the first stage cover screw, and second stage permucosal extension.

2.2 Influence of loading on Bone

Bone is a structural foundation for a load carrying implant [3]. Trabecular bone is a porous, structurally anisotropic, in homogeneous material. The hard outer layer of bones is composed of tissue, due to its minimal gaps and spaces called as compact bone. Compact bone may also be referred to as dense bone. Its porosity is 5to30%. Trabecular bone accounts for the remaining 20% of total bone mass but has nearly ten times the surface area of compact bone. Its porosity is 30–90%.

Bone density is directly related to the strength and elastic modulus of bone. In denser bone, there is less strain under a given load compared with softer bone. As a result, there is less bone remodelling in denser bone compared with softer bone under similar load conditions [18].

Bone has different strength and stiffness depending on direction of the load. It is weaker when loaded under an angled force. The greater the angle of load, the greater the stresses to the implant bone interface. Angled loads increase the amount of shear loads to the bone as shown in Fig.2, and the bone is weakest to shear type loads[19].

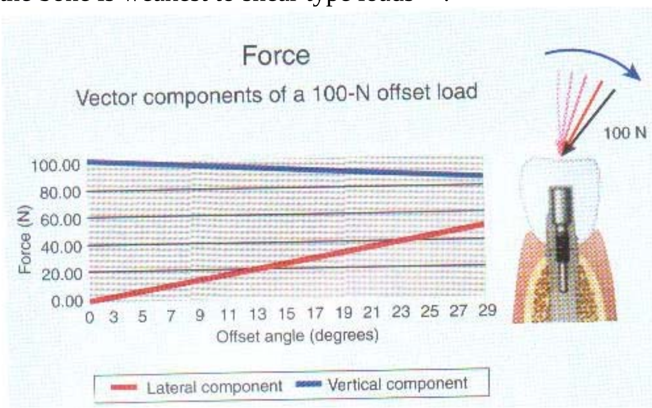

Figure 2: The Force Applied to an Implant Body with an Angled Load[19].

The noxious effect of angled loads to bone is further exacerbated because of the anisotropy of bone. Anisotropy refers to how the character of bones mechanical properties, including ultimate strength, depends on the direction in which the bone is loaded. Bone is strongest when loaded in compression, 30% weaker when subjected to tensile forces, and 65% weaker when loaded in shear as shown in Fig.3.

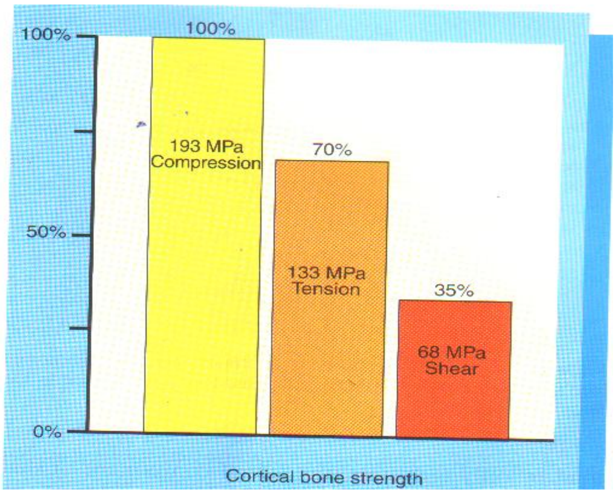

Figure 3: Bone is Strongest Under Compression Forces, 30% Weaker to Tensile Forces, and 65% Weaker to Shear Forces[19].

Cortical bone can withstand more stress but less strain. Trabecular bone can undergo more strain before fracturing. Dense cortical bone is 10 times stronger than the soft, fine trabecular bone. D2 bone is approximately 50% stronger than D3 bone. The stiffness of the bone is affected by the bone density. Youngs modulus for compact bone is 10 times larger than cancellous bone. The denser the bone, the stiffer the bone, and the less biomechanical mismatch to titanium during loading. The mechanical properties of the trabecular and cortical bone found within the oral environment exhibit a high degree of variation as a function of load direction, rate and duration. The structural density of the bone has a significant influence on its modulus of elasticity and ultimate strength.

