INVESTIGATION ON SUITABLE IMPLANTS FOR BIO-MEDICAL APPLICATIONS

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Abstract : **Implants for artificial hip replacement on patients with fractured or damaged acetabulam or femur bones are to be replaced with high wear resistance and bio-compatible materials. It should not affect the body tissues and fluids also the implants should be monitored continuously. High grade alumina alloys do offer a long life span and alloys of cross-chain linked polyethylene and structurally reliable ceramics too offer good life span and wear resistance. Implants are then tested by various testing methods like scanning electron microscope (SEM), simulating body fluids (SBF), Disc diffusion methods. Hopefully, these materials do offer a better resistance and life span for the implants. Rather than developing different structures with different materials, it is always preferred to replace the entire hip structure with homogenous alloys.**

IndexTerms - **Component,formatting,style,styling,insert.**

1. INTRODUCTION

Total Hip Replacement (THR) is one of the most common operations performed in the world today. An increasingly ageing population means that the number of people undergoing this operation is set to rise. The hip is a ball and socket joint formed by the head of the thigh bone (the femur) and a section of the pelvis called the "acetabulum". The stability of this structure is obtained by the bony configuration combined with a complex system of muscles and ligaments around the joint. The hip can wear out at different points during a person's life. Osteoarthritis is one of the most widespread causes for joint degradation, in middle-aged people hypertrophic changes in the bone and cartilage of joints can be as prejudicial to require surgery. A total hip replacement aims to provide the patient with a joint that functions as normally as possible, is resistant to dislocation, preserves as much bone as possible and which will last as long as possible. Advances in surgical techniques and manufacturing technologies allow increasing success in these procedures.

A biomaterial is a material that interacts with human tissue and body fluids to treat, improve, or replace anatomical elements of the human body. Biomaterials that are used in medical devices for orthopaedics application are commonly called implants; the main characteristics of these biomaterials are summarized. These are manufactured for a great number of orthopaedic applications. Clinical results in orthopaedics have demonstrated that a great need exists to find new and better biomaterials that would help to satisfy the minimum requirements for orthopaedic devices to perform correctly on a long-term basis.

The current life expectancy for a successfully implanted hip joint can nowadays be approximately estimated to be 15 years when a "standard" articulation includes a conventional polyethylene liner. However, with the most recent developed alternatives of low-wear bearings, one could expect (or, atleast, assertively hope for) achievement of longer lifetimes, owing to a significant decrease in the volume of wear (e.g., in the case of advanced ceramic-on-ceramic heads or ceramic composite heads sliding against ultra-high molecular weight polyethylene liners with engineered microstructures).

The burden to the lymphatic system could be reduced and the life expectancy for an artificial hip joint could be significantly improved as a result. Nevertheless, deeper scientific understanding and continuous technological monitoring need to be unremittingly pursued and updated, at least for a full establishment of biomaterials in vitro testing (cf. papers by Baikal and Kurtz and by Okazaki in this special issue) and the clinical procedures involved with the newly developed implants. Ceramic-on-ceramic bearings are quite innovative, but extremely sensitive to cup position and edge loading, for which reasons they could undergo significant wear degradation and, although in quite rare instances, fracture. Ceramic heads on highly cross-linked polyethylene acetabular cups might appear more forgiving than hard-on-hard bearings with regard to the degree of precision required in the surgical technique.

However, neck impingement arising from creep driven joint laxity and/or initial component malpositioning might increase the actual wear rate on the soft polyethylene side for these couples as well (i.e., although without emission of any detectable noise). Moreover, even in advanced polymer components, high contact stresses developed in very thin liners upon neck impingement might ultimately lead to component fracture. (Blumenfeld et al., 2011; D'Antonio et al., 2005; Duffy et al., 2009; Moore et al., 2008; Tower et al., 2007).

2. EVOLUTION THROUGH SEVERAL GENERATIONS OF HIP-JOINT MATERIALS

2.1. MAXIMIZING THE STRUCTURAL PROPERTIES OF BULK CERAMIC COMPONENTS

Nowadays the most used ceramic material in biomedical applications is alumina, which possesses hexagonal (corundum-type) crystallographic structure(α-Al_2O_3 phase). Its strong ionic bond and oxygen-rich stoichiometry make alumina chemically bioinert

and, thus, exceptionally stable in the human body, leaving its structure conspicuously unaffected by corrosive processes .Such stability is the key property in impeding Al-ion release from the bulk material and from wear debris. Bulk alumina ceramics are quite strong under a compressive load ,but they possess limited tensile strength being also quite brittle. These poor structural characteristics arise both from alack of any microscopic plastic deformation at the on set of fracture(at room temperature) and from the conspicuous absence of toughening mechanisms operating during crack propagation. [1]
Nevertheless, the fabrication of dense alumina bodies with a grainsize in the order of the single micrometer has indeed be come possible through many years of technological improvements owing to two important findings both related to the sintering process:

(i) The addition of a small fraction of magnesia (MgO) to the raw alumina powder. Such addition enhances mass transport during solid-states in tering ,thus allowing the ceramic body to reach full density at relatively low sintering temperatures, which in turn enables controlling grain growth during sintering(i.e., with a beneficial effect on strength).

(ii)The use of hot isostatic pressing (HI Ping) at some step of the production process. Such improved processing procedure provides the essential driving force for closing internal pores and pushes the density of the poly crystal close to its theoretical value at relatively low sintering temperatures and ,thus, with a fine grain size (i.e again with beneficial effects on strength).

