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Evaluation of Osseointegration Between Two Different Modalities of Hydroxyapatite Implant Surface Coatings: Plasma Sprayed HA Coated Implants and Electrophoresis Deposited Nano HA Coated Implants

Audrey Marie Selecman
University of Tennessee Health Science Center

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EVALUATION OF OSSEOINTEGRATION BETWEEN TWO DIFFERENT MODALITIES OF HYDROXYAPATITE IMPLANT SURFACE COATINGS: PLASMA SPRAYED HA COATED IMPLANTS AND ELECTROPHORESIS DEPOSITED NANO HA COATED IMPLANTS

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Abstract

Purpose: The objective of this study was to evaluate the osseointegration between plasma sprayed (PS) hydroxyapatite (HA) coated implants and electrophoresis deposited (EPD) nano hydroxyapatite coated implants in an animal model. The quantity of osseointegration was inferred by interfacial strength and percent tissue-contact length. **Materials and Methods**: Thirty-six cylindrical implants (18 PS and 18 EPD) were placed into dog mandibles. Nine implants from each group were evaluated with pull out testing and histological studies at 12 weeks after implantation. The remaining implants were occlusally loaded for 9 months and then evaluated by pull out testing and histological studies. **Results:** Twelve weeks after implantation, no statistical difference in pull out strength was observed between PS HA (70.43 ± 16.65 lbf) and EPD nano HA (86.35 ± 28.07 lbf) coated implants . After loading for 9 months, it was observed that the interfacial strength of implants coated by EPD nano HA (199.9 ∀ 35.1 lbf) was statistically higher ($P < 0.028$) than the PS HA coated implants (121.14 ± 38.45 lbf). At 12 weeks implantation, no statistical significant difference in bone contact length was observed between EPD nano HA (97.6 ± 3.2%) and PS HA coated implants (95.6 ± 4.6%). No statistical significant difference in tissue-contact length was observed between EPD nano HA(91.8 ± 8.2%) and PS HA coated implants (84.3 ± 7.2%) after loading for 9 months. **Discussion**: The advantages of PS HA coatings are widely recognized, but little data exists on rate and quantity of osseointegration relative to newer coating methodologies such as electrophoresis deposition. EPD nano HA shows promise as a HA implant coating process, achieving higher interfacial strength than the PS HA coating after 1 year in this study. **Conclusion:** Histological data and interfacial strength suggest that PS and EPD nano HA coated implants achieve similar bone responses short term. However, EPD nano HA coated implants attained a higher interfacial

strength after 9 months of loading, even though histological data was not statistically different from that of PS HA coated implants.

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Chapter 1. Introduction

1.1. Early History of Dental Implants

Tooth loss due to disease and trauma is as primitive as reality of replacement of teeth. Modalities of tooth replacement have evolved and today are only limited by available dental materials and technologies. The debilitating effects of tooth loss on general health and appearance were appreciated even in ancient times. Throughout history, teeth have been replaced, shaped, and veneered to show wealth and power as well as to enhance beauty. Ancient Incas and Egyptians were known to tap precious metals, sea shells and even human teeth into the jaw to replace missing teeth. Ancient Etruscans replaced missing teeth with the teeth of oxen or calves and even practiced dental prosthetics using gold bands affixed to artificial replacements (Ring 1985).

Dentistry as a profession was recognized in the Middle Ages. During the Early Middle Ages in Europe, monks were educated to practice medicine, surgery, and dentistry. However, from 1130 to 1163, a series of Papal edicts prohibited monks from performing any type of surgery. It was the barber who often assisted monks in their surgical ministry that assumed the monks' surgical duties: bloodletting, lancing abscesses, extracting teeth, etc (Ring 1985).

In 1210, the Guild of Barbers was established in France. Barbers eventually evolved into two groups: surgeons who were educated and trained to perform complex surgical operations; and lay barbers, or barber-surgeons, who performed more routine hygienic services including shaving, bleeding and tooth extraction (Ring 1985).

In 1809, the French dentist Maggiolo published *Le Manuel de l'Art du Dentiste.* He described how he cast a gold root-shaped implant, inserted it into a fresh extraction socket, and then attached a prosthetic tooth after a certain healing period (Maggiolo 1809). Sixty years later,

S.M. Harris described a technique in which he artificially created a socket to implant a roughed lead metal coated porcelain post (Dental Cosmos 1913). In 1886, Edmunds was the first clinician in the US to implant a platinum disc into the jawbone, to which he attached a porcelain crown. His efforts were accounted for at the First District Dental Society of New York (Ring 1995). Based on these and several other observations, many more implantation attempts were made, experimenting with different metal alloys and porcelain formulations; however, on the whole, the long-term success rates were still very poor. Modern dental treatment focuses on preservation of the oral hard and soft tissues as well as the replacement of defective or lost dentition. Because tooth loss is as common and problematic as it was thousands of years ago, considerable research has focused on the prosthetic replacement of teeth.

1.2. Modern History of Dental Implants

In 1937, Alvin Strock inserted the first dental screw implant made of cobalt-chromium-molybdenum alloy. This first successful biocompatible implant metal, otherwise known as vitallium, had been developed a year earlier by Charles Venable, an orthopedic surgeon. These were among the first somewhat successful oral implants. Strock published a paper on the physiological effects of vitallium in bone. He noted that implants made of vitallium had few post-operative complications or reactions in test animals and humans. Histologic sections demonstrated remarkable tissue tolerance to the vitallium implants. Strock documented long term success up to 15 years, until he passed away (Strock 1939, Strock 1949, Cranin 2006).

From the mid 1930's, implant concepts developed which formed the foundation for today's variety of implant modalities. Efforts to improve osseointegration influenced the fabrication of many implant geometries. These designs include subperiosteal, transosseous and endosteal implants.

The subperiosteal implant design is a nonosseointegrated, metal framework that rests beneath the mucoperiosteum of the jaw bone. The framework has posts which pierce the mucosa in order to attach to an overdenture. Problems associated with subperiosteal implants include damage to underlying bone, infection, difficulty of removal, and epithelial down growth with subsequent dehiscence of the frame (Worthington 1994). A retrospective study at the University of Southern California recorded follow up, maintenance, and treatment of complications in 81 mandibular subperiosteal implants placed at the Advanced Prosthodontic Clinic for periods up to 21 years. The 10-year survival rate for subperiosteal implants was calculated at 79% for 63 patients, and a 15-year survival rate of 60% was calculated for 34 patients. Researchers concluded that subperiosteal implants' long term survival rate declines over time. And although subperiosteal implants long term survival rates were poor, they were successful in providing support and retention to patients that that could not otherwise tolerate dentures (Yanase *et al.* 1994).

Transosseous implants are a surgically invasive implant design in which bicortical posts are inserted through a submental incision under general anesthesia. A common design is the transmandibular staple. The staple has a plate which fits against the lower border of the mandible. It is stabilized by posts that penetrate to the superior border of the mandible. These posts attach to an overdenture (Worthington 1994). Meijer *et al.* (1998) evaluated mandibular staple implants in a long-term retrospective study that included fifty-six edentulous patients over a mean evaluation period of 103 months. Results indicate a survival rate of 91%. Though long term success rates are comparable to other more common implant modalities, less invasive implant systems such as endosseous implants are preferred by operator and patient.

Endosseous implants as a whole have experienced an evolution of designs. Such designs include the blade and root shaped cylinder or threaded forms. The most common geometry used

today is the root shaped threaded implant. Root shaped implants can be placed in the maxilla or mandible and have vast restorative capabilities with good long term success rates (Worthington 1994). The long term survival of endosseous implants has been attributed to its ability to become osseointegrated. The process of osseointegration and implants was first described in the literature by Brånemark *et al.* in 1981. Initial research suggested direct bone-to-implant interface contact at the electron microscopic level. They concluded than osseointegrated implants were a reliable, cement-free anchor for permanent prosthetic substitutes (Albrektsson *et al.*1981).

1.3. Osseointegration

The mechanism of attachment of the implant to the bone has been evaluated though a variety of materials and geometries. For some years, clinicians sought to develop an analogue of the periodontal ligament to attach implants to bone. But as previous clinical evidence would hold true, replicating nature to artificially replace teeth is not always possible. The connective tissue sheath created in preliminary research histologically lacked organization and specialization necessary to truly function as the periodontal ligament. Loading implants sheathed in connective tissue increased the fibrous tissue layer, subsequently loosening the implant to failure (Weiss 1990).

In 1952 that Swedish professor Per-Ingvar Brånemark began his work on tissue integrated prostheses. His objective was to understand the healing and reparative capacity of hard and soft tissue in order to obtain a predictable tissue response to implant therapy. The pursuit of osseointegration and dental implants stemmed from microscopic studies bone marrow of the rabbit fibula. Intravascular examination revealed the intimate circulatory connection of bone and marrow during regeneration after surgical insult. In addition, mechanical, thermal, chemical and rheological tissue injury was reported in order to develop surgical procedures with

predictable healing of differentiated tissues (Brånemark 1983).

