

**DEVELOPMENT OF BIODEGRADABLE Mg-Zn/HA COMPOSITE VIA
MECHANICAL ALLOYING**

by

EMEE MARINA BINTI SALLEH

**Thesis submitted in fulfillment of the
requirements for the degree of
Doctor of Philosophy**

May 2016

ACKNOWLEDGEMENT

Bismillahirrahmanirrahim.

Alhamdulillah, a grateful gratitude only belongs to ALLAH for bestowing me a valuable opportunity in attaining this great experience in life and for HIS boundless blessings. An infinite praise is as well to HIS Messenger, Prophet Muhammad S.A.W. for his grand inspiration and pure bond of love. *Sollu'alan nabi*.

I am heartily thankful to my supervisor, Professor Dr.Zuhailawati Hussain, whose encouragement, guidance, expertise and research insight from initial to final level enabled me to develop an understanding of the subject. She quickly became for me the role model of a successful researcher in the study. Here, I wish to express my sincere thanks to my co-supervisor, Dr. Sivakumar A/L Ramakrishnan for his helpful advices and supports during my studies. *Thank you, Prof. Zuhaila and Dr. Siva.*

I am deeply thankful to Universiti Sains Malaysia (USM) for giving me this opportunity of being a postgraduate student in School of Materials and Mineral Resources Engineering (SMMRE). I am tempted to individually thank to Dean, Deputy Dean, lecturers, administrative and technical staffs for giving me a countless opportunity, space, facilities and time to work on this research project. I would also like to thank USM RU-PRGS Grant (No. 8046026).

I would like to take this opportunity to also convey my thanks and appreciation to all my friends for their various contributions and assistance, directly or indirectly in the running of this research. However, because the list might be too long and by fear of leaving someone out, I will simply say *thank you very much to all of you.*

I could not finish without saying how grateful I am with my family: brothers, sisters, nieces and nephews all have given me loving surroundings where to develop. A special word of thanks from me is also dedicated to my parents-in-law (Mr. Baharuddin Ismail, Mrs. Fuziah Arifin) and my grandparents-in-law (Mr. Ishak Buyong, Mrs. Fatimah Abdullah) for their continuous valuable encouragements. Particular thanks, of course to Major Ahmad Fitri Faizi Bin Baharuddin TUDM, my most beloved husband for his supports, understandings and endless loves for my whole life. *Barakallahulaka, Abang*. A gratefully wonderful thanks belong to my little caliph, Ahmad Adam Fitri Bin Ahmad Fitri Faizi. Thank you for being my new inspiration and motivation in performing my best in every single thing I do. *Ummmy loves you so much, Adam Fitri*.

My special words of heartiest appreciation also belong to my late dearest father, Allahyarham Salleh Bin Endut. Thanks for your everlasting loves and blessings. May Allah bless you till jannah. *Allahumma firlahu warhamhu wa'afih wa'fu'anh*. *Allahumma ameen*. Lastly, and most importantly, I wish to thank my dearly lovely queen of my heart, Mrs. Kasma Binti Majid. You have always heartened and encouraged me to do my best in all matters of life. *Jazakillahu khayran kathira wafi hifzillah, Ibu*. Thanks for being my great mother and my very best friend. To all of you I dedicate this dissertation. *I love you all*.

Last but of course not least, I offer my regards and blessings to all of those who support me in any respect during the completion of the Ph.D research project.

Emee Marina Binti Salleh,
15 May 2016

TABLE OF CONTENTS

Acknowledgement	ii
Table of Contents	iv
List of Tables	ix
List of Figures	xii
List of Abbreviations	xvii
List of Symbols	xviii
Abstrak	xix
Abstract	xx

CHAPTER 1: INTRODUCTION

1.1 Introduction	1
1.2 Problem Statement	5
1.3 Research Objectives	7
1.4 Research Outline	8
1.5 Scope of Thesis	9

CHAPTER 2: LITERATURE REVIEW

2.1 Introduction	10
2.2 Concern in Current Issues	10
2.3 General Properties of Biomaterials	12
2.4 Metallic Materials as Bioimplant	13
2.5 Magnesium Alloys as Biodegradable Metallic Implant	14
2.5.1 Characteristics of Magnesium	16
2.5.2 Clinical Properties of Magnesium	16

