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# Titanium as a Biomaterial for Implants

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## 1. Introduction

An ideal biomaterial is expected to exhibit properties such as a very high biocompatibility, that is, no adverse tissue response. Also, it must have a density as low as that of bone, high mechanical strength and fatigue resistance, low elastic modulus and good wear resistance. It is very difficult to combine all these properties in only one material.

Some metals are used as biomaterials due to their excellent mechanical properties and good biocompatibility. Since the metallic bonds in these materials are essentially non-directional, the position of the metals ions can be altered without destroying the crystal structure, resulting in a plastically deformable solid. This is also an advantage when thinking about the device manufacture technology.

The principal disadvantage of metals is its corrosion tendency in an in-vivo environment. Most metals can only be tolerated by the human body in small amounts even as metallic ions. The consequences of corrosion are the disintegration of the material implant, which will weaken the implant and the harmful effect of corrosion products on the surrounding tissues and organs.

Some metals are used as passive substitutes for hard tissue replacement such as total hip and knee joints, for fracture healing aids as bone plates and screws, spinal fixation devices and dental implants. Some metallic alloys are used for more active roles, as actuators such as vascular stents, and orthodontic archwires.

Metallic biomaterials can be conveniently grouped in the following categories:

- Stainless steel
- Cobalt base alloys
- Titanium base alloys
- Specialty metallic alloys

Examples of ASTM standards for some of these metallic biomaterials are shown in Table 1.

The first metal alloy developed specifically for human use was the “vanadium steel” but it was no longer used in implants because its corrosion resistance is inadequate *in vivo*. Later in the 1950s, 18-8sMo with very low carbon content (known as 316L) stainless steel was introduced and is actually widely used for implant fabrication. This alloy has a very good resistance to chloride solutions and poor sensitization.

The castable CoCrMo alloy has been used for many decades in dentistry and, relatively recently, in making artificial joints. The wrought CoNiCrMo alloy is relatively new, now used for making the stems of prostheses for heavily loaded joints such as the knee and hip. Both alloys have excellent corrosion resistance.

Attempts to use titanium for implant fabrication dates to the late 1930s. It was found that titanium was tolerated as was stainless steels and cobalt alloys. Titanium's lightness and good mechano-chemical properties are salient features for implant application.
Titanium was found the only metal biomaterial to osseointegrate (Van Noort, 1987). Also, there were even assumptions on a possible bioactive behaviour (Li et al., 1994) due to the slow growth of hydrated titanium oxide on the surface of the titanium implant that leads to the incorporation of calcium and phosphorous.

| Material designation | Common name | ASTM Standard |
|---|---|---|
| *Stainless Steel*<br>Fe-18Cr-14Ni-2.5Mo | 316L Stainless Steel | ASTM F 138 |
| *Cobalt base alloys*<br>Co-28Cr-6Mo<br>Co-35Ni-20Cr-10Mo | Cast CoCrMo<br>Wrought CoNiCrMo | ASTM F 75<br>ASTM F 572 |
| *Titanium base alloys*<br>Ti CP (grade 1 to 4)<br>Ti-6Al-4V ELI | Commercially pure Ti<br>Ti6Al4V | ASTM F 67<br>ASTM F 136 |
| *Specialty metallic alloys*<br>Ta<br>Ni-45Ti | Unalloyed Tantalum<br>Nitinol | ASTM F 560<br>ASTM F 2063 |

Table 1. Examples of metallic biomaterials (An introduction to biomaterials , 2006)

## 2. Titanium base alloys developments

### 2.1 Commercially pure titanium

Commercially pure titanium (Ti CP) and extra low interstitial Ti-6Al-4V (ELI) are the two most common titanium base implant biomaterials. These materials are classified as biologically inert biomaterials. As such, they remain essentially unchanged when implanted into human bodies. The human body is able to recognize these materials as foreign and tries to isolate them by encasing it in fibrous tissues. However, they do not promote any adverse reactions and are tolerated well by the human tissues.
These metals do not induce allergic reactions such as has been observed with some stainless steels, which have induced nickel hypersensitivity in surrounding tissues.
Titanium is very light with a density of 4.5 g/cm$^3$. The Ti CP is 98.9 - 99.6 % Ti, being the oxygen content (and other interstitial elements such as C and N) the main element influencing significantly its yield, tensile and fatigue strengths (Table 2). Interstitial elements strengthen the metal through interstitial solid solution strengthening mechanism, with nitrogen having approximately twice the hardening effect (per atom) of either carbon or oxygen.
Pure Ti is an allotropic metal having hexagonal α–phase (HCP) below 882 $^o$C and transforming to a cubic β–phase (BCC) over that temperature. As its typical microstructure is a single alpha phase, cold work is also an applied strengthening mechanism.
Its very good biocompatibility is due the formation of an oxide film ($TiO_2$) over its surface. This oxide is a strong and stable layer that grows spontaneously in contact with air and prevents the diffusion of the oxygen from the environment providing corrosion resistance.

