ELECTROCHEMICAL PROCESSES IN BIOMATERIALS DEGRADATION

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Stainless steels, cobalt-chromium and titanium alloys are the most common materials used in orthopaedic and dental surgery, either as temporary scaffolding during healing of bone fractures or as permanent replacement to bone. However, the human body constitutes, a very aggressive environment towards these materials, leading to their premature failure by corrosion, wear or a combination of both. These processes are, in turn, responsible for the release of toxic ions which accumulate in the tissues surrounding the implant or in other parts of the body.

Crevice corrosion accounts for about 90% of all the cases of corrosion found in orthopaedic surgery. Bone plates are particularly susceptible to this type of attack, which occurs on the surfaces which come into contact with fixing screws and not at the bone/plate interface. Around stainless steel plates high concentrations of chromium are found, while iron and nickel are carried away in the blood stream.

In vitro and in vivo electrochemical techniques are particularly useful in studying the processes of degradation of metallic materials. The necessity to keep the number of experiments with animals to a minimum and the need to test a wide range of new materials will lead to an increase in the use of electrochemical techniques. Their usefulness will be documented with some typical cases: selection of alloy composition for dental materials, use of metallic screws for the fixation of carbon plates, metallic porous coatings for cementless prostheses, and coating of prostheses with ceramic materials for increased biocompatibility and corrosion and wear resistance.

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1. Introduction

The types and applications of biomaterials are very wide, ranging from conventional materials, like 316L stainless steel used for fixation plates in orthopaedic surgery, to resorbable polymers, useful in drug delivery. Barenberg [1] gives a comprehensive list of fields of development and important issues in the production and testing of biomaterials. About 50% of the world production comes from the U.S.A.. A figure often quoted for the annual growth rate of the biomaterials industry is 15%. Metallic materials, and particularly stainless steels, Co-Cr alloys and Ti and its alloys, represent the greatest majority of the biomaterials market. In spite of the progresses made with bioceramics and biopolymers, the mechanical strength of the above metals makes them still unreplaceable for most orthopaedic applications. However, their high modulus of elasticity, in comparison with that of bone, is responsible for stress shielding, resulting in bone resorption. This is one of their main disadvantages, being often claimed that Ti and its alloys would be preferable to stainless steel and Co alloys because its modulus of elasticity is about half of that for the latter materials [2]. Another disadvantage of metallic materials is their susceptibility to corrosion in the body fluids, which will be addressed in the following section.

2. Degradation of metals in the body

Corrosion, wear and fatigue are the most frequent causes of biomaterials failure. The release of toxic ions and the incorporation of wear debries are reasons for concern, particularly because very high concentrations can be reached in the vicinity of the implants. Table I gives some figures for the contamination of tissues surrounding stainless steel, titanium and Co-Cr-Mo implants [3]. Very high concentrations are normally associated with presence of metallic particles which slowly dissolve and/or are phogocytized by macrophages. The values in Table I can be compared with the normal contents of Cr, Ni and Fe in blood, plasma (or serum), sweat and bone [4] given in Table II. The difference creates a concentration gradient, hence inducing a general increase in the metal content of blood and tissues away from the implant.

The most frequent type of attack is crevice corrosion, often accelerated by fretting. Retrievel studies, mainly on fixation plates made of stainless steel, have shown that crevice corrosion accounts for ca. 90% of all cases of corrosion [5,6]. In one of these studies [5] 70% of the devices were corroded. Attack is often located at the surfaces of contact between plate and screw (Fig. 1) and not at the bone/plate interface [3,5]. Fretting between material and bone may however occur, particularly in the case of loose hip prostheses [5]. Fig. 2 shows severe attack at the tip of the stem of a hip prosthesis.

TABLE I - CONCENTRATION OF METALLIC ELEMENTS IN TISSUES AROUND IMPLANTS[3]

MATERIAL	SAMPLE	CONCENTRATION (ppm*)
316L S.S	Fe	300 - 20000
	Cr	<d.l. - 10000
	Ni	<d.l. - 1400
Ti	Ti	<d.l. - 73000
Co-Cr-Mo	Fe	90 - 3650
	Co	<d.l. - 2200
	Cr	<d.l. - 10200
	Ni	<d.l. - 1500

*ppm of dry tissue

TABLE II - CONCENTRATION OF METALLIC ELEMENTS IN BIOLOGICAL SAMPLES[4]

ELEMENTS	SAMPLE	CONCENTRATION (ppb*)	
Cr	Blood	2-6	
	Plasma or Serum	0.04-0.5	
	Sweat	25	
	Bone	460	
Ni	Blood	2.9-7.0	
	Plasma or Serum	0.6-6.5	
	Sweat	7-270	
	Bone	900	
Fe	Blood	200-680	ppm
	Plasma or Serum	0.4-1.70	"
	Sweat	0.020-0.630	"
	Bone	91	"

* wet weight

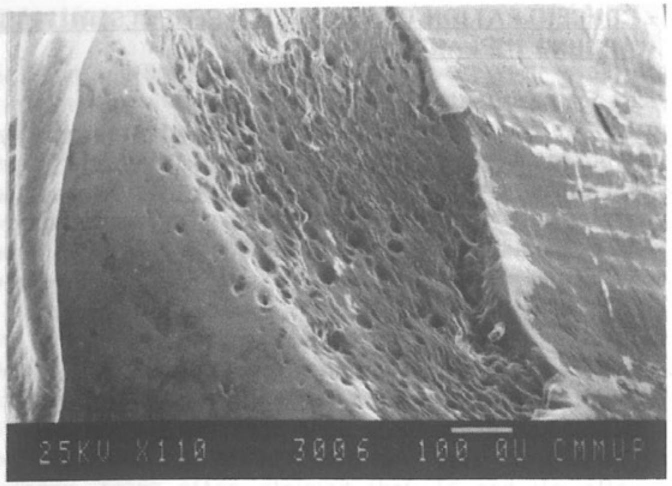

Fig. 1 - Intense Corrosion at a hole of a fixation plate made of AISI 316L stainless steel[5]

