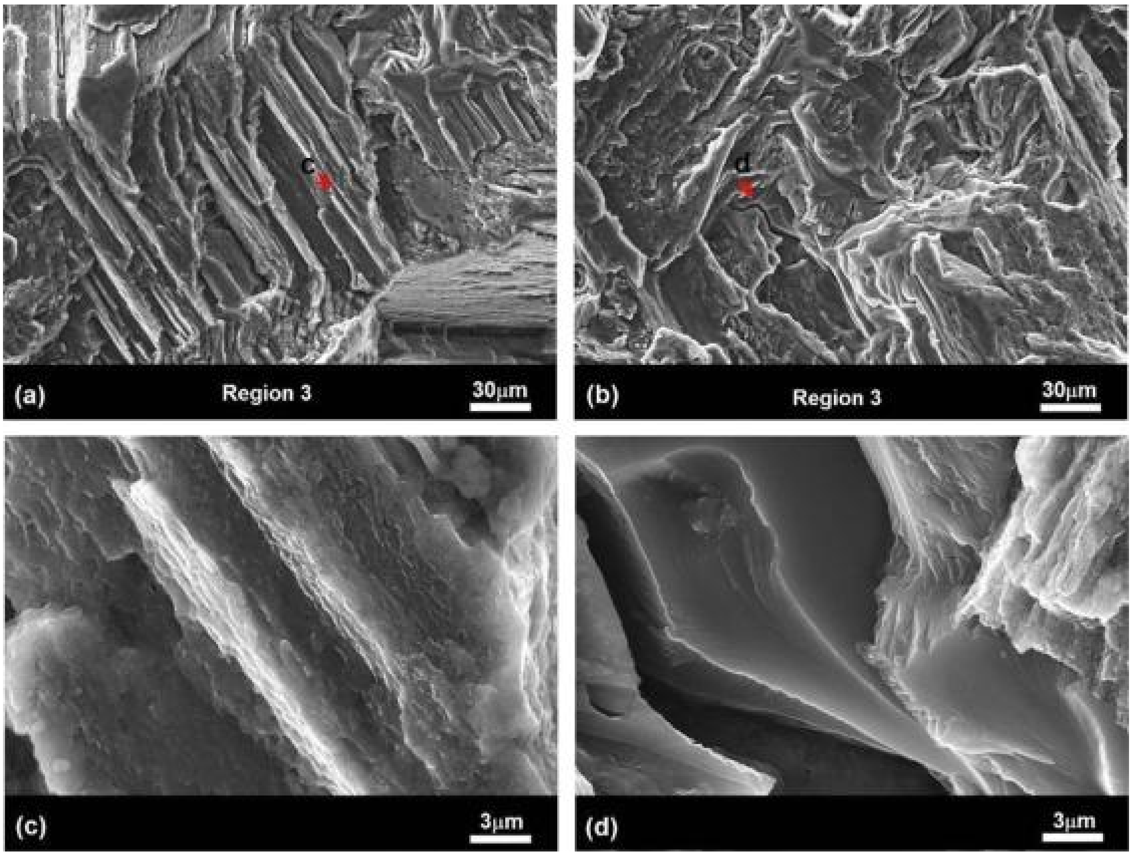

Figure 5. *(a) Fatigue striation and (b) parallel markings produced by stress corrosion cracking mechanism in AZ91 [36]. Permissions under CC NY 4.0 International Attribution*

HBF assisted cracking of Mg alloys

There are some studies on the critical temperature (CF) of magnesium alloys [51]. Since these investigations are restricted to alloys that include aluminum (Al), however, they are at best only vaguely applicable to applications involving body implants. In spite of the fact that it protects against corrosion and strengthens, aluminum is often believed to be poisonous to the human body [6]. Because of this, the Aluminum-Zinc alloys that include Magnesium (AZ-series) are often disregarded as potential candidates for use in biodegradable implant applications. These research used testing conditions that were vastly different from those encountered in vivo, including atypical frequencies and methods of loading, as well as artificial test environments and geometrically constrained samples [52]. This is another reason why the CF studies that have been reported are not very relevant.

Mode and frequency of loading

When placed in a real body setting, implants may be subjected to complicated state of stresses at the same time, which leads to failures such as twisting and bending. The components of the bending are also involved in the failure mechanism [53]. On the other hand, the CF tests that are most typically used are conducted under straightforward uniaxial loading. It has been reported in the literature that CF testing of Mg alloys [54-56] has usually carried out under load of fixed amplitude. Therefore, it is necessary to perform testing using waveforms that are more representtative of real-world settings in order to simulate the actual loading conditions.

