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Chapter 5

Laser Ablation of Biomaterials

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Additional information is available at the end of the chapter

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Abstract

Biomaterials, defined by high biocompatibility and biodegradability, are intensively used in medical applications, mainly to replace partial or total, damaged or destroyed hard or soft tissues. Most of them are used not only as coatings for implant coverage but also as parts for some medical devices. In the last decades, researchers sought to find the optimum processing methods and parameters to modify or deposit the biomaterial of interest. An important family of techniques, used to process a biomaterial, is represented by laser techniques, based upon laser ablation phenomenon. Laser ablation of biomaterials ensures the transference or modification with good precision and without or with minimal disruptions generated. To obtain thin coatings from biomaterials, one can use deposition techniques: pulsed laser deposition (PLD) or matrix-assisted pulsed laser evaporation (MAPLE). These techniques are chosen according to the selected biomaterial and desired performances of the obtained coating. Therefore, some sensitive biomaterials can be transferred only by MAPLE. Some results in the field of calcium phosphates deposited by PLD or MAPLE are presented, proving the usefulness of these biomaterials for medical applications.

Keywords: laser ablation, pulsed laser methods, spallation, biomaterials and medical applications

1. Introduction

In the last few years, the research on microfabricated devices or implants for biomedical applications has quickly advanced [1, 2]. The aim of biological implants is to reinforce or replace the damaged or diseased tissue [3]. The presence of a biomaterial in the body after

implantation procedure always causes a biological reaction [4]. This is a host response to the new living conditions and suggests that the body is trying to adapt at the given situation [5].

Every year, for a large number of patients, biomaterials save lives, relieve suffering, and improve the quality of life [6]. To achieve the needs of the biomedical society, materials comprise everything from metals and ceramics to glasses and polymers have been investigated [7].

Biomaterial performances can be improved by selectively modifying the surface properties. One simple solution could be the deposition of thin films useful for changing the chemical and physical properties of biomaterials and important in achieving the optimal surface [8]. There are many studies which demonstrated that the cell response is dependent on surface topography [4].

Laser processing of natural or synthetic biomaterials has been rapidly developed to produce biomimetic artificial organs, tissue engineering scaffolds, and other biomedical constructs [9]. The laser processing techniques are commonly grouped into three categories: polymerization (use of laser to induce cross-linking between biomaterial polymer chains), ablation (use of laser to selectively remove part of the biomaterial by thermal or chemical effects), and activation (use of laser to activate certain parts on the polymer chains for specific application) [9]. Relative to other methods, laser processing of biomaterials presents some advantages: reduced surface contamination and mechanical damage and the capability to produce three-dimensional components with complicated geometries by controlling the surface structuring [4].

The physics behind the interaction of laser pulse with the solid surfaces can help us to better understand the laser ablation mechanism [10].

This chapter will provide a brief overview in the field of laser ablation of biomaterials, starting with the explanation of ablation mechanism and then presenting some of its applications.

2. Biomaterials

Biomaterials science is a multidisciplinary domain, which involves various features of materials science, clinical science, chemistry, physics, biology, and medicine [11, 12]. It is also an exciting and quickly growing field. It is to be mentioned that a biomaterial is different from a biological material, the last being produced by a biological system [13]. A biomaterial is a natural or synthetic material used in a medical device, projected to interact with biological systems for direct medical treatment [6, 7, 14]. Their usage within a physiologic medium needs some specific characteristics such as biocompatibility (to perform the function with an appropriate host response), efficiency, and reliability [7, 14]. These representative features have provided with an appropriate combination of chemical, mechanical, physical, and biological properties [14]. Biomaterials used in medical applications are rarely used as simple material, being mostly integrated into devices [1].

The biocompatibility of a biomaterial implies its acceptance by the surrounding tissues and the whole body. Consequently, implanted biomaterials should not irritate the injured area,

provoke an abnormal inflammatory response, stimulate allergic or immunologic reactions, and cause other diseases [15]. Besides these characteristics, the selection of biomaterials for the manufacturing of an implant device and/or as thin films for the functionality improvement of implant should also depend on appropriate mechanical properties (strength, stiffness, and fatigue), proper optical properties, proper density, sterility, and long-standing storage [15].

The most commonly used metallic implants are based on titanium and its alloys due to their excellent mechanical properties and corrosion stability [16]. Unfortunately, studies in the field showed that bare titanium or other metallic implants can determine ion release into body [17]. To prevent, control or diminish the ion release, such implants are covered with thin films.

Nowadays, there are a lot of experimental studies focused on the growing of thin, uniform, and adherent films from polymers, organic materials, and biomaterials [18].

2.1. Hydroxyapatite

Hydroxyapatite (HA) named by Berndt et al. as "hydrated calcium phosphate mineral" is the main inorganic component of the hard tissues (bone and teeth) and is the most extensively studied materials for bone healing [1, 19, 20]. HA belongs to the "apatite" group of compounds, having the chemical formula $Ca_{10}(PO_4)_6(OH)_2$ and a Ca/P ratio of 1.67 [19, 21]. It can be termed as hydroxylapatite, calcium hydroxyapatite, or apatite and has a calculated density of 3.22 g/cm^3 [19]. The vacancies or substitutions, which could occur within the structural lattice, have therefore the non-stoichiometry (deficiency of Ca^{2+} and OH^- ions) of biological apatite. Research in the field demonstrated the poor crystallinity of the biological apatite [22]. On the contrary, synthetic HA is considered to be a stoichiometric material [19]. It is to be mentioned that the presence of some crystallographic sites in the structure of HA allows the atomic exchanges of specific elements with different ionic charges (Na^+, Mg^{2+}, HPO_4, K^+, CO_3^{2-}, F^-, Cl^-, or trace elements such as Sr^{2+}, Ba^{2+}, and Pb^{2+}) [19, 22].

The maximum sintering temperature of HA, up to which its structure remains unperturbed, is 1300°C. If one exceeds this threshold, the HA decomposition cannot be avoided [23].

Due to the similarity with the human bone, hydroxyapatite was largely used in medical applications. The influence of some HA parameters with respect to the structural, mechanical, and biological properties has been investigated [19].

One important feature of HA, used as thin film for medical implants, was the stability in the physiological media, presenting a dissolution rate of 0.1 mg/year [19, 24].

Hydroxyapatite can be considered a "smart" ceramic due to its functionalities such as fixation improvement and stabilization of implant [19].

2.2. Octacalcium phosphate

Octacalcium phosphate (OCP), with the chemical formula $Ca_8(HPO_4)_2(PO_4)_4{\cdot}5H_2O$ and a structure similar to calcium OH-apatite, is more soluble, being present in the first stage of tissue growth [25, 26]. OCP has a Ca/P ratio of about 1.33 and can induce the ectopic development of a new bone tissue [26, 27]. A disadvantage of OCP is represented by the thermal instability

of this material. The decomposition takes place at temperatures lower than those needed for ceramic sintering. It was observed that:

– at ~90°C, OCP lose water;

– at 180°C, the OCP structure undergoes changes by interlayer bond breaking;

– at 300°C, the pyrophosphate appears from OCP [28].

The immersion of OCP in physiological fluids induces the conversion into bone-like apatite spontaneously and irreversibly [29].

Related studies revealed the potential use of synthetic OCP as bone substitute material in different forms, such as coatings and granules. Furthermore, the stoichiometry of synthetic OCP controls the osteoconductivity and biodegradability of this material and recommends it to use in bone regeneration [30].