Biomedical grades of polycrystalline alumina produced in the period between1965and1977, which represents their first pioneering applications in joint arthroplasty and can be classified as
"first generation" alumina materials, possessed an average grain size of ≈ 4.5μm. After 1977, additional changes in the manufacturing process were made, which led to the so-called "second generation" alumina grades. These latter materials possessed an average grain size that could be successfully lowered to 3.4μm. Modern alumina materials ("third generation") are supplied with a grain size conspicuously more homogeneous and closer to the single micrometer as compared to the previous generation. Grain size distributions as measured in biomedical alumina grades manufactured before 1977 and after 1995. It should be noted that, by far more important than the average value of grain size, it is the right foot of the grain size distribution (i.e., the grain population with the largest sizes observed in the micro structural texture) that causes the most detrimental effect on the strength response of the polycrystalline body. In this context, a striking difference in the grain-size his to grams displayed in appears forthepre-1977 alumina material, in which large grains(from10to18 μm) are present in accumulative fraction as high as ≈ 5%. This datum should be compared with the largest grain sizes observed in "second" and "third" generation materials. [1]

The interesting feature is that one can still find a quite high fraction of grains (i.e., 420%) with size/4μm lying on the right foot of the histogram in materials belonging to the second generation.

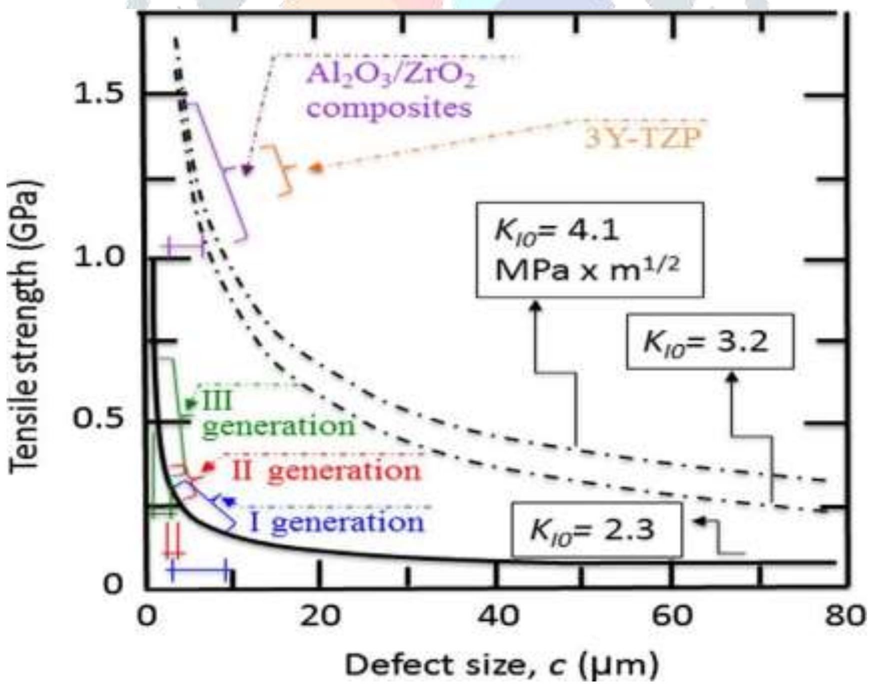

Figure 2.1- relation between tensile strength and fracture toughness

2.2. The multifaceted material evolution of polyethylene liners

The micro structural evolution of biomedical polyethylene materials during the last few decades represents a quite multifaceted argument, by far more complex than the evolution of biomedical ceramics described in the previous section. Arbitrarily and merely for the introductive purpose of this paper to the special issue on artificial hip joints, we shall assume as the apical stone of our discussion an early paper by Eyerer and Ke (1984) on the impact of gamma-irradiation retrieved liners. In light of a density increase, experimentally found for a large number of explanted ultra-high molecular weight polyethylene (UHMWPE) liners, the occurrence of a post-crystallization process was postulated, which in turn resulted in oxidative chain scission. The post-crystallization process led to a reduction in average molecular weight of the polyethylene material and to an increased extract ability of its constituents [2].

These micro- structural changes are now a days generally recognized as the main reason for aging(and failing)of early generations of gamma-irradiated UHMWPE liners .An increase of crystallinity during mechanical and chemical loading of the liner in vivo can be thus seen as a clear symptom of the presence of residual free radicals in the pristine material, triggered by the precursory occurrence of a time-dependent(or "delayed") oxidation process.

However, this aim is not pursued by means of a simple process of high-pressure re-melting. Two different manufacturing paths have been followed:

(i) To sequentially subject the UHMWPE microstructure to several partial steps of irradiation and successive annealing slightly below the melting point.

(ii) To anneal (rather than to re-melt) the irradiated micro- structure in a single step, but in the presence of a selected anti-oxidative substance(e.g., vitamin E)added at some stage of the manufacturing process. Note that the above process

(i) Of sequential irradiation/ annealing is expected to consistently improve the efficiency of free-radical elimination (whilst concurrently maintaining all the possible positive aspects of the annealing process in terms of micro structural development).But, unlike the remelting process, it lacks the physical "guarantee" of a full elimination of the free radicals formed during irradiation. This some what limitative a spectre presents the main motivation behind the above strategy

(ii) A series of commercially available acetabular liners made of biomedical polyethylene materials spanning what is usually defined as "conventional" polyethylene (i.e., earliest generation liners without post-irradiation processes) to remelted, annealed ,sequentially annealed, and vitamin E-doped brands[2].

2.3 New alternatives and technological solutions

2.3.1. Hybridmetal bulk/ceramic surfacefemoral heads

A novel approach has recently been proposed for obtaining artificial hip joint components that could combine the superior structural reliability of metallic components (including their highly reliable strength behavior and high toughness) with the advantages of oxide ceramics as articulating surfaces, including their greater lubricity and superior resistance to abrasive wear.

This new approach, demonstrated by the femoral head components commercialized as Oxinium (Smith &Nephew, Memphis, US),exploits the simple idea of using a joint bearing component, metallic in its bulk but covered with a strong ceramic layer, which would conceivably consent to retain the benefits of ceramic surfaces wedded to the mechanical resiliency of a bulk metallic component[1].

