Chapter 1

Introduction and Review of Literature

1.1 Introduction

Total Hip Arthroplasty (THA) or hip replacement is a standard surgical reconstruction procedure that offers relief from pain and restoration of hip joint functions for patients suffering from osteoarthritis (arthrosis), rheumatoid arthritis, congenital deformities or post-traumatic disorders. The working principle of hip joint is similar to that of a typical ball-and-socket joint, wherein the head of the femur rotates with respect to the acetabulum. In a standard hip replacement procedure, the proximal part of the femur is replaced by an implant, consisting of a femoral stem and a modular head, whereas on the acetabular side, a hemi-spherical cup is attached to the socket of the pelvic bone. Based on the fixation technique, THA can be broadly classified as cemented and cementless arthroplasties. In cemented fixation, the implant is made to adhere to the bone using bone-cement, whereas the implant-bone fixation in cementless prosthesis is attained through bone ingrowth into the porous-coated or fibre-meshed surface on the implant. Although both the techniques have their own advantages and disadvantages, the cementless THA is being increasingly preferred by surgeons owing to its more natural (biologic) fixation with bone (Yamada *et al.*, 2009).

The past few decades have seen an exceptional growth of hip procedures globally. Each year, over 800,000 THA operations are performed all over the world (Fraldi *et al.*, 2009). From 1967 to 2013, the total number of primary total hip replacements (THRs) in Sweden has skyrocketed from a humble 6 operations to a whopping 16,330 incidences (Swedish Hip Arthroplasty Register, Annual Report 2013). More than 285,000 THAs are performed each year in the United States (Source: Agency for Healthcare Research and Quality, U.S.A.) and the number of annual hip fractures in the country has been projected to surpass 500,000 annually by the year 2040 (Cummings *et al.*, 1990). The number of hip procedures in Australia has increased by 46.50% since 2003, and it is further anticipated that the rate of increase will continue in future (National Joint Replacement Registry, Annual Report

2014). Between the years 2003 and 2013, a total of 620,400 primary hip replacements were reported in the 11th Annual Report of National Joint Registry for England and Wales (2014), while osteoarthritis accounted for the majority (93%) of the cases. The registry also reported an increase in the use of cementless implants from 16.8% to 42.5% over the same period, and corresponding decrease in the use of cemented counterparts from 60.5% to 33.2%. In India, the annual incidence rate of the surgery is approximately 470,000 (Source: The American national Institute of Arthritis and Musculoskeletal and Skin Diseases). However, in a country of over 1.2 billion people, the yearly rate of incidences is postulated to rise (Pachore *et al.*, 2013). Furthermore, according to the report, 65% of the THAs in India were performed using cementless prostheses. In the United States, 60% to 90% of the THAs performed yearly involve both cementless cup and cementless stem (Dunbar, 2009; Lombardi *et al.*, 2009).

The immense success of THA notwithstanding, the failure rate is estimated to be 10% globally (Mancuso et al., 1997; Kurtz et al., 2007), and with the recent rise in incidences, there has been a significant increase in the absolute number of failed joints (Taylor and Prendergast, 2015). The principal measure of outcome of a THA is time to first revision surgery. Ahnfelt et al. (1990) reported that the failed 10% hip replacements need revision surgery after a mean duration of 10 years in use, depending on the patient's conditions, disorder and the type of implants used. Although these failures have been due to multifactorial reasons, the majority may be attributed to the biomechanical causes. Implant-induced adverse bone remodelling and excessive implant-bone interface stresses, leading to progressive interface debonding, are two major biomechanical failure mechanisms that may compromise the durability of cementless hip prostheses. Amongst the short-term failure mechanisms, lack of primary stability due to excessive interfacial micromotion strongly influences the success of cementless arthroplasties by compromising the biologic attachment between the implant and femur. Cumulative or individual effect of all of these mechanisms may lead to gross aseptic loosening of the implant or in extreme cases, femur fracture. Besides key aspects, such as surgical procedure and patient conditions, the design (or geometry) of the femoral implant is also known to affect these mechanisms (Huiskes and Boeklagen, 1988; Viceconti et al., 2001). Although a vast variety of cementless femoral implants are commercially available in the market (Fig. 1.1), for many of them the design outcomes remain unexplored primarily due to lack of clinical data. The mechanical designs of hip implant,

therefore, may be assessed preclinically and search for optimal geometry may be conducted in order to minimize the effects of all these failure mechanisms.

The design solutions of a prosthesis may be either structural or functional, or both. However, investigations have centered mostly around the structural aspects of a prosthesis. Among various structural aspects of hip prosthesis, the geometry of the femoral implant plays a critical role in determining the design outcome. Although the gross appearance of a femoral stem has hardly changed much since it was first mentioned (Gluck, 1891), the overall non-primitive shape offers a lot of scope for intricate study by altering the profile of stem transverse sections along the stem-length using the state-of-the-art solid modelling and finite element (FE) analysis software. Moreover, with the advent of modern day high-performance computers, a large number of stem shapes may be assessed in a relatively small amount of time and the most suitable one may be chosen based on its predicted design outcomes. Therefore, a suitable design optimization strategy in combination with a solid modelling and FE analysis may lead to a more improved prosthesis design.

Shape (or geometry) optimization is a particular stage of structural optimization, which deals with the search of the optimal configuration of a design domain (Fraternali *et al.*, 2011). In shape optimization, design variables are introduced to control the geometry of the structure and the methodology typically requires an FE model that changes during the course of the optimization. The growing interest in shape optimization reflects a realization of the effectiveness of shape changes for improving structural performance. By employing shape optimization as a design tool, stem geometries can be evaluated based on biomechanical cost functions, framed on the basis of the failure principles. Therefore, the formulation of suitable cost functions, representing the global effects of these failure mechanisms is necessary in order to search for optimal designs of the cementless femoral implant that would



Fig. 1.1: Variations in design of cementless femoral implants

enhance prosthesis durability. However, for *a priori* and practical understanding of the failure scenarios associated with THA, a study on the biomechanics of hip joint is required.

1.2 Anatomy and biomechanics of hip joint

1.2.1 Anatomical planes and directions

Biomechanics of the hip joint constitutes a basic understanding of the human anatomy. Typical orientations of different anatomical planes of the human body are presented in the Fig. 1.2a. The transverse (axial or horizontal) plane, which is parallel to the ground, separates the superior (top) from the inferior (bottom) part of the human body. The coronal (frontal) plane, perpendicular to the ground, separates the anterior (front) from the posterior (back). The sagittal (median) plane is perpendicular to both the transverse and coronal plane, and it separates the left from the right part of the body. The direction towards and away from the midline of the body are termed as the 'medial' and 'lateral' direction, respectively (Fig. 1.2b). The 'anterior' and 'posterior' directions are direction towards the front and the back side of the body, respectively. The term 'proximal' is used to describe the direction towards the limbs origin, while the part away from the origin of the limbs is termed as the 'distal'. This clinical definition is done based on their proximity with respect to the head. The 'superior' and 'inferior' directions indicate the top and the bottom parts of the body, respectively.



Fig. 1.2: Sketch showing (a) the anatomical planes of reference and (b) anatomical directions and movements of the hip joint (Martini and Bartholomew, 2000).

1.2.2 Hip anatomy

The primary connection between the bones of the lower and upper limbs of the human skeletal system is formed by the hip joint, scientifically referred to as the *acetabulofemoral joint*. The hip-joint supports the body weight and transfers load from upper limb to the lower limb. The main parts that constitute this joint are a ball (femoral head) and a hemi-spherical socket (acetabulum) (Fig. 1.3). The femoral head



Fig. 1.3: Natural hip-joint (http://www.ouh.nhs.uk/hipandknee/information/hip/)

is situated at the top of the thighbone (femur) and it fits into the acetabulum in the pelvis by means of bands of tissue called ligaments (hip capsule). The ligaments provide stability to the joint. A smooth durable cover of cartilage articular (a protein substance) cushions the ends of the bones between the two bones' surfaces (femoral head and

acetabulum), and enables them to move easily. All remaining surfaces of the hip joint are covered by a thin, smooth tissue called the synovial membrane. In a healthy hip, this membrane generates a small amount of fluid that lubricates and alleviates frictional resistance in the hip joint. All of these parts of the hip-joint work in harmony, allowing easy, painless movement.