It is a biomaterial with a high superficial energy and after implantation it provides a favourable body reaction that leads to direct apposition of minerals on the bone-titanium interface and titanium osseointegration (Acero et al., 1999).

| Element | Grade 1 | Grade 2 | Grade 3 | Grade 4 | Ti6Al4V[a] |
|---|---|---|---|---|---|
| N máx | 0.03 | 0.03 | 0.05 | 0.05 | 0,05 |
| C máx | 0.10 | 0.10 | 0.10 | 0.10 | 0,08 |
| H máx | 0.015 | 0.015 | 0.015 | 0.015 | 0,0125 |
| Fe máx | 0.20 | 0.30 | 0.30 | 0.50 | 0,25 |
| O máx | 0.18 | 0.25 | 0.35 | 0.40 | 0,13 |
| Ti | Balance | Balance | Balance | Balance | |

[a] Aluminium 6%, Vanadium 4%

Table 2. Chemical composition of Ti CP (ASTM F 67) and Ti6Al4V alloy (ASTM F 136)

### 2.2 Ti6Al4V alloy

Ti6Al4V alloy is widely used to manufacture implants and its chemical composition is given in Table 2. The addition of alloying elements to titanium enables it to have a wide range of properties because aluminium tends to stabilize the α-phase and vanadium tends to stabilize the β-phase, lowering the temperature of the transformation from α to β. The alpha phase promotes good weldability, excellent strength characteristics and oxidation resistance. The addition of controlled amounts of vanadium as a β-stabilizer causes the higher strength of beta-phase to persist below the transformation temperature which results in a two-phase system. The β-phase can precipitate by an ageing heat treatment. This microstructure produce local strain fields capable of absorbing deformation energy. Cracks are arrested or deterred at the particles. The mechanical properties of the Ti CP and the Ti6Al4V are given in Table 3.

| Property | Grade 1 | Grade 2 | Grade 3 | Grade 4 | Ti6Al4V |
|---|---|---|---|---|---|
| Tensile strength (MPa) | 240 | 345 | 450 | 550 | 860 |
| Yield strength (MPa) | 170 | 275 | 380 | 485 | 795 |
| Elongation (%) | 24 | 20 | 18 | 15 | 10 |

Table 3. Mechanical properties of Ti CP (ASTM F 67) and Ti6Al4V alloy (ASTM F 136)

The modulus of elasticity of these materials is about 110 GPa. This is much lower than stainless steels and Co-base alloys modulus (210 and 240 GPa, respectively (Dadvinson & Gergette, 1986). When compared by specific strength (strength/density) the titanium alloys exceed any other implant materials.

Titanium and titanium alloys, nevertheless, have poor shear strength, making them less desirable for bone screws, plates and similar applications. They also tend to gall or seize in sliding contact with itself or another metal.

### 2.3 Low modulus titanium alloys

The Ti6Al4V alloy has some disadvantages: its elastic modulus, although low, is 4 to 6 times that of cortical bone and has low wear resistance that is a problem in articulations surfaces.

Also, V can cause potential citotoxicity and adverse tissue reactions (Steinemann, 1980), and Al ions from the alloy might cause long-term Alzheimer diseases (Rao et al., 1996).
Briefly, a biocompatible titanium base alloy suitable for bone implant should meet at least the following requirements (Mehta, 2008):

- Potentially toxic elements, such as vanadium, cooper and tin, should be avoided completely
- Elements that may have potential toxicological problems, such as chromium, nickel and aluminium, should be used only in minimum, acceptable amounts
- The alloy should have high corrosion resistance
- The alloy should have, at least, the following desirable mechanical properties: low modulus, high strength and good smooth and notched fatigue strength
- The alloy should have good workability and ductility.

