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Up-to-Date Knowledge and Outlooks for the Use of Metallic Biomaterials: Review Paper

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Abstract

In all cases, when a material has to be used in medical applications, the knowledge of its physical, chemical and biological properties is of fundamental significance, since the direct contact between the biological system and the considered device could generate reactions whose long-term effects must be clearly quantified. The class of materials that exhibits characteristics that allow their use for the considered applications are commonly called *bio-materials*. Patients suffering from different diseases generate a great demand for real therapies, where the use of biomaterials are mandatory. Commonly, metallic biomaterials are used because their structural functions; the high strength and resistance to fracture they can offer, provide reliable performance primarily in the fields of orthopedics and dentistry. In metals, because of their particular structure, plastic deformation takes place easier, inducing good formability in manufacturing. The present paper is not encyclopaedic, but reports in the first part some current literature data and perspectives about the possibility of use different class of metallic materials for medical applications, while the second part recalls some results of the current research in this field carried out by the authors.

Keywords: biomaterials, metallic biomaterials, morphology, biocompatibility, implanted antenna

1. Introduction

1.1. Outline

Generally, materials are playing an important role in our society. Apart other materials, biomaterials became key elements for the human well-being. The beginning of biomaterials goes back to thousands of years (the use of metals in dental implants dates back to 200 A.D.), while after World War II, a real expansion of biomaterials can be considered [1].

Actually, biomaterials can be employed for both short-term or permanently, amplifying or partially or completely replacing some organs, tissues, or other elements of the body with the aim to save or to improve the people's well-being. The numbers of the aged people are quickly increasing that in turn requires a greater call to substitute failed body parts with artificial devices made of biomaterials. Additionally, the ongoing tendency to move in the direction of using short-term implants and/or degradable ones constitutes one of the key progresses in biomaterials research. Further important issues are correlated to the possibility to reduce as much as possible implant related infections by means of infection-resistant biomaterials, which usually involves the application of suitable coatings. The use of any biomaterial is influenced by different features, i.e., composition, mechanical strength, rate of degradation (in case of temporary application), growth rate of the human cells, osteoinduction, osteoconduction, etc. [2].

The present review is not encyclopedic and does not completely cover any features about biomaterials, regenerative medicine (RM), and tissue engineering (TE) aspects, but it shows some essential features belonging to such a wide-ranging subject. Some experimental results here presented deal with the outcome of some of the investigations performed by the authors on the development of metallic biomaterials and their possible use for medical implant development. It also includes the description of a multidisciplinary application where a data transmission system is referred to that has been possible to be developed because of the existence of large dimension implanted device. Nowadays, other similar applications are under focus: controlled drug delivery from swallowed capsules by means of external (to the body) control through a communication system that requires the use of an antenna, explicitly a metallic device, is just one of the many applications that could be nominated.

1.2. Metallic biomaterials

Metallic biomaterials show excellent structural functions, superior than ceramic and polymeric biomaterials and for a long time they have been usually employed to substitute unhealthy natural parts or to repair different organs of the human body. Extension of metal made implants was firstly determined by the request to repair bone and later on their application for orthopedic purpose has been increased together with the short-term applications of pins and screws, followed by the use of permanent implants for total joint substitution. Further evolution involves the use of metals and their alloys for dental applications and recently there is a higher tendency for their use in non-conventional reconstructive surgery of organs and hard tissue. There are only some areas where the use of metallic biomaterials is exploited such as joint prostheses, bone fixation plates, pins and screws, dental implants and materials, ocular, contact and intra-ocular lenses, vascular grafts, urinary and intra-vascular catheters, surgical meshes, and voice prostheses.

Within the different class of biomaterials, firstly, *stainless steel* has been used for medical purpose. This is principally due to their low cost, good corrosion resistance conferred by the presence of Cr, allowing the development of the passive and protective oxide layer. **Table 1** reports the composition of the most important/recommended stainless steels used for the manufacturing of orthopedic implants. ASTM F138 and F139 reveal higher biomedical interest because of a better fatigue strength, higher ductility, and better machinability [3], but the high Ni content determines possible toxicity and allergy [4].

Steels (ASTM)	C	Mn	P	S	Si	Cr	Ni	Mo	N	Cu
F138	0.03	2.0	0.025	0.01	0.75	17.0–19.0	13.0–15.0	2.25–3.0	0.10	0.50
F1314	0.03	4.0–6.0	0.025	0.01	0.75	20.5–23.5	11.5–13.5	2.0–3.0	0.20–0.40	0.50
F1586	0.08	2.0–4.25	0.025	0.01	0.75	19.5–22.0	9.0–11.0	2.0–3.0	0.25–0.50	0.25
F2229	0.08	21–24	0.03	0.01	0.75	19.0–23.0	0.10	0.50–1.50	0.90	0.25

Table 1. Composition (wt%) of some austenitic stainless steels [5].