1.2.3 Structure of the femur

The femur is the longest and strongest bone in the human skeletal system. It consists, primarily, of a central shaft or diaphysis and two wider and rounded bulges known as the epiphyses (Fig. 1.4). Each epiphysis is connected to the diaphysis via conical regions called the metaphysis. The diaphysis is mainly composed of hard cortical bone with a small spongy core, while the epiphysis and metaphysis contain mostly cancellous or spongy bone within a thin shell of cortical bone. The proximal part of the femur has a head, a neck, a greater trochanter and a lesser trochanter (Fig. 1.4). The head forms two-thirds of a sphere and is directed upward, medialward, and a little forward. The surface of the head is smooth and coated with cartilage tissue in the fresh state, except over an ovoid depression, clinically known as the *fovea capitis*

femoris. The neck is a flattened pyramidal part of the bone, connecting the head with the femoral shaft and forming with the latter a wide angle opening medial-ward (Gray, 1918). This angle of inclination of neck to the shaft in the frontal plane is known as neck-shaft angle. In addition to projecting upward and towards the medial

side of the femur body, the neck also projects a little forward. This angle of inclination of the neck to the shaft in transverse plane is known as the angle of anteversion. The neck of the femur bone has an irregular cross-section. It is almost circular at the upper end and roughly elliptical with major and minor axes in the ratio of about 1.6 at the lower end close to the femoral shaft. With aging, the femoral neck gradually undergoes degenerative changes. The greater trochanter is a large, irregular, quadrilateral eminence, situated at the junction of the neck with the upper part



Fig. 1.4: Anatomy of human femur bone

of the shaft, which provides attachment sites for a number of muscles and thus, forms the most palpable part of the femur. It is directed a little lateral ward and backward, and about 10 mm lower than the head in the adult (Gray, 1918). The lesser trochanter, on the other hand, is a conical eminence on the medial side of the femur.

1.2.4 Human gait cycle

The human gait is the way locomotion is achieved by forward propulsion of the body, while maintaining synergy with the help of human limbs. Though the nature of gait varies from individual to individual, it typically follows a common pattern. Human gait is bipedal and biphasic. The two distinct but interconnected phases that constitute the gait cycle are the stance phase and the swing phase. These phases can be further subdivided into eight different phases, as described in terms of percentage of each gait cycle. In the stance phase, which is approximately 60% of the normal walking cycle, the foot remains in contact with the ground. The cycle begins with the heel contact at the start of the right foot stance phase. The right foot then comes in flat contact with the ground before the heel rises. Lifting the toe off the ground marks the end of the

stance phase. The remaining 40% of the gait cycle is known as the swing phase, when the foot moves in the air. During the right swing phase, the left leg solely supports the body. The swing phase ends with heel contact and the cycle repeats itself. The same cycle applies to the left leg with a phase difference. The duration, when both feet remain in contact with the ground, is known as the double support. A typical gait cycle is presented in Fig. 1.5.

1.2.5 Musculoskeletal loading of hip joint

The movements of the hip joint are facilitated by a total of twenty one muscle forces. However, not all of them play a major role in load transmission (Nordin and Frankel, 2001). The muscles are grouped primarily as flexors, extensors, abductors and adductors, based on the movements they produce. The lines of action of the muscles are assumed to extend from the centre of the area of origin to the centre of the area of their insertion or to any bony surfaces over which they pass (Brand *et al.*, 1982; Duda *et al.*, 1996). The origin and insertion of the hip joint muscles are presented in Table 1.1. Musculoskeletal loading is known to play a significant role in the biological process of fracture healing, bone remodelling and primary stability of an implant



Fig. 1.5: A typical human gait cycle (Inman et al., 1981)

(Bitsakos *et al.*, 2005; Duda *et al.*, 1998; Weinans *et al.*, 2000). Apart from these muscle forces, the hip joint reaction force has a significant influence in transferring load between the two extremities of human body (Dalstra and Huiskes, 1995).

Several investigators have measured hip joint reaction (or hip contact) force either using force platforms and kinematic data combined with electromyography (EMG) for the normal hip joint (Crowninshield et al., 1978; Paul, 1967; Röhrle et al., 1984; van den Bogert et al., 1999) or using instrumented hip prostheses for the implanted hip joint (Bergmann et al., 1993; 1995; 2001; 2004; Davy et al., 1988; Kotzar et al., 1991; Taylor et al., 1997). A large amount of inter-patient variability was considered in these studies for the measurement of hip contact force. A summary of the range of peak hip contact forces and torsional moments during routine activities is presented in the Table 1.2. In a cohort study using four patients, Bergmann et al. (2001) measured the hip contact forces during most-frequent daily activities, such as normal walking and stair climbing, and also calculated the average joint forces. The average value of the peak hip contact force, reported in the study, was roughly 238% of body weight (BW) during walking (at a speed of 4 km/h). However, during stair climbing and going downstairs, the average measured hip contact forces were reported to be higher, 251% and 260% BW, respectively. The variation of the hip contact forces over the entire duration of a gait cycle is shown in Fig. 1.6. It may be noted that the hip joint force reaches its peak value right after the heel strike, which occurs approximately at 18% of the gait cycle.

Although accurate measurements of hip contact force on the femoral head were possible using instrumented prostheses, ethical complication associated with invasive



Fig. 1.6: Hip joint reaction: (a) joint reaction force diagram for the hip joint for normal gait; (b) comparison between calculated and measured joint reaction force (Bergmann *et al.*, 2001; Heller *et al.*, 2001).

Movement Muscle		Origin	Insertion	
	m.gracilis	Pubic bone	Anterior medial tibial condyle	
	m.pectineus	Pubic bone	Pectineal line	
Flexion	m.iliopsoas	Iliac fossa, anterior lumber spine	Lesser trochanter	
	m.sartorius	Anterior superior iliac spine	Anterior medial tibial condyle	
	m.rectus femoris	Anterior superior iliac spine	Tibial tuber via patellar tendon	
Extension	m.gluteus maximus	Posterior ilium, sacrum	Iliotibial band and gluteal tuberosity	
	m.biceps femoris	Ischial tuberosity, linea aspera	Fibular head	
	m.semitendinosus	Ischial tuberosity	Anterior - medial tibial condyle	
	m.semimembranosus	Ischial tuberosity	Medial tibial condyle	
	m.tensor fascia latae	Lateral to Anterior superior iliac spine	Inserts into iliotibial band	
Abduction	m.gluteus medius	Gluteal lines on posterior ilium	Greater trochanter Greater trochanter	
	ni.gracous minimus	Gluteal lines on posterior ilium		
	m.adductor magnus	Inferior pubis and ischium	Adductor tuberosity, linea aspera	
Adduction	m.adductor longus	Pubic bone	Linea aspera	
	m.adductor brevis	Pubic bone	Upper linea aspera	

Table 1.1: Principal actions of dominant hip muscles (adapted from Dowson *et al.*, 1981). The abbreviation m. stands for muscle.

methods served as a deterrent for quantitative *in vivo* assessment of muscle forces. This led to the wide use of mathematical optimisation algorithms for estimation of the complex distribution of *in vivo* muscle forces (Brand *et al.*, 1982; 1986; 1994; Crowninshield and Brand, 1981; Duda *et al.*, 1996; 1997; 1998; Glitsch and Baumann, 1997; Heller *et al.*, 2001; Pedersen *et al.*, 1997). The data obtained from the optimization method was found to be in good agreement with the measured EMG data of the muscle forces during normal gait (Crowninshield and Brand, 1981; Glitsch and Baumann, 1997). In a comprehensive optimization study, Heller *et al.* (2001) calculated the magnitude of muscle forces and the hip contact force for daily activities, e.g., walking and stairs climbing, for four patients.