Boanini et al. evaluate the possibility to deposit thin films of magnesium substituted OCP (Mg:OCP) and strontium substituted OCP (Sr:OCP) on Ti substrates [31].

2.3. Carbonated hydroxyapatite doped with manganese

Manganese is essential for a normal growth and function of bones and muscles [32]. Furthermore, Mn^{2+} ions increase the capacity of integrins (transmembrane receptors that mediate cellular interactions) to bind and facilitate cell adhesion [33]. The doping of hydroxyapatites with Mn^{2+} ensures a better connection of bone-implant and makes easier the regeneration of bone tissue. It was demonstrated that Mn-doped HA has a higher thermal stability than pure HA [32]. The results obtained by György et al. demonstrated that carbonated HA doped with manganese (Mn-CHA) films should determine a faster osteointegration of the implanted device [34].

To modify the biomaterial surface properties, several methods were applied [3]. One can mention chemical treatments, ion beam implantation, liquid immersion, thermal and plasma spray, electrophoretic processes or laser processing methods based on laser ablation [3].

3. Laser ablation

Laser, a versatile source of energy, is for sure an attractive tool with prospective applications in communication and military technology, industry, scientific research, and medicine [35, 36]. The continuous development of lasers and the impossibility of other techniques to deposit or modify some materials leaded to use lasers for material processing [3, 36]. The interaction between laser beam and material is a complex action, which needs the understanding of capabilities and limitations of this process. It involves the incidence of electromagnetic radiation on the material surface, followed by reflection, refraction, absorption, scattering, and/or transmission phenomena (**Figure 1**) [37]. Absorption of radiation in materials is an important and desired phenomenon and is governed by the excitation of free or bound electrons, inside the bulk material [35, 37].

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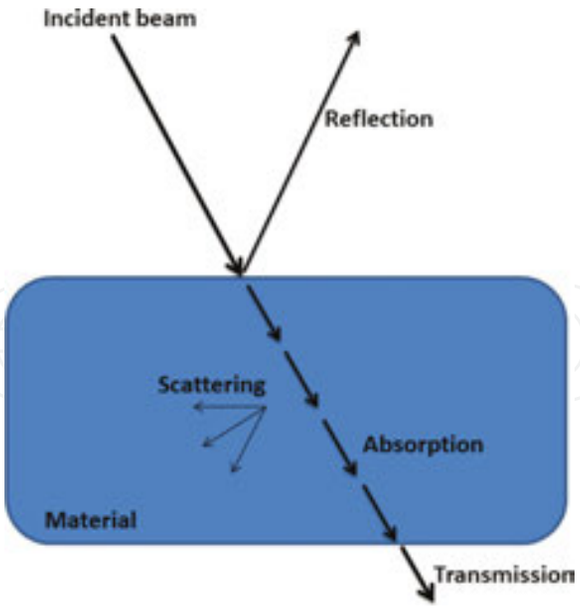

Figure 1. Possible phenomena generated by the interaction of laser beam with material.

Being the base of laser material processing, laser ablation consists in the removal of material from a substrate/target by absorption of laser energy [18, 38]. The ablation process always depends on the laser parameters like wavelength, energy, fluence, pulse duration, and the material properties, such as chemical structure, melting temperature, thermal diffusion rate, optical reflectivity, and the ambient medium [39, 40]. Laser ablation has significant applications in surface modification and deposition of thin films. Ablation is commonly determined by a phase transition from the liquid state to the vapor state. In function of the physical processes that take place, a classification of laser ablation in the following:

– thermal ablation: the predominant phenomenon is heat induced by laser and vaporization;

– photophysical ablation: the ablation rate is directly influenced by non-thermic excitation;

– photochemical ablation: the chemical bonds are interrupted by direct dissociation or by indirect transfer of energy via defects and impurities [41].

Laser ablation, also known as ablative photodecomposition, is of key importance in the field of material processing [38, 42]. Depending on the surface (from flat to rough), the ablation condition would gradually change [43]. The ablation process is also influenced by the laser beam parameters, thermo-optical properties, and ambient conditions [43]. The selection of proper parameters helps us to achieve the desired material modification [3]. In laser ablation, the stoichiometric transfer and controlled delivery of target composition to the substrate are possible, this process being a non-equilibrium process [44].

Laser ablation of biomaterials is physically defined by four main parameters: $d(F)$ (ablation rate per pulse), α_{eff} (effective absorption coefficient), F (irradiation fluence), and F_{th} (ablation threshold fluence), being frequently described by the equation [45, 46]:

$$d(F) = \frac{1}{\alpha_{eff}} ln\left(\frac{F}{F_{th}}\right). \quad (1)$$

The ablation threshold fluence initiates the plasma ignition and is described as the minimal energy of the laser pulse per surface unit [41].

Depending on the selected biomaterial and of laser processing parameters (wavelength, energy, and target-substrate distance), material removal during laser ablation is accompanied by a variety of mechanisms [3]. At high-laser intensities, a significant volume of the bulk material is directly excited producing plasma [3]. This can be considered as the fourth state of matter and consists of extremely excited atomic, molecular, ionic, and radical species [8]. The propagation and expansion processes of plasma are dominated by mechanical interactions [43].

It is to be mentioned that a complex cascade of strongly related processes happens in the vicinity of the target surface during and after laser-material interaction [47]. Laser ablation of biomaterials is also based on the strong absorption of laser photons by the investigated material. To achieve a maximum absorption, the wavelength has to be carefully chosen [48].

In the nanosecond regime, laser ablation is induced by rapid heating of the surface layer [49]. In our case, all experimental results were obtained on coatings deposited using a KrF* excimer laser (λ = 248 nm and pulse duration τ = 25 ns).

3.1. Pulsed laser deposition

The common method used for thin film synthesis and associated to laser ablation is pulsed laser deposition (PLD). In this process, the ablated material from target to substrate is pushed by the PLD plasma, which acts as a piston (**Figure 2**). Furthermore, the absorbed photons by irradiating a solid target with a laser beam of high intensity can generate the fusion and local vaporization of target. In PLD, ablation can be defined as the rapid boiling of material in a localized interaction volume on and in the vicinity of target surface (**Figure 3**) [50].

Figure 2. HA plasma expansion.

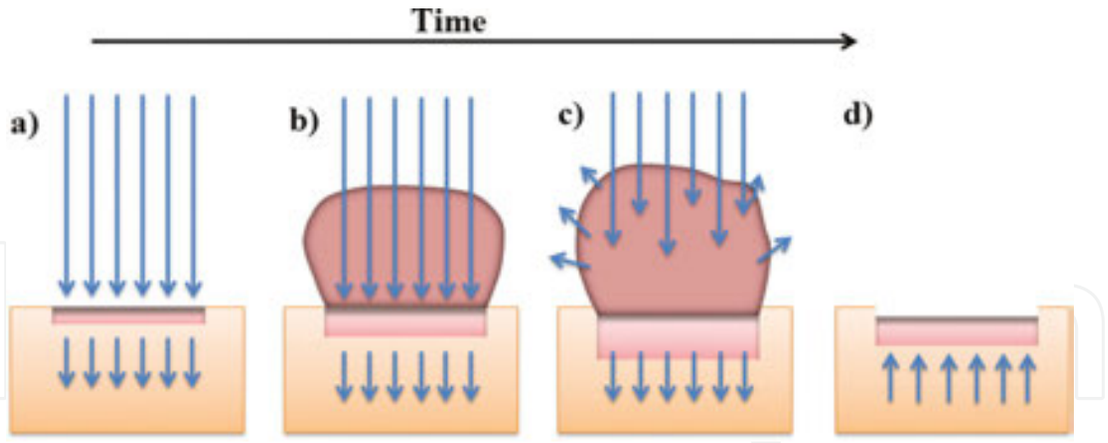

Figure 3. Schematically representation of laser ablation process. (a) The initial absorption of laser radiation (long arrows), the beginning of melting and vaporization. (b) The melted material propagates into solid, vaporization continues, and the interaction laser-plume becomes visible. (c) Absorption of incident laser radiation and plasma formation. (d) The melted material is ejected and re-solidification takes place [50].