2.6 Ways to Improve the Performance of Magnesium Based Biomaterials	17
2.6.1 Biological Effect of Alloying Elements.....	19
2.6.2 Zinc as Alloying Element	24
2.7 Considerations for Biodegradable Alloy for Biomedical Applications	28
2.7.1 Biocompatibility	30
2.7.2 Mechanical Consideration	32
2.8 Corrosion of Magnesium	35
2.8.1 Weight Loss Measurement	38
2.8.2 Electrochemical Testing by Tafel Extrapolation of Polarization Resistance	39
2.8.3 Inductively Coupled Plasma Optical Immersion Spectroscopy (ICP- OES)	42
2.9 Composite of Mg Alloy Based Matrix Incorporated with Bioactive Material ...	42
2.10 Fabrication Biodegradable Mg based Alloy using Mechanical Alloying	43
2.10.1 Mechanism of Mechanical Alloying	45
2.10.2 Factors Affecting Mechanical Alloying	47
2.10.3 Compaction	51
2.10.4 Sintering	53
2.11 Design of Experiment using Fractional Factorial Design	55
2.11.1 Optimization using Overlaid Contour Plot	57
2.12 Applications	58
2.13 Summary	59
 CHAPTER 3: MATERIALS AND METHODOLOGY	
3.1 Introduction.....	61

3.2 Raw Materials	63
3.2.1 Magnesium, Zinc and Hydroxyapatite.....	63
3.2.2 n-heptane	63
3.3 Preparation of Mg-Zn Alloy Matrix and Composite using Powder Metallurgy..	64
3.3.1 Preliminary Study: Selection of Variables.....	64
3.3.2 Design of Experiment	65
3.3.3 Fabrication of Mg-Zn Alloy Matrix Composite Incorporated with Hydroxyapatite	68
3.3.4 Compaction	69
3.3.5 Sintering	69
3.4 Characterization	70
3.4.1 X-Ray Diffraction	70
3.4.2 Microstructure Study	71
3.4.3 Density Measurement	72
3.4.4 Microhardness Measurement	73
3.4.5 Compression Test	74
3.4.6 Corrosion Test	74
3.4.6.1 Weight Loss Measurement	75
3.4.6.2 Electrochemical Polarization by Tafel Extrapolation	75
3.4.7 In Vitro Biodegradability in HBSS using ICP-OES	77
3.4.7.1 Sample Preparation for Morphological Study	79

CHAPTER 4: RESULTS AND DISCUSSION

4.1 Introduction.....	80
4.2 Raw Materials Characterization	80

4.3 Preliminary Study: Selection of Variables	82
4.3.1 Effect of Milling Time	84
4.3.2 Effect of Milling Speed	91
4.3.3 Effect of Ball to Powder Weight Ratio	98
4.3.4 Effect of Zinc Content	103
4.3.5 Summary of Variables Selection	108
4.4 Design of Experiment of Mg-Zn Alloy.....	109
4.4.1 Phase and Microstructure Analysis	109
4.4.2 Analysis of Variance of Fractional Factorial Design	112
4.4.3 Effect of Dependent Factors on Multiple Responses	122
4.4.4 Optimization of Multiple Responses using Overlaid Contour Plot ...	131
4.4.5 Validation Test of Mg-Zn Alloy Matrix	139
4.4.6 Summary	147
4.5 Fabrication of Mg-6.5wt%Zn Alloy Matrix Composite Incorporated by Hydroxyapatite	147
4.5.1 Characterization of HA Powder	147
4.5.2 Effect of HA Content on Mg-6.5wt%Zn Matrix Composite	148
4.6 Mechanical Properties and Corrosion Behaviour of Optimized Mg-Zn Alloy and Mg-Zn/HA Composite	154
4.7 <i>In Vitro</i> Biodegradation of Mg-Zn Alloy and MgZn-HA Composite	161
4.7.1 Surface Morphology of Pure Mg, Mg-Zn Alloy and Mg-Zn/HA Composite after Immersion in HBSS	165
4.8 Mechanical Integrity after Immersion in HBS Solution	177