In the early 1960's, Brånemark implanted screw shaped titanium chambers into animals. It was noted that chambers were "inseparately incorporated" into bone. Microscopic examination revealed bone had actually grown into the surface pores of the titanium. Separate studies evaluated healing and anchorage stability related to various sizes and designs of titanium implants. Brånemark found that titanium fixtures implanted into the marrow cavity would form a shell of cortical bone devoid of a fibrous tissue interface around the implant. This direct bone to implant interface between vital bone and the screw-shaped titanium implants was proven to achieve and maintain bone anchorage under loading. Brånemark and associates described this phenomenon as *osseointegration*, to describe the "direct structural and functional connection between ordered, living bone and the surface of a load carrying implant." In 1965, the concept of osseointegration and dental implants was proposed to maintain a dental prosthesis in an edentulated human (Albrektsson ***et al.*** 1981).

In 1983, Brånemark described osseointegration and its experimental background. Brånemark subsequently demonstrated that with a carefully controlled surgical technique, titanium could be predictably implanted and integrated into vital bone. In dentistry, these findings implemented a new direction of implant experimental design and research.

1.4. Bone Physiology

Endosseous implants rely on bone for support and anchorage. Therefore it is important to understand the physiology, structure and metabolism of unaltered bone as well as its response to surgical, healing and post operative conditions. Bone is connective tissue that consists of a complex composite of cells and extracellular matrix. Bone cells include osteocytes, osteoprogenenitor cells, osteoblasts and osteoclasts. Each of these cells is related to different

functions of cell activation, matrix maturation, mineralization and remodeling. The extracellular matrix contains type I collagen and ground substance. Ground substance is an aggregate of proteoglycans and noncollagenous glycoproteins that become mineralized. These minerals include calcium and phosphorus and are stored as hydroxyapatite crystals [Ca10 (PO4)6 (OH)3] (Roberts 1987). Through mineralization, bone obtains its rigid structure and function: support and protection of the musculoskeletal system and a physiologic reservoir of calcium and phosphorus. Other connective tissues included in bone are hemopoietic, vascular, neural and adipose tissue as well, hyaline and articular cartilage (Whitson 1998).

Bone is classified in several ways. Clinically, bone can be recognized by density and structure. Dense *compact bone* forms an outer shell for an inner meshwork of anastomosing spicules of *cancellous bone*. Its shape further identifies bone as long bones, short bones, flat bones or irregular bones (Whitson 1998).

Microscopically, bone can be subdivided into woven and lamellar bone. *Woven bone* is formed during growth and development and in response injury. It consists of highly cellular osseous tissue with low mineral content, random fiber orientation and minimal strength. It can be formed at a rate of 30-50um/day or more and is replaced with lamellar bone. Woven bone is essential to initial healing of endosseous implants. *Lamellar bone* is the mature load bearing bone that comprises most of the adult skeleton. This bone is densely mineralized and highly organized into three distinct structural units. These units create the Haversian system which encase the neural and vascular tissue of bone. *Circumferential lamellae* form a perimeter of compact bone that encases concentric and interstitial lamellae. *Concentric lamellae* encase Haversian canals to form osteons. *Interstitial lamellae* are irregular shaped remnants of concentric lamellae (Ross *et al.* 1995).

Although osseointegration occurs with predictable success, biological understanding of

early cellular events leading to osseointegration of implants is currently deficient. Histomorphometric and histologic analyses are often from animal models (Clokie and Warshawsky 1995, Sleats *et al.* 2006). However, the basic understanding of bone remodeling helps to describe the process of osseointegration.

Remodeling describes the dynamic ability of old or damaged bone to be replaced by new bone. The rate of turnover is approximately 5% for cortical bone and up to 15% for trabecular bone. Old bone is resorbed by osteoclasts which form a cutting cone. Osteoblasts differentiate behind the cutting cone to form the filling cone. Osteoblasts in the filling cone synthesize osteoid which becomes the mineralized osteons. Osteoblasts differentiate into osteocytes during mineralization and maturation of bone (Whitson 1998). The surgical placement of implants damages bone and initiates bone remodeling.

Block *et al.* (1997) divides bone healing and remodeling after implant surgery into three phases: the inflammatory, proliferative and maturation phase. Each phase is distinct, but there is overlap of specific occurrences in transition from one phase to another. The inflammatory phase occurs in the first ten days after surgical insult and is characterized by adsorption of plasma proteins, platelet aggregation, clotting cascade activation, cytokine release, nonspecific cellular inflammatory response, macrophage-mediated inflammation. The proliferative phase occurs from day 3 to 42 and is characterized by neovascularization, differentiation, proliferation and activation of cells and production of immature connective tissue matrix. The maturation phase occurs after day 28 and is characterized by remodeling of the immature bone matrix with coupled resorption and deposition of bone, in response to implant loading and physiologic bone recession.

Slaets *et al.* (2006) describes early cellular responses to titanium implants placed in the rabbit tibia. Upon placement of an implant, a coagulum of blood fill the fills the microgap

between the implant and bone. One week after insertion, osteoclasts and osteoblasts were observed at the bone surface. The osteocytic lacunae of the damaged bone appeared to be devoid of cells for up to 28 days ($P < 0.05$) after implant insertion. This region of altered nuclear morphology was accompanied by an invasion of basic multicellular units (BMUs) that initiated bone remodeling, which reached its maximum after 4 weeks ($P < 0.05$) but was ongoing 6 weeks after implant insertion. Intensive bone remodeling resulted in the formation of new bone, eventually leading to the osseointegration of the implant. Understanding bone remodeling and the capacity in which bone can heal is fundamental to formulating new implants that predictably integrate with reduced healing time.

1.5. Endosseous Implants

Endosseous implants have become a paramount option in the treatment planning of single tooth replacement, fixed partial dentures, removable partial and complete dentures, and prosthetic maxillofacial reconstruction. As implant technology evolves, so must the understanding of the biologic principles that control the ability of the body to tolerate and maintain the artificial replacement of lost tooth structure and adjacent periodontium.

1.5.1. Composition

In order to achieve success, implants must be made of a material that is biocompatible. Accepted materials used in the fabrication of implants are classified into three groups: metals, ceramics, and polymers. Each material can be further identified by is biocompatibility as bioinert, biotolerant, and bioactive. Bioinert materials allow close apposition on their surface, leading to contact osteogenesis. Materials considered bioinert include ceramics such as aluminum oxide and zirconium oxide, and metals such as commercially pure titanium and

titanium alloys. Biotolerant materials are not rejected when implanted in tissue, but are surrounded by a fibrous layer to form a capsule. Biotolerant metals and polymers encompass gold, cobalt-chromium alloys, stainless steel, zirconium, niobium, tantalum, polyethylene, polyamide, polymethylmethacrylate and polytetrafluorethylene. Bioactive materials allow for formation of new bone on their surface, exchanging ions to create a chemical bond. Polymers used as bioactive implant materials include hydroxyapatite, tricalcium phosphates, tetracalcium phosphates, calcium phosphates, fluorapatite, brushite, carbon: vitreous and pyrolytic, carbon-silicon and bioglass. Bioactive and bioinert materials are osseoconductive, meaning they form a scaffold for bone ingrowth, making them prevalent dental implant research and production (Sykaras *et al.* 2000).

The magnitude of parafunctional loads has eliminated most materials with adequate biocompatibility but poor physical properties such as silicone, hydroxyapatite and carbon from use as a primary implant biomaterial. However, the combination of excellent biomechanical and physical properties of titanium and titanium alloys in dental and orthopedic implants has yielded a long history of success (Cho and Park 2003). Titanium is the 22 element on the periodic table. Its anatomic structure is $1s^{2,}$ $2s^2$, $2p^6$, $3s^2$, $3p^6$, $3d^2$, and $4s^2$. The electrons $3d^2$ and $4s^2$ are highly reactive and form obstinate oxides, mainly TiO_2, responsible for its biocompatibility. Commercially pure titanium is available in four grades based on corrosion resistance, formability and strength. It contains dissolved oxygen, nitrogen, carbon, hydrogen and aluminum. The addition of aluminum and vanadium (Ti-6Al-4V) to titanium to create titanium alloys enhances the elastic modulus, ultimate tensile strength and yield strength (McCracken 1999).

Many surface modifications to titanium implants have been introduced in order achieve better bone apposition and implant to bone contact. Efforts have been made to enhance the oxide layer by thermal and anodic oxidation, sandblasting, acid etching and ionic implantation.

(Wisbey *et al.* 1991, Zhu *et al.* 2001, Cochran *et al.* 2005, Nayab *et al.* 2007). Another approach to surface modification is the addition of a biochemical surface coating to a titanium substrate. These substrates include bioactive peptides, proteins, and calcium and phosphate based ceramics. Hydroxyapatite is a calcium and phosphate based bioceramic of particular interest to the implant coating industry due to its similar structural and chemical composition to bone (Kato *et al.* 1979, Nanci *et al.*1998).