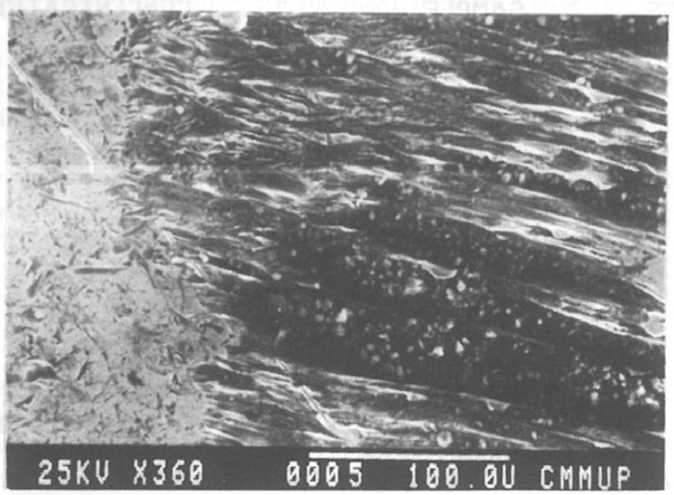

Fig. 2 - Attack of the stem of a Charnley type hip prosthesis made of 316L stainless steel[5]

Tissue reactions to released ions depend on their specific toxicity. In this respect Ti is practically non-toxic [2] and no cases of rejection of Ti implants are known. Presence of V in the common Ti6Al4V alloy is a matter for concern due to its toxicity. This has led to the development of V-free titanium alloys [7]. The toxicity of different metal salts on embryonic rat femura is given in Fig..3, showing that V has a toxicity 10 times greater than Ni, Cr and Co. Data on the toxicity of other elements is given by Tsalev [4]. The subject will be addressed again in this work in conjunction with the electrochemical properties of biomaterials.

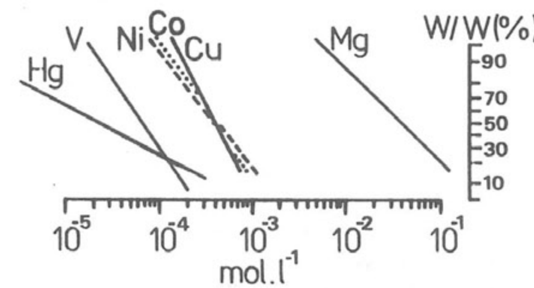

Fig. 3 - The influence of metal chlorides on the growth of embryonic rat femora. In the ordinates is represented the weight ratio between femora grown in media containing metal ions and those grown in metal-free media[7]

3. Passive surface characteristics

All metals used as biomaterials owe their corrosion resistance to the presence of an oxide/hydroxide film, known as the passive film. Its properties are therefore very important in guaranteeing a high degree of resistance to localized attack.

Comparison between corrosion resistances of different materials are often made with the help of polarization curves. Essential potentials in these curves are (see Fig. 4) the passivation potential, E_{pass}, and the pitting potential, E_p. Since localized corrosion is the most frequent cause of failure in these materials, comparisons based on E_p are often used.

Table III gives E_p values for several materials, showing that Ti and its alloys are far superior to stainless steel and Co-Cr-Mo alloys. In the absence of fretting Ti is in fact practically immune to corrosion, due to its high E_p in comparison with the free corrosion potential in the body. The high levels of Ti found in tissues surrounding Ti implants is often attributed to fretting and fretting corrosion. Several surface treatments can be employed to improve the resistance to these forms of degradation, some of which will be indicated below.

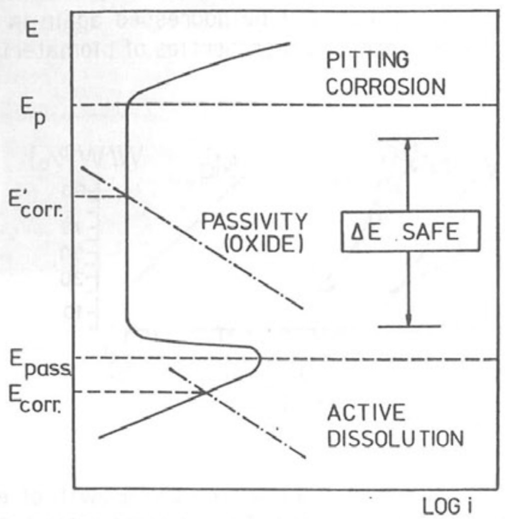

Fig. 4 - Schematic anodic polarization curve of a passive metal. The metal can be used safely within a potential interval in the passive region, not too close to the regions of film instability. Within crevices the corrosion potential can sometimes fall in the active region (E_{corr}). External surfaces of fixation plates are normally in the safe potential region (E'_{corr})

In the absence of mechanical stresses, passivation treatments are often used to improve the corrosion resistance of biomaterials. The treatments are carried out in solutions of HNO_3 and although the exact changes occurring on the surface are not known in detail the effect is normally an improvement of the resistance to breakdown. Significant improvements occur for stainless steel, (8-10) sometimes of the order of 600 mV in E_p (9). An explanation based on Cr enrichment of the film, due to selective dissolution of Fe, has been put forward by Asami & Ashimoto (9). An alternative explanation is removal of sulphide inclusions during the passivation treatment(10). Sulphides constitute the majority of inclusions in stainless steels and are often preferential sites for attack. For Co-Cr-Mo alloys the effect of passivation treatments is not as marked as