CHAPTER 5: CONCLUSION AND RECOMMENDATION

5.1 Conclusion 179

5.2 Contribution of Current Work 180

5.3 Suggestions for Future Work 181

REFERENCES 182

LIST OF PUBLICATIONS 196

APPENDICES

LIST OF TABLES

Table 2.1	Physical and mechanical properties of natural bone and some implant materials (Gupta and Sharon, 2011; Poinern et al., 2012)	13
Table 2.2	Mechanical and corrosion properties of different Mg based alloy system	18
Table 2.3	Clinical effect of alloying element added in Mg alloy (Yang et al., 2008; Poinern et al., 2012; Holzapfel et al., 2013)	20
Table 2.4	Effect of alloying element that are commonly used in Mg alloy (Gupta and Sharon, 2011; Gu et al., 2009; Biesiekierski et al., 2012; Poinern et al., 2012; Holzapfel et al., 2013)	22
Table 2.5	Physical properties of Zn (Mezbahul et al., 2014)	25
Table 2.6	Average corrosion rate for samples immersed in 3.5 wt% NaCl solution	39
Table 3.1	Physical properties of Mg, Zn and HA (Gupta and Sharon, 2011; Poinern et al., 2012; Mezbahul et al., 2014)	63
Table 3.2	Physical properties of n-heptane (Nouri et al., 2010)	63
Table 3.3	Details of mechanical alloying parameters	65
Table 3.4	Factors and levels evaluated in the experiments	67
Table 3.5	The 2^{4-1} experimental design of synthesis Mg-Zn alloy by MA	68
Table 3.6	Composition of blood plasma and consumed electrolytes (Kokubo and Takadama, 2006; Bohner and Lemaitre, 2009)	75
Table 3.7	Operating conditions of instrument and spectral line of analytes	79
Table 4.1	Distribution of average particle size of Mg and Zn powders	82
Table 4.2	Distribution of average particle size of as milled Mg-Zn powders at various milling times	88
Table 4.3	Distribution of average particle size of as milled Mg-Zn powders at various milling speed	95
Table 4.4	Distribution of average particle size of as milled Mg-Zn powders at various BPR	100

Table 4.5	Distribution of average particle size of as milled Mg-Zn powders at various Zn content	106
Table 4.6	Summary of range of selected variables	108
Table 4.7	The 2^{4-1} experimental design and actual responses of binary Mg-Zn alloy	113
Table 4.8	Analysis of variance of compressive modulus (coded units)	116
Table 4.9	Analysis of variance of weight loss (coded units)	116
Table 4.10	Effects and regression coefficient of compressive modulus	118
Table 4.11	Effects and regression coefficient of weight loss	118
Table 4.12	Internal strain of single replication of Mg-Zn alloy	124
Table 4.13	Dislocation density of single replication of Mg-Zn alloy	125
Table 4.14	Crystallite size of single replication of Mg-Zn alloy	127
Table 4.15	Lattice parameter of single replication of Mg-Zn alloy	129
Table 4.16	Relative density of single replication of Mg-Zn alloy	130
Table 4.17	Predicted and measured compressive modulus of Mg-Zn alloy	143
Table 4.18	Predicted and measured weight loss of Mg-Zn alloy	143
Table 4.19	Characteristic profile of Mg-Zn alloy at validation condition	145
Table 4.20	Density profile of Mg-6.5wt%Zn/HA composite	151
Table 4.21	Compressive characteristics of Mg-6.5wt%Zn/HA composite	153
Table 4.22	Corrosion behaviour of pure Mg, Mg-6.5wt%Zn alloy and Mg-6.5wt%Zn/HA composite	154
Table 4.23	EDX profile of pure Mg, Mg-Zn alloy and Mg-Zn/HA composite after immersion in HBSS	160
Table 4.24	Relative molecular mass of the pure Mg, Mg-Zn alloy and Mg-Zn/HA composite	164
Table 4.25	EDX profile of pure Mg after immersion in HBSS	167
Table 4.26	EDX profile of Mg-Zn alloy after immersion in HBSS	169

Table 4.27	EDX profile of Mg-Zn/HA composite after immersion in HBSS	172
Table 4.28	EDX profile and Ca: P ratio of ion dissolution in HBSS	176
Table 4.29	Compression test data of pure Mg, Mg-Zn alloy and Mg-Zn/HA composite after 7 days immersion	177
Table 4.30	Comparison of Mg based alloy with different elements	178