Consequently, the recent trend in research and development of titanium for biomedical applications is to develop alloys composed of non-toxic and non-allergenic elements with excellent mechanical properties (low modulus-high strength) and workability (Niinomi, 1999).
The first generation of design orthopaedic alloys try to replace the V and Al alloys with other non toxic components such as Nb, Fe and Mo (for the V) and Ta, Hf and Zr (for the Al).
Subsequent developments in orthopaedic Ti-base alloys have been motivated by the requirement of low elastic modulus. The stiffness of titanium and its alloys is still largely greater than that of cortical bone, although it is less than that of Co-Cr type alloys and stainless steels used for biomedical applications. This difference of rigidity produced the stress-shielding phenomenon. Stress shielding occurs because of the mismatch between the stiffness of the bone, which has a Young modulus of 7–25 GPa (Currey, 1998), and that of the metal implant stem.
Various methods of solving this problem have been considered, including changing the size and shape of the stem to reduce the differences in the structural stiffness of the implant and the surrounding bone and changing the implant material from steel to commercially pure titanium or Ti alloys with low modulus (Sarmiento et al., 1979).
Metastable β-Ti alloys were developed for this purpose, with low elastic modulus. In Table 4 it could be observed some old and new Ti-base alloys develop specifically for biomedical purpose (Guilemot et al., 2004).
Low modulus alloys are nowadays desired because the moduli of alloys are required to be much more similar to that of bone. These new alloys have an elastic modulus ranging 55-85 GPa, so it could be minimized the stress shielding phenomena because it is more proximally to the bone modulus. However, they are still greater than that of cortical bone.

| | |
|---|---|
| 1. Titanium CP (ASTM F 67) | 10. **Ti-12Mo-6Zr-2Fe** (ASTM F1813- low modulus) |
| 2. Ti-6Al-4V ELI (ASTM F 136) | 11. **Ti-15Mo** (low modulus) |
| 3. Ti-6Al-4V (ASTM F 1108) | 12. Ti-16Nb-10Hf (low modulus) |
| 4. **Ti-6Al-7Nb** (ASTM F 1295) | 13. Ti-15Mo-5Zr-3Al (low modulus) |
| 5. Ti-5Al-2.5Fe (ISO 5832) | 14. Ti-15Mo-2.8Nb-0.2Si-0.26O (low modulus) |
| 6. Ti-5Al-3Mo-4Zr | 15. Ti-35Nb-7Zr-5Ta (low modulus) |
| 7. Ti-15Sn-4Nb-2Ta-0.2Pd | 16. Ti-29Nb-13Ta-4.6Zr (low modulus) |
| 8. Ti-15Zr-4Nb-2Ta-0.2Pd | |
| 9. **Ti-13Nb-13Zr** (low modulus) | |

Table 4. Ti alloys developed for biomedical applications (in bold) and for other uses

## 3. Problems and solutions

Bulk titanium alloys used in implants present three main problems:
- High cost because the amount of processing energy and melting and casting difficulties
- Higher elastic modulus compared to bone
- Although the inert behaviour of Ti is a good property, its bone attachment is difficult because it do not react with the human tissues

### 3.1 The processing

A great problem of these new alloys is its fabrication processes because most beta titanium alloys contain considerable amounts of refractory elements with high melting temperatures. This results in heavily weight, difficult melting and solidification processing, low plastic deformability and high materials costs.

The various refractory materials employed in casting are attacked by titanium with such severity that sounds castings, possessing good mechanical properties are difficult to obtain. So, conventional methods are not practical with titanium.

The molten metal and the hot casting are susceptible to atmospheric contamination. Because Ti is very reactive with oxygen and other atmospheric gases, the melting and casting processes implies high temperature fusion and casting under vacuum or protective neutral atmospheres. Another casting problem is the maintenance of good flow over severe changes of dimensions or direction within the mold.

Powder metallurgy (P/M) is an alternative method of fabrication in which metal powders are utilized by compacting and sintering to form useful products. This method is employed primarily to produce simple shapes with good dimensional stability, to form shapes with material of extremely high melting temperatures and to produce parts not feasible by other means.

Production of cast titanium today takes 16 times more energy per tonne than the production of steel. Instead of conventional melting, milling and machining, P/M techniques implies powders that remain in solid form during the entire procedure. This saves a tremendous amount of processing energy with a reduction of over 50% (Mehta, 2008).

### 3.2 The elastic modulus

As was said, the elastic moduli and strength of titanium and its alloys are much higher than those of human bones, which may result in stress shielding and the failure of implants. People have tried to develop new types of titanium alloys, such as β-Ti alloys, to reduce the modulus of the implants to the level approaching human bones. On the other hand, the mechanical properties of porous titanium can be adjusted by pore fraction and morphology, and the stress shielding effect will be reduced. Porous titanium with porosity in a wide range can be prepared with powder metallurgy methods (Zhiguang et al., 2009), from which other kinds of powders as second phase in green bodies would be removed during subsequent heat treatment.

