ZIRCONIA REINFORCED HYDROXYAPATITE BIOCOMPOSITE FOR STRENGTH AND TOUGHNESS IMPROVEMENT

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by

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LIST OF ABBREVIATION

Al_2O_3	:	Alumina
BCP	:	Biphasic Calcium Phosphate
BET	:	Brunauer, Emmet and Teller
c- ZrO_2	:	Cubic Zirconia
Ca	:	Calcium
CaF_2	:	Calcium Fluoride
CaO	:	Calcia
CaO-ZrO_2	:	Calcia stabilized zirconia
CaP	:	Calcium Phosphate
CDA	:	Calcium Deficient Apatite
CeO_2	:	Ceria
CIP	:	Cold Isostatic Pressing
EDX	:	Energy Dispersive X-ray Spectroscopy
FESEM	:	Field Emission Scanning Electron Microscope
FSZ	:	Fully Stabilized Zirconia
HAp	:	Hydroxyapatite
ICDD	:	International Centre for Diffraction Data
m- ZrO_2	:	Monoclinic Zirconia
MgO	:	Magnesia
MOR	:	Modulus of Rupture
MPa	:	Megapascal
OH	:	Hydroxide
P	:	Phosphorus
PO_4	:	Phosphate
PSZ	:	Partially Stabilized Zirconia
SBF	:	Simulated body Fluid
SEM	:	Scanning Electron Microscope
SiC	:	Silicon Carbide
t- ZrO_2	:	Tetragonal Zirconia
TEM	:	Transmission Electron Microscopy

TTCP	:	Tetra-Calcium Phosphate
XRD	:	X-Ray Diffraction
Y_2O_3	:	Yttria
Y_2O_3-ZrO_2	:	Yttria stabilized zirconia
ZrO_2	:	Zirconia
ZrO_2/HAp	:	Zirconia reinforced hydroxyapatite biocomposite
β-TCP	:	Beta Tri-Calcium Phosphate
α-TCP	:	Alpha Tri-calcium Phosphate

LIST OF PUBLICATIONS

1. Firmandika Harda, Ch'ng Lay Ean, Aditianto Ramelan, Kunio Ishikawa and Ahmad Fauzi Mohd Noor. (2007) Physical and Mechanical Properties of ZrO_2-HAp Biocomposite Aided by CaF_2. *Proceeding. International Conference on Engineering and Environment (ICEE) 2007*. 10^{th}-11^{th} May 2007. Faculty of Engineering, Prince of Songkla University, Thailand.

2. Firmandika Harda, Hasmaliza Mohamad, Aditianto Ramelan, Kunio Ishikawa and Ahmad Fauzi Mohd Noor. (2007) Mechanical Properties Improvement of Zirconia-Hydroxyapatite Bio-Composite with Addition of CaF_2. *Proceeding. 10^{th} AUN/SEED-Net/JICA Field-Wise Seminar on Engineering Materials*, 5^{th}-6^{th} June 2007, School of Materials and Mineral Resources Engineering, Universiti Sains Malaysia, Malaysia.

3. Firmandika Harda, Hasmaliza Mohamad, Aditianto Ramelan, Kunio Ishikawa and Ahmad Fauzi Mohd Noor. (2007) Fracture Toughness Improvement of Zirconia-Hydroxyapatite Biocomposite. *Proceeding. International Conference On Sustainable Materials (ICOSM) 2007*. 9^{th}-11^{th} June 2007. School of Materials Engineering, Universiti Malaysia Perlis, Malaysia.

4. Firmandika Harda, Niki Prastomo, Wong Gar Shen, Siew Fook Wai, Zainovia Lockman, Hasmaliza Mohamad and Ahmad Fauzi Mohd Noor. (2007) Chemical Synthesis of Nanocrystals Tetragonal Zirconia Powders. *Proceeding. 12^{th} Asian Chemical Congress (12ACC)*. 23^{rd}-25^{th} August 2007, Institut Kimia Malaysia and Federation of Asian Chemical Societies, Malaysia.

5. Firmandika Harda, Aditianto Ramelan, Kunio Ishikawa and Ahmad Fauzi Mohd Noor. (2008) Investigation on Phase Stability of Zirconia/Biphasic Calcium Phosphate (ZrO_2/BCP) Biocomposite. *Proceeding. International Conference on X-Rays & Related Techniques in Research and Industry ICXRI 2008*, 2^{nd}-6^{th} June 2008, X-Ray Application Malaysia Society (XAPP) and Universiti Malaysia Sabah, Malaysia.

BIOKOMPOSIT HIDROKSIAPATIT DENGAN PENGUAT ZIRKONIA UNTUK PENINGKATAN KEKUATAN DAN KELIATAN

ABSTRAK

Biokomposit zirkonia/hidroksiapatit (ZrO_2/HAp) telah dihasilkan dengan objektif untuk meningkatkan kekuatan patah dan keliatan patah bioseramik HAp monilitik. Y_2O_3-ZrO_2 komersil dan CaO-ZrO_2 disintesis telah dipillih sebagai bahan penguat di dalam matrik HAp. Kaedah pemprosesan seramik yang lazim, termasuklah pencampuran, penekanan dan pensinteran, telah digunakan untuk menghasilkan sampel ZrO_2/HAp. Pada mulanya, HAp yang diperkuatkan dengan nisbah berbeza ZrO_2 komersil (5, 10, 15 dan 20% berat) telah dihasilkan menggunakan penekan isostatik sejuk (CIP) dan kemudiannya disinter pada suhu 1050°C hingga 1250°C dalam persekitaran udara selama 5 jam. Keputusan menunjukkan sifat fizikal dan mekanikal merosot dengan penambahan Y_2O_3-ZrO_2 yang lebih tinggi. Nilai kekuatan patah dan keliatan patah adalah 73 MPa dan 0.85 MPa.$m^{1/2}$ untuk sampel 5% berat Y_2O_3-ZrO_2/HAP. Ini berpunca daripada penguraian HAp ke β-TCP. Bagi meningkatkan keupayaan pensinteran dan kestabilan fasa biokomposit ini, sedikit CaF_2 telah ditambahkan. Penambahan CaF_2 didapati mampu mengekalkan fasa HAp dan menghadkan penukarannya kepada β-TCP. Kekuatan patah dan keliatan patah optimum untuk 20% berat Y_2O_3-ZrO_2 dengan 5% berat CaF_2 adalah 126 MPa dan 1.9 MPa.$m^{1/2}$ manakala bagi 5% berat Y_2O_3-ZrO_2 dengan 5% berat CaF_2 nilainya adalah 151 MPa dan 1.4 MPa.$m^{1/2}$. Pada masa sama, kaedah penguraian polimer telah digunakan untuk mensintesis serbuk tetragonal CaO-ZrO_2 bersaiz nano. Kajian awal menunjukkan penambahan CaO-ZrO_2 mampu meningkatkan kekuatan dan keliatan HAp berbanding penambahan Y_2O_3-ZrO_2. Nilai diperolehi adalah lebih tinggi berbanding sifat optimum HAp monolitik.