Finite element analysis of mandibular bone around implants indicated an increase in stresses and strains because of anisotropy. A compressive and shear anisotropy of 3% and 1% in cortical bone and 40% and 30% for cancellous bone respectively, increased stresses by 20% to30% in the cortical crest. Although tensile and radial hoop shear stress increased by threefold to fourfold in the cancellous bone along the lingual side, anisotropy decreased radial vertical interface shear stress by 40% on the buccal side near the apex in the cancellous bone[20]. Carter and Caler[21] have described bone damage or fracture caused by mechanical stress as the sum of both the damage caused by creep or time dependent loading and cyclic or fatigue loading and the relative interaction of these two types of damage ie creep and fatigue strength. Creep refers to the phenomenon whereby a material continues to exhibit increasing deformation as a function of time when subjected to a constant load. Carter and Caler[22] have reported the creep fracture curve as shown in Fig.4, for adult human bone at a constant stress of 60 Mpa. Fatigue strength of a material refers to an ultimate strength below which the material may be repetitively subjected for an infinite number of cycles without failure.

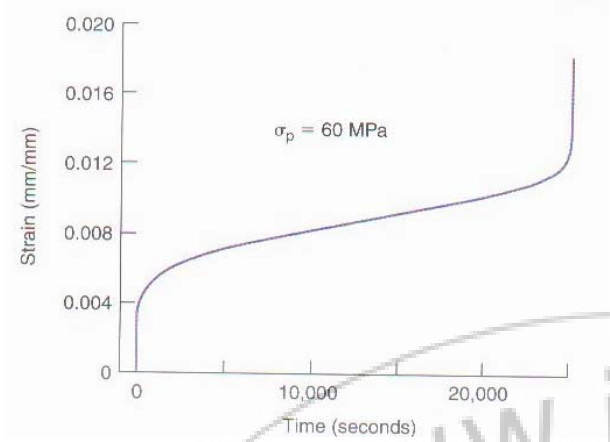

Figure 4: Creep Curve for Adult Human Cortical Bone at Constant Stress of 60Mpa[22].

Carter [23] have investigated the fatigue properties of human cortical bone. Fatigue failure has been reported for in vivo bone at relatively low cycles (10^4 to 10^8 cycles) [24]. Excessive cyclic loading on bones is known to cause microcrack growth and increase fracture risk[25]. Fatigue failure of cortical and trabecular bone has been characterized by a continuous reduction in modulus, with increasing number of cycles with a drastic drop closer to failure and increasing plastic strain[26].

Cortical bone has been observed to behave in an increasingly nonlinear form with the cyclic energy dissipation increasing with number of cycles in both tensile and compressive cyclic loading[27]. Large variations have been noted in experimental measurements of elastic modulus and ultimate compressive strength of trabecular bone. Occasional overloading of trabecular bone (upto 3% strain) degrades its mechanical properties and increase the risk of fracture[28]. The structural fraction of cortical bone is relatively stable. Outer portion of the cortex, the metabolic fraction is the highly reactive inner aspect as shown in Fig.5. The primary metabolic calcium reserves of the body are found in trabecular bone and the endosteal half of the cortices.

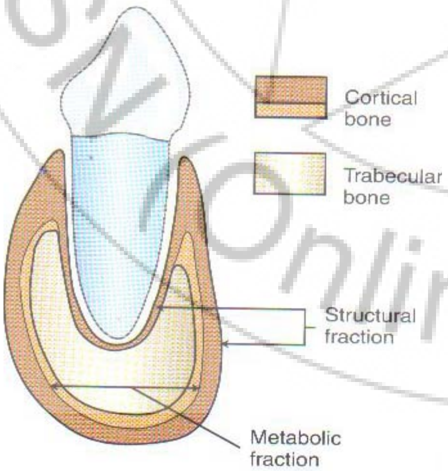

Figure 5: Structural and Metabolic Fractions of Bone in the Mandible[38].

In engineering terms, cross-sectional rigidity is related to the second moment of the area. The rigidity of the bone increases as the fourth power of diameter. Therefore when relatively rigid materials i.e. bone is doubled in diameter the stiffness increases 16 times. Structurally the mandible is modified tubes an optimal design for achieving maximal strength with minimal mass[37]. Within limits, loss of bone at the endosteal surface or within the inner third of the compacta has little effect on bone rigidity. The inner cortex can be mobilized to meet metabolic needs without severely compromising bone strength. The biomechanical response to altered function and applied loads depends on the metabolic status of the patient. Bone metabolism is an important aspect of clinical medicine that is directly applicable to implant dentistry.

