CHAPTER 1

INTRODUCTION

1.1 Shape memory alloys

Shape memory alloys (SMAs) are a group of smart metallic material which appeared to render a wide range of applications, due to their special thermo-mechanical behavior. They are eminent from conventional metallic material by their capability to recover its original shape after large deformations, when material is subjected to adequate thermal process. By the property of remembering their original shape, SMAs are called "memory" materials. The fundamental reason for such behavior of these alloys is associated to crystallographic phase transformation induced by stress or temperature and several alloys represent this behavior. Due to large deformation behavior these alloys are capable for fulfilling functions such as sensors and actuators. SMAs are widely used in domains and direct applications in numerous areas such as biomedical field, aviation industry, aerospace field, atomic-nucleation industry. The sensing application in which the material converts the mechanical input into non mechanical output. Conversely, in actuation application these alloys convert the non-mechanical variation to the mechanical output. SMA actuation implies that during the phase transformation process of a SMA component, large loads and/or displacements can be generated in a relatively short period of time.

Actuation application of an active material depends on many factors; two main key features are actuation energy density, which is known as the available work output per unit volume, and the actuation frequency. To work with a supreme activator material one should have highest actuation energy density and actuation frequency. However, in naturally occurring

actuators and man-made actuators, shape memory alloys possess the highest actuation energy density and the frequency (Figure 1.1 and Figure 1.2).

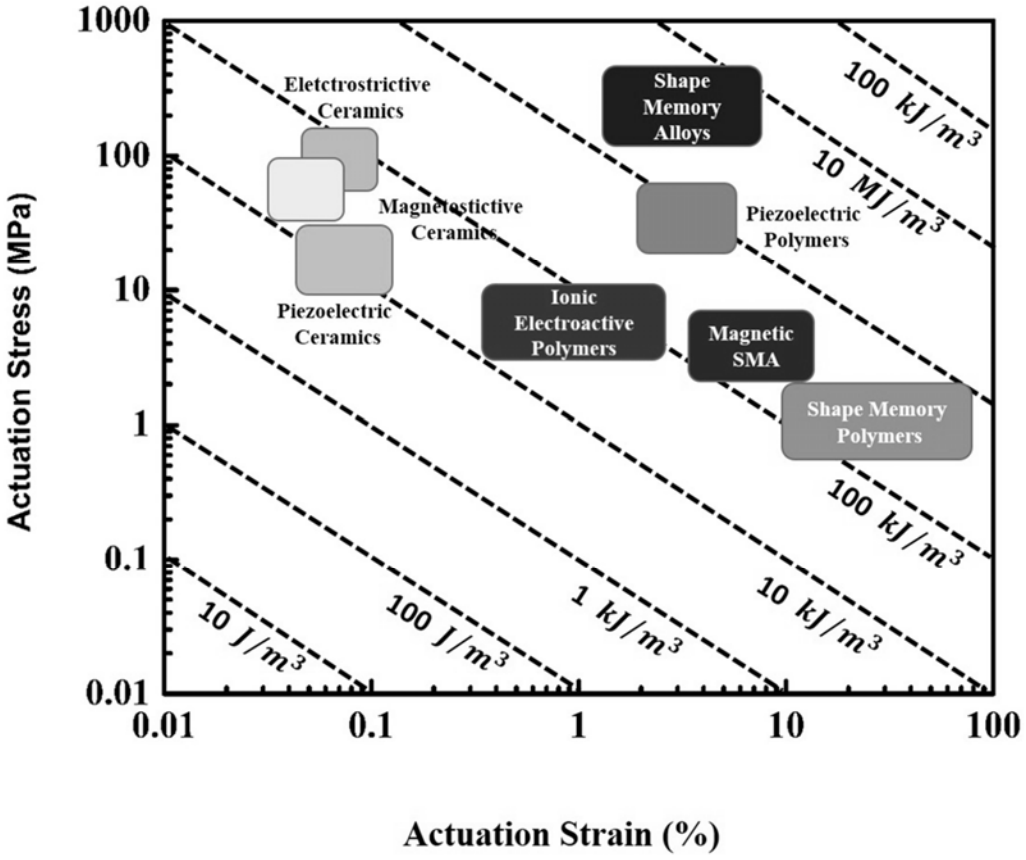


Figure 1.1: Illustration of typical strain, stress, and the actuation energy densities of different material (Lagoudas, 2008)

Since, shape memory alloys have the function of both actuator as well as a sensor, they are very suitable candidates for miniaturization of actuators such as micro actuators or micro-machines or human body implants. SMAs are a very specific type of smart materials which shows unusual properties, that is, after it is deformed the material can recover back to its original shape or in other way they can recover strain upon heating of material.

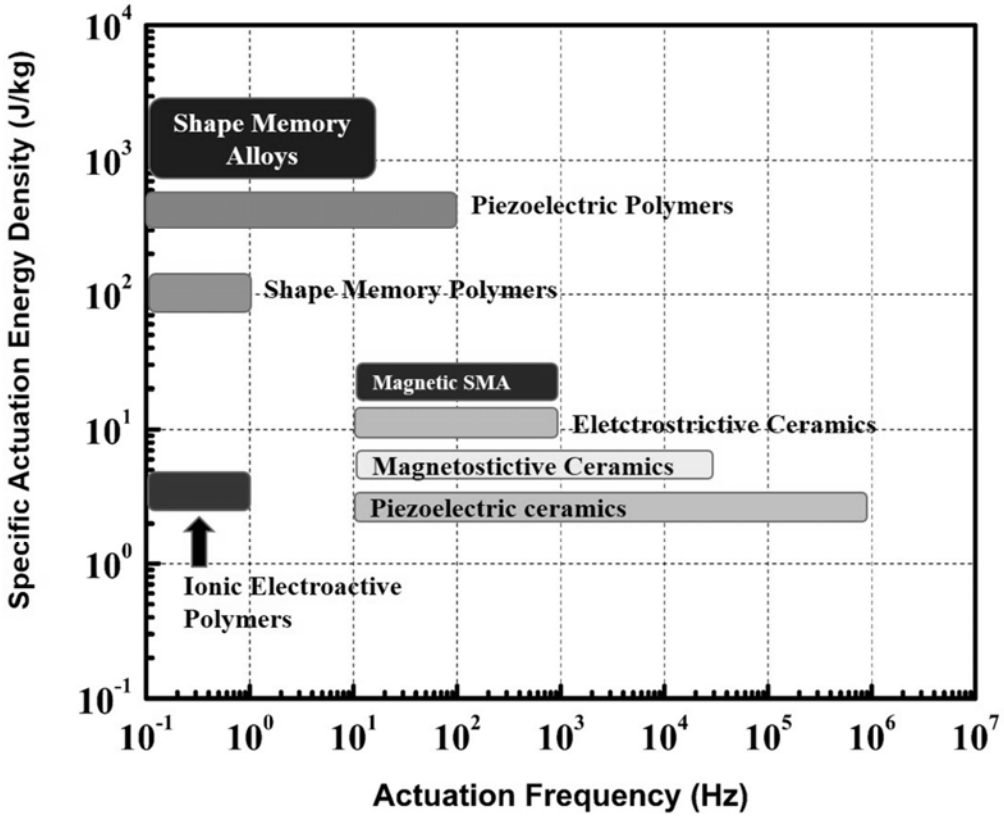


Figure 1.2: Typical actuation frequency chart of different actuators used for direct coupling (Lagoudas, 2008)

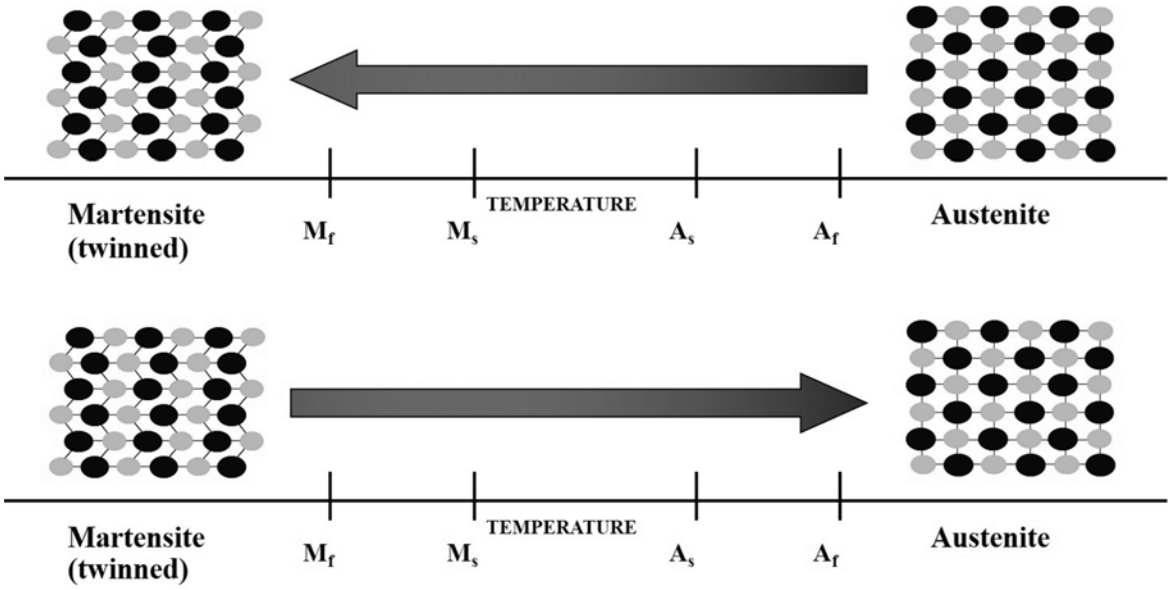
This unusual behavior of SMA is called shape memory effect (SME). First report on SME of shape memory alloy material was published by Buehler and Wiley at U.S Naval Ordinance Laboratory (Buehler et al., 1963). SME effect was first seen in NiTi alloy (Nitinol). SME property in shape memory alloys, makes the energy density high. Likewise, under particular conditions, the material experiences a hysteresis reversible transformation

which empowers the material to dissipate energy for energy absorption applications. These properties make shape memory alloys a decent contestant for a variety of uses in biomedical field. However, low frequency response of the material is a disadvantage which limits its applications.

Among different SMAs, NiTi alloy is one of the most widespread and implemented alloy for biomedical application for its two unique properties; shape memory (SM) and superelasticity (SE), in addition with a higher range of biocompatibility, good resistance with corrosion, wear and unique mechanical properties roughly equal to the bone mechanical properties.

1.1.1 Thermomechanical phenomena and phase transformation behaviors

Deformation behavior of shape memory alloys distinctively shows a strong temperature dependency. SMAs have two solid phases in normal working temperatures: the low temperature stage, or martensite (M) and the high temperature stage, which is known as austenite (A). The martensitic change is a solid state diffusionless phase transformation, in which molecules move agreeably, and frequently by a shear-like mechanism. Normally the parent stage (a high temperature stage) is cubic, and the martensite (a lower temperature stage) has a lower symmetry. The change is schematically shown in Figure. 1.3. At the point when the temperature is brought down below some critical one, martensite transformation begins by a shear-like mechanism, as shown in the Figure 1.3.



Characteristic temperatures: M_f = Martensitic Finish, M_s = Martensitic Start, A_s=Austenitic Start A_f= Austenitic Finish

Figure 1.3: Thermally induced phase transformation in SMAs

The martensite crystal can likewise be formed as twinned or detwinned or as reoriented by the agreeable movements of atoms. The martensites in twinning or detwinning have a similar structure, yet the orientations are distinct. These are known as the correspondence variations of the martensites. Since the martensite has a lower symmetry, numerous the event that it is crystallographically reversible, the martensite reverts to the original orientation. This is the cause of shape memory effect (SME) as shown in Figure 1.4 (a) (b). variations can be formed from a similar parent phase. Presently, if the temperature is raised and the martensite ends up noticeably unstable, the reverse transformation happens, and it is difficult to compress in the space accessible the boundless measure of new data about shape-memory alloys. Rather, in this thesis an effort is made to assess our present

understanding of the two essential mechanism of shape memory alloys; shape memory effect and superelasticity.

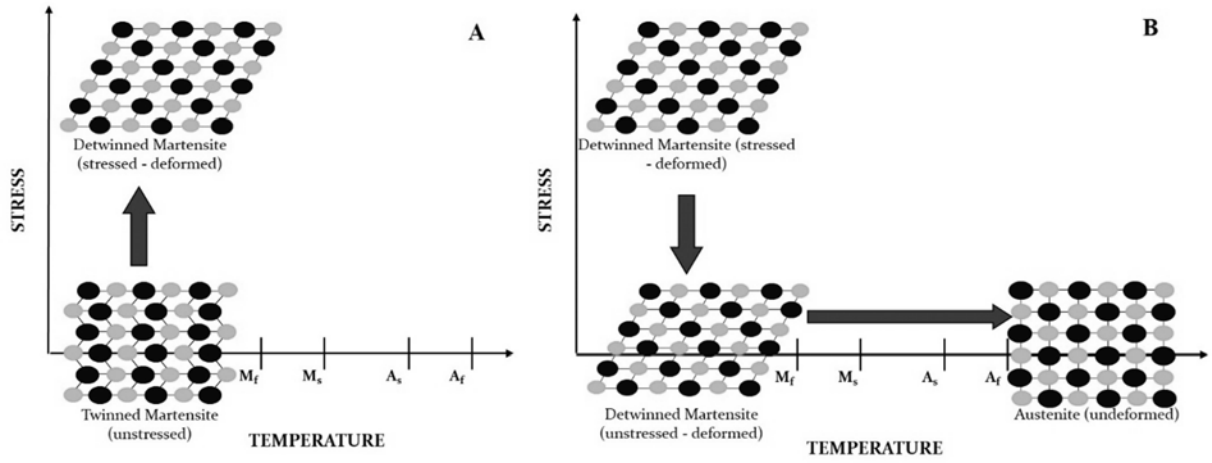


Figure 1.4: A simple model of martensitic transformation: Stress free shape recovery on the application of heat

1.1.2 Shape memory effect

Shape memory effect phenomena permits the alloy to come back to its previous shape, recovering from large strains through the increase of temperature by retaining an apparent plastic deformation. The formation of directional and solid interatomic bonds is responsible to pull back the deformed structures to their previous positions. The complete mechanism of shape memory effect can be better understood by the stress-strain-temperature space diagram as shown in figure 1.5. The diagram shows the typical behavior of shape memory effect of NiTi alloy.

Let us take a chance to begin from point B, the material starts from the origin of plot with a specific temperature and zero stress. At first instance when stress is increased, the

material behavior is linearly elastic, at that point apparent plastic deformation begin to start at almost constant stress. Here, shape memory effect is based on martensite phase transformation and effect occurs at low temperature; at point B material crystallographic structure is composed of twinned martensite. When the shape memory is loaded, the applied stress reaches its critical value (σ_s), reoriention process is initiated, resulting the completion of detwinning process, at this point the critical value of stress is (σ_f), the value of stress at which the detwinning process is completed is slightly higher with the starting stress level (σ_s). Toward the end of the plateau, the martensite is totally detwinned (Point C) and additional loading only causes elastic deformation of the new microstructure.