ZIRCONIA REINFORCED HYDROXYAPATITE BIOCOMPOSITE FOR STRENGTH AND TOUGHNESS IMPROVEMENT

ABSTRACT

Zirconia/hydroxyapatite (ZrO_2/HAp) biocomposites were fabricated with the objective to improve the strength and fracture toughness of monolithic HAp bioceramics. Commercial Y_2O_3-ZrO_2 and synthesized CaO-ZrO_2 were selected as the reinforcement material for the HAp matrix. Conventional ceramic processing route was used to prepare the samples of ZrO_2/HAp, which involved mixing, compaction and sintering. Initially HAp reinforced with different ratios of commercial Y_2O_3-ZrO_2 (5, 10, 15 and 20 wt%) were compacted by using cold isostatic press (CIP) and subsequently sintered from 1050°C to 1250°C in air atmosphere for 5 hours. The results show that the physical and mechanical properties had deteriorated with higher amount of Y_2O_3-ZrO_2. The value in fracture strength and toughness was 73 MPa and 0.85 MPa.$m^{1/2}$ for the sample 5 wt% Y_2O_3-ZrO_2/HAP. This was due to decomposition of HAp to β-TCP. To improve sinterability and the phase stability of these biocomposites, small amounts of CaF_2 were added. Addition of CaF_2 was able to retain the HAp phase, limiting its transformation to β-TCP. The optimum flexural strength and fracture toughness values for 20 wt% Y_2O_3-ZrO_2 with 5 wt% CaF_2 addition was 126 MPa and 1.9 MPa.$m^{1/2}$, respectively while that for 5 wt% Y_2O_3-ZrO_2 and 5 wt% CaF_2, the values were 151 MPa and 1.4 MPa.$m^{1/2}$. At the same time, a polymer precursor decomposition method was adopted to synthesize nanosized tetragonal CaO-ZrO_2. Preliminary study with the incorporation of synthesized CaO-ZrO_2 without CaF_2 was found to be effective to improve the strength and toughness of the HAp as compared with addition of Y_2O_3-ZrO_2. The values obtained were relatively higher than the optimum properties of monolithic HAp.

CHAPTER 1

INTRODUCTION

1.1 Background and Problem Statement

Hydroxyapatite (HAp), a very important bioceramic, has been studied and used extensively in medical applications for repair or replacement of bone tissues in human body system. With the chemical formula $Ca_{10}(PO_4)_6(OH)_2$, hydroxyapatite is chemically similar to the mineral component of bone and teeth, and it will form an artificial bone-like structure with the surrounding bone tissue when implanted [Hench and Wilson, 1993]. The reason for using hydroxyapatite as a bone substitute material is because natural bone is approximately 70% hydroxyapatite by weight and 50% hydroxyapatite by volume [Shackelford, 1999].

HAp is frequently used for reconstruction and replacement of damaged bone or tooth zones in plastic and dental surgeries as well as in coatings on dental and orthopedic implants [Muster, 1992]. Metals coated with hydroxyapatite have also been introduced as artificial bones. The hydroxyapatite coating will help the surrounding tissue to bond firmly with the implant while the metal provides the strength for the artificial bone [Oonishi, 1991].

Hydroxyapatite is reported to have a calcium to phosphate ratio of 1.67:1, and is relatively insoluble, thus suitable for prosthetic applications [Akao et. al., 1981]. It has consistent bioactive properties and therefore is well suited as a calcium phosphate coating for total joint arthoplasty and total knee arthoplasty. HAp has also been shown to be biocompatible, nontoxic, and capable of bonding directly to bone, thus allowing for true osteointegration [Søballe and Friedman, 1996; Epinette, 1999].

However, sintered hydroxyapatite material exhibits relatively poor mechanical properties than natural bone. For instance, it has a low fracture toughness value, low strength and high brittleness. The fracture toughness (K_{Ic}) of the HAp ceramics does not exceed the value of about 1 $MPa.m^{1/2}$ as compared with 2-12 $MPa.m^{1/2}$ for human bone [Suchanek et. al., 1997; Qing-Liang et. al, 2004]. Because of these disadvantages, HAp are not viable as heavy-loaded implants. Their medical applications are limited to small unloaded implants, coatings and low loaded porous implants.

For full utilization of bioactive HAp based implants, the improvement of mechanical properties are necessary. One of the possible methods to improve these properties can be by the incorporation of reinforcing phases into HAp, i.e. as a composite. The addition of bioinert materials such as alumina (Al_2O_3) was found to improve the mechanical properties of hydroxyapatite [Champion et. al., 1996; Gautier et. al., 1996]. However, other researchers have found that adding Al_2O_3, an amphoteric material, had encouraged reaction with the HAp, forming tricalcium

phosphate and calcium aluminate and thereby reducing the mechanical properties [Ahmad Fauzi and Saud, 2004].