Even in the loading conditions, corrosion frequency is the dominant factor which controls the rate of corrosion. It's possible that a less range will provide ample time for corrosion, which would in turn encourage a synergistic interaction between corrosion and mechanical stress, which will, in a nutshell, speed up the development of fatigue cracks [26]. On the other hand, this pattern may not always hold true. According to Rozali *et al.* [57] during the course of the investigation into the CF of AZ61 in NaCl solution, it was discovered that the influence of frequency was more apparent when the DK was low as shown in Table 1 [58,59]. In most cases, this frequency falls somewhere in the range of 1 to 3 Hz.

Table 1. *The noise figure for implants used in cardiac and bariatric practices [58,59].*

S. No.	Implants	Frequency, Hz	Activity
1	Orthopedic	1-3	Normal walking (vertical direction)
2	Orthopedic	0.5-1.5	Normal walking (lateral direction)
2	Cardiovascular	0.8-2	0.5-1.5 Normal heart

Chemistry of the corrosive environment

Although the effect of inorganic components on the deterioration of magnesium alloys has been studied and documented extensively [60-62]. However, the function of the organic components of real blood plasma, such as glycogen, organic molecules, and enzymes, is only recorded in a few studies. Yamamoto *et al.* [61] studied that bio sorption promotes calcium breakdown longer progressive, but proteins make dissolution more rapid. Xin *et al.* [62] indicated the creation of a bilayer as a consequence of the absorption of albumin on magnesium alloys, which further enhances the corrosion of the material. However, the effectiveness of this shield decreases gradually over the course of period [63]. However, there is no existing study on the potential function of endogenous substances in corrosion-assisted cracking phenomena like CF, hence this information is currently unavailable. As a result, it is of the highest importance to provide the appropriate state of the environment for conducting biological crack testing.

Physical form and symmetry of the sample

Sharp and smooth curves are often seen in the devices used in implant procedures. While CF and SCC cracks sometimes originate on clean metallic particles, pointed forms are frequently the sites where accelerated commencement occurs for the first time. The region of stress concentration may be found in fabricated components and devices like implants, for example, stents, screws, and plats (sharp contours). In addition, magnesium alloys are prone to pitting when exposed to chloride solutions, particularly HBF; pits are the most prevalent cause of corrosion fatigue (CF). This means that the CF and SCC data that are typically frequently used for layout causes need to be acquired but use the samples that have well before sharp deformation; to put it differently, it could be completely essential to use before the samples. To study a biomaterial's response to deformation, scientists use a variety of testing methods. These include tensile testing, compression testing, fatigue testing, and creep testing. Each of these tests measures different aspects of the material's properties.

Corrosion fatigue-stress corrosion cracking interaction

It's possible for localized corrosion, such pitting, to offer starting sites for fatigue cracks. The crack tip may also experience localized corrosion, which can further accelerate crack growth. Additionally, if the material is prone to SCC, the fracture propagation rate may be further accelerated, as stated in [3]. Thus, comprehending that CF and SCC affect the crack progression ($\mathrm{d}a/\mathrm{d}N$) in three circumstances is crucial. Figure 6 shows how such stress distribution ($K_{\max}$) and stress corrosion crack limit (K_{Iscc}) define these circumstances [64-66]. K_{Iscc} always exceeds $K_{\max}$ which measures fatigue. It is clear from looking at Figure 6(a) that the corrosive liquid lowered the threshold for the fatigue fracture to start propagating ($K_{\max}$). Even when $K_{\max}$ exceeds K_{Iscc}, the surroundings still cause fatigue cracks, rendering this realistic CF performance. When it comes to the second kind (shown in Figure 6(b), moderate fracture production levels have minimal environmental impact. The SCC framework reaches a peak; however, when K_{max} is greater than K_{Iscc}, a superposition of the SCC and CF mechanisms takes place. This suggests that the process is stress dependent. The third kind of behavior, seen in Figure 6(c), is a mix of time-dependent and externally induced stress mechanisms. Nevertheless, load ratio effects are not taken into consideration in this classification. Recent research [67] has suggested developing a more comprehensive version of this classification. Very little research has been done on the role that SCC plays in the progression of CF cracks in magnesium alloys.