### 3.3 The inert behaviour

Despite the great progress achieved in orthopaedic biomaterials, fixation of implants to the bone host remains a problem. As titanium has an inert behaviour, the body tries to encapsulate the Ti-based implant. However, titanium does not bond directly to bone resulting in micro-movements and, eventually loosening of the implant. Undesirable

movements at the implant-tissue interface results in failure cracks of the implant. As osseointegration starts with the cellular stage and continues with the nucleation of mineral and the structuring of the new vital bone, the overall required time is varying in a broad range. A proposed solution for a better control of osseointegration is the bioactive fixation.
One approach to improving implant lifetime is to coat the metal surface with a bioactive material that can promote the formation and adhesion of hydroxyapatite ($Ca_{10}$ $(PO_4)_6(OH)_2$), the inorganic component of natural bone.
The application of bioactive coatings to titanium-based alloys enhance the adhesion of Ti-based implants to the existing bone, resulting in significantly better implant lifetimes than can be achieved with materials in use today (Katti, 2004). Typically, several silicates glasses or hydroxyapatite (HA) are used as bioactive coatings. Some properties of hydroxyapatite are shown in Table 5.

| | |
|---|---|
| Hardness (Mohs) | 5 |
| Density ($g/cm^3$) | 3.1 |
| Elastic Modulus (GPa) | 100 |
| Ultimate Tension Stress (MPa) | 100 |
| Compression Stress (MPa) | > 50 (good) |
| Toughness $K_{IC}$ (MPa $m^{1/2}$) | 1 |
| Solubility | It has the less solubility in body fluids media, so it is impossible to have $Ca^{2+}$ or $PO_4^{3+}$ ions in water (PH=7) |

Table 5. Hydroxyapatite properties

An ideal bioactive coating would bond tightly both to the bone and the metal. Two problems arise when attempting to coat metals with ceramics. For one side, the thermal expansion coefficients of the ceramic and metal are usually different, and as a result, large thermal stresses are generated during processing. These stresses lead to cracks at the interface and compromise coating adhesion. In addition, chemical reactions between the ceramic and metal can weaken the metal in the vicinity of the interface, reducing the strength of the coated system. This problem is particularly important when coating Ti alloys, due to their high reactivity with most oxide materials. However, bioactive ceramics coatings on Ti implants further improves the biocompatibility of these implants.

## 4. Titanium-Hydroxyapatite composite

Biocomposite materials have been developed in order to combine bioactivity of ceramics and mechanical properties of metals. Hydroxyapatite (HA) is known for its weakness and brittles (see Table 5) but has an excellent biocompatibility and is a bioactive material. When HA is added to titanium, an improvement of the biomaterial chemical properties occurs. New developments try to aggregate hydroxyapatite as a second phase to the Ti alloy, with powder metallurgy techniques (P/M). In this composite material, particles of HA are incorporated in a porous titanium matrix providing points of good bone reaction. These solutions allow improved adhesion strength of the load bearing metallic component to the bone, resulting in shorter healing periods as well as predictable behaviour of the implant for longer periods of time.

Although this benefits, there are some problems in the manufacture of the composite material and some doubts about its biocompatibility.
One processing problem is that, as Ti is stable in vacuum or reducing atmospheres and HA is stable only in oxidizing atmospheres (Weng et al., 1994), sintering of this type of composites is difficult. Also, there are some reports indicating intense HA decomposition at temperatures lower than the decomposition temperatures of the monolithic powders, due the interaction with the Ti powder (Yang et al., 2004), which declines its bioactivity and mechanical properties According to literature the Ca titanates $CaTiO_3$ and $CaTi_2O_5$ are formed through reactions between HA and $TiO_2$ in vacuum, both when the titanium oxide was intentionally added or when it resulted from the oxidation of metallic Ti in Ti-HA composites (Yang et al., 2004).
Another problem of porous metals is its fatigue behaviour. The porosity of most implants is usually determined to compromise between maintaining the mechanical strength of the implant while still providing adequate pore size for tissue ingrowth. Although optimum pore size required for implant fixation remains undefined, the consensus is that in order to optimize mineralised bone ingrowth, pore sizes between 100 and 400 μm are necessary (Cameron et al., 1976). A major concern with the use of porous implants in highly loaded applications is the effect the porous matrix might have on fatigue strength. Ti alloys experienced drastic reductions in fatigue strength till one-third that of the solid alloy equivalent shape (Wolfrth & Ducheyne, 1994). Stress intensification due to these pores are major sources of weakness in the fatigue strength. This is sometimes referred to as the nocth effect. To achieve a functionally strong implant, porous implant design needs to account for these losses in metal strength.
Nevertheless titanium-hydroxyapatite porous structures are promising biomaterials to be used as replacement implants.