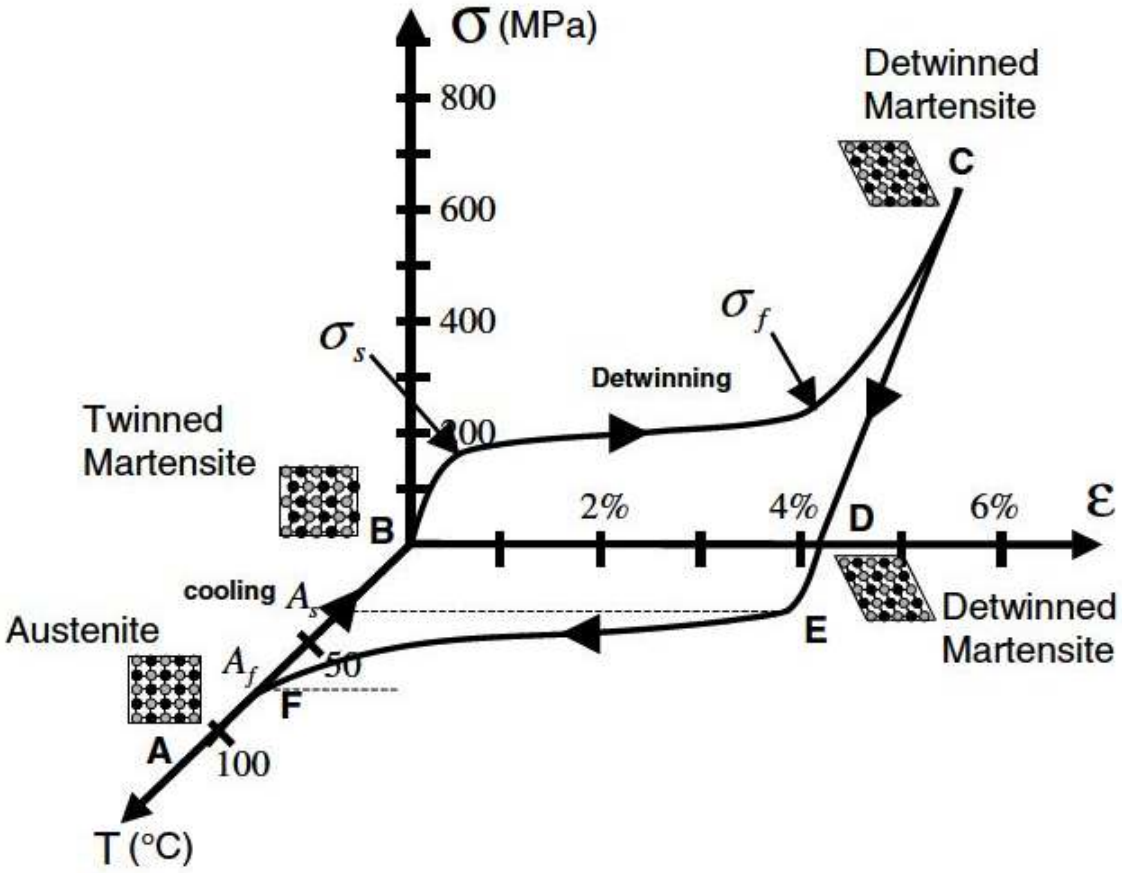


Figure 1.5: Schematic diagram of shape memory effect

When the material is unloaded with applied stress, it takes the linear path up to point D at the same temperature, and the martensite deformation is retained and all the martensite variants are equally stable. At point D the material keeps the residual strain. When material critical temperature is increased by heating the alloy, and the stage at point E, alloy temperature is reached to A_s, which is known as austenite start temperature. At point E, detwinned martensite phase starts transition to austenite crystalline phase. The transformation allows to recover the residual strain and completion of complete residual strain is finished at point F. At point F the temperature is known as austenite final temperature (A_f). Further, increasing the temperature above (A_f), the alloy will reach to the starting phase. Further cooling of material, the transformation from austenite to martensite happens under no load, in this manner no macroscopic change can be observed up to point B.

Shape memory effect is utilized for activation application. The process of actuation is represented by a straightforward case of SMA spring activation as shown in figure 1.6. At first SMA spring is stacked with constant weight at the one end and flip side is fixed. Actuation begins from left to right when the detwinned martensite phase of shape memory spring is extended to its maximum displacement by the placement of constant load. Upon increasing the temperature, by warming the spring to a specific degree, gradual transition from detwinned martensite to austenite phase happens inside the spring. Which empowers the spring to recover its strain, bringing about the progressive contraction of spring. After cooling the spring at this stage, the spring crystallographic structure begins bit by bit to

change from austenite stage to martensite stage, which comes about into the forward transformation and generating strains to transform the spring into the initial prolonged state i.e. detwinned martensite stage. Phase diagram of shape memory effect is shown in figure 1.7.

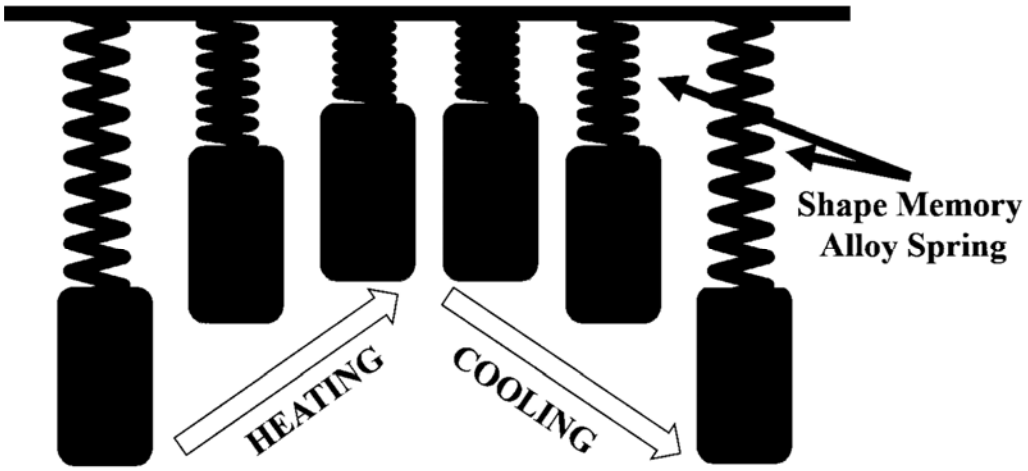


Figure 1.6: Schematic diagram of SMA spring actuation

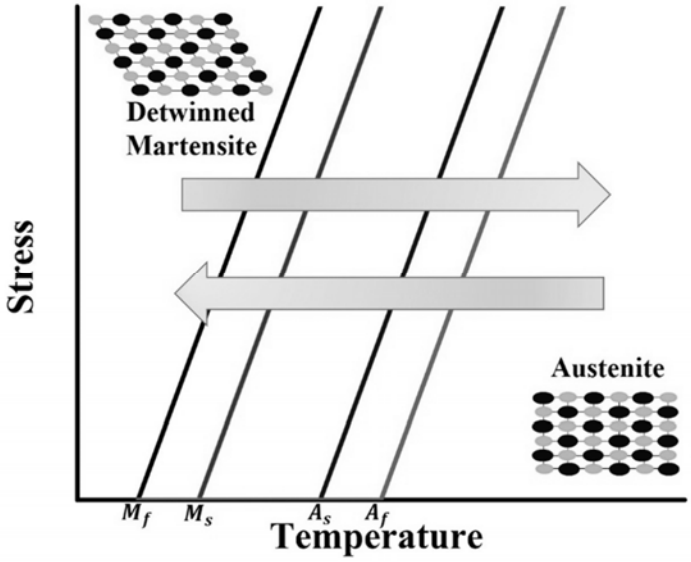


Figure 1.7: Shape memory effect and the associated crystalline changes presented as a phase diagram

1.1.3 Superelasticity (SE) or Pseudoelasticity (PE)

Superelasticity, sometimes called pseudoelasticity, is caused by stress induced martensite transformation from austenite to martensite. Upon loading, strain is generated in the material which is recovered by unloading process. Superelastic phenomena generally occurs above the austenite final temperature (A_f) (Figure 1.8). Superelasticity phenomena is well described by the mechanical loading and unloading of shape memory wire as shown in figure 1.9. Upon mechanical loading of wire the detwinned martensite transformation occurs. Strain is generated during austenite to martensite forward transformation and upon unloading the wire, reverse transformation occurs which turns the martensite crystallographic unstable phase to austenite stable phase and however, the original form is achieved (Figure 1.8 and 1.9). Typical stress-strain curve of superelastic material during loading and unloading is shown in figure 1.10.

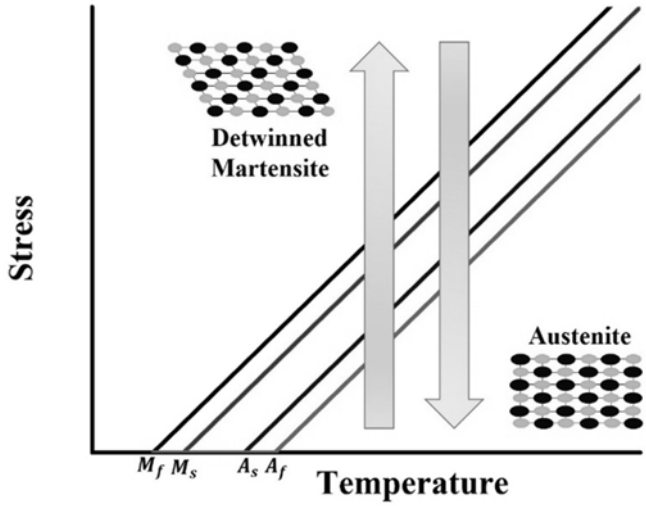


Figure 1.8: Superelasticity and the associated crystalline changes presented in a phase diagram

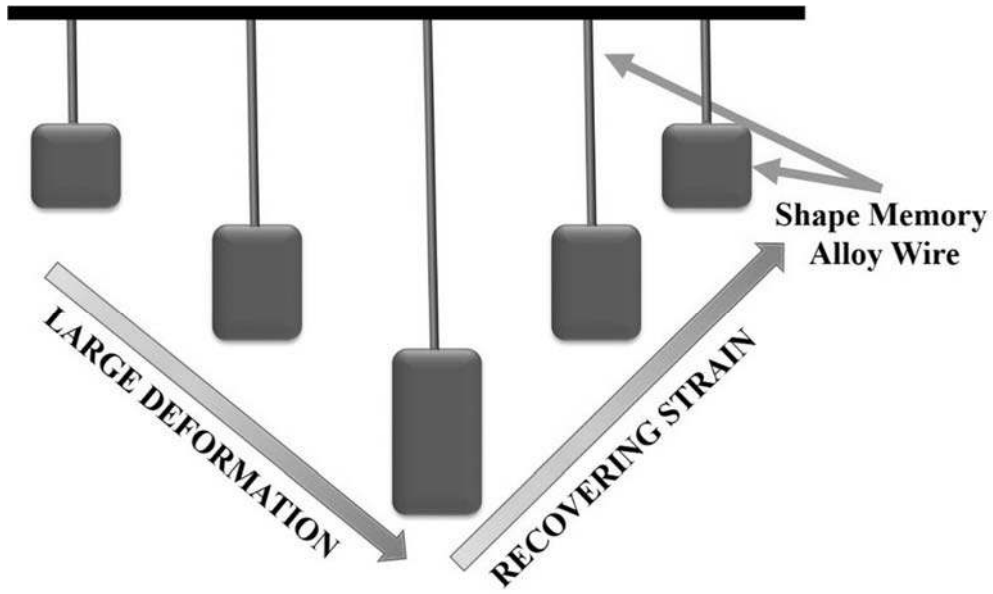


Figure 1.9: Schematic representation of large deformation and recovery of shape memory wire above the austenite final temperature (A_f)

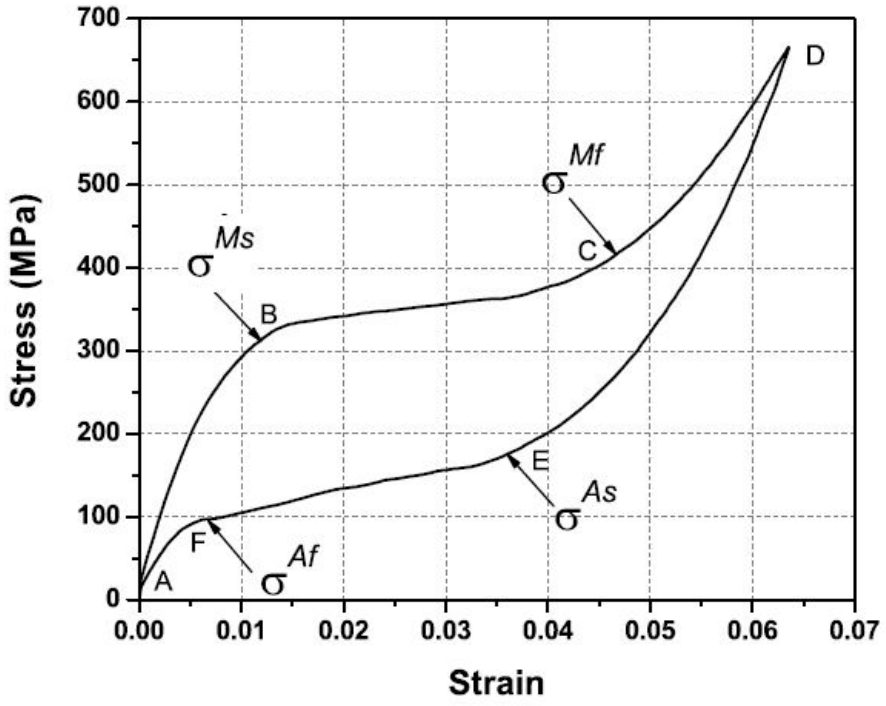


Figure 1.10: Loading cycle of superelastic shape memory alloy (Lagoudas, 2008)

1.2 Shape memory alloy applications and opportunities

In recent years, researchers have been gained interest on the use of shape memory alloy in various field such like aviation, biomedical and so forth. A biomedical literature analysis has been carried out to estimate the research in this field with key word "shape memory alloy" and "nitinol" using PUBMED search engine, which is well known for accessing primarily the MEDLINE database of references and abstracts on biomedical areas; results are shown in figure 1.11.

