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Chapter 28

Hydroxyapatite Thick Films as Pressure Sensors

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Electrical properties of hydroxyapatite (HA) in the form of screen-printed thick films that can be used as a biocompatible coating for bone and dental implants are reported. In particular, piezo- and pyroelectric behaviour of these films suggest that they can be used to promote faster healing of bones and prevent rejection of implants. Moreover, the reversible pressure-induced changes in their electrical characteristics can be employed for real-time *in vivo* pressure sensors implantable simultaneously, for example, with knee or hip prosthesis. The additional advantage of HA in the form of screen-printed thick films is that, due to the technology's versatility, it can be produced on flexible substrate in any shape and size to suit the needs of various patients.

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1. Introduction

Hydroxyapatite (HA) is a bioactive material with the ability to support bone growth. HA coated implants have been largely used in implants with bioactive fixation that rely on direct chemical bonding of the bone-implant interface.[1,2] It has high compressive strength but low flexural strength and fracture toughness. Due to the fact that HA does not exhibit good mechanical strength, it is often used as a coating material for different metal substrates. In this approach porous HA provides biocompatibility and space for bone ingrowth, while the metal substrate provides load bearing functions. HA coated implants are used in dentistry and orthopaedics due to their superior bonding with the host bone, achieved by bone ingrowth into the porous structure of HA coating. Polarisation of HA thick film and its effects on bone growth has been discussed in Chapter 9. In this chapter we report electrical properties of HA in the form of screen-printed thick films. Following the discussion of a theory behind pressure sensing, this chapter critically reviews the benefits of applying HA coatings on bone and dental implants. It then summarises the modern coating deposition techniques that are suitable for the biomedical industry, focusing on a special case of thick-film screen printing method, which was used to produce the bespoke pressure-sensitive HA coatings reported here. These can potentially be used as both the functional HA coating of any change and a real-time pressure sensor for implant performance monitoring.

2. Pressure Sensing Principles

The behaviour of screen printed HA thick films closely approximates the nature of standard capacitors. Since the tested HA samples can be treated as parallel-plate capacitors, pressure induced changes of capacitance can be explained simply. Application of pressure reduces the thickness of HA dielectric layer. Dependence of capacitance on the changes in the thickness can be described by

$$\mathbf{C} = \varepsilon_0 \varepsilon_r \left(\frac{A}{d + \Delta d} \right) \tag{1}$$

where C is the capacitance, A is the area of plates, which was $3.17 \cdot 10^{-5}$ m^2 in this set of experiments, d is the thickness of the dielectric material and Δd is the change in the thickness due to applied pressure; while ε_0 and ε_r represent the permittivity of free space and the dielectric material respectively.

3. Printed Pressure Sensors in Medical Applications

Medical technology is one of the most challenging markets for electronics. Very high demands are placed on the precision, reliability and long-term stability of electronic systems. The same also applies to the electronic components and sensors. A wide range of pressure sensor dye, pressure transducers and pressure transmitters are available for pressure measurement in the most diverse medical applications, ranging from basic blood pressure sensors to ingestible microsensors for *in vivo* gastrointestinal tract pressure measurements.

Probably the most effective material for pressure sensing application is lead zirconate titanate (PZT). It has excellent piezoelectric properties, however due to existence of a lead phase its utilisation in biomedical applications is limited. The novelty of the approach reported in this chapter is that the HA coating acts as both the promoter of bone-implant ingrowth and as a sensor that monitors this process by real-time assessment of a load experienced by that very coating.

4. Deposition Techniques of HA Coatings

New technologies, such as electron beam evaporation, provide for creation of complex interface structures to enhance bone ingrowth in cement-free implants. For example, in specimens with a simulated HA coating, a bone ingrowth depth of 500 μm yields substantial interface strength, whereas for the uncoated specimens a plateau was reached at 1500 μm of ingrowth depth, and deeper ingrowth did not enhance the interface strength considerably.[3] This confirms that the HA-implant interface itself provides the main support matrix, while the depth of in-bone ingrowth does not always enhance the bone-implant interface strength.

