Chapter-2

Literature review

2.1 Introduction

The prevailing chapter is a detailed review of research works available in the open literature related to the preparation of magnesium-based composite materials, their processing technique, various mechanical, physical properties and chemical properties and their applications. It explores the previous works done on magnesium-based composites. Also, the innovative areas of applications of magnesium-based bio composite have been elaborated.

2.2 Magnesium

Magnesium is the 8th most abundant element and constitutes about 2% of the Earth's crust by weight, and it is the third most available element dissolved in the sea. Magnesium is a metal with the symbol Mg, in group II A in the periodic table having atomic number 12 and an atomic weight of 24.312. Its density is 1.74 g/cc. Magnesium is a preferable lighter structural metal in the industry, due to its low weight and its capability of forming mechanically resistant alloys. It is chemically reactive, it takes the place of hydrogen in boiling water and a great number of metals can be produced by thermic reduction of its salts and oxidized forms with magnesium (M. M. Avedesian and H. Baker, 1999). The physical characteristics of magnesium are mentioned in Table 2.1.

TABLE 2.1 Physical properties of magnesium

S.N	Property	Value
1.	Atomic number	12
2.	Atomic weight	24.305 $g.mol^{-1}$
3.	Density	1.74 $g.cm^{-3}$ at 20 °C
4.	Velocity of sound, 20°C	3.67 km/s
5.	Melting point	650°C
6.	Boiling point, 1 atm	1107 °C
7.	Ionization potentials (first)	737.5 $kJ.mol^{-1}$
8.	Vander waals radius	0.16 nm
9.	Ionic radius	0.065 nm
10.	Electronic Shell	$[Ne]3s^2$
11.	Standard potential	-2.34V

2.3 Magnesium based alloys and composites

Magnesium alloys are one of the best known lightest structural alloys (T.T.T Trang, et al, (2018). Different alloying elements have been used to prepare Mg-based alloys and composites. It includes manganese, aluminum, zinc, silicon, zirconium, copper, and rare-earth metals (National Research Council, Washington, DC: The National Academies Press 1975). Magnesium has some favorable properties including a small specific gravity value and high strength-to-weight ratio. Therefore, the material is used in a range of automotive, aerospace, industrial, electronic, biomedical, and commercial applications. Alloying is a preferable used technique to enhance the properties of different elements. There are three major groups of Mg alloys: pure Mg; Aluminum containing alloys such as AZ91, AZ31 and rare earth elements (RE) such as AE21; and the final group consists of the Al free alloys such as Mg-Ca, WE, MZ and WZ (B.M. Girish, 2011). The use of alloying elements can significantly improve the physical and mechanical properties of the magnesium by 1) grain structure refinement, 2) enhancing the corrosion resistance, and 3) inter-metallic phase formation. In Mg-Al alloy Al is an alloying element in Mg and its concentration ranges between 2 and 9 wt % in commercially used Mg-Al alloys. Mg is alloyed with Al to increase both the strength and corrosion resistance (H.S. Brar, et al. 2009; M. Bamberger and G. Dehm, 2008).

2.4 Chemical properties of magnesium and its Alloys

The magnesium metal surface is covered with a thin layer of oxide that facilitates the protection of the metal from the air. On ignition, magnesium metal reacts in the air with a characteristic white flame to give a mixture of white magnesium oxide, MgO, and magnesium nitride. Magnesium is highly reactive towards the halogens such as chlorine and bromine and burns to form the dihalides magnesium chloride ($MgCl_2$) and magnesium bromide ($MgBr_2$) (H. Hu, et al. 2014). Magnesium when dissolved in water or by an electrochemical reaction leads to produces magnesium hydroxide and hydrogen gas. Such a mechanism varies with oxygen concentration, although the presence of oxygen is an important factor in atmospheric corrosion (H. Hu, et al. 2014). The dissolution of magnesium can be represented by the following equation:

$$Mg+2H_2O \longrightarrow Mg(OH)_2+H_2 \quad (2.1)$$

This reaction can be expressed as the summation of the following partial reactions:

Anodic reaction: $Mg \rightarrow Mg^{2+}+2e^-$ (2.2)