The optimisation procedure incorporated the criterion of minimising the sum of all muscle forces (Crowninshield, 1978), coupled with the inequality constraints imposed on the maximum muscle forces (Challis, 1997). Since maximum muscle activation during the mentioned activities was unlikely to occur, muscle forces were

Activity	Hip contact force (% BW)	Twisting moment (% BW.m)	Instrument	References	
	260 - 280	-		Davy et al., 1988	
	270	-	Instrumented	Kotzar et al., 1991	
	277	-	telemetric hip	Taylor et al., 1997	
Walking	$280 - 480^{\#\#}$	$1.30 - 4.40^{\#}$	prostitesis	Bergmann et al., 1993	
	211 - 285	1.20 - 190		Bergmann et al., 2001	
	220 - 280	-		van den Bogert et al., 1999	
	490 - 700	-	EMG/force	Paul, 1967	
	450 - 750	-	plate	Crowninshield et al., 1978	
	260	-		Davy et al., 1988	
Stair	320	-		Taylor et al., 1997	
climbing	$350 - 550^{\#}$	$3.70 - 5.70^{\#}$		Bergmann et al., 1995	
	227 - 314	1.80 - 3.00	Instrumented	Bergmann et al., 2001	
Jogging	550	5.30	30 telemetric hip	Bergmann et al., 1993	
Stumbling	870	5.40	prosthesis	Bergmann et al., 1993	
				Bergmann et al., 2004	
Single leg stance	210 - 280	-		Kotzar et al., 1991	
Standing up	181 - 220	0.80 - 1.21	-		
Sitting down	149 – 176	0.40 - 0.91		Bergmann et al., 2001	
Knee bend	117 - 177	0.58 - 0.83			

Table 1.2: Range of peak hip contact force and torsional moments in routine activities from selected studies (adapted from Pal, 2009).

Notes: ^{##} [#] Upper value for walking at 5 km·h⁻¹ [#] Upper value measured in one patient only and considered abnormally high

Table 1.3: Forces applied by major muscles and ligament on the femur in terms of body weight (BW) at approximately 20% of the gait cycle. The abbreviation m. stands for muscle.

Musala Nama	Force (%BW)			
Muscle Name	Normal walking	Stairs climbing		
m. abductor	104.2	113.8		
Iliotibial tract, proximal	0.0	16.8		
Iliotibial tract, distal	0.0	16.8		
m.tensor fascia lata, proximal part	19.0	6.5		
m. tensor fascia lata, distal part	19.0	6.5		
m. vastus lateralis	94.8	137.0		
m. vastus medialis	0.0	270.1		

Source: Heller et al., (2001).

restricted to below 85% of physiological muscle forces (Heller *et al.*, 2001). The forces were calculated as the product of each muscle's physiological cross-section area (PCSA) (Brand *et al.*, 1986; Duda *et al.*, 1996) and a muscle stress of 1 MPa (An *et al.*, 1989). Figure 1.6b shows a graphical representation of the measured and calculated values of hip contact force during normal walking. Although the optimization method yielded hip contact force values (Heller *et al.*, 2001), reasonably similar to measured values (Bergmann *et al.*, 2001), deviations in measured and calculated values were observed throughout the entire gait cycle. The major muscle forces calculated by Heller *et al.* (2001), corresponding to the peak hip contact force, are summarised in Table 1.3. These forces were based on the data from hip joint implanted with telemetric prostheses (Bergmann *et al.*, 2001; Heller *et al.*, 2001).

For biomechanical analyses of long bones, Duda *et al.* (1996) stressed on the importance of accurate quantification of the locations and size of muscle attachment. Different methods were adopted to quantify the muscles attachment size and locations in lower limbs (Brand *et al.*, 1982; Chao *et al.*, 1993; 1994; Crowninshiled *et al.*, 1978; Dostal and Andrews, 1981; Duda *et al.*, 1996; Lengsfeld *et al.*, 1994). Duda *et al.* (1996) employed a digitising method to determine reproducibly the muscle attachment area, centroidal location of the area and the muscle volume corresponding to six femoral specimens. Considerable amount of inter-specimen variability was reported in the measured muscle volume and computed area of muscle attachments



Fig. 1.7: Major group of muscles and their attachment points: P1, P2 and P3 are the attachment sites of the muscles (Bergmann *et al.*, 2001; Heller *et al.*, 2001).

(Brand *et al.*, 1982; Duda *et al.*, 1996). Typical attachment points for major muscles of the hip joint are presented in Fig. 1.7 (Bergmann *et al.*, 2001; Heller *et al.*, 2001). According to the location data provided by Brand *et al.* (1982) and Duda *et al.* (1996), muscles were represented as straight lines spanning from the centre of origin to the centre of insertion.

Both computational and in vitro studies indicated that the hip contact force and abductor muscle forces have the greatest influence on the strain distribution within the proximal femur during walking and stair climbing. The other significant contributors are the vastus (medialis and lateralis), tensor fasciae latae and iliotibial tract. Musculoskeletal hip loading data of Bergmann et al. (2001) and Heller et al. (2001) have been extensively used in most of the recent investigations on the proximal femur. The hip contact force acts at a distributed area on the surface of the femur head and makes an angle of 17° with the vertical in the frontal plane. Abductor muscle force also acts parallel to the hip contact force, albeit along an opposite direction and at a different location of attachment (P1) in the greater trochanter region. Surrounding the area around the point P1 attached are the iliotibial tract (proximal and distal) and the muscle tensor fascia latae (proximal and distal). The respective attachment sites for vastus lateralis and vastus medialis are P2 and P3. Musculoskeletal loading conditions for normal walking comprise of the hip contact force and the muscle forces of abductor, tensor fascia latae (proximal and distal) and vastus lateralis (Bergmann et al., 2001; Heller et al., 2001). The loading conditions for stairs climbing include the additional effects of iliotibial tract (proximal and distal) and vastus medialis, along with the hip contact force and the muscle forces applied during normal walking.

1.3 Bone structure and properties

Bone is a solid structural element of the human body that constitutes part of the vertebral skeleton. A connective tissue with unique structural and mechanical properties, bone protects the soft tissues and organs of the body, and transmits weight and muscle forces from one part of the body to another during daily activities, while maintaining shape of the body. The primary function of bone, however, is to bear load. From engineering perspective, bone is an anisotropic, non-homogeneous and viscoelastic material. Like most biological tissues, it is able to adapt its structure according to change in mechanical environment. Moreover, it exhibits wide variations in morphology depending on the porosity of the structure.

Macroscopically, bone exists in two basic forms depending on their relative densities or volume fractions of solids: cortical and cancellous (Gibson, 1985). The dense outer shell surrounding the core spongy cancellous part is known as the cortical or compact bone. Cortical bone has a volume fraction of solids greater than 70%, whereas the cancellous (or trabecular or spongy) bone is porous with a volume fraction of solids less than 70%. The distribution of cortical and cancellous bone varies considerably from bone to bone, and also from patient to patient. The mechanical properties of bone tissues are governed by the mineral and organic composition. The hydroxyapatite (HA) (scientifically referred to as *hydrated calcium phosphate*, $Ca_{10}(PO_4)_6(OH)_2$) contributes significantly to the stiffness of the cortical bone, while the collagen content determines the ductility (Guo, 2008). Cancellous bone has a distinctive lattice structured network of interconnecting rods and plates. This lattice of rods and plates is called trabeculae; hence the name trabecular bone.

From the molecular perspective, bone mimics a true composite material (Currey, 1984). It consists of 65% mineral (HA), 35% organic matrix (mostly collagen fibers), water, cells and vessels. The mineral content is largely impure HA in the form of small crystals with the shape of needles, plates and rods located within and between the organic matrix. The organic matrix is comprised of 90% collagen and 10% of various noncollagenous proteins (Jee, 2008). Depending on the differences of the architecture and arrangement of collagen fibres, bone can be categorised mainly into two types; woven and lamellar bone. Woven bone is premature temporary phase of bone. It grows rapidly and typically found in foetus, at younger ages and during fracture healing of bones. The collagen fibres in woven bone are comparatively loosely packed and randomly oriented, as compared to lamellar bone. Lamellar bone is more regularly arranged and the growth rate is not as fast as in woven bone. The fibres of collagen and associated calcium phosphate are oriented in forms of sheets, known as lamellae. The mineral content in this type of bone is usually lower than the woven bone (Currey, 1984). Lamellar bone can be further categorised into primary and secondary lamellar bone. Small cavities (lacunae) connected by their tubular canals (canaliculi) are found throughout the woven and lamellar bone. Entrapped bone cells (osteocytes) and their long cytoplasmic processes occupy the lacunae and canaliculi, respectively (Jee, 2008).