TABLE III - Breakdown potentials for implant metals in Hank's solution

Metal	Breakdown Potential, vs. Sat Calomel Electrode
Type 316L stainless steel	+0,2 to + 0,3[a]
Co-Cr-Mo	+0,42
Co-Ni-Cr	+0,42
Ti-Ni	+1,14
Ti6Al4V	+2,0
Tantalum	+2,25
Titanium (pure)	+2,4
Ti-4,5Al-5Mo-1,5Cr	+2,4

a Breakdown varies within this range

for stainless steel (11), whereas for Ti6Al4V the corrosion rate can decrease by a factor of 40 (12). Treatment in HNO_3 induces the transformation of TiO into TiO_2 (anatase) (13), a process that would ensure a higher degree of corrosion resistance. TiO_2 is also the oxide formed upon anodic oxidation of titanium (14); whereas TiO is found on surfaces exposed to water and 3.5% NaCl solutions (13).

Under fretting the process of degradation is accelerated. Small particles are released from the rubbing surfaces, damaging the passive layer, which is thus in a permanent process of rupture and repair. As a consequence, the dissolution rate increases and the tissues become impregnated with high amounts of metal. The poor wearing properties of Ti and its alloys are a major drawback to their use as hip prostheses. Several solutions have been attempted, including dense alumina and Co-Cr-Mo heads, normally wearing against UHMWPE (Ultra High Molecular Weight Polyethylene). Coating titanium surfaces with a hard coating, such as titanium nitride, has been used (15) but the process is still in the development stage. The implantation of nitrogen ions produces considerable improvements in corrosion and wear behaviour (12,16). Significant decreases in wear rate were found after implantation with C^+ ions (17). The surface modifications responsible for the effects of ion implantation are not yet well characterized. The hardness of surface layers may change as a result of modifications in the composition of the titanium oxide (which may be converted into an oxynitride) or of suboxide layer formation of hard compounds, e.g. titanium nitrides or carbides. Electrochemical changes in the properties of the implanted surfaces must be involved in the tribological processes occurring in liquid media, since tests have consistently shown a decrease in the corrosion rate by a factor of 40 upon implantation of Ti6Al4V with N^+ ions [12].

The electronic properties of passive films can be useful in interpreting the corrosion resistance of the parent metals [18], particularly if they provide information about the type of conduction and oxide structure. Passive films have semi-conducting characterists and thus concepts like donor density and flat band potential, used in the characterization of semi-conductors, have been introduced in the study of passive surfaces [19,20]. The Mott-Schottky relationship,

$$\frac{1}{C^2} = \frac{2}{\varepsilon \varepsilon_o e N_D}\left(E - E_{fb} - \frac{KT}{e}\right)$$

where

C = film capacitance
ε_o = permitivity of vacuum
ε = dielectric constant
e = electron charge
N_D = density of charge carriers
E = electrochemical potential
E_{fb} = flat band potential
K = Boltzmann constant
T = absolute temperature

is obeyed for passive films on stainless steel and titanium in contact with a physiological solutions [21]. A curve representative of the behaviour of AISI 316L stainless steel is given in Fig. 5. Extrapolation of the linear part gives E_{fb} (E_{fb}=-0.365 V/SCE). N_D can be obtained from the slope of the straight line (N_D=2.03x10^{21} atom/cm^3). The donor density for titanium oxide was more than one order of magnitude lower than for the film on stainless steel. Assuming that electron transfer is the rate determining step in passivation kinetics, the results can be used to explain the superior corrosion resistance of titanium. A low electronic conductivity is a necessary but not sufficient condition for low susceptibility to corrosion, since initiation of attack is often associated with non-metallic inclusions.

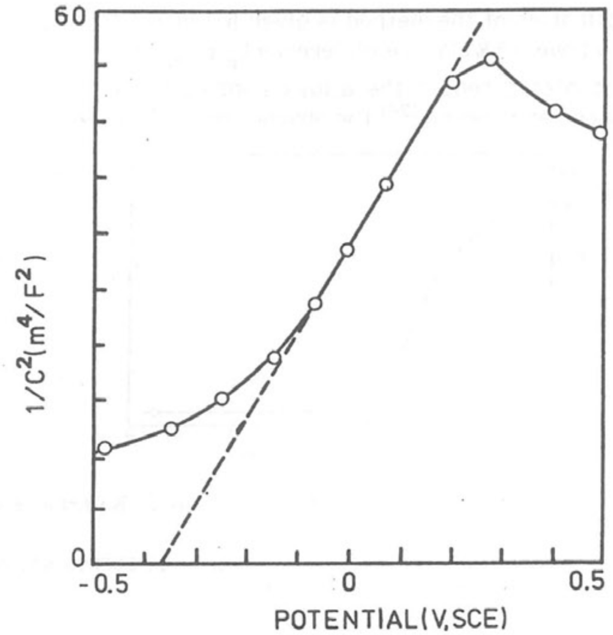

Fig. 5 - Mott-Schottky plot for AISI 316L stainless steel in a physiological solution containing 137 mM Na^+, 4.0 mM K^+, 6.6 mM Ca^{2+}, 5.0 mM Mg^{2+}, 110 mM Cl^- and 3.68 mM acetate[21]