## 5. Finite element analysis

As an artificial hip joints need to be designed to withstand the loads that they are expected to bear without fracture or fatigue, stress analysis is therefore required to ensure that all components of the device operate below the fatigue limit. For simple calculations, simple analytical calculations usually suffice. Unfortunately, analytical solutions are limited to linear problems and simple geometries governed by simple boundary conditions.
Implants as a hip joint involve some combinations of material or geometry non-linearity, complex geometry and mixed boundary conditions. Applying analytical methods to such a problem would require so many assumptions and simplifications. An alternative is the use of approximate or numerical methods. The most popular numerical method for solving problems in continuum mechanics is the finite element method (FEM), also referred to as finite element analysis (FEA).
FEA uses a complex system of points called nodes which make a grid called a mesh. The complex structure is divided into a large number of smaller parts, or elements, with interconnecting nodes, each with geometry much simpler than that of the whole structure. This mesh is programmed to contain the material and structural properties which define how the structure will react to certain loading conditions. Nodes are assigned at a certain density throughout the material depending on the anticipated stress levels of a particular area. Regions expected to receive large amounts of stress usually will have a higher node density than those which experience little or no stress.

The behaviour of the unknown variable within the element and the shape of the element are represented by simple functions that are linked by parameters that are shared between the elements at the nodes. Using boundary conditions, a large system of equations results and they are solved simultaneously using interactive means.
The essential steps in the FEM follow:

i. Discretization of the region of interest by the subdivision of the region (continuum) in small elements,
ii. Definition of the unknown variables (stress, strain and displacement) in each element within the continuum,
iii. Formulation in each element of the equations that define the behaviour of the unknown variables (stress-strain or strain-displacement relationships) in the form of matrices. These element matrices are assembled into a global system of equations for the entire discretized domain,
iv. Solving of the global system of equations by interactive means.

Following are presented some examples of the application of FEM in the estimation of fatigue life in a hip implant made in different materials. The materials implant analysed were: 316L stainless steel, Ti6Al4V, Ti-35Nb-7Zr-5Ta (low modulus β-Ti alloy) and sintered porous Ti.

### 5.1 Mechanical behaviour of hip implant

Currently, total hip arthroplasty (THA) is a common technique used in cases of reconstruction when the functionality of the natural hip joint and the leg is impaired. Despite great progress in biomaterials, fixation of the prosthesis to the bone remains a problem because the commercial metallic THA implants are five to six times stiffer than bone. The difference in elastic modulus between the bone and the implant material has been identified as the major cause of implant loosening from stress shielding.
The regenerative and remodelling processes in bone are directly triggered by loading, i.e. bone subjected to loading or stress regenerates and bone not subjected to loading results in atrophy. Thus, the effect of a much stiffer bone implant is to reduce the loading on bone resulting in the phenomenon called as stress shielding (Katti, 2004).
A stem of a lower stiffness material (e.g. a titanium alloy) will transfer more of the load to the femur proximally, reducing stress shielding, however, this is achieved at the expense of higher load transfer stresses at the cement interfaces with the bone and implant and the risk of cement failure (Gross & Abel, 2001). A goal to reach would be a low stress shielding and low interface shear stresses in this type of implant, but nowadays there are no means to reach that goal, so, the existing designs are based in a compromise between them.
There are many issues related to implants and prostheses, but in general the magnitude and direction of the load change and are not accurately known, and are patient dependant in any case. The average load on a hip joint is estimated to be up to three times body weight and the peak load during other activities such as jumping can be as high as 10 times body weight. Besides, hip joints may undertake cyclic loading as high as $10^6$ cycles annually. This led the setting of different standards for testing mechanical strength.

### 5.2 Simulation of the mechanical behaviour of uncemented femoral stem of a hip prosthesis

For the present study we considered two conditions. The first of them corresponds to the layout and loads used in the fatigue test established in the Standard IRAM 9422-3

*Determination of fatigue properties of the femoral stem components without the application of torsion*. The second condition corresponds to the implant attached to bone under the action of an arbitrary load. This last condition analysed the distribution of stresses and strains that produces the implant on the femur.
The proposed stem design was thought to minimise stress shielding.

### 5.2.1 Methods

Three materials were considered in the simulation: a low carbon stainless steel 316L in semi hard condition, a typical Ti-6Al-4V titanium alloy and a low modulus beta-Ti alloy, Ti-35Nb-7Zr-5Ta. The mechanical properties of these materials are given in Table 6.
For the stem support for fatigue testing, the material of the support was assumed to have an elastic modulus E of 2.7 GPa and a Poisson ratio of 0.3.
To characterize the mechanical behaviour of the bone, all materials were considered as isotropic. The cortical and cancellous bones were assumed to have an elastic modulus of 16,200 MPa and 380 MPa, respectively and to have a Poisson ratio of 0.3

### 5.2.2 Finite element models

For fatigue testing, we performed three finite element models in which, in each case, it was changed the material properties of the implant. The models accounted for stem, the test stand and also a piece to apply the load. Full 3D model were considered, with solid tetrahedral and hexahedral elements.