2.3.2. Ceramic resurfacing solutions

An increasingly spreading trend in total hip replacement consists nowadays in minimizing surgical exposure, which in turn involves a less traumatic impact on the muscles surrounding the hip joint and greatly helps to expedite post operative recovery. One such procedure is commonly referred to as hip resurfacing. Hip resurfacing has increasingly become a popular alternative to conventional hip replacement over the last decade in young and active patients. It involves replacing the femoral head surface with a cap (so far only metal has been used), while no stem is inserted into the proximal femoral shaft. The main advantage of this type of surgery is that the femoral bone stock is preserved for possible future revisions. Moreover, by just replacing the surface, the natural biomechanical function of the joint is restored with minimal alterations. Resurfacing surgical algorithms usually lead to a larger head size as compared to conventional hip arthroplasty, in order to lower the dislocation rate. On the other hand, the acetabulum is usually replaced with a (metal) socket as during standard hip replacements. No polyethylene surface is involved with the sliding process[2].

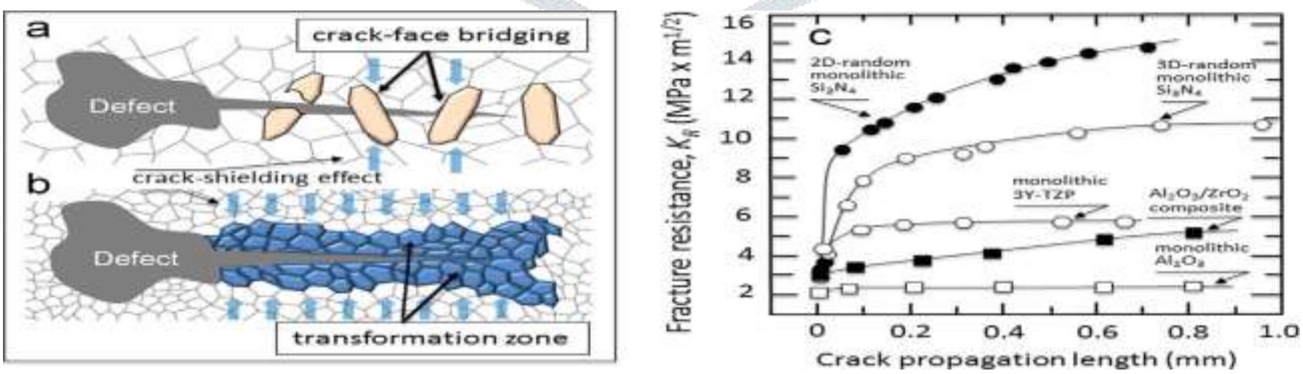

Figure 2.2- fracture development and transformation

2.4. THE CHALLENGE OF JOINT

According to the contents discussed in the previous sections, it might become clear that no single biomaterial has yet been capable to entirely meet all the strict requirements to be fulfilled in joint bearing surfaces. Moreover, it is not clear if such a "perfect" artificial biomaterial could actually be developed and the problems related to hip surgery finally and unquestionably solved. We have seen in for example, how even the most advanced polyethylene materials could only be developed with maximizing certain performances while accepting compromises on others. For these reasons, a wide group of scientists and biotechnologists nowadays keeps spending their full efforts in (rather than replacement with artificial biomaterials)of the damaged cartilage healing [3].

As further progress is searched for, more technological efforts (and finances) become needed, which gradually slow down the development rate. We have attempted a qualitative plot of such curve of progress ,also including future trends, The abscissa of the plot represents the technological efforts through the years, while the vertical axis refers to an hypothetical percentage of technological improvement, which should eventually end in a complete solution of the biomedical problem(i.e.,at100%).Revisiting what might actually be feasible in the middle range development of hip arthroplasty, and with what resources this could become available, one might come across the tangible probability that the medical device market, as presently structured, could turn back toward adopting less advanced [2].

3.1 MATERIALS AND METHODS

Design and proper material selection are equally important elements in the development of orthopaedic implants. Design is primarily concerned with the geometrical parameters; the main geometrical parameters considered in this study being radius of the femoral head (R1), acetabular cup radius (R2) and the thickness of the acetabular cup as shown in

Table 3.1- different cases in the study

Cases	Femoral head radius (mm) R1	UHMWPE cup radius (mm) R2	Radial clearance (mm)	UHMWPE thickness(mm)
A	11.1125	11.287	0.1745	14.115
B	11.1125	11.295	0.1820	7.770
C	13.000	13.124	0.1240	12.280
D	13.000	13.115	0.1150	7.070
E	14.000	14.079	0.0790	11.320
F	14.000	14.076	0.0760	5.720
G	16.000	16.098	0.0980	9.423
H	16.000	16.089	0.0885	5.910

The implant introduced in the human body must resist various biological and mechanical stresses, so material selection an extremely important criteria for the artificial hip prosthesis. Biocompatibility, corrosion resistance, similarity between the mechanical properties of the bone and the implant are some of the important factors that play a vital role in proper material selection. Biocompatibility is the ability of a material to perform effectively with the tissues of the human body without reacting with it. If the selected material is biocompatible there will be no irritation or infection. Otherwise, the human body will consider those materials as foreign agents and attempt to get rid of those foreign contaminant by digesting it using special cells called macrophages. The second important factor to be considered is corrosion resistance.

The human body is a harsh environment for the implanted materials because, after surgery the implanted material is constantly bathed in an extracellular tissue fluid. Those body fluids consist of large amount of sodium ions, chloride ions, and other electrolytes and the orthopaedic implant is considered to be a failure if it is affected by corrosion. Another important factor is the similarity of the mechanical properties of the selected material and the bone. The material selected for the implant should possess similar mechanical properties of the bone to have a similar response to the applied load at the hip joint as that of the pure bone. Taking the above factors into account, the most common choice for the bearing surfaces based on available literatures are metal on metal, metal on plastic, metal on ceramic, ceramic on ceramic, ceramic on plastic, ceramic on ceramic.