4. <u>In vitro evaluation of corrosion resistance</u>

4.1. General Applications

In the previous section a criterium for comparing corrosion resistance was given, using the value of the breakdown potential taken from anodic polarization curves. This method gives an indication of the resistance to attack initiation, which occurs as pitting corrosion. As seen before, this is not the most frequent type of attack, crevice corrosion being responsible for about 90% of all cases of corrosion in fixation plates. An adequate method of screening materials for resistance to crevice corrosion consists of analysing the hysteresis loops produced in anodic polarization curves after passivity breakdown. Briefly, the method consists of measuring the difference between the repassivation potential, E_{rep}, and E_p. E_{rep} is the potential for which the polarization current reaches again passive values, i.e. after the process of pit growth has ceased. The degree of crevice corrosion susceptibility in stainless steels[22] and Co-Cr-Mo[23,24] alloys has been related to the difference E_p-E_{prot}. As the difference becomes smaller the resistance to crevice attack should increase. An application of the method is given in Fig. 6 for Mo addition to Co-25Cr alloys. Above 3.5% Mo, the difference $E_p - E_{prot} = 0$, which signifies that higher Mo contents render the alloys immune to crevice corrosion. According to Cahoon and Cheung [24] the absence of Mo from wrought Co-Cr

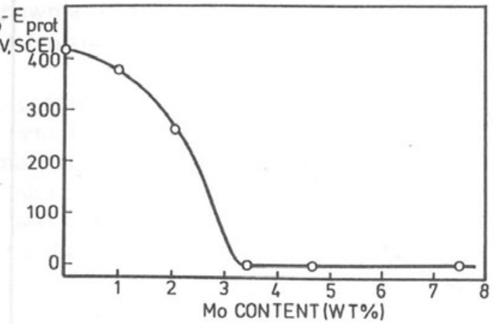

Fig. 6 - Effect of Mo on the protection potential of Co-25% Cr-Mo alloys[24]

alloys is responsible for their susceptibility to crevice corrosion, whereas cast Co-Cr-Mo are immune due to presence of Mo.

Both methods described above imply the application of large overpotentials and hence they do not provide information about corrosion rates under normal physiological conditions. The latter normally involve very low dissolution rates, assuming that localized attack has not installed itself. Under these circumstances the release of toxic species would be minimal and therefore measurements of corrosion rates would have little practical significance. Under fretting, however, the rate of metal ion release may become extremely large and then corrosion rate determinations may be important. They are often done by indirect methods, i.e. either analysing urine and blood for the wanted metallic elements or determining the polarization resistance, R_p,

$$R_p = \left(\frac{dE}{di}\right)_{E=E_{corr}}$$

either by linear sweep polarization or by the a.c. impedance technique, in which case

$$R_p = \left(\lim_{w \to 0} R_z\right)_{E=E_{corr}}$$

where R_z denotes the real part of the complex faradaic impedance and w the angular frequency of the a.c. signal. In both cases R_p and i_{corr} are related by the Stern-Geary equation

$$i_{corr} = \frac{ba.bc}{2.303\,(ba+bc)} \cdot \frac{1}{Rp}$$

where b_a and b_c are, respectively, the anodic and cathodic Tafel slopes. R_p measurements by the d.c. technique have been conducted on Ni-V-Cr casting dental alloys [25] and on alumina sprayed stainless steel [26]. Due to the small current densities encountered (ca. 1 $\mu A.cm^{-2}$) in the latter work the authors could not, however, discriminate the influence of passivation treatments and size of Al_2O_3 particles. Other applications of R_p measurements will be exemplified later on, when a possible correlation between biocompatibility and electrochemical parameters is discussed.

4.2. Applications to Systems Consisting of Different Materials

Avoidance of contact between different materials constitutes a generalized rule of thumb in corrosion engineering and biomaterials are, in this respect, no exception. In spite of the potential danger of acceleration of corrosion kinetics on the least noble material, one must consider the dangers of galvanic coupling in a correct perspective. Consider for example the anodic polarization curve of Fig. 4. Coupling a passive material to another material does not necessarily lead to catastrophic consequences, provided that E_{corr} does not fall or rise to potentials close to E_{pass} or E_p, respectively. Keeping this in mind the question one should ask is wether coupling between a particular set of materials can be considered safe or unsafe. The question is revelant for a number of applications, like for instance: carbon fibre reinforced carbon (CFRC) plates fixed to bone by metallic screws [27], titanium alloy hip prostheses with a cobalt alloy head [28], carbon coated metallic heart valves and carbon dental implants with metallic cores.

Direct coupling and measurement of the galvanic currents [28-30], as well as predictions based on the mixed potential theory [11,29-31], have both been used. The latter method is illustrated in Fig. 7 for a stainless steel/carbon couple. The anodic polarization curve of stainless steel was drawn in a deaerated solution, considering that corrosion occurs almost always at crevices, where a medium depleted in oxygen exists. The cathodic polarization curve of carbon was drawn in a solution saturated with oxygen.

A comparison between both methods is given in Table IV for couples of carbon with 316L stainless steel, a Co-Cr-Mo alloy and Ti6Al4V alloy[30]

TABLE IV – CORROSION CURRENTS (nA) FOR GALVANIC COUPLES WITH CARBON

METHOD	316L S.S	Co-Cr-Mo	Ti6Al4V
Z.R.A.	1300	120	5
M.P.T.	920	16	1035

Z.R.A. – Zero resistance ammetry
M.P.T. – Mixed potential theory

A good agreement is obtained for the first two alloys, whereas for Ti6Al4V the discrepancy is very significant. For this alloy application of the mixed potential theory gave too high corrosion rates for considering them representative of the actual currents flowing in a couple. The reason for the discrepancy is probably associated with the fast potentiokinetic technique used for determining the anodic polarization curve, which does not enable a sufficiently protective oxide to form on the electrode surface.

4.3. Applications to Coated Materials

Coated biomaterials are relatively new. Metallic beads or mesh have been used to providing a porous surface into which connective tissue can grow, as a means of strengthening the bonding of implants to bone. This method has gained great importance in femural hip replacement, in which smooth prostheses were previously fixed to the bone by an acrylic bone cement. High failure rates with use of cement in active patients [38], temperature rise during polimerization (leading to tissue necrosis), and toxicity of the monomer are leading progressively to the replacement of smooth prostheses by porous prostheses.