Other candidate material for the reinforcing phase for HAp is zirconia (ZrO_2) [Rapacz Kmita et. al., 2006]. Zirconia-based materials exhibit exceptional toughness due to martensitic transformation of tetragonal (t) to monoclinic ZrO_2 (m) ($t \rightarrow m$). Medical applications of ZrO_2 implants confirmed their satisfactory biocompatibility, although it cannot bond well to bone tissue. Nevertheless, it is also considered as a bioinert material [Park and Bronzino, 2003].

Zirconia is a well-known polymorph that occurs in three forms: monoclinic (m), tetragonal (t) and cubic (c). Pure zirconia is monoclinic at room temperature. This phase is stable up to 1170°C. Above this temperature it transforms into tetragonal and then into cubic phase above 2370°C. During cooling, a $t \rightarrow m$ transformation takes place in a temperature range of about 100°C below 1070°C. The phase transformation taking place while cooling is associated with a volume expansion of approximately 3-4%. Stresses generated by the expansion originate cracks in pure zirconia ceramics that, after sintering in the range 1500-1700°C, will break catastrophically into pieces at room temperature [Porter and Heuer, 1977; Piconi and Maccauro, 1999].

However, the addition of stabilizing oxides, like CaO, MgO, CeO_2 or Y_2O_3, to pure ZrO_2 allows generation of multiphase materials typically referred as Partially Stabilized Zirconia (PSZ), whose microstructure at room temperature generally consists of cubic zirconia with tetragonal zirconia precipitates. The precipitates may exist at grain boundaries or within the cubic matrix grains. The mechanical strength of PSZ was improved by a homogeneous and fine distribution of the tetragonal phase within the cubic matrix [Green et. al., 1989].

The tetragonal metastable precipitates finely dispersed within the cubic matrix were able to be transformed into the monoclinic phase when the constraint exerted on them by the matrix was relieved, i.e. by a crack advancing in the material. In that case, the stress field associated with expansion due to the phase transformation acts in opposition to the stress field that promotes the propagation of the crack. Enhancement in toughness will be obtained, because the energy associated with crack propagation is dissipated both in the $t \rightarrow m$ transformation and in overcoming the compression stresses due to the volume expansion.

With this background, this research was conducted to investigate the possibility of the improvement of HAp mechanical properties by using partially stabilized zirconia as the reinforcing phases. The transformation toughening mechanism in zirconia was expected to occur and thus improve the fracture toughness of HAp. The stability of the phases also needs to be studied carefully to maintain the biocompatibility of the system. Therefore, in this research, work was

focused to optimize the strength and fracture toughness of the reinforced hydroxyapatite biocomposite.

1.2 Objective of the Research

The objective of this research is to develop hydroxyapatite biocomposite with improved fracture toughness and strength with good physical properties and biocompatibility. With this main objective, the following studies were conducted:

1. Preparation of ZrO_2/HAp biocomposite bodies by conventional ceramic processing technique and characterizing the physical and mechanical properties. Commercial Y_2O_3-ZrO_2 and synthesized CaO-ZrO_2 were used as the reinforcing material.
2. Investigation of the bioactivity of Y_2O_3-ZrO_2/HAp biocomposite using simulated body fluid (SBF).

1.3 Project Overview

In this study, the conventional ceramic processing technique which involved wet mixing and dry pressing were selected. This technique was selected due to the simple process, low production cost, equipment availability and also easily adjustable parameters. Various parameters were investigated in preparing the ZrO_2/HAp biocomposite. The amount of ZrO_2 addition was varied in order to study the optimum composition of ZrO_2/HAp biocomposite. The sintering process was conducted in several different temperatures to investigate the most favorable sintering temperature to achieve optimum properties of the ZrO_2/HAp biocomposite which are comparable with human bone properties. The effect of ZrO_2 addition and sintering temperature to the phase stability of the HAp was also studied. Two types

of zirconia, i.e. commercial Y_2O_3-ZrO_2 and synthesized CaO-ZrO_2, were used in this study. The influences of both type of ZrO_2 were analyzed in this study.

The prepared ZrO_2/HAp biocomposite samples were characterized using X-Ray diffraction (XRD) for phase identification and scanning electron microscope (SEM) for topography and morphology analysis. The density and the shrinkage of the samples were measured, and the mechanical properties of the composite were determined by using Vickers hardness test and three point bending test.

CHAPTER 2

LITERATURE REVIEW

2.0 Introduction

Biomaterials are widely used to make devices to replace a part or a function of the body in a safe and biocompatible manner. The development and applications of biomaterials have been continually expanded since the first vanadium steel plates were introduced in 1912 by W.D. Sherman to help in the fixation of bone fractures [Park and Bronzino, 2003]. Over the years, various definitions of the term biomaterials have been proposed. Black (1999) defined biomaterials as materials intended to interface with biological systems to evaluate, treat or replace any tissues, organ, or function of the body. Other definition have simply defined biomaterials as synthetic materials used to make devices to replace part of a living system or to function in intimate contact with living tissue [Park, 2000].

Hydroxyapatite (HAp) is one of biomaterials that have received large attention due to its chemical composition which is similar to that of the mineral constituent of bone [Shackelford, 1999]. Bioactive properties of HAp allow the formation of tissue and bone-like layer on its surface hence encourages better fixation of implants [Chen et. al., 2002]. Therefore, many studies are continuously being carried out to investigate the properties of HAp and to expand its applications in wider usage as orthopedic implants.

This review starts with an overview about the structure and properties of bone tissue. Bioceramic materials will be described in more detail starting with its classifications and applications. Then the properties of HAp and other calcium phosphate materials are also the presented thoroughly in this chapter. This is followed by a review on ceramic biocomposite concept and explanation on the role of zirconia as a candidate for reinforcement phase for HAp. At the end, some evaluation techniques on the biological performance of materials are briefly explained.