The metals used for this analysis are Co Cr, and Co Cr Mo alloy, the ceramic is alumina/zirconia and the plastic used is UHMWPE. This study focuses mainly on optimising the best material selection pair that is metal in contact with plastic/ceramic/metal, based on four different combinations of non-linear materials with eight different cases of geometrical parameters. The four combinations of materials used to optimise the results are Co Cr UHMWPE (COMB 1), CoCr-CoCr (COMB 2), CoCr-CoCrMo (COMB 3), and CoCr-alumina (COMB 4) and their mechanical properties are listed in Table. 2.

Table 3.2- mechanical properties of materials

S. no.	*Alloys*	*Young's modulus (MPa)*	*Poisson's Ratio*	*Yield strength (MPa)*	*Resource for yield strength*
1	Co-Cr	230E3	0.3	720	Oguz et al. (2006)
2	UHMWPE	2200	0.33	21	UHMWPE hand book
3	Co-Cr-Mo	230E3	0.3	517	Oguz et al. (2006

3.2 NATURAL POLYMERS

Polymers can serve as a matrix having various properties including biodegradation [3]. There are two types of biodegradable polymers namely natural polymers and synthetic polymers [4].

Recently, natural polymer-based composites have been focused with more attention than synthetic polymer composites for bone tissue engineering applications. This is often because of the biocompatible and biodegradable behaviour of natural polymers. The natural-based materials are biopolymers which include polysaccharides (starch, alginate, chitin/chitosan, hylauronic acid derivatives) or proteins (soy, collagen, fibrin gels, silk) and a variety of bio-fibers, such as lignocelluloses [5]. Natural polymers often posses highly organized structures and may contain an extracellular substance, called ligand, which is necessary to bind with cell receptors. Natural polymers often possess highly organized structures which can guide cells to grow at various stages of development; they may stimulate an immune response at the same time [6]. Several natural polymers have been reported for their applications in bone tissue engineering [7].

3.3 HYDROXYAPATITE

Hydroxyapatite (Ca10(PO4)6(OH)2; HAp) is the major inorganic component of natural bone and has been used as an orthopaedic and dental material, a column packing material for affinity chromatography to separate various proteins, and in industrial catalysts . Nano-HAp (nHAp) has been applied widely in medical field as a bone repair material because of its excellent bioactive and biocompatibility properties. HAp is known for its biocompatible, bioactive (i.e. ability of forming a direct chemical bond with surrounding osteoconductive, nontoxic, non-inflammatory, non-immunogenic properties . Thus, HAp is one of the ideal materials for bone substitutions due to the nature of its biocompatibility and mechanical strength. Other calcium phosphate apatites including sintered HAp have been widely used for repair and replacement of damaged or traumatized bone tissues [8-10].

3.4 BIOCOMPOSITES

The scaffolds containing polymers and ceramic materials may have excellent properties such as biodegradation, biocompatibility and mechanical strength. Biodegradable polymers are polymers which are decomposed in a living body but whose degradation products remain in tissues for long-term [11]. Biodegradation can be obtained by synthesizing polymers that have hydrolytically unstable linkages in the back bone. An implant prepared from biodegradable polymers can be engineered to degrade at a rate that will slowly transfer load to the healing back bone [12]. These should posses some other requirements such as:

(i) polymers and their decomposition products should be free from immunogenicity or any toxicity;

(ii) degradation and absorption rates should compete with the curing rate of biotissues; and

(iii) these products should have good process ability and excellent mechanical properties to be compatible with human tissues [11].

Biocompatibility indicates the ability of a material to perform with an appropriate host response, in a specific application. This includes both surface compatibility and structural compatibility. Surface compatibility means the chemical, biological, physical (including surface morphology) suitability of an implant surface to the host tissue.

Structural compatibility is the optimal adaptation to the mechanical behaviour of the host tissues [24]. In the body, the mechanical properties of natural bone change with their biological location because the crystallinity, porosity and composition of bone adjust to the biological and biomechanical environments. The high tensile strength and fracture toughness of bone are attributed to the tough and flexible collagen fibers reinforced by HAp crystals [7].

Chitosan and its derivatives are very attractive candidates in the scaffold composites because it is expected that they degrade as the new tissues are being formed, eventually without inflammatory reactions or toxic degradation. Tang et al. [14] showed that the natural hydroxyapatite and chitosan composites have a good hard tissue biocompatibility and an excellent osteo conductivity. They also suggested that this composite may be suitable for artificial bone implants and frame materials of tissue engineering. Jayabalan et al. [14] have shown that the nano-composite containing calcined Hap nanoparticles is both biocompatible and osteocompatible. From the biomimetic point of view, composites of HAp could potentially improve both biocompatibility and mechanical properties of bone grafting materials. The electrospun nanocomposite nanofibers of HAp/chitosans with compositional and structural features close to the natural mineralized nanofibril counterparts were prepared by Zhang et al. [15] and they reported that these nanocomposite fibers are of potential interest for bone tissue engineering applications. A biopolymer-based novel nanocomposite chitosan/montmorillonite
(MMT)/hydroxyapatite (HAp) has been reported for biomedical applications [15].

This nanocomposite showed improved mechanical properties suggesting its potential applications in bone tissue engineering. The amount of nHAp in the bio composites appears to have a greater effect on the early stages of osteoblast behavior (cell attachment and proliferation) rather than the immediate/late stages (proliferation and differentiation) [16]. The addition of nHAp into the chitosan scaffolds improved cell attachment, higher proliferation, and well-spread morphology when compared to the chitosan scaffolds alone [16]. We have recently prepared and characterized the chitosan and gelatin scaffolds (CG) and the CG scaffolds containing nHAp.

The presence of HApin the CG scaffolds enhanced osteoblastic cell attachment and spreading [7]. The membranes containing chitosan hydrogel and hydroxyapatite [7], and the membranes containing _-chitin and hydroxyapatite [6] were reported to be suitable for osteoblastic cell attachment and proliferation. Kashiwazaki et al. [17] have fabricated novel chitosan/ hydroxyapatite nano composites with porous structure by the co-precipitation and porogen leaching method. These composites were found to have biocompatibility and biodegradation. Chitosan bio component nano fibers with an average diameter controlled from 100 to 50nm were successfully prepared by electrospinning chitosan and poly(vinyl alcohol) blend solution [17].