Prostheses with a metallic porous coating have, however, some drawbacks. One is bead loosening and the other is increased metal ion release. Loose beads were found in the synovial fluid after revision arthroplasty involving a cast cobalt chromium coated tibial plateau[39]. An increase in surface area, the occluded geometry of the pores and the possibility of occurrence of crevice attack at the points of contact between beads or

wires may all be responsible for enhanced metal ion release. The pitting potential of stainless steel decreases by ca. 400 mV upon coating with metallic beads of the same material [40], as seen in Fig. 8. The same figure shows that no film breakdown occurs for titanium. In this case the passive current density increases as a result of the greater surface. Based on these simple screening electrochemical tests it may be concluded that titanium can be used without risk of localized corrosion, whereas coated stainless steel is not recomended.

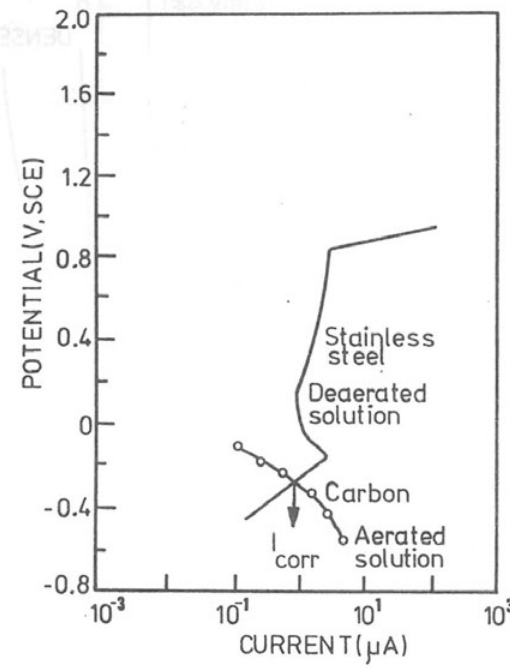

Fig. 7 – Application of the mixed potential theory to the prediction of galvanic corrosion currents flowing in stainless steel/carbon couples[30]

Dense alumina is being increasingly used for the head of femural prostheses due to its high wear resistance and to its practically inert behaviour in the body. Attempts to use it as a coating of stainless steel have not yet overcome two major difficulties: brittleness of the coating, and transformation, during plasma spraying, of relatively inert α-Al_2O_3 to soluble γ-Al_2O_3. Release of aluminium ions causes concern because of the mounting evidence that it may inhibit bone formation [41], leading to various forms of osteomalacia. In terms of corrosion behaviour of the substrate Santos and Monteiro [42] report a decrease in the passive current density of 316L stainless steel, associated with partial shielding of this material by a porous Al_2O_3 coating (see Fig. 9). Introduction of an intermediate sprayed coating for improved adhesion leads to a reduction in the corrosion resistance of 316L stainless steel [26].

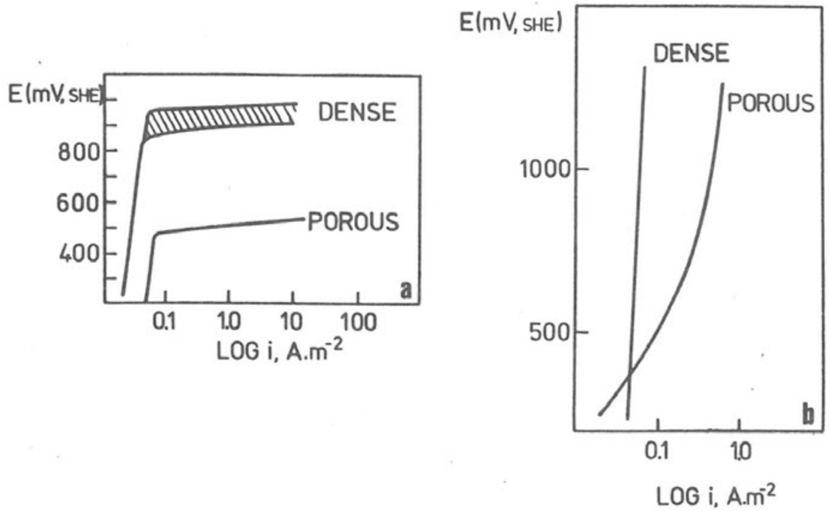

Fig. 8 - Anodic polarization curves of porous and dense (a) AISI 316L stainless steel and (b) titanium in Hank's solution at 37ºC[40]

5. Aggressive and Inhibitive Species in Physiological Media

The main ions present in the extra cellular fluid are Na^+, K^+, Ca^{2+}, Mg^{2+} and Cl^-. Of these only Cl^- is aggressive, its concentration being about 100 mM/l [32]. The role of chloride ions in localized corrosion is well known, and hence no special mention need be made about the way it influences passive film stability and corrosion kinetics. It is more interesting to consider the effect of organic species in these processes, and particularly that of proteins. In spite of the importance of the subject the number of works published is very scarse. From existing data it appears that their effect depends on wether static or fretting corrosion are taking place. They have a distinct inhibitory effect on the corrosion rate of stainless steel under fretting [33,34] but under static conditions the results are contradictory [34,35].