LIST OF FIGURES

Figure 2.1	Statistic of Syria conflict death tolls (Sources: Violations Documentation Center, Syrian Network for Human Rights, Syrian Center for Statistics and Research, Failing Syria aid agency report; Rodgers et al, 2015)	11
Figure 2.2	(a) Current condition in Aleppo, Syria and (b) innocently injured Syrian kids (Hurd, 2013)	11
Figure 2.3	Hexagonal close packed structure of pure magnesium (Avedusian and Baker, 1999)	16
Figure 2.4	Published tensile strength and elongation data for various magnesium alloys (Li et al., 2009; Zhang et al., 2010a; Witte et al., 2008; Gu et al., 2009; Zhang and Yang, 2008; Zeng et al., 2008; Zhang et al., 2009; Wang et al., 2012; Jihua et al., 2009; Hort et al., 2010; Guan et al., 2013)	21
Figure 2.5	Binary phase diagram of Mg-Zn system (Yao et al., 2014)	27
Figure 2.6	General composition of bone tissue (Alvarez and Nakajima, 2009)	29
Figure 2.7	Requirements of implants (Zeng et al., 2008; Yang et al., 2008)	30
Figure 2.8	Mg concentrations in SBF solutions at 3, 10 and 20 days of pure Mg and Mg-1X alloys for the (a) as-cast and (b) as-rolled samples (Gu et al., 2009)	31
Figure 2.9	Stress shielding in locking compression hip joint fracture (Lucas et al., 1999)	33
Figure 2.10	Typical true stress-strain curve for ductile material	34
Figure 2.11	Tafel extrapolation viewing anodic and cathodic components of corrosion (Shi et al., 2010)	40
Figure 2.12	Tafel curves of pure Mg and as-cast Mg-Ca alloys in SBF (Harandi et al., 2013)	41
Figure 2.13	Deformation characteristics of representative constituents of starting powders in mechanical alloying (Suryanarayana, 2001)	47
Figure 2.14	Configuration of horizontal ball thrown method of planetary ball mill (Suryanarayana, 2001)	49

Figure 2.15	Microstructural changes of (a) powder particles after pressing, (b) particle welding and pore formation as sintering begins, (c) pore changes in size and shape as sintering is prolonged	54
Figure 2.16	Typical sequence of operation in a sintering furnace	55
Figure 2.17	Schematic diagram of experimental design	56
Figure 2.18	Overlaid contour plot of steel, brass, cost (Gomes et al., 2015)	58
Figure 2.19	(a) Biodegradable stent made of WE43 alloy, (b) Mg-0.8Ca screw, ZEK100 plate, intramedullary LAE442 nail (from top to bottom), (c) cortical bone screw made of AZ31 and (d) interlocking healing caps and closure screws made of AZ31 alloy	59
Figure 3.1	Flow of overall experimental work	62
Figure 3.2	Sequence of DOE procedure on statistical design	66
Figure 3.3	Schematic profile of sintering process	70
Figure 3.4	Schematic diagram of electrochemical polarization corrosion test	76
Figure 3.5	Tafel extrapolation of the corrosion characteristics (Shi et al., 2010)	77
Figure 3.6	A submerged specimen in simulated body solution	78
Figure 4.1	XRD spectra and SEM (inserted images) of starting materials (a) magnesium and (b) zinc	81
Figure 4.2	DSC profile of as milled Mg-Zn powder	84
Figure 4.3	XRD pattern of sintered Mg-Zn alloy milled at (a) 1 hour, (b) 2 hours, (c) 5 hours, (d) 10 hours, (e) 15 hours and (f) 30 hours	86
Figure 4.4	Crystallite size and internal strain of Mg-Zn alloy for various milling time	87
Figure 4.5	SEM images of as-milled Mg-Zn powder milled at (a) 1 hour, (b) 2 hours, (c) 5 hours, (d) 10 hours, (e) 15 hours and (f) 30 hours	89
Figure 4.6	Density and microhardness of Mg-Zn alloy for various milling time	90