Chitosan–hydroxyapatite nanostructured composite films have also been prepared by solvent casting their hybrid suspensions for tissue engineering applications [18]. A chitosan/hydroxyapatite nanohybrid scaffold with high porosity and homogeneous nanostructure was fabricated through a bionic treatment combined with thermally induced phase separation. These nHAp particles had positive impacts on directing apatite crystallization in the scaffold and led to the good bioactivity of the nanohybrid scaffold [18].

Chitosan along with other natural polymers can be mixed with bioceramic material to prepare the biomimetic scaffolds. Zhao et al. [19] showed a biodegradable hydroxyapatite/chitosan–gelatin network (HAp/CS-Gel) composite of similar composition to that of normal human bone prepared as a three-dimensional biomimetic scaffold by phase separation method for bone tissue engineering. They reported that the cell/scaffold constructs had good biomineralization effect after 3 weeks in culture.

Zhao et al. [19] fabricated two types of biomimetic composite materials, Chitosan–gelatin (CG) and hydroxyapatite/chitosan–gelatin (HCG) and demonstrated enhanced protein and calcium ion adsorption properties of HAp in the CG polymer network which improved initial cell-adhesion and long-term growth, favour osteogenic differentiation upon induction, as well as maintained the progenicity of the 3Dhumanmesenchymal stem cell constructs.Anewartificial bone matrix with collagen and Ca(5)(PO(4))(3)OH (hydroxyapatite/ HAp) which are the main components of natural bone was prepared [20]. To improve the property of the artificial bone matrix, chitosan was cross-linked into the scaffolds. The authors reported that the above artificial bone matrix could be used as a bone substitute with outstanding properties.

The weight ratio of individual material in the composites maybe critical to support bone tissue growth. The composites containing nano-hydroxyapatite/chitosan/carboxymethyl cellulose (nHAp/CS/CMC) with weight ratios of 70/10/20, 70/15/15 and 70/20/10 through a co-solution method were prepared [20]. These novel composites of nHAp/CS/CMC have been shown to be a promising prospect used for bone repair materials in view of the good mechanical property, adjustable biodegradation rate and bioactivity in simulated body fluid (SBF). Liuyun et al. [3] reported the physico-chemical and biological properties of a novel biodegradable composite scaffold made of nano-hydroxyapatite and polymers of chitosan and carboxymethyl cellulose, namely, nHAp/CS/CMC, which was prepared by freeze-drying method.

The scaffold was non-toxic and had good cell biocompatibility, and the results of implantation experiment in vivo showed that the scaffold has good tissue biocompatibility and has a great potential to be used as bone tissue engineering material. It has been recently reported that the inclusion of hydroxyapatite into the chitosan–gelatin scaffolds could promote osteoblastic cell attachment, spreading and proliferation [7]. It has been recently shown that the porous scaffolds composed of alginate and chitosan were fabricated by combining the formation of polyelectrolyte complex with freeze-drying and they exhibited higher mechanical strength and better thermal stability.

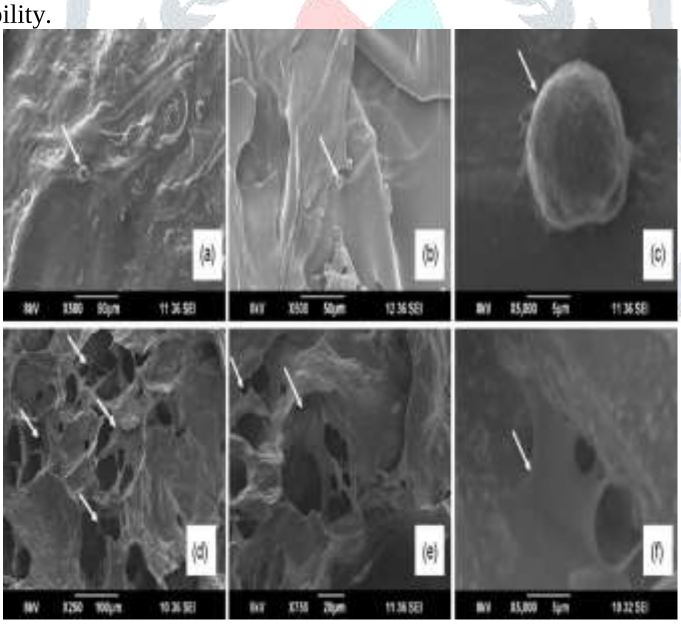

Figure -3.1 SEM images showing osteoblastic cell attachment

4. COATINGS AND TREATMENTS

4.1 CURRENT TREATMENT OF JOINT ALLOY SURFACES

The surfaces of virtually all alloys used for total hip replacement and total knee replacement in clinical practice are treated to enhance their functionality. Deliberately modifying surfaces effectively decouples the requirements placed on the bulk properties of the joint replacement from the characteristics required only for its surface. From a certain viewpoint, all surface treated alloys for joint applications can be viewed as bi-material composites or as functionally graded composites since they feature advantageous properties of both the components.

4.2 TREATMENTS TO REDUCE WEAR

Perhaps the most demanding requirement placed upon joint replacement materials is the need for their sliding interfaces to sustain high loads for millions of cycles without degradation. This requirement has been addressed by optimizing the combination of three interrelated engineering properties: hardness, wear resistance, and friction coefficient (Section 4.3). Numerical models of wear have been developed which consider the complex interactions of surface properties, geometric properties, and physiological conditions within joints [20]. Wear modeling of entire three dimensional joint systems provide insight into how contact pressures, temperature changes, fluid film shear stresses, friction coefficients, and asperity contact contribute to the volumetric and linear wear rate. Thus these models guide the development of materials and enable prediction of how surface treatments that alter hardness, roughness, or friction may improve the overall wear resistance.

