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Review

Current and future biocompatibility aspects of biomaterials for hip prosthesis

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Abstract: The field of biomaterials has turn into an electrifying area because these materials improve the quality and longevity of human life. The first and foremost necessity for the selection of the biomaterial is the acceptability by human body. However, the materials used in hip implants are designed to sustain the load bearing function of human bones for the start of the patient's life. The most common classes of biomaterials used are metals, polymers, ceramics, composites and apatite. These five classes are used individually or in combination with other materials to form most of the implantation devices in recent years. Numerous current and promising new biomaterials i.e. metallic, ceramic, polymeric and composite are discussed to highlight their merits and their frailties in terms of mechanical and metallurgical properties in this review. It is concluded that current materials have their confines and there is a need for more refined multi-functional materials to be developed in order to match the biocompatibility, metallurgical and mechanical complexity of the hip prosthesis.

Keywords: biocompatibility; biomaterials; hip prosthesis; mechanical properties; metallurgical properties

1. Introduction

Biomaterials are artificial or natural materials used in the fabrication of implants to replace the damaged or diseased biological structure, for example hip joint, to restore form and function effectively. These biomaterials provide help in the improvement of longevity and quality life of human being. The field of biomaterials has shown rapid growth to keep pace with the increased demand of unfortunate fractured bones and aged people. Therefore, the number of implants used for

spinal, hip and knee replacements are extremely high, however, other implants including artificial valves in the heart, stents in blood vessels, shoulders implants, knees, hips, elbows, ears and orodental structures are also on demand. The biomaterials used in the fabrication of different parts of the human body are fabricated from metals, polymers, ceramics and composites [1–3].The human joints well suffer from human degenerative diseases such as arthritis leading to pain or loss in function of a part in the body. Moreover, these diseases lead to degradation of the mechanical properties of the bone due to absence of normal biological self-healing process or excessive loading. n recent years, 90% of population over the age of 40 suffers from human degenerative diseases while an aged person has been suffering tremendously with an estimate of seven time's increase (4.9 million in 2002 to 39.7 million by 2010) [4]. The artificial or natural biomaterials are the only solution for these kinds of problems, as surgical implantation of appropriate shape and size can help in restoring the function of functionally compromised structures. However, the success of surgical implantation of a joint or bone replacement lies with the orthopaedic surgeon, not only to perform the surgery, but also to select the best suitable replacement for the patient. The selection process can be influenced by the age, weight, behavior of materials and the activity level of the patient post-surgery [5,6]. Furthermore, the selection of an appropriate biomaterial for the fabrication of implants is precarious for the effectiveness and success of the implant. There are various factors which influences the selection of biomaterial for a specific implant depends, for example, material properties, design of implant and biocompatibility of the material. In addition, the factors such as the surgical procedures, health condition of the patient, and the activities of the patient are also very important and considerable. The metallic and polymeric biomaterials used in total joint replacement (TJR) applications are discussed in this review.

2. Overview of Biomaterials for Hip Implants

Historically, a total hip replacement (THR) or the articulation of a human hip is simulated with the use of two components, a cup type and a long femoral type element. The head of the femoral element fits inside the cup to enable the articulation of human joint (see Figure 1). These two parts can be fabricated from metals, ceramics, polymers and composites (see Table 1 and Table 4). Typically, polymeric materials alone are too weak, therefore, not suitable to meet the requirement of stress deformation responses in THR components. However, metals have good mechanical properties but poor biocompatibility, release of dangerous metal ions and stress shielding effect causing eventual failure and may lead to removal of implants. Ceramics usually have good biocompatibility but poor fracture toughness and tend to be brittle. Composite materials with engineered interfaces resulting in combination of biocompatibility, mechanical strength and toughness, is the focus of many current studies [7,8].

The first total hip replacement was fabricated from stainless steel cup and head was developed in 1938 by Philip Wiles [9]. The cup was fixated using screws and the head was fixated using a stem, which was fixed to the neck of the femur by a bolt. The clinical results of this development were not known due to the intervention of World War II. Years later, metal-on-metal combinations were introduced by [8],[10], and [11]. The results from such implants were impractical due to loosening and high wear of the components as a result of increased frictional torque [12]. These MoM prostheses were manufactured to give a matching femoral head and acetabular cup with no clearance (small space between the head and cup). This small clearance could reduce friction between them by

creating a polar bearing [12]. In 1958, Sir John Charnley implanted first metal-on polymer total hip replacement [13] and the polytetrafluoroethylene (PTFE) was the first polymer articulating against a stainless steel head, which only was survived for two years due to the rapid wear of PTFE. Additionally, in 1961, Charnley adopted ultra-high molecular weight polyethylene (UHMWPE) as prosthesis, which provided a low interfacial friction against a metal head known as "Low Friction Arthroplasty" (LFA). The clinical results have encouraged the use of Charnley prosthesis and are still being used to date in less active elderly patients [14]. The metal-on-metal bearings were of great interest in the late 1980s after the long-term survivorship and the low wear. The improved tolerances and better surface finishes can be achieved by better manufacturing technologies, and may improve the outcome of metal-on-metal prostheses [15,16]. In early 1970s, Boutin in France, developed the first ceramic-on-ceramic (CoC) total hip replacement, widely used in Europe after the promise of ceramics as a highly inert material, good surface finish and excellent resistance to wear in vivo [17]. Although, CoC have shown enhanced wear performance, there are still concerns about the incidental fracture of the ceramic material [18].The outcome of most possible combinations of the femoral and socket component materials used in hip replacement prosthesis have been identified by [19,20].