1.3.1 Mechanical properties of cortical bone

Cortical bone represents nearly 80% of the body's bone mass in an adult human skeleton, but accounts for only 20% of the bone volume. It is a solid compact tissue constituting the diaphysis of the long bones and outer shell of the epiphyses and metaphyses. Cortical bone exhibits anisotropic material behaviour, with a larger ultimate strength and elastic modulus in the longitudinal direction than in the transverse directions. Moreover, it is stronger in compression than in tension. Ashman et al. (1984) measured the elastic moduli of cortical bone using a continuous wave ultrasound technique. They estimated the elastic moduli to be about 20 - 22 GPa along the axis of the long bone and 12 - 14 GPa transversely. This indicates that the cortical bone is transversely isotropic; about 1.5 - 2 times stiffer and stronger in the longitudinal direction than in either the radial or circumferential directions. In spite of its inherent anisotropy, an orthotropic or transversely isotropic constitutive relationship describes the cortical bone elastic properties fairly well (Guo, 2008). Elastic moduli of cortical bone, measured using mechanical testing, were reported to be in the range of 17.5 ± 1.9 GPa (Carter *et al.*, 1981). However, the reported values of Elastic moduli and strengths are only indicative as they may vary depending on ethnicity, age and sex of the patient.

The stress-strain behaviour of human femoral cortical bone was reported by Özkaya and Nordin (1999), who observed three distinct regions in the stress-strain curve. An Elastic modulus in the range of 17 GPa was found in the initial linearly elastic region. In the intermediate region, the bone exhibits nonlinear elastoplastic behaviour. This region is characterised by bone yielding with yield strength value reported to be around 110 MPa. The final region exhibits linear plastic behaviour with a strain hardening modulus of 0.9 GPa. The bone was found to have fractured when the tensile stress was 128 MPa and the corresponding tensile strain was 0.026 (Özkaya and Nordin, 1999). The study also reported that the elastic moduli and strength values of a bone specimen depends on the strain rate, which is indicative of the viscoelastic property of the bone (Özkaya and Nordin, 1999). Further investigations based on mechanical testing also suggested that the stress-strain behaviour of the bone is dependent on the orientation of the bone with respect to the direction of loading. Investigations were also carried out to provide data on the material properties of human femur under dynamic loading (Funk et al., 2004; Asgharpour et al., 2014).

1.3.2 Mechanical properties of cancellous bone

The mechanical behaviour of cancellous or trabecular bone varies extensively depending on the mode of loading. The stress-strain curve of cancellous bone under compressive load contains an initial linearly elastic region, followed by a plateau region of almost constant stress and finally, an increasingly steep region culminating into fracture (Gibson, 1985; Özkaya and Nordin, 1999). The material yielding tends to occur as the trabeculae begin to fracture. It should, however, be noted that the mechanical yield property of cancellous bone varies significantly with anatomical location (Morgan and Keaveny, 2001; Morgan *et al.*, 2003). As opposed to the cortical bone, cancellous bone fractures abruptly under tensile forces, exhibiting brittle material behaviour. The energy absorption capacity of cancellous bone is higher under compressive load than under tensile load (Kaneko *et al.*, 2004).

The material properties and the stress-strain characteristics of cancellous bone depend not only on the mode of loading but also on its apparent density. It is reported to exhibit mechanical behaviours similar to other solid cellular structures, e.g., polymeric foam (Gibson, 1985; Gibson and Ashby, 1988; Pugh *et al.*, 1973; Rajan, 1985). The apparent or relative density is equivalent to the volume fraction of solids in the cancellous bone, which is calculated from the cancellous bone density and volume of the trabeculae (or solid cell wall). At low relative densities, it has rods connecting to form open cells. At higher relative densities, more material is accumulated in the cell walls and the structure transforms into a more closed network of plates. The analysis by Gibson (1985) showed that the Young's modulus varies with the square of the density for an open cell structure, and with the cube of density for a closed cell structure. This prediction was further corroborated by Carter and Hayes (1977) who suggested a transition from rod-like to plate-like elements at a relative density of 0.20.

A series of experimental studies were carried out to find empirical power law relationships between apparent bone density and elastic modulus (Carter *et al.*, 1987; 1989; Morgan and Keaveny, 2001; Morgan *et al.*, 2003). Different equations were deduced, all based on the same basic relationship: $E = C\rho^{D}$. The values for constant C were found to vary in the range 3,000 – 30,000, whereas values for constant D ranged between 1.14 and 3.2, depending on the location of the bone (Morgan *et al.*, 2003). Evidently, a single elastic modulus-density relationship is not applicable across all the

Anatomic site	Apparent density	$E = C\rho^{D}$		
	(Range)	C (95% CI)	D (95% CI)	
Vertebra (T10 – L5)	(0.11 – 0.35)	4730 (3050 – 7320)	1.56	
Proximal Tibia	(0.09 – 0.41)	15520 (10830 – 22230)	1.93	
Greater Trochanter	(0.14 – 0.28)	15010 (7590 – 29690)	2.18	
Femoral Neck	(0.26 – 0.75)	6850 (5440 – 8630)	1.49	
Pooled	(0.09 – 0.75)	8920 (7540 – 10550)	1.83	

Table 1.4: Range of constant C and D used in the power-law regression between the elastic modulus (*E* in MPa) and apparent density (ρ in g.cm⁻³).

Source: Morgan *et al.* (2003), Helgason *et al.* (2008). CI denotes Confidence Interval

Table 1.	5: Trabecular	r bone mec	hanical	properties	(mean :	\pm standard	deviation)	by
anatomic	site and load	ling mode.						

Anatomic site- Loading mode	Apparent density (g·cm ⁻³)	Modulus (MPa)	Yield strain (%)	Yield stress (MPa)
X7 (1				
Vertebra				
Compression	0.18 ± 0.05	344 ± 148	0.77 ± 0.06	2.02 ± 0.92
Tension	0.19 ± 0.04	349 ± 133	0.70 ± 0.05	1.72 ± 0.64
p-value	NS	NS	< 0.001	NS
Proximal tibia				
Compression	0.23 ± 0.06	1091 ± 634	0.73 ± 0.06	5.83 ± 3.42
Tension	0.23 ± 0.10	1068 ± 840	0.65 ± 0.05	4.50 ± 3.14
p-value	NS	NS	< 0.001	NS
Greater trochanter				
Compression	0.22 ± 0.05	622 ± 302	0.70 ± 0.05	3.21 ± 1.83
Tension	0.22 ± 0.04	597 ± 330	0.61 ± 0.05	2.44 ± 1.26
p-value	NS	NS	< 0.001	NS
Femoral neck				
Compression	0.58 ± 0.11	3230 ± 936	0.85 ± 0.10	17.45 ± 6.15
Tension	0.54 ± 0.12	2700 ± 772	0.61 ± 0.03	10.93 ± 3.08
p-value	NS	NS	< 0.001	0.003

Source: Morgan and Keaveny (2001). NS indicate no significant differences (p>0.05)

anatomic sites (Table 1.4). A similar principle was adopted by Kaplan *et al.* (1985) and Stone *et al.* (1983), who determined the static strength of trabecular bone by relating it to the local bone density through power law relationship.