Although metal surfaces can be hardened, their ability to resist wear is intrinsically inferior to materials like ceramics or intermetallics. Therefore, surface treatments to add ceramics or to introduce ceramic-like properties are most prominent for reducing wear and wear debris. The results of numerical, experimental, and clinical studies also unambiguously point to the need to alter hard metal or ceramic surfaces to reduce wear in adjacent polymers. For example, TiN coatings were applied to Ti6Al4V and CoCrMo alloys via multiple deposition methods including Physical Vapor Deposition (PVD), Plasma Immersion Ion Implantation (PIII), Arc Evaporative Physical Vapor Deposition (AEPVD), Powder Immersion Reactive Assisted Coating (PIRAC), Arc Ion Plating (AIP), Arc Vapor Ion Deposition (AVID), Nitrogen Ion Implantation (NII), and Nitrogen Diffusion Hardening (NDH) [21].

The results of trials with these coatings varied. However, the reduction in the coefficient of friction and reduction in volumetric wear of UHMWPE was as much as 42% in one case. In another case, the presence of pin holes or partial delamination of the TiN layers increased the volumetric wear rate in UHMWPE by a factor of 4. Clinical studies of 76 patients with cemented CoCrMo hip joints with, or without, TiN coatings found loosening in 44% of implants with treated surfaces, as compared to only 21.6% with untreated surfaces [21]. The results of six other clinical studies showed varied results, but the evidence of TiN coating providing a significant advantage was insufficient. Decohesion of the TiN coating and the presence of pin holes were generally identified as factors contributing to the failure.

The thinness of the coating may have been a factor contributing to both pin holes and delamination. Also, the frictional characteristics of the coating depend on the presence of lubricant between the UHMWPE and the hard surface. For example,the presence of proteins on the surface has been shown to ameliorate the influence of surface roughness of TiN coatings [21]. Zimmer has tried nitrogen diffusion into Ti6Al4V. As with TiN, the coating layer was too thin to last long in service. Yildiz et al. [22] reported a very low wear rate for Ti6Al4V surface nitrided at 750 _C for 1 h (0.14 _ 10_6 mm3/Nm, pinon- disc tests, 141 m). Yet other coatings have been proposed, such as ZrN for coating CoCr. Studies of the effect of implantation of C and N ions on wear of the CoCrMo alloys generally show significant reduction in wear. Wang et al. [22] reported the wear rate of surface plasma nitrided CoCrMo samples to be less than 4 _ 10_8 mg/Nm at a load of 8.5 MPa. Wear rates of CoCrMo with implanted C and N atoms were reported to reduce by a factor of 3 in ball-on-disc wear testing with an alumina ball under the load of 5 N [23]. Ion implantation can also increase surface hardness in femoral heads, as introduced by Stryker in their Low Friction Ion Treated (LFIT) CoCr heads. The depth of hardening achieved was on the order of 0.2 lm [23].

4.3 TREATMENTS TO ENHANCE FATIGUE STRENGTH

The classic approaches to improve fatigue strength, both of which have been implemented through surface modification, are to reduce the stresses that initiate fatigue cracks and to reduce the presence of crack initiation sites. Peening of surfaces by either laser or mechanical means imparts compressive stresses into surface regions, the presence of which reduces or even eliminates the driving force for fatigue crack initiation and extension. For example, Abrasive Water Jet (AWJ) peening applied to treat surfaces of Ti6Al4V and AISI 304 stainless steel produced compression residual stresses as high as 460 MPa in the AISI 304 stainless steel surfaces. On the other hand, the fatigue endurance limit of Ti6Al4V increased by 25% to 845 MPa [23].

Other mechanical surface treatments such as ultrasonic shot peening, contact rolling, and laser shock peening were applied to alter near-surface residual stress profiles. Laser shock peening created biaxial compressive residual stresses greater than 500 MPa to depths of 0.5 mm in Ti6Al4V [23]. This treatment used 8 ns laser pulses with an intensity of 10 GW/cm2 to generate shock pulses greater than 7 GPa over 3 mm diameter surface regions. The induced microstructural changes were measurable to the depth of 1.6 mm and the associated increase in fatigue endurance limit was 17.2%, up to 639 MPa. The same authors also examined surface contact rolling using a 6 mm diameter hardened steel sphere rastered across the Ti6Al4V surface under the pressure of 30 MPa. This treatment induced the compressive stresses (exceeding 1000 MPa within the first 50 lm) to the overall depth of 0.6 mm. The fatigue endurance limit increased by 13.3% to 617 MPa, which was slightly less than achieved by the above-mentioned laser shock treatment.

However, the surface contact rolling provided an additional advantage of reducing the surface roughness. It decreased the arithmetic average roughness Ra by 64% compared to the untreated state. In examples of ultrasonic shot peening, ultrasonic energy at 20 kHz from 1 mm diameter ceramic beads [23] and a 2.6 mm tungsten carbide ball tip [23] were applied to surfaces of Ti6Al4V. The ultrasonic treatment with ceramic balls increased the surface hardness from 350 HV to 400 HV to a depth of 50 lm.

The corresponding increase in fatigue endurance limit was from 545 MPa to 600 MPa (10.1%). Comparable increases in surface hardness, up to 426 HV, were attained by ultrasonic surface treatments for cases in which the density of ultrasonic impacts was greater than 70,000 strikes/mm2 [23]. X-ray and SEM analyses of ultrasonically treated Ti6Al4V surfaces revealed a high density of twins and an increased volume fraction of b phase Ti within a highly deformed layer extending 120 lm beneath the top surface. Sub-surface dislocation density increased from 1.25 _ 1015 m_2 to 1.85 _ 1015 m_2 after ultrasonic treatment of 96,000 strikes/mm2. Fatigue performances for these microstructures were not reported.

