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Review

Computational biomechanical and biodegradation integrity assessment of Mg-based biomedical devices for cardiovascular and orthopedic applications: A review

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ABSTRACT

Magnesium and magnesium based alloys biomaterials are considered the most viable and pragmatic choice of selection as biomedical alternatives for tissue repair of damaged bones, vascular and other biological disorders. Considering the sensitivity and health implication of implanting foreign matter into the human system, extensive studies are required before clearance for clinical trials is given. This often requires rigorous experimental evaluations of biomechanical and biodegradation properties of such biomaterials, which are often expensive, time consuming and practically demanding, and requires several validation protocols. The technical advances in computing power as well as computational modeling techniques, such as finite element analysis, have been sort to address these limitations, but very little has been catalogued in reviews on the basis of their utilization and sundry issues. This paper attempts to fill this gap by assessing from literature, how effective the application of finite element analysis has been in the evaluation of the biomechanical and biodegradation behaviour of Mg-based biomedical systems which are considered for use as stents, screws, staples, and implants for orthopedic and cardiovascular applications. The review focuses on how factors such as geometric profile of the Mg based system, the material properties and applicable physical and constitutive laws, design parameters, boundary conditions stipulated, the analysis of modeling and simulation outcomes, and agreements with experimental data, influence the potentials for utilization for real life designs of Mg based biomaterial systems. Also, the challenges that could affect their effectiveness with directions on future studies for improvements, were covered. The review will be of immense benefit to researchers, biomaterial designers and clinicians.

1. Introduction

Musculoskeletal diseases and coronary heart diseases (CHD) rank among the most prevalent forms of human health diseases [1,2]. Coronary heart disease (CHD), has been reported to be the most deadly of all health issues in several advanced and developing

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nations in the world, and available statistics indicate that CHD health management is projected to rise by 21% by 2030, from expenditures estimated to be US$863 billion at 2010 [3,4]. Similarly, musculoskeletal disorders arising from injury, infection, disease, or abnormalities in skeletal development, are stated to be a source of disability for an estimated 1.71 billion people, globally [5,6]. Thus, there has been an age-long search for viable intervention strategies to help improve livelihood and global health concerns in these areas. The use of biomaterials to address some dimensions of these health issues has gathered attention in recent times.

Metallic biomaterials have been the most prominent biomaterials commercially developed for musculoskeletal and coronary heart disease management [7,8]. This is largely due to their generally good mechanical and biomedical characteristics [9,10]. In this regard, Ti and Ti alloys, cobalt-chromium alloys, stainless steels, and magnesium (Mg) and Mg alloys, have been the most utilized in biomedical applications [11]. However, the most significant drawback of most of these biometallic materials, excluding Mg, is their high elastic modulus, which causes stress shielding, and the proclivity to release ions that can be harmful to the host tissues and system [12]. Mg and its alloys and composites are however largely free of the aforementioned flaws, and hence, have found widespread use in orthopedics, tissue engineering, and regenerative medicine [13,14]. This is basically due to their exceptional mechanobiological properties such as excellent biodegradation with no deposition of toxic substances in the host, long-term biocompatibility with the host's physiological system, and more importantly, the close similarities between their mechanical properties and that of natural bone [15,16]. Despite these attractive spectral of biomedical properties, they are highly susceptible to corrosion attack when exposed to the host's body fluids [11,17]. In a quest to curb this corrosion-induced drawback and concurrently enhance the biomechanical properties of Mg alloys for cardiovascular and orthopedic applications, several kinds of research have been undertaken [16,18–20]. These researches encompass both experimental (in vitro and in vivo) and computational (*in silico*) studies focusing on innovative designs, development, characterization, and deployment of magnesium and its alloys for structural implant applications [11,15,18,21]. The present review, nevertheless, focuses on the *in silico* assessment of the biomechanical and biodegradation of Mg-based biomaterials, devices and implants. The motivation for the selected area of focus for this review lies in the fact that to date, there is no comprehensive literature on the computational biomechanical and biodegradation of Mg-based materials used in cardiovascular and orthopedic applications. The few reviews available on this subject matter are currently limited to experimental investigations on biomechanical properties, biocorrosion characteristics, alloy design, surface modification, coatings, and biological performance of Mg alloys [22–27]. This review work was thus undertaken in light of this observed gap in the literature on the state-of-the-art reviews on the application of computational methods in biomechanical and biodegradation assessment of Mg-based biomaterials. The next few paragraphs give a succinct account of the significance of computational methods in biomedical investigations.

Recently, there is rising interest among biomaterial scientists and biomedical engineers on the application of computational methods such as finite element analysis (FEA) for the assessment of biomaterials/implants properties before their physical development and clinical trials [21,28]. This can be linked to several factors such as high cost of experiments, perioperative and postoperative complications associated with some orthopedic surgery, unpredictable time-dependent integrity (in vivo) of the biomaterials/implants, the need to thoroughly understand the properties of the biomaterials/implants via parametric studies, and the non-destructive nature of computational assessment methods, among others [21,28]. In light of this, the application of computational methods in biomaterials developments has skyrocketed over the years [29]. Currently, computational methods such as continuum mechanics with finite element analysis, molecular dynamics simulation, computational chemistry, topology optimization, and machine learning have remarkably revolutionized the protocols of biomedical materials investigations [21,30]. The application of continuum mechanics with finite element analysis to the study of Mg-based biomaterials, devices, and implants is, however, the focus of this review paper. As a result, the next section provides some background on continuum mechanics to the readers.

2. Fundamentals of continuum mechanics and finite element analysis

Continuum mechanics is a field of mechanics devoted to the study of the mechanical behaviour of materials modelled on the consideration of the material as a continuous mass instead of discrete particles [31–33]. Fundamental physical laws such as the law of conservation of mass, the momentum conservation laws, and energy conservation are applied to develop models, expressed as differential equations, which describe the behaviour of such materials [34]. The equations which are essentially mathematical expressions of these laws, give rise to field equations, which hold true at all points and time on the continuum [33,35]. Apart from the applicability of these general principles to all continuum (solids, liquids, and gases), continuum mechanics also covers use of constitutive equations that define idealized materials [36–38].

Finite element analysis (FEA) is a popular technique for quantitatively addressing issues in continuum mechanics expressed as partial differential equations in two or three dimensions with two or three variables [37]. To address problems, the FEM divides a complicated system into smaller, basic components called finite elements [37]. This is accomplished by a specific space partitioning in the spatial coordinates, accomplished through the creation of a mesh that corresponds to the specified item: the solution's numerical domain, which is made up of a finite number of points. Thus, finite element formulation of boundary value problems generates a set of algebraic equations [39]. These equations are then integrated to form a larger set of equations that represents the entire scenario. The finite element approach then uses variational techniques derived from variational calculus to approximate a solution by eliminating an accompanying error function. The FEA method normally relies on computational processes for solving complicated engineering problems. Several commercial software programs for the finite element solution of complicated engineering problems, such as ANSYS, ABAQUS, COMSOL, DEFORM and NASTRAN, have been created since the approach is ideally adapted for the digital computer [21,37,40].

The finite element technique, in summary, consists of the following essential steps:

- Characterization of the geometric profile of the body as well as the material properties.
- The dissection of the body into nodes and elements (meshing).
- Assumption of a shape function to depict the elements' physical behaviour.
- Formation of equations to define an element.
- Grouping of the elements to explain the problem as a whole.
- Boundary conditions application.
- Instantaneous solution of a set of linear or nonlinear algebraic equations to produce nodal results
- Data visualization (post-processing of results) [41].

From available studies, there are substantial evidences, which show that FEA method is one of the most applied computational methods for studying biomechanical behaviour of materials [21,28,30]. The succeeding sections of this review has attempted to document the progress made so far in the use of FE protocols to understand and predict some selected properties and performance of Mg-based biomaterials and devices (MBBDs). Some of these properties are biomechanical/mechanical behaviour, biodegradation/biocorrosion, and damage evolution phenomena. The detailed accounts of this progress will be with emphasis on stents, screws, surgical staples and bone implants.

3. Prediction of the mechanical behaviour of Mg-based materials using finite element analysis (FEA)

The structural integrity of any biomaterial plays a vital role in its field-proven reliability assessment in terms of functionality, durability and life span [42]. The structural integrity of MBBDs has been numerically and experimentally studied. However, the scope of this section is limited to the numerical studies that have been undertaken using application-based implementation of FEA. It should be noted that the acronym "FEA" is interchangeably used with the term "*in silico*" herein.