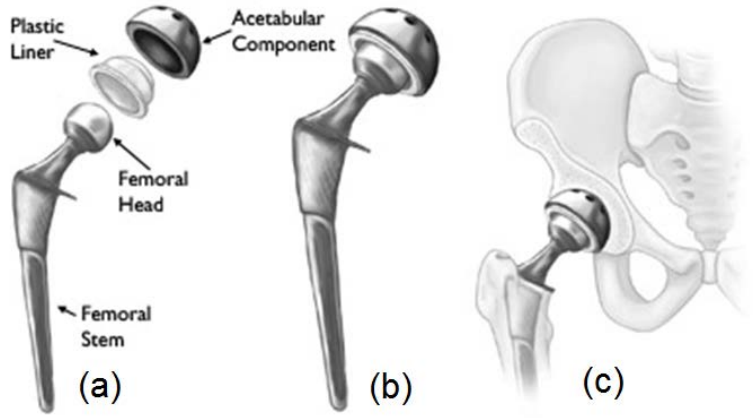

Figure 1. (a) Components of a total hip replacement; (b) The components merged into an implant; (c) The implant as it fits into the hip [15].

Table 1. Materials used in total hip replacement (THR) prosthesis.

Femoral component	Socket component	Results
Co-Cr-Mo	Co-Cr-Mo	High loosening rate and restricted use; new developments show minimum wear rate
Co-Cr-Mo	UHMWPE	extensively in use; low wear
Alumina/zirconia	UHMWPE	Very low wear rate; zirconia more impact resistant
Alumina	Alumina	Minimum wear rate (components matched); pain; not in clinical use in the United States
Ti-6Al-4V	UHMWPE	Reports of high UHMWPE wear due to breakdown of titanium surface
Surface-coated Ti-6Al-4V	UHMWPE	superior wear resistance to abrasion; merely thin treated layer attained

3. Requirements of Biomaterials: General Issues and Concerns

The choice of biomaterials basically depends on the site of implant and medical usage. The development of new biomaterials is an interdisciplinary research and often requires a combining effort of material scientists, biomedical specialists, pathologists and clinicians. In order to serve for longer time without rejection an implant should possess the following qualities.

3.1. Biomechanical compatibility

The mechanical properties that designate the type of material will be nominated for a precise application. The response of biomaterial to the repeated cyclic load is determined by the fatigue strength of the material which determines the long-term success of the implant. The biomechanical incompatibility is defined as the fracture produced in an implant due to the inadequate strength or mismatch in mechanical property (known as modulus) between the bone and implant. Hence, the biomaterial used for an implant should be biomechanical compatible and expected to have a modulus equivalent to that of bone for the success. Moreover, the biomechanical incompatibility leads to death of the bone cells known as "stress shielding". The bone modulus varies in the magnitude from 4 to 30 GPa depending on the type of the bone and the direction of measurement [21,22].If the implant material have higher stiffness than the bone, prevents the required stress being transferred to adjacent bone, results in bone resorption around the implant, consequently leads implant loosening. Thus a material with excellent combination of high strength and low modulus closer to bone has to be used for implant to avoid loosening of implants and higher service period for avoiding repeated surgery.

3.2. Biocompatibility

Table 2. Categorization of biomaterials based on its interaction with its surrounding tissue.

Categories	Examples	Response	Effect
Biotolerant materials	Polymer-polytetra-fluorethylene (PTFE), polymethylmethaacralyte (PMMA), Ti, Co-Cr, etc.	Formation of thin connective tissue capsules (0.1–10 μm) and the capsule does not adhere to the implant surface	Rejection of the implant leading to failure of the implant
Bioactive Materials	Bioglass, synthetic calcium phosphate including hydroxyl apatite (HAP)	Formation of bony tissue around the implant material and strongly integrates with the implant surface	Acceptance of the implant leading to success of implantation
Bioreabsorbable Materials	Polylactic acid and polyglycolic polymers and processed bone grafts, composites of all tissue extracts or proteins and structural support system	Replaced by the autologous tissue	Acceptance of the implant leading to success of implantation

The materials used as implant are expected to be highly nontoxic and should not cause any inflammatory or allergic reactions in the human tissues and cells. However, the success of the biomaterials is mainly dependent on the reaction of the human tissues with implant, and this also measures the biocompatibility of a material [23]. The two main factors that influence the biocompatibility of a material are the host response induced by the material and the materials degradation in body. Geetha et al. classified various biomaterials for hip prosthesis based on the response in human body (see Table 2) [24]. Bioactive materials are highly preferred as they give rise to high integration with surrounding bone, however, bio-tolerant implants are also accepted for implant manufacturing. When implants are exposed to human tissues and fluids, several reactions take place between the host and the implant material and these reactions also dictate the bio-acceptability of these materials. The issues related to biocompatibility are (1) thrombosis—which involves blood coagulation and adhesion of blood platelets to biomaterial surface, and (2) the fibrous tissue encapsulation of biomaterials that are implanted in soft tissues.