However, according to some research, e.g., [6], stainless steels can suffer pitting, crevice, corrosion fatigue, stress corrosion cracking, and galvanic corrosion within the human body accelerated by their reduced wear resistance. Additionally, appearance of allergic reaction limits their use as orthopedic joint prosthesis [7, 8]. Release of metallic ions, as corrosion product, in the human body can reduce the lifetime of the implant leading to an additional surgical procedure [9–12]. If high amount of Ni ions is released in the tissues, development of the most common contact allergy [13, 14] and cancer [15, 16] can be favored.

The corrosion resistance of stainless steels can be modified by the (i) different oxygen concentration (generally, low oxygen content contributes to a higher corrosion rate, because the growth of the protective oxide layer is delayed on the surfaces of the metallic implant), (ii) pH value which can be altered from a neutral condition to lower values due to the inflammatory cell secretions, and (iii) presence of aqueous ions level in the surrounding body fluid. If possible, the concentration of the released metal ions has to be as reduced as possible, and it has to be innocuous for the human body during a long service period [17]. The high elastic modulus of stainless steel is considered as one of the highest difficulties leading to the damage of the bone by stress shielding which is directly connected to the failure of the implant.

During the time, *Cobalt-Chromium alloys* have been developed, which reveal some superior properties and some drawbacks relating to their production costs than stainless steels [18, 19].

The alloys belonging to this family show high corrosion resistance attributable to the natural growth of the passive oxide film on the top of the metallic device which is resistant also in the chloride-rich background [20–23].

Generally, two basic types of such alloys are commonly employed:

- Cobalt-Chromium-Molybdenum (CoCrMo) alloy and
- Cobalt-Nickel-Chromium-Molybdenum (CoNiCrMo) alloy.

Table 2 reports the composition of the most important alloys belonging to these classes used for the production of orthopedic implants.

CoCrMo has been used for many years in dental applications and for the manufacturing of artificial joints, while for prosthetic stem productions, for hip and knee joints the alloy containing Ni has been employed [21]. CoCrMo alloy, because of its high corrosion and wear resistance, is used for joint prostheses for the femoral head in combination with an ultra-high molecular weight polyethylene cup.

Alloys (ASTM)	Cr	Mo	Ni	Fe	C	Si	Mn	W	P	S	Other
F15	27–30	5–7	1.0	0.01	0.35 max	1.0	1.0	0.2	0.02	0.01	0.25N 0.3Al 0.01B
F799 (low C)	26–30	5–7	1.0	0.01	0.05	1.0	1.0	-	-	-	0.25N
F799 (high C)	26–30	5–7	1.0	0.01	0.25	1.0	1.0	-	-	-	0.25N
F563	18–22	3–4	15–25	0.01	0.05	0.5	1.0	3.0–4.0	-	0.01	0.50–3.5 Ti
F562	19–21	9–10.5	33–37	0.01	0.25 max	0.15	0.15	-	0.015	0.01	1.0 Ti
F90	19–21	-	9–11	0.01	0.05– 0.15	0.40	1.0–2.0	14–16	0.04	0.03	-
F1058	19–21	6–8	14–16	0.01	0.15	1.2	1.0–2.0	-	0.015	0.015	0.10 Be 39.0– 41.0 Co

Table 2. Composition (wt%) of some CoCrMo and Co-Ni alloys [5].

However, Ni and Co ions, when released, are susceptible to induce allergic responses: Ni is carcinogenic and causes a toxicity problem, which can be avoided using an as much as possible low level of Ni.

Actually, *Ti and its alloys* have been received much more attention, even if they have been used since the late 1960s. Due to the fact that Ti is completely inert and resistant to corrosion in any fluids in the human body and tissues, these alloys exhibit the best biocompatibility among metallic biomaterials combined to good fatigue resistance, excellent in vivo corrosion resistance which is acquired by the growth on their surface of the stable passive oxide layer. In addition, showing some properties which are close to the human bones, i.e., strength, density, and relatively low elastic modulus, these alloys began to be extensively employed for joint replacements. Lower elastic moduli than other metallic biomaterial allow obtaining less stress-shielding phenomenon. Generally, Ti and its alloys are used for bone fixation, joint substitution, pacemakers, implants for dental, artificial heart valves, components and stents in rapid blood centrifuges due to their chemical stability and particular high strength [24]. Central position has been made for a commercially *pure Ti* (cp-Ti, ASTM F67), *Ti6Al4V alloy* (ASTM F136) strengthened by alloys obtained by the modification of them and *Ti-based shape memory alloys* (SMA) [25, 26]. Cp-Ti is typically used in dental and spinal surgery, Ti6Al4V for the manufacturing of artificial knee, hip and shoulder joints and for bone fixators, while SMAs are employed as orthodontic equipment and temporary spine and long bones fixations.