Biomechanical manipulation of bone is the physiologic basis of stomatognathic reconstruction. Implantology is bone manipulative therapy and favourable calcium metabolism is an important consideration. Because of the interaction of structure and metabolism, a thorough understanding of osseous structure and function is fundamental to patient selection, risk assessment, treatment planning and retention of desired dentofacial relationships[38,39].

2.3 Effect of Loading on Biomaterials

The functional aspects of biomaterials use include the transfer of force from the occlusal surfaces of the teeth through the crown, bridge and neck connector region of the implant into the implant for interfacial transfer to the supporting soft and hard tissues. This represents a very complex series of chemical and mechanical environmental conditions. The critical aspect of biocompatibility is dependent on the basic bulk and surface properties of the biomaterial. The disciplines of biomaterials and biomechanics are complementary to the understanding of device based function.

The physical, mechanical, chemical, and electrical properties of the basic material components must always be fully evaluated for any biomaterial application, because these properties provide key inputs into the interrelated biomechanical and biological analyses of function. It is important to separate the roles of macroscopic implant shape from the microscopic transfer of stress and strain along biomaterial- tissue interfaces. The macroscopic distribution of mechanical stress and strain predominantly controlled by shape and form of the implant device. One important material property related to design (shape and form) optimization is the elastic strain (one component of elastic modulus) of the material. The higher the applied load, the higher the mechanical stress and therefore the greater the possibility for exceeding the fatigue endurance limit of the material. In general, the fatigue limit of metallic implant materials reaches approximately 50% of their ultimate tensile strength[29,30]. Ceramic materials are weak under shear forces because of the combination of fracture strength and no ductility, which can lead to brittle fracture. Metals can be heated for varying periods to influence properties, modified by the addition of alloying elements or altered by mechanical processing such as drawing, swagging or forging, followed by age or dispersion hardening, until the strength and ductility of the processed material are optimized for the intended application. A general rule is in mechanical process hardening procedures results in an increased strength but also invariably correspond to a loss of ductility. This is especially relevant for dental implants.

Corrosion is a special concern for metallic materials in dental implantology because implants protrude into the oral cavity where electrolyte and oxygen compositions differ from that of tissue fluids. The pH can vary significantly in areas below plaque and within the oral cavity. This increases the range of pH that implants are exposed to in the oral cavity compared with specific sited in tissue[31]. Plenk and Zitter[32] state that galvanic corrosion could be greater for dental implants than for orthopedic implants. Galvanic process depend on the passivity of oxide layers, which are characterized by a minimal dissolution rate and high regenerative power for metals such as titanium. Laing, willert and lemuns[33] have extensively studied the corrosion of metallic implants. Fontana and greene[34] have presented many of the basic relationships specific to implant corrosion. Mears[35]addressed concerns about galvanised corrosion and studied the local tissue response to stainless steel and cobalt chromium- molybdenum (Co-Cr-Mo) and showed the release of metal ions in the tissues. Williams[36] suggested that three types of corrosion were most relevant to dental implants has effect of stress corrosion cracking, galvanised corrosion and fretting corrosion. The combination of high magnitudes of applied mechanical stress plus simulations exposure to a corrosive environment can result in the failure of metallic materials by cracking where neither condition alone would cause the failure.

Most traditional implant body designs under 3D finite element stress analysis show a concentration of stresses at crest of the bone support and cervical third of the implant. This tends to support potential stress corrosion cracking at the implant interface area. Galvanic corrosion occurs when two dissimilar metallic materials are in the contact and are within an electrolyte resulting in current flowing between the two. Fretting corrosion occurs when a micro motion and rubbing contact occurs within a corrosive environment e.g. The perforation of the passive layers and shear directed loading along adjacent contacting surfaces. It has been to occur along implant body abutment superstructure interfaces. Dental implants are typically fabricated from titanium or its alloy. The modulus of elasticity of titanium is five to ten times greater than that of cortical bone. An engineering principle called the composite beam analysis states that when two materials of different elastic modulus are placed together with no intervening material and one is loaded, a stress contour increase will be observed where the two materials first come into contact[40].

3. Conclusion

A growing field of research is implant biomechanics due to the fact that many aspects of implant treatment are based on biomechanical principles. Research has been done on the different parameters which affected biomechanics of dental implant. The prognosis of dental implant treatment is related to the influence of mechanical load on the biomechanics of implant supported prosthesis. Further study could aim at understanding how different engineering techniques employed for evaluating mechanical and biomechanical behaviour of implant.

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