1.5.2. Design

The design of an implant can be characterized by combination of elements that compose its three-dimensional structure. Elements and characteristics that compose implant design include the type of prosthetic interface, the presence or absence of threads, additional macroirregularities, and the shape/outline of an implant (Sykaras *et al.* 2000). Additional features such as vents, grooves, flutes, indentations, and perforations have been added by many implant manufacturers to accentuate or replace the effect of threads. Parallel, tapered (conical), or stepped implant outlines can be made hollow or solid, and with a flat, round, or pointed apical end (Steigenga *et al.* 2003).

1.5.3. Surface Topography

The submerged portion of the implant can be described by its topography. Ideal implant topography would enhance osseointegration, thus many unique surfaces are commercially available. A consensus of literature indicates that implant enhancements associated with surface roughness improve adhesion to bone, if only by mechanical interlocking (Baier and Meyer 1988). Roughening an implant has been demonstrated to increase the surface area available for bone to implant contact. The increased surface area of a roughened implant has been positively

correlated by Buser *et al.* (1991) to increased biomechanical anchorage to bone. Some in vitro studies suggest surface roughness may influence osteoblast proliferation, differentiation, attachment and matrix production (Martin *et al.* 1995). Cooper *et al.* (1999) demonstrated a diminished expression of bone sialoprotein and osteocalcin with decreased mineralization of fetal bovine mandibular osteoblasts when adjacent to a plasma sprayed titanium surface as opposed to titanium oxide grit-blasted surface. They concluded that different surface topographies do in fact influence unique organic and inorganic deposits during matrix formation.

Few quantitative recommendations for surface topography are documented owing to the vast influx of commercially available surface topographies. MacDonald *et al.* (2004) expounded on the ambiguity of quantifying surface texture. Examiners sent seven implants of significantly different surface textures to three internationally renowned laboratories for surface texture characterization. Measurement techniques included contact profilometry, two- and three-dimensional laser profilometry, and atomic force microscopy. Four to thirteen parameters were reported. Thus, implant surfaces are often described by the mechanism in which they are made.

Roughened surfaces are achieved either by many mechanisms including plasma spraying, blasting, etching or sintering implants. Plasma spray (PS) coatings of titanium or hydroxyapatite are most commonly applied to the metallic core of an implant. PS particle size, speed, time of impact, temperature and distance impacts the thickness of the coating (Ong *et al.* 2006).

Titanium oxides and aluminum oxides of various particle diameters are used to abrade the surface of implants. Grit blasting creates a surface of irregular pits and depressions. The particle size, time, pressure and distance create a range of desired surface roughness (Sykaras 2000). Feighan *et al.* (1995) affirmed that blasting polished titanium implants with 30 and 60 grit aluminum oxide resulted in significantly more bone formation on the implant as well as a higher shear strength at the implant-bone interface as compared to a polished titanium surface alone.

The disadvantage to grit blasting is the potential for particles to remain embedded on the implant surface disrupting osseointegration (Orsini *et al.* 2000). Chemical exposure of a metallic implant to acid etches may enhance osseointegration without adding particulate matter or surface contaminants, eroding pits of specific dimension and shape (Sykaras 2000). Cho and Park (2003) concluded that acid etched implants have a greater resistance to torque removal compared to machined surface titanium. These results are interpreted as an increase in implant to bone contact. A combination of blasting and etching has recently been introduced. Buser *et al.* (1998) found that blasted and etched surfaces has increased removal torques over etched only surfaces, suggesting increased bone to implant contact and anchorage.

Porous surfaces are differentiated from roughened surfaces by the lack of sharp or jagged topography. Pores are created by sintering spherical powders of metallic or ceramic materials to the metallic core of an implant. The pore size, shape, depth, and volume are affected by the particle size and temperature and pressure of the sintering process (Pilliar 1998). Maximum bone ingrowth appears to occur at a poor depth of 150-300 μm (Bobyn 1980). The pore volume (porosity) directly affects the strength of the sintered coating to contact metal (Hoffman 1997).

1.6. Hydroxyapatite

1.6.1. Background

In 1788, A.G. Werner identified a brilliantly colored crystalline mineral which he would term apatite (Dana 1951). Its name in Greek is translated “to deceive”, because it had been mistaken for precious gems such as beryl, tourmaline, chrysolite and amethyst (Wei *et al.* 2005). Some 40 years later, G. Rose would postulate that apatite, a natural phosphate, may contain "atoms" of either chloride or the fluoride. Apatite is one of few minerals that are produced and

used by biological systems (Dana and Dana 1920).

Hydroxyapatite (HA) is composed of calcium, phosphorus, oxygen and hydrogen [Ca10 (PO4)6 (OH)2]. The chemistry of HA lends itself to be similar to the mineral content of hard tissues such as bone and teeth in vertebrates (Whitson 1998). Therefore, significant efforts in multidisciplinary research have sought to implicate the use of HA for numerous scientific applications such as separation and purification of proteins, drug delivery systems and bone implants (Koutsopoulos 2002, Suen *et al.* 2004, Wahl and Czernuszka 2006). For each application, the use of the calcium phosphate based material is optimized by specifying its geometry, dimension, density, pore size, mechanical strength, purity, and chemical phase.

1.6.2. Dental Implant Uses

Implant surface chemistry and topography are correlated to tissue response and fixation in bone. Therefore, the inherent bioactivity of HA lends itself to much research for implant composition (Hacking *et al.* 2002). Although HA has excellent biocompatibility, the brittle nature of HA makes it unacceptable for load bearing applications. Therefore, HA was developed as coating for dental and orthopedic implants to combine the bioactive surface of HA with the strength of titanium (Sykaras *et al.* 2000). The advantages of HA coatings lie within its reported osseoconductive ability to improve the rate and strength of initial implant integration. Enhanced integration could result in earlier and improved stability for prosthetic loading (Jarcho 1992).

Many clinical and laboratory studies have been undertaken to evaluate the biocompatibility of HA coated implants. Laboratory research by Thomas *et al.* (1989) concluded that early bone growth and apposition are accelerated by implants coated with HA. A study by Weinlaender M. *et al.* (1992) was designed to quantify the amount of bone in contact with three different implants placed in mongrel dogs. These implants included threaded titanium, flame

sprayed titanium cylinder and HA coated titanium cylinder. Results indicated that HA coated implant had a significantly greater contact with bone than the other implants. Block *et al.* (1997) found that CaP coated Ti implants stimulated bone formation at an earlier and faster rate than non-coated implants. However, Hacking *et al.* (2002) attributes the increased osseoconductive tissue response a possible result from surface topography and not bioactivity of HA coatings. This article suggests that the similarity of surface roughness in HA and titanium is often unaccounted for in research designs. Therefore, they designed an experiment to mask HA chemistry while maintaining equivalent surface topographies. Although HA surface coating showed significantly more bone apposition, 80% of apposition could be attributed to the surface topography. Wheeler (1996) cited no statistically significant difference in survival rate between HA and grit blasted titanium.

1.6.3. Preparation

HA is a stable calcium phosphate salt found in bone and naturally produced as a result of cartilage arthritis, formation of renal and bladder stones, and calcification of cardiac valves (Schwille *et al.* 2005, Suvorova and Buffat 2005, Reginato and Olsen 2007). Its medical and pharmacological implications have necessitated the production of well-defined, synthetically pure HA crystals. Several methods of production are reported, including solid state reactions, plasma techniques, crystal growth under hydrothermal conditions, layer hydrolysis, and sol-gel crystallization. However, non-stoichiometric products are the result of synthetic processes. The presence of crystal lattice vacancies and ions such as carbonates, hydrogen phosphates, potassium, sodium, nitrates and chloride combined with the high affinity of HA molecules to these ions easily contaminates and changes the cystallograpic characteristics and morphology of HA (Koutsopoulos 2002).

Commercially available HA grain size is available from 45 to 160 micrometers (Zheng 2000). However, the large grain size of conventional HA may contribute to its physical weaknesses. Smaller, denser HA feedstocks are the focus of commercial production of bioceramics today.

The concept of technological opportunity on the scale of atoms and electrons was introduced at the annual meeting of the American Physical Society in 1959 by Caltech scientist Richard Feynman (Appenzeller 1991). Feynman coined this science Nanotechnology, which today refers to the field of science focused on the development of nanometer sized materials, devices and systems. The size dependant physical properties of materials reduced to the nanoscale can suddenly show very different characteristics compared to what they exhibit on a macroscale, enabling unique applications (Salata 2004). With this knowledge, the use of nanotechnology by the implant industry has lead to the creation of nanosized bioceramics such as nano HA to overcome physical, chemical and biological limitations of conventional HA.

1.6.4. Composition

HA is chemically similar to bone mineral, but not in physical properties and microstructure. Bone is a nanocomposite composed of a collagen-fiber-reinforced microstructure, lending itself to be mechanically tough (2-12 $MPa.m^{1/2}$) as well as plastic. However, HA is a brittle polycrystalline ceramic with low fracture toughness (.6-1.5 $MPa.m^{1/2}$) (Hench 1991, Salata 2004).

The chemical composition of HA coating can vary in percent composition of crystalline HA, alpha and beta tricalcium phosphates, calcium oxides, calcium pyrophosphates, and amorphous CaP. HA coating crystallinity, phase composition, quantity and type of porosity all determine the reactive properties and subsequently the osseointegration potential of the coating.