| Material | Young's modulus (GPa) | Tensile strength (MPa) | Yield strength (Mpa) | Poisson ratio |
|---|---|---|---|---|
| 316L | 196 | 861 | 620 | 0.3 |
| Ti6Al4V | 115 | 860 | 795 | 0.33 |
| Ti35Nb7Zr5Ta | 55 | 596 | 547 | 0.33 |

Table 6. Mechanical properties of implant materials considered in the simulation

Because the stem consists of two parts and a fastener, frictional contact was modelled at the interface. The remaining interactions were assumed as tied. A total of 106,195 elements and 26,192 nodes were used in the analyses. The meshed finite element model is shown in Figure 1.
For implant attached to bone finite elements models were realized, in which the hip implant and the femur were represented. Four nodes solid elements were used in the models, to realize 4 finite elements models. Three of them were developed with the implant and an additional model was analyzed without the implant. This was considered as a control solution for evaluation of stress shielding.
One of the meshed finite element model is shown in Figure 2.
For those models with implant, the implant was completely fastened to the bone through an interaction in which "slave" nodes are tied to the master surface of the bone. So the degrees of freedom in the exterior side of the implant associates to the degrees of freedom of the bone surface in contact to it.

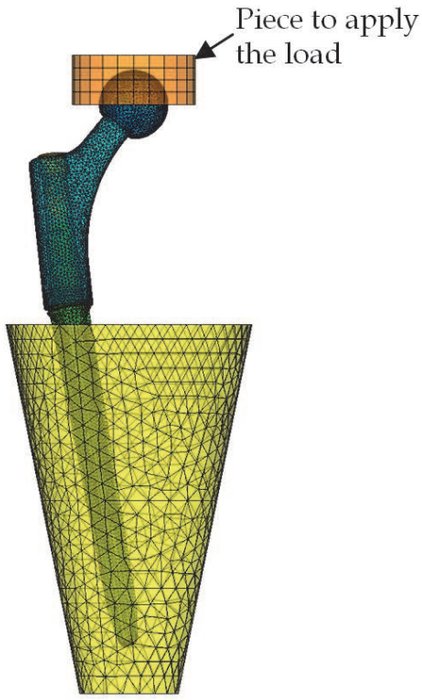


Fig. 1. Fatigue testing. The 3D finite element model developed in the analysis

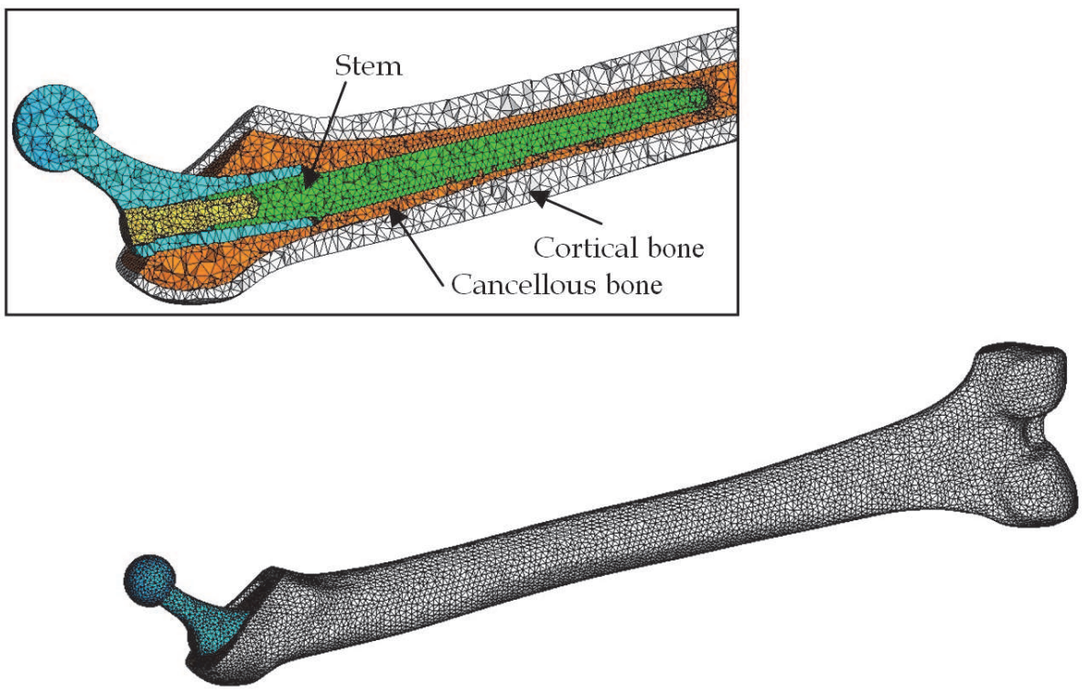


Fig. 2. Implant attached to bone. Finite element model developed in the analysis

All the models were created, analyzed and afterwards the results were processed using Abaqus 6.4.5.