2.1 Bone Tissue

Most bioceramics implants are in contact with bone, and therefore it is necessary to understand the characteristic of bone. Bone is a living material composed of cells and a blood supply covered by a strong, interwoven composite structure [Hench and Wilson, 1993]. Each bone organ is made up of bone tissue with cartilage tissue, other connective tissue and nerves. The major roles of bone are to serve as a support for the body's muscles, protection for some parts of the body, and also act as a reserve of calcium for the body [Krajewski and Ravaglioli, 1991].

2.1.1 Composition of Bone Tissue

Bone tissue is a composite made of 60 to 70% mineral substances and 30 to 40% organic tissues. This ratio holds true although the properties of a bone varies from point to point, and the ratio of the substances in the skeleton also differ from one bone organ to another [Krajewski and Ravaglioli, 1991]. The composition of bone depends on a large number of factors, e.g., species, age, sex, type of bone tissue and the location from which the sample is taken [Katz, 2000]. However, in general

there are three major components of bone tissue, and they are [Hench and Wilson, 1993]:

1. Collagen, which is flexible and very tough
2. Biological apatites, bone mineral, which is the reinforcing phase of the composite
3. Bone matrix or ground substance, which performs various cellular support functions.

Collagen is a natural, polymeric protein and the most important structural material in vertebratas. Collagen constitutes 36% of bone by weight, and the form of collagen found in bone is termed Type I. It is the dominant form throughout the body [Shackleford, 1999]. Collagen, which can be considered as the matrix of bone, is a triple helix structure and each strand being made up sequences of amino acids [Meyers et. al., 2008]. Some data for the composition of adult human and bovine cortical bone are given in Table 2.1.

Table 2.1 Composition of adult human and bovine cortical bone [Katz, 2000]

Species	% H_2O	% Apatite	% Dry weight collagen
Bovine	9.1	76.4	21.5
Human	7.3	67.2	21.2

Biological apatites, which are the mineral phases of bone tissue, are usually referred to as calcium hydroxyapatite, although biological apatites are usually calcium deficient and carbonate substituted [LeGeros and LeGeros, 1993; Landi et. al., 2003]. Small amounts of magnesium, fluoride, carbonate, chloride and potassium

are also found in the mineral substance of bone. Table 2.2 shows the comparative composition of human enamel and bone.

Table 2.2: Enamel and bone component of the human adult [LeGeros and LeGeros, 1993]

Constituent	Enamel	Bone
Calcium, Ca^{2+}	36.0	24.5
Phosphorus, P	17.7	11.5
Ca/P molar ratio	1.62	1.65
Sodium, Na^{+}	0.5	0.7
Potassium, K^{+}	0.08	0.03
Magnesium, Mg^{2+}	0.44	0.55
Carbonate, CO_3^{2-}	3.2	5.8
Fluoride, F^{-}	0.01	0.02
Chloride. Cl^{-}	0.30	0.10
Total inorganic	97.0	65.0
Total organic	1.0	25.0
Absorbed H_2O	1.5	9.7

2.1.2 Structure and Properties of Bone Tissue

The complexity of bone's properties comes from the complexity of its structure [Katz, 2000]. Therefore it is important to understand the bone structure in order to analyze the properties. There are two principal types of bone, i.e. cancellous bone and cortical bone. Cancellous bone, also called trabecular or spongy bone, is less dense than cortical bone (compact bone). Cortical bone is found in long bones (femur, tibia, fibula), while cancellous bone is found in the core of bones, in flat bones and occurs across the ends of the long bones [Meyers et. al., 2008].

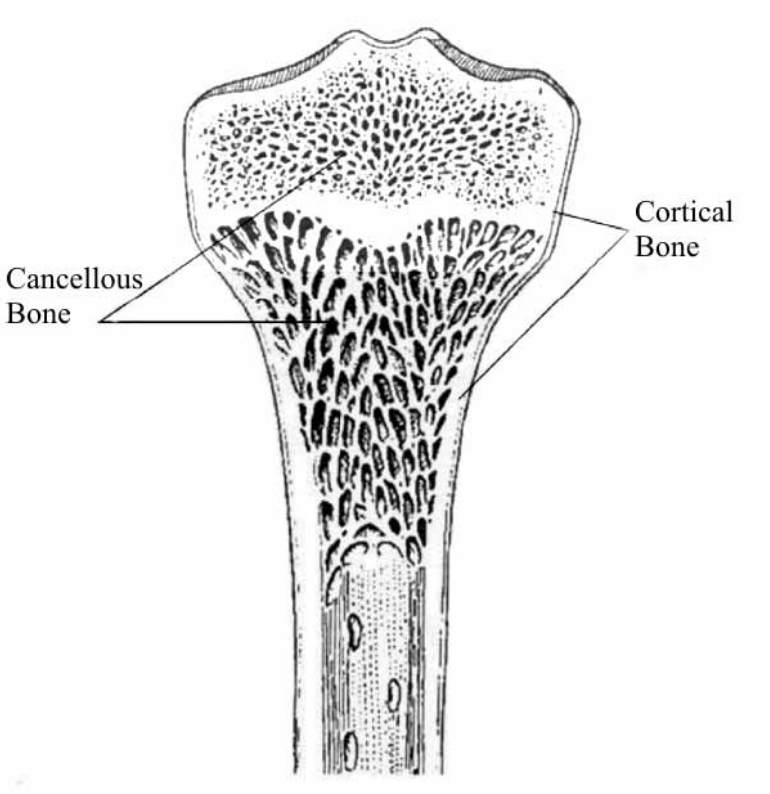

Figure 2.1: Longitudinal section of a human femur [Meyers et. al., 2008]

Figure 2.1 shows the structure of a long bone. The surface regions consist of cortical bone, while the inside is porous and is called cancellous bone. The porosity reduces the strength of the bone, but also reduces their weight. Bones are shaped in such a manner that strength is provided only where it is needed. The porosity of cancellous bone provides interesting mechanical properties. Table 2.3 gives some information about mechanical properties of bone tissues. The mechanical strength is determined by the porosity and the manner in which the porosity is structured, therefore cortical bone have higher strength compared to cancellous bone. Cancellous bone has a lower modulus of elasticity and higher strain to failure than cortical bone and this is due to its lower density. The pores also perform other physiological functions and contain the marrow [Meyers et. al., 2008].