The product of this interaction, known as laser ablation phenomenon, produces plasma of complex composition: photons, electrons, ions, atoms, molecules, clusters, and even liquid or solid particles [51]. The ejected material from the target is in a thermodynamic equilibrium and can be associated to thermic and non-thermic mechanisms. A substrate parallel with the irradiated target was used to collect the vaporized material. The deposition of a thin layer on the substrate is the result of repeated laser pulses [52]. The target ablation and deposition rate are dependent on the material or combination of materials present in the target. The ablation threshold is dependent of absorption coefficient and wavelength [53]. As compared to films deposited by other methods, PLD films also exhibit superior performances.

An important feature of PLD is the stoichiometry preservation of the target in the deposited thin films by choosing proper ablation and deposition parameters [54, 55]. Furthermore, in PLD, the deposited material exhibits an excellent adherence to substrate, with a controlled growth rate and without contaminations. This technique also offers the possibility to deposit multi-layers and doped films with great versatility [54]. The variation in the experimental parameters offers the possibility to deposit crystalline structures from various complex materials deposited at room temperature [55].

Nevertheless, PLD presents some limitations as low deposition rate, use of compounds that are not sensitive to thermal decomposition and degradation, and restricted deposition area [52].

3.2. Matrix-assisted pulsed laser evaporation

A specific tool appropriate for ablation of organic and inorganic materials is matrix-assisted pulsed laser evaporation (MAPLE; **Figure 4**). The material ejection and film formation in MAPLE process are also generated by photophysical interaction between laser and target material [56]. MAPLE technique proved to be an appropriate method to obtain high quality thin films by gentle laser transfer onto any substrate of interest [57]. MAPLE emerged as an

alternative to PLD in order to preserve the chemical bonding or conformation of delicate materials. On the other words, MAPLE, a less damaging technique, is used to transfer, from the condensed phase to the vapor phase, organic and polymeric compounds, including small and large molecular weight species [58].

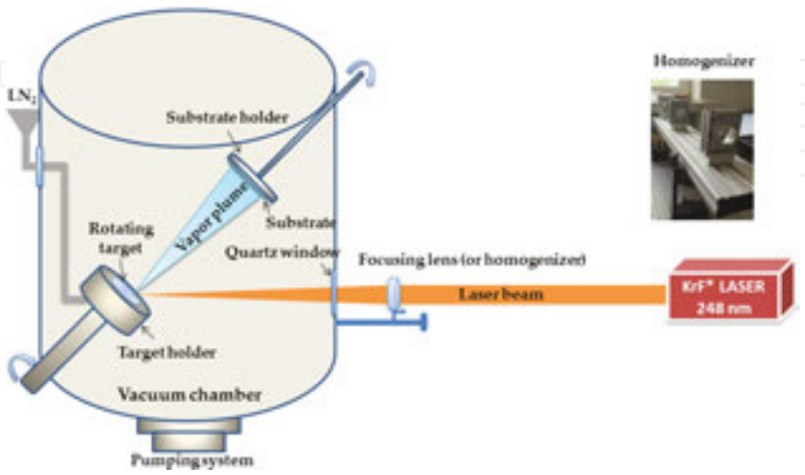

Figure 4. Typical MAPLE experimental set-up.

MAPLE, a non-contact technique, preserves the PLD advantages, allowing a better control of film thickness and morphology and enhancing the adherence of film to substrate [59].

To avoid significant material decomposition, one can optimize the MAPLE deposition conditions: laser wavelength, fluence, solvent type, concentration, target-collector separation distance, temperature, repetition rate, background gas, and pressure. The selected solvent should absorb most of the laser energy and should not react with the studied material [60].

In case of MAPLE, a cryogenic complex target of a dilute/suspension mixture of a material to be deposited and an appropriate solvent is irradiated using a laser beam of low energy. The target is maintained frozen during the laser irradiation and deposition process using liquid nitrogen [59]. This technique is governed by two photothermal processes, which takes place in the matrix, the evaporation of frozen complex target and material release into the chamber. The conversion of photon energy (absorbed by solvent) in thermal energy generates material heating and solvent vaporization [61].

MAPLE offers the possibility to produce uniform, ultrastable, and nanostructured coatings [59].

3.3. Laser spallation

In an extension of laser ablation, laser spallation is a progressive process in which the laser pulse is applied on the rear surface [62]. As known, spallation is defined as a damage occurred at the interface between inner film and substrate. Some examples are interface delamination, film spallation, and film expulsion. As illustrated in **Figure 5**, spallation can be described as a dynamic tensile fracture [63].

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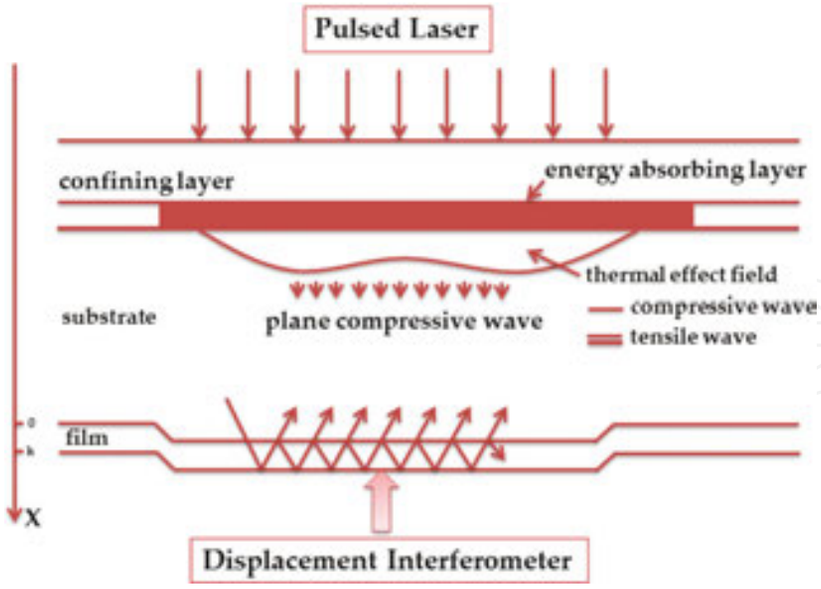

Figure 5. Laser-induced film spallation process: schematic diagram [63].

In 1992, Gupta et al. [64] used laser spallation to measure the strength of planar interfaces. To achieve this, a laser pulse of a high enough energy and a pre-determined duration was converted into a pressure pulse of a critical amplitude and width that was sent through the substrate toward the free surface with the coating [64].

Extreme superheating/deformation conditions generated by laser processing are related to mechanisms such melting and/or photomechanical damage/spallation [65]. By increasing the fluence over the spallation threshold, the composition of the ejected plume rapidly changes. The modifications consist in conversion from liquid deposits and large droplets to a blend of individual atoms, small droplets, and atomic clusters [66].