3.3. High corrosion and wear resistance

The low wear and corrosion resistance of the implants in the body fluid results in the release of non-compatible metal ions. These released ions are found to cause allergic and toxic reactions [25]. The service period of the material is mainly determined by its abrasion and wear resistance. However, the low wear resistance also results in implant loosening and wear debris deposited in tissue causes several adverse reactions [26]. Thus, the development of implants with high corrosion and wear resistance material is of prime importance for the longevity in the human system.

3.4. Osseointegration

Table 3. Requirements of biomaterials and problems resulting from inadequate requirements.

Significant requirements	Consequences of not fulfilling the requirements
Long fatigue life	Implant mechanical failure and revision surgery
Adequate strength	Implant failure, pain to patient and revision surgery
Modulus equivalent to that of bone	Stress shielding effect, loosening, failure, revision surgery
High wear resistance	Implant loosening, severe inflammatory response, destruction of the healthy bone and producing wear debris which can go to blood.
High corrosion resistance	Releasing non compatible metallic ions and allergic reactions
Biocompatibility	Body reaction and adverse effects in the organic system
Osseointegration	Fibrous tissue between the bone and the implant, not well integration of the bone and implant and finally implant loosening

The inability of an implant surface to integrate with the adjacent bone or the other tissues due to micro-motion, results in implant loosening. A fibrous tissue is formed between the bone and the

implant, if the implant is not well integrated with the bone. Hence, materials with an appropriate surface are highly essential for the implant to integrate well with the adjacent bone [27]. The surface chemistry, roughness and topography play a major role in the development of good osseointegration. The general requirements of biomaterials and the problems occurring due to non-compliance is summarized in table 3 [28].

4. Current and New Promising Biomaterials

Most synthetic biomaterials used for implants are common materials familiar to the average materials engineer or scientist. The section below discusses application of various currently and new promising biomaterials used for total hip joint replacement and fabrication of implants.

4.1. Metallic biomaterial

Metallic implants are the primary biomaterials used for joint replacement and become gradually important. The metallic implants used for orthopedic applications can be fabricated from stainless steel, CoCr alloys, Ti and Ti alloys (see Table 4).These metallic materials have several promising properties, for example, excellent thermal conductivity, high strength, high fracture toughness, hardness, corrosion resistance and biocompatibility, which make them an excellent choice for total joint replacement [29]. However, the disadvantage with metallic implants is their high elastic modulus, which causes stress shielding and corrosive nature. The consequence of corrosion is loss of material, which will weaken the implant, and probably the most importantly, the corrosion products and metal ions released into the tissue resulting in undesirable side effects [30]. They also have other additional drawbacks such as low bio-compatibility, too high stiffness compared to tissues and high density.

4.2. Stainless steel

Stainless steel (316 and 316L) is the first material used to fabricate artificial bone and easily cast into different shape and size. 316L has a better corrosion resistance than 316 due to lower carbon content. However, it may corrode inside the body under certain circumstances such as highly stressed oxygen-depleted environment. Both 316 and 316L are suitable to fabricate temporary devices such as fracture plates, screws, and hip nails. Due to ease of fabrication and desirable assortment of mechanical properties, corrosion behavior, stainless steel become the predominant implant alloy [31].The lower the carbon content the more is the corrosion resistant to the physiological saline in the human body. Based on this reasoning, the American Society of Testing and Materials (ASTM) have recommended 316L as a principal alloy for implant fabrication in comparison to other SS grade [32]. However, the other alloying elements include nickel (Ni), which is used to increase corrosion resistance in more aggressive environments and molybdenum (Mo) which improves localized corrosion resistance against pitting, fretting, and crevice corrosion [33].The minimum amount of Ni for maintaining austenitic phase is approximately 10%. In the last few years, development has been made toward nitrogen-rich austenitic stainless steels such as Rex734 (ISO 5832-9: E) [34] and nickel-free high-nitrogen austenitic steels such as PANACEA P558 [35] for medical devices. These alloys have a higher Mo and N than 316L and

hence, more resistant against pitting and crevice corrosion. In situations where the original cast, wrought, or forged alloy requires a physical or chemical alteration, heat treatments such as annealing, cold working, and hot forging, can be of service [36]. The advantages of 316 and 316L in surgical implants fabrication include good hot-and cold-working mechanical properties, ultimate tensile strengths, yield strengths, elongations, and low cost [37] (see Table 4). Though SS maintains fine biocompatibility properties, however, there is a tendency to fall short in the viewpoint of fatigue resistance because of low proportional limits that lead to initiation and propagation of fatigue cracks [38].