5.1 EXPERIMENTAL VALIDATION

The finite element result from the present work is validated with the experimental observations of Jin et al. (1999). Their work compares the experimental and simple elasticity results with the FEA results for a metal to plastic elastic contact case. However the present work extends their work for all possible biocompatible contacting materials combinations as mentioned above with their non linear material properties. Jin et al. (1999) presented an experimental methodology to measure the contact radius of the acetabulam/femoral head while loading. Figure 3a shows the comparison between the experimental and the present FEA non linear results for the contact radius. The results are obtained for the five intermittent load values
i.e. 500 N, 1000 N, 1500 N, 2000 N and 2500 N.

It is found that the contact radius calculated after the deformation results with the increase in the load. Good agreement has been found for the complete contact radius results between the experimental and also the FEA observations with a maximum difference of 13.4% in the maximum loading region, but the difference is negligible in other areas. The important design parameter which affects the contact radius is that the radial clearance between the femoral head and also the acetabulam cup.

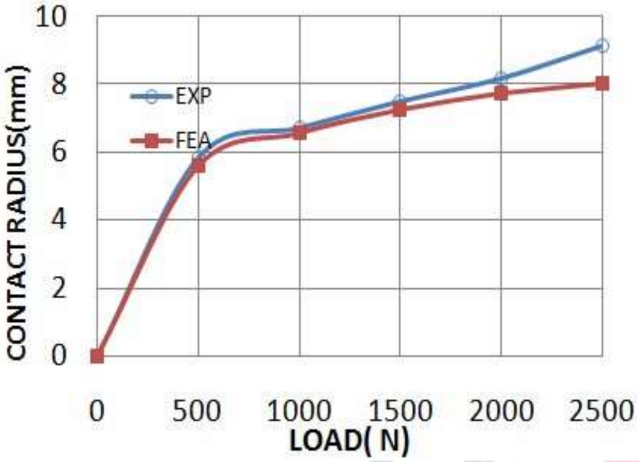

Figure 5.1- contact radius

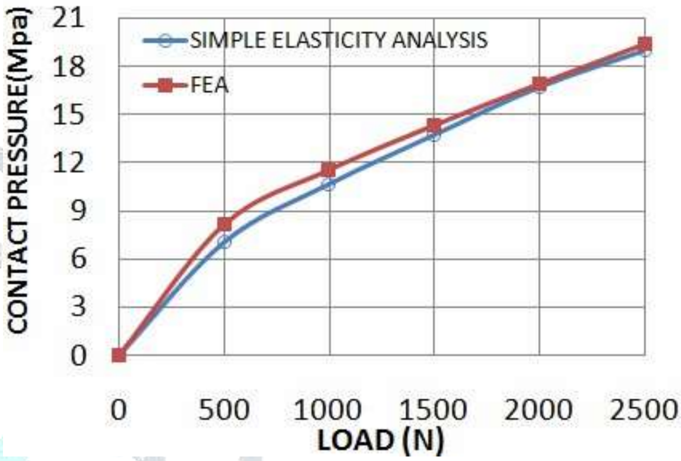

Figure 5.2- contact pressure

As the loading gradually increases, the deformation over the acetabulam contact increases, at some stage the load crosses the critical (elastic limit) load postulated by Hertz theory. Pressure between the contacting regions also increases with increases in loading. The FEA contact pressure results are validated with the results obtained from the simple elasticity analysis performed by Jin et al. (1999). The contact pressure increases gradually with the increase in load on the acetabular cup as shown in Figure 3b. Good agreement has been found for all the expected maximum contact pressure values in the present study between the simple elasticity analysis and the present finite element results. It is found that a maximum difference of 13% is visible at the 500 N load region between the two results. The comparison is absolutely reliable since all the geometrical parameters considered in the two studies are the same.

5.2 IMPACT OF CONTACT STRESSES ON DIFFERENT BEARING COUPLES

The actual contact stresses that occurring at the acetabular cup depends upon the material and design of the prosthesis. Hip joint is subjected to several forces throughout the static and dynamic loading conditions. These forces tend to migrate the femoral component towards the acetabular component which ends up in excessive contact stresses. The high contact stresses developed at these areas causes the localised plastic deformation. Figure 5 shows the maximum von Mises stress developed for the different bearing couples when it is subjected to 2500 N. By analysing the different bearing couples like metal on plastic, metal on metal, metal on ceramics, ceramics on plastic, ceramic on metal and ceramic on ceramic, it is concluded that the metal in contact with plastic i.e.CoCr head paired with UHMWPE generates lower contact stresses at the cup. It is because of the lower yield strength of the plastic (UHMWPE) material. The yield strength of UHMWPE is 21 MPa. The maximum stress value developed for the CoCr-UHMWPE pair is 7.62 MPa which is lower than the yield strength of UHMWPE. This reduction in contact stresses lead to an improvement in the performance of the prosthesis which is an achievement for orthopedics. Metal on ceramics contact case produces the maximum von Mises stress of more than 76 MPa when compared to the other material combinations.

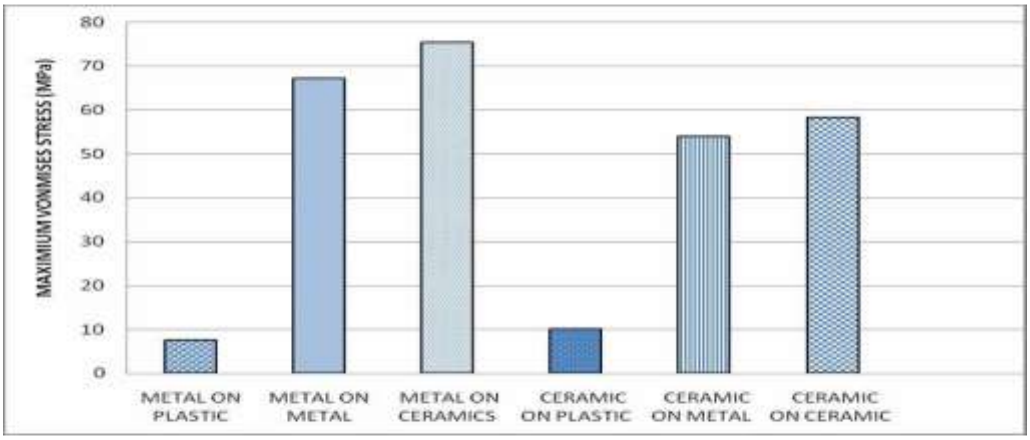

Figure 5.3- maximum von mises stress for different combination of materials

5.3 VARIATION OF CONTACT PRESSURE FOR DIFFERENT BEARING COUPLES

Radial clearance strongly influences the contact pressure at the interface regions and it is found from the previous study that the reduction in the radial clearance value decreases the contact pressure values. Figure 6 shows the variations in the maximum contact pressure at the acetabular cup with different combinations of bearing materials It is inferred that the metal on plastic combination i.e. CoCr head paired with UHMWPE combination yields the lowest contact pressure at the interface region.

