HAL
open science

Plasma-Sprayed Apatite Coatings: Review of Physical-Chemical Characteristics and Their Biological Consequences

Imane Demnati, David Grossin, Christèle Combes, Christian Rey

To cite this version:

Imane Demnati, David Grossin, Christèle Combes, Christian Rey. Plasma-Sprayed Apatite Coatings: Review of Physical-Chemical Characteristics and Their Biological Consequences. Journal of Medical and Biological Engineering, 2014, vol. 34 (n°1), pp. 1-7. 10.5405/jmbe.1459 . hal-01105362

HAL Id: hal-01105362

https://hal.science/hal-01105362

Submitted on 20 Jan 2015

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L'archive ouverte pluridisciplinaire **HAL**, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d'enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.

Open Archive TOULOUSE Archive Ouverte (OATAO)
OATAO is an open access repository that collects the work of Toulouse researchers and makes it freely available over the web where possible.

This is an author-deposited version published in : http://oatao.univ-toulouse.fr/
Eprints ID : 12046

To link to this article : DOI: 10.5405/jmbe.1459
http://dx.doi.org/10.5405/jmbe.1459

To cite this version : Demnati, Imane and Grossin, David and Combes, Christèle and Rey, Christian *Plasma-Sprayed Apatite Coatings: Review of Physical-Chemical Characteristics and Their Biological Consequences.* (2014) Journal of Medical and Biological Engineering, vol. 34 (n°1). pp. 1-7. ISSN 1609-0985

Any correspondance concerning this service should be sent to the repository administrator: staff-oatao@listes-diff.inp-toulouse.fr

Plasma-Sprayed Apatite Coatings: Review of Physical-Chemical Characteristics and Their Biological Consequences

Imane Demnati David Grossin Christèle Combes Christian Rey*

University of Toulouse, Cirimat,Umr Ups-Cnrs-Inpt, Ensiacet, 31030 Toulouse, France

Abstract

Hydroxyapatite (HA) is widely used as a coating on orthopedic medical implants to improve anchoring to bone tissue and facilitate bone repair. The use of plasma-sprayed HA coatings on metal implants has raised as much controversy as interest over the last decade. Although faster and stronger fixation and enhanced bone growth have been reported, the long-term performance of HA-coated implants has been questionned This ambivalent behaviour is related to the thermal decomposition of HA during the plasma spraying of HA powder, which has several observable consequences for the physical-chemical and biological properties of the coatings. This review presents the current knowledge on the characteristics of plasma-sprayed HA coatings and their biological consequences. Various schemes of chemical reactions and physical-chemical processes that can occur during HA-particle plasma spraying, HA-coating formation, or after implantation of HA coatings are presented. Standard and emerging methods developed for the characterization of HA coatings are reviewed, with a focus on Raman spectroscopy, which allows high-quality imaging. Finally, several promising trends that have been explored in recent years to improve the mechanical resistance of coatings and their adhesion on the substrate, to improve the coating microstructure and crystallinity, or to confer a therapeutic activity (e.g., antibacterial activity) on plasma-sprayed HA coatings are reviewed.

Keywords: Hydroxyapatite, Plasma-sprayed coatings, Thermal decomposition, Characterization

1. Introduction

Plasma spraying hydroxyapatite (HA) remains the most commonly used process for giving biological properties to orthopedic and dental metal implants, especially osteoconductive capabilities and efficient primary biointegration into bone tissue. Plasma spraying techniques have the ability to construct thick, adherent, and biologically active HA coatings [1-5]. However, it is well established that HA decomposes during plasma spraying, leading to the formation of various phases [1-13]. This decomposition has several drawbacks for the behavior of coatings after implantation, such as degradation, release of particulates, and delamination due to a loss of cohesion and poor metal-coating bonding. Plasma-sprayed coatings are usually composed of large crystalline HA particles embedded into a relatively soluble amorphous phase.

Several physical-chemical properties of plasma-sprayed HA coatings have been standardized with regard to their chemical composition (Ca/P ratio), phase composition, and crystallinity [14]. Most of these parameters are based on X-ray diffraction (XRD) analyzes. However, several additional parameters that are more difficult to determine, such as crystalline phase distribution within the coating, the amount of amorphous phase, and the oxyapatite (OA) fraction, are not considered in the standards although they could be related to the coating behavior *in vitro* and *in vivo*.

This review presents the current knowledge on the characteristics of plasma-sprayed HA coatings and their biological consequences. The reactions involved in the thermal decomposition of HA are summarized, and the consequences of these reactions for the coating characteristics are discussed. The biological consequences and recent developments are also reviewed.

2. Thermal decomposition of apatites

The thermal decomposition of HA occurs in two steps: an irreversible decomposition reaction in the range 200-800 °C and a reversible decomposition above 1000 °C.