There is a broad range of technologies to produce HA coatings depending on the type and location of implants, with bone and dental implants being the main example. The main methods of HA coatings and issues associated with them in medical applications are briefly reviewed in the following section. Thermal spraying techniques of HA have been long and widely established in the research community. Other most commonly used techniques include High Velocity Oxy-Fuel (HVOF), high velocity suspension flame spraying (HVSFS), chemical vapour deposition (CVD), physical vapour deposition (PVD), and electrophoretic deposition (EPD). The following sections will focus on these techniques and will describe them in more detail. Another novel technique, CoBlast^{TM}, will also be discussed.

4.1. *HA coatings for bone and dental implants*

HA coated implants have been used in dentistry and orthopaedics since the 1980s due to superior bonding between the host bone and HA coated implants that can be achieved by bone ingrowth into porous structure. HA coated implants reduce the risk of osteolysis and subsequent implant failure.

However, there are still concerns related to long-term *in vivo* stability of HA coated implants, although there is more evidence.[4] that porous HA coated acetabular components significantly enhanced bone in-growth in the presence of wear particles, preventing their migration and reducing osteolysis. Since HA particles are biodegradable and do not produce any inflammatory reaction in the surrounding bone, fears of osteolysis or third body wear due to HA debris have not been confirmed.[5] Furthermore, a superior survival (up to 11 years) but accompanied by a higher wear in the polyethylene part of HA-coated Mallory-Head cups was reported,[6] along with the excellent long-term fixation of the ABG-1 femoral stems derived from the osteointegration and proximal seal around the HA coating.[7] HA coated titanium based endosteal implants have gained significant use in dental applications. Titanium plasma sprayed (TPS) HA is most widely used as coating material for endosseous dental implants. Although coated implants with both TPS and HA show appreciable rate of success, coating metallic implants with bioactive HA can accelerate the integration process during initial stages of osseointegration and reduce recovery time.[8]

4.2. *HA coatings by plasma spraying technique*

Plasma spraying deposition technique is frequently utilised to deposit the HA on a metal substrate. It involves deposition of HA powder in a heating compartment where it is subsequently melted. Molten fragments of HA powder are pushed towards and deposited on metal substrate. To date, the plasma spraying method has enjoyed dominance in application of HA coatings on metal substrates. Considerable amount of HA coated implants that were used in early the 1990s in the US were produced by a plasma spraying technique. Furthermore, plasma spraying technique is a commercially approved method by US Food and Drug Administration (FDA) for HA coatings on metal substrates.

The plasma spraying process involves transfer of electrical energy to a plasma generator. This device is sometimes referred to as a "plasmatron", which is an X-ray tube like device that consists of cathode and anode. Different types of ionising gases are used in a plasmatron. A direct current supplied to the cathode causes the ionisation of gases and a plasma flame is produced. The gas forms unstable plasma that is transformed into gas again. This process is characterised by withdrawn thermal and kinetic energy which is subsequently used to deposit the HA particles on a substrate of any shape.

Plasma spraying equipment is somewhat complex due to the fact that many parameters should be carefully chosen in order to attain the desirable coating results. Spraying HA at low particle temperature results in low degradation properties, however it lacks mechanical properties due to poor adhesion to the substrate. On the other hand, spraying HA at high particle temperatures results in good mechanical properties but exhibits poor biocompatibility due to formation of calcium oxide (CaO). In addition, one of the disadvantages of plasma sprayed HA coated implants is their degradation. On their way from plasmatron to substrate, HA particles are subjected to high temperature fluctuations that provoke its degradation. Formation of highly dissoluble tricalcium phosphate (TCP), tetracalcium phosphate (TTCP), and amorphous calcium phosphates (ACP) represents the consequence of degradation.[9]

Low pressure plasma spraying (LPPS), aims to ameliorate an unacceptable porosity and chemical modification of deposited coatings resulted from stereotyped plasma spraying techniques. Increasing the volume of the chamber allows decreasing the pressure in the range of 2–13 kPa. This subsequently allows increasing the particle velocity to promote better adhesion and to hinder adverse effects of gas–metal interaction. However, the LPPS process has drawbacks in terms of raising the substrate temperature. Heimann and Vu[10] outlined the characteristics of HA coatings on Ti-6Al-4V substrates deposited by a LPPS process. They concluded that subjecting the HA powder to preliminary sintering at $\approx 1300°C$ promotes better adhesion strength and prolonged in vitro stability of the coated layer. An increased density of HA particles attained by prior sintering was believed to be the major contributing factor in the final product's properties.