According to ISO 5832-2, four grades of cp-Ti are recognized for medical applications. Cp-Ti can contain a minor level of interstitial elements (O, N, and H) affecting the mechanical properties

by interstitial solid solution strengthening. Cp-Ti is available in four grades, where (i) grade I corresponds to the lowest O content and yield strength and at the same time to the highest ductility and (ii) grade IV reveals the highest O content and the highest strength combined with the lowest ductility. **Table 3** reports the grades of different cp-Ti with the maximum limits of the interstitial elements and some mechanical properties. Due to the capacity for stimulating fast osseointegration, coming from the incorporation of OH⁻ ions inside the passive TiO₂ layer and due to the reaction of the hydroxylated surface area with the inorganic phase constituents (prevalently Ca²⁺ and (PO₄)³⁻) present in the bone, cp-Ti is principally used for endosseous dental implants manufacturing.

Ti6Al4V alloy reveals outstanding mechanical strength, high biocompatibility and consents good implant-bone integration. However, over the years, some weaknesses during the use of such alloy have been noticed: (i) it has higher elastic modulus than bone, producing stress-shielding effect, (ii) it shows lower wear resistance too, and (iii) the release of Al and/or V can determine permanent diseases, i.e., osteomalacia and Alzheimer's disease [27].

The research community has dedicated high attention to these problems, and the trend is to develop new Ti alloys, by alloying the basic alloy and reducing as much as possible or even totally removing the release of any toxic elements. In this scenario, alternatives alloy with ($\alpha + \beta$) structure, without V have been obtained, i.e., Ti6Al7Nb, Ti5Al2.5Fe, Ti6Al6Nb1Ta, and Ti5Al3Mo4Zr, which at the beginning have been developed for aerospace application. These alloys reveal higher corrosion resistance, fatigue properties than other biomaterials, principally governed by the size and the distribution of the α and β phases [28]. Such structure can be obtained during mechanically working till to obtain the preferred form followed by fast cooling at room temperature and annealing for recrystallization within the ($\alpha + \beta$) two-phase field and determine a fatigue strength more than 650 MPa.

β -Ti and near β -Ti alloys, i.e., Ti12Mo6Zr-2Fe, 35.5Nb7.3Zr5.7Ta, and Ti13Nb13Zr, introduced in the 1990s and their continuous improvement have been one of the leading topics of orthopedic materials research, because they contain superior amount of β -stabilizing elements and they are characterized by a considerably lower elastic modulus than other biometals (44–51 GPa for water-quenched and cold-worked Ti-13Nb-13Zr alloy compared to 110 GPa for Ti6Al4V). Additionally, good formability, high hardenability, excellent corrosion resistance, and better notch sensitivity (measure of how sensitive a material to notches or geometric discontinuities is) than the ($\alpha + \beta$) Ti alloys characterize such alloys. In order to amplify the wear resistance, surface hardening can be carried out by aging in an oxygen-rich environment

Ti grade	O	N	H	σ_{yield} (MPa)	σ_{ultimate} (MPa)	%E
1	0.18	0.03	0.015	170	240	24
2	0.25	0.03	0.015	275	345	20
3	0.35	0.05	0.015	380	450	18
4	0.40	0.05	0.015	483	550	15

Table 3. Maximum limits of the interstitial elements and mechanical properties of different cp-Ti grades.

determining the growth of a hard oxide film on the surface with an interstitial solid solution strengthening of the subsurface section atmosphere [29].

A class of special Ti alloys, developed in Japan, with exclusive physical-mechanical properties and large range of possibilities in medical applications are Gum alloys, belonging to the β -type of Ti alloys, fundamentally expressed as Ti (Ta, Nb, and V) + (Zr, Hf, and O). These alloys have excellent mechanical behavior at room temperature: an ultra-low Young's modulus (60–70 GPa) and a non-linear elastic behavior, an extended elastic limit, ultra-high strength (>1 GPa), superplastic-like deformability, Invar-like thermal expansion, and Elinvar-like thermal dependence of the elastic modulus making available several prospects for their application [30]. Such alloy, with no hazardous elements, combines extremely low elastic modulus with extremely high strength and can be considered a perfect candidate for many medical implants.

The capacity to return to the initial memorized shape varying the temperature is the most important features of *shape memory alloys*. Such a particular characteristic can be exploited in critical medical application, when recovering the original shape after large deformations induced by mechanical load and for conserving the deformed shape up to the heat-induced recovery of the original shape has primary importance [31]. Generally, in orthopedic, dental, and cardiovascular uses NiTi alloy (Nitinol) is employed because its good corrosion resistance and pseudoelastic property (reversible elastic response to an applied stress). However, the study [32] reports that at temperatures equal to the human body the formation of a passive layer on NiTi is not as much protective than the passive layer on Ti6Al4V. Additionally, there are some limitations too in the widespread use of such alloy because even if some studies have revealed that NiTi alloy is biocompatible [31, 33, 34], the possible release of Ni ions represents a serious apprehension [13, 15] because Ni is highly allergenic element, and it can induce intense inflammatory reactions.