It is evident from the yield strength data for various anatomical locations and loading conditions that bone is a non-homogeneous anisotropic material (Table 1.5). However, the site-specific differences between the yield strain results were found to be small. Turner (1989) observed that the correlations between yield strain and the trabecular orientation, structural density and bone density vary little for differing definitions of yield. This suggests that yield strain in cancellous bone is independent of the structural anisotropy. A further investigation suggested that the apparent density in bovine cancellous bone adapts in a manner such that the continuum-level strains are uniform (Turner et al., 1996). The uniform strain criterion was found to reproduce realistic density distributions in the proximal femur and was also applicable to human bone. In a recent experimental study by Morgan and Keaveny (2001), it is suggested that strain-based criteria for human trabecular bone may be more mathematically simple and statistically powerful. They tested cylindrical specimens of human trabecular bone taken from different anatomic site under both uniaxial tensile and compressive loads and subsequently, confirmed that the yield strains of this bone depend on anatomic site. Moreover, due to the weak dependence on the apparent density, the yield strains can be considered uniform within a single site despite substantial variation in elastic modulus and yield stress.

1.4 Review of literature: hip arthroplasty

Hip replacement surgery has over 100 years of operative history. However, the procedure has undergone a series of revolutionary changes and modifications over the years. The earliest recorded attempts at hip replacement happened in Germany back in 1891 by Themistocles Gluck, with results presented at the 10th International Medical Conference. Later in the late 19th and early 20th century, surgeons experimented with interpositional arthroplasty, which involved placing various tissues (fascia lata, skin, pig bladders submucosa) between articulating hip surfaces of the arthritic hip (Learmonth *et al.*, 2007). Surgeon Marius Smith-Petersen may be credited for the development of the first glass-made 'mold arthroplasty' in 1925. The prosthesis consisted of a hollow hemisphere which could fit over the femoral head to provide a new smooth articulating surface. However, the surface failed to withstand the hip joint

forces and ruptured. Marius Smith-Petersen and Philip Wiles later independently tested cobalt-chrome (Co-Cr) implant, a dramatic improvement in the field of arthroplasty, in the first THR that was fitted to bone with bolts and screws (Smith-Petersen, 1948; Wiles, 1957). While this new metal proved to be a success, the actual resurfacing technique was found to be inadequate. In order to address the problem of the femur fracture and arthritic femoral head, Frederick R. Thompson and Austin T. Moore separately developed replacements for the entire femoral head (hemi-arthroplasty). In spite of being hugely popular in the 1950s, unpredictability in outcome and damage of the acetabular cavity persisted. Moreover, no effective method of securing the component to the bone was found as yet.

English surgeon George McKee was the first to use a metal-on-metal prosthesis on a regular basis. In 1953, he started to use modified Thompson stem with a new one-piece Co-Cr socket as the new acetabulum. McKee, together with Watson-Farrar, further developed on this design. This prosthesis was reported to have good survival rate, with one study recently showing survival rate of 74% in a 28-year follow-up (Brown *et al.*, 2002). Nevertheless, the popularity of the method waned by the mid-1970's, owing to high frictional resistance producing wear particle debris (McKellop *et al.*, 1996). As early as in 1938, Judet brother, working in Paris, attempted to use an acrylic material to replace arthritic hip surfaces. Although a smooth surface was achieved using this technique, the occasional loosening remained an issue. Hence, the search for alternative prosthesis design continued.

Surgeon Sir John Charnley is considered the father of the modern THA. In 1962, while working at the Manchester Royal Infirmary, he developed the Charnley THR, a gold standard of primary hip replacement. Following unsatisfactory experience with Teflon cups (1959-62), Charnley introduced his first low friction arthroplasty (LFA) (Charnley, 1970), which was identical, in principle, to the prostheses used today. It consisted of three parts: a metal femoral stem, a high density polyethylene acetabular component and acrylic bone cement. This work of his resulted in a design that completely replaced the other designs by the 1960s. It was called 'low friction arthroplasty' as Charnley advocated the use of a small femoral head leading to considerable reduction in wear due to smaller surface area. Long-term results of the Charnley LFA, including 35-year follow-up, reflected a 78% femoral implant survival.2, although the design suffered from poor joint stability. Despite this, the Charnley LFA design was the most accepted artificial hip over two decades.

However, the original Charnley prosthesis has undergone further modifications, since 1969, with regard to stem shape, surface texture and femoral head material to further reduce failure of the THA.

1.4.1 Hip replacement: cemented vis-à-vis cementless fixation

Until about late 1970s and early 1980s, cemented fixation was the predominant technique used for THA. Although the Charnley LFA presented a successful cemented design and technique, the outcomes of the other cemented THA procedures were distinctly poor with high and early loosening rate (Yamada *et al.*, 2009). These failures were found to be associated with localised areas of bone destruction and resorption (osteolysis). The root cause of these failures was initially believed to be infection (Charnley, 1979). However, subsequent revelations attributed the cause to a local inflammatory response initiated by cement particles. Histological examination of tissue taken from these localised areas of osteolysis showed the presence of 'polymethyl methacrylate debris' (Harris *et al.*, 1976), which led to the conclusion that the premature loosening of cemented components was related to so-called 'cement disease' (Jasty *et al.*, 1986; Jones and Hungerford, 1987; Maloney *et al.*, 1990). This prompted researchers to find alternative method of fixation in THA.

Later investigations further pointed out that cementing the prostheses has certain disadvantages; prominent among them are necrosis and bone thermal injury (Mont *et al.*, 2006; Beaulé *et al.*, 2007). Moreover, cement is strong in compression, but weak in tension. The implant-cement interface has often been reported as the weakest link in the implanted bone, leading to interface debonding. Cement particles abraded from the cement mantle may lead to particulate reactions by macrophages, osteolysis, soft tissue interposition and eventual loosening of the implant. These issues, although not very well understood earlier, might have led to the development of cementless, press-fitted implant designs. The principles of cementless fixation have evolved since the first outcomes were reported in 1979 (Lord *et al.*, 1979). The design aim was to obtain stability through biologic fixation occurring naturally at the implant-bone interface. The anatomic medullary locking (AML) stem was the first cementless femoral implant approved for use in the United States. The design featured a straight, extensively porous coated Co-Cr stem that employed diaphyseal fixation. The stem has an exceptional track-record, including up to 98% survivorship at 20 years

(Belmont *et al.*, 2008). However, disadvantages include proximal stress shielding and occasional thigh pain due to distal cortical hypertrophy (Belmont *et al.*, 2008).

The success of AML notwithstanding, the clinical outcomes of majority of early cementless designs were equally poor compared to their cemented counterparts. This is partly because of the implant designs having smooth surface, which failed to adhere strongly to the bone. As a result of these smooth-surfaced cementless THAs, aseptic loosening was reported to occur several years after surgery, and in many cases required revision surgery (Kawamoto et al., 1998). The development of material engineering during the 1980s, however, revolutionised cementless implants with the advent of materials that allow biologic fixation. This biologic fixation can be achieved through bone ingrowth or ongrowth, or a combination of both. Porous-coated implants facilitate the host bone to grow inside the pores of the implant surface and a stable biologic fixation through osseointegration occurs naturally. This process is clinically termed as bone ingrowth. For grit blasted implants, the bone grows onto the microdivots of the textured implant surface and the biologic fixation is attained through bone ongrowth, and finally osseoincorporation. Nevertheless, porous-coated ingrowth type implants are more common in hip replacement procedures. Different type of surface processing, such as titanium (Ti) fibre mesh, Co-Cr beads, Ti plasma spray, are presently in use for the development of bone-ingrowth type implants. Fig.



Fig. 1.8: Cementless total hip arthroplasty: (a) the surgical procedure and (b) the typical hip replacement components (Source: *http://www.oakazoo.com/allHips/hip_replacement.htm*).

1.8 illustrates the modern THA procedure (cementless) and the corresponding hip replacement components including a proximally porous-coated femoral stem.

The relative superiority between cementless and cemented technique, however, remained a matter of debate in the field of THA. Nevertheless, the long-term survival rate of cemented implants was reported to have surpassed that for any cementless implant system (Malchau *et al.*, 1996). The reason behind the relatively inferior results for cementless technique may, in part, be ascribed to the apparent paucity of understanding of the fixation method and hence, further investigations – clinical as well as preclinical – are warranted. Nevertheless, the enormous computing abilities of the modern-day computers have enabled researchers to dig deeper into the understanding of the performance of cementless implants. The recent advancement of image processing techniques and 3-D printing is further ushering a new age of cementless implant design.

