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Mg-based biodegradable alloys for orthopedic implants - A review

ANSU J KAILATH* and ARVIND SINHA
Advanced Materials and Processing Division CSIR-National Metallurgical Laboratory Jamshedpur 831007

Abstract : Traditional metallic biomaterials for orthopedic implants require materials exhibiting excellent corrosion resistance in human body. Recently, implants made of biodegradable metallic materials are thought to be potential for orthopedic implant applications as they can circumvent revision surgeries. These implants are considered as the third generation implants as they are expected to provide adequate mechanical strength to support the bone during restoration; have excellent in vivo biocompatibility and controlled degradation rate. These implants would degrade within the body after completing its mission without leaving any residues within the body. Biocompatible elements like Mg, Fe and Zn and their alloys have been considered for bone implant applications due to their biodegradability. Amongst these, Mg alloys are preferred due its high specific strength, low elastic modulus that are close to human bone and low density minimizing the risk of stress shielding. However, Mg alloys have fast degradation rate in biological fluids leading to the release of hydrogen which would lead to premature failure of implants. This paper reviews the research efforts towards the development of Mg-based biodegradable alloys for orthopedic implants. Concepts followed in designing Mg-based biodegradable alloys, the mechanical properties of developed Mg-based alloys, degradation mechanism of Mg-based alloys and the efforts to reduce the degradation rate, and status of Mg-based alloys in orthopedic applications are compiled along with the existing problems and future research directions.

Keywords: *Biodegradable alloys, Orthopedic implants, Mg-based alloys*

INTRODUCTION

Demand of orthopedic implants are dramatically increasing because of the increasing aging population and also due to the rise in accidents and injuries[1,2]. Some of the orthopedic procedures require permanent implants which need to stay within the body as they replace a body part; whereas most of the orthopedic procedures require only temporary implants to support the bone during healing and needs to be removed once the healing completes[3,4]. Partial functional substitutes of natural bones have been done by permanent implants made of titanium alloys, stainless steel and cobalt-chromium alloys[5]. However, when these alloys are used as temporary support during healing, they need to be removed through a second surgery after the patient recovers. The revision surgery would certainly burden the patients with additional expenditure and pain. Therefore, use of biodegradable implants as temporary implants can avoid revision surgeries. These implants would degrade within the body and are replaced gradually by the growing tissues[6,7]. The materials suitable for these applications should have similar or slightly higher elastic modulus, strength and hardness as compared to human bone. These materials should not induce any sort of irritation, inflammation, and toxicity within the body; on the other hand, they should be biocompatible and promote cell growth (osteogenesis)[8,9]. The degradation products produced from these materials should be non-toxic and capable of entering into the metabolic activities of the body. Biodegradable polymers viz. polylactic acid (PLA), polyglyclid acid (PGA), biodegradable bioglasses, bioceramics and biodegradable metallic materials have been developed as the third generation implant materials[10]. Biopolymers have excellent biocompatibility but their mechanical strength is not sufficient for load bearing applications[11,12]. Metallic biomaterials have better mechanical properties as compared to bioglasses and biocermaics.

Recent research efforts have been focused on developing magnesium, iron and zinc based metallic biodegradable materials[12]. Mg alloys are easy to machine and have low density, high specific strength and comparable modulus of elasticity with human bone [13,14]. Therefore, Mg-based alloys are preferred over Fe-based and Zn-based alloys for implant applications. Mg is a biocompatible element when implanted within the body which degrades to

*Corresponding Author E-mail: ansu@nmlindia.org

produce an oxide which is soluble and non-toxic. Mg^{2+} ions are one of the most abundant cations present in human body and is involved in many metabolic reactions. However, Mg being a group IIA element, is highly active and form porous oxide film on its surface with the evolution of hydrogen gas[3,15]. These films fall off in solutions containing chlorine ions thus accelerating degradation.

Efforts on the development of Mg-based biodegradable alloys for orthopedic implant applications are reviewed in this paper. Alloy design strategy, mechanical properties of the developed alloys, degradation mechanism and the methods to lower the degradation rate are reviewed. Present status of Mg-based biodegradable alloys in orthopedic applications and the unresolved problems are discussed along with the future research directions.