### 5.2.3 Loads

In all cases for fatigue testing, the applied load was the load specified in the standard, of 2300 N. The load was applied on the upper surface of the piece that applys the load.
For implant attached to bone the applied load was an arbitrary load of 1000 N in the acetabular component of the prosthesis. Besides the pre-tensioning bolt load was simulated.

### 5.2.4 Results

**a. Fatigue testing:**

Due to the assembly load between the stem components, high levels of compressive contact stress were obtained. Therefore, and to wean the study of the influence of these stresses, we considered in the analysis the stress components S33, which would be responsible for a possible material fatigue. The simulation results obtained for the three models using different materials are given in Table 7.
Figure 3 shows the S33 stress distribution in the implant made of beta titanium alloy.

| Material | Implant maximum flexural stress S33 (MPa) | Vertical displacement in the head of the implant (mm) |
|---|---|---|
| 316L | +379<br>-602 (*) | - 0.57 |
| Ti6Al4V | +397<br>-624 (*) | - 0.94 |
| Ti35Nb7Zr5Ta | +455<br>-580 (*) | - 1.87 |

(*) The indicated compression component was mainly due to the contact stresses.

Table 7. Simulation results for the three models

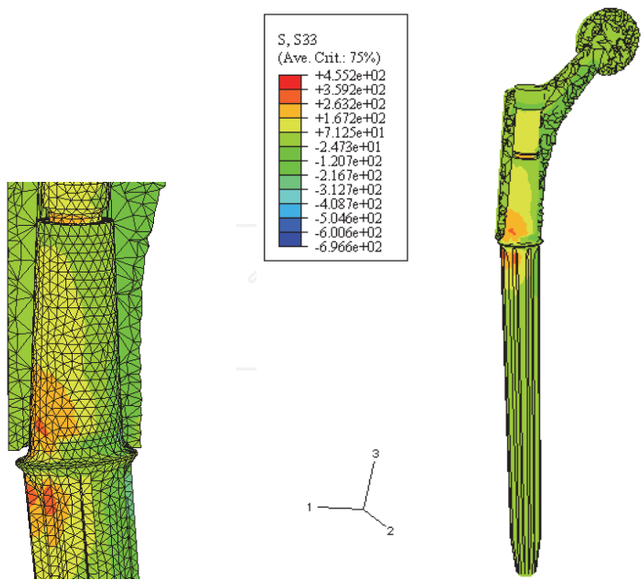


Fig. 3. S33 stress distribution (MPa)

Whereas the applied load varies between zero and 2300 N, the minimum stresses are related to those produced by the preload of the bolt connecting both parts of the implant, and the maximum stresses are due to the load of 2300 N.

Using the mechanical properties of the materials, were drawn Goodman diagrams and plotted the points corresponding to the applied loading (Figures 4 and 5). From them, it could be seen that the implants have unlimited duration of life with the geometry, loads and properties considered. However, it can be seen in the diagram for the stainless steel implant that the point representing working conditions is located near the boundary of the safe zone.

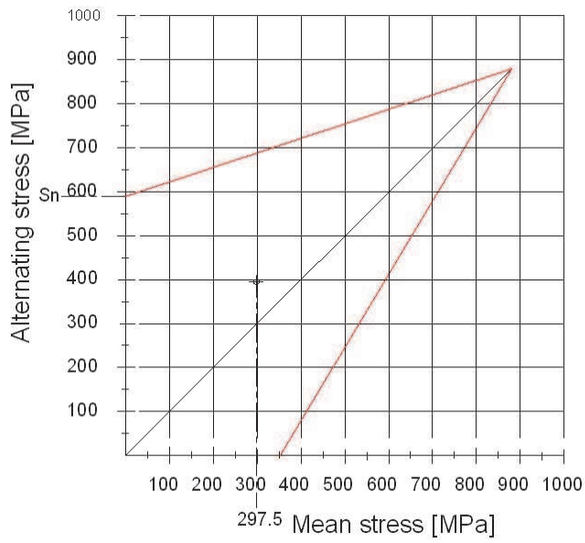


Fig. 4. Goodman's diagram. Ti6Al4V

It should be noted that the environment in which the implant should work would influence the final fatigue behaviour.