The maximum contact pressure is induced for the metal on ceramic pair i.e. stainless steel in contact with alumina. It is due to the fact that for material with the higher hardness and higher elastic modulus the initial Hertzian contact area will be lesser also having the lower elastic deformation.

Both stainless steel and Alumina are the harder materials, but in the case of UHMWPE, it has lower hardness value and lower elastic modulus when compared to the other materials. So the contact area will become more for this combination thereby decreases the contact pressure. The contact pressure value decreases by 91% when metal on plastic combination is considered instead of metal in contact with the ceramics case.

So from the present study, it is evident that the presence of metal on plastic contact pair i.e. CoCr as a head and UHMWPE as a cup, with higher femoral head radius and cup thickness, lower radial clearance value gives the lowest contact pressure than the other combination of bearing surfaces.

The main objective of the today's implanting is to relieve the patent's pain and discomfort by providing more life. Figure 7 shows the variations in the maximum contact radius at the acetabular cup with the different combinations of bearing surfaces when it is subjected to
2500 N. Radial clearance also plays a crucial role in predicting the contact radius values.

The contact radius between the cup and the femoral head should be more to get the maximum contact area that would prevent the failure of the material due to the excessive stress values.

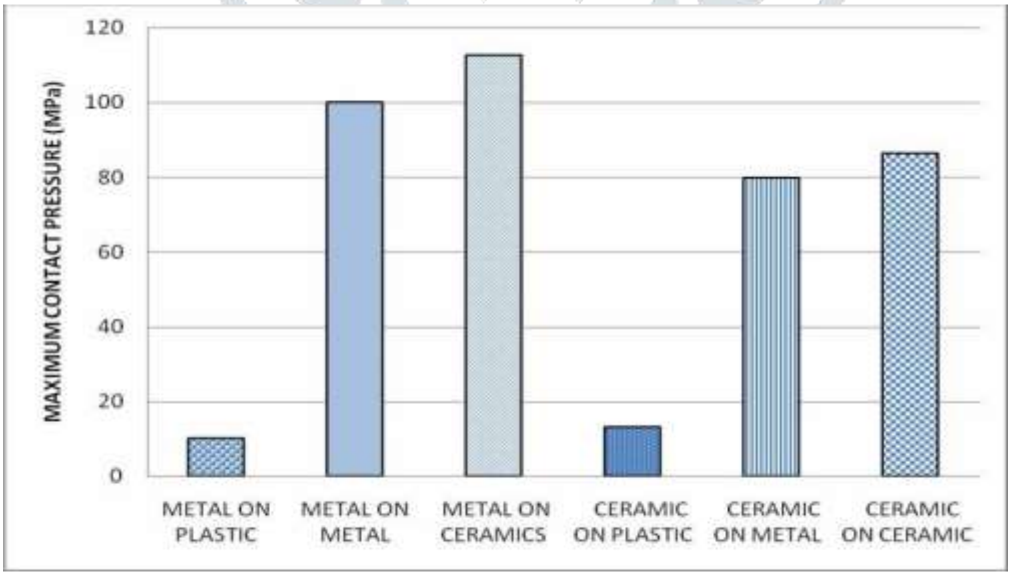

Figure 5.4- maximum contact pressures for different combination of materials

Both stainless steel and Alumina are the harder materials, but in the case of UHMWPE, it has lower hardness value and lower elastic modulus when compared to the other materials. So the contact area will become more for this combination thereby decreases the contact pressure.

6. CONCLUSION

Thus the different material behaviours involved in the hip joint anthroplasty is studied. Materials with good structural reliability and bio- compatibility are more preferred for the future work. Components like non-oxide ceramics do have a high strength and toughness factor in them. The fracture development are observed effectively by advanced spectroscopy methods. People have received arthroplastic surgeries extending their mobility and quality-of-life by years. Titanium, stainless steel, and CoCr remain the most extensively incorporated alloys in joint replacement. Enhancements to each of these alloy systems continue to be explored, as well as new alloys based on alternative metals such as Zr and Ta.

Even though natural polymer-based biomaterials such as chitin, chitosan, collagen have the biocompatible and biodegradable properties, their applications as scaffold materials for bone cell seeding in tissue engineering are often limited or restricted by their poor mechanical strength. The inclusion of nanoparticles of hydroxyapatite into the biopolymer matrices improves the mechanical properties and incorporates the nanotopographic features that mimic the nanostructure of natural bone.

Tribological assessment of different bearing couples were investigated in terms of maximum von Mises stress, contact pressure, and contact radius by using the finite element concepts. The bearing surfaces includes metal on plastic, metal on metal, metal on ceramic, ceramic on plastic, ceramic on metal and ceramic on ceramic. The significant findings from this present study are the metal (CoCr) on plastic (UHMWPE) bearings couple will facilitate within the development of the hip implant. This combination reduces von Mises stress by 90%, contact pressure by 91% and provides three times the contact radius than other material combinations. It is due to the lower yield strength, lower hardness value and lower elastic modulus of the plastic (UHMWPE) material when compared to the other materials.

7. ACKNOWLEDGMENT

Authors are grateful to the Head of the Department and the faculties of Mechanical Engineering of Sri Ramakrishna Engineering College, Coimbatore for their valuable suggestions and guidance.

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