BIODEGRADABLE METALLIC MATERIALS

Though the medical field has been using biodegradable polymer sutures for many years, the concept of metallic implants that degrades can be considered as a novel concept. This has broken the conventional paradigm that metallic biomaterials for implant applications should be corrosion resistant[16]. Biodegradable metallic materials should have the following characteristics; (i) it should be made of metals, (ii) it should provide adequate mechanical support during the healing process, (iii) the mechanical properties of the implant material should be comparable with human bone or slightly better, (iv) it should corrode within the body by releasing corrosion products which are non-toxic, (v) it should not leave any implant residues. This restricts the biodegradable metallic materials to be made of (at least) an essential metallic element that can be metabolized by the human body[17]. The mechanical properties of conventional metallic biomaterials, Mg-based biodegradable alloys and human cortical bone are compiled in Table-1 along with their advantages and disadvantages. Last few decades have witnessed active research on the development of biodegradable Mg alloys because of the similarity in their mechanical properties with human bone. However, their applications are constrained due to the high degradation rate with hydrogen evolution. An overview of the research efforts on developing Mg-based biodegradable alloys are given in the following sections.

Mg-BASED BIODEGRADABLE ALLOYS

Mg alloys have been well investigated and established for automobile and aerospace applications; however, these alloys are not suitable for biodegradable implants due to the presence of toxic elements. Therefore, several newer Mg alloys have been developed for biodegradable implant applications as revealed by the huge number of publications dealing with microstructure studies, mechanical property evaluation, degradation characteristics and in vitro and in vivo biocompatibility studies.

Concepts of alloy design

Though pure Mg is highly biocompatible and attractive for bio-implant applications, its mechanical property is relatively low; yield strength of as-cast Mg is ~ 21 MPa which increased to ~ 105 MPa on extrusion[19]. Degradation rate of pure Mg is 2.89 mm/year in 0.9 % NaCl solution[20]. Therefore, several alloys have been developed to improve their mechanical and degradation properties. Mechanical property enhancement can be achieved through (i) solid-solution strengthening, (ii) precipitation strengthening, and (iii) grain refinement[21]. Generally, alloying elements with high solubility in Mg are chosen for strengthening. Atomic diameter of Mg is 0.320 nm and is hexagonal close packed structure with a c/a ratio 1.624[22]. It forms solid solution with a wide range of elements; the solubility limits of the main alloying elements are given in Table 2. Mg alloys, containing elements (Al, Zn, RE) having high solubility in Mg, exhibits precipitation hardening. On the other hand, the alloys made of elements having poor solubility (Si, Ca) do not exhibit strengthening by heat treatment. Grain refinement is another method to increase the mechanical properties and corrosion resistance of Mg alloys. Grain strengthening is governed by the famous Hall-Petch equation (1).

$$\sigma = \sigma_0 + kd^{-1/2} \qquad ...(1)$$

where, yield strength is σ material constant is σ_0, k is the strengthening coefficient and d is the average diameter of the grain. The strengthening coefficient of Mg alloys (280-320 MPa $\mu m^{1/2}$) is several times higher than many face centered cubic (for Al, k is 68 MPa $\mu m^{1/2}$) and body centered cubic metals. Therefore, strengthening of Mg alloys by grain refinement is much more effective[23]. Studies have revealed excellent grain refinement by the addition of Zr, Ca and Si in magnesium. Plastic deformation and/or severe plastic deformation also induces grain refinement; in addition, it can create high density of dislocations and stacking faults in the material.