Figure 4.7	XRD pattern of sintered Mg-Zn alloy milled at (a) 100 rpm, (b) 200 rpm, (c) 300 rpm and (d) 400 rpm	92
Figure 4.8	Crystallite size and internal strain of Mg-Zn alloy at various milling speed	94
Figure 4.9	SEM images of as-milled Mg-Zn powder milled at (a) 100 rpm, (b) 200 rpm, (c) 300 rpm and (d) 400 rpm	95
Figure 4.10	Density and microhardness of Mg-Zn alloy at various milling speed	96
Figure 4.11	XRD pattern of sintered Mg-Zn alloy milled using BPR of (a) 5:1, (b) 10:1, (c) 15:1 and (d) 20:1	99
Figure 4.12	Crystallite size and internal strain of Mg-Zn alloy using different BPR	100
Figure 4.13	SEM images of as-milled Mg-Zn powder milled using BPR of (a) 5:1, (b) 10:1, (c) 15:1 and (d) 20:1	101
Figure 4.14	Density and microhardness of Mg-Zn alloy using different BPR	103
Figure 4.15	XRD pattern of sintered Mg-Zn alloy added with (a) 3 wt%, (b) 5 wt%, (c) 10 wt% and (d) 15 wt% of Zn	104
Figure 4.16	Crystallite size and internal strain of sintered Mg-Zn alloy for different Zn content	105
Figure 4.17	SEM images of as-milled Mg-Zn powder with (a) 3 wt%, (b) 5 wt%, (c) 10 wt% and (d) 15 wt% of Zn	106
Figure 4.18	Density and microhardness of Mg-Zn alloy using different Zn content	108
Figure 4.19	XRD pattern of sintered Mg-Zn alloy for a single replication of (a) Mg, (b) sample 1, (c) sample 2, (d) sample 3, (e) sample 4, (f) sample 5, (g) sample 6, (h) sample 7 and (i) sample 8	110
Figure 4.20	Micrograph and EDX profile of (a) Mg-3wt%Zn (Sample 3) and (b) Mg-10wt%Zn alloy (Sample 10)	111
Figure 4.21	Residual plots of compressive modulus	119
Figure 4.22	Residual plots of weight loss	119
Figure 4.23	Pareto chart compressive modulus	121

Figure 4.24	Pareto chart of weight loss	121
Figure 4.25	Main plot effects of compressive modulus	122
Figure 4.26	Main plot effects of weight loss	123
Figure 4.27	XRD spectra of as milled alloy powder of (a) Mg-3wt%Zn and (b) Mg-10wt%Zn	131
Figure 4.28	Contour plot corresponded to compressive modulus of Mg-Zn alloy	133
Figure 4.29	Contour plot corresponded to weight loss of Mg-Zn alloy	134
Figure 4.30	Overlaid contour plot of multiple responses corresponding to compressive modulus and weight loss	137
Figure 4.31	Optimization plot of compressive modulus and weight loss global solution	139
Figure 4.32	Variation of milling time for experimental validation test	140
Figure 4.33	XRD spectra of validation test for various milling time of (a) 3 hours, (b) 4 hours, (c) 5 hours, (d) 6 hours, (e) 7 hours and (f) 8 hours	141
Figure 4.34	Compressive modulus and weight loss of validation test for various milling time	142
Figure 4.35	Relative density and microhardness of validation test for various milling time	144
Figure 4.36	XRD spectra of validation test of Mg-Zn alloys added with (a) 10 wt%, (b) 12 wt%, (c) 14 wt% and (d) 16 wt% Zn	146
Figure 4.37	(a) XRD spectrum and (b) SEM image of raw HA powder	148
Figure 4.38	XRD spectra of MgZn alloy matrix composite incorporated with (a) 5 wt%, (b) 10 wt%, (c) 15 wt% and (d) 20 wt% of HA	149
Figure 4.39	SEM images of sintered compact and as milled powder (inserted images) incorporated with (a) 5 wt%, (b) 10 wt%, (c) 15 wt% and (d) 20 wt% HA	150
Figure 4.40	Relative density and microhardness of Mg-Zn/HA composite	152
Figure 4.41	Compressive stress strain curves of pure Mg, Mg-Zn alloy and Mg-Zn/HA composite	156