Cathodic reaction: $2H_2O+2e^- \rightarrow H_2+2OH^-$ (2.3)

Products formation: $Mg^{2+}+2OH^- \rightarrow Mg(OH)_2$ (2.4)

The reduction of hydrogen ions and the hydrogen overvoltage of the cathode act as a key factor in the corrosion of Mg. Low overvoltage cathodes facilitate hydrogen evolution, causing a substantial corrosion rate (N. D. Tomashov, 1966). In water at elevated temperatures, the corrosion rate in water increases with increasing temperature, corrosion becoming particularly very high above 100°C (W. A Ferrando, 1989). Magnesium gets dissolved by most acids. Also, in dilute solutions of strong and moderately weak acids, magnesium dissolves vigorously. There are some exceptions, like chromic acid and hydrofluoric acid (N. D. Tomashov, 1966). The slow dissolution of magnesium in chromic acid is due to its becoming passive in this acid. An insoluble surface film of MgF_2 is formed which protects it against further attack, and that is why magnesium is resistant to hydrofluoric acid (H. Tamai, et al. 2000). Generally, the corrosion resistance of magnesium alloy is far better than that of pure metal, because corrosion-resistant phases exist in case of alloys. The analysis of the films formed when Mg alloys containing Al, Mn, or Zn are exposed to the atmosphere, shows an enhancement of the secondary phases. It was found that there is a formation of oxide on Mg-Al alloys and has a layered structure composed of MgO/Mg-Al-oxide/substrate, with the Mg-rich oxide becoming thinner with increasing Al content. It is likely that this benefit of Al is related to the strong tendency for Al to form a stable passive film (G. Song and A. Atrens, 1999). The addition of alloying elements more than the solubility limit results in the precipitation of additional content on cooling that subsequently leads to galvanic corrosion at the interface of grain and grain boundaries. This precipitation of additional content due to solubility limit can be avoided by the formation of rapid cooling of molten alloys (X.N. Gu, et al. 2014).

2.5 Characteristics of bioactive glasses and their applications

Bioactive glasses are amorphous silicate-based materials that are compatible with the human body bone and can stimulate new bone growth after dissolution. They, therefore, have the potential to restore the diseased or damaged bone to its original state and function i.e. bone regeneration (L.L. Hench, et al.1971). The glass composition of the first bioactive glass was 46.1% SiO_2, 24.4% NaO, 26.9% CaO and 2.6% P_2O_5, in mol%, termed 45S5 Bioglass (W.F. Enneking, *et al.* 1975; L.L. Hench, *et al.* 1977). Additive manufacturing has also been be used for preparing bioactive glass scaffolds and is used in bone and tissue biomedical implants (I. D. Xynos, et al., 2000). In general, many bioactive glasses are silica-based and form the glass chain with calcium and phosphorous for new bone growth. 45S5 scaffolds were the first robocast in 2013 from melt-quenched glass powder. Sol-gel-synthesized bioactive glasses have great merit over melt-produced glasses, but until recently were never robocast as scaffolds, due to inherent problems, until 2019 when high-silica-content sol–gel bioactive glasses were robocast for the first time (I. D. Xynos, et al., 2000). Bioactive glass is a potential area of research of biomaterials and biomaterial engineering. Bioactive glass binds to interfacial body tissues and may be prepared to also act therapeutically (L. L. Hench, 2013). Recently, it was discovered that bioactive glass has the capability to activate the genes that facilitate osteoblast proliferation and differentiation and promote bone regeneration. Bioactive glass such as 45S5, when implanted in the body reacts with body fluid, exchanging the Na^+ and Ca^{2+} ions on their surface with H^+ ions in the fluid to hydrolyze and form Si-OH bonds, accompanied by an initial increase in pH (P. Paliwal, 2018). Bioactive coatings on the surface are used to diminish the corrosion rate of magnesium composites, but there is a case that coating may be separated during application. Another approach to develop bioactivity and to increase the corrosion resistance of magnesium metal alloys is to add certain additives, which improve not only chemical resistance but also bioactivity (J.M. Anderson, 2000; L Xu, et al. 2009; M. Thomann, et al. 2010; J. Yang, et al. 2012; T. Kokubo,et. al, 2007).