Surface reactions include the following cascade of events including the dissolution of HA, precipitation of the apatite, ion exchange accompanied by absorption and incorporation of biological molecules, cell attachment proliferation and differentiation, and extracellular matrix formation and mineralization (Xue 2005).

Dissolution is reported by Ducheyne *et al.* (1993) to be crucial to the induction of apatite precipitation on the surface of HA coatings to initiate osseointegration. However the rate of dissolution must be balanced between enhancement of bioactivity and detrimental effects of coating disintegration, resulting in the loss of bond strength and implant fixation. (Schepers *et al.* 1991, Wheeler 1996). The rate of dissolution is associated with the crystallinity of a prepared HA coating. The crystallinity of HA describes the percent of chemical composition in crystalline form within any given HA coating. The rate of dissolution is affected by the ratio of crystalline HA to metastable compounds such as amorphous calcium phosphate (ACP), calcium oxide (CaO), α-tricalcium phosphate (α-TCP), β-tricalcium phosphate (β-TCP), and tetracalcium phosphate within HA that are either inherent or produced during the application of the coating to an implant subtrate. These metastable constituents are more soluble that crystalline HA; therefore, increase dissolution (Xue 2005).

1.6.5. HA Coating Methodologies

Implants are coated with HA through various methodologies. An ideal coating must be biocompatible at its desired phase and crystallinity to enhance biological response, have sufficient mechanical characteristics to maintain the integrity of the coating-implant interface, and possess adequate porosity to promote bone in growth and fixation (Cahn *et al.* 1993). HA is most commonly applied by plasma spraying and electrophoresis deposition. The investigation for optimal coating properties has led to other experimental coating methodologies that include

ion beam sputter deposition, sol-gel processing and high velocity flame sprayed deposition.

The PS process is a type of thermal-spray technology that uses a device to melt and deposit a coating material at a high velocity onto a substrate. PS HA onto dental implants is accomplished under normal atmospheric conditions. A direct-current electric arc created by a high current, low voltage electrical discharge between two electrodes produces a plasma flame. The arc super heats a carrier gas stream that contains the molten HA powder. The HA is deposited onto the implant by the plasma flame. Adhesion of the HA to titanium is purely mechanical and can be enhanced by a roughened substrate surface (Herman 1988, Ong *et al.* 2006).

Structural and chemical properties are modified by power, currents, plasma work gas rate and composition, carrier gas rate, spraying time, particle size, speed of the plasma, distance between substrate and nozzle and cooling rate. The plasma work gas composition is the most important spraying parameter in regulating the desired crystallinity, heat content and coating thickness of HA. An argon carrier gas can be infiltrated with other elements to change coating properties. For example, introducing hydrogen will result in a higher crystallinity and hotter plasma, whereas nitrogen can produce a thicker coating layer (Ong *et al.* 2006). Commercially available PS coatings are reported to have a thickness of greater than 30 micrometers (Herman 1988). Surface roughness of PS HA coated implants can range from R_a= 3 to 8 micrometers (Hacking 2002).

The advantages of PS include simplicity, rapid deposition rate, low substrate temperature, low cost and variable coating porosity, phase, and structure and low cost. The disadvantages associated with PS HA coatings are attributed to over 100 operational variables associated with spraying process that ultimately affect the final coating's microstructure and physical properties (Cheang and Khor 1996). Reported problems include poor bond strength between coatings, HA

adhesion to its substrate, structural and chemical variation within the coating process, and variation between commercial vendors of HA coatings (Wennerberg *et al.* 1993, Zheng *et al.* 2000, Khor *et al.* 2003).

To improve the mechanical integrity and biological and physical properties of PS HA, research has focused on incorporating functionally graded coatings and post-spray heat treatments. Functionally graded HA coatings incorporate dissimilar materials such as titanium alloys enhance mechanical properties. Chen *et al.* (2006) determined functionally graded HA/Ti composite coatings had superior mechanical properties over monolithic HA coatings. Zheng *et al.* (2000) concluded that incorporating titanium to HA coating would significantly improve the bond strength of the PS coating. Khor *et al.* (2003) found that a graded distribution of HA and Ti-6Al-4V phases in FGM coatings improved the tensile adhesion strength significantly.

Post spray heat treatment affects the HA coating phase by increasing crystallinity. The desired crystallinity of HA coating is of much dispute. Ding *et al.* (2003) found that annealing HA coatings in air at 650° C led to recrystallization of amorphous CaP, increasing crystallinity and reducing residual stresses of the apatite. However, Porter *et al.* (2002) noted annealing HA coatings resulted in prolonged timing of apatite deposition due to the reduced dissolution of the heat treated coating.

Electrophoresis deposition (EPD) is a process in which colloidal particles such as HA nano precipitates are suspended in a liquid medium migrate under the influence of an electric field and are deposited onto a counter charged electrode. Pressure is concurrently applied to HA nano particles against the electrode. The coating is simply formed by pressure exerted by the potential difference between the electrodes (Birks and Schulman 1959).

The operational parameters of EPD can be changed to alter HA surface coating morphology and composition. These parameters include the voltage/intensity relationship,

deposition time, and composition, pH and temperature of the electrolyte (Meng *et al.* 2006, Wang *et al.* 2006). The process of EPD is described by Hamaker's equation, which relates the electrophoretic velocity to a function of the electric field and particle size:

$$\upsilon = Q\,E\,/\,4\,\pi\,r\,\eta \qquad \text{(Eq.1)}$$

where Q is the charge, r is the particle radius, E is the potential difference applied to the suspension of the electric field, and η is the suspension viscosity (Mondragon-Cortez and Vargas-Gutierrez 2004). It is inferred that small HA particles gain the highest electophoretic velocity and are deposited at a lower velocity to create a denser coating. A higher suspension velocity would deposit larger HA particles on a substrate at a faster rate resulting in a more porosity of the coating (Meng *et al.* 2006).

Studies done by Meng *et al.* (2006) confirmed that a low voltage of 20 V created a coating of fine, dense HA particles (150-200 nm). However, a constant high voltage of 200V created a porous surface composed of large HA particles (>400nm). By applying a dynamic voltage, authors were able to create a graded coating of particle size and porosity. They also noted that post deposition sintering resulted in a denser, more adhesive coating on the substrate.

The reported advantages of EPD encompass its low cost, simple methodology capable of producing coatings of variable thicknesses, high deposition rate, formation of highly crystalline deposits with low residual stresses and ability to uniformly coat non-line-of-site, irregularly shaped, or porous objects such as threaded implants due to its high throwing power (Meng *et al.* 2006, Wang *et al.* 2006). EPD can produce HA coatings ranging from <1 micron to >500 microns thick (Wei *et al.* 2005). Surface patterns created on the EPD cathode have been shown to create a patterned HA coating on an implant substrate to change surface topography and

enhance osseointegration (Wang and Hu 2003).

The major disadvantage is EPD is the need for post deposition heat treatment to densify the coating (Wei *et al.* 1999). Conventional HA feedstocks require temperatures of at least 1200° C to be densified (Ruys 1995, a). Temperatures above 1050° C affect the oxide layer and mechanical properties of a stainless steel or titanium alloy, as well as decompose HA affecting the interfacial strength between the metal and coating (Ducheyne 1990). To obtain lower temperatures for densification, ultrafine HA powders were developed in the range of 0.01 to 0.1 micrometers (nanoparticulate), achieving a plateau density of 100% at 1000° C (Ruys 1995, b). Wei *et al.* (1999) evaluated the optimization of the properties in production of HA nanoparticles for use in EPD methodology, concluding ambient-aging ripening for 10 days eliminated cracking in the electrophoretic coating. Further research by Wei *et al.* (2005) noted several methodologies for producing HA nano precipitates, confirming the metathesis produced highly equiaxed nano particles, lending to less cracked coatings obtained by the electrophoretic deposition technique. Hu *et al.* (2007) confirmed with the Raman and IR spectra that the main component of EPD coatings is well crystallized with excellent bioactivity and biocompatibility.

Chapter 2. Problem and Research Objectives

2.1. Research Problem

Hydroxyapatite coatings have significant biologic potential as a surface modification for dental implants. However, an optimal HA application modality must be identified. The current study evaluated two HA coating methods and intrinsic differences in coatings relative to application processes.

2.2. Research Objectives

The goal of this research was to compare electrophoresis deposited nano HA coated implants with plasma sprayed HA coated implants. The following aims and hypotheses were developed for this study:

Specific aim I: To evaluate the interfacial strength and bone-contact length of the two different HA coated implants following 12 weeks in situ before loading.

Hypothesis I.I: No difference in bone contact length will be observed between the electrophoresis deposited nano HA coated implants and plasma sprayed HA coated implants following 12 weeks in situ before loading.

Hypothesis I.II: No difference in and interfacial strength will be observed between the electrophoresis deposited nano HA coated implants and plasma sprayed HA coated implants following 12 weeks in situ before loading.