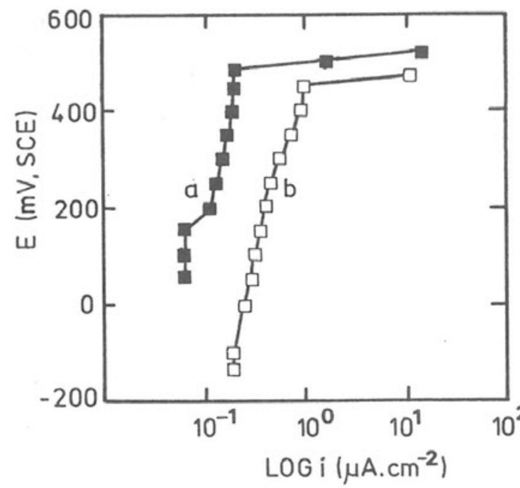

Fig. 9 - Anodic polarization curves of AISI 316L stainless steel before (b) and after (a) plasma spraying with α-Al_2O_3[42]. The solution composition is the same given in Fig. 5; room temperature

Early results of Revie and Greene [36], which show that isotonic saline solution (0.9% NaCl) is considerably more corrosive towards titanium than living tissue (rabbit and dog), could be interpreted as a sign of the favourable influence of proteins on corrosion kinetics. Their results, obtained by linear polarization measurements, are shown in Fig. 10. The complexity of the processes occurring in living tissue do not permit, however, a straightforward correlation, since many other factors may play a significant role. At present it is not legitimate to conclude that a living organism is less aggressive than current physiological test solutions, simply on the account of the above results. In fact exactly the opposite conclusions have been reached in fatigue testing of stainless steel [37].

6. Application of Electrochemical Techniques in Biocompatibility Testing

Severe restrictions exist in most european countries regarding animal experimentation. This, together with the variability of results and costs, have provided strong arguments in favour of a series of laboratory experiments aimed at predicting the behaviour in vivo. Animal experimentation will always have to be done but improvement of in vitro techniques will gradually reduce the number of such experiments.

A few attempts have been made to apply simple electrochemical techniques in predicting biocompatibility of metallic materials. The works carried out by Sawyer and coworkers [43,44] at the State University of New

York, constitute perhaps the best and most complete example of the usefulness of these techniques so far. As an example, Table V relates the tendency for thrombus formation on various metals with the corresponding corrosion potentials, E_{corr}. Thrombus deposition does not occur on those metals that have a E_{corr} higher than the equilibrium potential. Metals with a tendency to go into solution will therefore tend to be antithrombogenic.

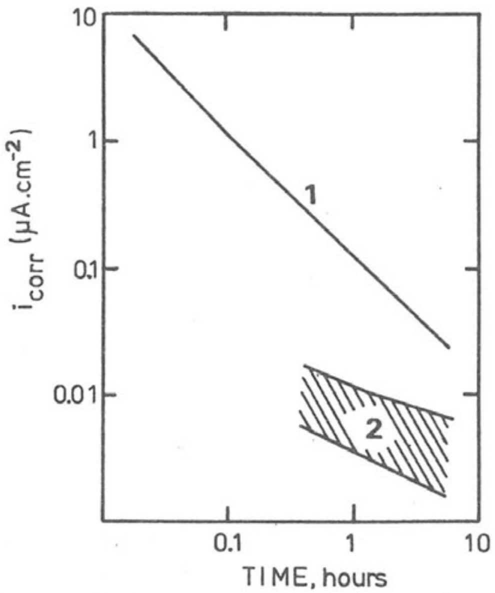

Fig. 10 - Comparison of the in vivo and in vitro corrosion rates of titanium[36]. 1 - isotonic sodium chloride; 2 - rabbit and dog

More recently Steinemann [45] has related toxicity of various materials with their polarization resistance, R_p, in the rabbit. Fig. 11 shows that toxic metals tend to have lower R_p than inert materials. Intermediate behaviour involves those materials that form a connective tissue membrane around the implants. Gold and silver are in this group. They are not as resistant to degradation in tissues as other materials, often considered less noble, like 316L stainless steel. This is probably associated with complex formation.

The polarization resistance method has also been employed by Zitter and Plenk [46], who measured R_p in a $K_4Fe(CN)_6/K_3Fe(CN)_6$ redox system. They

TABLE V - DEPENDENCE OF THROMBUS DEPOSITION ON E^0 AND E_{corr}

METAL	E^0 M/M^{n+} (V/NHE)	E_{corr} (V/NHE)	THROMBUS
Mg	-2.375	-1.360	No
Al	-1.670	-0.750	No
Cd	-0.402	-0.050	No
Cu	0.346	0.025	Yes
Ni	-0.230	0.029	Yes
Au	1.420	0.120	Yes
Pt	1.200	0.125	Yes

found this system appropriate because the free corrosion potentials were close to those measured in tissue culture fluid. Differentiation between materials behaviour was greater, however, for what they call the "straddle test". The test applies a constant potential difference of 250 mV between two identical electrodes and measures the current vs. time curves. The idea is to simulate the behaviour of materials in contact with tissues with different degrees of oxygenation. Gold, stainless steel and

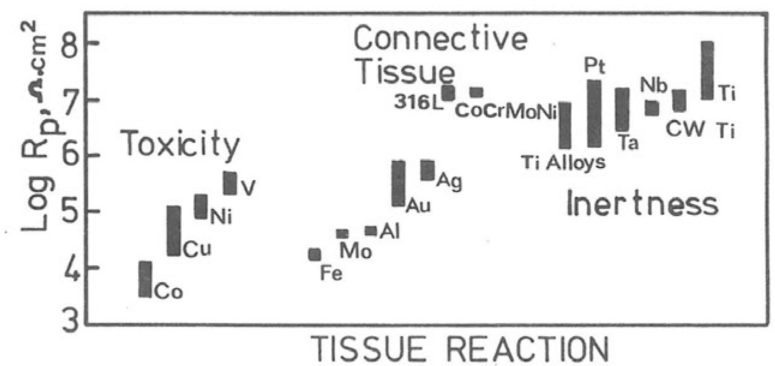

Fig. 11 - Polarization resistances grouped according to tissue reaction of metals[45]

cobalt-based alloys gave the highest currents, both in the "straddle tests" and by the polarization resistance method. Separate experiments have shown that human fibroblast growth is inhibited by gold and stainless steel and that bone-healing is disturbed by these materials. On the basis of their work they conclude that the higher the current densities the worse the biocompatibility. If further work confirms this conclusion, electrochemical methods could become invaluable in predicting the behaviour of biomaterials in vivo.