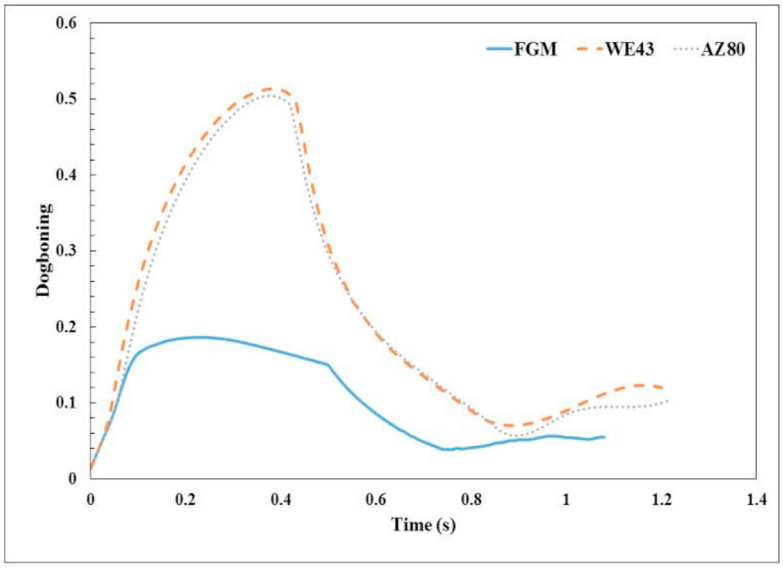
3.1. Mg-based stents

3.1.1. In silico biomechanical behaviour of Mg-based stents

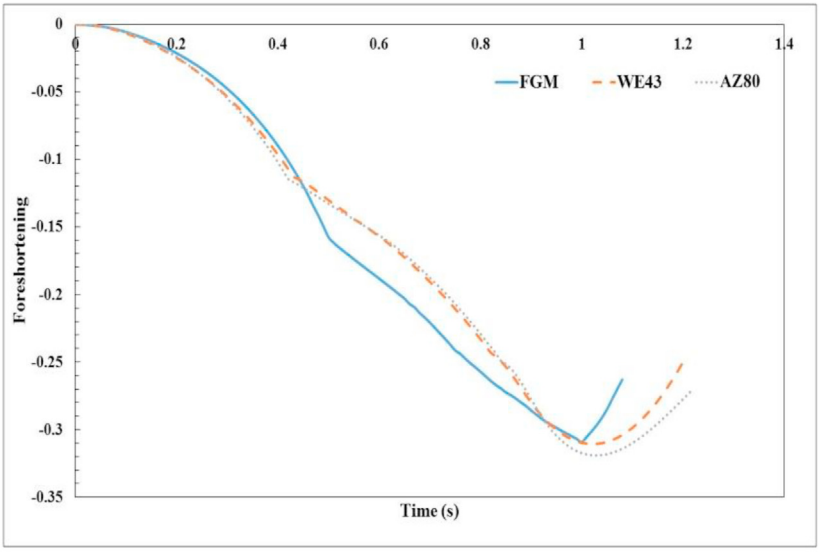
One area of biomedical research and cardiovascular medicine where the applications of MBBs is widespread is in the design, development as well as deployment of stent for the treatment and management of coronary artery diseases (CADs) [43,44]. In general, the stent provides mechanical support to the artery after percutaneous coronary interventions (PCIs), which in turn, aids in the resolution of post-PCI issues such as acute recoil and late negative remodeling [45,46]. On the one hand, MBBs are found suitable for stents' application, primarily due of their outstanding biocompatibility, bioabsorbability, and appreciable mechanical characteristics [43]. On the other hand, stents made of MBBs are susceptible to dog-boning and foreshortening when employed for long-term use which in turn affects the arterial walls adversely [47]. Thus, a comprehensive understanding of the mechanisms of deformation and degradation of these stents is essential. To understand these mechanisms, and concurrently improve the mechanical properties of stents before clinical trials, several researchers have adopted FEA based on continuum mechanics for their investigations [46–49]. In continuum approaches, partial differential equations solve the given problems, and thus they are more suitable for modeling bulk materials' behaviour [50]. On this note, when compared with discrete models, they are more desirable in terms of facile implementation in the FEA framework for the modeling and simulations of real-world stent performance. More significantly, they can make it easier to find some desired mechanical features of arteries, stents and their interactions, which can be difficult to detect using conventional approaches [46]. Finally, in the absence of many experiments, FEA offers a rapid and cost-efficient basis for assessing stent performance, providing detailed data on the structure's effective limits, expanded geometry, and stress-strain distribution profiles (SSDPs) within the stent for various loading states [51]. Interestingly, a large number of scholars have delved into this territory and reported promising results, which have the potential to provide solid foundations for future study in this area. Some selected studies and their findings are presented in the succeeding paragraphs.

Bahreinizad et al. [48] used a three-dimensional (3D) FEA model (FEAM) combined with numerical simulations of an idealized balloon expandable Palmaz-Schatz stent to forecast the mechanical behaviour of bioabsorbable Mg alloys stents during implantation. The bioabsorbable Mg alloy stents were developed using parabolic Functionally Graded Materials (FGM) (AZ80 50.40% and WE43 52.31%). They compared the FGM stents to two other uniform/homogeneous stents composed of WE43 and AZ80 Mg alloys. COMSOL Multiphysics FE software was adopted to model the mechanical behaviour of the stents, and constitutive elastoplastic law and large plastic strain theory in the COMSOL environment was used to simulate the bioabsorbability of the stents. About 3845 s order tetrahedral elements were used to mesh the entire geometry. Additionally, the bound optimization by quadratic approximation (BOBYQA) technique was used to discover the optimal heterogeneous index for the FGM, which can reduce the stent's maximal dogboning. Because the Palmaz-Schatz stent is symmetric, 1/24 of the total geometry was employed in this investigation. The stent's length, thickness, and unexpanded inner radius, respectively, were 9.86 mm, 0.05 mm, and 0.70 mm. From the FEA results, not only was the maximal dogboning of the FGM stent significantly lower than that of the stents constructed of homogeneous materials (as shown in Fig. 1(a)), its dogboning throughout the stenting procedure was also significantly reduced. This position is supported by the FGM stent's more uniform deformation when compared to homogeneous stents, as seen in Fig. 2(a–c). The FGM stent, however, had a comparable foreshortening to the other two stents, as seen in Fig. 1(b). The elastic characteristics of AZ80 and WE43, the two magnesium alloys employed in this work, are similar, however their plastic mechanical behaviour was noted to be different. The reason behind the FGM stents (composed of these two Mg-based alloys) optimal performance is because of the stent's high plastic strains. Large plastic straining is a property required of viable stents as it must have capacity to expand from a restricted blood vessel's small diameter to its own full diameter and maintain its shape after the balloon is withdrawn. This underscores the importance of the stent's plastic mechanical behaviour. The authors suggested that the models' accuracy might be enhanced if alternative geometrical models (apart from the Palmaz-Schatz geometry) were evaluated and material deterioration was factored into the simulation procedure.

One common problem of stents is their inhomogeneous deformation-induced fracture during their deployment and expansion process (DEP) which can provoke damage of the blood vessel and aggravates patient suffering. In an attempt to thoroughly understand the deformation mechanism and fracture-to-failure behaviour of WE43 Mg Alloy Stents (MAS) during the deployment and expansion process, Yoshihara et al. [52] developed a finite element analysis method coupled with a ductile failure criterion to evaluate and predict the aforestated phenomena of the WE43 MAS in ANSYS/LS-DYNA environment. The alloy was modelled as an elastoplastic material which satisfies the n-power law with the constitutive equation ($\sigma = C\varepsilon^n$), where C is the plastic modulus, ε is the strain and n is the work-hardening exponent. The stent thickness was set at 0.10, 0.15, and 0.20 mm, while the breadth was 0.10, 0.15, and 0.20 mm. During the initial step, the inside pressure was linearly raised (0–0.8s). Pmax was established as the highest pressure at which the plaque's inner diameter is 3.0 mm. The frictional coefficient in between the stent and the plaque, as well as between the plaque and the artery, was adjusted at 0.1. In addition, the influence of the stent's thickness and width on the stress and strain behaviour was simulation, in order to determine the best stent design for optimal performance. It was shown numerically, that the inner diameter, width and thickness of the stent have great influence on its material deformation behaviour and fracture during the expansion process. In comparison with experimental results, there was a significant deviation in terms of maximum



(a)



(b)

Fig. 1. A comparative plot of (a): variation of the dogboning among different stents versus the simulation time, and (b) variation of the foreshortening among different stents versus the simulation time (Bahreinizad et al. [48], culled with permission from Elsevier).

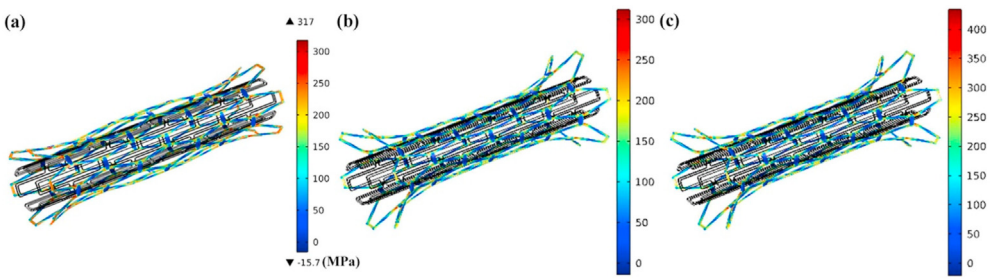


Fig. 2. The (a) FGM stent showed more uniform deformation compared to the (b) WE43 or (c) AZ80 magnesium alloy stents (Bahreinizad et al. [48], culled with permission from Elsevier).

pressure at fracture for the stent (attributed to more surface roughness of the experimental stent due to manufacturing conditions) and strong correlation between both results in terms of shape expansion. Noteworthy is the fact that the study established that combining stent shape optimization with FEA-predicted findings can be utilized to design and manufacture MAS that are reliable and efficient.