Nevertheless, it was found that certain laser processing conditions can generate undesired and very detrimental rear surface spallation effect at the textured front surface, even if there are used laser conditions well below the spallation threshold for planar surfaces [67].

4. Study of thin film growth mechanisms by laser ablation

One should take into account that coatings are obtained by successive ejected material from the target by each laser pulse. We will start the discussion by presenting some representative results from our studies related to the processing of hydroxyapatite by laser ablation.

In **Figure 6**, the surface of a HA target before and after laser irradiation is presented. The morphology is characteristic to a material melted and then resolidified. The details presented in **Figure 6c**, at a higher magnification, are specific to HA solidification, in fractal form. All circular bumps indicated the presence of the bubbling phenomenon. Moreover, the cracks appeared on target surface are due to expansion-contraction cycles as a result of repeated heating/cooling processes.

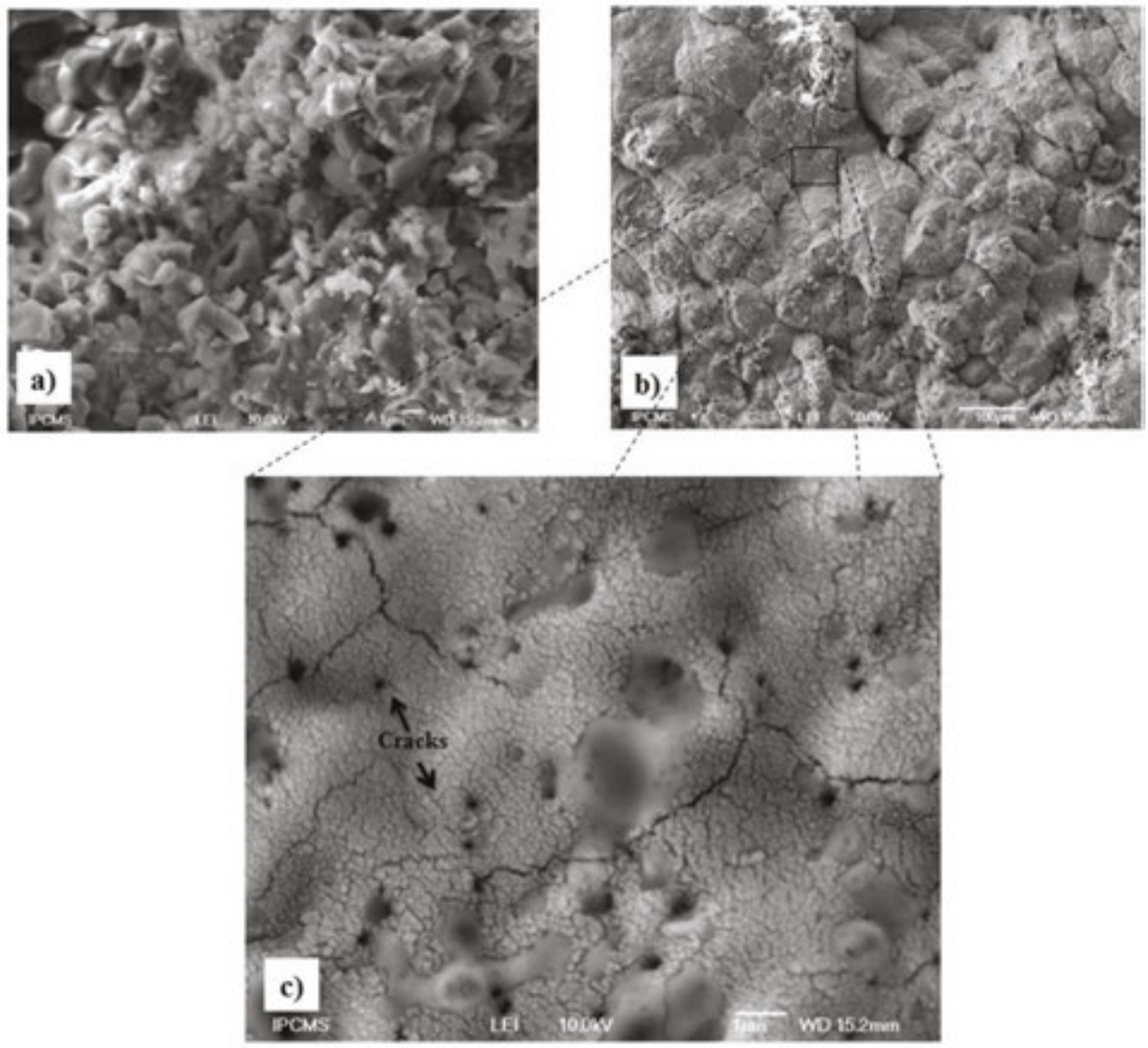

Figure 6. SEM micrographs of HA target before (a) and after (b and c) laser irradiation.

Typical HA structures deposited by laser ablation at different laser fluences and studied by Scanning Electron Microscopy (SEM) are presented in **Figure 7**. It is obvious that the aspect of deposited coatings varies from acicular at low fluence to cauliflower aspect at high fluence, respectively.

In case of samples HA1 and HA2, the coatings seem to be the product of material condensation originated from plasma. The droplets, even if they are present, are of small dimensions.

These coatings were also investigated By Energy Dispersive X-ray Spectroscopy (EDS) and X-ray Photoelectron Spectroscopy (XPS). EDS gives us information about the composition, on the entire thickness of the layer, while XPS provided information exclusively from surface.

Only three of the samples (HA1, HA3, and HA5) have been biologically investigated *in vitro* by cell growth. The cells used for these analyses were osteoprogenitor human cells (HOp) from bone marrow. HOp were cultured in ISCOVE (Sigma I 3390) medium, supplemented with 10%

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(v/v) fetal bovine serum, 1% (v/v) glutamine, and 1% (v/v) penicillin and streptomycin. After 5 days, the samples were prepared for SEM investigation (**Figure 8**).