2.6 Biological applications of magnesium-based composites

In the present developing world, traumatic cases are increasing day by day, and bone fracture is one of the most common injuries. For bone fracture curing, fixed implants have been used for the last 30 years. The steel rods, screws and plates made up of titanium and stainless steel have also been used as a bone fastener (M. Geetha, et al. 2009; T. Juutilainen, et al. 1997; Q. Chen and G. A. Thouas, 2015; R. L. Kuncicka and T. C. Lowe, 2017). Metal implants are preferred to repair bone fracture injuries because of their better mechanical properties (S. Wu, et. al, 2013). Stainless steel, Co and Ti-based alloys are generally available for bone implants. Metals are preferred for long-term, durable and load-bearing implants since they exhibit high strength and outstanding ductility that lead to high resistance to fracture.

Among permanent orthopedic implants, the best one is the titanium-based implant, which is widely used. Ti alloys are preferred because of their lower modulus varying from 110 GPa to 55 GPa compared to stainless steel (210 GPa) and chromium cobalt alloys (240 GPa) (M. Geetha, et. al, 2009; T. Juutilainen, et al. 1997). Titanium-based alloys doped with Cr, Mo, Zr, Mn, are used as implant materials in tissue and damaged organs because of their better biocompatibility, high mechanical strength and low corrosion rate (Y. Li, et al. 2014). But the permanent orthopedic implants (Ti and stainless steel based) used, have some added disadvantages, such as post-operative treatment and infections to the surrounding tissues (J.B. Park, 1984; L. Sennerby, et al. 1992; H. Nakashima, et al. 2012; K. Abumi, et al. 2000). It increases the cost of surgery and post-treatment causes discomfort to the grieved patient. The problems concerning permanent implants caused the researchers to think about the development of biodegradable implants, which rule out the subsequent surgery requirement (F. Witte, 2010; M.J. Imola, et al. 2001). The biodegradable implants tend to degrade with time without harming the bone tissues and excrete out of the human body thru metabolic processes (Y. Jang, et al. 2013; M.P. Staiger, et al. 2006; F. Witte, et al. 2006; F. Witte, et al. 2007; W.D. Mueller, et al. 2010).