4.3. *HVOF technique*

HVOF technique exploits mixture of oxygen and fuel (Propylene, Hydrogen, and Propane) fed into the combustion chamber. The continuous combustion process propels partially melted HA powder from the nozzle at approx. 800 m/s and deposits it on the substrate surface. The maximum thickness of HA coating from HVOF deposition technique can reach 6.35 mm with low porosity. Lima *et al.*[11] conducted a study of an *in vitro* behaviour of HVOF sprayed HA on Ti-6Al-4V via immersing the coatings in simulated bodily fluids (SBF) for seven days. It was reported that HA coatings are highly crystalline and formation of TCP, TTCP, and CaO were not observed.

The potential use of HVOF as an alternative technique to attain the similar or better quality HA coatings, compared to that achieved by conventional plasma spray technique is widely considered. By varying the parameters of HVOF process, such as, oxygen flow rate, propylene flow rate, air flow rate, spray distance and powder flow rate, the desired final product properties can be achieved. However, further studies, such as on bioactivity and adhesion strength should be conducted to fully assess the potential of HVOF process.

Fernandez *et al.*[12] have performed heat treatment to monitor its possible effects on bioactivity and chemical/mechanical properties of HVOF deposited HA coatings. Comparing the several important factors of raw powder, sprayed, and heat-treated coating as a function of bond strength, crystallinity, degradation products, phase analysis, porosity, and surface morphology yielded the following results. Heat-treated coatings exhibited superior adhesion strength. In addition, the bonding strength of heat treated coating remained unaltered when subjected to SBF. On the other hand, as-sprayed coatings exhibited significant worsening of adhesive strength. It has to be mentioned that heat treated coatings showed limited bioactivity compared to that of the as-sprayed one. Lower biomimetic ability may be a contributing factor for decreased porosity of heat-treated coatings.

4.4. ***HVSFS***

Another lately emerged thermal spray technique is the HVSFS process, which typically produces HA films of 75–90 μm in thickness. In this process the feeding system draws the suspension out of a reservoir and pushes it continuously with controlled flow rate into the combustion chamber of a HVOF spraying modified torch. Comparing the properties of HA coatings deposited with different thermal spray techniques, such as Atmospheric Plasma Spray (APS), HVOF, and HVSFS, the efficiencies of HVOF and APS deposited coatings were higher ($42.5 \pm 1.5\%$ and 52.5 ± 2.5 respectively as compared to 13.0 ± 1.5 for HVSFS), while the HVSFS coatings exhibited better microstructure. In addition, a higher level of crystalline HA was observed in the coatings deposited by HVOF, compared to that of APS and HVSFS. The suspension type, for example water based or diethylene glycol (DEG) with dispersed nano-HA that can be employed during the HVSFS process, affects the final microstructural properties of coatings.[13]

4.5. *EPD technique*

The method of depositing HA via EPD technique has received great attention in biomedical applications due to its simplicity, relatively low equipment cost, uniformity, and ability to control the thickness of the coatings. EPD can be used to process wide range of materials including metals, ceramics and polymers. Furthermore, by combining ceramic materials with polymers or metals, it is possible to manufacture unique composite coatings. EPD process is split in two parts: electrophoresis and deposition. Applied electrical field to stable colloidal suspension promotes movement and subsequent deposition of electrically charged HA powders on a metallic substrate.

In order to acquire the desired homogeneity of HA coatings and its close integration to a substrate, several deposition parameters like sintering duration, temperature profile, and applied current density should be carefully chosen. Ti-6Al-4V is typically used as cathode with gold or gold plated glass plate as anode for constant voltage electrophoteric deposition. A deposition of several layers is also recommended to avoid poor mechanical behaviour of HA coatings.

4.6. *Physical and chemical vapour deposition techniques*

PVD represents yet another technique used for HA deposition. Deposition of materials during PVD process is carried out under vacuum conditions. The process is broken into several steps: (1) Deposition material is converted into vapour by high energy ion beams; (2) Vapourised atoms are transported towards the substrate; (3) Condensation process of vapourised HA atoms leads to formation of thin film on metal substrate.