Zr, Ta, and Nb have been employed as constituent elements of Ti alloys for medical applications because of their good biocompatibility and high corrosion resistance. Zr belongs to the same group as Ti and exhibits similar properties [35]. Recently, Zr-Nb [36, 37] and Zr-Mo [38] have been developed. Ta and Nb are non-toxic elements, and they exhibit very similar physical and chemical properties. Ta exhibits excellent chemical stability and good biocompatibility similar to that of Ti. Ta has been employed in dental and orthopedic applications such as radiographic bone markers, vascular clips, and as a material for cranial defect repair and nerve repair since the 1940s [39, 40]. In particular, porous Ta has been developed for bone in-growth applications, i.e., hip and knee arthroplasty, spinal surgery, and bone graft substitutes [41, 42]. Introduction of Ta in TiNi shape memory alloy determines a low incidence of artefacts, according to [43, 44], the blood compatibility of Ta is higher than that of other metals.

Recently, among other research, *biodegradable metals and their alloys* have attracted high interest, because of their possibility of degradation in biological surroundings. The main benefit of this class of materials is related to the fact that after the tissue has appropriately restored and they are no longer useful, the removal of follow-up surgery is avoided allowing higher protection since physical irritation and chronic inflammatory local reactions are totally excluded [45–47].

Fe, Mg, and Zn are all vital nutritional elements for a healthy body and generally reveal excellent biocompatibility in the human body, with no any sign of local or systemic toxicity biodegradable materials than polymers for load-bearing applications. Both pure Fe and pure Mg have been reported to possess excellent biocompatibility in the human body and show no signs of local or systemic toxicity [45, 48, 49].

Mg and its alloys (i.e., (AZ91, WE43, AM50, and LAE442) can be considered as favorable alloys for medical applications, and they have already been positively verified *in vivo* and in clinical studies and it has been reported that development of new bone is facilitated [50–52], when they are implanted as bone fixtures. The elastic modulus of Mg (41–45 GPa) is closer to that of natural bone (3–20 GPa). However, the mechanical properties of such alloys, i.e., strength and the elongation to fracture are not always acceptable which combined to the development of significant quantities of hydrogen limits their widespread use. Some studies [53–55] have reported as alternative solution the use of Fe-based alloys. However, the implants realized using such alloys reveal some reactions like those developed in long-term uses which is predominantly due to the low-degradation rate of pure Fe in biological atmosphere [56].

In Refs. [53, 54] the authors have developed Fe35Mn alloy with a higher degradation rate compared to the pure Fe; however, compared to Mg-based alloys, the interested value is still at least one order of magnitude inferior.

Even with the significant evolution carried out over the time on the implants made of by biodegradable alloys, many tasks are still unexplained.

1.3. Biomaterials in tissue engineering and regenerative medicine

Current evolutions of biomaterials and the constantly increasing demand for new technologies have transformed the field of TE and RM too. In such areas, biomaterials have received more and more attention and can be placed in pole position for repairing tissue functions by joining materials design and engineering with cell therapy. RM is considered new frontline of medical research; however, the idea of generating synthetic tissues is not so recent, since it goes back to 1938 [57]. Biomaterials can offer physical sustain for engineered tissues and can control and guide the cells. RM involves the replacement or restoration of human cells, tissue or organs, to repair or create typical function [58, 59].

Biomaterials engineering involves production, processing, characterization, and knowledge of innovative materials, including metallic ones. The most important limitations of RM is related to the generally health-care programs: the inclination to substitute conservative methodologies with new therapies is not so immediate. A large use of RM would be possible when a deep knowledge related to all connecting aspects (positive and negative outcomes) will be absolutely solved [60]. For the use on large scale RM, the ideas developed at laboratory level have to be inserted and totally transformed into widespread commercial products. In the past century, achievements in medicine have included different approaches and fundamentally, three different lines can be really able to follow the objective of RM: (i) cell-based therapy (meaning of introducing new and healthy cells in pathologic tissues), (ii) use of both biological

or synthetic materials (use of cells and extra cellular matrix providing structural and functional support), and (iii) implantation of scaffolds seeded with cells (mixture of the earlier mentioned two possibilities) [61].

Generally, metallic ions have a significant role for some day-to-day living biological processes, being essential for many biochemical reactions: i.e. K, Na, Ca, Mg. Ca^{2+} ions level governs the intra- and inter-cellular communications, blood coagulation and muscle contraction, Mg^{2+} ions support photosynthesis, the electron transfer routes are generally based on Fe proteins, transport of oxygen requires Fe and Cu containing proteins, Zn has the main role in regulation of DNA transcription are only some cases which highlight the importance of them [62–66].

In addition, metallic ions have interesting capacity and can be exploited as therapeutic agents in tissue engineering. There are a large number of metallic ions which are vital cofactors of enzymes (Co, Cu, Fe, Mn, Zn, Ag, Sr, V, and Ga), and their specific properties can be correlated to their use for therapeutic purpose. The geometry and valence of metal ions, their hydrolytic and redox activity, Lewis acidity, radiochemical properties directly involve fast kinetics, higher affinity to interact with other ions with high flexibility to be inserted in engineered biomaterials allowing the modification/revision of cellular functions and their metabolism. Actually, there is a continuous growth in medical field of the use of metal ions in RM and tissue engineering, with a special attention concerned to their therapeutic properties. However, the possible toxicity of metal ions in case of their eventually local release has to be carefully considered, when exploiting them [67–70].