The deformation and delamination behaviour of coatings used on biodegradable Mg alloy cardiovascular stents, was investigated by Chen et al. [28]. They used a cohesive approach based on FEA to forecast the adhesive integrity between the coating and the stent during the deployment and crimping process in the study. Three-dimension (3D) FEAM was used to model the deformation mechanism and stress distributions of the coatings while also forecasting and assessing the coatings' durability as well as delamination susceptibility. The modeling approach was based on peeling model-based cohesive zone approach, which can reliably mimic the adhesive behaviour between two surfaces like stents and coatings. The stents were made of Magnesium alloys AZ31 and Mg-Nd-Zn-Zr, while the coatings were polymeric - made of poly(L-lactic acid) PLLA and poly(D,L-lactic-co-glycolic acid) PLGA. Time (s), Inner Surface (expansion) Diameter (mm), Outer Surface (crimping) Diameter (mm), and Contact state were the boundary conditions for the stent-coating deformation. The stents having an outside diameter of 1.3 mm were crimped on the fold-balloon. The coated stents with an outer diameter of 3.2 mm in air were then expanded by inflating the balloon to a pressure of 8 atm. Eight-node brick components with reduced integration (C3D8R) were used to mesh the coating and stent, with 10 and 2 layers in the stent and coating thickness directions, respectively. Eight node 3-dimensional cohesive elements (COH3D8) with an average maximum edge length of 15 μm were used to mesh the cohesive layer. The balloon was also meshed with four nodes surface element that had a lower integration (SFM3DR). The experimental findings were utilized to simulate and validate the FEA scenarios, and there was strong correlation between numerical and experimental results. The verified simulation showed the robustness, correctness, and compatibility of the proposed CZM framework. Furthermore, the numerical framework provided by the study, was useful in determining the most essential material characteristics and standards for coating and stent design, as well as comprehending deformation mechanism of both coatings and stent struts, and predicting delamination behaviour. However, there were some observations which future studies should resolve. Firstly, in the 3D modeling, the zero-thickness cohesive components are sensitive to mass scaling. The goal time increment in this study is 2×10^{-6} s - a greater target time increment might result in an unstable deterioration process of the cohesive components, implying that the computing time of 3D simulation is significantly longer than in 2D space. Secondly, is the balloon's surface is reduced to a cylindrical surface. This will result in increased friction on the coated surface, particularly in the circumferential directions. Thirdly, the validation test was conducted in-vitro without liquid, implying that the polymer coating property will remain stable as in dry circumstances. However, when a stent is implanted, the material properties of PLGA and PLLA alter during immersion in blood, and the hydration that ensues, reduces the interface strength between the coating and the stent.

In summary, the evaluated articles in this section demonstrate that the application of FEA methods could be used to study and understand how design parameters such as stent thickness and breadth influence the integrity of stents. It was possible to establish from FEA evaluation that for elasto-plastic considerations, plastic material properties exerted a greater influence on the mechanical behaviour of the Mg-based stents than their elastic characteristics. Furthermore, FEA methids could be applied to forecast delamination and deformation behaviour of coatings used in Mg based stents.

3.1.2. In-silico biodegradation/corrosion of magnesium-based stents

Computational modeling techniques, such as Finite-Element-Analysis, have received scientific publicity as very valuable tool in creation, optimizing design, and studying the material behaviour of various biomedical implants [53]. In stent applications, the active reaction of artery tissue within a certain time frame following stent placement, that is, neointimal remodeling, has not received much attention in the field of computational modeling. The phenomenon of neointimal remodeling in degradable stents is remarkably interesting, because both the stent degradation and neointimal remodeling can take place simultaneously, allowing for mechanical interaction and load transfer between the deteriorating stent and the remodeling artery to occur. A few studies reviewed in the succeeding paragraphs in this section, highlights how the use of FEA can be applied to study such phenomena in Mg based stents.

Boland et al. [46] developed a mechanobiological FE model to study Mg alloy AZ31 degradation and neointimal remodeling simultaneously in stent deployment. This was motivated by the need to address research questions about the applicability of computational methods to neointimal remodeling (depiction of the dynamic reaction of arterial tissue in the immediate aftermath of stent placement). In the study, a mechanobiological model for neointimal development was launched in EXPLICIT/ABAQUS software, incorporating two distinct Mg corrosion models that were used in tandem (pitting and uniform corrosion models). The choice of the software was informed by its better contact enforcement technique. For the contacting surfaces of the artery stent and neointima, general contact in ABAQUS/EXPLICIT was utilized, with “hard” contact specified for normal contact behaviour and a frictional coefficient of 0.1 presumed for tangential contact behaviour. The three-layer artery (magnesium stent (Le = 0.0173 mm), characteristic element length (Le = 0.073 mm), and ghost mesh (Le = 0.0173 mm) were all represented using eight-node linear reduced integrations brick elements with distortion control. The mesh density was concentrated on the region of interest, which in this case was the stent and ghost mesh, which have 125,000 and 1,400,000 elements, respectively; while the arterial mesh has 68,000. Three factors (material degradation rate kU, Weibull function parameter, c, as well as growth rate parameter, b) were examined in the pitting corrosion model. These parameters were calibrated, and the model's performance was verified using data from separate corrosion and loading studies on Mg AZ31. The proposed framework was found useful in predicting the dependence of biodegradable metallic stents' long-term and short-term mechanical and corrosion behaviour, which could serve as a useful guide in stent design. The gap in the study is the application of isotropic material models to represent the three-layer artery. It is well known that arterial walls are heterogeneous and have anisotropic characteristics. An anisotropic material's model would have provided a more realistic depiction of the three-layer artery, perhaps affecting the model's conclusions. Also, the stent corrosion rate is constant in the uniform and pitting corrosion model, and it is unaffected by arterial remodeling around the stent struts. However, in vitro corrosion studies on pure magnesium samples have revealed that covered samples degrade at a slower pace than uncovered ones, indicating that using a constant stent corrosion rate might not simulate the practical expectations reliably.

Gao et al. [21] acknowledged insufficient scaffolding time due to intense corrosion process, as the major issue with magnesium alloy stents (MAS). In their research, the role of multi-dimensional corrosion on the biomechanical integrity of MAS structures was

investigated using finite element technique. A corrosion model was created phenomenologically for this purpose, while immersion tests performed on Mg alloy (AZ31B) cubes with varied number of exposed surfaces, was used to investigate size changes in the cubes. In the Finite Element simulation, every element's corrosion was considered to be a hybrid of uniform micro galvanic corrosion and stress-corrosion processes. The stent model comprised two ring designs, an 80-m strut width, a 150-m strut thickness, and a 1.5-mm outer diameter. Based on mesh convergence research, it was meshed using 48,992 linear-reduced brick elements (C3D8R) with characteristic lengths of 20 μm. An arterial vessel emerged from the stent. It had a 4 mm outside diameter and a 2.2 mm interior diameter. In the vascular wall, an idealized atherosclerotic plaque shape with a thickness of 0.3 mm was discovered. C3D8R elements were used to mesh arterial and plaque models. Surface components were meshed into the rigid-cylinder within the stent. The friction factor of all tangential contacts was 0.2. The stent was made of AZ31B alloy, a material that has a modulus of 44 GPa, density of 1.7 g/cm^3, as well as a Poisson's ratio of 0.35. A von-Mises plasticity model with isotropic hardening was utilized during stent expansion, and it was replaced by the corrosion model once the expansion was completed. For the artery and plaque, an isotropic hyperelastic material with a sixth-order reduced polynomial strain energy density functions was utilized. The tissue densities (plaque and artery vessel) were 1.1 g/cm^3 on average. The normalized time, t, was utilized, which was set to 0 at the start of the deterioration process and t equaled 1, when the stent totally lost its scaffolding to the vessel. It was discovered that the cube corrosion rates scaled proportionally with the number of exposed surfaces, particularly when pitting corrosions are not visible. Under uniform or multi-dimensional corrosion, damage rates were proportional to the number of exposed surfaces. The application of FE simulation with the recommended model offered detailed information on mechanical behaviour and morphological changes within the scaffolding time which were in agreement with the experimental data.

According to Shen et al. [54] magnesium alloys have shown tremendous promise for usage in medical devices, because of their good biocompatibility and biodegradability. However, quantitative assessment of the degrading behaviour of Mg alloys with a view to regulate and meet performance demands for practical purposes has been largely unaddressed. A diffusion theory based numerical model that can capture the degrading Mg alloys' behaviour, was proposed in the research. ABAQUS′ ALE adaptive meshing and user subroutine UMESH-MOTION were used to convert the numerical model into a finite element scheme and calibrated using corrosion data of rolled Mg-3Ge and as cast Mg−1Ca binary alloys. In order to demonstrate the model's reliability, the calibrated model was used to analyze the degradation behaviour of a pin implant.

The task of numerically addressing the corrosion problem is complicated by the fact that the solid-liquid contact moves as corrosion progresses, making it a moving boundary problem. The moving boundary challenge was addressed in this study using the ABAQUS finite element software's Arbitrary Lagrangian-Eluerian (ALE) adaptive meshing capabilities. The corrosion behaviour of as-cast Mg−1Ca alloy as well as as-rolled Mg-3Ge alloy implants in vivo was predicted using a Finite-Element model. This model uses 3-D linear coupled temperature-displacement reduced integration brick elements (C3D8RT) to mesh all of its components. Because diffusion equations have the same form as heat conduction equations, the temperature degree of freedom is employed to represent the user-defined parameter (C), the concentration. As a boundary condition, the concentration of solid components in the model is fixed to an invariable value. Only one-eighth of whole geometry was modelled when symmetry is taken into account. Two cylinders make up the corrosion environment. One is 50 mm in length and 12.5 mm in diameter, while the other one is 50 mm in length and 11 mm in diameter. SBF solution is utilized as the corrosion medium for Mg−1Ca alloy while Hanks solution is utilized for Mg-3Ge alloy to complement the calibrating process. Fig. 3 shows the variations in Mg ion concentration in the corrosive medium at 0 h, 50 h, 100 h, 150 h, 200 h, and 250 h after corrosion. Each of the pictures in Fig. 3(a)–(f) is made up of two sections. The 3-D representation of the Mg−1Ca alloy pins in the SBF environment is the core component, while the cross-section of the pin is depicted in the included image at bottom left. The varied colors represent different Mg ion concentrations, and the non-corroded pin is shown by the red area. The steady diffusion of ions into the environment and the isotropic nature of along the cross-section can be easily discerned. In addition, as the 3-D image shows, corrosion has a significant effect in the end. It means that diffusion is inhomogeneous along the axial direction. In Fig. 4(a), it can be seen that the cross-section dimension decreases roughly equally in all directions, whereas the curvatures on both endpoints dramatically changes over time. Fig. 4(b) has the same features as Fig. 4(a), with the exception that the implant's surface is rough. The appearance of a rough surface is due to the fact that the physiological environment of an animal's true body is complex. Diffusion corrosion, galvanic corrosion, stress corrosion, and other mechanisms have all occurred on the implant. The overall profile, however, remains the same as the simulated result. The simulated profiles are clearly homogeneous in cross section, but heterogeneous in the axial direction. The approach was validated by comparing with a large amount of data from in vitro degradation studies on Mg alloys, and found to correctly capture the degrading behaviour of the alloys.