Radiofrequency (RF) magnetron sputtering is frequently used to deposit thin films with good adhesive properties. RF magnetron sputtering uses a RF field that generates plasma between a target and a substrate holder. The target represents the source from where the materials are ejected, and the substrate is placed in a vacuum chamber. Negative charge applied to the source causes plasma discharge. The plasma region produces gas ions which are bombarded towards the negatively charged target plate. This in turn produces a momentum transfer and ejected atoms are accelerated towards the substrate. Upon collision with the substrate HA particles form a thin film.

Other deposition techniques compatible with HA deposition for medical applications include ion beam deposition (IBD), sol-gel, pulsed laser deposition (PLD) and ion beam mixing (IBM). However, due to their slow deposition rate, these techniques have been rarely used for HA coatings on a large scale.

CVD differs from PVD, in that it deposits the solid material rather than vapourised one. CVD is a deposition technique where precursor gases are decomposed upon interaction with the heated substrate. During decomposition they form solid phase. CVD has several advantages over PVD in that it does not need ultra-high vacuum for operation. Controlling the stoichiometry can be easily performed by monitoring flow rates of precursors. Notably, novel flame assisted chemical vapour deposition (FACVD) technique is believed to have a huge potential for HA coatings. Principles of FACVD approximate that of CVD, however, avoiding the utilisation of chamber for promoting the chemical reactions makes this technique relatively inexpensive.

4.7. *CoBlast^{TM} process*

CoBlast$^{\text{TM}}$ process represents the novel and promising coating technology with reported excellent osteoconductivity and osseointegrity of HA films.[14] This process of coating is chemistry free, non-vacuum and is performed at ambient temperatures. Dopant materials can be deposited on the surface of any reactive metal. CoBlast$^{\text{TM}}$ is a variation of standard grit blasting. Conventional grit blasting is mainly used for surface roughening and material removal, while CoBlast$^{\text{TM}}$ can be employed for HA material deposition. The significant difference between this process and other coating deposition techniques is that dopant is trapped in the metal oxide layer of the surface. Another distinct advantage of the CoBlast$^{\text{TM}}$ process is that during deposition HA is unaltered. These features eliminate the need of HA heat treatment prior to its deposition and avoid formation of the calcium phosphate, which in turn hinders bonding between host bone and implant. The deposition process is fairly simple and easy to implement in manufacturing environment.

During CoBlast^{TM} process, two blast jets are employed simultaneously. One jet blasts an abrasive grit that abrades and churns up the metal surface, baring the metal beneath the oxide layer, while the second jet is used to deposit HA coating material on exposed surface. In addition, compared to widely used plasma spray technique, production flow of CoBlast^{TM} combines roughening and deposition of HA in a single step. Therefore, CoBlast^{TM} allows pointing the coating on particular area of substrate and eliminates the masking procedure. The process integrates the coating material into the reformed oxide layer of metal surface by tribochemical bonding and interlocking. Thus, the deposited bioceramic material is not prone to alteration and delamination.

Layer of HA deposited by CoBlast^{TM} is $\leq$10 μm in thickness and elicits better adhesion properties, based on the results of American Society for Testing and Materials (ASTM) tensile (79 MPa) and shear (48 MPa) stresses. Notably, EnBIO has dispatched OsteoZip^{TM} HA surface that aims to promote rapid integration and fixation of orthopaedic implants with host bone. It may be used as a coating material for metal implants in orthopaedic applications. Osteozip^{TM} is HA surface sprayed by novel CoBlast^{TM} technique that aims to enhance osseous integration and speed up the bonding process between the orthopaedic implant and bone.

5. Thick Film Technology

The manufacture of thick film structures for medical sensing devices can be achieved using a number of approaches, with screen-printing being the most popular. This is due to the fact that this technique is cost-effective, robust, and versatile, giving the opportunity to produce complex structures with a range of materials, from biocompatible metals and coatings to polymers, printed on virtually any substrate.

Therefore, the screen-printing technique was chosen as a way to produce thick HA films so that their pressure sensing potential can be assessed. This in turn leads to the unique opportunity to deposit functional HA films in a broad range of medical applications.

Thick-film technology remains a popular manufacturing method for many years and its main application is in the production of hybrid microelectronic circuits for use in telephones, automotive electronics, and especially for a broad variety of sensors, including gas sensors, pressure, humidity, radiation, and for biomedical applications.