Specific aim II: To evaluate the interfacial strength and bone-contact length of the two different HA coated implants after loading for 9 months.

Hypothesis II.I: No difference in bone contact length will be observed between the

electrophoresis deposited nano HA coated implants and plasma sprayed HA coated implants after loading for 9 months.

Hypothesis II.II: No difference in interfacial strength will be observed between the electrophoresis deposited nano HA coated implants and plasma sprayed HA coated implants after loading for 9 months.

Chapter 3. Materials and Methods

3.1. Experimental Plan

Thirty-six cylindrical implants (4 mm diameter by 8 mm long) were obtained ("O" Company, OCO Biomedical, Inc., Albuquerque, NM). Eighteen implants were plasma sprayed (PS) with HA and 18 were coated with electrophoresis deposited (EPD) nano HA coatings. Interfacial histological and mechanical strength were recorded through qualitative methods to compare coating methodologies.

3.2. Samples

Six adult male foxhound dogs (approximately 2 years old), weighing between 20 to 25 kg, were used in this study. All animals used in the study were managed in compliance with USDA program and NIH publication # 85-23, "Guide for the Care and Use of Laboratory Animals." Policies, standards and guidelines for the proper use, care, handling, and treatment of animals were followed. Surgery intervention consisted of two operations; mandibular posterior sextant tooth extractions and implantation. All dogs were kept NPO 12 hours prior to surgery. Prior to extractions, the dogs were maintained on Purina Canine Lab Chow (Purina Mills, St. Louis, MO). Following extractions, the dogs were fed Pedigree (Kal Kan Foods, Inc. Vernon, CA) soft canned dog food.

3.3. Instrumentation and Equipment

Three months prior to implantation, extraction surgery was performed. Anesthesia was induced using sodium pentothal (Abbott Laboratories, Chicago, IL). The dogs were intubated and maintained with vaporized 0.80-1.0% halothane (INEOS Fluor Americas LLC, St. Gabriel

LA). The animals were maintained on a ventilator. Lactated Ringer's solution was administered at a slow drip. Prior to surgery, the dogs received Penicillin BP (3 cc every other day) and Gentocin (2 cc SID) subcutaneously as an antibiotic prophylaxis. Extractions were accomplished using standard oral surgery techniques, including the use of elevators, forceps, and a high speed handpiece. Each animal was edentulated in each posterior mandibular sextant (first through fourth premolars). The surgical flaps were closed with a continuous 3/0 Vicril suture. Post operative analgesia was accomplished with 1 mg/Kg Nubane bid subcutaneously. The dogs continued to receive Pen BP and Gentocin for 7 days post surgery.

At the time of implantation, anesthesia (IV and vaporizer), ventilator settings, IV fluids, antibiotics and analgesics were the same as described above. The oral cavity was rinsed twice with Betadine solution to obtain a relatively clean environment. The healed alveolar ridges of the mandibular posterior quadrants were exposed by raising subperiosteal flaps. When the crest of the alveolar ridge was less than 4 mm wide, the height of the ridge was reduced to provide an adequate width of bone for implant placement.

A contra angle handpiece was used with 3i drills (BIOMET 3i, Palm Beach Gardens, FL) for implant site preparation. Three surgical implant sites were prepared in each quadrant. A maximum cutting drill speed was used to minimize surgical trauma to the bone, and new drills were used for each animal. Three PS HA coated implants and three EPD nano HA coated implants were placed in each dog. A total of six implants were placed in each mandible. After placement of the implants, the screw holes on the superior surface of the implants were covered with cover screws. The surgical area was liberally irrigated with normal saline to remove bone fragments and the tissue flaps are to be closed with continuous 3/0 Vicril sutures.

At 12 weeks after implantation, 3 animals euthanized with potassium chloride, administered IV, under deep sodium pentothal anesthesia. Testing was performed within 24

hours after sacrificing the animals. The tissue over the alveolar ridge was elevated and the most distal implant site identified. The mandibles were resected 20 mm posterior to the distal most implants, stripped of soft tissues and sectioned with a Gillings Holmes Sectioning Machine using a diamond wafering blade with copious normal saline irrigation.

To evaluate the interfacial strength of the implants at the bone-implant interface, pull out testing was performed using a MTS test frame (Model: Alliance RF/150, MTS). Prior to testing, the bone-implant blocks were kept in cold normal saline. In preparation for testing, excess tissue was removed. Each specimen was dissected with a table saw equipped with a small diamond blade until the implant was just at the surface and the implant axis perpendicular to the base plane of the bone–implant complex. Prior to each experiment, implant and transcortical contact dimensions were measured with a caliper gauge to determine the cortical bone–implant contact area. An implant pulling fixture was inserted through the topside of a stainless steel metal plate with a hole just large enough for the middle portion of the pulling fixture to tightly slip through. A bone-implant block was placed beneath the metal plate, and the implant pulling fixture was screwed into the implant. The top of the implant pulling fixture protruding through the hole in the fixture was secured in a Jacobs Chuck attached to the MTS instrument. At rate of 1 mm/min at a load of 5 kN, testing was performed until the bone–implant interface was ruptured. Significant differences in interfacial strength between the groups were analyzed using ANOVA, with Sheffe's procedure as the post hoc test. Differences were considered significant if $P < 0.05$.

To quantitatively evaluate bone formation, specimens were embedded in methylmethacrylate and sectioned parallel to their long axis using a Buehler Isomet equipped with a slow speed and diamond wafering blade. Multiple-stain (Paragon Stain, Polysciences, Warrington, PA) was used as the primary stain. The length of direct contact of bone to implant was measured histomorphometrically over a distance that corresponds to the average contact

length. This length was calculated as a percentage of the axial length. This measurement technique has been referred to as the profile contact length (PCL). The length of integration and area of integration along a 100 μm boundary at the implant tissue interface was measured with digitally processed images. Any interface exhibiting osseous tissue within 10 μm of the implant surface was considered osseointegrated. Significant differences between the bone-implant contact lengths were analyzed using ANOVA, with Sheffe's procedure as the post hoc test. Differences were considered significant if $P < 0.05$.

At 12 weeks, the remaining animals were anesthetized using the aforementioned procedures. Custom made implant crown were cemented to implant abutments. The implants were loaded for 9 months before sacrifice and data collection. Interfacial strength and histological analysis of the bone to implant contact lengths were measured according to the same testing protocol described at 12 weeks.

Chapter 4. Results

4.1. Pull Out Strength Testing

All surgeries were uneventful, with no post-operative complications. At 12 weeks after implantation, no statistical difference in pull out strength was observed between EPD nano HA (86.35 ± 28.07 lbf) and plasma sprayed HA coated implants (70.43 ± 16.65 lbf) (Figure 1 and Table 1).

After loading for 9 months, it was observed that the interfacial strength of EPD nano HA coted implants (199.9 ± 35.1 lbf) was statistically higher ($P < 0.028$) than the PS HA coated implants (121.14 ± 38.45 lbf) (Figure 1 and Table 2).

4.2. Histological Evaluation of Bone Contact Length

At 12 weeks implantation, no statistical significant difference in bone contact length was observed between EPD nano HA (97.6 ± 3.2%) and PS HA coated implants (95.6 ± 4.6%) (Table 3). No statistical significant difference in bone contact length was observed between EPD nano HA coated implants (91.8 ± 8.2%) and PS HA coated implants (84.3 ± 7.2%) after loading for 9 months (Table 4).

Figures 2 and 3 are representative histology of tissue-implant interface after 12 weeks implantation. Figures 4 and 5 are representative histology of implant-bone interface after 1 year implantation. Mature bone at the tissue-implant interface was observed for both implant groups at both time intervals.