Debusschere et al. [49] reported that in cardiovascular angioplasty, the use of bioresorbable stents is a relatively new area of interest. Long-term adverse reactions of non-degradable stent, such as late stent thrombosis, fatigue induced strut fracture, as well as in-stent restenosis can be avoided as long as the transient mechanical properties of bioresorbable stents are understood. An implicit finite element solution was used in this study to predict corrosion of biodegradable magnesium stents. They were able to merge two models created for use with an explicit solver into a material user subroutine UMAT that was compatible with the implicit FE solver Abaqus/standard. An isotropic plasticity model served as the foundation for the framework. The stent was made up of ten zigzag hoops that are linked laterally by thin connections. The stent has a nominal outside diameter of 3 mm and an overall length of 11 mm when fully inflated. The stent's struts are 0.1 mm wide and thick, with rounded edges. 200500 hexahedral elements with a mean element size of 0.02 mm make up the final mesh geometry. Likewise, 5 distinct mesh geometries of stenotic blood arteries were constructed to serve as simulation input. The blood vessel geometries are 20 mm in length, 2 mm in global inner diameter, 0.3 mm in thickness, as well as 1.2 mm in central stenotic diameter. Each vessel has a distinct curvature, ranging from 0 to 90°. The blood arteries were given incompressible hyperelastic material characteristics (neo-Hookean material) for the purpose of simplicity, with the shear modulus set to result in a 5% arterial distension for a 40 mmHg (5 kPa) rise in inner pressure. Between the stent and the blood artery, a surface to surface contact with a friction coefficient of 0.1 was used. It was discovered that the appropriate amount of time steps required to completely simulate as well as to validate the total deterioration time during corrosion process, output stresses and deformation can be reduced with the use of the implicit FE solution approach. Even for more complicated vessel-stent configurations, the developed finite element approach was adaptable to analyze localized corrosion of stent struts.

Summarily, the FEA models used in this section of the review effectively capture the degradation mechanisms and give further

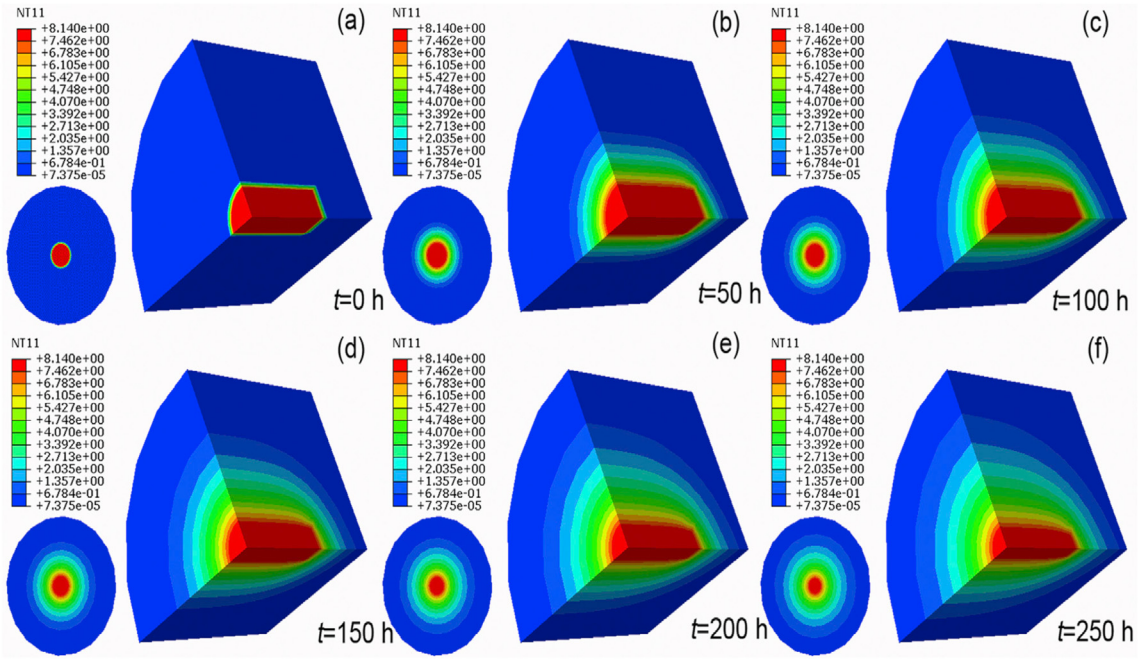


Fig. 3. Contour plots of the corrosion environment concentration at different time: (a) 0, (b) 50 h, (c) 100 h, (d) 150 h, (e) 200 h, (f) 250 h. The main picture in each figure is 3D view of the Mg−1Ca alloy pin, the inserted picture is cross section of the Mg−1Ca alloy pin (Shen et al. [54], culled with permission from Elsevier).

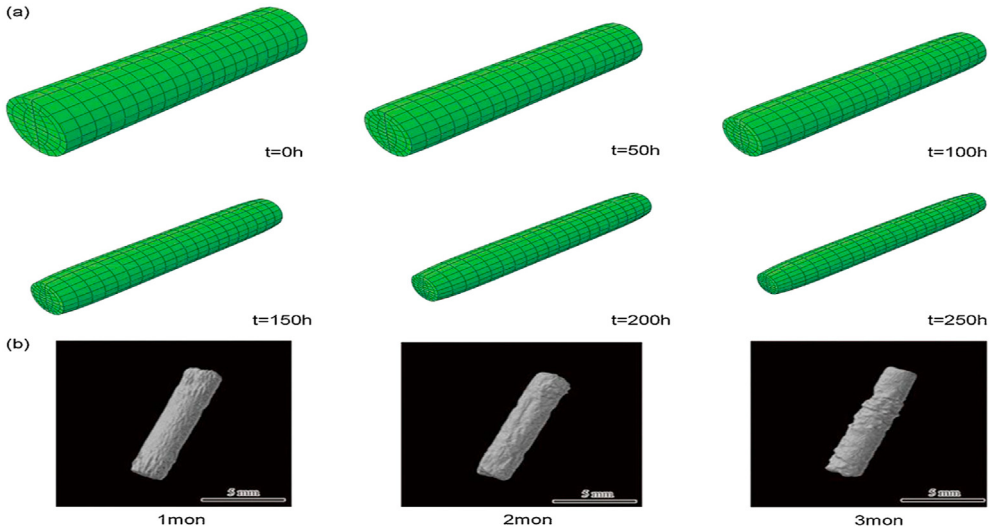


Fig. 4. Morphology feature of the Mg−1Ca alloy pin over time: (a) simulation, (b) experiment (Shen et al. [54], culled with permission from Elsevier).

insights on its influence on the mechanical performance of biodegradable Mg-based implants. The findings suggest that improving the pitting corrosion resistance of these materials through the development of improved magnesium alloys, manufacturing techniques, innovative designs and surface treatments has the potential to greatly improve the development of functional biodegradable magnesium alloy-based implant.

3.2. Mg-based screws

It is acknowledged that internal fixation failure mechanisms such as gradual screw loosening can be caused by stress shielding [55]; a condition in which a biomedical screw absorbs greater stress levels inside the screw-bone design, which results in loosening. However, the closeness of Mg and bone elastic moduli can potentially lower screw loosening by allowing optimal stress transmission between the screw and bone, thereby aiding bone remodeling around the screw.

Tetteh et al. [55] used a 2-D FE model to simulate the influence of thread profile on stress transmission in a Mg fixation. Stress transmission across three thread profiles was estimated using modified stress parameters from a previous study. An axial pull-out test of an interference screw in an artificial bone was simulated using three 2-D axisymmetric Finite-Element models. Each screw featured a 5.45 mm head diameter, a 15 mm stem length, and a 2 mm thread pitch. Each tip was tapered at an angle of 65.1° to the horizontal, and all screws were entirely cannulated with a 2.54-diameter circular hole. For the sake of simplification, the

materials in the model were assumed to be linear elastic and isotropic, allowing the model to concentrate on the influence of the thread design. Yield Strength (MPa), Elastic Modulus (GPa), as well as Poisson Ratio (for Magnesium and for Cancellous bone) are material characteristics utilized in the finite element model. This simulation utilized a final element size of 0.039 mm, which corresponded to 129622 nodes as well as 64415 elements. The result showed that the trapezoidal-shaped Mg screw thread demonstrated the most stress transmission. The findings also emphasized the relevance of thread profile design in improving the stress transmission capabilities of Mg-based screws. Trapezoidal thread profiles, in particular, were noted could help improve stress transmission. Hence, variations in screw thread parameters such as the proximal and distal half angles, for example, could have substantial influence on the thread profile and, as a result, the thread's stress transmission capabilities.