* Corresponding author: Christian Rey
Tel: +33-5-34323484; Fax: +33-5-34323498
E-mail: christian.rey@ensiacet.fr

2.1 Irreversible decomposition

Generally, HAs used in plasma spraying are prepared in aqueous media either by using a double decomposition reaction between soluble calcium and phosphate salts, for example ammonium phosphate and calcium nitrate, or by neutralization of a calcium hydroxide suspension with a phosphoric acid solution [13]. After their synthesis, HAs may contain several impurities such as counter ions as well as HPO_4^{2-} and carbonate ions related to the weak dissociation constant of HPO_4^{2-} and contamination of the solutions with atmospheric CO_2. According to several studies, the chemical composition of such apatites can be represented as:

$Ca_{10-x-y+z}\,(PO_4)_{6-x-y}\,(HPO_4)_x(CO_3)_y\,(OH)_{2-x-y+2z}$
with $0 \leq x+y \leq 2$ and $0 \leq 2z \leq x+y$

The heating of unstable species like HPO_4^{2-} and CO_3^{2-} can lead to several irreversible reactions:

$2\,HPO_4^{2-} \longrightarrow P_2O_7^{4-} + H_2O$	200-600 °C range
$CO_3^{2-} + 2\,HPO_4^{2-} \longrightarrow PO_4^{3-} + H_2O + CO_2$	200-600 °C range
$CO_3^{2-} + P_2O_7^{4-} \longrightarrow PO_4^{3-} + CO_2$	300-700 °C range
$P_2O_7^{4-} + 2\,OH^- \longrightarrow PO_4^{3-} + H_2O$	600-800 °C range

At temperatures between 850 and 1000 °C, the mineral phases coexisting with HA depend only on the Ca/P ratio of the apatite (Fig. 1).

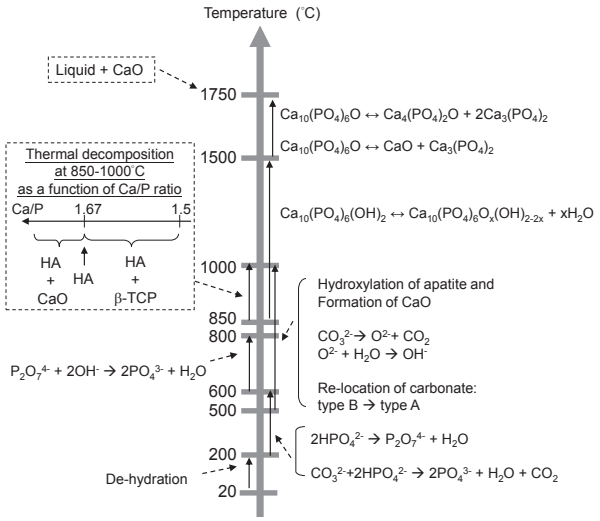

Figure 1. Thermal decomposition of apatite. Reversible and irreversible reactions as functions of temperature.

2.2 Reversible high-temperature decomposition

Several phase diagrams have been provided for the CaO-P_2O_5 system, with water vapor not considered in most cases [15,16]. However, it is well established that the decomposition of stoichiometric HA begins with dehydroxylation and the formation of OA related to the loss of water [17]:

$Ca_{10}\,(PO_4)_6\,(OH)_2 \leftrightarrow Ca_{10}\,(PO_4)_6\,O + H_2O$

This reaction is reversible and depends on the water vapor pressure. It seems to begin at about 850 °C in vacuum. Pure OA is very difficult to obtain and generally the prepared samples contain residual OH^- ions. Between 1050 °C and 1550 °C, oxy-hydroxyapatite (OHA) begins to decompose into tricalcium phosphate (TCP) and tetracalcium phosphate (TTCP):

$Ca_{10}\,(PO_4)_6\,O \leftrightarrow Ca_3(PO_4)_2 + Ca_4(PO_4)_2O$

Various values of this decomposition temperature, which increases with the water partial pressure, have been reported [16,18]. At higher temperature (1570 °C), a liquid phase, which is believed to correspond to a eutectic mixture is obtained. However some reports mention the existence of CaO resulting from the decomposition of HA within the liquid phase [5].

3. Foreign phases in plasma-sprayed coatings

The plasma spraying processes used by several medical manufacturers consist of the acceleration and heating at very high temperatures of HA crystalline particles introduced in a plasma flame produced by an electrical discharge between electrodes. The partly (or totally) melted HA particles impact the substrate, where they rapidly cool, resulting in a coating microstructure with a lamellar structure. These thermal events lead to a partial decomposition of the apatite structure and several foreign phases are generally observed in the coatings after cooling (Table 1). The possible formations of these phases are shown in Table 1. Multiple thermal events have to be considered: the rapid heating of in-flight particles, the quenching of the HA particles upon impact with the target, and the partial re-heating and cooling in multiple torch passes. The first layer is different from subsequent ones since it is deposited on a metal (most generally titanium or one of its alloys), which has good thermal conductivity. The other layers impact thermally insulating splats that have already been deposited, and thus their quenching is much less efficient and is associated with local re-heating and slow cooling.

Table 1. Phases found in plasma-sprayed HA coating and solubility products of various calcium phosphates.