4.3. Cobalt-chrome

Ever since the use of stainless steel as the first biomaterial in THR surgeries, biomedical manufacturers have aimed research towards the fabrication of additional alloys with superior mechanical properties [39]. However, cobalt-based alloys are among the safest biomaterials for orthopedic prostheses, because of their exceptional corrosion resistance in chloride environments, which is due to the specific weight percentages of base elements and alloy additions in their compositions, the formation of the chromium oxide Cr_2O_3 passive layer [40] and mechanical strength. Compared to the wrought alloys, cobalt-based casting alloys are characterized by higher contents of high melting point metals such as chromium, tungsten, tantalum, titanium, zirconium, and higher carbon contents. The CoNiCrMo alloy originally called MP35N (Standard Pressed Steel Co.) contain approximately 35% Co and Ni each is highly corrosion resistant to seawater under stress. The abrasive wear properties of the wrought CoNiCrMo alloy are similar to the cast CoCrMo alloy (about 0.14 mm/yr in joint simulation tests with ultra-high molecular weight polyethylene acetabular cup); however, the former is not recommended for the bearing surfaces of joint prosthesis because of its poor frictional properties with itself or other materials. However, the superior fatigue and ultimate tensile strength of wrought CoNiCrMo alloy made it suitable for the applications which require long service life without any fracture or stress fatigue, for example, in case of stems of the hip joint prostheses. This advantage is better appreciated when the implant has to be replaced, since it is quite difficult to remove the failed piece of implant embedded deep in the femoral medullary canal. The typical microstructure of cobalt-based alloys consists of a cobalt-rich solid-solution matrix containing carbides (i.e., Cr7C3, and M23C6) within the grains and at grain boundaries, where chromium, tungsten, tantalum, silicon, zirconium, nickel, and cobalt, may be present in a single carbide particle [40,41]. The early version of cobalt-based alloys used for hip implants contains relatively high carbon (0.2%), and typically fabricated by investment casting [42]. Depending on the casting method, the manufacturing process has the ability to produce at least three micro structural features that can strongly influence implant properties, both positively and negatively [40]. These features include: (1) if not the typical F75 microstructure, then the formation of inter-dendritic regions that become solute (chromium, molybdenum, cobalt) rich and contain carbides, while dendrites become depleted in chromium and richer in cobalt, (2) dendrite formation and relatively large grain sizes that decrease yield strength, and (3) casting defects [43]. More recently, low-carbon wrought versions of cobalt-based alloys have excellent mechanical properties and corrosion resistance and tend to be stronger than cast alloys [42,44].

Table 4. Properties of materials used for Hip Prosthesis.

Name of material	Property/Grade	Young's Modulus E(GPa)	Fatigue Limit s_{end}(MPa)	Ultimate Tensile Strength s_{UTS}(MPa)	Yield Strength s_y(MPa)	Elongation (min.%)	Density (g/cm^3)	Hardness (Hv)	Poisson's ratio	Bending strength (MPa)	Reference
Stainless steel	316	190	240–820	515	203	40	8.02	155	0.25	—	[30,31]
		193	260	619	310	35	8	275–340	0.3	—	[32,86]
	316L	190	260	860	685	12	8	225	0.25	—	[38,86]
		190	260	503	195	40	8	199	0.3	—	[38,86]
		190	260	603	294	35	8	199	0.3	—	[31,87]
Co-based alloys	Cast CoCrMo	280	208–950	660	448–517	10	7.8	298	0.3	—	[38]
	Wrought CoCrMo	210	207–310	858	448–648	30	9.15	239	0.3	—	[38]
		210	586	1500	1606	9	9.15	445	0.3	—	[30]
		232	600–896	1000	965–1000	12	8.3	280	0.3	—	[31]
	Wrought CoNiCrMo	210	207–310	794–1000	240–655	50	8.1	—	0.3	—	[50]
		232	689–793	1794	1585	8	9.2	—	0.3	—	[30,31]
	Wrought CoNiCrMoFe	210	586	1515–1794	1606	2–4	8.5	—	0.3	—	[38,50]
		232	689–793	1862–2273	1500	1.0–17	8.3	—	0.3	—	[50]
	Wrought CoNiCrMoWFe	210	207–310	600	448–517	50	8.3	—	0.3	—	[38,41]
		210	586	1172	1606	12	8.3	—	0.3	—	[38,50]
Ti and its alloy	Grade1	107	300	240	170	24	4.5	122	0.34	—	[19]
	Grade2	105	425	345	275	20	4.51	145	0.37	—	[50]
	Grade3	107	240	450	380	18	4.5	280	0.36	—	[36,50]
	Grade4	103	250	550	485	15	4.5	280	0.39	—	[19,50]
	Ti6Al4V	116	620	860	795	10	4.43	349	0.342	—	[51]
	Ti13Nb13Zr	64	—	1030	900	15	4.66	245	0.3	—	[36]