The following were noted as some of the limitations of the approach adopted in the study: Firstly, the 2-D methodology limits the results' applicability to real-world 3-D screw-bone direct interaction. The authors claimed, however, that stress transfer parameters (STP) from the 2-D method may be used in 3-D screw-bone interactions based on the work of [56]. Second, this model assumed that the cancellous bone had homogenous and isotropic material characteristics, as well as a completely bonded contact between the screw and the bone rather than friction. The overall contact surface between bone and screw is limited because cancellous bone is made up of a complex network of trabeculae having anisotropic material characteristics. However, the authors claimed that hitting bone during screw penetration causes bony fragments to break off from the trabecular network, filling gaps between bone and screw and so reducing the impact of open pores [57]. The use of a completely bonded contact surface derives from the fact that the bone-screw interface must be firm, with no micromotions, to prevent aseptic loosening as well as implant failure shortly after implantation [58]. Finally, the simulated pull-out test was not physically validated. The model, on the other hand, created a stress transmission pattern that fits with physical pull-out test results obtained from Ref. [59], who found greater stresses in the apical threads, comparable to regions of high stress concentration in the first thread.

Wang et al. [13] studied the use of device in form of a ring as mechanical support to assist Anterior Cruciate Ligaments (ACL) repair. The aim of the investigation was to determine the amount of load and stress in the ACL and its femoral insertion site after ring repair. A 3D FE model of a goat stifle joint was created to determine the load and stress levels in the Anterior Cruciate Ligaments as well as its insertion point in the femur. FEBio 2.0 was used to define the mechanical characteristics of the joint tissues. The properties of bone (Young's modulus = 17 GPa and Poisson's ratio v = 0.33) as well as cartilage (Young's modulus = 5 MPa and Poisson's ratio v = 0.46) were assumed to be linear isotropic elastic. They also assumed Menisci to be orthotropic-elastic tissues. Strain energy functions were used to classify ligaments as isotropic hyperelastic. The ACL as well as Posterior Cruciate Ligament (PCL) were defined using the Veronda-Westmann function, whereas the Medial Collateral Ligament MCL as well as the Lateral Collateral Ligament LCL were defined using the Mooney-Rivlin material model. The LCL coefficients were presumed to be the same as the MCL coefficients. The Magnesium rings were specified as isotropic and elastic (Young's modulus to be 42,000 MPa and Poisson's ratio to be 0.3). The model was meshed with four-node tetrahedral element. The ligaments had a 0.4 mm element size, while the other joint tissues had a 0.8 mm element size (Fig. 5). There were 261,882 elements in total for the complete model. The results showed that, when compared to the model with a defective ACL, ring repair resulted in 50% reduction in anterior-posterior tibial translation. Following ring reconstruction, the ACL's in-situ force was recovered to around 60% of that of the intact Anterior Cruciate Ligaments. The stress distribution as observed by comparing Fig. 6a and b, show that they are almost the same, except for the additional stress concentrations around the vicinities of the sutured areas. At the femoral insertion point, the maximum Von Mises stress attained 71% of the intact ACL stress, resulting in improved healing. The data collected on loading level in the rehabilitated Anterior Cruciate Ligaments and its insertion spot that resulted in regeneration were found to be potentially useful for building innovative devices to help humans mend their ACLs, mechanically. There are however some observations noted which could have influence on the results. Firstly, the goat's stifle joint is not anatomically or biomechanically equivalent to the human knee. The authors, though argued that goat was used based on the findings from literature, which indicated that anterior tibial loads at joint flexion angle of 90° were nearly identical to those in the human ACL [60]. Second, during model construction, the goat stifle joint was simplified. Muscles were not included in the design, however, to investigate the functioning of the ACL, a typical clinical testing approach was mimicked, in which the knee joint was moved passively and muscles had minimal influence on joint kinematics. Furthermore, in order to simplify the model and

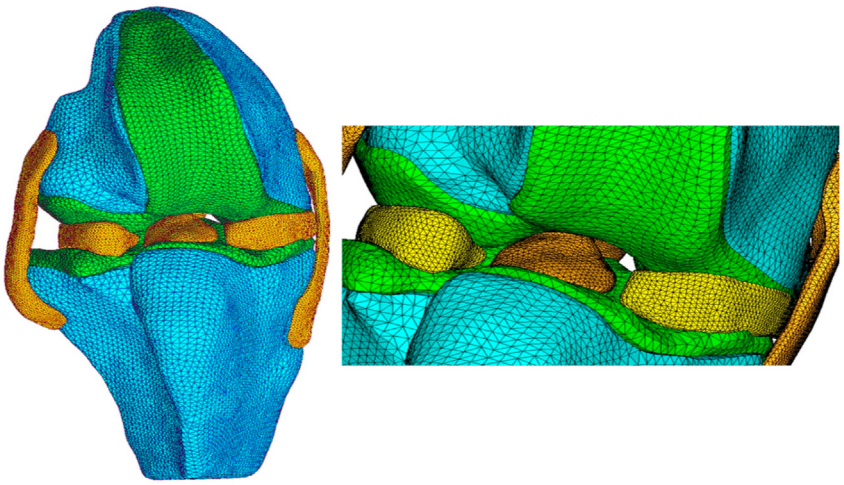


Fig. 5. Meshed model of goat stifle joint (Wang et al. [13], culled from Elsevier).

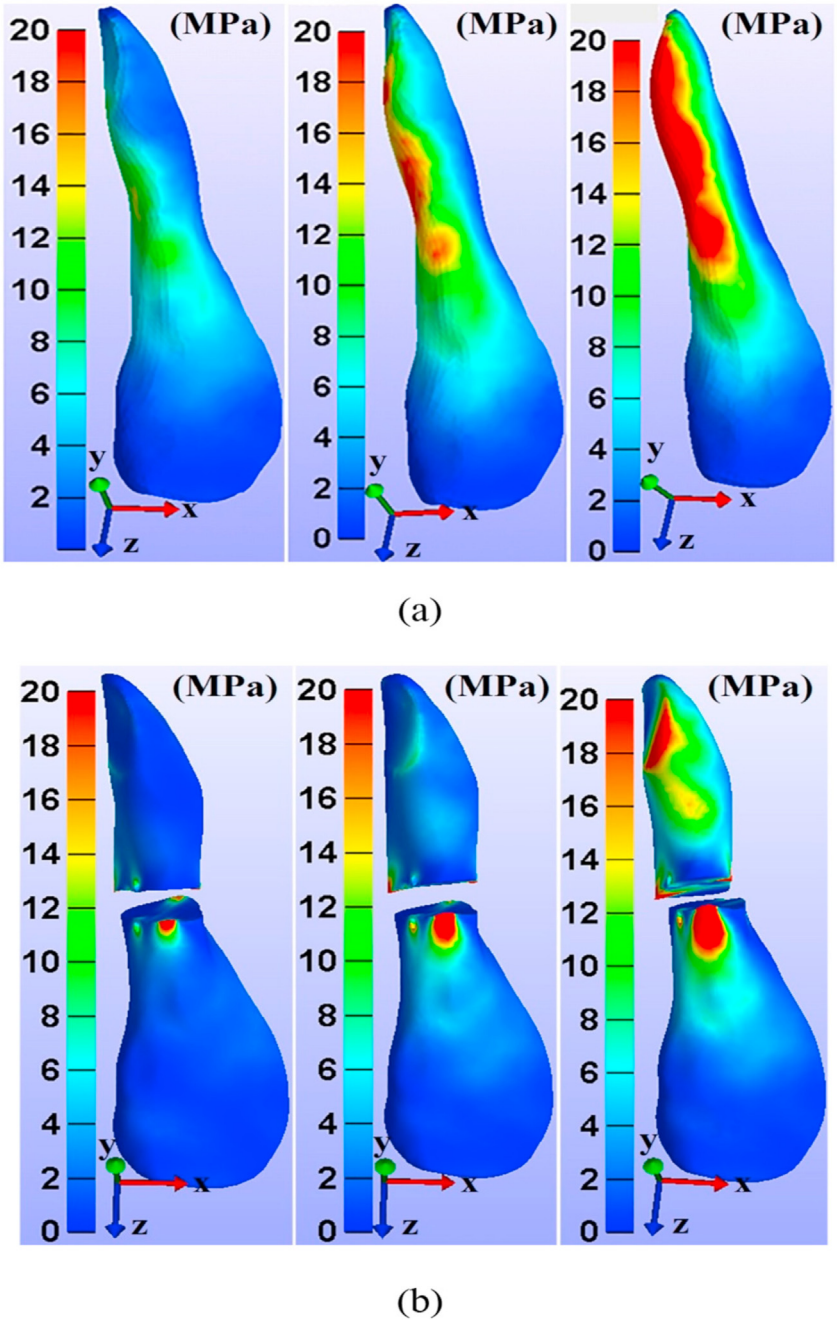


Fig. 6. Stress distribution on the ACL under a 67 N anterior tibial load at joint flexion angles of 37°, 60° and 90° with (a) ACL intact (b) after ring repair. Axis z shows the anterior (+) posterior (−) direction, axis y shows the proximal (+) distal (−) direction and the axis x shows the medial (+) lateral (−) direction in the knee joint (Wang et al. [13], culled from Elsevier).

promote model convergence, friction between tissues was omitted. Also, the loading condition utilized in this investigation was relatively basic: a 67 N Anterior-Posterior Tibial Load (APTL), with no internal or valgus tibial moments, primarily because the goal was to improve model reliability in forecasting A-P movements and ACL load. Finally, the mechanical characteristics of the ligaments in the goat models were derived using published data derived from human tissues.