1.4.2 Reasons for hip replacement

The primary symptom of a diseased hip joint is pain, which, in extreme cases, calls for hip replacement. The pain can be mild to debilitating; however the ability to perform normal activities becomes restricted nonetheless. Subsequent to hip replacement, the patient gets a new lease of life with improved mobility and considerable reduction in pain. The most common ailments that lead to hip surgery are described as follows:

Osteoarthritis (OA) is the most common reason for hip replacement surgery, accounting for over 75% of the THAs performed globally. Primarily associated with aging, the disease degenerates cartilage exposing the bone surfaces to come into direct contact of each other, causing pain and restricted mobility. Gait abnormalities and limitation of motion can also cause muscle atrophy and weakness over time. Although post-traumatic arthritis is caused by trauma or injury, the diagnosis and symptoms are similar to that of OA. The trauma may be induced in the form of a fracture or dislocation, altering the mechanics of the hip joint.

Rheumatoid Arthritis (RA) is a chronic and progressive joint disease that causes inflammation in the cartilage of the joints, eventually leading to swelling, pain and immobility to the patient. Tipped as an autoimmune disorder, the root cause of RA is not properly known yet. Multiple joints, both weight-bearing and non weight-bearing,

are affected by the disease. RA is known to inflame the synovial membrane which, as a result, starts to release enzymes. These enzymes digest bone and cartilage, leading to eventual deformation of the joint and inflammation of the joint lining, and thereby causing damage to the joint surface.

Osteonecrosis (ON), sometimes called as aseptic or avascular necrosis, is a condition in which the part of the femoral head loses its blood supply resulting in bone cells dying due to the lack of oxygen and nutrients. Patients having fractured or dislocated hip and having been on steroids or having a history of heavy alcohol use are at higher risk of developing this condition. Evidences suggest that early stages bear little or no X-ray findings, while at advanced stages, joint damage identical to OA can be observed. Though the major symptom of ON is pain in groin, often the patient carries the disease for a considerable period of time without any noticeable symptom. Initial treatments typically include the use of walking aids to reduce the load on the joint. However, no preventive medicines or treatments for ON have been established as yet. Consequently, surgical intervention, such as joint replacement, osteotomy of the bone and bone grafting, becomes the major course for ON treatment. Other reasons for hip surgery include femoral neck fracture and inter-trochanteric fracture owing to trauma or low bone quality, or a combination of both.

1.4.3 Hip replacement: failure scenarios

Although failure of a hip prosthesis may be due to multifactorial reasons, the initiation of the failure process can be attributed to mechanical events. Three dominant failure scenarios, as suggested by Huiskes (1993) for reconstructed hip joints, are presented as follows:

Accumulated damage failure scenarios: The likelihood of mechanical failure depends on the stresses induced within an implant material or at various material interfaces within the implant-bone structure, as compared to the strength of the material or the respective material interface. These implant materials or interfaces are too weak to sustain the effect of long-term, dynamic loads due to normal physiological activities. As a result, mechanical damage, typically micro-cracks, is gradually accumulated within the implanted bone structure, eventually causing failure.

Particulate-reaction failure scenarios: There are three possible sources of generation of wear particle debris in joint replacement: (1) wear of articulating surfaces, (2) abrasion of cement/prosthesis/bone interfaces, and (3) fretting between metal parts in

modular prostheses. As a result of the generation and migration of these wear particle debris, the cement-bone interface gradually disintegrates, causing interfacial micromotion and eventual loosening.

Among other aseptic failure mechanisms of the implanted hip joint, dislocation, migration, and effect of stress shielding and bone remodelling need special mention:

Dislocation: Surgical intervention disturbs the stable structure of the normal hip joint, enhancing the risk of dislocation. The small bearing size at the articulating surface may also be responsible for dislocation. Impaction allografting, a technique to treat cavitary defects, is also reported to cause dislocation (Dattani, 2007).

Migration: All cemented and cementless implants are subject to some extent of initial migration relative to the bone. However, cementless implants are more prone to migration due to lack of initial adhesive aids, e.g., bone cement. A portion of the implant support may be lost due to microcracking of the trabeculae in the bone, allowing it to subside (Burr *et al.*, 1985). This is observed in the regions of high stress, particularly around the tip of the femoral implant. Rapid early migration is associated with premature loosening of the implant (Glyn-Jones *et al.*, 2004).

Stress shielding and bone remodelling: In the implanted femur, the implant carries bulk of the load, which was formerly transferred by the bone itself. This shields the bone from the mechanical load and brings forth abrupt changes in the mechanical environment within the bone, eventually causing bone resorption and osteolysis. This phenomenon, known as adaptive bone remodelling, is related to long-term failure of THA.

In brief, the multi-axial stresses (tensile, compression and shear) generated at the material interfaces should not be high enough to initiate interface failure. Moreover, implant-induced adverse changes in stress distribution should be reduced, such that a more biologic load transfer, as prevailing in the natural bone, may be attained. Furthermore, the initial migration and implant-bone relative displacements, especially for cementless implants, should also be minimal.

1.4.4 Bone remodelling

Bone is capable to adapt its structure, both externally and internally, in response to change in mechanical loading by bone apposition (formation) and bone resorption (loss) through the activities of cells called osteoblasts and osteoclasts, respectively.

This process is known as bone remodelling. Internal bone remodelling refers to changes in internal morphology, whereas external remodelling is described as occurring relative to periosteal geometry. Internal remodelling, thus, is expressed as a change of density or porosity of the cancellous bone (Carter *et al.*, 1989; Huiskes *et al.*, 1987), whereas the geometric (or shape) changes of the cortical bone are referred to as external remodelling (Hart *et al.*, 1984; Hart and Davy, 1989). For an adult, the cancellous bone usually has a higher rate of metabolic activity and hence, appears to respond more rapidly to changes in mechanical loads than the cortical bone (García *et al.*, 2002). Therefore, the geometry changes are considerably less in comparison to internal adaptation or internal remodelling of the bone. The combined effect of internal and external remodelling models has been investigated by several researchers (Beaupré *et al.*, 1990a; Fridez *et al.*, 1998; García *et al.*, 2001; Huiskes *et al.*, 1987; Weinans *et al.*, 1993).

No net changes in bone morphology may be observed in the natural bone, since the rate of bone resorption and that of bone apposition remain in equilibrium (Frost, 1964; Parfitt, 1984; Weinans, 1991). However, any mechanical intervention disturbs the normal state of equilibrium between apposition and resorption processes, and subsequently, the bone tries to reach a new state of equilibrium. Surgical reconstruction of bone with prosthesis alters the mechanical environment within the bone by changing the load transfer mechanism. As a result, the prosthesis starts to share the joint load in the implanted situation, which was otherwise carried exclusively by the bone in the pre-operative stage.

Bone remodelling process is known to be regulated by external loading condition (Duncan and Turner, 1995; Huiskes *et al.*, 1989; Mullender *et al.*, 2004; Nomura and Takano-Yamamoto, 2000; Turner and Pavalko, 1998). Way back in the 17th century, the famous Italian physicist, Galileo Galilei first observed a certain relationship between mechanical forces (body weight) and bone morphology (as cited by Treherne, 1981; Carter, 1984). Considerable scientific interest developed thereafter to describe the relation between the bone structure and its function. A notable contribution came from Wolff (1892), who further developed the theory of functional adaptation originally conceived by Roux (Roesler, 1981; Roux, 1881). These studies concluded that the combination of bone apposition and resorption is a biological control process that depends on the local state of stress (Roux, 1881). According to Wolff's hypothesis, every change in the form and the function of a bone is followed

by certain definite changes in their internal architecture and equally definite secondary alterations in their external geometry, based on certain mathematical laws (Wolff, 1892). This 'law of bone transformation' by Wolff was later referred to as 'Wolff's Law'. A series of investigations followed to describe this law in terms of mathematical formulation and to further quantify the process of bone remodelling (Beaupré *et al.*, 1990a; Cowin and Hegedus, 1976; Doblaré and García, 2002; Fyhrie and Carter, 1986; Hart *et al.*, 1984; Hart and Davy, 1989; García *et al.*, 2001; Huiskes *et al.*, 1987; Jacobs *et al.*, 1997).