Phase	Formula	Formation	-log K_{sp} at 25°C
Calcium oxide	CaO	Decomposition of HA and TTCP (volatilization of P)	-
Calcium hydroxide and calcium carbonate	$Ca(OH)_2$	Reaction of CaO with humidity and CO_2 (on contact with air or body fluids)	-
Amorphous calcium phosphate (ACP)	Variable	Melting of HA and quenching	-
α-Tricalcium phosphate (α-TCP)	α-$Ca_3(PO_4)_2$ (monoclinic)	Thermal decomposition of HA	25.5
β-Tricalcium phosphate (β-TCP)	β-$Ca_3(PO_4)_2$ (trigonal)	Phase transition on cooling from α-TCP	28.9
Tetracalcium phosphate (TTCP)	$Ca_4(PO_4)_2O$	Thermal decomposition of HA	38
Oxyhydroxyapatite (OHap)	$Ca_{10}(PO_4)_6O_x(OH)_{2-2x}$	De-hydration of HA	-
Hydroxyapatite (HA)	$Ca_{10}(PO_4)_6(OH)_2$ (stoichiometric)	-	116.8

The formation of some of the phases found in coatings can be easily explained by the thermal stability of apatite. The amorphous calcium phosphate (ACP) phase, which is generally the most abundant foreign phase found in coatings, is considered to originate from the fast quenching of the liquid phase upon reaching the target, as schematized in Fig. 2. This amorphous phase is more abundant in coatings made with one pass of the plasma torch (Fig. 2). The TTCP and TCP phases are considered to correspond to phases formed during the heating of the HA particles, inside the grain, in areas where the temperature is in the narrow gap between the maximum stability temperatures of the apatite phase and the liquid phase. Another possibility is that these phases are formed during the cooling period if the rate of cooling of the liquid fraction is slow enough to allow the nucleation of these phases. In coatings formed by multiple torch passes, re-heating and a slower cooling rate could favor the re-formation of apatite from the decomposition products, as it is the most stable phase, having a large domain of thermal stability.

OA is considered by most researchers to exist in solid solution with HA [16-18]. It likely exists in deeper zones of the in-flight particles where the temperature is lower (in the range 1050-1350 °C). It seems reasonable to consider that the OA content varies within the flying particles (Fig. 2). Rehydration of the OA fraction during cooling in air, which can be considered as re-hydroxylation, has been shown to be active at temperatures as low as 400 °C [17].

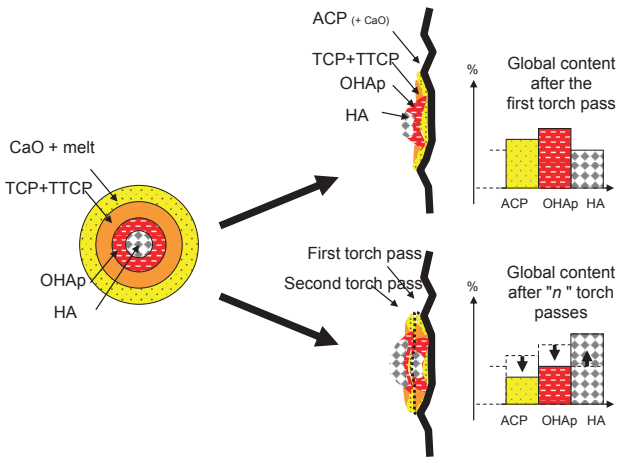

Figure 2. Sequence of events during deposition of a semi-molten HA particle impinging on a roughened substrate.

The formation of CaO, which is often observed in coatings rich in amorphous phase obtained at high power, remains, however, unexplained by phase diagrams. Some researchers consider that the liquid phase may in fact contain CaO. This assumption is supported by experimental data on the fast quenching of totally melted HA particles, which shows only an amorphous phase and CaO [6]. In this case, however, one may wonder why CaO is not observed systematically in all coatings containing an amorphous phase but is only seen in some of them. Several hypotheses can be proposed: CaO forms only nanocrystals which are difficult to detect; CaO reacts with water and/or CO_2 from the atmosphere once the coating has been formed, giving poorly crystalline or amorphous phases; the solubility of CaO in the melted phosphate varies with temperature. CaO is the most refractory phase in coatings with a melting point of 2613 °C.

Most coatings obtained in argon plasma do not show any variation in their chemical composition and after plasma spraying their Ca/P ratio is the same as that of the initial apatite powder. However, very high powers and/or reductive plasmas, containing hydrogen, may induce a loss of phosphorus and are associated with an increase in the Ca/P ratio.

The global chemical phase composition and crystallinity of the coatings are determined, in most countries, according to ISO standards, based on the analysis of XRD patterns. However, certain characteristics of the coatings which have to be considered for their biological behavior have not been standardized, such as OA content, amorphous phase content (not only crystallinity, which is a different parameter), and phase distribution.

The most important parameter is certainly the amorphous phase content. Old standards (AFNOR S 94-068) distinguished between crystallinity and amorphous phase content. Although a large amount of amorphous phase affects the crystallinity of the coating, other parameters are also involved in the crystallinity, such as the re-formation of poorly crystalline apatite during cooling or re-heating in multiple torch passes and heterogeneous solid solution formation between OHAs and HAs, which increase the level of strain and contribute to the XRD background. The determination of the real amount of amorphous phase by XRD is rather tedious and imprecise; however, Raman spectroscopy allows a clear identification of the amorphous component, which is distinct from the crystalline apatite phase. The phase distribution is also crucial for the evolution of the coating after implantation.