Lee et al. [61] investigated the use of a Mg–Ca–Zn alloy as a fixing material in maxillofacial surgery. Using finite element analysis simulation, this study examined three distinct materials – Mg–Ca–Zn alloy, PLLA, and titanium – for the reliable fixation of Sagittal Split Ramus Osteotomy (SSRO) in the rehabilitation of mandibular prognathism, and assessed the viability of the Mg–Ca–Zn alloy for use in the fixation of SSRO in the rehabilitation of mandibular prognathism. Because information on organic components were not available, the model was made up of 354,309 nodes and 203,642 elements and presumed to be homogeneous, isotropic, and linear elastic. Young's modulus (MPa) and Poisson's ratio are mechanical characteristics of the mandible and screws utilized. To mimic actual stress distribution and eliminate the impact of bone contact, a 0.5 mm gap was established between the proximal and distal portions. Screw sizes in titanium and magnesium were estimated to be 2.0 mm in diameters, and 2.5 mm in

biodegradable polymer. All screws were lengthened to 14 mm in order to be bicortically positioned in the mandible. When stress distribution across the cortical bone is compared between the polymer and Mg alloys groups (Figs. 7 and 8), it seems that the Mg alloys screws can handle more loads, resulting in less stress being transferred to other biological parts like the cortical ramus and condyle of the mandible. Irrespective of the material employed, the screw exhibited little deformation (< 0.21 mm after surgery, during the mandible's initial functional load) as a function of functional load. Hence, the degradable Mg alloys screws may withstand higher load and reduce the negative impact of sagittal cleft ramus osteotomy setbacks surgery on stability. The triangular design was shown to be the most beneficial screw location in the investigation, leading to the lowest amount of stress on the cortical bone. Summarily, the absorbable Mg alloy screw can withstand higher stress and has a less negative impact on SSRO setback surgical stability.

One drawback of this study's FEA simulation is that the number of variables used to assess various circumstances complicates the model, making interpretation of the simulation outcomes challenging. According to the authors, FEA modeling of SSRO setback is a well-established model based on literature data. Although, the SSRO setback has solid bone contacts and the parameters to consider are straightforward. If the limits of FEA simulation are to be addressed, additional orthognathic techniques should be investigated. Many physicians favor the plate fixation device in SSRO, however bicortical screw fixations are also utilized. Future research should incorporate the simulation models for plate

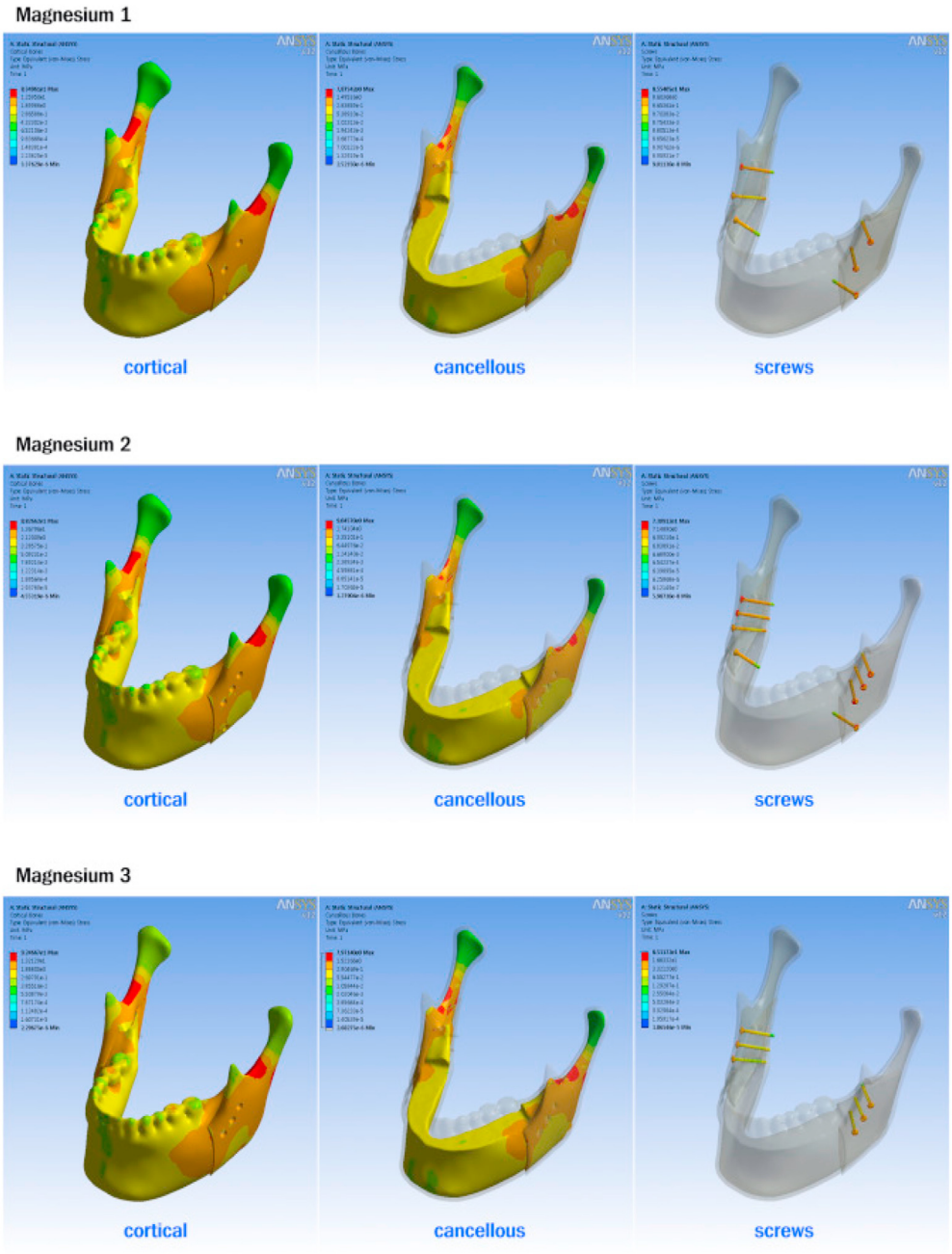


Fig. 7. The distribution of stress in magnesium alloy screws according to different screw configurations: 1 –triangular, 2 – reinforced triangular, 3 – linear ((Lee et al. [61], culled with permission from Elsevier).

fixation. In this work, bicortical screw fixation was chosen for SSRO simulation because it does not need sophisticated modeling, as opposed to plate fixation, which requires numerous variables and is more difficult to simulate reliably.

3.3. Mg-based staples

Owing to their biodegradability and biocompatibility, magnesium-based biomaterials have been ranked among the most viable biomaterials. Magnesium ions from magnesium alloy degradation have been reported to enhance the release of transforming growth factor beta 1 (TGF-β1) in the intestinal tract, which raises the biosynthesis of cytokines such as basic fibroblast growth factor (b-FGF) and a signal protein known as vascular endothelial growth factor (VEGF), that boost angiogenesis as well as wound healing [62]. Medical devices made of magnesium have been clinically utilized as vascular stents as well as orthopedic screws. However, due to the lack of ductility of Mg-based alloys, their use as staples is restricted. It has also been reported that due to their susceptibility to stress corrosion, Mg staples have been observed to readily fracture or deteriorate in the corners of the B-shape after stapling. Another source of worry is the generation of hydrogen gas as a result of Mg corrosion in physiological settings.

To address the challenges outlined above, Amano et al. [63] presented a new staple made of magnesium (Mg) alloy that is biodegradable and detailed the safety and feasibility tests that were performed to establish its suitability. The study assessed the theoretical deformation characteristics of Magnesium alloy staples using FEA modeling performed on MSC MARC2017 simulation code. For both the traditional U–shaped design in clinical use and the suggested design, FEA was used to evaluate the theoretical deformation behaviour of the Magnesium alloy staples during stapling and anastomosing using the MSC MARC2017 simulation code. The first model replicated stapling deformation from a U-shape to a B-shape by striking the staple head at 1 mm/s and determining the distribution of equivalent plastic strain and stress. By pushing the legs horizontally by less than 0.1 mm, the second model was used to simulate the deformation of the B-shaped staples anastomosing the gut. To decrease the concentration of tension produced by stapling as well as anastomosing, the staples were developed with radial curvature of greater than 0.4 mm. The staple had a diameter of 0.25 mm, a breadth of 3 mm, a height of 2.5 mm, a weight of 0.6 mg, and a surface area of 0.06378 cm^2. The distribution of comparable elastic strain and stress, as well as plastic strain, was obtained with the aid of the simulation. They had an elastic–plastic body with a Young's modulus of 45 GPa, a yield stress of 0 = 260 MPa, a Poisson's ratio of v = 0.3, and a coefficient of friction of 0.3. The suggested staple has an appropriate stress distribution when stapling and preserving closure, based on finite element analysis result. In vitro analysis shows a good biodegradability, mechanical stability as well as bio-compatibility in an immersion test utilizing synthetic intestinal juice. Only slight amounts of hydrogen arising from fast Mg corrosion were seen during the immersion's first week, and the majority of staples retained their