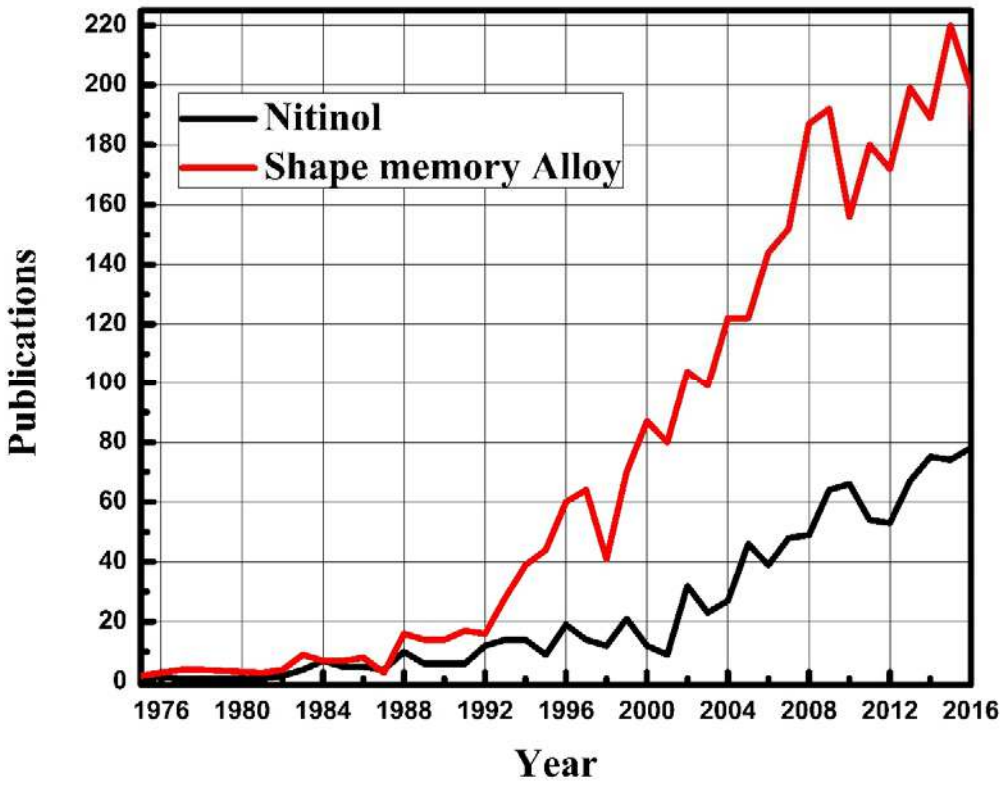


Figure 1.11: Publications of shape memory alloy and nitinol from 1975 to 2016 (Accessed from: PUBMED on April 24, 2017, keyword: "shape memory alloy" and "nitinol")

In this section we will put light on usage of shape memory alloy in biomedical field. Researchers are consistently putting effort to improve and develop prosthetic hand for better execution. Prosthetic hands are furnished with sizable servo engines to supplant the function of hand joints. In any case, these engines are bulky and loud in nature, which makes an improper segment for prosthetic hand. An attempt has been made by Loh et al. in 2005 to overcome with this issue by replacing servo motors with shape memory alloy actuators as appeared in Figure 1.12 (a) (b). In complete assembly the two actuators of shape memory alloy are utilized to actuate the robotic finger, which can practically replicate the actions of the human finger activities (flexion and extension).

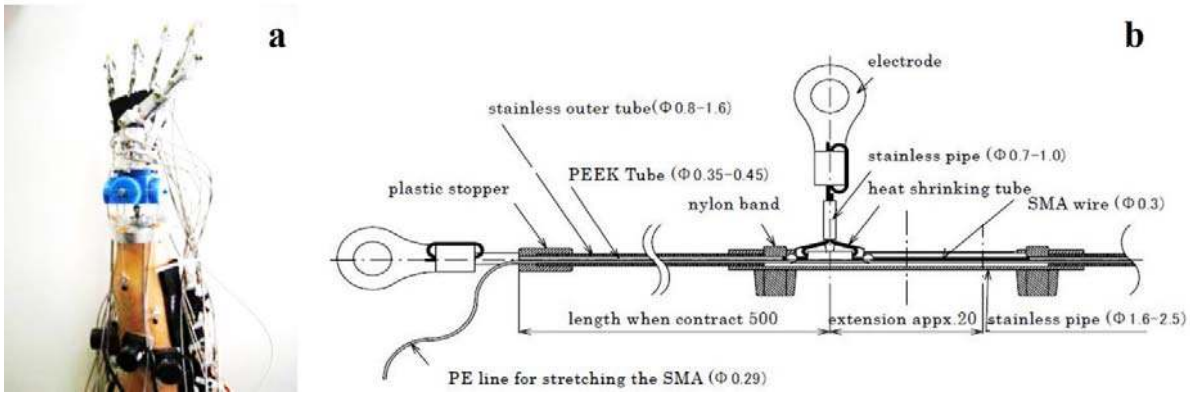


Figure 1.12: Prosthetic hand (a) Prosthetic hand equipped with servo motors (b) Detailed sketch of the shape memory alloy actuator (Loh et. al., 2005)

Prevalence of shape memory alloy application in biomedical devices has been increasing from past few decades, because of their special thermomechanical behavior and its biocompatibility. However, there are many biomedical devices which had been developed

because of the discovery of shape memory alloy and it was extremely hard to develop from different materials like other grade titanium alloy, stainless steel and so forth.

Although, shape memory alloy is significantly expensive from other materials for the use of medical application, shape memory exhibits a wide use in medical industry. In application of shape memory alloys, it can be divided in two categories: active application and passive application. In active application, the temperature of alloy is controlled to induce phase transformation between martensite and austenite, which results in desired thermomechanical behavior. However, passive application corresponds to both the shape memory characteristics: shape memory effect and superelasticity. These phenomena allow the shape memory alloy, especially in the form of NiTi to undergo the mechanically induced transformation and then recovery of original shape.

A famous and most commercialized application of NiTi shape memory alloy in medical field is self-expanding stents. The “stent” word is kept on the name of dentist, Dr. C. T. Stent, who developed the dental device which helps in forming an impression of teeth (Duerig et. al., 1999). Actual definition is made from its functionality as, a tiny tube that is placed inside duct or blood vessel, to keep blocked pathway open. Nitinol is used to develop stents which uses its shape memory and superelastic properties passively, to make them self-expanding after inserting into the body. Stents are shape set according to their use. Normally the external diameter of the shape set form is set 10% bigger from the vessel or duct diameter to guarantee that the stent can't move. These stents are typically made by laser slicing a tube to make the latticed wall. Apart from shape memory effect, the superela-

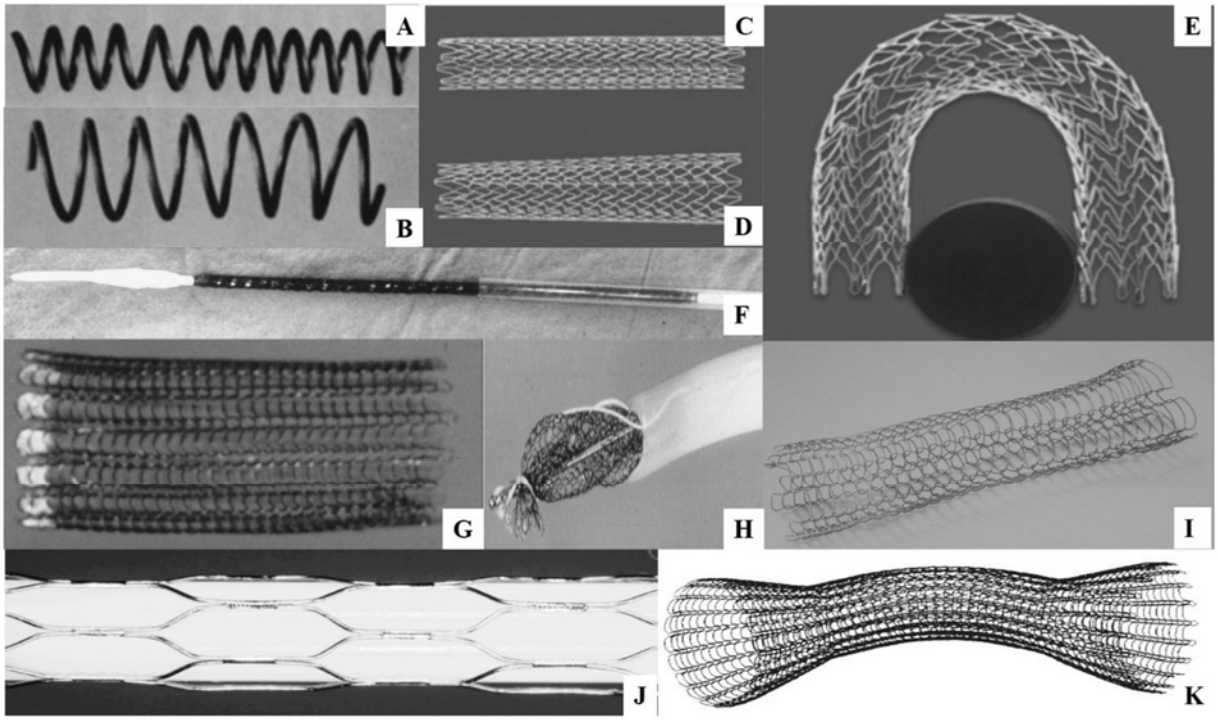


Figure 1.13: Nitinol stents (a) First coil stent in initial form before placement (b) Expended shape after placement in lumina of blood vessels (Dotter et. al, 1983) (c) Initial shape of Abbott Acculink stent made by Abbott. (d) Stent shape after placement, complex pattern and tapered design (e) Illustration of kink resistance of a Nitinol biliary stent (www.abbott.com) (f) Digestive tract stent encased in gelatin and its proper sheath used to treat esophagus anomalies (g) Bunch of self-expanding stents when gelatin has been dissolved with warm, sterile saline solution (h) Procedure to insert the self-expanding nitinol stent through a tracheal tube by flexible tube (Yanagihara et. al, 1997) (i) The memotherm® prostatic stent/bulbar stent used for treating subvesical obstructions in urinary tract (Image courtesy of Bard, Angiomed, Karlsruhe, Germany) (j) Demonestration of Stent actuation with hexagonal matrix design (Boston Scientific Co., Natick, MA) (k) Rectosigmoidal nitinol stent used for treating rectosigmoidal cancer (Tack et. al, 1998).

sticity of the nitinol stents inside the body is another solid point for them, which gives a greater flexibility for about 10-20 times then the stainless steel. There are several stents which are deployed in individual anatomy of human body according to their use as shown in figure 1.13.

Inside the wide group of shape memory alloy, NiTi partners shape memory and pseudo-elastic effects, described by large plateu and stress hysteresis, with the below listed interesting property makes its wide applicability in medical field.

1. Good resistance to corrosion (Trepanier et. al, 1999)
2. Biocompatibility (Ryhnen, 1999)
3. Fatigue resistance (Ryhnen, 1999)
4. Magnetic resonance compatibility (Duerig et. al, 1999)
5. Kink resistance (Duerig et. al, 1999)
6. Dynamic interference (Duerig et. al, 1999)
7. Uniform Plastic deformation (Ezaz et. al, 2013)
8. Hysteresis (Pruski & Kihl, 1993)
9. Constant unloading stresses (Ryhnen, 1999)
10. Biomechanical compatibility (Ryhnen, 1999)
11. Thermal deployment (Duerig et. al, 1999)
12. Elastic deployment (Ryhnen, 1999)

Superelastic phenomena gives the elastic deployment or additional flexibility for nitinol. This feature of shape memory alloy is utilized as a part of performing surgery with small

entry points or small incisions: known as minimally invasive surgery (MIS). MIS has been made conceivable after the assessment of endoscopic instrument. In endoscopic surgery or MIS, littler size of surgical instrument is required, and the superelastic and shape memory effect of SMAs give more flexibility, enhancing viability in narrow cavities. Moreover, superelasticity gives high strain recovery and a wide constant stress plateau over a huge range of strains. These one of a kind SMA attributes have driven, amid the previous couple of decades, to the design and manufacturing of novel improved instruments particularly suited for MIS, examples of which are ablation devices, tongs, suture passers, grippers, deflectable graspers, and scissors. Such upgrades in surgical instrumentation have additionally prompted setups that are significantly more flexible than those obtainable with conventional endoscopic devices, thus giving greater expertise and usability to the surgeons, improving surgical precision. Among the devices that advantages from elastic deployment of nitinol, the RITA (Radiofrequency Interstitial Tissue Ablation) device is shown in figure 1.14. In this device, shape memory alloy's curved tubular needles can be deployed from the straight shape upon insertion.

Thermal deployment is another unique property which is the reason of shape memory effect. Simon vena cava filter is the principal vascular device of the blood vessel in the treatment of blood clot is as of now being supplant by devices presented by means of a peripheral vein (Simon et. al, 1977). It is inserted as a straight thin wire by means of the little bore catheter while the material is in the martensite stage outside the body so it keeps the shape. Passing through the vessels, the flushing chilled saline solution keeps the

catheter in the martensite stage, after reaching the lumen of the vena cava (large blood vessel) and detecting body temperature, it returns to its preset original filter shape and locks into place permanently (Figure 1.15). It will trap blood clot throughout its route. Superelasticity property of nitinol keeps the vessel in open position. This device is frequently utilized for bedridden patients. The comparative technique is likewise utilized for stents.