4. Phase distribution in coatings

Plasma-sprayed coatings have been shown to be very inhomogeneous. Several methods have been proposed for evaluating the distribution of phases in coatings, including optical microscopy [7], scanning electron microscopy (SEM), backscattered electron imaging, and scanning cathode-luminescence microscopy [8]. A few studies have used microdiffraction [9]. The amorphous phase appears as the major foreign phase in most commercial HA coatings. It has been identified in close proximity to the metal surface, whose high thermal conductivity facilitates quenching of the molten phase. TCP and TTCP phase formation resulting from the decomposition of OA between 1050 °C and 1550 °C should occupy the same areas; however, these phases are generally present in very small amounts in commercial HA coatings and their distribution is not well known [10]; in some coatings they do not appear at all. Several explanations can be given, such as the existence of supercooled liquid in the narrow window of stability of TCP and TTCP, which would favor the formation of OHA, whose nucleation would be facilitated by its abundance in the coating. The reformation of apatite from these phases, which should be in close proximity in multiple coatings, also seems likely [11]. The OA fraction is also an important characteristic

of the coatings, but as its XRD pattern is very close to that of apatite, it is very difficult to identify.

Information about the morphological changes at the surface and interface of coatings can be obtained from SEM observations, but a detailed structural chemical analysis of the coatings in these regions is still lacking. Raman scattering allows local investigations of the coating's heterogeneity. It is a unique tool providing information on the structure and short-range order of amorphous and crystalline phases [19-21]. Moreover, Raman imaging based on different strategies from a given dataset may provide a powerful means of studying structural changes and visualizing the distribution of phases within heterogeneous plasma-sprayed HA coatings. All mineral phases present in a coating can potentially be distinguished by Raman spectroscopy in the ν_1PO_4 domain. An example of micro-Raman spectroscopic imaging collected across an HA coating cross-section is shown in Fig. 3 The ACP is characterized by a broad Raman line at 950 cm^{-1} and all crystalline phases show lines distinct from that of apatite at 961 cm^{-1} (Fig. 4). OA, difficult to distinguish by other methods, shows very specific Raman scattering lines at about 951 cm^{-1} and 966 cm^{-1}, which have been assigned respectively to the vibration of phosphate ions close to a vacancy and close to an O^{2-} ion [17]. CaO is the only phase which cannot be detected in this domain.

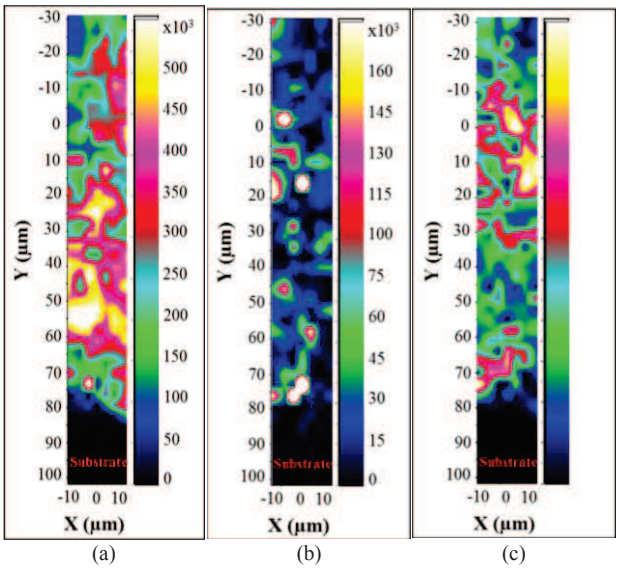

Figure 3. Representation of micro-Raman spectroscopic images of a HA coating cross section showing the phases distributions: ACP (a); HA (b) and OA (c).

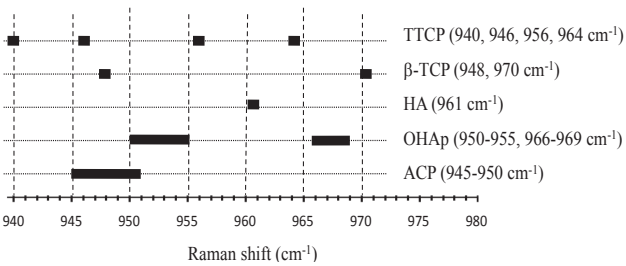

Figure 4. Raman spectral characteristics of phosphate ions in various calcium phosphate salts between 940 and 970 cm^{-1}.

The most accurate mapping of the phases in HA coatings can be obtained by Raman imaging. Imaging of the phase distribution using Raman scattering involves curve fitting in the ν_1PO_4 domain, extraction of the characteristic line(s) associated with specific phases, and mapping of their relative amounts [12]. It should be noted that without standardization, the data represent only the relative integrated intensities of lines associated with a specific phase and their relative variation, not their content in the coating. Figure. 3 shows the distributions of the amorphous component (Fig. 3(a)), HA (Fig. 3(b)), and OA (Fig. 3(c)) in a cross section of a HA coating obtained using a low-energy plasma spraying device [11]. The lamellar structure of the plasma-sprayed coating appears clearly on all these images; however, the limits of individual splats are difficult to distinguish. Heterogeneous structures much smaller than the particles are also observed, suggesting rupturing of grains and splitting on impact. The amorphous phase appears in close proximity to the metal surface but not directly on the surface, probably because the plastic deformation of the solidifying liquid phase on the impact results in a rejection of this phase to the periphery of the splat, whereas the solid core, containing mostly HA, reaches the surface. The domains rich in OA are relatively narrow and dispersed. Two types of OA-rich domain can be distinguished as those found within the amorphous phase, which may correspond in part to a recrystallization process, and those contiguous to HA domains, possibly formed by dehydration during particle flight. HA-rich domains can be found directly in contact with the metal. It is clear that HA-rich domains exist in zones with low amorphous content and inversely.