The major steps in the screen-printing of thick-film structures are printing, drying and firing/curing the functional layer at elevated temperature.[15] Modern screen-printing technological process normally automatically forces an ink or paste through a stainless steel mesh, using a squeegee onto the substrate below it. The substrate acts as a physical support for the thick film and the choice of the substrate material is also dictated by the considerations of mechanical strength, smoothness of surface texture to promote good film adhesion, chemical, and physical compatibility with the fired thick film, high electrical insulation resistance to prevent electrical leakage currents between closely spaced conductor lines, low thermal expansion coefficient to prevent thermal mismatch, high thermal stability to prevent decomposition during processing, biocompatibility if required, and not least the desire of cost-effectiveness.

The quality of the printed layer is largely affected by the choice of mask. Elements such as the material used, the type of coating, the accuracy of the pattern, the alignment of the pattern to the mesh, the frame type and method of mounting and supporting the frame for printing are important. Polyester mesh was used in this work and the semi-automatic screen printer model was DEK 1022.

The main requirement for the thick film paste is that it must be able to flow through the screen and retain its intended shape on the substrate beneath. This depends largely on the pastes flow properties, viscosity, and particles size in particular.[16] Cermet pastes have a functional ingredient, solvent, temporary binder and permanent binder and require firing at high temperatures. For polymer pastes, there is no temporary binder and pastes are dried at temperatures in the region of 120°C–250°C.

The main purpose of the squeegee is to bring the screen into contact with the substrate, push the paste through the stencil to the substrate, to shear the paste level with the top of the screen in order to obtain uniform thickness and to control the rate at which the screen peels away from the substrate. A classic 'trailing edge' squeegee type was used throughout this work to print the HA layers and Ag electrodes, as it is flexible and exerts uniform pressure on the screen.

After the screen-printing the substrates were left to stand in air for 5 min to allow the paste to settle. To remove the organic solvents from the printed layer, so that it can take its final form and be immune to smudging, the substrates with printed silver electrodes were then placed in a conventional oven at temperature of 100°C for 1 h. Drying also improves adhesion of the printed layers to the substrate.

The term "firing" refers to a high temperature cycle of up to 1000°C, the purpose of which is to remove the temporary or organic binder from the film, sinter the permanent or inorganic binder, and to develop the electrical properties of the paste, while ensuring the film's adequate adherence to the substrate. However, low-temperature silver paste was used for this work to avoid the firing step. This reduced the cost of manufacturing without compromising the films quality and is more in line with the requirements of medical devices industry.

The film layers produced using the screen-printing technology typically measure 5–200 μm, and can be deposited on virtually any suitably prepared substrate. The main attractions of screen-printing for medical devices are its versatility, reliability and reproducibility, and cost-effectiveness even at small or medium scale production. For highly specialised application, thick film technology offers the manufacture of devices that are robust, can be miniaturised, can be integrated onto the same substrate as the electronics and the printed pattern can act as an active sensing component.[15,17]

6. Manufacture and Testing of Pressure Sensing Properties of HA Thick Films

To prepare HA paste for screen-printing, 3 g of HA powder with a typical particle size less than 25 μm, supplied by Cam Bioceramics (the Netherlands) was mixed with 7 wt.% of polyvinyl butanol

(PVB) as organic binder and 3 wt.% of glass frit. Further, addition of diethylenglycolmonobutylether as a solvent improved the rheological properties of paste.

Upon deposition of each layer, films were allowed to dry at room temperature for 10 min and then cured for 1 h in an oven. These steps were successfully repeated before depositing additional layers on each sample to produce films with various thicknesses.

Characterisation of electrical properties based on real-time changes in the capacitance values as a function of applied pressure was performed at 10 kHz using the HP Impedance Analyser 4192A LF. Area of HA layer subjected to pressure application was $3.17 \cdot 10^{-5}$ m^2. Pressure ranging from 15 to 140 MPa was applied on samples covered with 14 μm, 20 μm, and 32 μm thick HA films. During the experiments the pressure applied on the samples was gradually increased and the changes in the values of capacitance were recorded with the sampling rate of 1 s.