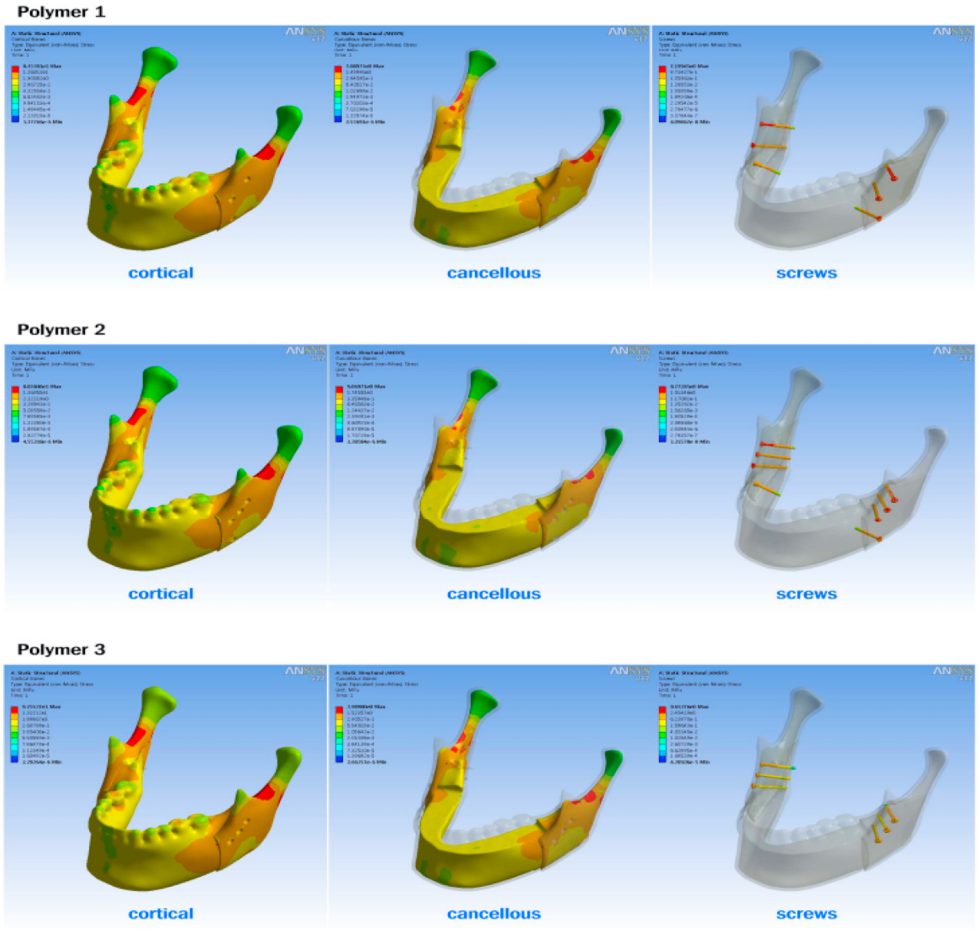


Fig. 8. The distribution of stress in polymer screws according to different screw configurations: 1 – triangular, 2 – reinforced triangular, 3 – linear (Lee et al. [61], culled with permission from Elsevier).

forms at least for the first four weeks. In addition, the tensile load was sustained for a period of time beyond one week. After four weeks, it is roughly one-half of its original size. Furthermore, the cytotoxic limit for magnesium was not exceeded in the intestinal synthetic juice. The magnesium alloy staples did not induce technical failure or problems such as anastomotic leakage, hemorrhage, or adhesion in swine intestinal anastomoses. Histopathologically, there was no evidence of necrosis or a severe inflammatory response. As a result, the suggested Magnesium alloy staple appears to be a potential replacement for Titanium alloy staples. The authors used artificial intestinal juice to simulate the biological environment, but its composition may have differed from that found in real-life situations, which could be a downside. The anastomosis location, such as the stomach, small intestine, or large intestine, should have different exposure.

Wu et al. [64] stated that in gastrointestinal anastomosis, surgical staples produced from pure titanium and titanium alloys are commonly utilized. However, the titanium staple, are not bioabsorbed by the human body, causing abnormalities as observed on computed tomography (CT) and other imaging investigation, as well as, the danger of erroneous diagnosis. In the study, biodegradable staples made from high purity Mg for gastric anastomosis was investigated. U-shape staples were designed with two different interior angles, of 90 ◦ (original) and 100 ◦ (modified). When the original staple approached B-shape, residual stress was centered on the arc section, whereas it was centered on the feet for the modified staple following closure, based on the FEA results. In vitro studies revealed that the original staple's arc initially broke after seven days of immersion, but the modified one remained intact, indicating that residual stress had a significant impact on the HP-Mg staples' corrosion rate. The modified Mg staples were implanted in vivo with acceptable biocompatibility and no inflammatory response nine weeks after surgery. During the deterioration, the Mg staples maintained good closure to the anastomosis. There was no leakage or bleeding, and the staples showed no breakage or significant corrosion fractures. The modified HP-Mg surgical staple demonstrated sufficient closure strength and adequate biocompatibility in gastric anastomosis, based on the in vivo implantations. In the in-vitro experiment, modified simulated body fluid (m-SBF) was employed, whereas in-vivo was done by gastric anastomosis. The authors did not provide enough detail on the finite element analysis which formed the basis of the data generation and results interpretations.

In summary, the reviewed articles in this section show that Mg alloy staples are a viable alternative to titanium alloy staples for gastric anastomosis. The results show that Mg alloy staples with innovative design configurations based on FEA may satisfy the criteria of intestinal anastomosis both in-vitro and in-vivo based on the uniform corrosion behaviour and excellent biodegradation they exhibited. The redesigned surgical staples had enough closure strength as well as excellent biocompatibility in stomach anastomosis, according to in-vivo implantation.

3.4. Mg-based bone implants

In medical research, one approach to addressing the challenges of fractures is to substitute the bones with artificial implants. Metallic materials used in bone implants must be durable, corrosion-resistant, be able to sustain large loads, and have good biodegradability [25,65]. Mg alloys, with their excellent biodegradable characteristics, have found prominence as ideal materials for bones replacement or repair. This is supported by the fact that Mg occurs naturally within the body and is found in bones [62]. It is present in extracellular fluid ranging from 0.7 to 1.06 mmol/L when blood serum is at normal level; also, it is the second most abundant element in intracellular ions as well as the fourth most abundant cation in the human body [66].

Andoko et al. [67] assessed the mechanical behaviour of a Mg alloy hip joint implant utilizing experimental as well as Finite Element Analysis (FEA). With time and load changes, the simulation technique was created to assess deformation in total, maximum shear stress and maximum main stress. FEA simulations were performed using ANSYS 18.1 software. Walking, leaping, and moving down a staircase for 0–4.5 s were used to conduct the simulation. A mesh of 8333 nodes and 4534 elements is produced by the implant design. Density (kg/mm^3), Poisson's ratio, Young's modulus (MPa), Bulk modulus (MPa), and Shear modulus (MPa) are the material properties used (Mg–Zn–Ca). The hip-joint replacement had a cumulative distortion of 0.38 nm, 0.82 nm, and 0.90 nm in strolling, in leaping, and in descending the stairwell respectively, based on finite element analysis findings. Strolling, leaping, and descending the stairwell produced maximum shear stresses of 125.98 MPa, 264 MPa, and 291 MPa, respectively. Walking exercise produced 192.76 MPa of maximum main stress, leaping produced 397.48 MPa, and descending the stairwell produced 438.85 MPa. The simulation findings revealed that the action of walking down the stairs caused the most deformation (0.90 mm), while walking activities caused the least deformation, which was 0.39 mm. Descending the stairs produced the highest main principal voltage, based on the maximum principal stress assessment. Walking exercise produced the least principal voltage. The going down the stairs action maximum shear stress was the greatest, based on the investigation while the walking exercise had the least amount of maximum shear stress. The authors however, did not provide enough details on the finite element model procedures adopted in the study.

Garttzke et al. [68] emphasized the importance of understanding mechanical behaviour of implants during the degrading process in general and at distinct sites, as it guides design and selection of suitable implants. In their study, FE approach was used to construct a simulated deterioration model of Mg-based bone replacement materials. The simulation model is based on the premise that deterioration is a diffusion-controlled process fueled by Mg dissolution. Using the level set method (LSM) and finite element modeling (FEM), a three-dimensional bio-degradation model for complicated implant structures composed of magnesium alloys is created. The model is the first effort at simulating the creation of a degradation layer surrounding a non-degraded core in an open -pored three - dimensional structure. The level set technique is used to model the interface movement between the not-deteriorated and degraded materials. A numerical model largely dependent on the diffusion equation is used to simulate the deterioration. The diffusion coefficient in alloy ($1\cdot 10^{-9}$), Magnesium concentration in alloy ($6.478\cdot 10^{-5}$), diffusion coefficient in environment ($1\cdot 10^{-8}$), magnesium concentration in environment ($8.5\cdot 10^{-10}$), Young's modulus of alloy (45), mass transfer coefficient ($1\cdot 10^{-8} - 1\cdot 10^{-6}$), and Poisson's ratio (0.35) are all material parameters utilized in this study. The friction coefficient was assumed to be 0.1. From the FE simulation, it was possible to establish that the effect of environment is more pronounced during degradation and the structure loses stiffness more slowly. Furthermore, the nonuniform degradation noted could be linked to the accumulation of Mg in the puss of the structure. As observed in Figs. 9 and 10 for the in-vitro and in-vivo experiments, respectively; the SEM images show a nonuniform degradation pattern at some locations, which was opined to occur due to heterogeneity in microstructure from manufacturing variation, and environmental influence such as bone accumulation. The degradation features are noted to be consistent with the predictions from the model. Thus, the model may be used to

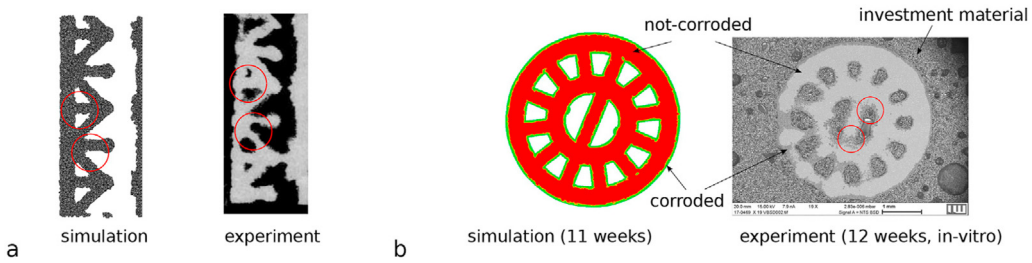


Fig. 9. In-vitro experimental results and FE simulations for the effect of corrosion on the implant structure. (a) without corrosion layer. (b) with corrosion layer ([68] culled with permission from Elsevier).