In addition to a heterogeneous distribution of phases in terms of thickness in the deposit cross section, there is also a heterogeneous distribution on the surface itself and very different Raman spectra are collected depending on the location (Fig. 5) [12]. Therefore, Raman micro-spectroscopy can be used as a routine non-destructive tool to obtain rapid analysis of the composition of apatite coatings and grains. This technique could be adapted for the chemical characterization of coatings applied directly on biomedical implants.

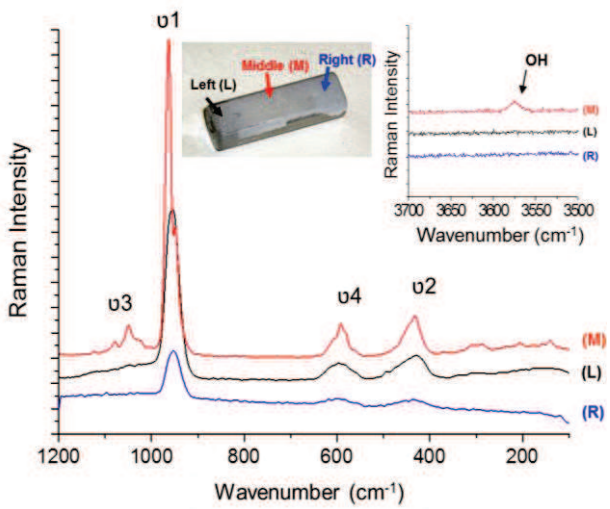

Figure 5. Raman spectra at various points on the surface of a HA coating showing its heterogeneity.

The heterogeneities of plasma-sprayed coatings affect their biological behavior.

5. Physical-chemical properties of HA coatings and their effect on biological behavior

Several physical-chemical properties of the coatings affect biological behavior, namely the solubility of mineral phases, hydration characteristics, and propensity to hydrolyze or transform spontaneously into more stable compounds.

5.1. Solubility of mineral phases and biodegradation

The biodegradation of calcium phosphates has often been related to their solubility, and this relationship has been confirmed by experimental data [22]. According to their solubility product, the phases found in plasma-sprayed HA coatings can be arranged in order of decreasing biodegradability (Table 1). Among these phases, one may however distinguish those which should spontaneously dissolve, because their solubility product is higher than the ion product of body fluids, and those for which the dissolution is mediated by cellular activity. The phases belonging to the former type, namely CaO, ACP, and α-TCP, cannot remain as long as they are in contact with body fluids. Due to the spraying process, however, some phases can be embedded in the accumulated splats, where they are not accessible to body fluids or cells, allowing them to remain unaltered until cracks or dissolution paths reach them.

5.2. Hydration and hydrolysis reactions

Several phases formed at high temperature are not stable in aqueous media and may transform more or less rapidly into other compounds. This is the case of CaO, which hydrolyzes into $Ca(OH)_2$. This reaction leads to a 40% increase in volume and may lead to a rupture of the coating. In addition, in the presence of phosphate ions from body fluids, a precipitate may form with the calcium ions released from the very soluble $Ca(OH)_2$.

The amorphous phase can also hydrolyze into apatite. This reaction first involves a hydration of the amorphous phase [22]. It has been suggested that this hydration process could also lead to fracture of the coating [23]. Once hydration has occurred, ACP transforms rather rapidly into apatite [24]. Several investigations have shown that the ACP content decreases and the crystallinity increases after contact with body fluids or their model solutions (Ringer solution or simulated body fluid (SBF) solution) due to the hydrolysis of ACP into hydroxylated apatite.

α-TCP and TTCP hydrolyzes have also been observed and, like ACP, they lead to the formation of hydroxylated apatite [25,26]. These reactions can be rapid at body temperature (a few tens of minutes when the phases are in a powder state) and they are involved in the setting of several hydraulic mineral cements for orthopedic applications.

Although β-TCP has been shown to hydrolyze into apatite in aqueous media, the reaction at body temperature seems extremely slow and it is generally not mentioned; however, when considering months of implantation, it might be involved.

The transformation of OA into HA by hydrolysis requires the diffusion of water molecules into the apatite structure. Surface hydrolysis may be responsible; however, the diffusion of water molecules inside the lattice at body temperature would be much too slow. Another possibility is a dissolution-reprecipitation process, as OA has been claimed to be more soluble than HA [18]. However, very often, no accurate phase analysis of coatings is conducted and thus this question remains open. These reactions and their effects on coatings are summarized in Fig. 6.