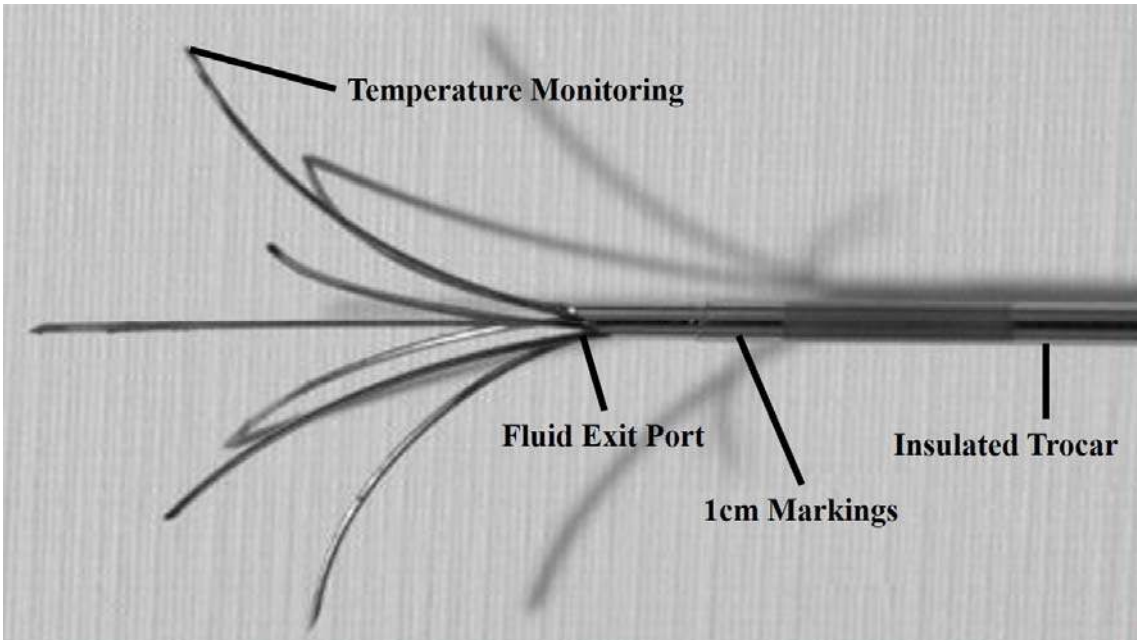


Figure 1.14: RITA multitine electrode system (Rita Medical Systems, Mountain View, California)

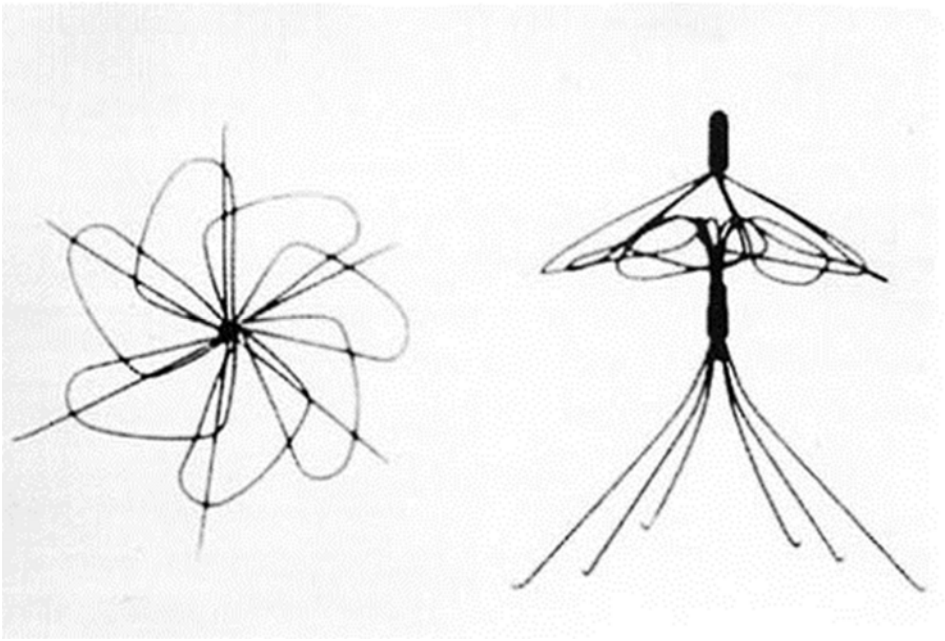


Figure 1.15: Simon vena cava filter used for trapping blood clot in blood vessels (Duerig et. al, 1999)

Another utilization of superelastic properties is kink resistant, because of the pathway of stress-strain curve of nitinol which shows long strain plateau (up-to 8% strain) trailed by quickly increasing stress. The capacity of nitinol to endure large deformations without permanent deformation, which can diminish the steer ability of the device, settles on it a decent choice for the wire material. These features permit bending through convoluted pathways while avoiding strain localization and plastic deformation. An extremely acclaimed medical application of kink resistant nitinol device incorporate retrieval basket, the endovascular radiation probe and the intra-aortic balloon pump (IABP) as shown in figure 1.16. Different device has likewise been developed from this property of nitinol, for example, angioplasty guide wires, these wires are normally long which are utilized to get

to the access of brain. The baskets are made using Nitinol wire and the distal end are shape set into the basket configuration. Later two devices grandstand the advantages of superelastic Nitinol tube and guide-wire, which are utilized to put radioactive material in exact areas close tumors in the body. The balloon pump is utilized to help heart beating in patients waiting for heart surgery. Nitinol grants smaller profiles for the device without worry of tube kinking or buckling upon insertion through vessels in the body.

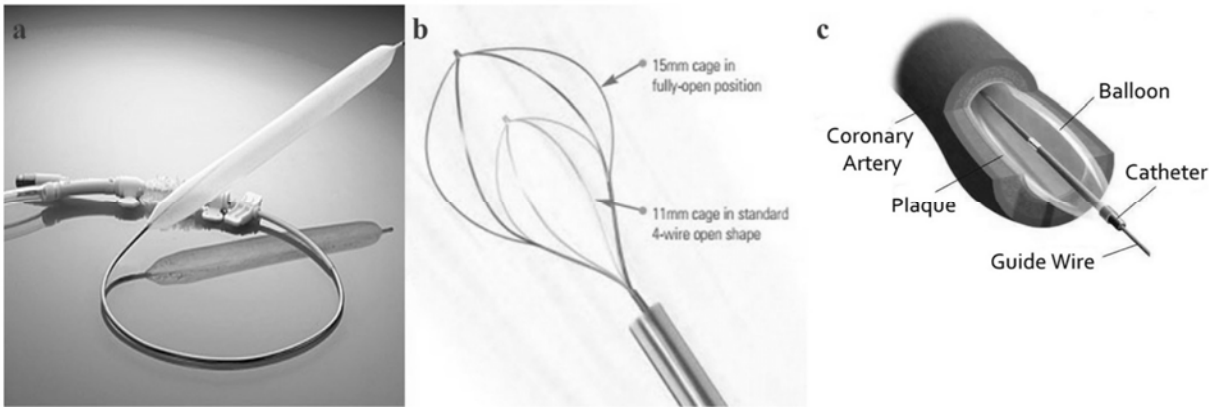


Figure 1.16: Kink resistant of nitinol in medical devices (a) Intra-Aortic balloon pump (Image courtesy: Teleflex, Carrington Mill Boulevard, Morrisville, NC) (b) Nitinol baskets in final configuration during deployment (Kesler et al, 2008) (c) Nitinol angioplasty guide wires (Image from Radiac Abrasives, A Tyrolit Company, Salem, IL)

Biocompatibility is the ability of a material or device to remain biologically inactive during the implantation period. Examines on biocompatibility of nitinol has prompted material of choice for some medical device applications including yet not restricted to endovascular device, cardiovascular stent, orthopedic implant, orthodontic arch wires (Duerig et. al, 1999). The purpose behind a biocompatibility test is to evaluate potential toxicity which

comes out by the contact of the device with the body. The biocompatibility of a nitinol must incorporate the biocompatibility of the material constituents. As nitinol corrode, metallic particles are discharged into the nearby tissues or, then again fluid by a typical mechanism other than corrosion (Autian, 1977). Despite the fact that nitinol contain more nickel than 316L stainless steel, nitinol indicate great biocompatibility, high corrosion resistance in due to naturally formed homogenous TiO_2 coating layer, which has an exceptionally low concentration of nickel. In spite of the fact that nitinol compounds have the corrosion resistance of titanium, the passivated oxide film will dissolve at some rate; moreover, the oxide layer does not give a totally impenetrable barrier to the diffusion of nickel and titanium ions (Williams, 1975). Ptentiodynamics outcomes demonstrate that the biocompatibility of nitinol positions between that of 316L stainless steel and Ti6A14V, even after sterlization as shown in figure 1.17.

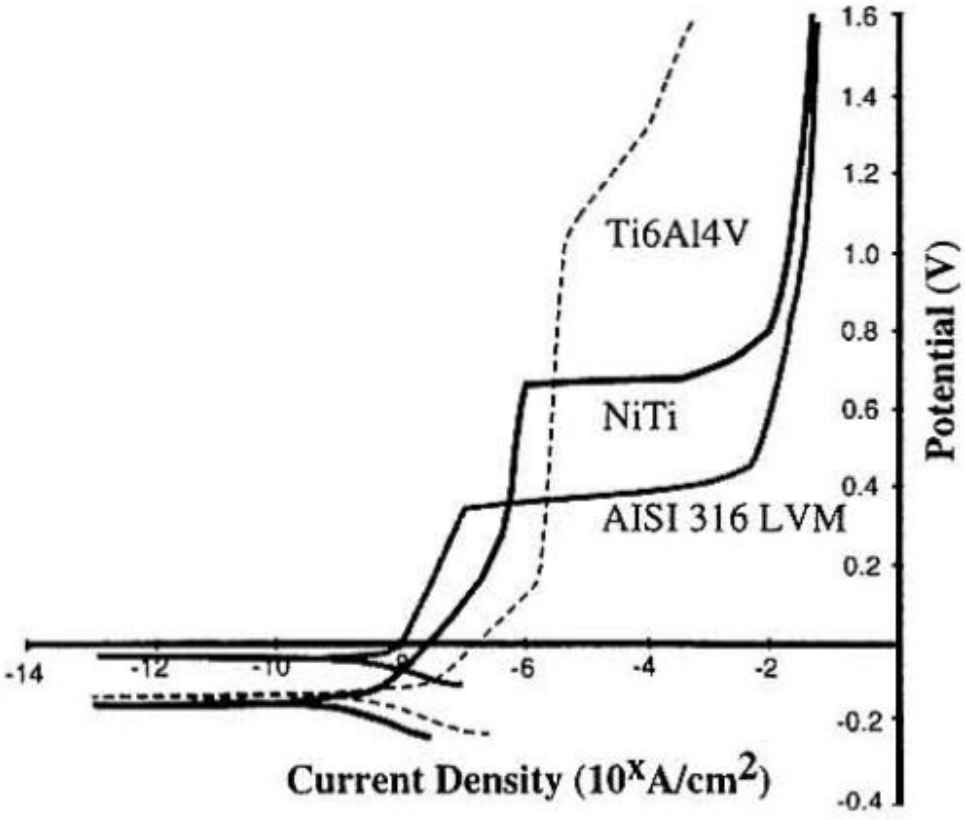


Figure 1.17: Ptentiodynamics result: Ryhanen revealed that nitinol is nontoxic, nonirritating, and fundamentally similar as stainless steel and Ti–6Al–4V alloy (Ryhnen, 1999).

Unlike other materials, nitinol has been found similar mechanical property as of the biological materials as shown in figure 1.18. This quality of nitinol prompts the utilization in orthopedics. Dynamic compression bone plates and hip implants are a popular amongst the most well-known orthopedic uses of nitinol, trailed by intramedullary fixation nails. Fracture healing in long bones can be quickened when bone closures are held in position with compression between the bone fragments. Using this technique, the undesirable surface damage and wear of the openings that happen in a conventional dynamic bone plate

are avoided, while constant compression is guaranteed, regardless of the possibility that bone resorption happens at the fracture sites.

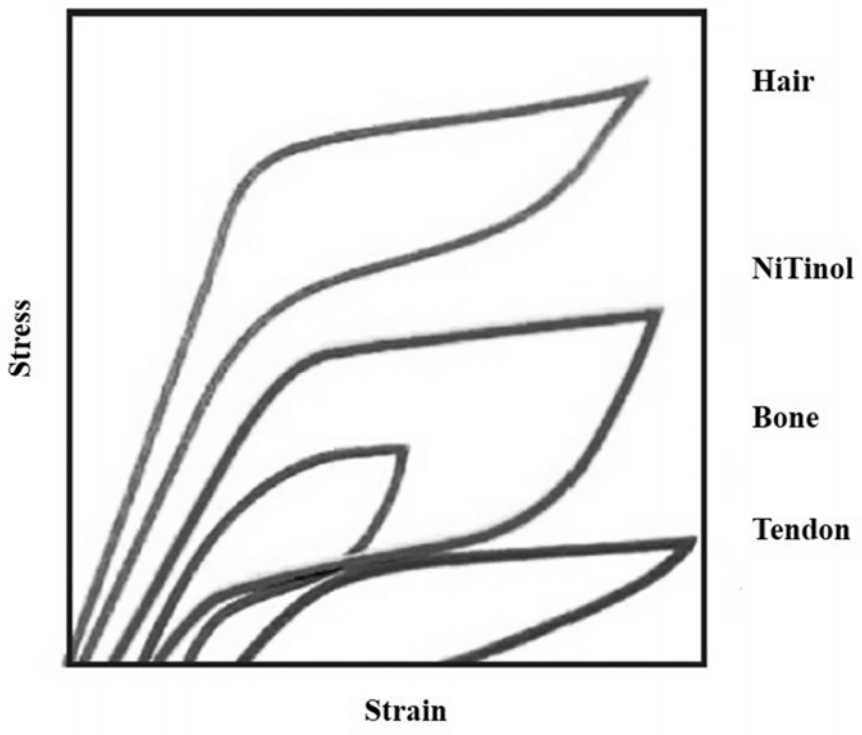


Figure 1.18: Schematic stress–strain curves of Hair, NiTi, bone, and tendon (Stoeckel et al., 2000).

The unusual phase transformation of nitinol identifies the nitinol implant device during magnetic resonance imaging (MR). A spin echo and a gradient echo strategy are utilized to gauge the magnetization changes related with the phase transformation of nitinol. This little change can be recognized by MRI measurements, which are nondestructive for the implant and performed noninvasively. MR images are inclined to artifacts brought by metallic implants: however, nitinol showed less artifact instead of other metallic material

like stainless steel 316L (Melzer et al, 2004). A magnetic resonance image of stent is indicated clearly in figure 1.19.

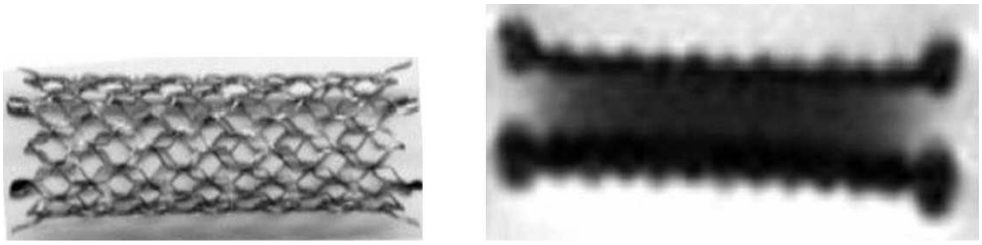


Figure 1.19: Magnetic resonance Image of stent (a) Stent (b) Laser-cut closed cell (Sinus) (OptiMed, Ettlingen, Germany).