Table 2.3: Mechanical properties of bone tissues [Hench and Wilson, 1993]

Property	Cortical Bone	Cancellous Bone
Compressive Strength (MPa)	100-230	2-12
Flexural Strength (MPa)	50-150	10-20
Strain to failure	1-3	5-7
Young's Modulus (GPa)	7-30	0.5-0.05
Fracture Toughness, K_{IC} ($MPa.m^{1/2}$)	2-12	N.A.

The variation of the properties values of the bone tissue is dependent upon age, location of the bone and direction of measurement. The maximum compressive strength of cortical bone is around 200 MPa, and it will show a progressive loss of strength with age. Figure 2.2 illustrates that the increase in age would lead to a decrease in strength. Hence, elderly people will have higher risk of bone fracture when they fall.

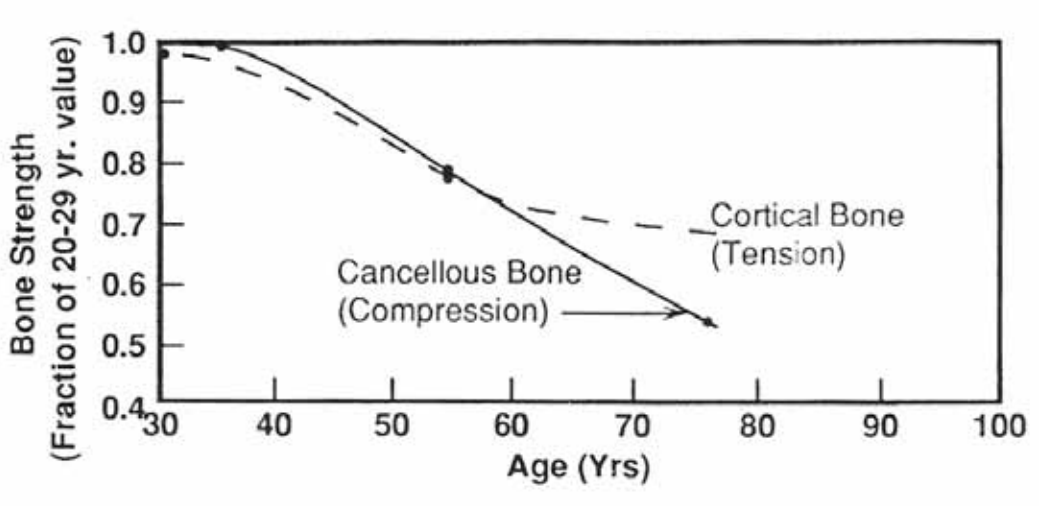

Figure 2.2: Effect of age on strength of bone [Hench and Wilson, 1993]

2.2 Ceramics and Bioceramics Implants

Ceramic is defined as the art and science of making and using solid articles that have as their essential component being inorganic nonmetallic materials [Kingery et. al., 1976]. Ceramics are refractory, polycrystalline compounds, usually inorganic, including silicates, metallic oxides, carbides, nitrides, and various refractory hydrides and sulfides.

Unlike metals and polymers, ceramics are difficult to shear plastically due to the ionic nature of the bonding and the minimum number of slip systems. Ceramics are nonductile materials and are generally hard. Ceramics are very susceptible to microcracks because, instead of undergoing plastic deformation, they will fracture elastically on initiation of crack. At the crack tip the stress could be many times higher than the stress in the material. Other characteristics of ceramic materials are their high melting temperatures and low conductivity of electricity and heat. These characteristic are due to the chemical bonding within the ceramics [Park and Lakes, 1992].

During the last couple of decades, ceramics has been use to improve the quality of human life. Innovative techniques for fabricating ceramics have led to their use as advance materials. Specially designed and fabricated ceramics for the repair and reconstruction of diseased, damaged or worn out parts of the body are widely developed. Ceramics used for the purpose to improve the quality of human life are called *bioceramics*. Most clinical applications of bioceramics are related to the repair of the skeletal tissues, composed of bones, joints and teeth, and to augment both hard

and soft tissues. Ceramic are also used to replace parts of the cardiovascular system, especially heart valves [Hench and Wilson, 1993].

In order to be classified as a bioceramic, the ceramic material must meet or exceed the following desired properties of implantable bioceramics [Bilotte, 2003], that is they should be:

1. Nontoxic
2. Noncarsinogenic
3. Nonallergic
4. Noninflammatory
5. Biocompatible
6. Biofunctional for its lifetime in the host

Although bioceramics seem to be a suitable candidate for implant materials, the natural problem in conventional ceramics are also the challenge that inhibits the application of bioceramics. Primary drawbacks of bioceramics are their brittleness, high Young modulus, and inferior workability. Brittleness becomes a significant problem when bioceramics are used in positions with high stress loading. Stress shielding is observed when bone tissue is bonded with materials with high Young's modulus, where the bone around the materials is resorbed [Ishikawa et. al., 2003]. Because the bioceramics workability is not good, it is difficult for the surgeon to shape the bioceramics when necessary during the surgery [Shackelford, 1999].

2.2.1 Classification of Bioceramics

Bioceramics can be classified independently in terms of their behavior in the body environment. Such a classification method is especially useful and relevant since bioceramics are used in human body. This method classifies bioceramics as relatively bioinert or nonabsorbable, bioactive or surface reactive, and biodegradable or bioresorbable ceramics [Bilotte, 2003]. It is also critical that any bioceramics implant avoid a toxic response that kills cells in the surrounding tissues and can cause systemic damage to the patient [Hench and Wilson, 1993].

The fundamental principle in the classification of bioceramics is based on chemical reactivity with the physiological environment. Bioinert ceramic, such as alumina (Al_2O_3), have a tendency to exhibit inherently low levels of reactivity which peak on the order of 104 day (over 250 days). Bioactive ceramics have a substantially higher level of reactivity peaking on the order of 100 days, while bioresorbable ceramics have even higher levels of reactivity peaking on the order of 10 days as shown in Figure 2.3 [Hench, 1991].