Table -2 : Main alloying elements in Mg and their solubility limits[6, 24-26]

Element	Solubility limit (wt.%)	Element	Solubility limit (wt.%)	Element	Solubility limit (wt.%)
Ag	15.0	Ho	28.08	Sm	~6.4
Al	12.70	In	53.2	Sn	14.5
Ca	1.34	La	0.23	Sr	0.11
Cd	100	Li	5.5	Tb	24.0
Ce	0.74	Lu	~41	Th	4.75
Dy	25.8	Mn	2.2	Tm	31.8
Er	33.8	Nd	3.6	Y	12.4
Eu	0	Pr	~0.6	Yb	8.0
Ga	8.5	Sc	~24.5	Zn	6.2
Gd	23.49	Si	~0	Zr	3.8

Therefore, selection of alloying elements for developing biodegradable Mg alloys should consider (i) the toxicity, (ii) the strengthening ability, and (iii) the influence of corrosion behavior. With regard to the toxicity level, elements are broadly classified as (a) elements (Ca, Mn, Si, Sn, Zn,) that are nutrients to humans, (b) elements (Ag, Al, Bi, Li, Sr, Zr) that are nutrients to plants and animals, (c) elements (Al, Ce, Co, Cr, Cu, La, Ni, Pr) that create allergy and liver toxicity in humans and (d) toxic elements (Ba, Be, Cd, Th, Pb)[25,27,28]. According to the strengthening ability, elements are divided as: (1) elements that can enhance both the strength and ductility (Al, Zn, Ca, Ag, Ce, Ni, Cu, Th), (2) elements that enhance the ductility reasonably with very scarce improvement in strength (Cd, Li, Tl), (3) elements that augment the strength and reduce the ductility (Bi, Pb, Sb, Sn), (4) impurities (Co, Cu, Fe, Ni)[24,22,29]. Alloying elements influence the corrosion behavior of Mg alloys; the electrochemical potential of magnesium is -2.37 V. Corrosion resistance can be improved by reducing the internal galvanic corrosion by choosing elements (Ce: -2.48 V, Nd: -2.43 V, Y: -2.37 V) with similar electrochemical potential as Mg or the elements that form intermetallic phases. A brief overview of the different Mg-alloy families developed for biodegradable application is given below:

MECHANICAL PROPERTIES OF Mg ALLOY FAMILIES

Mg-Al based alloys

Solubility of Al in Mg is very high (12.7 wt.%) and hence many Mg-Al alloys have been developed for automotive and aerospace applications. Al addition in Mg results solid solution strengthening due to the formation γ-$Mg_{17}Al_{12}$ and α-Mg phases. These alloys possess good castability, reasonable mechanical properties and good corrosion resistance[21]. Zinc and manganese are added to Mg-Al alloys to produce AZ and AM series alloys to improve the strength and ductility at room temperature[30-34]. As-cast AZ91 exhibited 145 MPa yield strength (YS), 275 MPa ultimate tensile strength (UTS) and 6% elongation. This alloy on undergoing (two-step) equal channel angular pressing (ECAP) resulted in enhancing its YS to 290 MPa, UTS to 417 MPa and elongation to 8.45 %[35]. Typical

Table-1 Mechanical properties of conventional metallic biomaterials, Mg-based biodegradable alloys and human bone with their advantages, disadvantages and applications[10,18]

Material	Mechanical Properties				Advantages	Disadvantages	Applications and Remarks
	Density gm/cm^3	Elastic Modulus (GPa)	Yield Strength (MPa)	Tensile strength (MPa)			
Stainless Steel (316L)	7.9	190	221-1213	586-1351	Cheap & readily available Good machinability High fracture toughness Biocompatible	High modulus, results in stress shielding Poor resistance to wear & corrosion Allergy around the implant	1st generation implants Bone plates, Bone screws, Bone Pins & Wires
Co-Cr alloys	7.8	210-253	448-1606	655-1896	Superior corrosion resistance, fatigue & wear High strength Long term biocompatibility	High modulus, results in stress shielding Expensive Difficult to machine Co, Cr & Ni ion toxicity	1st generation implants Popular implant material for joint bearing applications
Ti alloys	4.4	110	485	760	Low density and low modulus of elasticity High specific strength Superior corrosion resistance High biocompatibility	Poor wear resistance Poor bending ductility Expensive	2nd generation implants Fracture fixation plates, screws, rods, nails & wires Total joint replacement (TJR) arthroplasty- hips & knees
Mg-based biodegradable alloys	1.74	45	130	220	Biodegradable/Bioabsorbable Biocompatible, Density & young's modulus close to bone, no stress shielding High tissue growth	High degradation rate with evolution of Hydrogen	Categorized as 3rd generation implants
Human Cortical Bone	1.8-2.0	5-23	104.9-114.3	35-283		-	-

AM60 alloy exhibiting an ultimate tensile strength of 160 MPa and elongation 3%, on rolling (80% reduction) yielded UTS of 378 MPa and elongation of 12% [35]. Grain refinement and precipitation ($Mg_{17}Al_{12}$) strengthening are the reasons for improving the mechanical properties [35]. However, Al is neurotoxic and is not suitable for developing biodegradable alloys.