2. Real case study

Some fundamental aspects related to metallic biomaterials and their use in RM and TE are integrated with some experimental results obtained by the authors [71–75] during the years spent studying the development and characterization of metallic materials for medical purpose. In particular, some results, (1) prevalently related to the corrosion resistance, which can be directly associated with their biocompatibility of some (i) Ti-enriched Co-based traditional alloys to be submitted for dental application and (ii) TiNb alloys for load-bearing implant production will be reminded here; (2) the third line (iii) involves the possibility of using an implanted two-element antenna array placed on a metallic cylinder (metal made implant) embedded in a polymer dielectric that acts as the substrate between the ground plane and radiators with the purpose to use it for in-body communications.

2.1. Some details and the corrosion resistance of the experimentally prepared biometals

Cold crucible levitation melting technique with induction heating system has been used for the metallic alloys manufacturing, which guarantee high purity melts, important features for materials used for surgical implants. Enhancement with Ti has been performed in the case of the Co-based basic alloy. **Table 4** reports the chemical compositions of the alloys for dentistry, while **Table 5** reports the composition of the alloy used for load-bearing implant production.

Alloys	Chemical composition (wt%)					
	Cr	Mo	Si	Ti	Other	Co
CoCrMoTi4	26.5	6.0	1.0	4.0	1.5	bal.
CoCrMoTi6	28.0	4.5	1.0	6.0	0.5	bal.

Table 4. Chemical composition of the experimentally developed alloys.

Alloy	Ti	Nb	Ta	Zr	O
TiNb	71.4	24	1	3	0.6

Table 5. Chemical composition (at %) of the Ti-Nb alloy.

Wirobond®280 (Co60.9, Cr25, Mo4.8, W6.2, Ga2.9) and Ti6Al4V have been considered as reference material. The microstructures of the alloys investigated are reported in **Figure 1**, where a relatively homogeneous structure has been observed. Major details are reported in Refs. [71–73].

Following some structural and microstructural characterization (**Figure 1** reports the general morphology of the investigated alloys) combined with some mechanical properties evaluation, which can be found in [71–73], some words are spent about the corrosion behavior, verified by static immersion test, of the metallic alloys prepared, because such line is directly associated to the biocompatibility behavior of the metal alloys. The test has been carried out according to the route specified in the Standard ISO 10271/2011 at 37°C (±1°C). According to the standard, immersion of the samples has been realized in acid solution (7.5 ml lactic acid, 5.85 g NaCl, 300 ml H₂O di grade 2 purity, and 700 ml H₂O) simulating the biological environment. The pH value has been corrected to arrive to the neutral situation. The samples have been monitored after the permanence in the acid solution for 28 days. As a reference solution, an acid solution maintained in the same condition without any metallic alloy inside has been used. The sample surfaces, following corrosion test, have been investigated by scanning electron microscopy

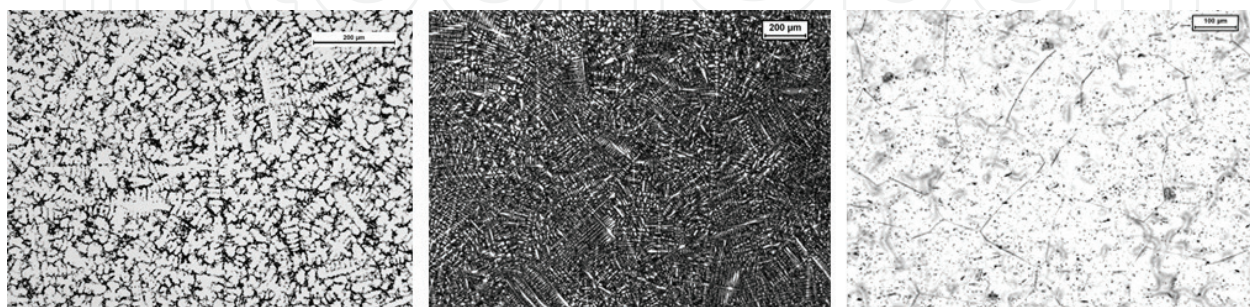


Figure 1. Optical micrographs of the investigated alloys.

analysis. The hydrogen ions measurements in the solution with the metallic alloys inside and the weight loss measurements are directly correlated to damage arisen and indicate the progress of the corrosion which involves the metallic alloys. After a regular time, the weights of the metallic samples have been measured, and the pH values have been monitored. Comparison of the results obtained for the solution with and with no metallic alloys inside has been carried out. The investigated alloys do not present any significant weight alteration after 28 days and no significant release of metal ions was observed (**Figure 2**). The same results have been confirmed by the pH value measurements (**Figure 3**). When the dissolution of the passive film developed on the surface of the alloy has a very low rate, the layer is able to protect the alloy. The release of metal ions is lower than 0.1% within the week, and this is below the threshold level indicated in the Standard (according to the Standard ISO 10271/2011 the allowed deviation from the original state could be only 1%).