7. Experimental Results Overview: Reversible Pressure-induced Changes in HA

Figure 1 depicts a plot of real time changes of capacitance values as a function of gradually applied pressure to 14 μm thick HA film. Screen printed sample exhibits the direct relationship between the

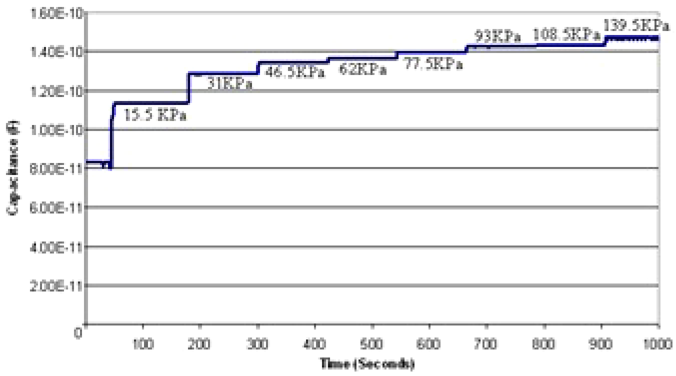

Fig. 1. Real-time capacitance changes due to pressure application in 14 μm thick HA film.

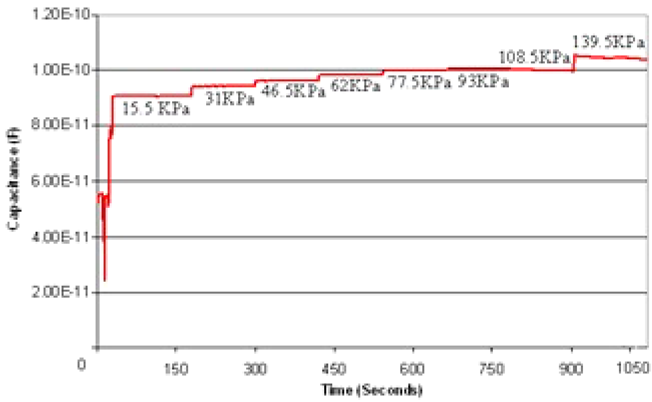

Fig. 2. Real-time capacitance changes due to pressure application in 20 μm thick HA film.

increase in pressure and the value of capacitance. The plot can be clearly partitioned in two parts. Application of 15.5 kPa yields rapid change in the value of capacitance and this phenomenon remains the same for further loading. However it can be discerned that rapid changes in capacitance values of 14 μm thick HA sample seems to dissipate. Minor changes in capacitance can be observed as the pressure overlaps the 46.5 kPa value and reaches the value of 147 pF as it is subjected to 139.5 kPa.

Real-time capacitance changes as a function of applied pressure to 20 μm thick HA layer are depicted in Figure 2. A sudden shift in capacitance value from 55.2 to 90.5 pF in the first stage of pressure application is clearly noticeable. Further increases in the capacitance values are minor, similarly to 14 μm film behaviour. In addition, capacitance value seems to stabilise in the pressure range from 93 to 108.5 kPa, while an upward shift is noticeable once the pressure equalises the value of 139.5 kPa. At the final stage of measurement capacitance reaches the value of 10 nF.

In line with thinner films, 32 μm thick HA layers showed an appreciable increase in capacitance from 46.6 to 63.8 pF when subjected to 15.5 kPa, with graduate but less pronounced further increase in capacitance with load, reaching 70.5 pF at 139.5 kPa. Notably, these results correlate with the theory as set in Section 2, specifically the

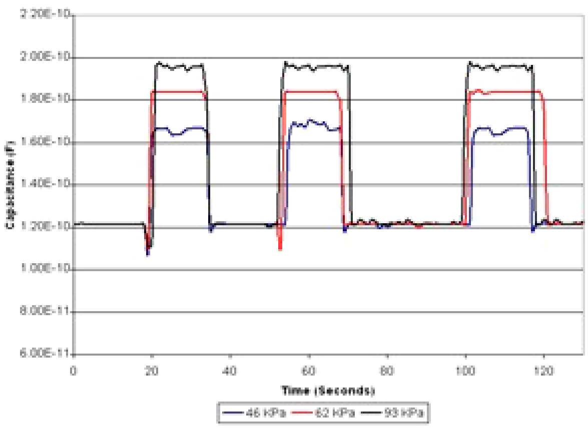

Fig. 3. Application and removal of various amount of pressure on 14 μm HA sample.

applied pressure results in increased capacitance values, which is inferred to be due to the film thickness reduction/film compression.