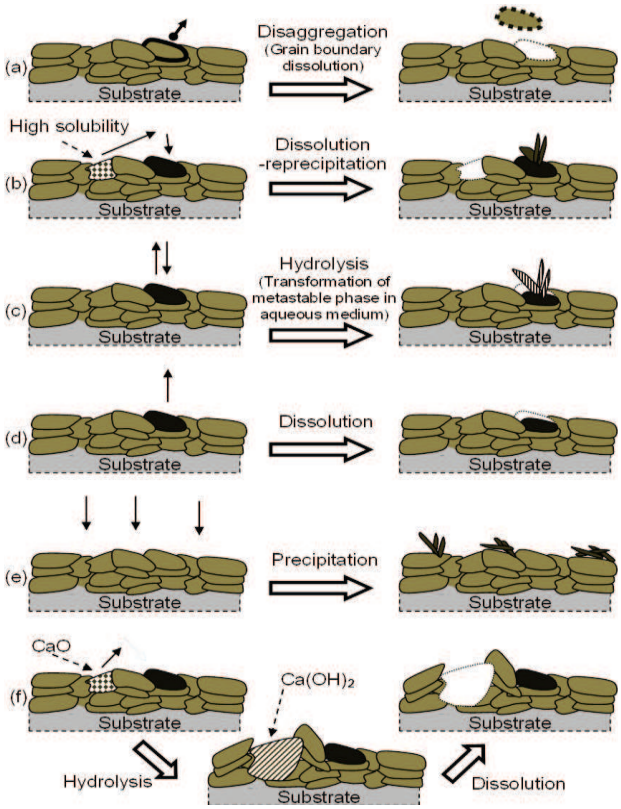

Figure 6. Scheme of reactions at the coating aqueous medium interface.

6. Biological role and biological consequences of HA coating composition

The main role of a coating is to improve bone integration and the stability of the interface between the implant and the tissue. This role is essential in the first months after implantation, as most failures of implants are related to poor stability in the biological tissue. However, as bone remodels continuously, it seems important for some biological activity to occur on the implant surface during its life span. Several studies have shown the positive effect of coatings on bone integration during the first months after implantation [28-30]. In the long term, however, the difference between coated and uncoated implants generally tends to diminish and coated implants have sometimes been reported to deteriorate [31].

The preferential dissolution of foreign mineral phases weakens coating integrity and cohesion. The events associated with mechanical stress may lead to delamination and scaling and

to the release of particles [31-33]. Potentially, such a release of small particulates may lead to inflammatory reactions, which do not seem to have been observed in most cases. Nevertheless, the actual tendency is to prefer coatings with a high crystallinity to avoid weakening by dissolution of the heterogeneously distributed amorphous phase. Another precaution is to ensure that either very large particles or small degradable particles, which would have little inflammatory power, are generated during degradation.

However, the presence of hydrolyzable phases has also been seen as an advantage as it allows the formation of a neo-formed layer of carbonated apatite analogous to the mineral phase of bone. The development of such layers is considered an essential step in the colonization of the implant surface by osteoblast cells and bone reconstruction. Another cellular effect of the hydrolysis reaction, especially in the case of CaO and TTCP, is the local alkalinization of the surrounding media, which could inhibit osteoclast activity.

Most studies have shown that a weakness of plasma-sprayed coatings is the attachment to the metal surface. Although chemical reactions with metal oxide layers cannot be excluded, their role in the adherence of the coating seems marginal and the main parameters to consider are the surface roughness and the thickness and composition of the coating, as discussed in previous studies [2,13]. This mechanical weakness plays a role in degradation, as underlayers rich in a soluble amorphous phase can be uncovered. A second problem with HA-coated dental implants is the possible promotion of bone infection. Recent developments have addressed some of these drawbacks.

7. Recent developments

Three types of development can be distinguished: that aimed at improving the mechanical properties and adhesion to the substrate, that claiming a therapeutic activity, and that aimed at improving microstructure and crystallinity.

7.1 Mechanical properties and adhesion

Several approaches have been made to improve the mechanical resistance of coatings and their adhesion to the substrate. The most successful incorporate ZrO_2 [34] or other ceramic oxides and use graded co-spraying of HA and titanium [35] or titanium oxide. In these coatings, the proportion of apatite co-deposited increases progressively from the metallic substrate to the surface. However, the implantation of coatings with ZrO_2 has led to poor biological results with lower adhesion of bone to the coating [36]. Other coatings have apparently not been tested *in vivo*. There are still improvements to be made in this domain.

7.2 Coatings with therapeutic activity

The second field of development concerns the biological activity of coatings. Two approaches have been proposed: the association of coatings with growth factors or active molecules and the use of active mineral ions. The first one is rather limited as the surface area of plasma-sprayed HA coatings is low. The second one seems more promising and several biologically active elements or their combinations have been proposed: Ag^+, Cu^{2+}, and Zn^{2+} ions for their antibacterial properties, Sr^{2+} for its anti-osteoporotic activity, and silicon oxide to improve bone formation.