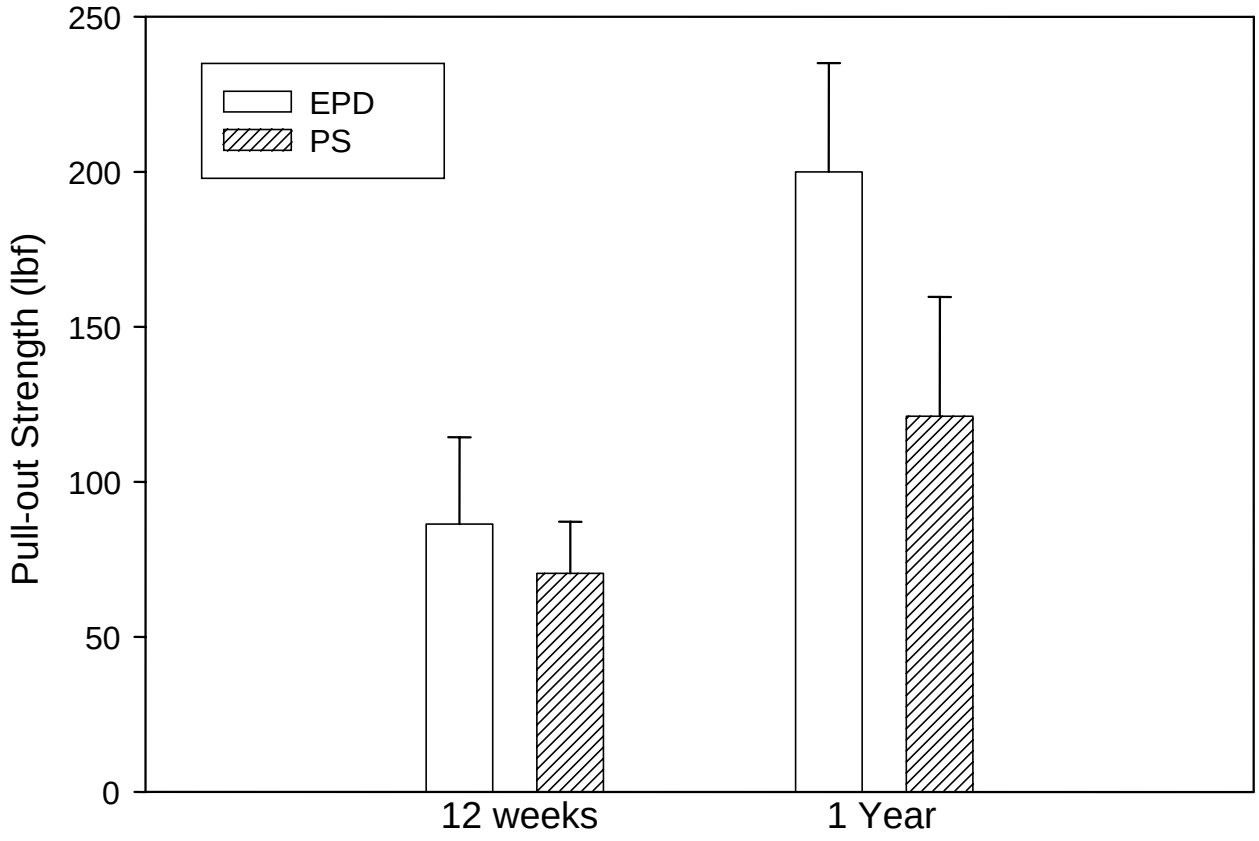

Figure 1. Mean interfacial strength with standard deviations of EPD nano HA and PS HA coated implants after 12 weeks and 1 year of implantation.

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Table 1. Pull out strength (lbf ± SD) of different implant surfaces after 12 weeks implantation.

Sample	Pull Out Strength (lbf ± SD)
EPD – 12 wk	86.35 ± 28.07
PS – 12 wk	70.43 ± 16.65

Table 2. Pull out strength (lbf ± SD) of different implant surfaces after 1 year implantation.

Sample	Pull Out Strength (lbf ± SD)
EPD – 1 year	199.87 ± 35.13
PS – 1 year	121.14 ± 38.45

Table 3. Mean bone contact length (% ± SD) of different implant surfaces after 12 weeks implantation.

Sample	Percent Bone Contact Length (± SD)
EPD – 12 wk	97.6 ± 3.2
PS – 12 wk	95.6 ± 4.6

Table 4. Mean bone contact length (% ± SD) of different implant surfaces after 1 year implantation.

Sample	Percent Bone Contact Length (± SD)
EPD – 12 wk	91.8 ± 8.2
PS – 12 wk	84.3 ± 7.2

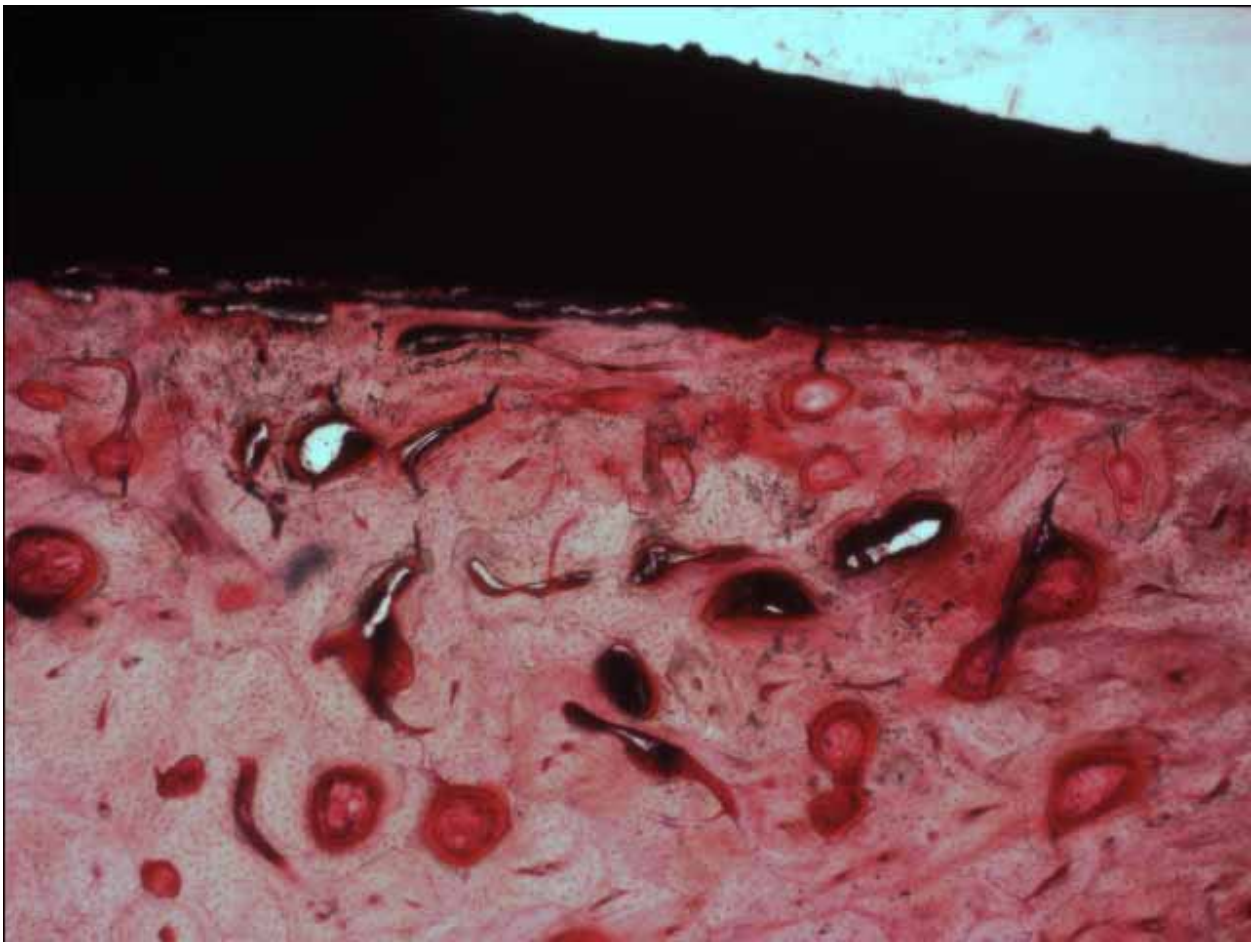

Figure 2. Tissue-implant contact of plasma sprayed HA after 12 weeks.