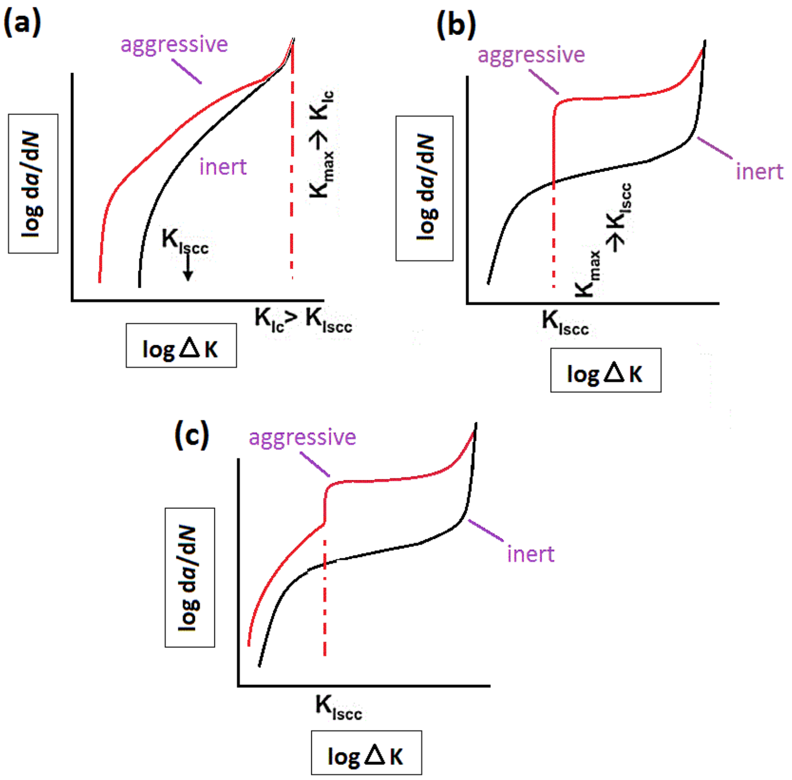

Figure 6. *The genesis of a fracture as depicted in a schematic under a variety of conditions, including concurrent cyclic and tensile loads, in addition to an innocuous and hostile environment.*

Reducing Mg alloy transplant body-fluid-assisted corrosive failure

CF and SCC originate from just a dynamic relationship among surface properties and enhanced localized stress. The prevention of such cracking requires the implementation of strategies that are suited to address each of these elements or any combination of them.

Procedure of composites making and structure of metals

This is necessary in order to use these alloys for implant applications. A few important alloying components are covered in this section. Aluminum, which offers resistance to corrosion in addition to strengthening, is the primary alloying element in the most prevalent types of magnesium alloys (*i.e.* AZ series). However, due to the fact that aluminium is known to be harmful to the human body, alloys of magnesium that include aluminium.

Nutrients are an essential component of skeletal tissue, particularly in bone fragments [68]. Ca is also necessary for the transmission of chemical signals across the human cytosystem [68]. Additionally, the incorporation of Ca into bone was discovered to be aided by the incorporation of Mg. Ca helps to reduce the grain size of magnesium alloys, which enhances both the mechanical characteristics and the corrosion resistance of the alloys. However, when present at a weight percentage of P 1 %, calcium causes the precipitation of magnesium calcium along grain boundaries, which results in embrittlement [68]. Zinc (Zn) in the amount of 15 mg per day is needed by the human body [69]. A solid solution of magnesium may be strengthened via alloying with zinc. However, if the percentage of P reaches 6.3 wt.%, Zinc will begin to form secondary phases, which will again result in brittlement.

Rare earth elements, often known as REs, are regarded not only be poisonous. [68], but they are also said to 'display anti-carcinogenic qualities [70]. This is the broad consensus among scientists. Mg alloys with an adequate amount of RE content may generate a surface film that is resistant to corrosion.

For example, the Elektron 31 alloy, which contains 2.3 % neodymium and 1.6 % gadolinium, produces a mixed oxide film. These intermetallics can cause localized corrosion as well as embrittlement. It has also been noted that adding a very little amount of RE may increase the fatigue strength [71].

The function of alloying additives in CF of magnesium alloys in SBF has received very little research attention. Refining the microstructure of Mg alloys is a potent method that may significantly improve the materials' resistance to corrosion as well as fatigue, in addition to the positive benefits that can be achieved by appropriate alloying [72]. In addition to this, there are claims of improvements in both tiredness and corrosion characteristics of magnesium alloys that were produced by these techniques [73].