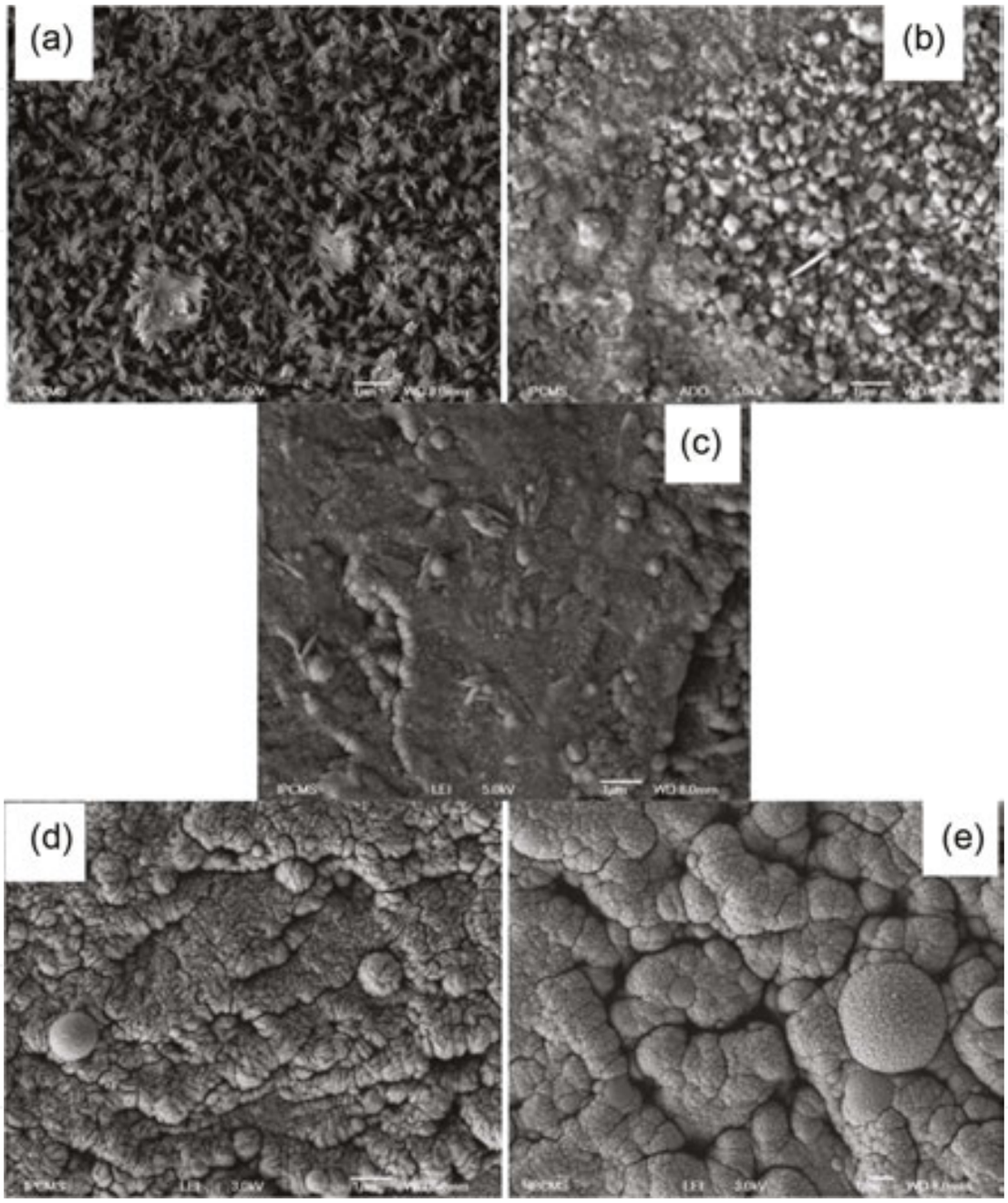

Figure 7. SEM micrographs for HA coatings deposited at (a) 1.2 J/cm^2 (HA1), (b) 1.8 J/cm^2 (HA2), (c) 2.7 J/cm^2 (HA3), (d) 5 J/cm^2 (HA4), and (e) 7.5 J/cm^2 (HA5).

From **Figure 8a**, one can observe that HA1 sample was destroyed, the coating being completed and dissolved in the culture medium. We could not identify cells on the sample surface. **Figure 8b** demonstrated a partial dissolution of HA3 coating. No cell was present on the sample surface. In case of HA5, one can observe an important coverage rate of the coatings deposited

at 7.5 J/cm^2 in the presence of cells. The morphology of the coating was not modified, while the osteoblasts present a polygonal flattened form.

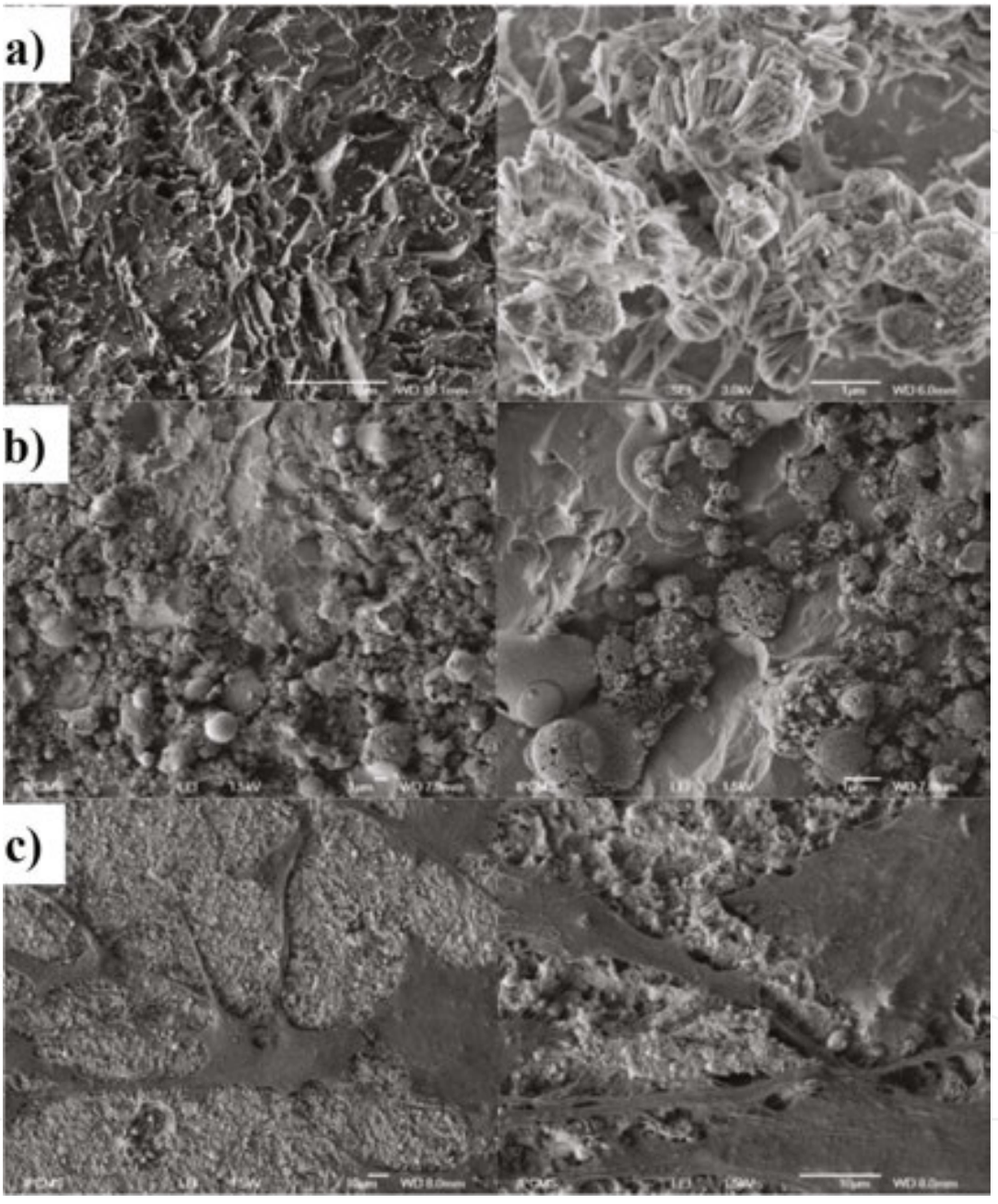

Figure 8. SEM micrographs for HA coatings deposited at (a) 1.2 J/cm^2 (HA1), (b) 2.7 J/cm^2 (HA3), and (c) 7.5 J/cm^2 (HA5) after 5 days of cell growth.

A similar investigations on HA coatings were conducted by Zhu et al. They analyzed the behavior of MC3T3-E1 cells cultured on the specimens after 7 days [68].

In another study, some of the HA coatings grown by PLD using a laser fluence of 3 J/cm^2 were thermally treated (at 400°C for 6h) in order to improve the crystallinity.