1.3 Hip joint and its anatomy

Hip joint is one of the most stable joint in human body, its stability originates from the rigid ball and socket joint of femoral head and acetabulum of the pelvis (Byrd, 2004). The articulating surfaces of femur head and acetabulum makes a stable joint contact for load bearing and provides a range of movement of joint as shown in figure 1.20

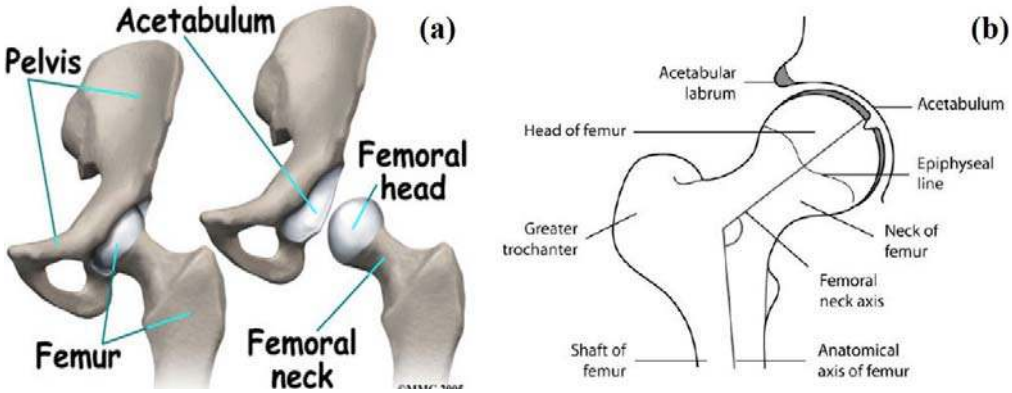


Figure 1.20: Hip Joint Anatomy (a) 3D view of hip joint (b) cross sectional view of hip joint

1.3.1 Hip Joint bones and its mechanical properties

Hip joint is composed of two bones; femur and pelvis. Bones are a type of rigid connective tissue which has either cortical and cancellous bone or spongy bone. Cortical bone is hard in nature that means the density is high ($\simeq$ 1.8 gm/cm^3), it provides the protective layer of these bony structures. Cancellous bone is spongy in nature and provides support to the cortex, density of cancellous bone is lower than the cortical bone ($\simeq$ 0.9 gm/cm^3), spongier nature of cancellous bone helps to reduce the weight of bony structure (Figure 1.21). Mechanical properties of bone are highly dependent on age, gender, location and many pathological reasons.

Since, the bone is highly anisotropic, means its mechanical properties is considered in two orthogonal directions; longitudinal and transverse.

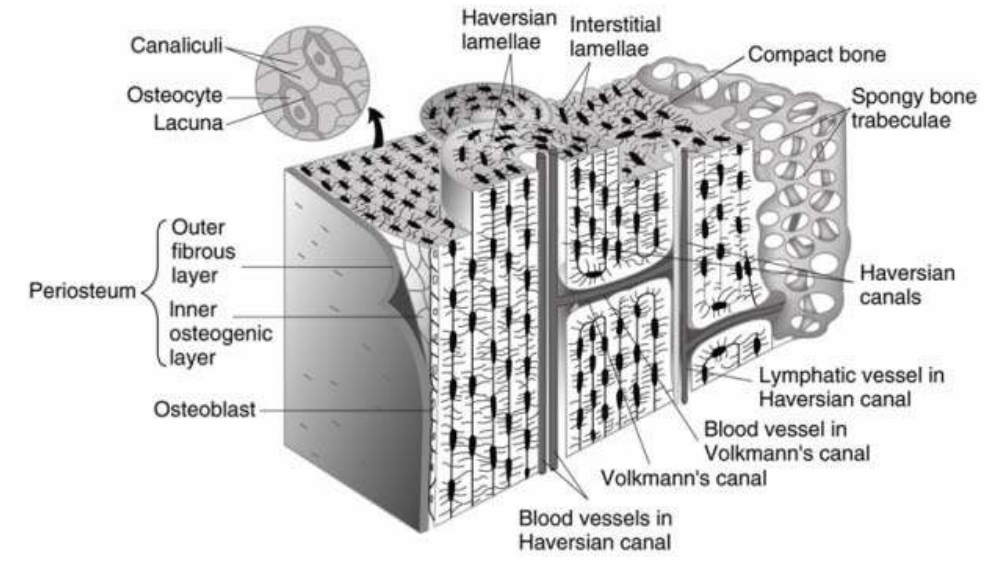


Figure 1.21: Schematic illustration of long bone, showing cortical and cancellous layer (Tortora, 1983)

Human bone consists of mainly 35% of organic and 65% of inorganic substances. Organic substances are composed of cells, matrix of collagen fibers composites and ground substances, whereas inorganic are mineral crystals which includes primarily calcium phosphates. Despite of fact, fiber composites are composed two or more different components, so modulus of elasticity of bone can be estimated by using rule of mixture and inverse rule of mixture which gives the value for cortical bone as transversely isotropic ($E_1 = E_2$ = 11:5 GPa, E_3 = 17 GPa; G_{12} = 3.6 GPa, G_{13} = G23 = 3.3 GPa; υ_{12} = 0.51; υ_{13} = υ_{23} = 0.31) and cancellous bone is considered as isotropic material with (E=2.13 GPa and υ=0.3) (Cowin,1989). These properties were obtained from long bones such as the femur or the tibia, as they are easier to test and their properties are uniformly distributed along the bone.

1.3.2 Total Hip Replacement: Joint Reconstruction

Total hip replacement (THR) also known as Total hip arthroplasty (THA) or Total hip surgery (THS) is a surgical procedure to correct the mobility of hip ball-socket joint function by replacing damaged or arthritic bone and cartilage with metal components (prosthesis) (Postel et al, 1987) as shown in figure 1.22.

1.4 Brief history of THR

The earliest documented attempts to treat trauma and hip disease date back to the 1790s and consisted of amputation. In the beginning of the 19th century, hip deformities were treated by osteotomy and arthrodesis. These methods involved fusing the hip joint, resulting in a joint with no effective mobility (Morrey, 2003).

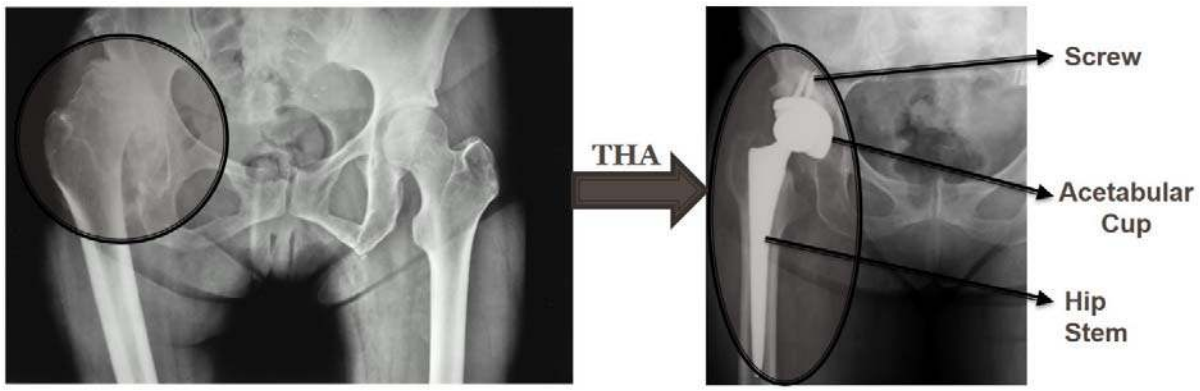


Figure 1.22: Total hip arthroplasty and its component

The 20th century saw the development of modern THR, which was introduced in two major steps. The first step was the development of "Mold Arthroplasty", also known as "Cup Arthroplasty", by the surgeon Smith-Petersen in the 1920s. This technique consisted of interposing material in the joint between the femoral head and the acetabulum. He first used glass and then Pyrex after he observed soft tissue ongrowth on an explanted piece of glass. However, these materials fractured easily in the hip so he turned to Vitallium, an alloy of chrome, cobalt and molybdenum, which was unique with regards to its biocompatibility in living tissue. Only half of the surgeries performed using this Vitallium mould (Figure 1.22) successfully relieved pain, however, this was a major step in the development of acetabular cups in THR (Morrey, 2003).

The second step was the introduction of endoprostheses in the 1940s, which consisted of a femoral head attached to a short stem and implanted in the intertrochanteric region of the femur (Figure 1.23). These implants were initially made of acrylic but, because of wear issues, they were later made of a cobalt-chrome (CoCr) alloy. This change was one of many

modifications made to endoprostheses; however, most of these endoprostheses became loose and failed because of their defective load bearing capacity (Ling, 1989). The small stem was later replaced by a longer intramedullary stem to give the head more mechanical support (Ling, 1989; Morrey, 2003).

Figure 1.23: A Smith-Petersen Vitallium mould arthroplasty from 1939 (left); a Wiles stainless steel endoprosthesis from 1938 (centre); a Judet Brothers acrylic endoprosthesis from the late 1940s. (Source; Reynolds et al., 2007)

In 1938, Philip Wiles performed the first THR by implanting a stainless steel ball-and-socket hip prosthesis, which was attached to the bone with screws and bolts. This prosthesis, as with other similar ones, was prone to mechanical failure. It was not until the 1950s, when Sir John Charnley began his extensive research, that the modern THR was created. His innovations included the use of a dental bone cement called polymethyl-methacrylate to anchor the prostheses, and "Low Friction Arthroplasty", which comprised a stainless steel intramedullary stem with a 22 mm diameter femoral head articulating into a high-density polyethylene polymer cup (Figure 1.24). This was a major turning point for orthopedics as his innovations, which are still used today, could for the first time

successfully treat major hip disabilities (Morrey, 2003; Callaghan et al., 2007; Jones et al., 1987).

Figure 1.24: Charnley low-friction THR with small stem and ultra-high molecular weight polyethylene (UHMWPE) socket from 1962. Picture from (Reynolds et al., 2007)

1.5 Modern approach and developments in THR

Using bone cement was initially a very popular method to anchor the prostheses as any errors in bone resection or reaming could be accommodated (Ling, 1989; Jones et al., 1987; Sakellariou et al., 2013). The short and mid-term results were excellent with a low rate of complications, failures and revisions. However, the long-term results were less satisfactory; high loosening rates and loss of bone stock were commonly reported in early THR, increasing exponentially after 5 years of implantation and particularly affecting younger patients (Morrey, 2013; Ling, 1989; Sakellariou et al., 2013). These problems were thought to be linked to the use of cement. Exothermic reactions were known to occur

during cement polymerisation and methyl methacrylate particles had been discovered in soft tissues surrounding loose implants during revision surgeries. The cause of implant loosening was therefore widely attributed to "Cement Disease" [Jones et al., 1987; Schmalzried et al., 1992; MacInnes et al., 2012; Mjöberg, 1994). In addition to this, several studies revealed that methyl methacrylate undergoes ageing in the human body; with time, the cement becomes more rigid and fragile, and is increasingly affected by creep deformation as its elastic modulus decreases (Jones et al., 1987). Operative and cementing techniques were continuously improved in the hope of reducing implant loosening. Improvements such as thorough cleaning of the bone surface using brushing and pressurized lavage, control of bleeding, vacuum mixing of the bone cement, and the use of cement guns for the introduction of the cement all have resulted in improvements in cementing techniques (Mulroy et al., 1990; Jones et al., 1987; Mjöberg, 1994). Today, using cement for implant fixation remains a viable option and is widely used for older patients, providing a reproducible and cost-effectivemtechnique (Mulroy et al., 1990; Sakellariou et al., 2013; Blaha and Tanifuji, 2011).

As a response to cement disease, a new anchoring philosophy, cementless fixation, was introduced in the 1980s. This included the development and use of new materials that encourage osseointegration, thereby removing the need for cement in THR (Morrey, 2013). Osseointegration is the biological process in which an implanted prosthesis is integrated with the surrounding bone and is achieved through bone ingrowth or bone ongrowth (Morrey, 2013, Ling, 1989). Bone ingrowth relies on the bone growing into the porous

surface of an implant to secure it, while bone ongrowth is provided by the direct adhesion of the bone to the implant (Blaha and Tanifuji, 2011; Watt et al., 2006; Kienapfel et al., 1999).

Implant modularity was another development of THR, which was brought on by cementless fixation, where the femoral head is separate from the stem and the acetabular cup is composed of a shell and a liner (Figure 1.25). Both the stem and the acetabular shell are made of metal, usually CoCr or Titanium alloys, and covered with a porous coating for osseointegration. The head and the liner come in a multitude of materials to form different bearing couples such as ceramic on polyethylene, ceramic on ceramic, metal on polyethylene and metal on metal. Furthermore, modularity allows the surgeon to decide on the implant parameters, such as head size and liner material, during surgery, as well as change a worn out liner or head without removing a well fixed shell or stem, respectively, during revision surgery (Jones et al., 1987; MacInnes et al., 2012; Watt et al., 2006; Young et al., 2002).

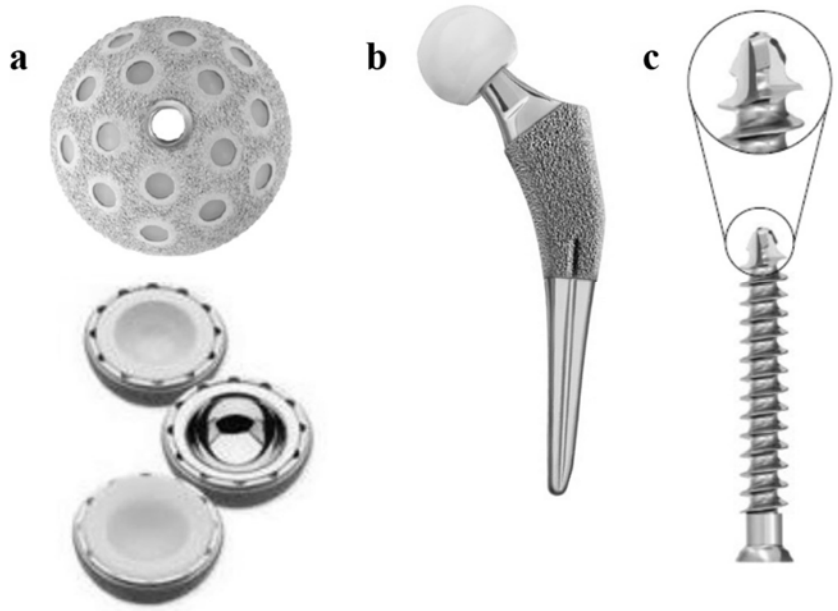


Figure 1.25: Modern total hip prosthesis components a) Acetabular cup with a ceramic, a metal and a polyethylene liner (Courtesy of United Orthopedic Corporation, Taiwan and Exactech Gainesville, Florida, United States) b) Hip stem (Courtesy of Exactech Inc. Gainesville, Florida, USA) c) Acetabular screw (Courtesy of Depuy, Warsaw, IN, USA.)

National joint registries and clinical studies have reported a lower percentage of revisions of cementless THR than cemented THR in younger patients (Figure 1.26) ("Australian Orthopaedic", 2014; "National Joint", 2014; Watt et al., 2006). This difference increases with duration of implantation ("Australian Orthopaedic", 2014; National Joint, 2014; Watt et al., 2006). Hence, cementless fixation has become a popular fixation method for THR (Figure 1.26), especially in younger and more active patients (Morrey, 2013; Jones et al., 1987).