The mathematical formulations were based on the common assumption that bone has certain sensing capability to measure the change in internal mechanical conditions (stimulus) within it and respond to that change (combined with other biological factors) by the action of osteoblasts and osteoclasts. Although most of these models considered apparent density (ρ) as variable to represent the remodelling state, the choice of mechanical stimulus varied depending on the model. Several diverse mechanical stimuli have been defined as a function of strain, stress, strain energy density (SED), elastic strain energy per unit bone mass to predict bone adaptations (Carter et al., 1989; Cowin and Hegedus, 1976; Fyhrie and Carter, 1986; Huiskes et al., 1987; Weinans et al., 1992). Cowin and Hegedus (1976) proposed a 'site-specific' remodelling objective, described as a normalisation of the active local strain values to the strain values occurring under normal physiological conditions at the same locations. According to this model, the amount of local bone resorption or apposition depends on the local differences in strains between an actual and the corresponding normal situation. In a 'non-site specific' theory, Fyhrie and Carter (1986) proposed that the tissue strives to optimize its state of stresses and strain to a uniform stimulus level over its entire volume. They also suggested that the SED normalised to the apparent density can be used as the mechanical stimulus for bone adaptation. Huiskes et al. (1987) predicted bone adaptation phenomenon around intramedullary implants using the local SED as the remodelling signal.

Bone, however, does not respond to small deviations in the mechanical stimulus (Frost, 1964). A bare minimum threshold value of the inhibitory signal, the difference between the mechanical stimuli for altered and natural situations, is required for the bone remodelling process to initiate (Huiskes *et al.*, 1987). The range of values of mechanical stimuli, where bone does not respond, is known as the 'dead zone' or

'lazy zone'. Huiskes *et al.* (1987) accounted for an inclusion of the 'dead zone' in their mathematical formulation for bone remodelling. Iterative computer-simulation procedures, in combination with the FE analysis, were described to predict adaptive bone remodelling around prostheses (Huiskes *et al.* 1987; van Rietbergen *et al.* 1993). Nevertheless, these theories were based on an assumption that bone is an isotropic material.

Trabecular bone orientation is heterogeneous, and hence, can be described by anisotropic continuum material properties. Using anisotropic strain data reported by Carter (1978), Beaupré et al. (1990a; 1990b) defined a daily tissue level stress stimulus as the equilibrium state for a time dependent remodelling theory. An anisotropic model was developed by Jacobs et al. (1995a; 1995b; 1997) based on density adaptation and anisotropy reorientation, considering the principal stresses as the mechanical stimulus. Damage-based theoretical models have also proved to be capable of successfully predicting some aspects of bone remodelling (Prendergast and Taylor, 1994; Prendergast and Huiskes, 1995). Doblaré and García (2002) proposed an anisotropic bone remodelling theory based on damage repair theory, where microdamage in the bone surface was considered to be the remodelling stimulus. Moreover, combined strain and microdamage has also been proposed as remodelling stimulus (McNamara and Prendergast, 2007). Several studies have investigated structural topology optimization for bone remodelling simulation (Hollister et al., 1994; Fernandes et al., 1999; Bagge, 2000; Jang and Kim, 2008; Jang et al., 2009). More recently, a 3-D orthotropic adaptation algorithm was proposed and implemented on an FE model of femur, wherein bone was modelled as optimised strain-driven continuum with local orthotropic symmetry (Geraldes and Phillips, 2014).

1.4.4.1 Adaptive bone remodelling: numerical simulations

The mathematical formulation of bone remodelling is based on 'Wolff's law' (Wolff, 1892). The well-known theory of adaptive bone remodelling assumed the 'elastic strain energy per unit mass of bone' as the corresponding mechanical stimulus (Cowin and Hegedus, 1976; Carter *et al.*, 1989; Huiskes *et al.*, 1987). A detailed description of the numerical simulation based on FE models of intact and implanted bone is provided in the following paragraphs.

The reference stimulus S_{ref} of each bone element is obtained from the intact model, which is subsequently compared with the remodelling stimulus S of the

corresponding bone element of the implanted model. The amount of bone remodelling depends on the difference between *S* and S_{ref} , and the dead zone *s*. At successive iterations, a new model having updated bone material properties is obtained, which, on subsequent analysis, gives rise to a new stimulus *S*. The iterative procedure is allowed to continue until an equilibrium state in terms of density is reached. It should be noted that the elements having mechanical stimulus within the dead zone and limiting density values of 0.01 g.cm⁻³ (no-bone condition) and 1.73 g.cm⁻³ (cortical bone) are not allowed to take part in the remodelling process.

According to the theory, the reference stimulus (S_{ref}) and the remodelling stimulus (S) are the local (per element) elastic strain energy (U) per unit of bone mass averaged over a loading history (n), for an intact and an implanted bone, respectively. The mechanical stimulus for each bone element is estimated from the output of FE models. The strain energy density, U, varies in each location over time during a gait cycle, owing to variations in the hip-joint force and muscle loads. In order to take some of these variations into account, an average strain energy density, U_a , for a number of loading cases was used to calculate the remodelling stimuli (Carter *et al.*, 1989; van Rietbergen *et al.*, 1993), expressed as follows:

$$S = \frac{1}{n} \sum_{i=1}^{n} \frac{U_i}{\rho} = \frac{U_a}{\rho} \qquad ... (1.1)$$

As discussed previously, bone is unresponsive in the region known as 'lazy zone' or 'dead zone' (Beaupré *et al.*, 1990b; Huiskes *et al.*, 1987; van Rietbergen *et al.*, 1993). The region between $(1-s)S_{ref}$ and $(1+s)S_{ref}$ represents the dead zone, as



Fig. 1.9: The relationship between remodelling stimulus and rates bone resorption and apposition used in the adaptive bone remodelling simulation.

shown in Fig. 1.9. The dead zone (*s*) is usually taken as ±0.75 of S_{ref} for human (Engh *et al.*, 1992; Huiskes *et al.*, 1992). A certain threshold stimulus value is required to overcome this zone and initiate remodelling. Furthermore, for remodelling prediction, estimation of free surface area in terms of apparent density values, i.e., $(A = A(\rho))$ is required. The internal free surface area per unit volume of the whole bone, $a(\rho) = A(\rho)/V$, is calculated using Martin's assumptions (Martin, 1972). It is assumed that $a(\rho) = 0.0$ for $\rho = \rho_{max} = 1.73$ g.cm⁻³. Therefore, no remodelling occurs in the cortical bone with an apparent density of 1.73 g.cm⁻³.

The adaptive quasi-static process in the operative bone is expressed in terms of the rate of change of bone mass as follows:

$$\frac{dM}{dt} = \tau A(\rho) \left[S - (1-s)S_{ref} \right], \quad \text{if} \quad S \le (1-s)S_{ref} \quad \dots (1.2a)$$

$$\frac{dM}{dt} = 0$$
, if $(1-s)S_{ref} < s < (1+s)S_{ref}$... (1.2b)

$$\frac{dM}{dt} = \tau A(\rho) \left[S - (1+s)S_{ref} \right], \quad \text{if } S \ge (1+s)S_{ref} \qquad \dots (1.2c)$$

$$0.01 \le \rho \le 1.73 \text{ g.cm}^{-3}$$

The parameter τ is a time constant given in gm/(mm²(J/gm) per month), $A(\rho)$ is the free surface at the internal bone structure, S represents the bone remodelling stimulus. The time t is given in units of one month. The rate of change of bone mass $\frac{dM}{dt}$ can be expressed as a rate of change in the internal bone mass due to porosity change, and represented in mathematical terms as follows:

$$\frac{dM}{dt} = V \frac{d\rho}{dt} \qquad \dots (1.3)$$

where V is the volume in which the change in bone mass change takes place (the volume of the element) and $\frac{d\rho}{dt}$ is the rate of change in apparent density. Thus, by rearranging Eq. 1.2, the mathematical description of adaptive bone density change (non-zero) can be expressed as:

$$\Delta \rho = a(\rho) \left[S - (1 \pm s) S_{ref} \right] \tau \Delta t , \quad \text{if } S \le (1 - s) S_{ref} \quad \text{or } S \ge (1 + s) S_{ref} \quad \dots (1.4)$$