After a first examination, the SEM images of this type of structures did not revealed significant differences (**Figure 9**).

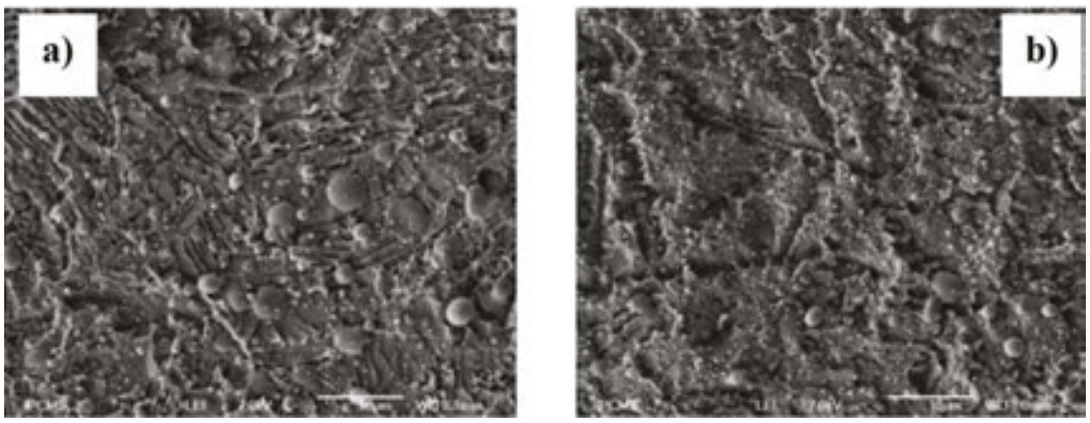

Figure 9. SEM micrographs of HA coatings deposited by PLD (a) with (HAt) and (b) without thermal treatment (HA).

SEM micrographs, at higher magnification, showed the differences induced by the thermal treatment (**Figure 10**).

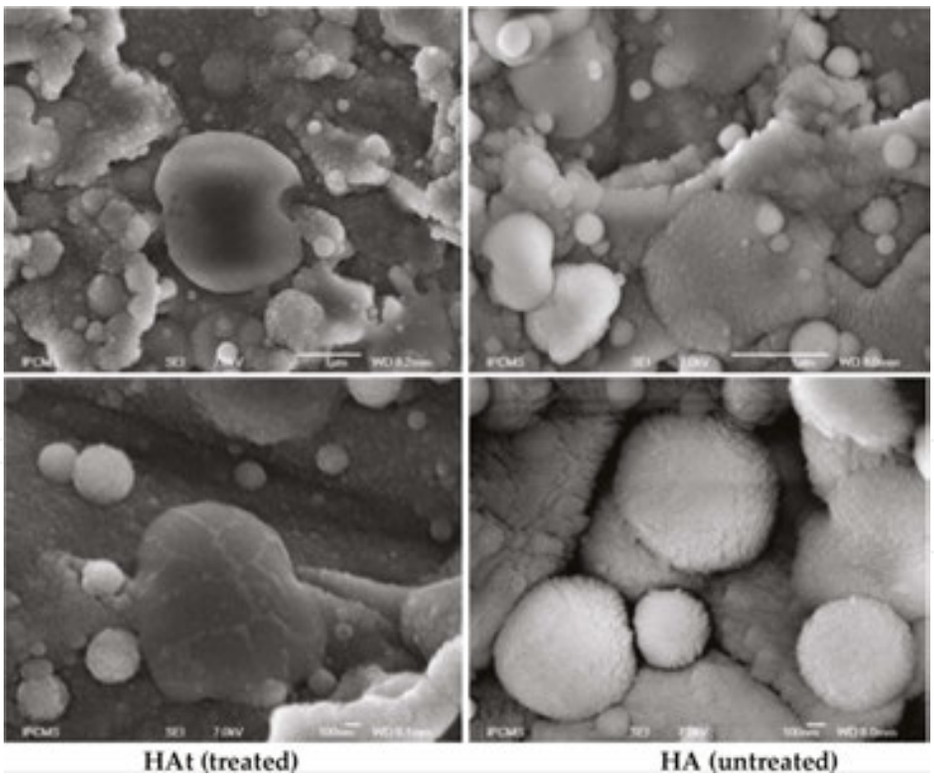

Figure 10. SEM micrographs of HA coatings synthesized by PLD with (left) and without (right) thermal treatment.

It can be seen that HA coatings present a more rough morphology at nano level. The droplets' shape is similar to that of snowballs, and the surface is made up of parallelepipedic structure.

The thermal treatment induced the surface structure reorganization. There are no longer irregularities, and the appearance of the droplets is smooth.

The Atomic Force Microscopy (AFM) analysis revealed differences between the two types of structures related to their roughness (**Figure 11**). The decrease in the rough value from 1.01 to 0.8 nm, at nanometric scale, proves the smoothing of the target.

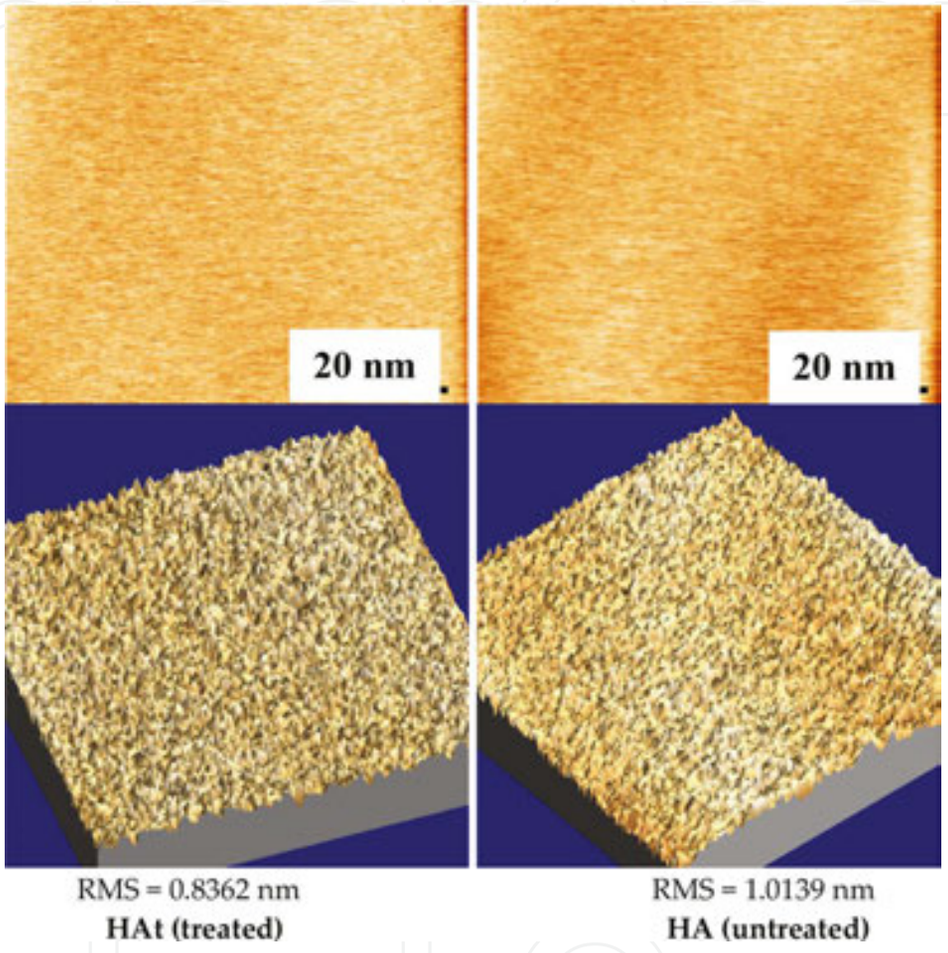

Figure 11. AFM imaged for HA coatings deposited by PLD with (left) and without (right) thermal treatment.