Ceramic	Alumina	375	—	350	—	—	3.9	2000–3000	0.22	379	[36,88]
	Zirconia	150–199	—	200–495	—	—	5.9	1000–3000	0.3	500	[88]
	Pyrolytic carbon	18–28	—	280–555	—	—	1.7–2.2	—	—	—	[89,90]
	Bioglass-ceramics	22	—	56–83	—	—	—	—	—	—	[64,90]
	Calciumphosphates	40–117	—	69–193	—	—	—	—	—	—	[89,90]
	Graphite(LTI)	20–25	—	—	—	—	1.5–1.9	—	—	—	[65,89]
	Vitreous Carbon	24–31	—	70–207	—	—	1.4–1.6	150–200	—	—	[89,64,88]
	Bioactive HAP	73–117	—	120	—	—	3.1	350	—	—	[65,90]
	Bioglass	≈75	—	50	—	—	2.5	—	—	0.31–4.03	[64,90]
	AW Glass Ceramic	118	—	215	—	—	2.8	679	—	215	[50,88]
Carbon	Glassy	24	—	—	—	—	1.5	—	—	—	[5]
	Graphite	24	—	—	—	—	1.5–1.9	—	—	—	[50]
	Pyrolitica	28	—	—	—	—	1.5–2.0	—	—	—	[89]
Calcium	Calcium phosphate	4.0–115	—	30	—	—	3.16	3.43	0.27	147	[64]
Polymers	Poly (methylmethacrylate) (PMMA)	1.8–3.3	19.3–38.5	77	—	2.5–6	1.12–1.2	10–22	0.4–0.43	148–120	[68]
	Polycaprolactam	2.8	—	28–50	—	80–130	—	11–18	0.39–0.44	85	[50]
	Poly(lacticacid)	1.2–3	22–31.9	2.8	—	2–6	1.02–1.15	10–16	0.38–0.42	108	[65]
	Polysiloxane	Upto0.01	—	>35	—	100–1200	1.05–1.22	2–8	—	66	[65,90]
	Ultra-high-molecular-weight polyethylene(UHMWPE)	0.9–2.7	—	53	17	140–500	0.93	62–66	—	27	[65,68]
	Poly(ethylene terephthalate)	2.2–3.5	30–43.2	28–36	—	50–300	0.95–0.96	9.7–21	0.38–0.43	80	[50,65]
	Polypropylene	1.1–1.6	11–18.2	17–28	—	100–300	0.9–0.91	6–10	0.4–0.45	40	[68,79]
	Polytetrafluoroethylene	0.3–0.7	9–18	30	—	120–345	2.1–2.2	2.7–9.0	0.44–0.47	5–6	[65,79]

Composites	Poorly crystalline carbonate-apatite + tetracalcium phosphate + collagen	0.66–2.24	—	6.08–11	—		—	—	—	—	[91]
	Directmineralized collagencomposite(0-39%calciumphosphate)	0.44–2.82	—	34–53	—	—	—	—	—	—	[92]
	Decalcifiedbonecomposite (10-15%calciumphosphate)	0.68	—	44.87	—	—	—	—	—	—	[92]
	PHB/HAP(30%HAP)	2.52	—	67	—	—	—	—	—	—	[93]
	Polyacrylic acid/HAP(40-70%HAP)	1–1.8	—	—	—	—	—	—	—	—	[94]
	UHMWPE-collagen hydroxyapatite(23-40%HAP)	0.11–0.17	—	11.0–17.0	—	—	—	—	—	—	[95]
	Chitosan-polygalacturonic acid-hydroxyapatite (50%HAP)	2.06	—	—	—	—	—	—	—	—	[96]
	Chitosan/hydroxyapatitecomposite(50%HAP)	1.02	—	74.08	—	—	—	—	—	—	[96]
	Chitosan / hydroxyapatite (70%HAP)	—	—	—	—	—	—	—	—	—	[97]
	Self-hardening chitosan/hydroxyapatite	0.88–4.29	—	—	—	—	—	—	—	—	[98]
	Chemically coupled PE/HAP	—	—	18.34–20.67	—	—	—	—	—	—	[99]
	Biphasic calcium phosphate/polylactic acid	0.296–2.48	—	30–60	—	—	—	—	—	—	[100]
	Polylactic acid/HAP	0.66–2.24	—	—	—	—	—	—	—	—	[101]

4.4. Titanium and its alloys

Ti and Ti based alloys are lighter in weight than the other metals and have good mechano-chemical properties. Ti has poor shear strength, making it less desirable for bone screws, plates and similar applications, however, Ti alloy has an excellent biocompatibility and currently the most widely used [38]. Ti alloys due to the combination of its excellent properties such as high strength, low density, good resistance to corrosion, complete inertness to body environment, enhanced biocompatibility, moderate elastic modulus of approximately 110 GPa are a suitable choice for implant fabrication [24]. Ti and its alloy, also have ability to tightly integrate into bone and other tissues, considerably improves the longevity of the implanted devices, decreasing the risks of loosening and failure. Importantly, good clinical outcome from rough surfaces of Ti and its alloy is resulted due to the good osseointegration between the bone and the implant when compared to smooth-surfaced implants [45]. The mechanical properties of a specific implant depend not only on the type of metal but also on the processes used to fabricate the material and device. Titanium alloy (Ti6Al4V) is widely used to fabricate implants for biomedical applications contains aluminum (5.5–6.5%) and vanadium (3.5–4.5%).It has approximately the same fatigue strength (550 MPa) as of CoCr alloy [46].