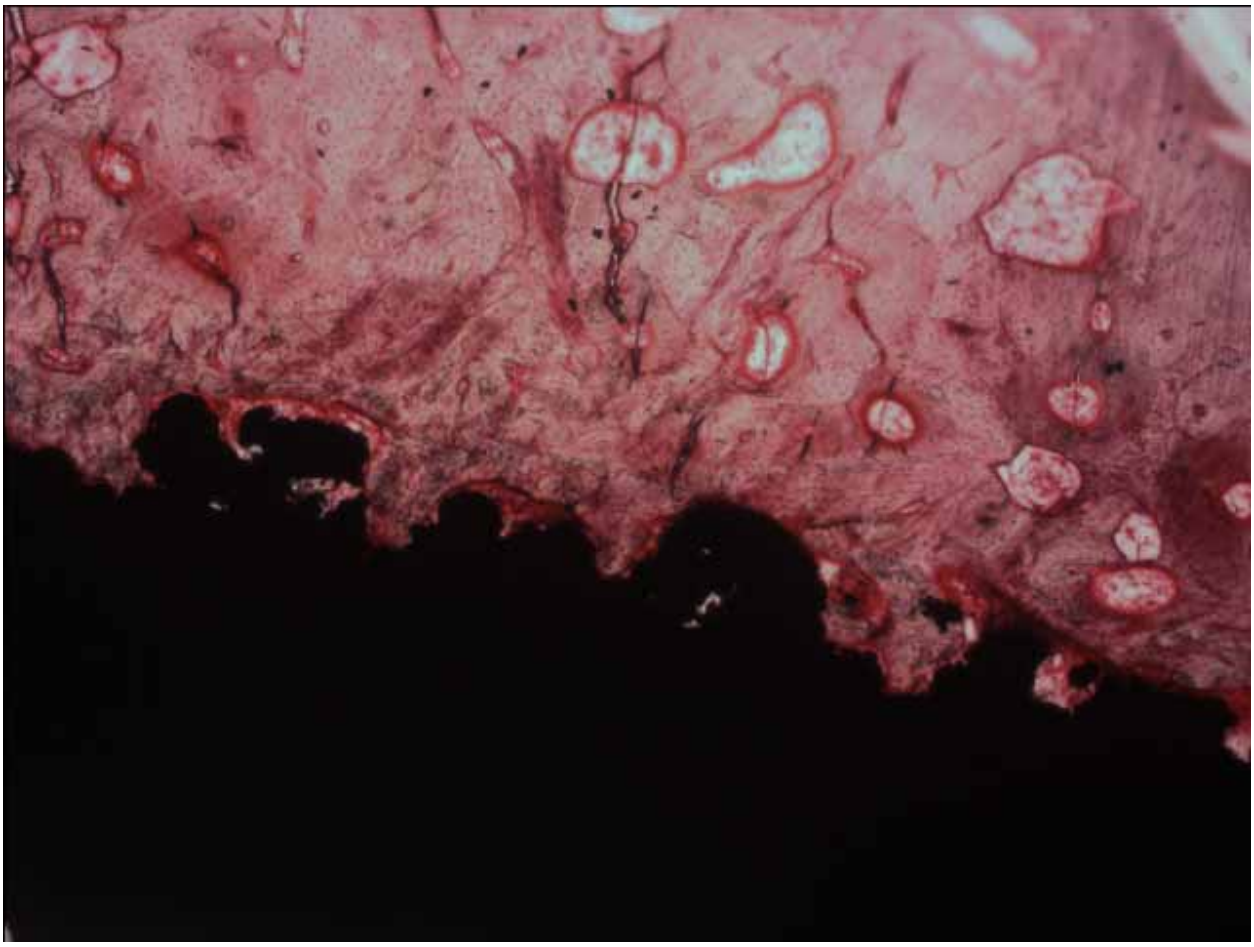

Figure 3. Tissue-implant contact of EPD nano HA after 12 weeks.

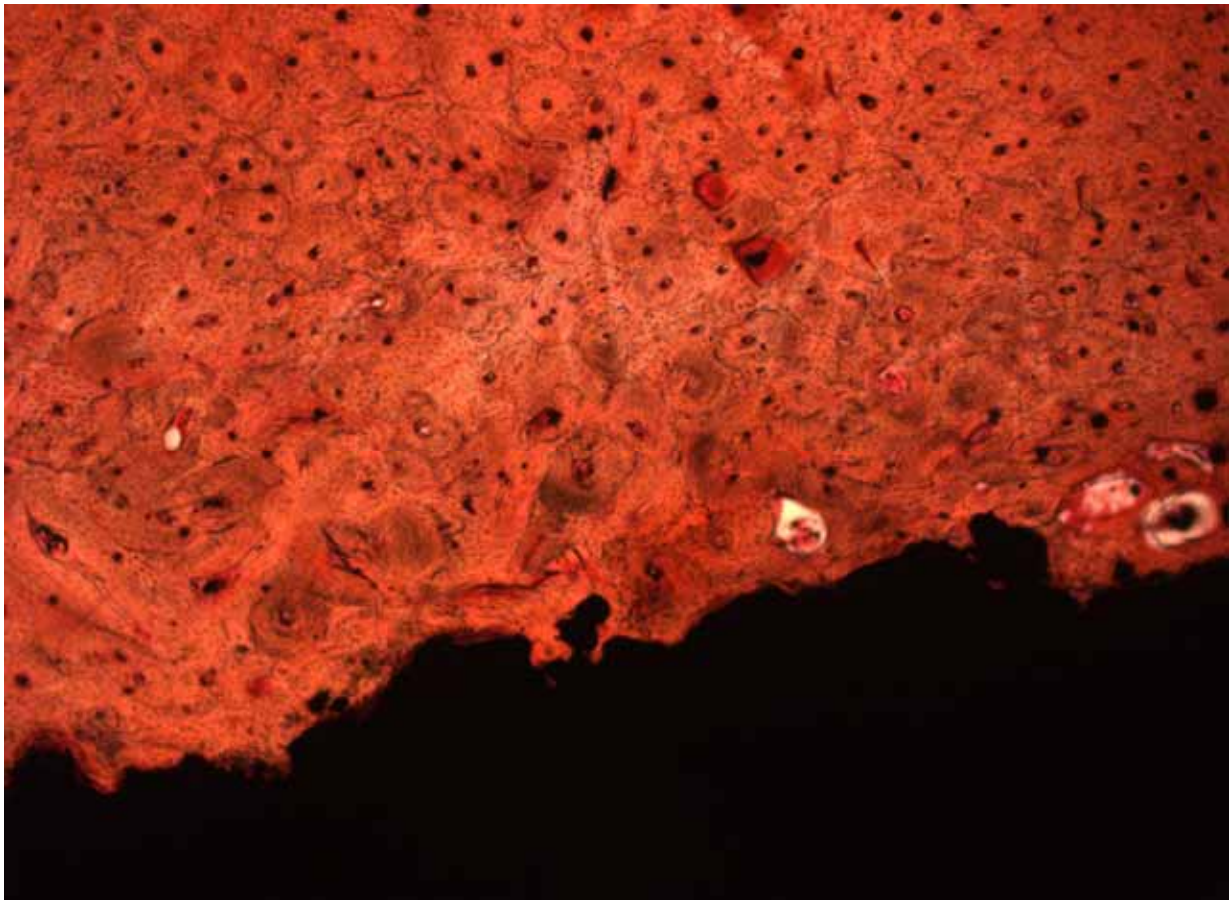

Figure 4. Tissue-implant contact of plasma sprayed HA after 1 year.

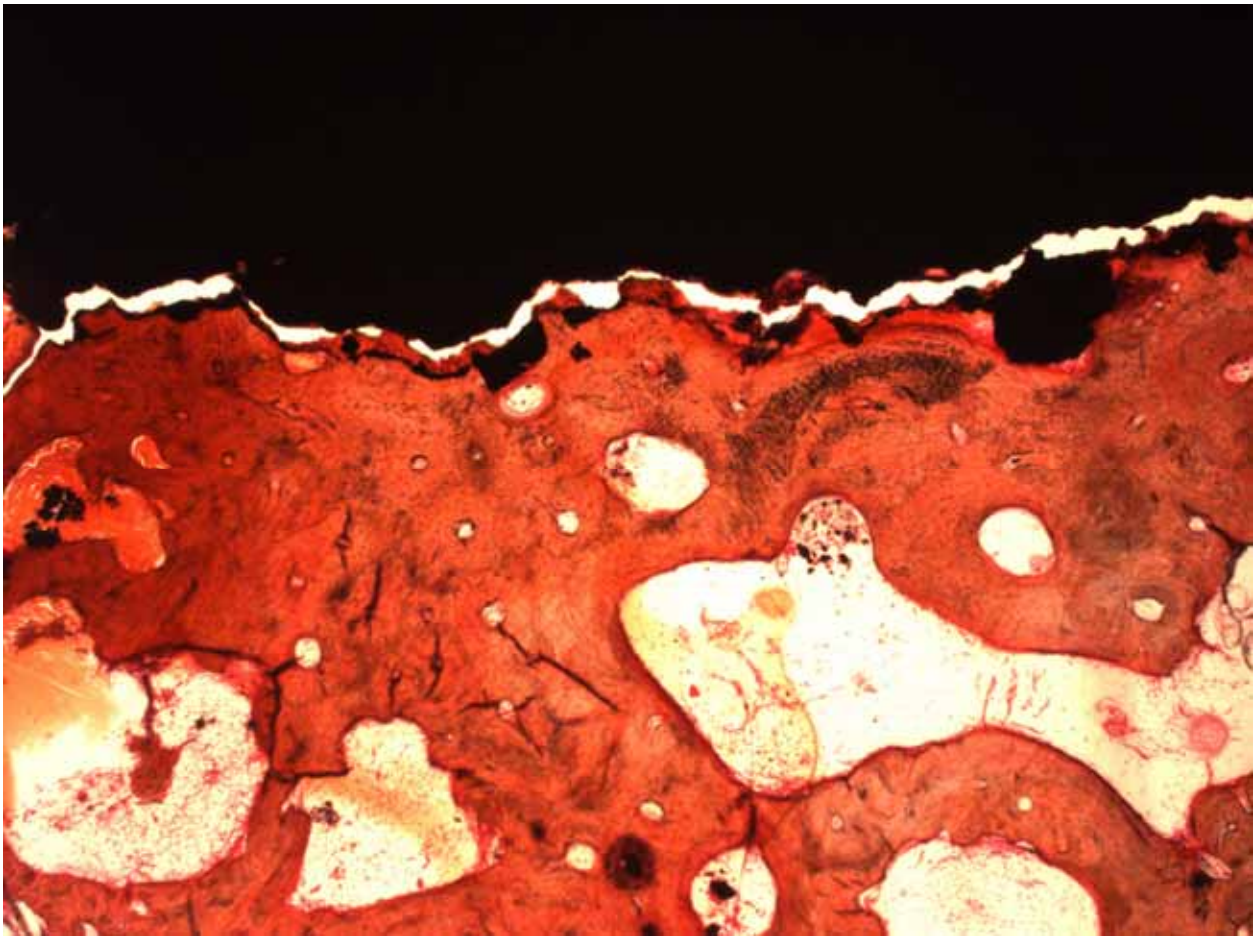

Figure 5. Tissue-implant contact of EPD nano HA after 1 year.