Mg-Ca and Mg-Sr based alloys

Calcium and strontium are Gr II elements with similar properties. The solubility of Ca and Sr in Mg are 1.34 wt.% and 0.11 wt.%; addition of these elements in more quantities results in the precipitation of Mg_2Ca and $Mg_{17}Sr_2$ along their grain boundaries. The mechanical properties including creep at elevated temperature improves due to the formation of thermally stable intermetallic phases. The tensile strength and elongation of Mg-Ca alloys reduced when the Ca content increased from 1 to 3 wt.%[36]. Addition of 0.6 wt.% Ca improved the bending and compressive strength in Mg-Ca alloys; further addition weakened the properties[37]. Addition of 1 wt.% Y in Mg-1Ca decreased the compressive strength but increased the ductility[38]. The incorporation of 2.31 wt.% Zn in $Mg_{-3}Ca$ alloy improved the strength and ductility[39]. In binary Mg-Sr alloys, the strength increased and elongation reduced till the Sr content increased to 3 wt.%; at higher Sr content the properties deteriorated. Investigation on as-rolled Mg-Sr alloys containing Sr 1-4 wt.%, revealed that the $Mg^{-2}Sr$ alloy has the highest UTS. Therefore, Mg-Ca alloys containing Ca < 1 wt.% and Mg-Sr alloys containing Sr < 2 wt.% are preferred for biodegradable Mg-alloy applications.

Mg-Zn based alloys

Zinc has a solubility of 6.2 wt.% in Mg and forms α-Mg matrix and γ-MgZn phase[40]. The addition of Zn (1 to 6 wt.%) results increasing yield strength (YS); however, Zn content should be limited to 4 wt.% to achieve maximum ultimate tensile strength (UTS) 216.8 MPa and elongation 15.8 %[40]. Mg-Zn alloys with less than 4 wt.% Zn are also alloyed with third alloying elements like Ca[40,41], Mn[42], Sr[43], Y[44], and Zr[45]. When the Ca content in Mg-based alloys is more than 1 wt.%, the strength and ductility of the alloys decreased with increasing Ca content[36]. In $Mg_{-4}Zn$ based alloys, the UTS and ductility decreased when the Ca content is above 0.5 wt.%[40]. Though the microstructures of $Mg^{-2}Zn$ alloys refined with the addition of Ca, Sr and Y, their mechanical properties had very little enhancement[40,41,43,44]. Zirconium is a good grain refiner in Mg-Zn alloys; addition of 0.4 to 0.6 wt.% Zr to $Mg_{-3}Zn$ and $Mg_{-6}Zn$ resulted in grain refinement effectively to produce ZK30 and ZK60 alloys respectively. ZK 30 alloy exhibited yield strength of 215 MPa, ultimate tensile strength of 300 MPa and elongation of 9%. The yield strength, ultimate tensile strength and elongation of ZK60 alloy are 235 MPa, 315 MPa and 8% respectively[45]. Addition of 0.2 wt.% Zr in $Mg^{-2}Zn$ alloy enhanced UTS from 146 MPa to 187 MPa, and elongation from 12 % to 18 %. $Mg_{97}Zn_1Y_2$ (at.%) produced by rapid solidification processing is the strongest (YS: 610 MPa & elongation: 5 %) Mg alloy produced so far[46]. $Mg_{97}Zn_1Y_2$ alloy on extrusion resulted in getting 390 MPa yield strength and 5% elongation[47]. The high strength of Mg-Zn-Y alloys is attributed to the fine grain size, dispersion of hard lamellar phase and occurrence of long-period stacking ordered (LPSO) structure[47].