Magnesium (Mg) based biodegradable materials are a new generation orthopedic implant materials that possess identical mechanical properties as that of human bone. Mg alloys are considered prime substitutes to permanent implants due to their biodegradability in the physiological environment. However, a fast corrosion rate is one of the main constraints of applying Mg alloys in clinical applications in spite of their good biocompatibility. It has been used in medical applications since 1878 (G. He, et al. 2015). Mg is a biodegradable material that degrades completely in the human body after medical use. However, these alloys may suffer from bacterial infections due to their insufficient antibacterial capability. To reduce post-surgical infections, biocompatible alloys were fabricated with the addition of antibacterial Zn with variable contents and evaluated in terms of their biocompatibility and antibacterial properties (G. He, et al. 2015). However, rapid degradation of these materials in physiological environments may lead to gas cavities, hemolysis, and osteolysis and thus, hinder their clinical orthopedic applications. Therefore, these materials are required to have excellent physical and mechanical properties and must be biocompatible, biofunctional, bioadhesive, corrosion-resistant, osteoconductive and have low friction coefficient and wear rates (J.L. Katz, 1980; R. Hedayati, et al. 2000). Different alloying elements have been added to magnesium to improve its mechanical, corrosion and biological characteristics (S.M Rabiee, et. al. 2015). (S. Teoh, 2000). Magnesium alloys that are commercially used in biological applications include the AZ (Mg-Al-Zn), WE (Mg-REZr), and ZK (Mg-Zn-Zr) series alloys. AZ series alloys have been extensively studied both in vitro and in vivo in recent years (H. Wang and Z. Shi, 2011; Y. Song et al. 2009; T. Yan, et al. 2010). It was found that AZ31 and AZ91 alloys release hydrogen after degradation in physiological environments, leading to an increase in both pH and Mg ion concentration (Y. Ding, et al. 2014). In Hank's solution, the AZ31 alloy degrades slower than the AZ91 alloy, but there is no significant change in vivo (F. Witte, et al. 2006). In vivo studies of AZ31 and AZ91 alloys have also indicated that a biocompatible calcium phosphate protective film layer covers their surfaces and increases the formation of new bone mass around the implants (F. Witte, et al. 2005; R. Walter and M. B. Kannan, 2011). It has been found that WE54 (1.58 Nd, 4.85 Y, 0.28 Zr, 0.08 Ce, 0.13 Gd, 0.16 Er, 0.13 Yb, and balance Mg (in wt.%) has marginally higher resistance to corrosion in vitro than pure Mg and heat treatment impacts its degradation (Walter and M. B. Kannan, 2011). Witte et al. investigated the in vivo degradation of four different Mg alloys and confirmed that WE43 (4.16 Y, 3.80 RE, 0.36 Zr, 0.20 Zn, and 0.13 Mn, all in wt.%) has excellent biocompatibility. However, at higher Al concentration in the brain, there are chances of occurrence of Alzheimer's disease and severe hepatotoxicity has occurred after the administration of RE elements, such as Y, Ce, and Pr (L. Tan, et. al, 2013). Mg-based materials have also been used to synthesize cardio vascular stents and can achieve required angiographic results after 4 months by complete and safe desorption (N.T. Kirkland, et al. 2012; R. Erbel, et al. 2007). The time duration for coronary stents to complete the remodeling process of arterial vessels and degrade with optimal mechanical integrity is from 6 to 12 months (M.M. El-Omar, et al. 2001; A. Schcomig, et al. 1994). Many Mg- based bio absorbable stents are used in clinical applications (P. Peeters, et al. 2005; B. Heublein, 2003; W. Ding, 2016) to recover the function of diseased vascular arteries. Rare earth metals alloys have also been observed to have better mechanical and corrosion characteristics, New Mg-RE alloys, such as MgY, Mg-Nd, Mg-Gd, Mg-Ce, and Mg-Ld, also have been studied. Among these, Mg-Nd alloy has a very slow corrosion rate than the other alloys. Mg-Y alloy has been prepared using a zone solidification method and improved corrosion and mechanical characteristics (Q. Peng, et al. 2010). Mg-Y-Zn alloy contains combination of preferred microstructural, mechanical, electrochemical, and biological properties, making it very promising for use as a biodegradable implant material (A. C. Hanzi, et al). It has been found that the addition of Al element having the property of solution and precipitation strengthening by which it significantly improves the strength and corrosion resistance of Mg alloy (P. J. Marie, et al.

1985; P. J. Marie, 2005; S. G. Dahl, et al. 2001). It has been observed that continuous release of trace amounts of aluminum during the degradation process can be tolerated (P. J. Marie, et al. 1985). Also containing Al generally possesses a better aspect of mechanical properties and corrosion resistance (S. Tournis, 2007). Another alloying element zinc is an essential mineral component for hundreds of biological enzymes and transcription factors and is widely associated with the normal function or the structure of more than 300 proteins (E. G. Brandt, et al. 2009; G. D. Jin, et al. 2014). Moreover, many researchers (A. M. Diez-Pascual, et al. 2014; T. J. Wood, et al. 2012; A. Tarushi, et al. 2009; J. A. Xua, et al. 2010; T. N. Phan, et al. 2004) found that Zn exhibits an antibacterial effect on Staphylococcus aureus (S. aureus). The addition of Zn in the Mg–Ca–Sr based alloys may produce a biomaterial with antibacterial properties (H.S. Brar, et al. 2012). Mg alloys that contain zinc as alloying element exhibited good biocompatibility with high cell viability. The antibacterial studies reveal that the number of bacteria adhered on all of these Mg alloy samples diminished remarkably compared to the Ti–6Al–4V control group. We also found that the proliferation of the bacteria was hindered by these Mg alloy extracts. Among the prepared alloys, the Mg–1Ca–0.5Sr–6Zn alloy not only exhibited a strong antibacterial effect, but also enhanced the growth of MC3T3-E1 osteoblasts, making it a promising alloy with both good antibacterial properties and good biocompatibility suitable as an implant (G. Xn, et al. 2010). Zinc is also an essential nutrient in the human body and is good for use in biomedical applications (J. Li, et al. 2010). The Mg corrosion rate can be controlled by increasing the mass fraction of Zn mixed with Mg, thus strengthening the mechanical characteristics of Mg through solid solution strengthening (S. Zhang, et al. 2010). Cai et al. reported that a Zn content of up to 5 wt.% in Mg-Zn binary alloys exhibits grain boundary, solid solution, and secondary phase strengthening, resulting in improved resistance to corrosion and mechanical properties (S. Cai, et al. 2012). Mg-6Zn alloy has good biocompatibility in vitro based on hemolysis and MC3T3-E1 cell adhesion assays (S. Zhang, et al. 2009). Corrosion tests on zinc have been observed to decrease the amount of hydrogen gas evolved during degradation (Z. J. Li, et al. 2008). Zinc can reduce the degradation rate of pure Mg by filling the internal voids and diffusing into Mg matrix. It has been observed that the addition of zinc in magnesium, from 2–6 wt. %, can retard the corrosion rates of the alloys and enhance their mechanical properties (P. Yin, et al. 2013; M. Hambidge, 2000).