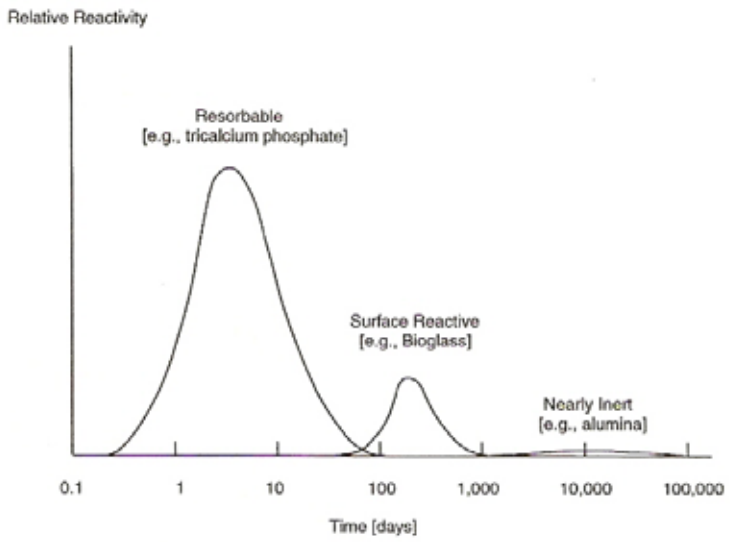

Figure 2.3: Bioactivity spectrum for various bioceramics implants [Hench, 1991].

2.2.1.1 Bioinert ceramics

Bioinert ceramics are defined as ceramics that are stable in human body and do not show harmful response or bioactivity. Bioinert ceramics maintain their physical and mechanical properties while in the host [Bilotte, 2003]. In general, the living body recognizes artificial materials as foreign materials when the latter are implanted in the living body and consequently the artificial materials are covered with fibrous layer. The thickness of fibrous layer depends on the tissue compatibility of the implanted materials. Biologically inactive, nearly inert ceramics (like alumina and zirconia) will elicit thin fibrous capsule at their interface, while metals and most polymers may cause a total encapsulation of the implant within the fibrous layer. For an implant made of bioinert materials, a tight mechanical fit with the host tissue is very important in order to prevent interfacial movement and subsequent clinical failure [Hench and Wilson, 1993].

Example of relatively bioinert ceramics are dense and porous alumina, zirconia ceramics, and a single-phase calcium aluminates. Bioinert ceramics are usually applied as structural support implants, such as bone plates, bone screws, and femoral heads. Examples of a non-structural support uses are ventilation tubes, sterilization devices, and drug delivery devices [Bilotte, 2003].

2.2.1.2 Bioresorbable Ceramics

Bioresorbable or biodegradable ceramics degrades upon implantation in the host. The resorbed material is replaced by endogeneous tissues. Instead of replacing the tissues, the material encourages the regeneration of tissues to take their place, and the degradation rate varies from material to material. Examples of bioresorbable

ceramics are aluminum calcium phosphate, coralline, plaster of paris, and beta-tricalcium phosphate (β-TCP) [Bilotte, 2003].

There are a few issues or criteria that need to be taken into consideration in order to successfully implant the bioresorbable ceramics. The constituent of the materials need to be of a composition that can be broken down chemically by body fluids. The degradation products must also be non-toxic chemical compounds that can be easily disposed of without damaging the surrounding cells or harming the host's health. Moreover, the resorption rate of the materials must also match the restoration rate of the tissues even as the material provides a sufficient mechanical strength to support the host tissue while the regeneration of tissues takes place [Hench and Wilson, 1993].

2.2.1.3 Bioactive ceramics

Bioactive ceramics are defined as the material that bond directly with bone without having fibrous connective tissues between them. Upon implantation in the host, bioactive ceramics form a strong bond with adjacent tissue. This characteristic of direct bonding is extremely useful, especially when the biomaterial is used in an area where the material will be bonded with bone.

In order for an implant to perform optimally, its properties, such as a controlled rate of chemical reactivity, morphology and phase composition need to be carefully engineered according to its function and rate of bonding to the host tissue. A small change in composition can change the properties of the bioceramics from nearly inert to bioresorbable to bioactive. In bioactive ceramics, unlike bioresorbable

ceramics, chemical reactions only occur at the surface while the rest of the implant remains largely stable [Hench and Wilson, 1993]

Examples of bioactive ceramics are dense porous glasses, Bioglass and Ceravital, and hydroxyapatites. One of their many applications is the coating of metal prostheses. This coating provides a stronger bonding to the adjacent tissues, which is very important for prostheses [Bilotte, 2003]. It is still not fully explained why bioactive ceramics bond directly with bone. One hypothesis is that apatite shows good osteoconductivity since it is an excellent adsorbent, and thus adsorbs necessary factors including the adhesion factor that is required for the wandering osteoblast [Ishikawa et. al., 2003].

2.2.2 Applications of Bioceramics

Bioceramics are produced in a variety of forms and phases and serve many different functions in repair of the body. These are summarized in Figure 2.4. In many applications, bioceramics are used in the form of bulk of a specific shape called implants, prostheses, or prosthetic devices. Bioceramics are also used to fill space while the natural repair processes restore function. In other applications, bioceramics are used as coating on a substrate, or a second phase in a composite, combining the characteristics of both materials into a new material with enhanced mechanical and biochemical properties [Hench and Wilson, 1993].