In order to further investigate the possible piezoelectric effect in screen printed HA thick films, they were subjected to a series of loading and subsequent unloading. In particular, HA films were subjected to 46, 62, and 93 kPa pressure. Upon changing the capacitance values, the load was kept for several seconds for reliable measurements. Once the changes were observed, applied load was released to monitor the reversal of capacitance values. Steps of loading and unloading were repeated for three consecutive times to address response-repeatability issues. Figure 3 illustrates the behaviour of 14 μm thick sample subjected to various amount of pressures during the loading/unloading tests.

Analysing this data reveals that the application of pressure yields sudden shift in the capacitance, while upon removal of it, the capacitance falls to its value in unloaded state. Furthermore, behaviour of the sample upon loading with the same amount of pressure and subsequent withdrawal of it closely approximates that exhibited in the previous load/unload step. This type of behaviour is replicated for various amounts of pressure. Loading/unloading while utilising the same values of pressure was also performed on 20 μm and 32 μm samples with identical trend.

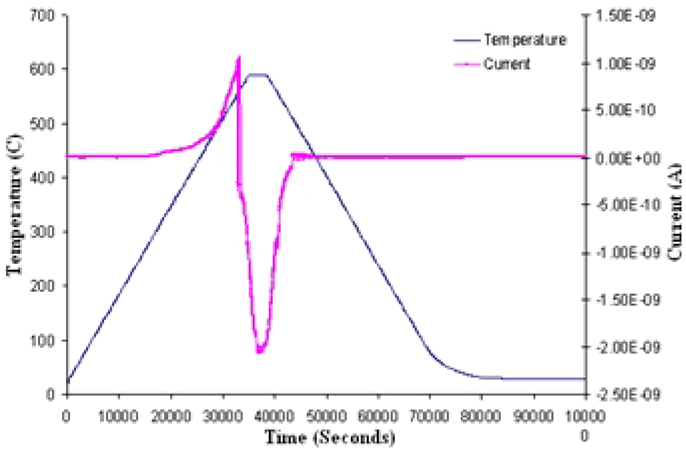

Fig. 4. Pyroelectric surface charge in screen-printed HA thick films (Reprinted with permission from Ref. 18.)

Another objective of pressure application experiments was to identify the possible pyroelectric effect in screen printed HA samples. Capacitance values of all the samples tested in aforementioned experiments underwent noticeable changes upon application of pressure. In addition, removal of pressure was instantly followed by returning the capacitance values to their initial values.

To further confirm the pyroelectric effect in screen-printed HA thick films, they were subjected to poling at $4\,\mathrm{kV/cm^2}$ for 1 h.[18] Subjecting the sample to the peak temperature of 550°C while the heating/cooling rate of 3°C/min showed a pyroelectric behaviour of HA thick films. Figure 4 reproduces the pyroelectric effect that took place during poling the sample in an aforementioned environment and shows the charge storing capabilities of HA films.[18]

Pressure induced changes of capacitance further confirms piezoelectric and pyroelectric nature of HA postulated by Tofail *et al.*,[19] who measured pyroelectric constant of poled HA ceramics to be in $0.1 - 40$ nC $\mathrm{cm^{-2}\,K^{-1}}$ range at $300 - 500$°C. Piezo and pyroelectrical properties of un-poled HA thin films was noted in the study by Lang *et al.*,[20] during which HA thin films were spin-coated on silicon wafers.

These properties of screen printed HA thick films make them attractive option for real-time *in vivo* pressure sensors, as response and recovery times were ranging from 0.9 to 3.2 s for HA films, with shorter time for thinner films. However, thicker sample showed higher sensitivity to pressure as compared to relatively thin ones, not least due to a porous nature of thick HA films.

8. Conclusions

Thick film technology, namely screen printing technique, was utilised to deposit various thickness of novel HA thick film on alumina substrates, with the aim of their possible implementation in the field of biomedical industry. Specifically, HA films can be used as a coating for bone and dental implants, as well as biocompatible pressure sensors, due to their reversible changes in electrical properties under the influence of applied pressure. The additional advantage of the coating in the form of screen-printed thick films is that due to the technology's versatility, it can be produced on flexible substrate in any shape and size to suit the needs of various patients.

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