Figure 4.42	Electrochemical polarization curves of pure Mg, Mg-Zn alloy and Mg-Zn/HA composite	157
Figure 4.43	XRD spectra of corrosion product of (a) pure Mg, (b) Mg-Zn alloy and (c) Mg-Zn/HA composite	158
Figure 4.44	Microstructure and EDX profile of (a) pure Mg, (b) Mg-Zn alloy and (c) Mg-Zn/HA composite after immersion test	159
Figure 4.45	Mg concentration of pure Mg, Mg-Zn alloy and Mg-Zn/HA composite after immersion test in HBSS	162
Figure 4.46	Element concentrations of pure Mg, Mg-Zn alloy and Mg-Zn/HA composite after 4320 minutes of immersion in HBSS	163
Figure 4.47	Morphology of pure Mg surface after immersion in HBSS at (a) 30 minutes, (b) 60 minutes, (c) 120 minutes, (d) 240 minutes, (e) 1440 minutes and (f) 4320 minutes	166
Figure 4.48	Schematic illustration of the corrosion of Mg in an aqueous environment	168
Figure 4.49	Morphology of Mg-Zn alloy surface after immersion in HBSS at (a) 30 minutes, (b) 60 minutes, (c) 120 minutes, (d) 240 minutes, (e) 1440 minutes and (f) 4320 minutes	170
Figure 4.50	Schematic of galvanic corrosion between the anodic Mg matrix and the cathodic second phase	171
Figure 4.51	Morphology of Mg-Zn/HA composite surface after immersion in HBSS at (a) 30 minutes, (b) 60 minutes, (c) 120 minutes, (d) 240 minutes, (e) 1440 minutes and (f) 4320 minutes	173
Figure 4.52	Schematic illustration of the formation of HA deposit on Mg alloys: (a) a reduction of H_2O produces many OH^- , and then a deposition of Ca-P phase is triggered, (b) an enrichment of OH^- promotes a transformation of HPO_4^{2-} and PO_4^{3-} ; (c) DCPD deposits on the Mg surface in alkaline environment provides a degree of supersaturation and (d) HA finally forms converted from the DCPD	175

LIST OF ABBREVIATIONS

Ar	Argon
BPR	Ball to powder weight ratio
BSE	Back scattered electron
DSC	Differential Scanning Calorimetry
EDX	Energy Dispersive X-ray
FESEM	Field Emission Scanning Electron Microscopy
ICP-OES	Inductively Coupled Plasma Optical Emission Spectroscopy
MA	Mechanical Alloying
MMC	Metal Matrix Composite
PM	Powder Metallurgy
Rpm	rotation per minute
XRD	X-ray Diffraction

LIST OF SYMBOLS

\AA	Angstrom
2θ	Diffraction angle
e^-	Electron
a, c	Lattice parameter
H_v	Vickers hardness
V	Voltage
E_{corr}	Corrosion potential
R_p	Polarization resistance
T_m	Melting temperature

PEMBANGUNAN KOMPOSIT BIODEGRADASI Mg-Zn/HA MELALUI PENGALOIAN MEKANIKAL

ABSTRAK

Kajian ini bertujuan untuk membangunkan bahan logam biodegradasi menggunakan pengalioian mekanikal (MA). Magnesium (Mg) adalah calon yang paling menjadi tumpuan bagi aplikasi bioperubatan berdasarkan kelebihan sifat-sifatnya berbanding bahan bio yang lain. Tetapi kadar degradasi yang cepat dalam persekitaran fisiologi menghadkan prestasinya. Oleh itu, Mg telah dialoikan dengan zink (Zn) bagi meningkatkan kerintangan kakisan dan mengekalkan integritimekanikal. Dalam mencapai sasaran ini, bahan bio berasaskan Mg telah difabrikasi melalui MA diikuti dengan pemadatan di bawah 400 MPa dan pensinteran pada 350°C. Empat parameter MA iaitu masa pengisaran, kelajuan pengisaran, nisbah berat bola kepada serbuk (BPR) dan kandungan Zn telah disiasat. Ketumpatan 1.80 hingga 1.99 g/cm³ yang setara dengan tulang manusia dan kekerasan mikro yang lebih baik daripada Mg tulen (39.30 HV) iaitu antara 53.76 hingga 94.37 HV telah diperolehi. Berdasarkan rekabentuk faktorial pecahan (FFD), keadaan MA optimum dalam menghasilkan aloi Mg-Zn dicapai dengan menambah 6.5 wt% Zn yang dikisar selama 5 jam pada 200 rpm dengan 7: 1 BPR. Kekuatan mampatan yang lebih tinggi (249.28 MPa) dan kadar kakisan yang lebih rendah (1.13×10^{-2} mm/y) daripada Mg tulen (178.04 MPa dan 13.77×10^{-2} mm/y) telah diperolehi. Penambahbaikan sifat-sifat tersebut telah dicapai dengan menambah 10 wt% HA ke dalam aloi Mg-6.5wt%Zn. Kekuatan mampatan (292.33 MPa) dan kadar degradasi (0.72×10^{-2} mm/y) yang bagus diperolehi. Komposit Mg-Zn/HA memberikan bioaktiviti paling tinggi dengan nisbah Ca:P sebanyak 1:1.46 diikuti oleh aloi Mg-Zn 1:1.29 memenuhi keperluan pemineralan awal tulang iaitu 1:1 kepada 1:1.67.