Mechanical surface treatment

It is well knowledge that the application of compressive pressures may enhance fatigue strength and life. According to Zhang *et al.* [74], Shot peening compressed residual stresses increased AZ80 alloy fatigue life. Khan *et al.* [75] Shot blasting also strengthens AM60 alloy. Roller burnishing AZ80's clean surface and residual stress improved CF endurance in solvent [75]. However, physical imperfections, such as quick zones, formed by such procedures may make CF cracks easier to begin and progress. Biomaterials can be treated with various techniques to render them more biocompatible. This can include surface treatments such as chemical modification, deposition of thin films, and thermal treatments. Chemical modification involves treating the surface of the biomaterial with a chemical agent such as an acid or alkali. This can be done to alter the surface energy, which can make the material more hydrophilic or hydrophobic. It can also be used to create a functionalized surface that can interact with biological molecules such as proteins or cells. Thin film deposition involves depositing a thin protective layer of material onto the surface of the biomaterial. This can be done by different methods. This layer can be used to create a more inert surface that is resistant to corrosion and wear. Thermal treatments such as plasma etching, annealing, and laser ablation can also be used to modify the surface of the biomaterial. These treatments can be used to alter the surface roughness and morphology of the material, which can affect its biocompatibility.

Biocompatible surface chemical treatment

Biocompatible surface chemical treatments involve the use of a variety of chemicals and processes to modify the surface properties of a material to make it more compatible with living tissue. These treatments can include deposition of proteins, polymers, and other molecules, as well as modification of the surface energy of the material through etching, plasma treatments, or other methods. The ultimate goal of these treatments is to create a surface that is non-toxic, non-fouling, and promotes cell adhesion and proliferation. The best approach to avoiding corrosion and highly corrosive cracking is to apply a surface coating. This is because magnesium has a high chemical reactivity, making it one of the most reactive elements. According to Bhuiyan and colleagues [43] findings, many different coating techniques that are able to increase the fatigue life of magnesium alloys when exposed to corrosive conditions [76]. Surface engineering methods provides a safeguard against the corrosion, biodegradation, and erosion [77-93]. Various coating methods are employed to protect the surface of different biomedical materials [94-97]. Spray pyrolisis deposition [98], Electroplating [99], electrodeposition [100], hot dipping [98-104], physical vapor deposition (PVD) & chemical vapor deposition (CVD) [105-117], electrochemical vapor deposition (EVD) [118], cold spray [119], thermal spraying [120-125], sputtering [126], claddings [64-75], plasma spraying [79,80,88,91-93,127-150], and electrophoretic deposition are all methods used to deposit coatings onto metals (EPD). However, the coatings that were used in these trials were not always biocompatible since the focus of the coatings was on addressing the CF resistance in settings that were not biological [88]. It has been observed that such coatings not only increase the implant's biocompatibility but also improve its resistance to corrosion. Calcium orthophosphates were shown to increase the corrosion resistance and surface biocompatibility of magnesium-based metallic biomaterials, as reported by Dorozhkin [151]. There are several findings that demonstrate an increase in corrosion resistance as a result of electro-deposited Ca-P coatings on a variety of magnesium alloys [152].

It is possible that referring to a research by Srinivasan and colleagues [153] on the impact of silica-based plasma electrolytic oxidation (PEO) coating on the surface-corrosion-corrosion (SCC) of magnesium alloy in a chloride solution is not completely out of place. Although the PEO coating made the alloy more resistant to general corrosion and pitting corrosion, it did not make a significant difference in the alloy's resistance to SCC. When the AZ61 Mg alloy was tested in chloride solution, a recent research reveals that PEO coating resulted in about a 56 % drop in the fatigue strength of the material [154].

Conclusion and future prospective

In order for a material to be considered suitable for use as an implant, it must first satisfy a number of criteria, including those pertaining to its mechanical, electrochemical, and biological properties. Alloys made of magnesium provide a favorable balance of ductility and strength. Since magnesium is both biocompatible and biodegradable, it is a strong contender for use in applications involving temporary implants. When it comes to implants, however, the resistance to cracking might occur because of the synergistic action.

However, the existing studies are mostly inappropriate for body implants for the reasons listed: (a) the alloys involved in the earlier research typically include Al, which is known to be poisonous to humans; and (b) the lab testing variables adopted in the in vitro tests, such as the regularity and mode of loads, testing process, and specimen geometries, were distinct in the preclinical studies from those in the real in vivo execution of the tests. Both issues may be traced back to the fact that the alloys used in the earlier research were, for the most part, generic.

Recent research has unequivocally shown that separate corrosion films form in in vivo circumstances as opposed to the in vitro conditions that are often used, with in vivo corrosion rates being much lower than in vitro rates. Therefore, it is of the greatest priority to first ascertain the in vivo model, such as the solubility and maximum load spectral response, and then to conduct virtual crack experiments on the Mg alloys that have been primarily designed for use as body implantable devices under conditions that imitate the in vivo solubility and loading spectral range on the Mg alloys that have been developed specifically for use as body implants.

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