EDS results showed that the value of Ca/P ratio diminishes from 2.04 (untreated sample - HA) to 1.63 (treated sample - HAt). The corroborated results demonstrated the importance of thermal treatment in obtaining crystalline hydroxyapatite, biocompatible, having a structure similar to stoichiometric HA.

HA coatings were also grown by other techniques, such as thermal spray, high velocity oxy-fuel (HVOF) techniques, and plasma spraying trying to find, as in PLD depositions, the optimal conditions for good film with applications in medicine [19, 69].

A thermal treatment was also applied to Mn-CHA and OCP films obtained by PLD. In case of Mn-CHA coating, the Ca/P atomic ratio obtained by XPS and EDS investigations was 1.64–1.66, close to the stoichiometric values.

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The morphologies of the two structures, OCP and Mn-CHA, are quite different. OCP has a porous and arborescent-like structure (**Figure 12a** and **b**), and Mn-CHA has a granular and more compact (**Figure 12c** and **d**). The surface morphologies of both calcium phosphates are well matched for bone tissue growth and osteointegration.

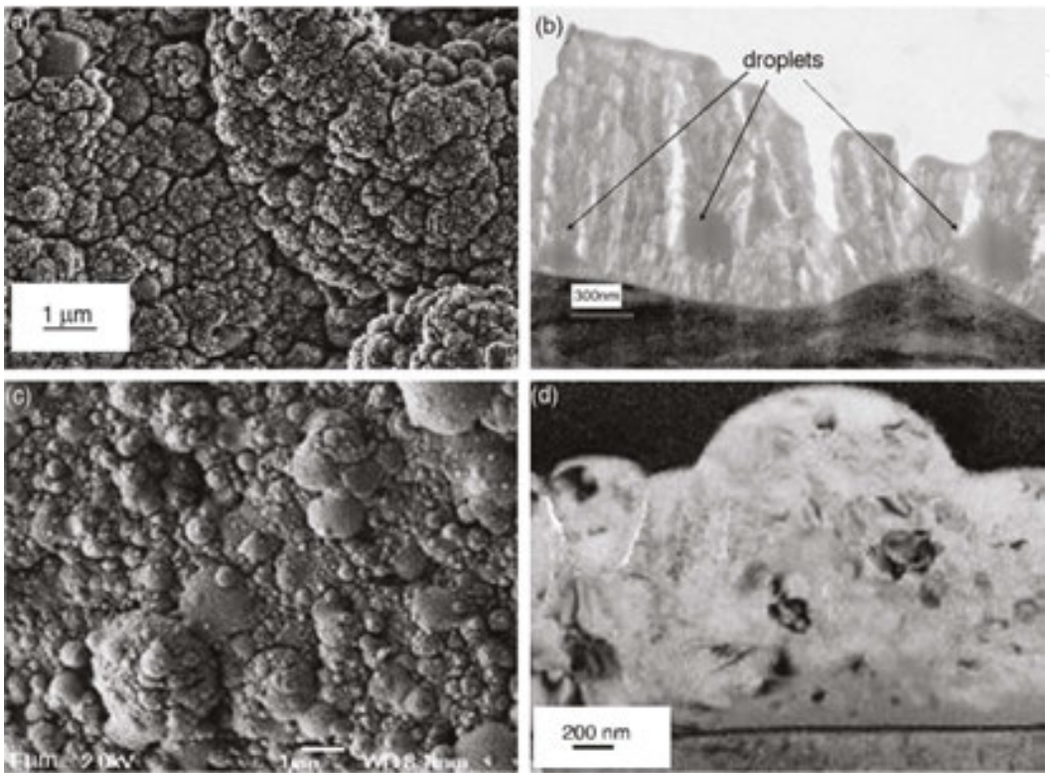

Figure 12. (a) Scanning and (b) Transmission Electron Microscopy (TEM) images of OCP coatings; (c) SEM and (d) TEM images of Mn-CHA coatings (reproduced with permission from Ref. [25]).

In case of OCP coatings, XPS measurements showed the total dissolution and disappearance of coating after 7 days of immersion in simulated body fluid (SBF) (**Figure 13a**). SEM investigations confirmed this advanced dissolution. As for Mn-CHA, the SEM and XPS investigations demonstrated that the coatings preserve the basic composition even the intensity of Ca and P peaks decreased (**Figure 13b**). After 7 days of immersion in SBF, the surface becomes slightly smoother.

Some interesting results were obtained by laser ablation of Mg:OCP and Sr:OCP compounds using the MAPLE technique [31]. The X-ray Diffraction (XRD) patterns revealed that all MAPLE coatings are constituted of OCP. This remark is sustained by the presence of the strong low angle reflection 2θ of 4.7° and the series of reflections in the range of 30–34°. Comparing these results with the previous one related to OCP deposition by PLD [70], one can remark that the gentle deposition conditions of MAPLE offer a higher degree of OCP crystallinity with respect to PLD [31]. The homogeneous distribution of magnesium and strontium on the thin film surface was evidenced by EDS analysis (**Figure 14**).

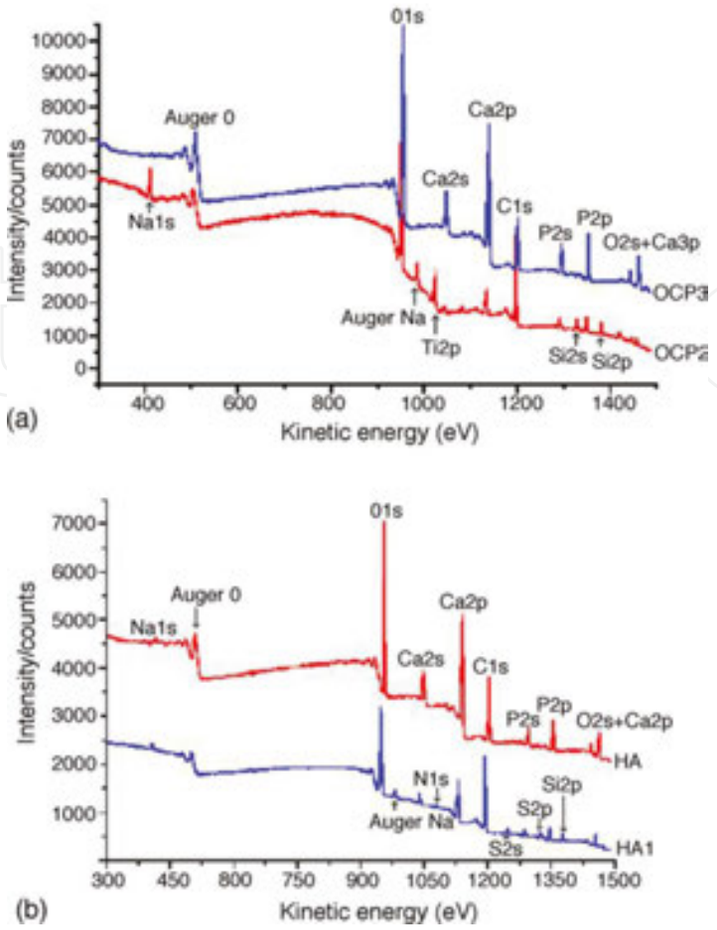

Figure 13. XPS spectra of (a) OCP and (b) Mn-CHA before and after degradation tests (reproduced with permission from Ref. [25]).