Microstructural observation and compositional analysis after corrosion test confirm the analytical results, since on the surface of the alloys extracted from the physiological solution, no evidence of corrosion product has been found. The experimental alloys are used for some crown (**Figure 4** left) and load bearing implant (**Figure 4** right) manufacturing. The research in this direction is ongoing to further investigate additional performance of the alloys as much as possible in functional environment.

2.2. Some details about the use of the implant itself as ground plane for implanted antenna

Biotelemetry, increasingly exploited in the recent healthcare systems, involves the application of telemetry in medicine and health care, allowing remote checking of various vital functions

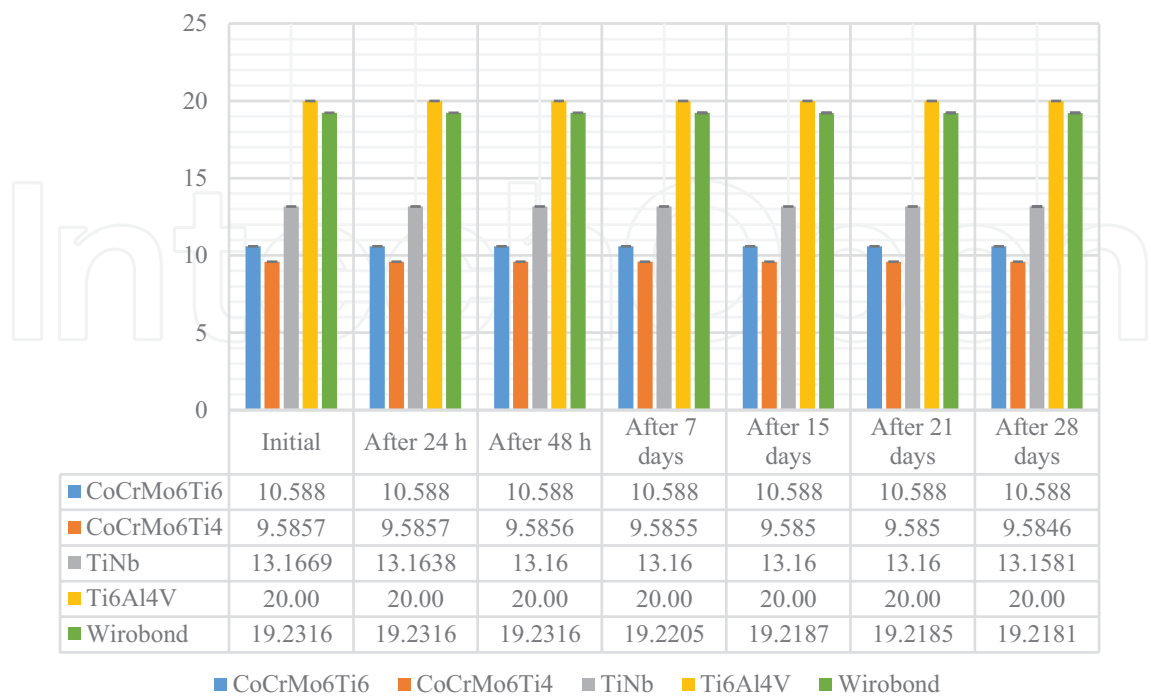


Figure 2. Weight loss measurement evaluation results during 28 days of investigation.