4.5. Ceramics

Specially designed ceramics for the repair, reconstruction and replacement of diseased or damaged parts of the body are termed “bioceramics” [47]. Ceramic materials possess several useful properties; such as they exhibit high stiffness, inert behavior under physiological environment, and superior wear resistance as compared with metallic and polymeric bearing surfaces. Additionally, Ceramics may be bioinert (e.g. alumina and zirconia), resorbable (e.g. tricalciumphosphate), bioactive (e.g. hydroxyapatite, bioactive glasses, and glass ceramics), or porous for tissue in growth (e.g. hydroxyapatite-coated metals). However, the brittleness is one of the limiting properties of ceramic materials (see Table 4). Since, the mechanical properties of ceramic materials are highly dependent on their density; small voids left in the implant during processing severely affect their longevity. Ceramics have been used in orthopedic implants, for example, dental crowns owing to their inertness to the body fluids, high compressive strength, and good esthetic appearance [44]. The various biomedical applications of ceramics include replacement of hips, knees, teeth, tendon, ligaments, repair for periodontal disease, maxillofacial reconstruction, augmentation, stabilization of the jaw bone, spinal fusion and bone repair after tumor surgery. By using bioactive ceramics high tissue bonding between ceramic and soft tissues can be achieved.

4.6. Alumina (Al_2O_3)

High density high purity alumina (Al_2O_3) was the first ceramic widely accepted clinically because of excellent corrosion resistance, good biocompatibility, high wear resistance and high strength. The reasons for the excellent wear and friction behavior of Al_2O_3 are associated with the surface energy and surface smoothness. Noiri et al. valuated the biocompatibility of alumina-ceramic material histopathalogically for eight weeks by implanting eye sockets in albino rabbits. The results showed no signs of implant rejection or prolapse of the implanted piece. Interestingly, fibroblast

proliferation, vascular invasion and tissue growth was noted in the pores of the implant after eight week [48]. The main source of high pure alumina (aluminum oxide, Al_2O_3) is bauxite and native corundum. The American society for testing and material (ASTM) specifies alumina for implant application should contain 99.5% pure alumina and less than 0.1% combined with SiO_2 and alkali oxides (mostly Na_2O).However, the strength of polycrystalline alumina depends on its grain size and porosity. Generally, smaller the grains lower the porosity and higher the strength. The ASTM standards (F603-78) require a flexural strength greater than 400 MPa and elastic modulus of 380 GPa (see Table 4). Aluminum oxide has been used in the area of load-bearing hip prostheses, dental implants, and orthopedics for more than 25 years [43] while single crystal alumina has been used in orthopedics and dental surgery for almost 20 years. Alumina is usually a quite hard material; however, its hardness varies from 20 to 30 GPa, which permits its use as an abrasive (emery) and bearings for watch movements [44]. Interestingly, aluminum oxide hip prostheses with an ultra-high molecular weight polyethylene (UHMWPE) socket have been claimed to be better device than a metal prostheses with a UHMWPE socket [49].

4.7. Zirconia (ZrO_2)

Zirconia is one of the biomaterial that has a bright future because of its high mechanical strength and fracture toughness. Zirconia ceramics have several advantages over other ceramic materials due to the transformation toughening mechanisms operating in their microstructure that can be expressed in components made out of them. The research on the use of zirconia ceramics as biomaterials commenced about 20 years ago and now zirconia is in clinical use in total hip replacement, however, the developments are still in progress for application in other medical devices. Interestingly, in the current scenario, the main application of zirconia ceramics is THR ball heads fabrication and replacement [50]. Pure zirconia can be obtained from chemical conversion of zircon ($ZrSiO_4$), which is an abundant mineral deposit. Zirconia has a high melting temperature (T_m = 2953K) and chemical stability with a = 5.145 Å, b = 0.521 Å, c = 5.311 Å and β = 99º14’ in [5]. It undergoes a large volume change during phase changes at high temperature in pure form; therefore, a dopant oxide such as Y_2O_3 is used to stabilize the high temperature (cubic) phase. 6 mole% $Y_2O_3$is used as dopant to make zirconia for implantation in bone [51]. Zirconia produced in this manner is referred to as partially stabilized zirconia [52]. However, the physical properties of zirconia are somewhat inferior to that of alumina (see Table 4).

4.8. Pyrolytic carbon

Good compatibility of carbonaceous materials with bone, other tissue and the similarity of the mechanical properties of carbon to those of bone indicate that carbon is an exciting candidate for orthopedic implants [53]. Unlike metals, polymers and other ceramics, these carbonaceous materials do not suffer from fatigue. However, their intrinsic brittleness and low tensile strength limits their use in major load bearing applications. The mechanical bonding between the carbon fiber reinforced carbon and host tissue was developed three months after intra bone implantation and is accompanied by a decrease of the implant strength [54]. Shi et al., studied the thromboembolic rates in the mitral and aortic positions of omni-carbon valve constructed entirely of pyrolyticcarbon [55]. They found a total of 569 aortic omni-carbon valves had thromboembolic events of 0.5% and a total of 298 mitral