7.3 Improvement of microstructure and crystallinity

The third type of development is more specifically related to the physical-chemical properties of the coatings and their biological consequences. Increases in coating crystallinity may result from the choice of plasma-spraying conditions and several reports have been made. The use of mini- or even micro-torches working at low energy has been promoted [12]. These systems need smaller particulates but they produce coatings with high crystallinity and have several other advantages such as better homogeneity of composition within the particle and within the coating and thus a lower amount of metastable phases along with better control of coating microporosity. The use of phases other than HAs, especially fluorapatites, fluorhydroxyapatites, and more recently chlorapatites [12], has also been proposed. Unlike HA, fluor- and chlor-apatites melt without decomposition. Although some of these coatings have been tested with some success, they have not yet been commercially applied. The improvement of crystallinity can also be obtained by post-treatments: thermal and hydrothermal treatments and treatments in solutions have been applied [37]. There is however a risk of deterioration of the mechanical integrity of the coating.

8. Conclusion

Plasma-sprayed coatings are largely used in many countries. Since the first experiments, considerable progress has been made, especially in the understanding of the characteristics of coatings and their biological behavior. Several means of improvement have been explored; however, the problems of coating adhesion and mechanical strength remain. Other coating processes have been investigated, especially low-temperature coating, which offers more flexibility concerning, in particular, the types of apatite used (e.g., non-stoichiometric, biomimetic, and carbonated apatites), but without dethroning, at present, plasma spraying.

Acknowledgments

Part of the work reported in this review was carried out within the framework of a European Community funded project, Brite-Euram II, and a Midi-Pyrénées region MNT ERA-Net funded project, NANOMED2.

References

[1] S. C. G. Leeuwenburgh, J. G. C. Wolke, J. A. Jansen and K. de Groot, "Calcium phosphate coatings," in: T. Kokubo (Ed.), *Bioceramics and their clinical applications*, Cambridge: Woodhead publishing limited, 1: 464-484, 2008.

[2] L. Sun, C. C. Berndt, K. A. Gross and A. Kucuk, "Material fundamentals and clinical performance of plasma sprayed hydroxyapatite coatings: a review," *J. Biomed. Mater. Res.*, 58: 570-592, 2001.