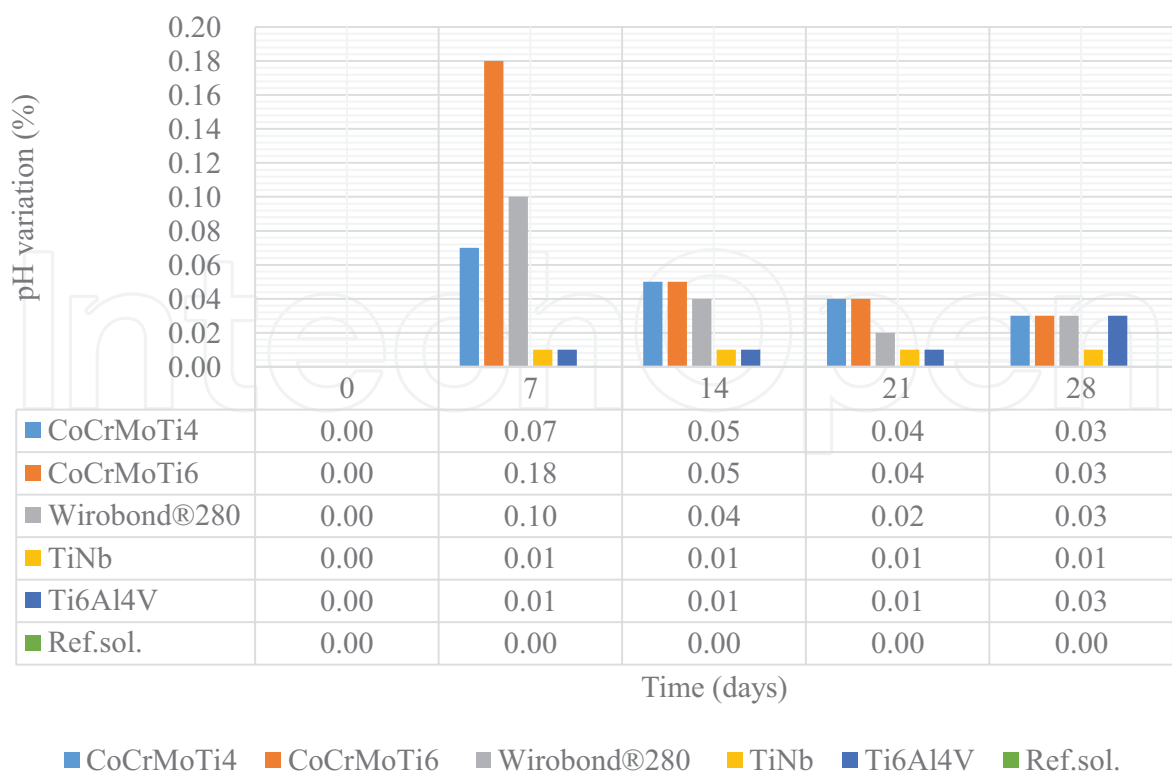


Figure 3. Variation of the pH value during 28 days of investigation.

(i.e., temperature, blood pressure, and cardiac beat) in patients. The use of such technologies, firstly, can considerably decrease hospitalization time and then allows the constant collection of the patient’s biological data. When the information has to be obtained from inside the body, it should be firstly revealed by a dedicated sensor and in a second step to be transmitted to an external (to the body) receiver. This operation requires the use of antennas that are made of conductive materials, i.e., metal. Such devices are in a direct contact with the different tissues, so special attention to their biocompatibility has to be paid. On the other hand, to increase the efficiency of the transmission, directive antennas are to be used, which implicitly means a large dimension. Such space is not always available inside the body. An alternative solution consists of using low-profile printed antennas, but they require the use of a large ground

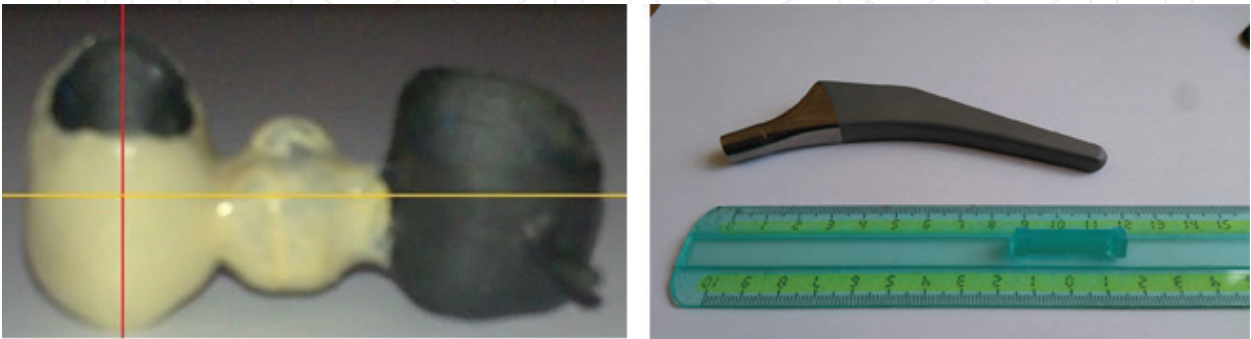


Figure 4. Photographs of the produced elements for medial application.

plane. A recently proposed solution by the author consists of the use of the implant itself as ground plane for the antenna. The rectangular radiator is made of a similar metal as the implant, i.e., biocompatible, and it is conformal to the cylindrical bone structure. The reported results in [74] refer to the case when a piece of bone is fully substituted by a biometallic cylinder. To maintain the distance between the ground plane and radiator constant, a biocompatible polymer, namely polydimethylsiloxane (PDMS), has been considered both as dielectric and superstrate. The geometry is presented in **Figure 5**.

The thickness of each of the two PDMS layers is of $h_{PDMS} = 2\text{ mm}$ with respect to the radius of the overall geometry of 106 mm (similar to an adult arm). Details on the dielectric properties of the tissues and materials involved in the numerical experiment can be found in [74].

As a result, a wide beam, approximately 100° half power beam width (HPBW) radiation pattern in the orthogonal to the bone axis plane has been obtained. Such characteristics allow a significant freedom to the patient to move without losing the communication link between the implanted antenna and fixed external base station.