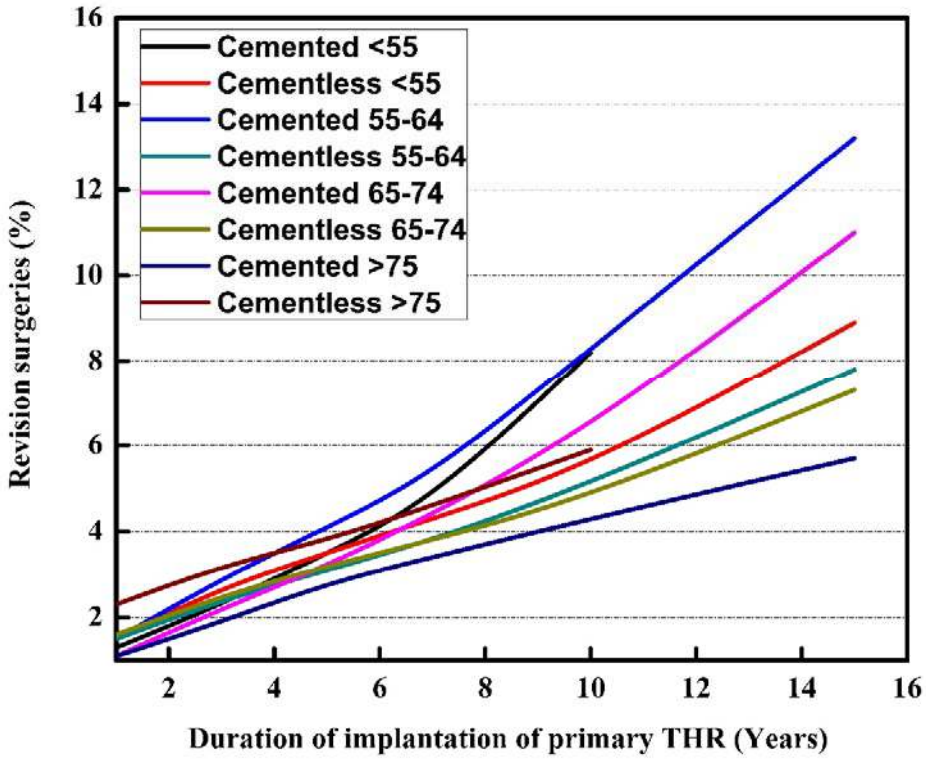


Figure 1.26: The percentage of revision surgeries on both primary cemented and cementless THR in younger patients with respect to the duration of implantation ("Australian Orthopaedic", 2014).

Due to the high success rates of cemented implants using modern cementing techniques and the promising results of cementless implants, hybrid fixation is also available. Standard hybrids comprise a cemented femoral stem and a cementless acetabular cup (Cook et al., 1992), while reverse hybrids comprise a cementless stem and a cemented cup ("National Joint", 2014); Hybrid fixations are used in 15% to 30% of all THR (Figure 1.26); however, the type of hybrid used varies between countries. In England and Wales, 85% of hybrid procedures are standard hybrids ("National Joint", 2014), whilst in Sweden, 87% of these procedures are reverse hybrids ("Swedish Hip", 2013).

1.6 Failure of total hip implants

1.6.1 Various national THR registries report

THR is a successful surgical procedure with excellent results. A number of hip replacement registers are available from different countries. These registers record the reasons for hip failures and can thus provide information on the possible failure criteria to consider during design development. They can only be used as indicator, as they use revision as a criterion for failure, which may be an underestimation as some hips may not be revised owing to poor patient health and patients may be living with painful hips (Sundfeldt et al., 2006). The most recent UK hip register found that the highest rate of failure occurred when large bearing metal on metal press-fit acetabular cups (46 mm diameter) were implanted, where 25% of THR implants had to be revised after 10 years, and 6.3% of hip resurfacings; compared to 2.0% of the patients receiving a cemented cup and 3.4% of those receiving a nonmetal on metal cementless cup (NJR, 2010). This indicates that metal on metal press-fit cups are a good candidate for analysis as they are evidently not performing as intended.

Table 1.1 shows the prevalence of different failure modes in a variety of hip registers. The hip registers clearly show aseptic loosening to be the most prevalent problem. Aseptic loosening describes non-infection related loosening of an implant from bone; the mechanism of failure, but not the cause. A number of theories exist concerning the cause of aseptic loosening and are discussed in next section

Table 1.1: Reasons for implant failures from different national registries

Country	Year	No. THR	No. Revisions	Failure mode	%	Source
USA	2012-2015	169,060	17,180	Infection Dislocation Fracture mechanical loosening Other	18.9 18.1 18.4 4.6 3.1 36.5	AJRR, 2016
UK	2003-2015	796,598	88,822	loosening Pain Lysis Dislocation Infection Bone fracture Implant fracture wear Malalignment Other	51.0 22.0 15.6 15.1 3.3 9.6 3.6 14.1 5.7 8.6	NJR, 2016
Australia	2003-2015	363561	57819	Loosening Dislocation Fracture Infection Wear Other	28.3 21.7 16.2 11.0 2.5 20.3	AOANJRR, 2016
Sweden	1979-2000	207,311	15,960	Loosening Infection Dislocation Fracture(bone) Technical error Fracture(implant) Wear Other	75.3 7.6 5.8 5.1 3 1.5 0.9 0.8	(Malchau et al., 2002)
India*	2006-2015	8246	614	Loosening Infection Dislocation	74 15 11	ISHKS, 2015

*Registered hospitals in this registry are very less so the accuracy of data is very low.

1.6.2 Failure scenarios

There is not total agreement in the cause of aseptic loosening, though it is likely to be the result of a number of different failure scenarios (Sundfeldt et al., 2006). The small percentage of THR that do fail need to be revised. Furthermore, THRs are being increasingly performed on younger and more active patients who therefore outlive the lifetime of their implant. There are various causes for revision surgeries, none of which are mutually exclusive. The four main causes of revision are osteolysis and aseptic loosening, dislocation or instability, infections, and fractures. Huiskes, (1993) developed a framework of failure scenarios to describe the possible processes leading to implant failure: accumulated damage; particulate reaction; failed ingrowth; stress shielding; stress bypass; destructive wear. High fluid pressure (Van der Vis et al., 1998) and surgical error (Ferney et al., 2007) have also been highlighted in the literature as possible causes of failure.

1.6.3 Accumulated damage

Repetitive dynamic loading such as that generated by walking can cause mechanical damage to implant materials and their interfaces. High local stresses can cause cracks and damage which in turn can cause micromotion. Micromotion between the implant and bone can lead to loosening. Failures occur when locally elevated stresses occur in the same place as weak points in either material or interfaces (Huiskes, 1993).

1.6.4 Particulate reaction

Particulate reaction occurs when the implant-bone interface de-bonds owing to the formation of macrophages at the interface. Osteoblasts and osteoclasts are two types of cell which generate and eliminate bone respectively. The balance between the presences of these two cells controls bone remodelling and if this equilibrium is disturbed then either too much bone may be eliminated or not enough bone formed (Sundfeldt et al., 2006). The presence of wear debris is thought to activate osteoclasts (or possible reduce osteoblasts (Van der Vis et al., 1998)) and cause bone resorption (Bauer and Schils, 1999b). Wear debris also stimulates macrophages which absorb the foreign body and contribute to inflammation at the osteolytic site. The bone loss can lead to an area of weakened bone, which may increase risk of fracture. The bone loss and inflammation weaken the interface bond between implant and bone and can lead to loosening. This process is commonly referred to as osteolysis, but some hypothesise that osteolysis may also be caused by high fluid pressures (Aspenberg and Van der Vis, 1998).

Sundfeldt (2006) found that wear particles on their own could not initiate loosening, and that additional factors such as motion and infection had to also be present. The size of particle (Green et al., 1998), the individual (Matthews et al., 2000), the material (McEwen et al., 2005) and manufacturing process (Sundfeldt et al., 2006) all contribute to the inflammatory reaction experienced from wear debris. This means that prediction of implant failure owing to wear is not simply correlated with the amount of wear debris.

Particulate reaction occurs when debris finds its way to the bone-implant interface. Weak points such as cracks in cement, gaps caused by lack of bone ingrowth or screw holes (some

cup designs incorporate screw holes within the cup walls to secure the cup into the surrounding bone) all offer pathways for the debris. Mechanical and hydrodynamic forces may also guide the particles to the interface (Van der Vis et al., 1998)

1.6.5 Failed ingrowth

Press-fit cups rely on the ingrowth of bone onto the implant to fix the cup in place. Without this ingrowth, the cup may migrate, although mechanical interlock due to the press-fit still has mechanical function. Early prosthesis migration (~0.85mm) is a predictor of implant failure (Karrholm et al., 1994) and occurs before any other failure scenario, suggesting its importance. The long term fixation of a press-fit cup is dependent on osseointegration in the initial stages of implantation, but the criteria for initial fixation may differ to that for long term fixation (Bauer and Schils, 1999a).

There are two main approaches for encouraging bone ingrowth: osteoconductive porous coating, where an irregular, rough surface finish provides easy attachment for bone; and osteoinductive Hydroxyapatite (HA) coating which encourages bone to grow. Studies have shown that micromotion greater than 150µm at the implant-bone interface will prevent bone growing into the implant and instead the implant will be surrounded by a fibrous membrane (Pilliar et al., 1986). This magnitude of micromotion has been found in implants with failed ingrowth (Engh et al., 1992). The latter study also showed micromotions of up to 40µm in successfully ingrown prostheses, while Pilliar (1986) found movement up to 28 µm could be permitted. The load transfer through a new prostheses has also been shown to contribute to areas of bone ingrowth (Engh et al., 1993); areas without adequate stresses

lack bone ingrowth onto the implant. Even with successful early bone ingrowth bone resorption can still be initiated through inappropriate loading, infection, debris, movement and hydrodynamic pressure (Bauer and Schils, 1999b).

1.6.6 Stress shielding

The introduction of an implant can redirect the transfer of forces through the bone. As bone remodels itself in response to mechanical loading (Frost, 1987), the distribution of strong and weak bone may therefore change throughout the local trabecular bone. If the implant is very rigid compared to the bone, then the bone in close proximity may become very weak as it is being 'shielded'. Though clearly noticeable on an X-ray, and therefore commonly commented upon, few clinical failures are reported owing directly to stress shielding and it generally stabilizes after 2 years (Huiskes, 1993; Laursen et al., 2007). Weak bone however may contribute to fracture and the potential voids and loss of ingrowth may contribute to osteolysis and loosening (Sundfeldt et al., 2006). Press-fit cups are generally made from stiffer material than cemented cups therefore are more susceptible to stress shielding than cemented UHMWPE cups.

1.6.7 Stress bypass

The phenomenon of stress bypass is similar in concept to stress shielding, but rather than redirecting load away from the implant, a mismatch in stiffness between the reamed bone and implant leads to bone being loaded where it was not before, rather than being shielded. The bone may therefore become overloaded.

1.6.8 Wear

Wear is mechanical process which has been linked to aseptic loosening (Howie et al., 2007) and is directly correlated to use, rather than time. Even if loosening does not occur, wear without loosening is still a problem. Wear of the main bearing surfaces can be predicted and planned for in implant design, but wear of secondary surfaces, such as impingement, can cause accelerated wear. Impingement is influenced by implant design, surgical positioning and patient variables (Malik et al., 2007).

Metal on metal hip replacements have a much lower wear rate than those manufactured from UHMWPE (Anissian et al., 1999), but there has been recent concern about the type of debris produced from a metal on metal bearing. High levels of metal ions have been found in the blood of patients who have metal on metal hip replacements, along with a number of occurrences of fluid-filled masses around the hip joint area (Hart et al., 2009) and tissue necrosis. The severity of a failed metal on metal hip joint has lead to increased attention from the Medicines and Healthcare Products Regulatory Agency and the subsequent withdrawal of two metal on metal resurfacing cups; the ASRTM (DePuy Inc) and the DUROMR (Zimmer Inc).

1.6.9 High fluid pressure

Studies have shown that high fluid pressure (700 mmHg) between implant and bone can cause bone resorption (Van der Vis et al., 1998). There is also evidence that a lack of joint fluid access to the bone is linked to osteoarthritis (Sundfeldt et al., 2006). High fluid pressure is not commonly thought to be a primary cause of bone resorption, despite Van der Vis and Aspenberg's study and hypothesis that high pressure kills osteoblasts, thus preventing new bone generation (Aspenberg and Van der Vis, 1998). High fluid pressures

are more commonly thought to contribute to the transport of wear debris to the bone-implant interface (Sundfeldt et al., 2006).

1.6.10 Surgical error

Misalignment of the implant, failure to 'bottom out' (fully seat the cup within the pelvis) or inadequate reaming could all lead to a number of further problems, such as excessive loading, stress shielding, impingement leading to wear as well as pain and dislocation.

1.6.11 Aseptic loosening

The hip registers highlight aseptic loosening as the most prevalent reason for revision surgery, thus failure of implant. The prevalence of loosening at the femoral stem and acetabular cup is comparable (Havelin et al., 2000). It is evident from Section 2.2.3.2 that loosening can result from a number of different mechanisms. Cement disease, wear particles, stress shielding, micromotion, fixation, high fluid pressure, and individual variations have all been highlighted as causes of loosening (Sundfeldt et al., 2006). It does not seem possible to isolate one single cause of loosening, though it seems plausible that without initial fixation, loosening is inevitable, and that excessive micromotion indicates a lack of initial fixation.

1.6.12 Infection

Infection is a clinical complication resulting from surgery. The mechanisms of failure through infection are physiological and thus are not possible to model using an engineering approach.

1.6.13 Dislocation

Implant dislocation can be linked to implant design. Studies show that higher risk of dislocation is associated with a smaller femoral head (Conroy et al., 2008) as well as the surgeon's positioning of the acetabular cup and the patient's anatomy (Kristiansen et al., 1985).

1.6.14 Fracture (implant)

Fracture of the acetabular cup is rare and normally only occurs in cases of high energy traumatic loads. It has also been linked to cases of fatigue failure from excessive wear and excessive stresses owing to bad fixation and thus unusual load transfer (Wroblewski et al., 1998).

1.6.15 Fracture (bone)

Periprosthetic femoral fracture most commonly occurs after minor trauma such as a low energy fall at the same level (Lindahl et al., 2005). Patients with weak bone are at higher risk, therefore osteoporosis, osteolysis, loosening, age, gender and revision surgery have all been linked to higher rates of fracture (Franklin and Malchau, 2007). Implant design, primarily of the femoral stem can also influence risk of fracture (Franklin and Malchau, 2007).