Calcium is one of the major constituent in the human body, and is also the major component of human bones (P. Yin, et al. 2013). Also, Ca can to increase the corrosion resistance of pure Mg (Z. J. Li , et al. 2008). Calcium acting as a grain-refning agent in Mg alloys, can stabilize grain size at levels up to 0.5% of the Ca content and cause slight decreases at higher concentration (Y. Li, et al., 2010). It has been investigated that Mg-1Ca alloy does not cause any cytotoxicity and osteoblasts and osteocytes are highly active around Mg-1Ca alloy pins implanted in rabbit femoral shafs, thus showing good biocompatibility and bioactivity (Z. Li, et al. 2008). It is reported that strontium (Sr) has the capability to stimulate bone formation and Sr ranelate has shown to be quite efficient for treating osteoporosis by raising bone mineral density and bone strength (H. S. Brar, 2012). It has been verified that adding a small amount of Ca and Sr into Mg-based alloys can improve the mechanical characteristics of pure Mg by the grain refinement (C. Zhao, et al. 2017). Berglund et al. (I. S. Berglund, et al. 2012) found that low amounts of alloying elements enhanced the corrosion resistance properties, with an optimum composition of 1 wt% Ca and 0.5 wt% Sr. The ternary alloy consists of a higher compressive strength than that of binary Mg–Ca alloys with similar calcium contents and shows negligible toxicity (J. Nriagu, 2011). Brar et al. found that the Mg-0.5Sr alloy corroded the slowest (H. S. Brar, 2012). Zhao et al. and Gu et al., respectively, found that the as-extruded Mg-0.5Sr and as-rolled Mg-2Sr alloys had the best combination of corrosion resistance, high strength, and in vivo biocompatibility (C. Zhao, et al. 2017; X. N. Gu, et al. 2012). Recent research focused on Mg-based amorphous alloys such as Mg-Cu-Y, Mg-Zn-Y (E.L. Zhang, et al. 2008; I. Shogo, et al. 2009; J.Wang, et al. 2001), Mg-Cu-Gd (Q. Zheng, et al. 2007; G.Y. Yuan, 205), Mg-Zn-Ca [(M. Ramy, et al. 2015; H.X. Wang, et al. 2010; N. Hua, et al. 2018). Among these alloys, Mg-Zn-Ca amorphous alloy exhibited significant biocompatibility and mechanical characteristics (H.X. Wang, et al. 2010). Ramya et al. reported that Mg66Zn30Ca4 exhibited higher glass-forming ability than the Mg60Zn35Ca5 (M. Ramy, et al. 2015). Loffler (J.F. Löffler, 2003) found that with the increase of the Zn content, the H_2 evolution rate of Mg60+xZn35−xCa5 (x=0, 3, 6, 7, 9, 12, 14, 15) was gradually decreased. When the Zn content was above 28 %, there was no H_2 liberation in vivo for MgZnCa amorphous alloys.