omni-carbon valves had a thromboembolic rate of 1.6% [56]. Zimmerman et al, studied the compatibility of filamentous carbon fibre speculated that it does not corrode and elicit almost no foreign body response and is also an efficient electrical conductor *in vivo* [55].The increased mechanical properties of pyrolitic carbon depend mainly on the aggregate structure of the material. Graphite and glassy carbon have a much lower mechanical strength than pyrolitic carbon. However, the average modulus of elasticity is almost the same for all carbons. The strength of pyrolitic carbon is quite compared to graphite and glassy carbon (see Table 4). Pyrolitic carbon can be deposited onto finished implants from hydrocarbon gas in a fluidized bed at a controlled temperature and pressure. The anisotropy, density, crystallite size and structure of the deposited carbon can be controlled by temperature, composition of the fluidized gas, the bed geometry, and the residence time (velocity) of the gas molecules in the bed. The microstructure of deposited carbon should be highly controlled, since the formation of growth features associated with uneven crystallization can result in a weaker material, however, silicon (10 to 20 w/o) is co-deposited (or alloyed) to increase hardness for applications requiring resistance to abrasion, such as heart valve discs. Recently, the success was achieved in depositing pyrolitic carbon onto the surface of blood vessel implants made of polymers. This type of carbon is called ultra-low temperature isotropic (ULTI) carbon instead of low temperature isotropic (LTI) carbon and is thin enough not to interfere with the flexibility of the grafts.

4.9. Bioglass and glass ceramic

In the last few decades various authors have been used numerous bioglass and ceravital glass ceramics. Usually, silicon oxide based glass ceramics with or without phosphorous pentoxide was used for implantation. In the early 1960s, Glass ceramics are polycrystalline ceramics fabricated by controlled crystallization of glasses developed by S D Stookey of Corning Glass Works. These were first utilized in fabrication of photosensitive glasses with small amounts of copper, silver, and gold are precipitated by ultraviolet light irradiation. These metallic precipitates helps to nucleate and crystallize the glass into a fine grained ceramic which possesses excellent mechanical and thermal properties [57]. A common characteristic of such bioactive materials is a modification of the surface that occurs upon implantation. Bonding to bone was first demonstrated for a range of bioactive glasses, which contained specific amounts of SiO_2, CaO, and P_2O_5 [58]. Bioglass has been widely used for filling bone defects; however, the porosity of bioglass is beneficial for resorption and bioactivity [59]. The interface reaction was interpreted as a chemical process, which includes a slight solubility of the glass ceramic and a solid-state reaction between the stable apatite crystals in the glass ceramic and the bone [60].

4.10.Calcium phosphate ceramics

Calcium phosphate has been used in as an artificial bone. This material has been synthesized and used for fabrication of various styles of implants, and solid or porous coating on other implants as well. Calcium phosphate can be crystallized into salts such as hydroxyapatitie and β-whitlockite depending on the Ca: P ratio, presence of water, impurities, and temperature. Different phases of calcium phosphate ceramics are used depending upon whether a resorbable or bioactive material is desired. One of the main characteristics of calcium phosphate is porosity, which provides ideal pore size for bioceramic similar to that of spongy bone [61]. The prime requirement for calcium

phosphate material to be bioactive and bond to living bone is the formation of a bone like apatite layer on their surface [62]. However, the major drawback to use ceramics and glasses as implants are their brittleness and poor tensile properties. Although they can have outstanding strength when loaded in compression but fail at low stress, when loaded in tension or bending. The wide variations in properties of polycrystalline calcium phosphates are due to the variations in the structure and fabrication processes. The calcium phosphate can be calcium hydroxyapatite or β-whitlockite, depending on the final firing conditions during fabrication. Hydroxypatiteused as biomaterial has excellent biocompatibility. It appears to form a direct chemical bond with hard tissues. On implantation of hydroxypatite particles or porous blocks in bone, new lamellar cancellous bone forms within 4 to 8 weeks [63,64].

4.11.Polymeric materials

A polymer is a substance composed of molecules characterized by the multiple repetitions of one or more species of atoms linked to each other in sufficient amount to provide a set of preferred properties. Polymeric materials have a wide variety of applications for implantation since they can be easily fabricated into many forms, such as fibers, textiles, films, rods, and viscous liquids. These have a close resemblance to natural tissue components such as collagen. In some cases it is possible to achieve a bond between synthetic polymers and natural tissue polymers. However, polymers tend to be too flexible and too weak to meet the mechanical demands in orthopedic surgery and they may absorb liquids and swell, leach undesirable products (e.g. monomers, fillers, plasticizers, antioxidants), depending on the application and usage. Moreover, the sterilization process (autoclave, ethylene oxide, and Co-irradiation) may affect the polymer properties. Several polymers have been used for orthopedic applications such as acrylic, nylon, silicone, polyurethane, UHMWPE, and polypropylene (PP) (see Table 4) [65, 66].

4.12.Poly (methyl methacrylate), PMMA

PMMA is a hard fragile polymer that emerges to be inapt for most clinical applications; however it has numerous critical attributes. It can be prepared under ambient conditions so that it can be manipulated in the working theater or dental clinic, explaining its use in dentures and bone cement. The relative accomplishment of numerous joint prostheses is reliant on the performance of the PMMA cement, which is prepared intra-operatively by blending powdered polymer with monomeric methylmethacrylate, which forms dough that can be positioned in the bone.