**b. Implant fixed to bone:**

In Figure 6 the S33 stress is shown for the bone, from the analysis of the stemmed Ti35Nb7Zr5Ta implant. Comparing whit the model without the implant, in the stemmed ones it was observed a change in the stress pattern, because the stem shielded the bone from

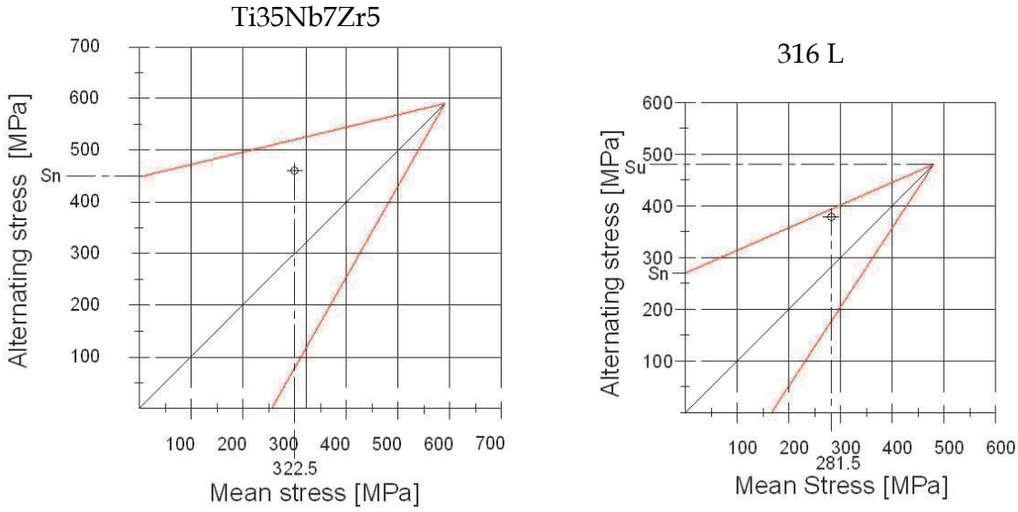


Fig. 5. Goodman's diagram. 316L and Ti35Nb7Zr5Ta

the loads. Comparing the stemmed models, from the point of view of stress and displacements that were produced in the bone, with the titanium Ti35Nb7Zr5Ta implant, the behaviour of the whole bone-implant is nearer to the natural bone, and the effect of stress shielding was the smaller.

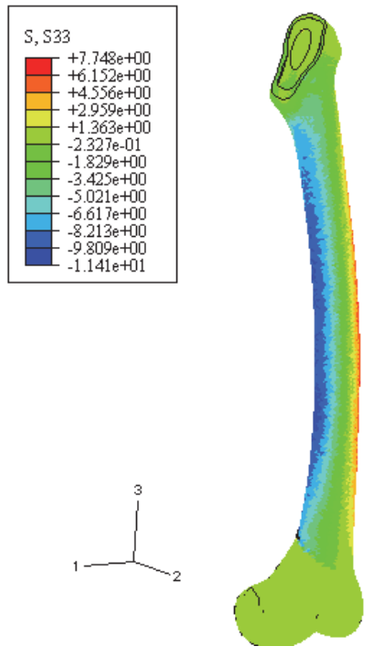


Fig. 6. S33 Stress in the bone, stemmed Ti35Nb7Zr5Ta implant model

### 5.3 Fully porous material in hip implants

It is well known that porosity decreases the Young's modulus of a material, thus it could be thought as a means to reduce stress shielding. It is difficult to get the properties for FEA analysis but it could be used some approach to represent the Young's modulus of a material with a given fraction of porosity. However, the effect the porosity on fatigue strength discourage using fully porous material in joint arthroplasty implants because porous metal alone does not provide sufficient mechanical strength to sustain the physiological loads (Ryan et al., 2006).

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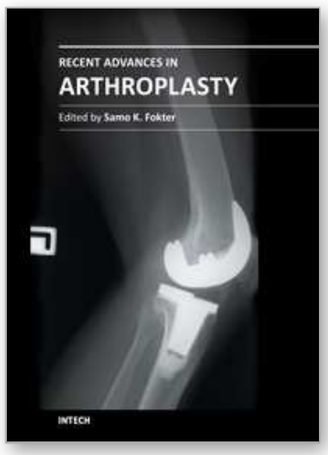


**Recent Advances in Arthroplasty**

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The purpose of this book was to offer an overview of recent insights into the current state of arthroplasty. The tremendous long term success of Sir Charnley's total hip arthroplasty has encouraged many researchers to treat pain, improve function and create solutions for higher quality of life. Indeed and as described in a special chapter of this book, arthroplasty is an emerging field in the joints of upper extremity and spine. However, there are inborn complications in any foreign design brought to the human body. First, in the chapter on infections we endeavor to provide a comprehensive, up-to-date analysis and description of the management of this difficult problem. Second, the immune system is faced with a strange material coming in huge amounts of micro-particles from the tribology code. Therefore, great attention to the problem of aseptic loosening has been addressed in special chapters on loosening and on materials currently available for arthroplasty.



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