The desired characteristics of an implant material for being developed as an implant are the formation of apatite after being implanted into living body. Reinforcement of bioactive materials into magnesium up to 20% increases the bioactivity of magnesium by its capacity to form a bond with living bone tissue (F.S.Cai, et al. 2012). Bioactive glass is the best-known material having bioactive behavior in all bioactive materials (R. Hill, 1996). The bioglass addition results in the formation of apatite layer to the Mg–Zn after immersing in the SBF solution (P. Sepulveda, 2002). The higher percentage of bio glass in the composite increases the porosity and apatite formation on the sample surface. More apatite will lead to the rapid bonding between artificial materials and living bones, 1393 BAG is added to increase the bioactivity of the sample and control degradation rate (H. Tripathi, et al. 2019). Biodegradable bone substitute alloy of magnesium (Mg–Zn–Ca) have been developed as materials in different researches (M.G. Jiang, 2016). There are several techniques by which corrosion properties of magnesium alloys can be enhanced. One such technique is alloying non-toxic elements into magnesium and magnesium based alloys to produce new alloy (Q. Zhao, 2013).

In vivo testing on animals helped in characterizing local tissue reactions to Mg-based implants through follow-up testing, including serum analysis, radiographic examination, micro CT investigations, histology analysis, and implant examination (D. Dziuba, et al. 2013). Local bone reactions to biodegradable magnesium alloys depend on the rate of corrosion, corrosion products, and stability of the magnesium alloys. Zhang et al. implanted Mg-Zn-Mn alloy into rats to analyze the in vivo corrosion of Mg alloy, bone response to the magnesium implant, and effect of the degradation of Mg alloy on blood composition and organs. Mg-Zn-Mn alloy was found to corrode at different rates in the marrow cavity and cortical bone. The degradation of the Mg-Zn-Mn implant caused little change to the blood composition, liver, and kidneys (E. L. Zhang, et al. 2009; R.A. Ahmed, et al. 2013). The alloys of Mg can gradually be dissolved and absorbed in the human body after implantation when healing process is complete. Recently corrodible Mg-based alloys have been preferable as coronary stents and orthopedic devices (R.A. Ahmed, et al. 2013; D. Zander and N.A. Zumdick, 2015). In vivo testing on Mg–4.0Zn–0.2Ca has shown that Mg–4.0Zn–0.2Ca alloy extracts have no cytotoxicity on osteoblast cells. The in-vivo behavior of Mg–4.0Zn–0.2Ca alloy was investigated on rabbits. The alloy did not induce inflammation reactions nor affect the new bone formation (Y. Sun, 2011). Three months after the in-vivo experiment, about 35-38% magnesium alloy implant was degraded, and a degradation layer which was composed of Ca, P, O and Mg were formed on the magnesium alloy implants. The in-vitro and in-vivo tests indicated that the Mg–4.0Zn–0.2Ca alloy had good biocompatibility (Y. Sun, 2011). Recent studies by Witte et al. examined the corrosion behavior of Mg-based alloy rods and polymer-based control rods in animal models. The percentage compositions by weight of the Mg alloys investigated consisted of two aluminum-zinc alloys composed of 3% Al and1% Zn {AZ31} and 9% Al and 1% Zn {AZ91} with the balance of the alloys composed of pure Mg. In addition, two RE alloys were studied, the first consisted of 4% yttrium and a 3% rare earth mixture composed of neodymium, cerium and dysprosium (WE43) and the second composed of 4% lithium, 4%, aluminum and a 2% rare earth mixture of cerium, lanthanum, neodymium and praseodymium (F. Witte,et al. 2007; F. Witte,et al. 2011). The implants were harvested at 6 and 18 weeks, with complete implant degradation occurring at 18 weeks. All magnesium-based alloy implants were found to be suitable and promoted new in situ bone tissue formation, while the polymer control rods produced a less significant effect. The LAE442 alloy had the greatest resistance to corrosion, while the other alloys all had identical, but lower values of corrosion resistance and degraded at similar rates (F. Witte, et al. 2007; F. Witte, et al. 2011).