DEVELOPMENT OF BIODEGRADABLE Mg-Zn/HA COMPOSITE VIA MECHANICAL ALLOYING

ABSTRACT

This work aims to develop biodegradable metallic material using mechanical alloying (MA). Magnesium (Mg) is the most highlighted candidate for biomedical applications because of its advantageous properties as compared with other biomaterials. But a rapid degradation rate in physiological environment limits its performance. Hence, Mg was alloyed with zinc (Zn) in order to improve its corrosion resistance and sustain its mechanical integrity. In achieving the target, Mg based biomaterials were fabricated using MA followed by compaction under 400 MPa and sintering at 350 °C. Four MA parameters namely milling time, milling speed, ball-to-powder-weight ratio (BPR) and Zn content were investigated. The density of 1.80 to 1.99 g/cm³ which is comparable to human bone and improved microhardness of 53.76 to 94.37 HV as compared to pure Mg (39.30 HV) were attained. By fractional factorial design (FFD), an optimized MA condition in producing Mg-Zn alloy was achieved by adding 6.5 wt% Zn and milled for 5 hours at 200 rpm with 7:1 BPR. A higher compressive strength (249.28 MPa) and lower corrosion rate (1.13×10^{-2} mm/y) than pure Mg (178.04 MPa and 13.77×10^{-2} mm/y) were acquired. A further improvement of those properties was attained by incorporating 10 wt% HA into optimized Mg-6.5wt%Zn alloy. An enhanced compressive strength (292.33 MPa) and degradation rate (0.72×10^{-2} mm/y) was attained. Mg-Zn/HA composite provided the highest bioactivity due to highest Ca:P ratio of 1:1.46 followed by Mg-Zn alloy of 1:1.29 which is in agreement with the required Ca:P ratio of 1:1 to 1:1.67 for initial bone mineralization.

CHAPTER 1

INTRODUCTION

1.1 Introduction

The use of metallic materials for medical implants can be traced back to the 19th century, leading up to the era when the metal industry began to expand during the Industrial Revolution (Kraus *et al.*, 2012). The development of metallic implants was primarily driven by the demands for approaches to bone repair, typically internal fracture fixation of long bones. However, almost no attempts of implanting metallic devices, such as spinal wires and bone pins made from iron, gold or silver, were successful until Lister's aseptic surgical technique was implemented in the 1860s (Xin *et al.*, 2011). Since then, metallic materials have predominated in orthopaedic surgery, playing a major role in most orthopaedic devices, including temporary devices (e.g. bone plates, pins and screws) and permanent implants (e.g. total joint replacements) (Zeng *et al.*, 2008).

The conventional metallic implant materials namely titanium (Ti) alloys, stainless steels and (Co-Cr) alloys possess excellent mechanical capabilities and highly resistance to corrosion (Hermawan *et al.*, 2010; Castellani *et al.*, 2011; Anghelina *et al.*, 2013). However, when these conventional alloys are used as temporary implant devices, a second surgical procedure is required for the implant removal after the traumatized tissues have healed which markedly increases the health care cost. Besides, there is an increased risk of local inflammation due to potential release of cytotoxic ions as well as the physical irritation due to the rigidity of these conventional implants (Li *et al.*, 2012).

Currently, the development of new biodegradable metallic biomaterials combining excellent strength retention properties and improved biocompatibility for several applications such as stents for blood vessels and screws and plates for fixing hard tissues are highly desirable (Hort *et al.*, 2010). The main driving force to develop biodegradable implants is an elimination of secondary surgical procedure as they have ability to biodegrade in the bioenvironment during the implantation duration. Hence, the paradigm of metallic implants must be highly inert and corrosion resistance has now been challenged by advent of the new class of degradable biomaterials (Lei *et al.*, 2012).