4.13.Ultra high molecular weight polyethylene (UHMWPE)

UHMWPE is one of the most preferred polymers as an orthopedic implant because of its high mechanical strength, low wear rate, and good biocompatibility. Much research is moving ahead in examining the wear properties of UHMWPE. It is used as the bearing surface in total joint arthroplasty, and it has found 90% success rates in 15 years with metal on polyethylene. However, submicron particles found in periprosthetic tissues when polyethylene wear present [67,68]. The mechanical properties of polymers depend on several factors, including the composition, structure of the macromolecular chains and molecular weight. The wear resistance of UHMWPE can be

improved by increasing crystallinity and cross linking density [69].Although, cross linking improves wear resistance, but at the same time it also degrade tensile strength, fracture toughness and fatigue crack propagation resistance [70,71].Meanwhile, increasing crystallinity of UHMWPE also improves elastic modulus and resistance to fatigue fracture resistance [72,73].

4.14.Composite

Composite is encompassed of two or more metals, polymer or ceramic structures which are separated by an interface. Composite materials have been widely used for a long time in innovative technological applications due to their superior mechanical properties. Some synthetic composites can also be used to produce prosthesis, able to simulate the tissues, to compromise with their mechanical behavior and to restore the functionality of the damaged tissues and structures. Many matrix and reinforcement components of composite materials have been tried by several researchers in tissue engineering to advance the mechanical features, biological functions and to deliver special implants. Biocompatible polymers have been mostly applied as matrix for composite materials associated with ceramic fillers in tissue engineering. Although ceramics are generally stiff and brittle but polymers are known to be flexible and exhibit low mechanical strength and stiffness [74,75].Composite materials are fabricated from various material combinations with different mechanical properties, resulting in structures with superior behavior as compared to structures made of alone [76]. This is normally achieved by the application of a flexible resin reinforced by stiff fibers. The commonly used resins are thermoplastic and thermosetting polymers. Thermoplastic polymers have a good biocompatibility due to a good intermolecular bond, which can be increased by cross-linking. Poly (sulfone) (PSU) PEEK [77], polyaryletherketone (PAEK) [78] and poly-etherimide (PEI) [79] have good mechanical properties, low water absorption and can be sterilized due to the chemical resistance. Thermosetting polymers, such as epoxy resin, allow for more sophisticated products due to lower viscosity during manufacturing, although proper selection of the epoxy resin and total curing of all monomer is determinative with respect to the biocompatibility and in-vivo durability [80].The composite structures offer the possibility to adjust the mechanical properties of an implant not only by means of geometrical changes but also by the design of the material. More or less, the material will be simultaneous developed with the structure by selecting the proper material combinations, fiber alignment and volume fraction the stiffness of the prosthesis can be adapted within a large range. Combination of carbon fibers in a poly(etheretherketone) (PEEK) or Poly(sulfone) (PSU) resins can have a stiffness ranging from 1-170 GPa [81] and strength ranging from 80 MPa to 2.13 GPa [76]. Presently not many composite orthopedic prostheses have been implanted, but many researchers have been explicitly working [82,83]. Composite materials can also be broadly classified based simply on the matrix material used such as the polymer-matrix composites (PMCs), ceramic-matrix composites (CMCs), or metal-matrix composites (MMCs) and carbon/carbon composites (CCCs). Recently, PMCs are the most commonly preferred class of composites. There are important medical applications of other types of composites which are indicative of their great potential in biomedical applications [84,85].

5. Conclusion

In this study, reviewed different metallic, ceramic, polymeric, composite and natural materials used in design of elements for the orthopaedic hip implant. The mechanical and material issues are imperative in the design, selection and fabrication of materials to plan bioprostheses. Ceramics are attractive biological implants due to their good biocompatibility while Alumina with high mechanical strength produce negligible tissue reaction, nontoxic to tissues and blood compatibility tests were also adequate could be a good candidate as well. Moreover, carbon with alike mechanical properties of bone is an exciting candidate due to good blood compatibility, no tissue reaction and non-toxicity to cells. The accessibility of an extensive variety of polymers significantly affected the growth of tissue engineering and controlled drug delivery technologies. Innovations in the composite material design and fabrication processes are raising the possibility of realizing implants with improved performance. Therefore, there is a need to develop more refined multi-functional materials in order to match both the biocompatibility and mechanical complexity of the hip implants. However, for effective application, surgeons must be persuaded with the long term durability and reliability of composite biomaterials.

In the future, we can anticipate to see novel biomaterials developed that will increase the span of orthopaedic implants life. Therefore, it is vital to accentuate the need for precise studies that will determine the behavior of these novel materials prior to their clinical use and determining an approach to improve the biocompatibility (i.e. biological reactions) that occur instantly after implantation. However, close alliance between orthopaedic surgeons, biologists and engineers is vital in order to attain success with the challenging future of joint replacements.

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Conflict of Interest

All authors declare no conflict of interest.

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