However, considering the low efficiency, basically due to the lossy medium, and of the deep positioning of the antenna (with respect to subcutaneous solutions), and exploiting the low profile configuration, as a second step two radiators have been inserted, aiming to realize a two-element array with scanning possibility. It should be mentioned that this solution does not require additional space, since the second radiator is embedded in the already present

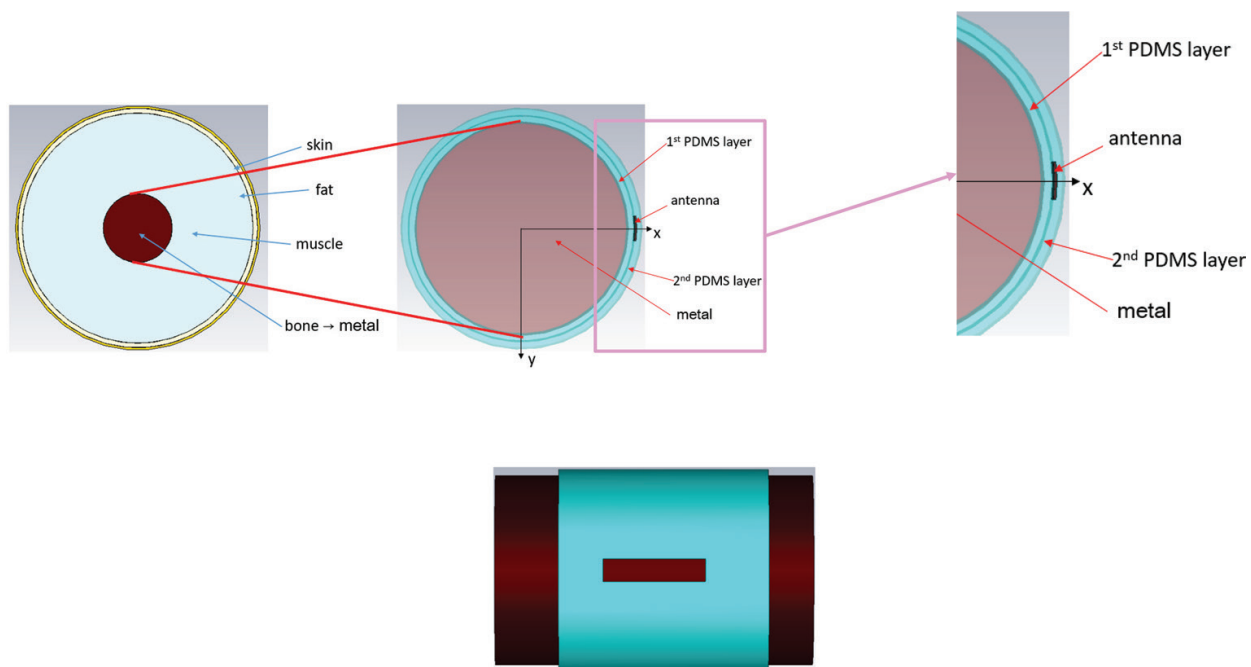


Figure 5. Numerical adult leg model: the bone is substituted by an equal diameter cylinder (left); the conformal antenna is positioned on a constant thickness PDMS dielectric layer of ring shape around the biometallic implant (central), side view of the structure (with removed tissues and outer PDMS layer for a better rendering (right)).

PDMS layer. Because of the available space, different configurations have been tested to reduce the mutual coupling between the two radiators. The parametric study allows quantifying the coupling for different angular distances between the two radiators in the presence and absence of the coupling reduction elements. In all cases, metallic parts made of the same biometal have been considered. Due to the low coupling a more directive, i.e., higher gain, system can be built. The higher gain associated to the possibility to scan the beam further increases the freedom of movement of the patient the system is implanted in. All these features could be obtained at a very low extra space demand, basically due to the exploitation of the metallic implant as ground plane of the planar antennas.

From the manufacturing point of view, the proposed solution, i.e., antenna and coupling reduction structure, can be implemented on the biometallic cylinder before it is implanted. Moreover, the metallic cylinder could also incorporate the necessary electronics, sensors, beam forming network, etc. and power supply (batteries).

Similar solution using two dipole antennas has also been discussed in Ref. [75] where conical implant has been considered. The presence of the bone surrounding the implant eliminates the use of any additional materials, i.e., the PDMS layers, in the previous case. In particular, two orthogonal dipoles in turn-style like configuration have been considered. This solution allows to generate a circularly polarized radiated field, further increasing the freedom of movement of the patients.

3. Conclusions

The aim of this chapter was twofold. The first section was dedicated to offering a short outlook and to demonstrating on how metals and their alloys can be used for medical purpose focusing on their role and their influence on the surrounding body environment. The chapter has illustrated some significant features which are appropriate to such a widespread topic. Secondly, some experimental results achieved by the authors over time and illustrated here were proposed to partially support the theoretical line by providing a real case study.

In particular, some results about the biocompatibility of Ti enriched Co-based traditional alloys and TiNb alloy for medical implant production and the possibility of using an implanted two-element antenna array placed on a metal made implant inserted in a polymer dielectric acting as substrate among the ground plane, and radiators with the purpose to use it for in-body communications were highlighted.

There is a continuous need to obtain a deep understanding about the roles and the effect of the metallic ions in body environment. The interdisciplinary character of such research was pointed out with the aim to encourage the teamwork between material scientists, electromagnetic engineers, biologist, tissue engineers, and biomedical and medical researchers to develop new biomaterials and to acquire sufficient data about biological processes as a function of foreign biomaterial introduced in the human body.

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