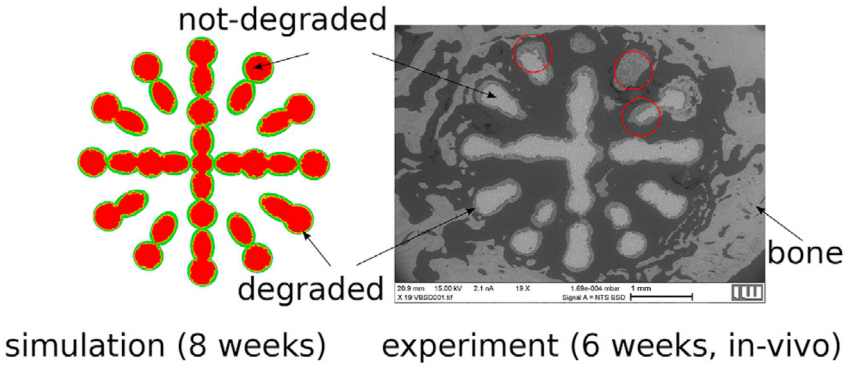


Fig. 10. In-vivo experimental results and FE simulations for the effect of degradation on the implant structure ([68], culled with permission from Elsevier).

anticipate how the mechanical behaviour of implants evolves during its deterioration phase will in use.

Amerinatanzi et al. [69] created a model that can predict the in vitro corrosion rate of magnesium-based alloys while taking pitting corrosion into consideration. To achieve this purpose, a FORTRAN users material subroutine (or VUMAT) was built which is coherent with the Abaqus/Explicit Finite Element (FE) solver. A continuum damage mechanism (CDM) Finite Element model was built using the established subroutine to phenomenologically predict the pace at which the corrosion of biodegradable Magnesium–Zinc–Calcium alloy occurred. Furthermore, Magnesium–Zinc–Calcium samples were immersed into a glassy reactor containing SBF solution (simulated bodily fluid) at 37° Celsius and at a pH value of 7.40 to assess mass loss over time. The corrosion Finite Element model parameters (Beta (β), Gamma (Υ), Psi (Ψ), as well as the kinetic parameter (K_u)) were then calibrated using RSM (Response Surface Methodology). The best recorded readings for Υ, K_u, β, and Ψ were reported to be 2.74898, 0.1005, 5.1, and 2.60477, respectively. Lastly, the numerical framework was shown to properly depict the impact of corrosion on mass loss over time, based on the strong agreement between the FE projections and the experimental results.

The customized FORTRAN user material subroutine (VUMAT) code, took into account the influence of the Mg–Zn–Ca alloy's pitting corrosion damage parameters (Dp). As a result, rather than taking into consideration the micro or nanoscale physical processes that drive deterioration, the simulation was predicated on the alloy's overall macroscopic behaviour. They used all four key parameters (i.e., γ, ψ, β, and K_u) to calibrate the model rather than the three most commonly reported in the literature. These four parameters are β (the dimensionless scaling parameter controlling pitting growth acceleration), γ (the dimensionless distribution shape factors characterizing the probability density function (PDF)), ψ (the dimensionless distribution shape factors characterizing the probability density function (PDF)) and K_u (the kinetic parameter representing the uniform corrosion process). To acquire these parameters, the data of mass loss over time of coupons with a diameter of 15 mm and a thickness of 3 mm in a simulated bodily fluid must be gathered. The coupons were constructed using an eight-node linear brick (C3D8) in Hypermesh, which depicts three (3D), solid, hexagonal, and deformable element types. Lastly, instead of using an arbitrary tuning strategy, they used response surface methodology (RMS) to tune and calibrate the four key parameters. It is important to note that the model created in this study has a number of drawbacks. Its predictions are particular to a given alloy because of its phenomenological foundation, which means it does not physically represent the influence of electrochemical processes on corrosion behaviour. As a result, the effective degradation parameters for other biodegradable Mg-alloys must be recalibrated. The model also can't be utilized to forecast the impacts of precipitate formation and grain size on corrosion because the predictions aren't based on the alloy microstructures. Furthermore, if the alloy is implanted within the body system, the model does not account for the impact of tissues coverage on alloy corrosion.

4. Challenges and future scope

Finite element analysis has been established as a viable tool for gaining a better knowledge of the biomechanical characteristics and biodegradation behaviour of magnesium-based implant devices. The comparison of Finite-Element-models to in-vivo and in-vitro strain demonstrates that given well-characterized material characteristics and functional load, Finite Element Analysis can reproduce stress-strain patterns, direction, and amplitude. However, it should be emphasized that there is still a significant issue with lack of interoperability from CAD to FEA, CT to CAD, CT to FEA, and even CAD to CAD data transfer. Multiple file formats, provide significant challenges to effective and accurate Finite Element

model development. As a result, improved scan data transmission to FE-models is required.

The design of a multi-physical and multi-scale modeling framework is crucial for future researches. A multi-physical framework, that is, a collection of multiple degradation models, may be more appropriate and beneficial to the modeling of magnesium based implantable devices. The mechanical, structural, and chemical characteristics of magnesium based implants would alter over time as they degrade. Thus, in the case of modeling of magnesium based medical implants, a single model cannot capture several features at the same time.

It is commonly recognized that soft tissues, in certain situations, operate as critical tension members to promote kinesis within the skeleton, and so play an important biomechanics function. As a result, modeling the motion of soft tissues, either geometrically or via finite element analysis that may behave as springs, will become more significant. Bone has linear elastic properties under typical loading conditions. Soft tissues exhibit nonlinear elasticity, necessitating sophisticated material qualities and model conditions. Existing studies have almost entirely concentrated on the implant component, with little regard for the influence of the adjoining soft tissues on biomechanical function and degradation [70]. Future research should explore the accommodation of the biomechanical behaviour of such soft tissues in model development. Also, laboratory studies that do not include tissues were also shown to provide less accurate results than computational projections [71]. As a result, when using laboratory data to verify a model, care must be taken. Furthermore, more robust solid models with enhanced capacity to modify CAD elements would enable additional study in this field. The capacity to readily construct models with slight differences and rectify minor faulty geometries will considerably minimize the time necessary to simulate various biomechanical scenarios.

It is expected that as the process of magnesium-based implant's degradation advances, the magnesium-based implant will be substituted by the newly formed adjacent tissue. Hence, both the implant device and the newly developed tissue will subsequently share the mechanical stress. Obviously, a large load will have a detrimental influence on tissue development if the magnesium implant degrades too fast, and the degradation products will also affect the growth of the tissue [72]. Combining the models of magnesium-based implant degradation with the models of biological tissue development, and taking into account biological parameters to replicate physiological scenarios in vivo, might be one way to accomplish harmonization. By having a better understanding and taking into account of how magnesium implants interact with biological tissue could be of immense benefit in designing magnesium-based implants, as it has been observed that local concentrations of Mg ions, hydrogen, and other elements, may play important roles in biological tissue function.

Additional considerations such as surface modification and alloying should be taken into account in future computational modeling, which would be a promising path given that pure Mg-based implants are not ideal for substantial number of medical uses. Controlling the deterioration rate of magnesium has necessitated the development of new techniques such as alloying [73,74] and surface treatment like coating [75] as well as surface nanocrystallization [76]. Most material damage models, on the other hand, are calibrated based on selected experimental data, with an assumption that the magnesium is pure. A model that could take into consideration the degradation modification approaches would be extremely valuable in evaluating numerically the performance of various Mg-based systems in order to improve future design and development.

5. Conclusions

This paper has presented a comprehensive review on the application of finite element analysis in biomechanical and biodegradation appraisal of Mg based biomaterial systems, which are intended for use as stents, screws, staples, and implants for orthopedic and cardiovascular applications. The need, purpose, effectiveness and potentials for practical implementation of computational modeling and simulation techniques as not only a tool for assessing material behaviour but also a guide for optimal biomaterial product design was demonstrated using biomechanical and biodegradation studies carried out using finite element analysis. From the reviewed articles, it was established that factors such as geometric profile of the Mg based system, the material composition and properties, assumptions and deployed physical and constitutive laws, design parameters, stipulated boundary conditions, basis for analyzing modeling and simulation result outcomes, and nature of validation process, influenced the degree of confidence and agreement between predicted and experimental results. The outcomes were seen to largely predict the biomechanical behaviour and biodegradation rate and mechanisms of the Mg based biomaterial systems. Also, the challenges militating against their effectiveness were highlighted with directions for future development proffered. Overall, the use of outcomes from FEA based biomechanical and biodegradation studies of Mg based systems, is concluded to have huge potentials for reliable practical applications beneficial to biomaterial designers and experts, and clinicians.

Conflicts of interest

The authors declare that there is no conflicts of interest. The authors acknowledge the funding supports received from their institutions.

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