[3] K. De Groot, R. Geesink, C. Klein and P. Serekian, "Plasma sprayed coatings of hydroxylapatite," *J. Biomed. Mater. Res.*, 21: 1375-1381, 1987.
[4] R. LeGeros, J. LeGeros, Y. Kim, R. Kijkowska, R. Zheng, C. Bautista, J. Wong, G. Fischman, A. Clare and L. Hench, "Calcium phosphates in plasma-sprayed HA coatings," *Ceram. Trans.*, 48: 173-173, 1995.
[5] R. B. Heimann, "Thermal spraying of biomaterials," *Surf. Coat. Technol.*, 201: 2012-2019, 2006.
[6] X. Ranz, "Développement et caractérisation de dépôts d'apatite obtenus par projection plasma sur prothèses orthopédiques," *PhD. Thesis at Institut national polytechnique of toulouse*, 1996.
[7] K. A. Gross, C. Berndt and H. Herman, "Amorphous phase formation in plasma-sprayed hydroxyapatite coatings," *J. Biomed. Mater. Res.*, 39: 407-414, 1998.
[8] K. A. Gross and M. Phillips, "Identification and mapping of the amorphous phase in plasma-sprayed hydroxyapatite coatings using scanning cathodoluminescence microscopy," *J. Mater. Sci. Mater. Med.*, 9: 797-802, 1998.
[9] B. Y. Chou, E. Chang, S. Y. Yao and J. M. Chen, "Phase transformation during plasma spraying of hydroxyapatite-10-wt%-zirconia composite coating," *J. Am. Ceram. Soc.*, 85: 661-669, 2002.
[10] M. Carayon and J. Lacout, "Study of the Ca/P atomic ratio of the amorphous phase in plasma-sprayed hydroxyapatite coatings," *J. Solid State Chem.*, 172: 339-350, 2003.
[11] I. Demnati, M. Parco, D. Grossin, I. Fagoaga, C. Drouet, G. Barykin, C. Combes, I. Braceras, S. Goncalves and C. Rey, "Hydroxyapatite coating on titanium by a low energy plasma spraying mini-gun," *Surf. Coat. Technol.*, 206: 2346-2353, 2012.
[12] I. Demnati, "Développement et caractérisation de revêtements bioactifs d'apatite obtenus par projection plasma à basse énergie: application aux implants biomédicaux," *PhD. Thesis at Institut national polytechnique of toulouse*, 2011.
[13] P. Cheang and K. Khor, "Addressing processing problems associated with plasma spraying of hydroxyapatite coatings," *Biomaterials*, 17: 537-544, 1996.
[14] "Implants for surgery-Hydroxyapatite-Chemical analysis and characterization of christallinity and phase purity," ISO 13779-3, 2008.
[15] J. C. Elliott, *Structure and chemistry of the apatites and other calcium orthophosphates*, Amsterdam: Elsevier, 1994.
[16] P. V. Riboud, "Composition et stabilité des apatites dans le système CaO-P_2O_5-oxyde de fer-H_2O," *Ann. Chim.*, 8: 381-390, 1973.
[17] J. C. Trombe, "Contribution à l'étude de la décomposition et de la réactivité de certaines apatites hydroxylées, carbonatées ou fluorées alcalino-terreuses," *PhD. Thesis at University Paul Sabatier of Toulouse*, 1972.
[18] K.A. Gross, C.C. Berndt, P. Stephens and R. Dinnebier, "Oxyapatite in hydroxyapatite coatings," *J. Mater. Sci.*, 33: 3985-3991, 1998.
[19] L. Yan, Y. Leng and L. T. Weng, "Characterization of chemical inhomogeneity in plasma-sprayed hydroxyapatite coatings," *Biomaterials*, 24: 2585-2592, 2003.
[20] H. Podlesak, L. Pawlowski, R. d'Haese, J. Laureyns, T. Lampke and S. Bellayer, "Advanced microstructural study of suspension plasma sprayed hydroxyapatite coatings," *J. Therm. Spray Technol.*, 19: 657-664, 2010.
[21] M. Weinlaender, J. Beumer, E. B. Kenney and P. K. Moy, "Raman microprobe investigation of the calcium phosphate phases of three commercially available plasma-flame-sprayed hydroxyapatite-coated dental implants," *J. Mater. Sci.*, 3: 397-401, 1992.
[22] R. Z. LeGeros, "Biodegradation and bioresorption of calcium phosphate ceramics, " *Clin. Mater.*, 14: 65-88, 1993.
[23] S. Somrani, C. Rey and M. Jemal, "Thermal evolution of amorphous tricalcium phosphate," *J. Mater. Chem.*, 13: 888-892, 2003.
[24] J. Weng, J. G. C. Wolke, X. Zang and K. de Groot, "Environmentally induced cracks and detachment of plasma-sprayed hydroxyapatite coatings under ultrasonic treatment," in L. Hench (Ed.), *Bioceramics 8*, New York: Elsevier, 1: 339-344, 1995.
[25] S. Somrani, M. Banu, M. Jemal and C. Rey, "Physico-chemical and thermochemical studies of the hydrolytic conversion of amorphous tricalcium phosphate into apatite," *J. Solid State Chem.*, 178: 1337-1348, 2005.
[26] H. Monma, S. Ueno and T. Kanazawa, "Properties of hydroxyapatite prepared by the hydrolysis of tricalcium phosphate," *J. Chem. Tech. Biotechnol.*, 31: 15-24, 1981.
[27] L. Chow, M. Markovic, S. Fukhtbeyn and S. Takagi, "Hydrolysis of tetracalcium phosphate under a near-constant-composition condition, effects of pH and particle size," *Biomaterials*, 26: 393-401, 2005.
[28] M. J. Coathup, G. W. Blunn, N. Flynn, C. Williams and N. P. Thomas, "A comparison of bone remodelling around hydroxyapatite-coated, porous-coated and grit-blasted hip replacements retrieved at post-mortem," *J. Bone Joint Surg.-Br. Vol.*, 83: 118-123, 2001.
[29] N. Aebli, J. Krebs, H. Stich, P. Schawalder, M. Walton, D. Schwenke, H. Gruner, B. Gasser and J. C. Theis, "In vivo comparison of osseointegration of vacuum plasma sprayed titanium- and hydroxyapatite-coated implants," *J. Biomed. Mater. Res. Part A*, 66: 356-363, 2003.
[30] A. Moroni, V. Caja, E. Egger, L. Trinchese and E. Chao, "Histomorphometry of hydroxyapatite coated and uncoated porous titanium bone implants," *Biomaterials*, 15: 926-930, 1994.
[31] A. E. Porter, P. Taak, L. W. Hobbs, M. J. Coathup, G. W. Blunn and M. Spector, "Bone bonding to hydroxyapatite and titanium surfaces on femoral stems retrieved from human subjects at autopsy," *Biomaterials*, 25: 5199-5208, 2004.
[32] T. W. Bauer, R. C. T. Geesink, R. Zimmerman and J. T. McMahon, "Hydroxyapatite-coated femoral stems-Histological analysis of components retrieved at autopsy," *J. Bone Joint Surg.-Br. Vol.*, 73: 1439-1452, 1991.
[33] H. W. Denissen, C. P. Klein, L. L. Vish and A. van den Hooff, "Behavior of calcium phosphate coatings with different chemistries in bone," *Int. J. Prosthodont.*, 9: 142-148, 1996.
[34] E. Chang, W. Chang, B. Wang and C. Yang, "Plasma spraying of zirconia-reinforced hydroxyapatite composite coatings on titanium: Part I Phase, microstructure and bonding strength," *J. Mater. Sci.-Mater. Med.*, 8: 193-200, 1997.
[35] C. C. Chen, T. H. Huang, C. T. Kao and S. J. Ding, "Characterization of functionally graded hydroxyapatite/ titanium composite coatings plasma-sprayed on Ti alloys," *J. Biomed. Mater. Res. Part B*, 78: 146-152, 2006.
[36] T. M. Lee, C. Y. Yang, E. Chang and R. S. Tsai, "Comparison of plasma-sprayed hydroxyapatite coatings and zirconia-reinforced hydroxyapatite composite coatings: in vivo study," *J. Biomed. Mater. Res. Part A*, 71: 652-660, 2004.
[37] C. Y. Yang, T. M. Lee, C. W. Yang, L. R. Chen, M. C. Wu and T. S. Lui "In vitro and in vivo biological responses of plasma-sprayed hydroxyapatite coatings with posthydrothermal treatment," *J. Biomed. Mater. Res. Part A*, 83: 263-271, 2007.