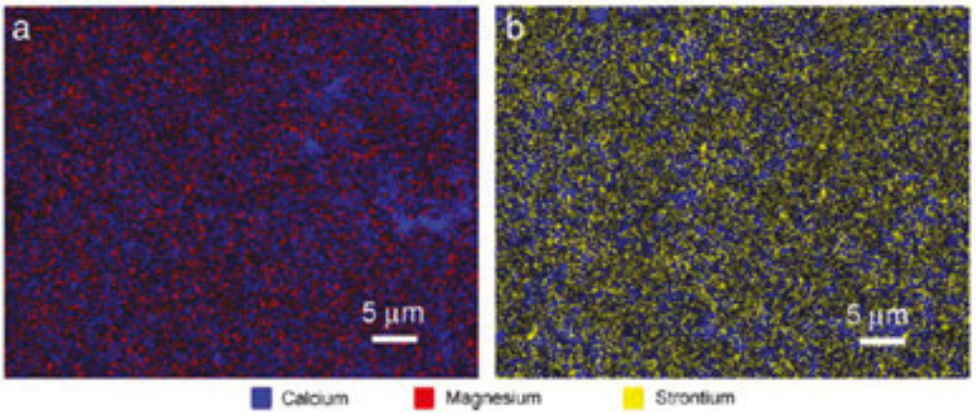

Figure 14. EDS maps recorded for (a) Mg:OCP and (b) Sr:OCP coatings (reproduced with permission from Ref. [31]).

To evaluate the proliferation and morphology of MG63 cells, one can also perform phalloidin staining on OCP, Mg:OCP, and Sr:OCP coatings deposited by MAPLE. The surface topography and chemical composition can influence cell behavior. Looking the images of MAPLE coating

stain with phalloidin for 14 days, no visible differences were observed (**Figure 15**, left). Similar results are visible from SEM images of the same structures (**Figure 15**, right). Deeper studies showed that Mg:OCP and Sr:OCP coatings improved the proliferation and differentiation of MG63 cells.

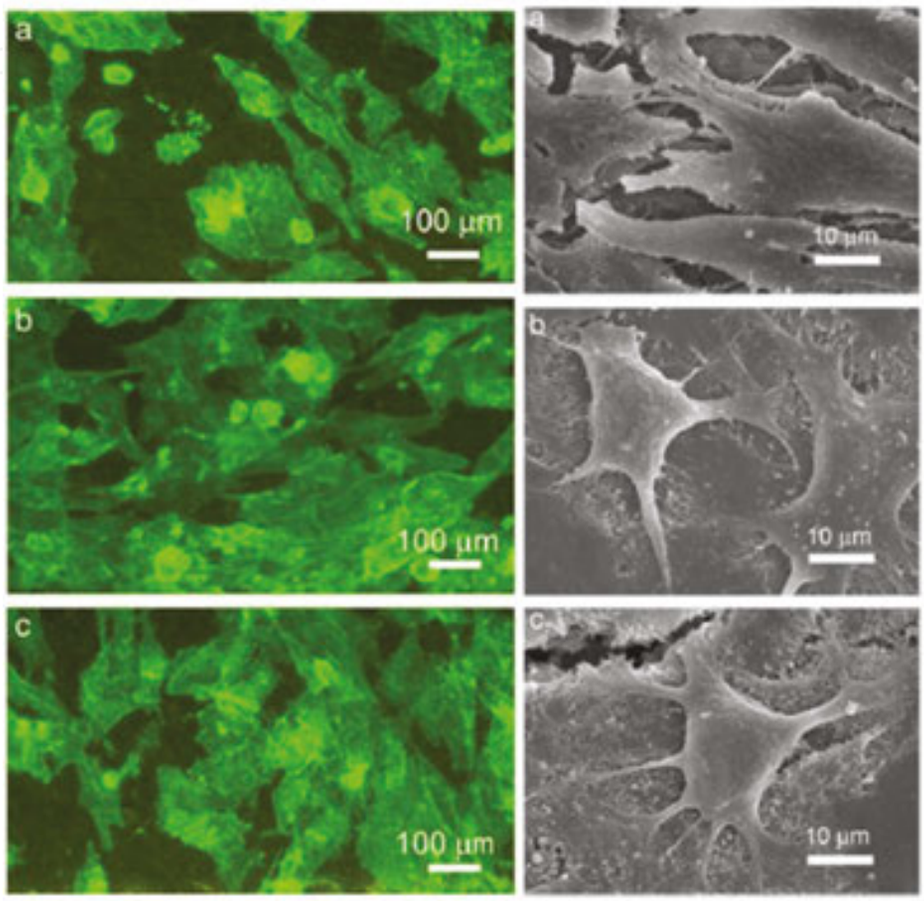

Figure 15. Phalloidin staining (left) and SEM images (right) of MG63 cells after 14 days of culture grown on (a) OCP, (b) Mg:OCP, and (c) Sr:OCP (reproduced with permission from Ref. [31]).

The biological analysis conducted on Mg:OCP and Sr:OCP coatings revealed no visible cytotoxic or inflammatory effects on osteoblast-like cells in all experimental times (**Table 1**).

Test MG63	3 days	7 days	14 days
WST1 (450/625 nm)	1.564 ± 0.125	3.132 ± 0.081	3.303 ± 0.047
Alkaline Phosphatase (ALP) (µg/mg proteins)	1.20 ± 0.17	2.33 ± 0.19	1.65 ± 0.02
Collagen Type I Production (CICP) (ng/mg proteins)	9.9 ± 1.2	11.3 ± 0.3	11.1 ± 0.3
Osteocalcin (OC) (ng/mg proteins)	2.00 ± 0.09	2.09 ± 0.18	2.63 ± 0.22
Interleukin-6 (IL-6) (pg/mg proteins)	37 ± 9	109 ± 5	97 ± 3
Transforming Growth Factor -β1 (TGF-β1) (pg/mg proteins)	454 ± 34	537 ± 67	370 ± 26

Table 1. Control values of proliferation and differentiation for MG63 osteoblast cells at 3, 7, and 14 days culture.

Mroz et al. also evaluated the performances of HA and OCP coatings deposited by PLD. The biological assays revealed that both layers are biocompatible with respect to human osteoblast cells, offering favorable conditions for their proliferation [71].

5. Conclusions and perspectives

This chapter highlighted the importance of laser ablation phenomenon that underlies the processing of biomaterials.

Laser ablation of biomaterials is an important and complex field. This domain was explored since long time, but there are a lot of natural or inorganic biomaterials, which wait to be developed and understood.

PLD and MAPLE techniques are of interest for thin film deposition, allowing varying some parameters to achieve the optimum conditions for the selected biomaterial. They permit the stoichiometric transfer. In case of PLD, the depositions are obtained from a solid target, as compared to MAPLE in which the target is a cryogenic mixture.

Sensitive biomaterials can be processed only by MAPLE, to avoid the chemical decomposition. The deposition of calcium phosphates (CaP) as thin films can be done by both techniques. Physical-chemical and biological analyses in the field of CaP recommend these coatings as potential biomaterial for the development of medical implants.

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