Magnesium (Mg) and its alloys have attracted increasing attention as innovative biodegradable materials for temporary orthopaedic implants due to their excellent biological performance and biodegradability in bioenvironment. In terms of mechanical properties, Mg is well compatible with natural bone. Its density (1.74 g/cm^3) and Young's modulus (40 - 48 GPa) are closer to those of bone ($1.8 - 2.1 \text{ g/cm}^3$ and 40 - 57 GPa) than in the case of other currently used biomaterials for fixation of fractured bone, like Ti alloys, stainless steels or Co-Cr alloys at approximately 100, 180 and 210 GPa respectively (Li *et al.*, 2004; Sudhakar, 2005; Gupta and Sharon, 2011). In term of biocompatibility, Mg ions are present in a large amount in the human body and they are involved in many metabolic reactions and biological mechanisms. The human body usually contains Mg at approximately 35 g per 70 kg body weight and the daily demand for Mg is about 375 mg (Gill *et al.*, 2011).

An attractive characteristic of Mg due to its corrodibility makes it as a potential biodegradable metallic implant. However, the fast degradation rate of Mg in human bioenvironment containing chloride anions (Cl^-) about 100 mmol/l

concentration limited its clinical application (Wang *et al.*, 2012). A rapid degradation rate during implantation results in a deterioration of its mechanical performance which then causes a worst injury to a traumatized hard tissue. Elemental alloying is one of the most effective way to improve corrosion resistance as well as mechanical properties of Mg. Mg alloys containing aluminium (Al) and rare earth (RE) elements showed relatively high strength and good corrosion resistance against sodium chloride (NaCl) solution such as AZ61 (Mg-6wt%Al-1wt%Zn) and WE43 (Mg-(3.7-4.4wt%) Y-(2.4-4.4 wt%) E-0.42wt%Zr) alloys (Gupta and Sharon, 2011). However it has been reported that the Al release from Mg alloy into human body could induce Alzheimer's disease, allergic reaction and neurological disorder (Silva *et al.*, 2004; Zhang *et al.*, 2010b).

The exploration of high strength biodegradable Mg alloys without Al for medical implants has gained great attention in the past years and it is still ongoing. In certainty, there are only a small number of elements that can be tolerated in human body and can also retard the biodegradation of Mg alloys including calcium (Ca), manganese (Mn), zinc (Zn) and perhaps very small amount of low toxicity rare earth including niobium (Nb) and tantalum (Ta). Zn has been found to be next to Al in strengthening effectiveness as an alloying element in Mg. Adding Zn to Mg may improve both tensile strength and the corrosion resistance (Yin *et al.*, 2008). Biologically, Zn is a necessary microelement and component of many amino acids and nucleic acids syntheses of human body. Moreover, Zn is an inexpensive alloying element with a potential of accelerating the metabolism of cells and bone healing (Jang *et al.*, 2013). Therefore, it is a contributing approach to develop Mg-Zn alloy with good corrosion resistance for a temporary bioimplant in biomedical applications.

Typically, metallic materials including Mg based alloy have been produced using conventional liquid state processing such as casting (Zeng *et al.*, 2011; Lei *et al.*, 2011; Kubok *et al.*, 2013). However due to some defects that are usually found in cast Mg alloy causing poor final properties, further treatments may be required to improve the situation. Recently, powder metallurgy (PM) process coupled with mechanical alloying (MA) to synthesize Mg based alloys is of growing interest (Gonzalez *et al.*, 2012; Patel and Morsi, 2012). This technique is a solid state powder metallurgical process in which elemental powders are being alloyed by repeated deformation mechanism under frequent mechanical impacts (Suryanarayana, 2001). MA is one of the simplest and most economical routes for the fabrication of nanocrystalline materials. In addition, MA offers the possibility to scale up the quantity of processed material to tonnage amount and be employed for the processing of nearly all types of materials (Yadav *et al.*, 2012). This makes MA the ideal processing route for small as well as for large scale production of nanocrystalline materials.

Another important point for a biomaterial is the ability of the implant to establish bonding with the surrounding bone tissue which is the bioactivity of the implant (Khanra *et al.*, 2010). Therefore, it seems necessary to increase the bioactivity of Mg based alloys by introducing bioactive materials into the matrix (Khalil, 2012). Bioactive ceramic such as hydroxyapatite (HA; $C_{10}(PO_4)_6(OH)_2$) has been widely used as an implant material for hard tissues owing to its excellent biocompatibility to human tissues because it has similar structure with the mineral part of bone. HA has calcium (Ca) and phosphorus (P) elements in its hexagonal structure (Veljovic *et al.*, 2009). These elements present the inorganic of the bone. Therefore, strong bonds are spontaneously generated to living bone via an apatite