Mg-Si based alloys

Solubility of Si in Mg is very low (0.003 wt.%) and it reacts with Mg forming intermetallic compound Mg_2Si. Therefore, Mg-Si alloys have initially been developed as magnesium matrix composites[48]. Mg_2Si has low density, high melting temperature, high hardness and low thermal expansion. The precipitation of Mg_2Si phase can efficiently strengthen Mg alloys. However, coarse Chinese script Mg_2Si reduce the ductility; with 0.3 to 2.3 wt.% Si addition ductility limits to < 10%. The highest YS, UTS and elongation obtained with 0.8 wt.% Si addition are 52 MPa, 152 MPa and 9.5 % respectively[49]. Though, the addition of Ca in Mg-Si alloys can refine the grains and change the morphology of Mg_2Si precipitates, their strength and ductility could not be improved[50]. Addition of 1.5 wt.% Zn to $Mg_{0.6}Si$ alloy modified the coarse eutectic structure of Mg_2Si to dot or small bar like structure. As a result, the mechanical properties of the alloy improved.

Mg-Sn based alloys

The ultimate tensile strength and elongation of the as-cast Mg-Sn alloys increased with Sn content up to 5 wt.%, further addition of Sn deteriorated the strength and ductility[51]. Addition of Ca in Mg_5Sn alloy refined the dendritic microstructure and inhibited Mg_2Sn formation resulting in improved creep resistance and shear strength[52]. The strength and elongation of as-cast Mg1Sn alloy is higher than pure Mg[25]. Mg-Sn-Mn alloys with Sn content varying from 1 to 3 wt.% and Mn content varying from 0.5 to 1 wt.% are designed for stent applications; best combination of mechanical properties and corrosion resistance is shown by as-rolled $Mg_3Sn_{0.5}Mn$[53].

Mg-Zr based alloys

Maximum solubility limit of Zr in Mg is 3.8wt.% and is a powerful grain refiner for Mg alloys. Zr is added in Mg alloys containing zinc, rare earth elements, yttrium and thorium. However, Zr could not be used in alloys containing aluminum and manganese as it forms stable compounds with them[19]. Because of its high specific damping capacity Mg-Zr alloys have attracted research interests as it can reduce the vibrations created during movement at the interface of bone and implant[54]. Addition of 1 wt.% Zr in Mg improved the UTS to 172 MPa and elongation to 27 %[25]. Effect of the addition of Sr and Zr is investigated in $Mg_{-x}Zr_{-y}Sr$ alloys (x and y $\leq$ 5 wt.%). Microstructure of the alloys revealed α-Mg matrix, $Mg_{17}Sr_2$ intermetallic phase and unalloyed Zr. The compressive yield strength and the ultimate compressive strength and the ductility obtained from the alloys are 65 to125 MPa, 200 to 290 MPa and 14 to 38 % respectively[55]. The tensile strength and the elongation of Mg-Zr-Ca alloys increased when the Zr content increased from 0.5 wt.% to 1 wt.%[56]. As cast Mg1Ca1Zr exhibited a UTS of 125 MPa and elongation of 8%. Zhou et al reported that the $Mg_{0.5}Zr_1Ca$ alloy is a single phase (α-Mg) but, $Mg_{0.5}Zr_2Ca$ and $Mg_1Zr(1,2)Ca$ alloys consist of α-Mg and Mg_2Ca phases. The strength and ductility of these alloys are $\leq$ 135 MPa and $\leq$ 8% respectively; Mg_2Ca formation along the grain boundaries deteriorates the mechanical properties[56].

Mg-Y and Mg-RE based alloys

The addition of rare earth elements to Mg helps in improving the high temperature properties, creep resistance and corrosion resistance[57-59]. Rare earth elements can be divided into two groups according to their solid solubility in Mg as (i) elements possessing high solubility in Mg (Y, Gd, Tb, Dy, Ho, Er, Tm, Yb and Lu, (ii) elements possessing limited solubility in Mg (Nd, La, Ce, Pr, Sm and Eu[1]. These elements can also be classified as light REE (La, Ce, Pr, Nd and Pm) and heavy REE (Sm, Eu, Gd, Tb, Dy, Ho, Er, Tm, Yb and Lu)[60].