REFERENCES

1 - S.A. Barenberg, "Abridged report of the Committee to Survey the Needs and Opportúnities for the Biomaterials Industry", J. Biomed. Mater. Res., 22, 1267-1291 (1988)

2 - D.F. Williams, "Titanium and Titanium Alloys", Biocompatibility of Clinical Implant Materials (Ed. D.F. Williams), Vol. I, p. 9-44, CRC Series in Biocompatibility, 1981

3 - O.E.M. Pohler, "Degradation of Metallic Orthopaedic Implants", Biomaterials in Reconstructive Surgery (Ed. L.R. Rubin), Chap. 15, the C.V. Mosby Co., 1983, pp. 158-228

4 - D.L. Tsalev and Z.K. Zaprianov, Atomic Absorption Spectrometry in Occupational and Environmental Health Practice, Vol. I, C.R.C. Press, Boca Raton, 1983

5 - R.A. Silva, M.A. Barbosa, A.S. Costa, J.A.C. Ribeiro, "Estudo Analítico de Implantes e Próteses Ortopédicas", Proc. "Materiais 85", Vol. V, S.P.M., 1985

6 - S.D. Cook, K.A. Thomas, A.F. Harding, C.L. Collins, R.J. Haddad Jr., M. Milicic and W.L. Fischer, "The in vivo Performance of 250 Internal Fixation Devices: a Follow-Up Study", Biomaterials, 8, 177-184 (1987)

7 - S.M. Perren, V. Geret, M. Tepit and B.A. Rahn , "Quantitative Evaluation of Biocompatibility of Vanadium Free Titanium Alloys". Biological and Biomechanical Performance of Biomaterials (Eds. P. Christel, A. Meunier and A.J.C. Lee), Elsevier, 1986, pp. 397-402.

8 - E.J. Sutow, D.W. Jones and E.L. Milne, "In vitro Crevice Corrosion Behavior of Implant Materials", J. Dent. Res., 64, 842-847 (1985)

9 - K. Asami and K. Hashimoto, "An X-ray Photo-Electron Spectroscopic Study of Surface Treatments of Stainless Steels", Corros. Sci., 19, 1007-1017 (1979)

10 - M.A. Barbosa, "The Pitting Resistance of AISI 316 Stainless Steel Passivated in Diluted Nitric Acid", Corros. Sci., 23, 1293-1305 (1983)

11 - R.A. Buchanan, J.E. Lemons, C.D. Griffin, N.G. Thompson and L.C. Lucas."Effects of Carbon Coupling on the In vitro Corrosion of Cast Surgical Cobalt-Base Alloy". J. Biomed. Mat. Res., 15, 611-614 (1981)

12 - R.A. Buchanan, E.D. Rigney, Jr. and J.M. Williams, "Ion Implantation of Surgical Ti-6Al-4V for Improved Resistance to Wear-Accelerated Corrosion", J. Biom. Mat. Res., 21, 355-366 (1987)

13 - A.C. Fraker, A.W. Ruff, P. Sung, A.C. Van Orden and K.M. Speck, "Surface Preparation and Corrosion Behavior of Titanium Alloys for Surgical Implants". Titanium Alloys in Surgical Implants, ASTM STP 796 (Eds. H.A. Luckey and F. Kubli, Jr.), ASTM, 1983, pp 206-219

14 - P.A. Mausli, S.G. Steinemann and J.P. Simpson, "Properties of Surface Oxides on Titanium and some Titanium Alloys", 6th World Conference of Titanium, Cannes, June 1988

15 - C.D. Peterson, B.M. Hillberry and D.A. Heck, "Component Wear of Total Knee Prostheses Using Ti-6Al-4V, Titanium Nitride Coated Ti-6Al-4V, and Cobalt-Chromium-Molybdenium Femoral Components", J. Biom. Mater. Res., 22, 887-903 (1988)

16 - R.A. Buchanan, E.D. Rigney, Jr., and J.M. Williams,"Wear-Accelerated Corrosion of Ti-6Al-4V and Nitrogen-Ion-Implanted Ti-6Al-4V: Mechanisms and Influence of Fixed-Stress Magnitude", J. Biomed. Mater. Res., 21, 367-377 (1987)

17 - P. Sioshansi, R.W. Oliver and F.D. Mathews, "Wear Improvement of Surgical Titanium Alloys by Ion Implantation", J. Vac. Sci. Technol. A, 3, 2670-2674 (1985)

18 - M.H. Dean and U. Stimming, "The Electronic Properties of Disordered Passive Films", Corros. Sci., 29, 199-211 (1989)

19 - A.M.P. Simões, M. Cunha Belo, M.G.S. Ferreira and B. Rondot, "Characterization of Passive Films on Stainless Steel by Photoelectro-chemical and Impedance Techniques", Portugaliae Electrochimica Acta, 5, 59-69 (1987)

20 - D. Gorse, B. Rondot and M. Cunha Belo, "Influence de l'Irradiation sur les Propriétés du Film de Passivation Formé sur un Acier Inoxydable du Type 18-8", C.R. Acad. Sci. Paris, 308, 369-375 (1989)

21 - R. Silva, M.A. Barbosa, B. Rondot and M. Cunha Belo, "Impedance and Photoelectrochemical Measurements on Passive Films Formed on Some Metallic Biomaterials", submitted to Br. Corros. J.