In the case of Mg-Ca alloy, Ca is one of the most important minerals in the composition of bones (J.W Agna, 1958). This fact makes Ca a desirable alloying element in Mg for biomedical applications. The alloying of Mg with Ca leads to the refinement of grain sizes, which results in enhanced mechanical properties of Mg (X.N. Gu, et al. 2014; Y. Wan, et al. 2008).

2.7 Processing methods of magnesium-based Composites

There are various preparations methods for magnesium-based alloy and composites based on their usage and applications. The preparation method consists of Melting method, Sintering, Hot-Dip Galvanization, etc. The method of sintering and temperature of sintering provides an important effect on the strength and properties of the composite prepared via the powder metallurgy route. The sintering temperature is a key factor in the fabrication of composite materials, which can significantly affect the mechanical and physical characteristics of materials (A. Gokce and F. Findik, 2008). The powder metallurgy route for the development of composites is preferred as compared to the melt cast method because of the advantage of less energy usage, a proper blending of the powder with the extra advantage of uniform distribution of additive material and hence uniform mechanical properties and uniformity of reinforcement distribution (K. Yamaguchi, et al. 1997). The powder metallurgy method is based on the principle of mixing of powder and reinforcing elements followed by cold pressing and then sintering (J.W. Kaczmar, et al. 2000; C.P. Ling, et al. 1995). For improvement of the mechanical and physical characteristics, parameters affected by sintering such as density, porosity, grain size, etc. must be optimized (C.R.K Mohan and P. K. Bajpai, 2008; L. Hallmann, et al. 2012). Sintering results in the development of solid dense matter with a defined microstructure (N.J. Shaw, 1989). The process of sintering of powder consists of densification and grain growth, which significantly affects the mechanical and physical properties of a material (D.E Garcia, et al. 2012). It is desirable to control the total densification process of the material as well as the pore presence and to hinder grain growth, during the sintering period, a considerable micro-structural development takes place. The grain size distribution of particles affects the grain boundary which in turn affects the mechanical and physical properties (N.J. Shaw, 1989; D.E Garcia, et al. 2012). In the final stage of sintering, most of the pores disappears (Q. Wang, et al. 2010; K. Krishnan, et al.2006). The strength achieved by any composite material depends on the extent of the optimized temperature at which sintering is done. At lower temperature, a material is prone to lack of integrity and strength, and as the temperature increases grain size is controlled to some extent at some intermediate temperature and then increases as sintering temperature further increases (R. Bjork, et al.2013; T. Hungria, et al. 2009).

2.8 Summary

In the present chapter extensive literature survey is reported and various applications of magnesium and magnesium based alloys were discussed. Study of different load bearing implants was done. Magnesium based alloys and composites were found to be extremely biocompatible and have similar mechanical properties as that of natural bone. This makes them an attractive material for the manufacture of biodegradable implant, with the capability to replace many currently used orthopedic materials such as biodegradable biopolymers.

And despite having the potential to function as an osteoconductive and biodegradable substitute in load bearing applications, the practical application of Mg based alloys faces the serious challenge of overcoming the rapid corrosion rates that occur within the physiological environment of the body. The types of biological corrosion occurring within the body environment and the influence of body fluid pH, concentration of ions, protein adsorption on the implant surface and the influence of the surrounding tissues was discussed. To overcome the effects of biological corrosion, a number of treatment methods designed to reduce the corrosion rate, such as the addition of alloying elements and surface modification techniques were discussed.

On the basis of previous studies it was found that magnesium based alloys and composites are potential biomaterials and can substitute permanent implants. It was found that addition of aluminum, zinc, calcium and manganese improves the antibacterial responses, strength and corrosion characteristics of magnesium based composites. Bioactive glasses when added have shown tremendous bone tissue bonding and observed to have developed layers on metallic surfaces and tend to deteriorate corrosion rates. The main problem associated with magnesium based implants is their high corrosion rate and low mechanical characteristics. Present work focuses on alloying method for improving the properties of magnesium composites and by adding different ingredients.