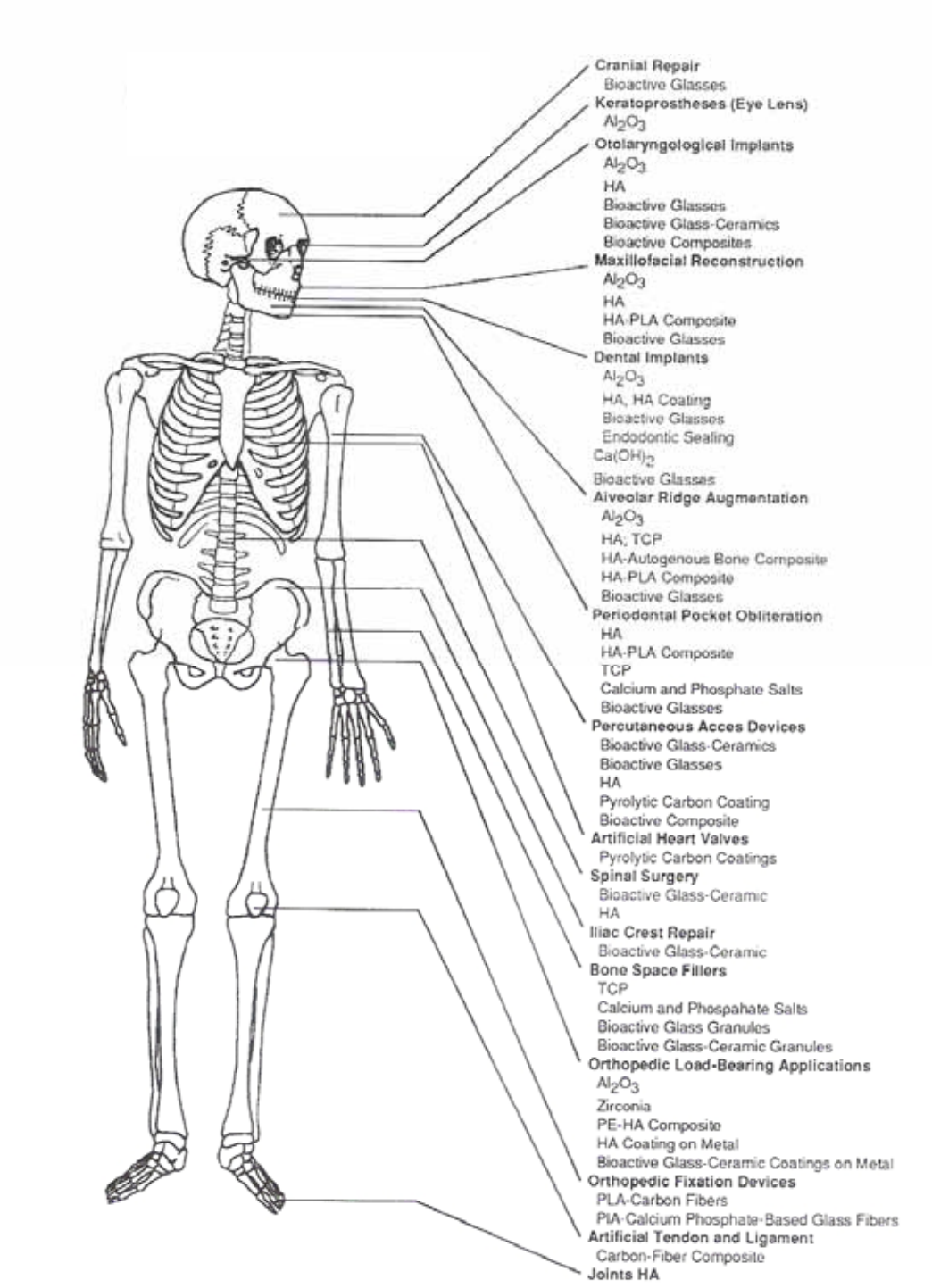

Figure 2.4: Clinical uses of bioceramics [Hench and Wilson, 1993].

2.3 Calcium Phosphate (CaP) Bioceramics

Calcium phosphate (CaP) has been synthesized and used for manufacturing various forms of implants, as well as for solid coatings on other implants. There are several forms of calcium phosphate and they typically form a wide group of compounds called apatite. It is said that an Australian mineral researcher Werner named a mineral as apatite in 1786 based on the Greek word “apataw”, which means puzzled, since it was confused with several other similar looking minerals.

Calcium phosphate bioceramics include hydroxyapatite from synthetic, natural (from coral) and biological (from bovine bone) origin, calcium-deficient apatite (CDA), tricalcium phosphate (β-TCP), biphasic calcium phosphate (BCP) and calcium phosphate cements consisting of mixtures of different CaP phases, e.g., β-TCP, tetracalcium phosphate (TTCP), monocalcium phosphate monohydrate (MCPM), dicalcium phosphate dehydrate (DCPD or brushite), dicalcium phosphate anhydrous (DCPA or monetite) and octacalcium phosphate (OCP) [LeGeros and LeGeros, 1996].

The types of calcium phosphate phase formed depend on the Ca/P ratio, presence of water, impurities and temperature [Hench, 1998]. For example, hydroxyapatite is more likely to form in a wet environment and at lower temperature ($<900^{o}C$), while in a dry atmosphere and at higher temperature, β-TCP will be formed [Park and Lakes, 1992]. Table 2.4 gives the summary of various calcium phosphate ceramics and their Ca/P ratio.

Table 2.4: Calcium phosphate ceramics and their applications [LeGeros and LeGeros, 2003]

Calcium Phosphate	Ca/P	Applications
Monocalcium phosphate monohydrate, MCPM $Ca(H_2PO_4)_2$	0.5	Cements, polyphosphates
Dicalcium phosphate dehydrate, DCPD $CaHPO_4.2H_2O$	1.0	Cements, coating
Dicalcium phosphate anhydrous, DCPA $CaHPO_4$	1.0	Cements, coating
Octacalcium phosphate, OCP $Ca_8H_2(PO_4)_6.\ 5H_2O$	1.33	Coatings, bonegraft
Alpha tricalcium phosphate, α-TCP $Ca_3(PO_4)_2$	1.50	Cements
Beta-tricalcium phosphate, β-TCP $Ca_3(PO_4)_2$	1.50	Cements, bonegraft composites
Calcium-deficient apatite, CDA $Ca_3(PO_4)_2$	<1.67	Bonegraft, cements, composites
Hydroxyapatite, HA $Ca_{10}(PO_4)_6(OH)_2$	1.67	Bonegraft, coatings
Biphasic calcium phosphate, BCP HA/β-TCP	1.55-1.66	Bonegraft, coatings
Fluorapatite, FA $Ca_{10}(PO_4)_6F_2$	1.67	Bonegraft, coating
Carbonate hydroxyapatite $(Ca,Na)_{10}(PO_4,CO_3)_6(OH)_2$	1.7-2.6	Bonegraft
Tetracalcium phosphate, TTCP $Ca(PO_4)_2O$	2.0	Cements