1.7 Osteoporosis

Osteoporosis is type of disease that affects the bone strength, which causes weakening and loss of bone mineral density (BMD). This disease is termed as "silent killer", as it always progresses without any symptom or pain. The first sign is collapsing of bone with minor

fall or an accident, it occurs specially at the location of hip and spine (Cooper et al., 2011). Osteoporosis occurs less common in men than women, elderly people are affected more with this disease which even cause morbidity and mortality (Raisz, 2005). As per the report published by International Osteoporosis Foundation (IOF), 200 million elderly people worldwide are affected by this disease. Osteoporosis relating the cause of fractures having wide discrepancies between the incidence of men, women and age. Correlation and incidence between sex and age was derived from European Osteoporosis survey as shown in table 1.2 (Staa et al., 2002). Noninvasive diagnosis of osteoporosis level involves many techniques which are listed below:

a) Dual energy X-ray absorptiometry (DEXA) scanning: Level of bone mineral density in bone is measured noninvasively.
b) Quantitative ultrasound (QUS)
c) Quantitative computer tomography scan (QCT)
d) Fracture risk assessment tool (FRAX)
e) The Bayes' theorem
f) The radiological assessment of vertebral osteoporosis: Jikei osteoporosis Grading Scale
g) Single-energy X-ray absorptiometry (SXA)
h) Peripheral dual energy X-ray absorptiometry (PDEXA)
i) SPA: Method which uses a single-energy photon beam
j) Dual-photon absorptiometry (DPA)

k) Radiographic absorptiometry (RA)

l) Magnetic resonance imaging (MRI)

1.7.1 Effect of osteoporosis on implant stability

Stability of prosthesis is main concern for surgeons during surgery. Bone quality plays an important role for implant stability. Poor bone quality is the major factor for implant loosening and migration, postoperatively or at the time of surgery. The adverse effect of osteoporotic bone on the stability of implant (acetabular cup and screw) is shown in figure 1.27

Table 1.2: Age and sex specific fracture occurrence of femur/hip, pelvis and vertebra.

Fracture Location	Women			
	No. of cases	**Rate per 10,000 Person Years**	**No. of cases**	**Rate per 10,000 Person Years**
Femur/hip	5755	5.3	19,179	17.0
Pelvis	1086	1.0	3527	3.1
Radiographic Vertebra	3406	3.2	6195	5.6

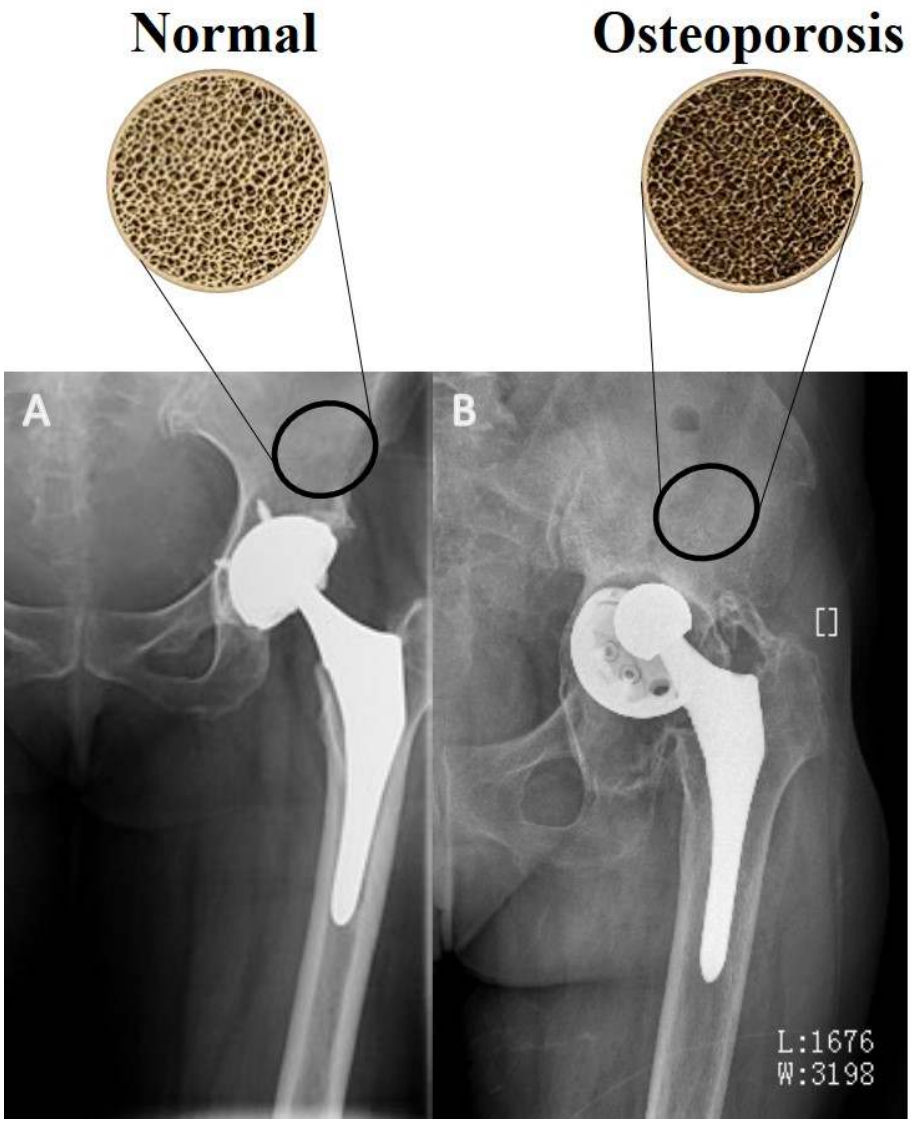


Figure 1.27: THR of 55-year-old female patient a) THR surgery b) Migration of screw and acetabular cup, due to osteoporosis after 2.8 years of surgery (Source: Zhu, et al., 2015).

Osteoporosis is the main cause for postoperative THR bone fractures, improper fixation, migration, loosening of total hip components. The management of these sophisticated and serious problems after THR are treated in several ways. At the time or after surgery, if

periprosthetic fracture occurs, the hip stem is removed and cables are placed around femur shaft, following with bone fixation plates and bone grafting. Treatment of acetabular fracture due to weakening of bone is highly dependent on the types of fracture and preoperative surgical planning and use of THR prosthetic component. Osteoporosis adverse effects and use of prosthetic component to overcome this issue is shown in figure 1.28.

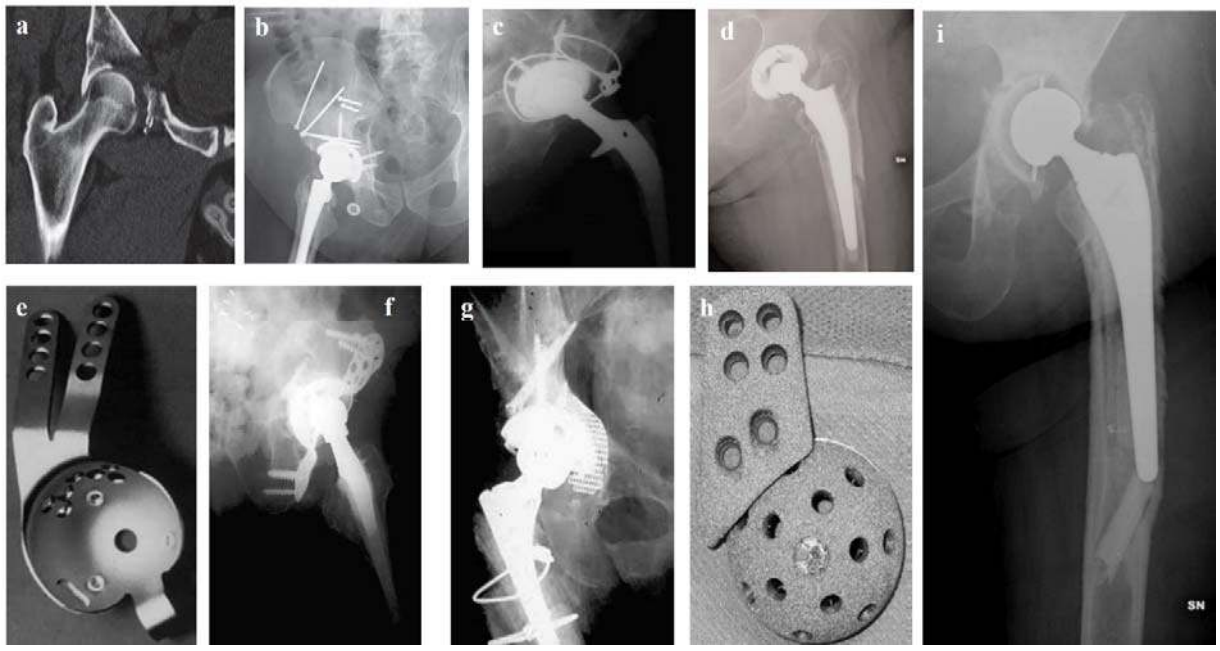


Figure 1.28: Adverse effect of osteoporosis and treatments by modular implants a) Image of 79-year-old patient with bone discontinuity in acetabulum due to reduced bone mineral density b) Repair by modified implant in THR surgery c) Acetabular roof-ring anchored with total hip prosthesis d) Fracture due to primary osteoporosis (Bottai et al., 2015) e) Illustration of U-shaped ischial supporting flange (Peters et al., 1995) f) Implementation of U-Shaped flange with sacetabular screws in THR revision (Peters et al., 1995) g)

Implementation of titanium mesh to support the bone graft in bone defect (Sloof et al., 1999) h) Acetabular augment blocks to treat diverse acetabular defects (Del Gaizo et al., 2012) i) Loosening of implant in patient (Bottai et al., 2015)

THR surgery is performed by arthroscopy surgeons in which the head of the femur is replaced by femoral stem and the acetabulum is reamed and fixed with hemispherical cups. Prosthesis required in THR is acetabular cup, femur stem. However, to achieve the good bone and implant interface, the fixation of these prosthesis is carried out by two types; cemented or uncemented. In cemented THR, bioresorbable and biocompatible fast drying cement is used to affix the prosthesis with bone. Whereas, uncemented fixation is fixed without cement, these type of fixation is also called press-fit fixation. Uncemented prosthesis are developed likewise to allow bone ingrowth onto it to achieve a proper fixation and reduced micromotion between implant and its surrounding bone.

In some situation, deficiency of bone occurs, which concludes improper fixation of prosthesis, these types of THR surgery uses many additional options, if bone loss is found in femur than “mega-prostheses”, “distal fixation taper fluted stem”, short stem, completely custom made stem is used as prosthesis (Sakellariou and Babis, 2014). Suchlike, deficiency of bone in acetabulum offers many acetabular cups with dome-screw capacity and a peripheral screw, bone augments, bilobed cups, flanged cups and patient specific custom made cups (Gonzalez, 2014).

Aside from these adjustments in prosthesis, researchers have demonstrated that geometric optimization of traditional hip components is straightforwardly relative to the increased

strength of implant inside the bone, which defeats the disadvantages of osteoporosis in implant fixation.

1.8 Screw in THR

1.8.1 Acetabular Screw

Fixation of acetabular cup by screw during THR surgery is disputable, as there are relative advantages and inconveniences. However, the initial stability of implant is exceptionally fundamental to accomplish the bone ingrowth, this stability is accomplished by inferring screws with acetabular cup (Lachiewicz et al., 1989). Regardless of certainty, neurovascular injury in pelvis region is uncommon issue brought about by using screws (Hwang, 1994), which is maintained a strategic distance from by quadrant system developed by Wasielewski et al. in 1990.

As discussed earlier in section 1.3, the hip is a load bearing joint, so it is essential for the hip implant to be fixed tightly with surrounding bone, to avoid micromotion and loosening of implant. Weakening of bone, causes low implant purchase is the main concern (Iorio et al., 2010).

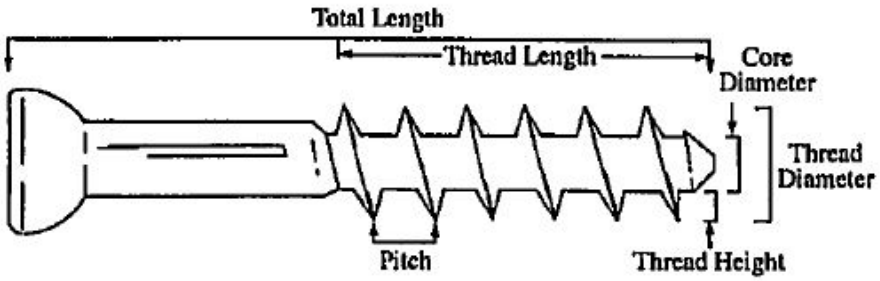


Figure 1.29: Schematic illustration of basic screw design parameters (Joel et al, 1997)

Intent on the acetabular screw (Figure 1.29), testing of metallic medical bone screws are performed by a standard ASTM test method (ASTM F543 - 17). In this test, axial pull-out test is performed to find out the strength of bone screw fixation. These tests are being performed by pulling out the screw from cadaveric, fresh or foam (artificial) bone in vertical direction as shown in figure 1.30.

Figure 1.30: Schematic illustration of pullout test for metallic medical bone screws.

Numerous study has been performed to evaluate the pull out strength of screw but the main factors which effect the screw pull out strength of the screw is disputable. A computational analysis was performed by Zhang et al, in 2006 to investigate the correlation between bone mineral strength and the pull out strength of screw. The study predicts the higher pull out strength of screw in the bone with high mineral density Battula (et al., 2006) (Figure 1.31).

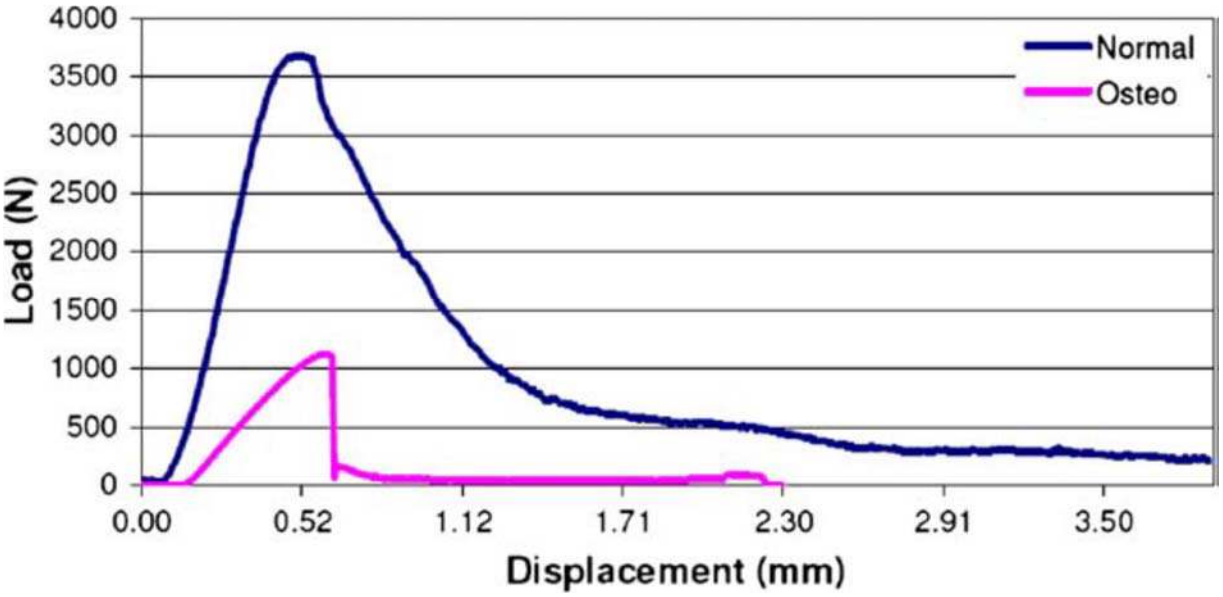


Figure 1.31: Screw pullout profile in normal and osteoporotic bone (Battula et al, 2006)

The effect of screw thread count on pull out strength has also been investigated, which showed that increasing the number of threads on screw, increases the pull out strength of screw as shown in figure 1.32.