Solubility of La, Ce and Nd in Mg are 0.23 wt.%, 0.74 wt.% and 3.6 wt.% respectively; addition of these elements in Mg would result in the precipitation of intermetallic phases leading to galvanic corrosion. The volume of the intermetallic precipitates increased with the REE content (0 to 5 wt.%) which results in faster degradation in 0.1 M HCl solution[61,62]. A series of Mg-Nd-Zn-Zr alloys, containing 2 – 4 wt.% Nd, 0.1 to 0.5 wt.% Zn and 0.3 to 0.6 wt.% Zr are developed for biomedical applications. These alloys after extrusion exhibited very good mechanical properties (UTS: 300 MPa and elongation 30 %)[63]. The YS, UTS and elongation of as-cast $Mg_3Nd_{0.2}Zn_{0.4}Zr$ alloy are 90 MPa, 194 MPa and 12 % respectively. On extrusion, the mechanical properties of this alloy improved due to grain refinement and dynamic precipitation of $Mg_{12}Nd$ phase[64]. Long-period stacking ordered (LPSO) structure is reported in Mg-RE based alloys (RE = Dy, Er, Gd, Ho, Tb, Tm) when Zn is added as the third alloying element[65]. This LPSO structure exhibits very high plasticity and toughness. The critical resolved shear stress (CRSS) of basal slip (0001)<11-20> of the LPSO structure is 10 to 30 times than that of pure Mg at room temperature[66]. To cite an example, the YS and UTS of extruded $Mg_{7.25}Y_{0.31}Zn$ alloy are 149 MPa and 246 MPa respectively. The extruded $Mg_8Y_1Er_2Zn$ alloy containing LPSO exhibited much higher strength (YS: 275 MPa and UTS: 359 MPa)[67].

The elements possessing very high solubility at eutectic temperatures (Y: 11.4 wt.%, Dy: 25.3 wt.% and Gd: 23.49 wt.%) in Mg are selected for designing biomedical alloys to avoid intermetallic phase formation. Mechanical and degradation properties can be tailored over a wide range by the addition of a single (above mentioned) REEs. Since solid solubility is temperature dependent, properties of Mg-REEs can also be modified by subsequent heat treatment. In general, the strength of the alloys increased with increasing amount of Y, Dy and Gd, but ductility decreased. The degradation rate of these binary alloys depend on the concentration of each element alloyed[68-70]. The YS, UTS and elongation of the extruded $Mg_{11.3}Gd_{2.5}Zn_{0.7}Zr$ alloy are 281 MPa, 341 MPa and 13.5 %. Though this alloy has good bio-corrosion rate, it has slight cytotoxicity[71].

CORROSION

In spite of knowing the electrochemical potential and the corrosion behavior of pure Mg, the degradation behavior of multi-component Mg alloys are difficult to predict. This is due to the microgalvanic corrosion which occurs between the intermetallic phases and the matrix because of the potential difference[72]. The details of the various phases and their potential are not available in most of the newer alloys. Mg alloys undergo degradation in aqueous environment through electrochemical reactions.

$$Mg_{(s)} + 2H_2O \longrightarrow Mg(OH)_{2(s)} + H_{2(g)} \qquad (2)$$

As seen in equation (2), magnesium hydroxide and hydrogen gas is produced[73]. This magnesium hydroxide forms a protective layer in water preventing corrosion; however this changes to $MgCl_2$ (soluble) when the concentration of chloride ions >30 $mmol^{-1}$[73]. The biggest constraint of the Mg-based alloys for orthopedic application is the evolution of hydrogen gas which accumulates around the implant[74]. Evolution of hydrogen can be minimized by improving the Mg-alloy quality; one approach is to develop Mg-based metallic glasses with high Zn content beyond the alloying threshold of Zn, second approach is to improve the corrosion resistance of the alloys which in turn minimize the evolution of hydrogen. The corrosion rate of pure Mg is very high as compared to some of the newly developed Mg-alloys. Corrosion rate can still be improved by alloy design, post processing methods viz. heat treatment, extrusion, rolling etc. and surface modification.