Chapter 5. Discussion

Tooth loss due to disease, trauma or otherwise congenitally absence is as primitive as the idea of replacement. The early 1900's marked the beginning of an era defined by modern methods of implant therapy. The principal advancement was the identification of the ability of an implant to become "inseparately incorporated" into bone by Brånemark and associates, coining the term osseointergration. They noted the importance of osseointegration to achieve and maintain implant bone anchorage under loading (Albrektsson *et al.* 1981). Many implant studies at present concentrate on understanding the physiology of bone response to a variety of implant characteristics in order to enhance osseointegration for more dynamic treatment therapy involving dental and medical implants.

Contemporary endosseous implants rely on bone for support and anchorage. The composition of bone includes cells, extracellular matrix and minerals including calcium and phosphorus which are stored as hydroxyapatite (HA) crystals. These components work together to heal and remodel bone after surgical insult associated with implant placement. Block *et al.* (1997) described the integration of implants into bone by three phases; inflammatory, proliferative, and maturation phase. Within the first 42 days (inflammatory and proliferative phases), there is an increase of cellular activity, neovascularizton, differentiation, proliferation, and production of connective tissue matrix that will ultimately be mineralized and remodeled. Sleats *et al.* (2006) confirmed the invasion and maximum activity of basic multicellular units in vivo after 4 weeks.

A basic knowledge of the cascade of events associated with bone remodeling correlated with the ability to synthetically reproduce HA has led to the development HA within the implant industry. As implant technology evolves, so must the understanding of basic biologic principles

that control the ability of alveolar bone to tolerate and maintain artificial replacements of teeth. Implant design, composition and surface topography are among the most researched variables in finding the fastest and most durable bone anchorage for implants.

Classified as a bioactive material, HA has the potential to allow for formation of new bone on its surface, forming a scaffold for bone ingrowth by exchanging ions to create a chemical as well as mechanical bond (Sykaras *et al.* 2000). HA has been developed as a coating to combine its bioactivity with the strength of a metal substrate. Its use continues to be a popular topic of research and debate. Weinlaender *et al.* (1992) recognized that HA coated implants had a significantly higher percentage of bone contact as compared to different titanium surfaced implants, histologically noting a different pattern of bone growth.

Among HA coatings are a multitude of depositional, compositional and topological differences. Cahn *et al.* (1993) described ideal HA attributes to include biocompatibility at its desired phase and crystallinity to enhance biological response, sufficient mechanical characteristics to maintain the integrity of the coating-implant interface, and adequate porosity to promote bone in growth and fixation. Several methodologies of HA surface deposition now achieve these fundamental elements to produce surface coating capable of enhanced osseointegration. Such methodologies include plasma spray (PS) and electrophoresis deposited (EPD) nano HA. However, no general consensus is recognized about the superiority of one methodology's ability to enhance osseointegration. A multitude of tests exist to interpret each coating's ability to augment osseointegration.

Pull out testing is often used to infer information about the interfacial bond strength between bone and the implant interface. The pull out test results observed in the present study for PS HA coated implants (70.43 ± 16.65 lbf) was not statistically different from that found for EPD nano HA coated implants (86.35 ± 28.07 lbf) at 12 weeks after implantation, suggesting

comparable interfacial bond strengths. This was supported by the histological findings, which indicated no statistical difference in percent bone contact length between PS HA coated implants (95.6 $\pm$ 4.6%) and EPD nano HA coated implants (97.6 $\pm$ 3.2%) at 12 weeks after implantation. Hypothesis I.I and I.II were accepted: No difference in bone contact length will be observed between the EPD nano HA coated implants and PS HA coated implants following 12 weeks in situ before loading and no difference in and interfacial strength will be observed between the EPD nano HA coated implants and PS HA coated implants following 12 weeks in situ before loading.

The results at 12 weeks are similar to other short term healing studies. Wang *et al.* (2006) evaluated early bone apposition in vivo on PS HA and EPD nano HA coatings on titanium alloy's. Their research concluded that HA coatings only accelerated early stage mineralization ($\leq$ 7 days). No difference was found at later stages (14 days) of mineralization rates. They attributed these results to the difference in surface morphology and crystallinity created by unique coating processes. Their ESEM images revealed plate-like shaped HA crystallites with a surface roughness of 1300 $\pm$ 400 nm compared to the large globules of amorphous HA and indistinguishable HA crystallites with a surface roughness of 480 $\pm$ 120 nm for the EPD nano HA coating.

The interfacial strength of EPD nano HA coated implants (199.87 $\pm$ 35.1 lbf) was statistically higher ($P < 0.028$) than the commercial PS HA coated implants (121.14 $\pm$ 38.45 lbf) after 1 year of integration and 9 months of loading. However, histological analysis reveal no statistical significant difference in bone contact length between EPD nano HA coated implants (91.8 $\pm$ 8.2 %) and PS HA coated implants (84.3 $\pm$ 7.2 %) after 1 year of integration and 9 months of loading. Histologically, no statistical difference in percent bone contact length exists even though EPD nano HA coated implant mean percent bone contact is 7.5 % higher.

However, pull out testing suggests stronger interfacial bond strength, inferring superior osseointegration of the EPD nano HA after 1 year of integration and 9 months of loading. Hyposthesis II.I was rejected: no difference in bone contact length will be observed between the EPD nano HA coated implants and PS HA coated implants after loading for 9 months. However, hypothesis II.II was accepted: no difference in interfacial strength will be observed between the EPD nano HA coated implants and PS HA coated implants after loading for 9 months. To date, there is a lack of literature evaluating the superiority EPD nano HA coatings to other methodologies.

PS HA and EPD nano HA coating methodologies have the ability to dramatically change structural and chemical properties by changing operational parameters. PS HA coatings are modified by power, currents, plasma work gas rate and composition, carrier gas rate, spraying time, particle size, speed of the plasma, distance between substrate and nozzle and cooling rate. The plasma work gas composition is the most important spraying parameter in regulating the desired crystallinity, heat content and coating thickness of HA (Ong *et al.* 2006). EPD nano HA coatings are modified by the voltage/intensity relationship, deposition time, and composition, pH and temperature of the electrolyte (Meng *et al.* 2006). Commercially available PS and EPD nano HA coatings can substantially differ in crystallinity, thickness and surface roughness among proprietors. Each of these variables in itself can have a distinct affect in tissue response. Therefore, an evaluation of these coating attributes would be valuable to quantitatively and qualitatively differentiating and interpreting results.

Osseointegration is dependant on the bioactivity and stability of the coating material. Many authors address these properties of coating materials, suggesting that HA's bioactive influence expires after the initial stages of osseointegration (Wang *et al.* 2006). Therefore, thick, soluble coatings as those achieved by PS may be unnecessary. The ability of EPD nano HA to be

thinner and more adherent than conventional PS HA coating could beneficially impact long term stability, even though its typically higher crystallinity values make it less soluble. Although PS HA is an accepted and proven surface coating, EPD nano HA shows industrial and biological promise in the implant industry.

Chapter 6. Conclusion

This study determined that the ultimate interfacial strength of plasma sprayed (PS) HA coated implants were similar to that of electrophoresis deposited (EPD) nano HA coated implants at 12 weeks. In addition, histomorphometric evaluation at 12 weeks indicated that the percent bone contact length for PS HA coated implants were similar to that of EPD nano HA coated implants. However, the ultimate interfacial strength of PS HA coated implants was statistically less than that of EPD nano HA coated implants at 1 year, even though histomorphometric evaluation at 12 weeks indicated that the percent bone contact length for PS HA coated implants was similar to that of EPD nano HA coated implants. It is concluded that EPD nano HA coating technology is a viable and promising process in use for medical and dental implants.

Commercially available PS and EPD nano HA coatings can substantially differ in crystallinity, thickness and surface roughness among proprietors. Each of these variables in itself can have a distinct affect in tissue response in any bioactive or biotolerant material. Therefore, an evaluation of these coating attributes would be valuable to quantitatively and qualitatively differentiating and interpreting results in future studies.

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Vita

Audrey Marie Selecman was born in Atlanta, Georgia, on December 19, 1976 as Audrey Marie McFadden. At age twelve, she moved to Knoxville, Tennessee, where she would graduate from Karns High School in 1995. Audrey attended Maryville College in Maryville, Tennessee, where she received a Bachelor of Arts degree in Biology in June 1999. She was accepted at The University of Tennessee Health Science Center College of Dentistry in Memphis, Tennessee, and was awarded a Doctorate in Dental Surgery degree in May 2004. The following July, Audrey entered as a graduate student in the Department of Prosthodontics and is expected to earn her Master of Dental Science degree in May 2007. Audrey and her husband plan to remain in Memphis, Tennessee, after her graduation.