22 - B.E. Wilde and E. Williams, "The Use of Current/Voltage Curves for The Study of Localized Corrosion and Passivity Breakdown on Stainless Steels in Chloride Media", Electrochim. Acta, 16, 1971-1985 (1971)

23 - J.R. Cahoon, R. Bandyopadhya and L. Tennese, "The Concept of Protection Potential Applied to the Corrosion of Metallic Orthopaedic Implants", J. Biomed. Mater. Res., 9, 254-264 (1975)

24 - J.R. Cahoon and C.T.F. Cheung, "The Susceptibility of Metallic Surgical Implant Materials to Crevice Corrosion", Canadian Metallurgical Quarterly, 21, 289-292 (1982)

25 - J.M. Meyer, J.M. Wirthner, R. Barraud, C.P. Susz and J.N. Nally, "Corrosion Studies on Nickel-Based Casting Alloys". Corrosion and Degradation of Implant Materials, ASTM STP 684, (Eds. B.C. Syrett and A. Acharya), ASTM, 1979, pp. 295-315

26 - M.L. Escudero, J.A. González and J. Ruiz, "In Vitro Corrosion Behaviour of Ceramic Sprayed Stainless Steel Prostheses", Br. Corros. J., 22, 182-184 (1987)

27 - A.D. Haubold, H.S. Shim and J.C. Bockros, "Carbon in Medical Devices", Biocompatibility of Clinical Implant Materials, Vol. II (Ed. D.F. Williams), C.R.C. Series in Biocompatibility, Boca Raton, 1981, pp. 3-42

28 - L.C. Lucas, R.A. Buchanan and J.E. Lemons, "Investigations on The Galvanic Corrosion of Multialloy Total Hip Prostheses", J. Biomed. Mater. Res., 15, 731-747 (1981)

29 - N.G. Thompson, R.A. Buchanan and J.E. Lemons, "In Vitro Corrosion of Ti-6Al-4V and Type 316L Stainless Steel when Galvanically Coupled with Carbon", J. Biomed. Mater. Res., 13, 35-44 (1979)

30 - R.A. Silva, M.A. Barbosa, G.M. Jenkins and I. Sutherland,"Electroche-mistry of Galvanic Couples Between Carbon and Common Metallic Biomaterials in the Presence of Crevices", submitted to Biomaterials.

31 - C.D. Griffin, R.A. Buchanan and J.E. Lemons, "In Vitro Electrochemical Corrosion Study of Coupled Surgical Implant Materials", J. Biomed. Mater. Res., 17, 489-500 (1983)

32- A.C. Guyton, Textbook of Medical Physiology (Portuguese Translation), 5th Edition, p. 39

33 - K. Merritt and S.A. Brown, "Effect of Proteins and pH on Fretting Corrosion and Metal Ion Release", J. Biomed. Mater. Res., 22, 111-120 (1988)

34 - R.L. Williams, S.A. Brown and K. Merritt, "Electrochemical Studies on the Influence of Proteins on the Corrosion of Implant Alloys", Biomaterials, 9, 181-186 (1988)

35 - S.A. Brown, L.L. Farnsworth, K. Merritt and T.D. Crowe, "In Vitro and in Vivo Mètal Ion Release", J. Biomed. Mater. Res., 22, 321-338 (1988)

36 - R.W. Revie and N.D. Greene,"Comparison of the In Vivo and In Vitro Corrosion of 18-8 Stainless Steel and Titanium", J. Biomed. Mater. Res., 3, 467-470 (1969)

37 - M. Morita, T. Sasada, H. Hayashi and Y. Tsukamoto,"The Corrosion Fatigue Properties of Surgical Implants in a Living Body", J. Biomed. Mater. Res., 22, 529-540 (1988)

38 - R.M. Pilliar, "Porous-Surfaced Metallic Implants for Orthopaedic Applications". J. Biomed. Mater. Res., 21, A1, 1-33 (1987)

39 - H.W. Wevers and T.D.V. Cooke, "Bead Loosening in Porous Metal Coated Orthopaedic Implants: a case study". Clinical Materials, 2, 67-74 (1987)

40 - P. Ducheyne, "In Vitro Corrosion Study of Porous Metal Fibre Coatings for Bone Ingrowth", Biomaterials, 4,.185-191 (1983)

41 - M.R. Christoffersen and J. Christoffersen, "The Effect of Aluminium on the Rate of Dissolution of Calcium Hydroxyapatite - A Contribution to The Understanding of Aluminium-Induced Bone Diseases", Calcif. Tissue Int., 37, 673-676 (1985)

42 - J.D.S. Santos and F.J. Monteiro,"Estudo de Resistência à Corrosão por Picadas In Vitro do Aço Inoxidável AISI 316L Revestido com Al_2O_3", Book of Abstracts, 4th Meeting of the Portuguese Electrochemical Society, Costa do Estoril, March 1989

43 - P.N. Sawyer and S. Srinivasan,"Essential Criteria for the Selection of Metallic and Non-Metallic Vascular Prostheses", Adv. Biomed. Eng. Med. Phys., 3, 273-294 (1970)

44 - M.E. Lee, T. Murakami, B. Stanczewski, A. Parmeggiani, S. Srinivasan and P.N. Sawyer, " Etiology of Thrombus Formation on Prosthetic Metal Heart Valves - The Role of Spontaneous In Vivo Interfacial Potentials".The J. of Thoracic and Cardiovasc. Surg., 63, 809-819 (1972)

45 - S.G. Steinemann,"Corrosion of Titanium and Titanium Alloys for Surgical Implants; Titanium-Science and Technology". Proc. 5th Intern. Conf. on Titanium, Vol. 2, DGF,1985, pp. 1373-1379

46 - H. Zitter and H. Plenk Jr., "The Electrochemical Behavior of Metallic Implant Materials As An Indicator of Their Biocompatibility", J. of Biomed. Mater. Res., 21, 881-896 (1987)