Corrosion resistance also depends on the processing methods followed for the development of the Mg-based alloys. Investigation of the corrosion behavior of as-cast, as-rolled and ECAPed AZ31 alloy is done in Hank's solution. It is reported that the corrosion rate of as-rolled AZ31 is much lower than that of the as-cast AZ31. Further reduction in corrosion rate is not observed in ECAPed AZ31, though the ECAPed AZ31 has much finer grain size than the as-rolled AZ31[75]. The corrosion rates of extruded Mg-Ca[36], Mg-Nd-Zn-Zr[76], and ZK60[77] alloys is lower than the corresponding as-cast alloys. Increase in (i) the density of grain boundaries, (ii) dislocation density and (iii) redistribution of the second phase contribute to the observed improvement of corrosion resistance.

Selection of proper alloying elements is very critical to improve the corrosion resistance of Mg-based alloys. Research efforts have shown that the addition of Al[31], Ca[50], Mn[74], Sn and Sr[78], Zn[41,50], Zr[55] and most REEs including Gd[68] and Nd[64] improved corrosion resistance. The majority of elements have shown a critical limit of alloying addition up to which the corrosion resistance increased and beyond which corrosion resistance deteriorated[36,41,68]. Corrosion resistance can also be controlled by adopting suitable heat treatment protocols to generate single-phase microstructure, and to produce fine and uniform distribution of precipitates[64,79]. Another approach for improving corrosion resistance is by adopting suitable surface modification[80,81]. On the other hand, excessive corrosion occurs when the protective coating develops a minor defect or failure[75].

Orthopedic applications of Mg-based biodegradable alloys

Requirement of orthopedic implants are increasing due to the increased fractures and injuries occurring worldwide. Increased quality of life demands biodegradable bone implants; thus the market for the plates, screw, nails, pins, wires and needles made of Mg-based biodegradable alloys is huge. Mg-based alloy screws made of MgCa0.8[82] MgYREZr[83], LAE442[84] and ZEK 100[85] alloys have been fabricated for animal studies and clinical trials. Osteogenic properties of Mg-based alloys are confirmed in animal trials. Hydrogen gas evolution is not reported in both $MgCa_{0.8}$ and LAE442 alloys. The bone-implant contact in MgCa0.8 alloy is better than that of LAE442 indicating the superior biocompatibility of MgCa0.8 alloy[86]. The mechanical properties of MgCa0.8 screws in the first 2-3 weeks after implantation in adult rabbits are comparable to SS316L[82]. MAGNEZIX (R) screw made of MgYREZr alloy is observed to be clinically similar to standard titanium screw used for the treating the deformities of mild hallux valgus[83]. There are no reports of any inflammatory reactions or other complications in this clinical trial.

CONCLUSIONS AND FUTURE RESEARCH DIRECTIONS

This review summarizes the progress made in the development of Mg-based biodegradable alloys for bio-implant applications. Mg-alloys for implant applications should provide adequate mechanical properties, degradation rate and biocompatibility[87,88]. Therefore, development of Mg-based alloys with superior properties is very crucial. It is known that the microstructure of the alloys depend on alloy composition, alloy preparation method, post processing techniques (heat treatment, mechanical working) and the amount of impurities present in it. Studies so far has shown that Mg-RE- based alloys have exhibited sufficient mechanical properties and reasonable corrosion resistance. As a result, stents and screws made of these materials have passed animal trial and are in clinical trials. Mg-Zn-based alloys are also potential alloys for bio-implant applications because of their mechanical properties. Therefore, future research directions towards development of Mg-based alloys should focus on the following aspects: (i) to develop Mg-based alloys possessing suitable mechanical properties and degradation rate by alloy design (element selection & controlling impurities), method of alloy preparation, post processing techniques (heat treatment, mechanical working), (ii) to develop functional Mg-based alloys using the nutritional elements of the human body viz. Ca, Mn, Sn, Sr, (iii) to understand the biological degradation of the implant especially at the implant / tissue interface.

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