2.3.1 Hydroxyapatite

Hydroxyapatite (HAp) is the dominant inorganic component if the hard tissue of the human body, such as dental and bone, and it represent the bioactive ceramics. The term hydroxyapatite is used generally to represent $Ca_{10}(PO_4)_6(OH)_2$ in the field of biomaterials, although the term apatite originally indicates a much wider composition [Shackelford, 1999].

Although the term HAp with the stoichiometric chemical formula, $Ca_{10}(PO_4)_6(OH)_2$, are used generally for simplicity, apatite found in enamel, dentin, and bone show more complex composition [Ishikawa et. al., 2003]. In spite of a large variety of compositions, calcium hydroxyapatite belongs to the hexagonal system, with a $P6_3/m$ space group. The unit cell contains a complete representation of the apatite crystal, consisting of Ca^{2+}, PO_4^{3-}, and OH^- groups closely packed together in arrangement shown in Figure 2.5. The substitute ion of OH^- with fluoride gives HAp greater chemical stability due to the closer coordination of fluoride (symmetric shape) as compared to hydroxyl (asymmetric) by the nearest calcium. This is why fluoridation of drinking water helps in resisting caries of the teeth [Park and Lakes, 1992].

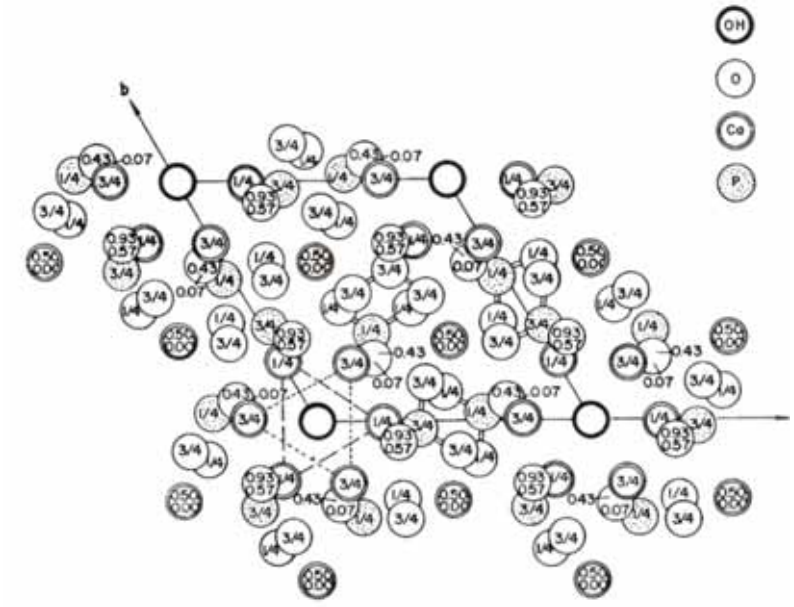

Figure 2.5: The atomic arrangement of calcium hydroxyapatite [Hench and Wilson, 1993].

The theoretical composition of pure HAp is 39.68 wt% Ca and 18.45 wt% P with Ca/P molar ratio of 1.667. In dense HAp material, variations in Ca/P ratios after sintering are reflecting the β-TCP/HAp ratios which in turn reflect the purity. If the Ca/P is 1.67, only HAp will be observed in the X-ray diffraction. If the Ca/P ratio is lower than 1.67, β-TCP and other phases such as tetracalcium phosphate (TTCP) will be present with the HAp phase, depending on the temperature and condition of sintering. If Ca/P is higher than 1.67, CaO will be present with the HAp phase [Hench and Wilson, 1993].

2.3.1.1 Hydroxyapatite as an implant materials

Its similarity to the inorganic phases of bone and teeth makes hydroxyapatite a very attractive candidate for implant materials. Hydroxyapatite is often used as bone substitute because of its superior biocompatibility, showing a capacity for direct

chemical bonding with bone [Muster, 1992]. Hydroxyapatite is not resorbed in the human body upon implantation and will stay in the human body while positively influencing bone formation and thus speeding up recovery [Ishikawa et. al., 2003].

Hydroxyapatite is often used as coatings on metallic implants, usually titanium, titanium alloys, and stainless steels, due to its surface reactivity. The body responds to metallic implants by surrounding it with fibrous tissues in order to isolate it. The coating of HAp provide a bioactive surface which allows the implant to bond with the bone tissues while retaining the mechanical properties of the metal within [Oonishi, 1991, Lim et. al., 1999]. Hydroxyapatite is also commonly applied as filler materials and scaffolds for bone reconstruction. The need for these fillers occur when there is a bone loss due to diseases or accident or when bone augmentations or replacement of fragments on non-loaded bones are required, such as ridge augmentations, ear implants and repair of periodontal defects [Muster, 1992]. Hydroxyapatite acts as a frame or scaffold that facilitates the growth of tissues across the void. It is readily incorporated into the bone structure even as it encourages bone growth and speeds up the healing process [Akao et. al., 1981].

2.3.1.2 Mechanical properties and manufacturing techniques of hydroxyapatite

In order to fabricate HAp bioceramic implants with desired shapes and properties, researchers investigates the use of conventional and advanced ceramic manufacturing techniques. The techniques used to fabricate HAp will depend on the application of the implants and the desired properties. For applications as hard tissue replacement, the most important and immediate property is the strength since the