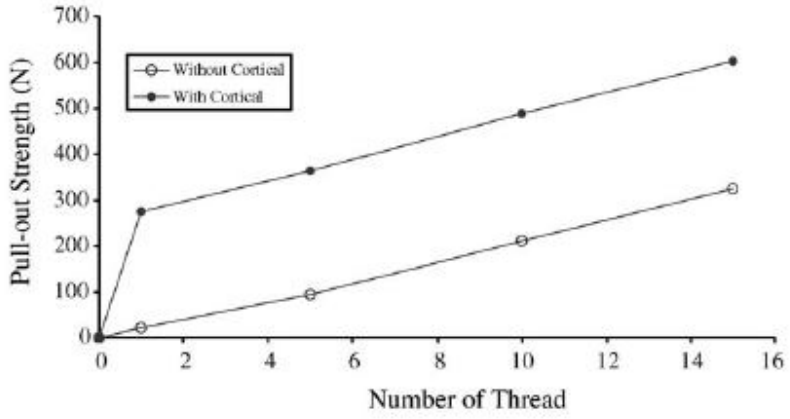


Figure 1.32: Typical screw thread versus Pullout strength curve (Zhang et al, 2006)

Defino, et al., in 2007 performed experimental analysis to access the relation between the screw inner diameter and pilot hole diameter to fix the screw in bone. Study, found that wider pilot hole (72% of the screw diameter) with respect to inner diameter of screw reduces the pull out strength. Moreover, Data from previous studies also showed that, screw length and thread shape influence the pull out strength of screw and screw osseointegration within bone to a vast majority (Chapman, et al, 1996, Chang et al, 2012, Steigenga et al., 2004, Geng et al.2004, Chun et al., 2002, McAllister et al., 2012, Arnhart et al., 2012). Kim et al, in 2012 investigated the difference between pull out strength of screw between square shape, buttress shape and V shape (Figure 1.33). They showed that V shaped thread screws showed the highest pullout strength and lowest in squared thread.

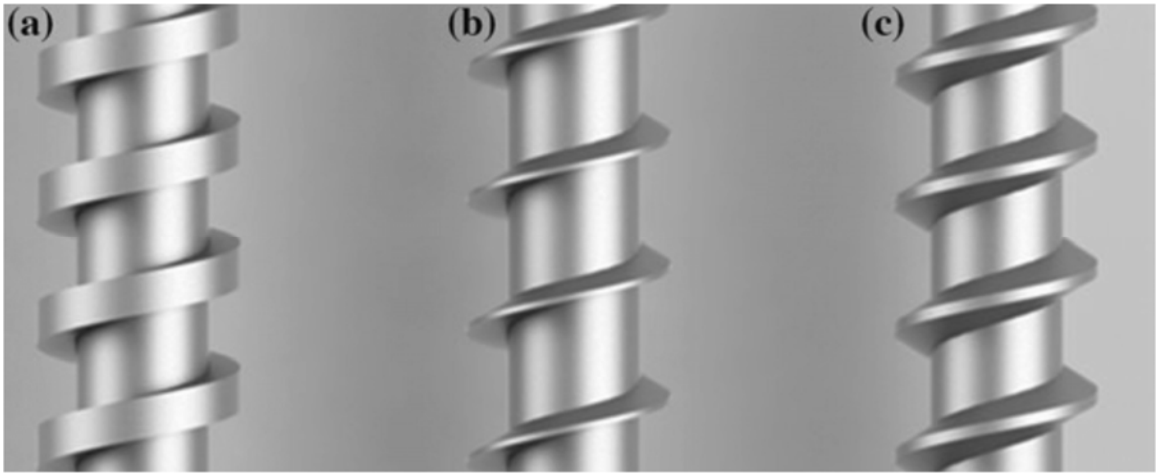


Figure 1.33: Illustration of a) square shape b) buttress shape and c) V shape screws

On the other hand, a special type of screw design is often used in osteoporosis case, known as cannulated screws (Takigawa et al, 2007; Arslan et al, 2012; Choma et al, 2012; Demir, 2014). Cannulated screw design parameter differs from standard screws on following

parameters like, increased outer diameter, decreased outer diameter to core diameter ratio, to accommodate guide wires and cement augmentation through its hollow core (Figure 1.34). These cannulated screws give the best performance then the other available standard screws. These screws are often used with bone cement to give the excellent pull out strength. However, Demir, investigated that cannulated screw could be a better option without cement and expected to give the better result other than the standard screws (Demir, 2014).

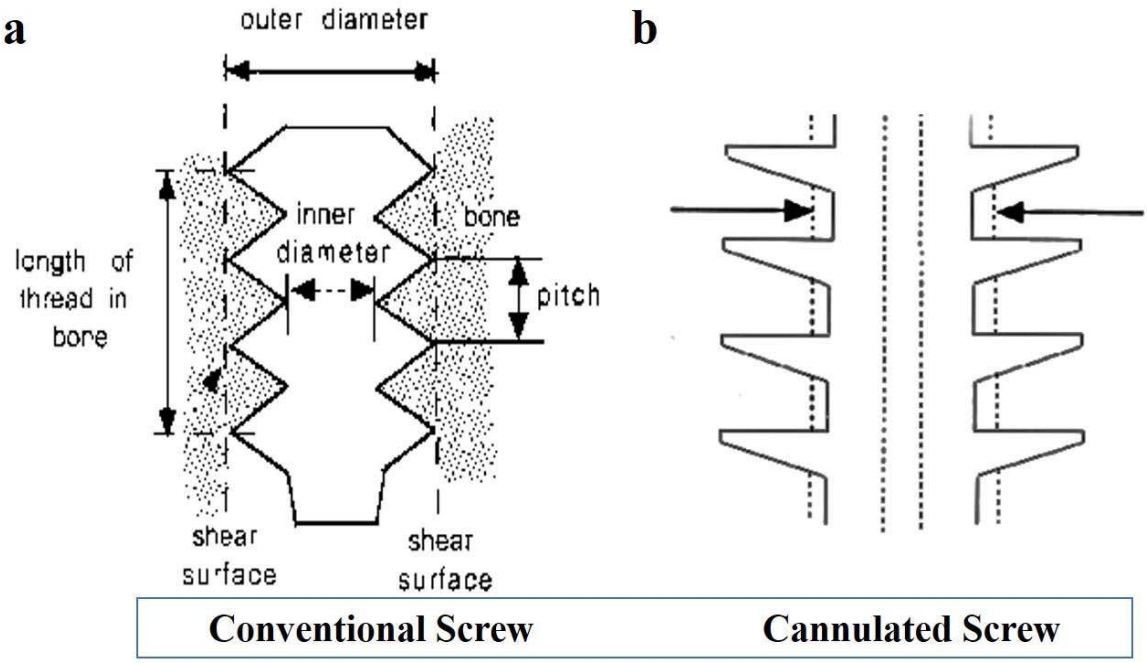


Figure 1.34: Conventional screw and Cannulated screw anatomy

Rather these above technique, bone cement were additionally utilized to achieve the excellent pullout strength and screw-bone interface for severe osteoporosis add up to hip joint (Goldhahn et al., 2008, Kock, et al., 2001, Battula et al., 2006). Research on the use

of cement with cannulated screws has been done which results a very high pullout strength increase (300%) (Yazu et al., 2005). Biocompatible bone cements were infused through screws which acts as a paste between screw and bone interface. Polymethylmethacrylate (PMMA) is most popular biocompatible polymer bone cements because of their excellent outcomes (Flahiff et al., 1995, Collinge et al., 2007, Motzkin et al., 1994). Numerous potential novel materials, similar to bone cements with calcium phosphate, have as of now been developed to take care of the fixation issues in osteoporotic bone (Yánez et al., 2010, Espigares et al., 2002, Low et al., 2010). However, cement augmentation showed good performances in pull out test of screw and fatigue strength, it possesses several complications such as thrombosis and surrounding bone damage. Use of cement causes exothermic reaction which generates unwanted heat, which result in damage of surrounding bone (Stadelmann et al., 2010 and Collinge et al., 2007). Moreover, cement leakage at the time of setting, causes immediate transitory nerve root palsy, which is another major drawbacks of cement augmentation of screws (Chang et al, 2013). Finally, revision surgery is required for some patients, in which removal of prosthesis is performed with new one; fixing the bone-screw with cement is associated with difficulty in removing of screw (Xie et al., 2011). In order to encourage screw pullout strength, another method is plasma sprayed hydroxyapatite (HA) coated screws, experimental studies and retrospective clinical analysis showed higher screw bone interface other than the uncoated screws (Durham, 2007).

Researchers have shown an alternate technique to improve the initial pull out strength of screw by expanding mechanism. In this technique the screw height is bifurcated into four legs as shown in figure 1.35. It is obvious that if distal portion of these legs are expanded, it will imply additional stress to the bone-screw contact, resulting the increase of screw stability. Vishnubhotla et al., showed the increased pullout strength in expandable screws rather than standard titanium screws as shown in figure 1.36 (Vishnubhotla et al., 2011). Then again, revision of expandable screw is risky because of bone in-growth through extended legs of screws and another factor is legs, are just expandable not retractable.

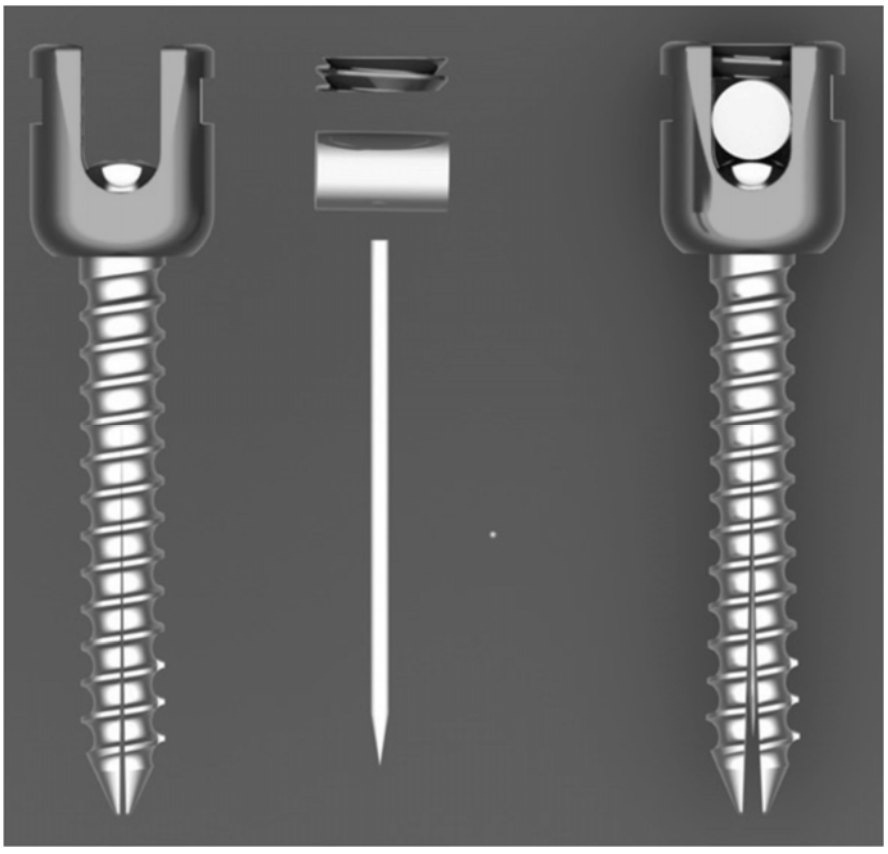


Figure 1.35: Schematic illustration of expanding mechanism of expandable screw

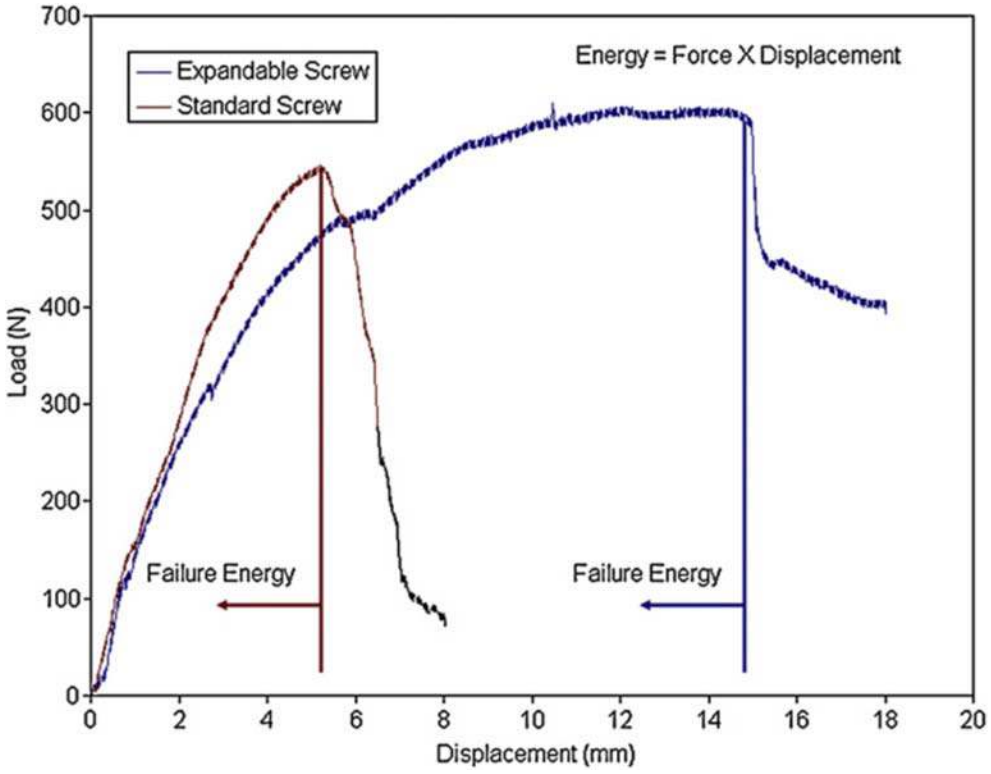


Figure 1.36: Load-displacement curve of both expandable screw and standard titanium screw Acetabular Cup