Using Euler's integration, Eq. 1.4 can be solved iteratively to yield a new value of apparent density. Thus, a chosen time step Δt and apparent density in each element can be determined using the following equations for the above two conditions:

$$\Delta \rho_i = \tau \ a(\rho) \{ S - (1 \pm s) S_{ref} \} \qquad \dots (1.5a)$$

$$\rho_{i+1} = \rho_i + \Delta \rho_i \qquad \dots (1.5b)$$

The integration was carried out in steps of 'simulation time scale' $\tau \Delta t$ (Weinans *et al.*, 1993). The time step (Δt) was variable and was determined in each iteration using the following equation, where the maximum bone density change in the most highly stimulated element was assumed to be equal to the half of maximum bone density ($\frac{1}{2}\rho_{max} = 0.865$ g.cm⁻³) (Weinans *et al.*, 1993):

$$\tau \Delta t = \frac{0.865}{\{a(\rho)(S - (1 \pm s)S_{ref}\}_{\max}} \qquad \dots (1.6)$$

The adaptation rate (τ) was assumed to be equal to 129.6 g.mm⁻² (J/g) per month for calculating Δt (Weinans et al., 1993). Bone is modelled as continuous material at all times, the porous structure of bone is accounted by the apparent density variable, which is related to the Young's modulus (*E*) according to $E = C\rho^{D}$ (Table 1.4), where C and D are constants; *E* is expressed in MPa and the ρ in g.cm⁻³.



Fig. 1.10: Computational scheme for iterative bone remodelling simulation.

It was observed that the size of time-step had minor influence on the predicted final configuration, as long as it is sufficiently small to guarantee convergence of the process (Weinans, 1993). However, the relationship between iterative time-step for bone density predictions and actual time is not properly known yet. Comparison of these simulation results with dual energy X-ray absorptiometry (DEXA) scans involving same human femoral specimen may lead to more precise validation, allowing iterative time-step to be related to physical time. A schematic description of iterative bone remodelling combined with FE analysis is presented in Fig. 1.10.

1.4.5 Fixation of cementless implant: primary and secondary stability

The biologic fixation is an important criterion associated with the long-term survival of cementless THA. The absence of biologic fixation has often been reported as one of the major causes of aseptic loosening of cementess hip implant (Maloney et al., 1989; Sugiyama et al., 1989; Philips et al., 1990; Pal and Gupta, 2011). A combination of both primary as well as secondary stability is necessary for long-term success of the THA. During early post-operative period, when the bone ingrowth is yet to commence, the primary stability or the mechanical stability is typically governed by the amount of bone-implant relative micromotion, induced by the physiological loading (Viceconti et al., 2006). Secondary stability indicates the implant-bone relative micromotion occurring under physiological loading conditions, after mechanical interlocking through bone ingrowth is completed. This biologic interlocking is clinically known as 'osseointegration'. On the basis of human retrieval studies, Albrektsson et al. (1981) described 'osseointegration' as the attachment of lamellar bone to implants without intervening fibrous tissue. Further animal studies and human retrieval analyses of implants have led to a better understanding of this process (Galante et al., 1971; Zweymüller et al., 1988).

Subsequent to implantation, the implant-bone interface undergoes an adaptive process of bone ingrowth upto a point, when the two parts become osseointegrated (Kienapfel *et al.*, 1999; Lintner *et al.*, 1986). The long-term mechanical stability (or secondary stability) of the prosthesis is determined by this amount of osseointegration. However, the initial post-operative micromotion (i.e., lack of primary stability) has predominant influence on the bone ingrowth, and subsequent osseointegration (Manley *et al.*, 1995; Spears *et al.*, 2000; Callaghan *et al.*, 1992). Lack of primary stability of cementless implants inhibits bone ingrowth, but promotes

fibrous tissue formation (Pilliar *et al.*, 1986; Engh *et al.*, 1992; Bragdon *et al.*, 1996; Jasty *et al.*, 1997). On exceeding certain threshold value, micromotion may lead to the formation of fibrous layer at the interface (Engh *et al.*, 1992; Jasty *et al.*, 1997). This may impair the implant-bone interface stiffness, resulting in poor secondary stability of the prosthesis. A threshold micromotion value of $40 \,\mu\text{m}$ was reported by Engh *et al.* (1992), whereas Jasty *et al.* (1997) suggested a more conservative value of $20 \,\mu\text{m}$. In essence, a desired level of primary stability should prevail postoperatively in order to promote secondary (i.e., long-term) fixation through osseointegration (Burke *et al.*, 1991; Søballe *et al.*, 1993).

1.4.6 Cementless femoral implant designs: clinical findings

A large variety of cementless femoral implants are available in the market. Barring a few, majority of these fall into three broad design categories: cylindrical, tapered and anatomic (Fig. 1.11). Early designs of femoral implants were cylindrical, with extensive coating at the distal leg of the implant. Consequently, a considerable amount of diaphyseal bone ingrowth could be attained with these implants. However, many of these designs were associated with a high rate of cortical atrophy, proximal stress shielding and bone resorption. Additionally, patients sometimes complained of thigh pain, presumably owing to elastic mismatch between the stiff stem and the biologically flexible femur (Bourne *et al.*, 1994; Lavernia *et al.*, 2004).

In order to provide an enhanced physiological proximal loading of the femur to counter stress shielding, a porous ingrowth surface was provided metaphyseally,



Fig. 1.11: Different designs of cementless femoral implant: (a) fully-coated cylindrical, (b) tapered round and (c) anatomic (adapted from Khanuja *et al.*, 2011).

while keeping the distal shape still cylindrical (Keaveny and Bartel, 1995). The porous coatings in these early stems were not applied circumferentially, but rather in patches throughout the proximal region. These designs, however, recorded significant distal osteolysis. The cause of this osteolysis was believed to be polyethylene particles; gaining access to the distal femur through channels between the areas of porous coating (Schmalzried *et al.*, 1992). This led to the development of femoral implants with circumferential proximal porous coating to eliminate access channels for particulate debris.

Distal fixation in femoral stems not only imparts immediate stabilisation, but also allows for a greater lever arm to resist torsional forces as compared to proximally

coated stems. In order to achieve distal fixation in cylindrical stems, cortical support was a requirement. Consequently, the prosthesis needed to be canal-filling, with implants having large diameter. Stem stiffness depends on the Elastic modulus of the material and is proportional to the fourth power of the diameter. Therefore, an increase in the stem diameter resulted in greater stem stiffness - a factor that has been linked to distal thigh pain and proximal stress-shielding (Vresilovic et al., 1996). A 2-year radiograph, shown in Fig. 1.12, demonstrates pronounced proximal bone remodelling due to stress shielding in a femur of a 66-year old woman, fitted



Fig. 1.12: Radiographs demonstrating bone-remodelling: (*left*) the immediate post-operative image, (*right*) the 2-year radiograph (Bugbee *et al.*, 1997).

with distal fixation type cylindrical stem (Bugbee *et al.*, 1997). The frequency of thigh pain has been reported between 1.9% and 40% in some studies, wherein the cause of pain is attributed to large stem-size, distal porous coating, and material composition (Bourne *et al.*, 1994; Lavernia *et al.*, 2004).

In order to minimize stem stiffness, some implants were designed with coronal slots within the distal third of the stem and longitudinal grooves that can enhance stem strength without increasing the diameter. Another attempt was to use proximal cancellous bony ingrowth along with three-point stem fixation to obtain immediate

stability. This led to the development of tapered stems. Clinical results of straight tapered stems with at least 10-year follow-up have been fairly good, with stem survivorship reported between 92% and 100% (Bourne *et al.*, 2001; Reitman *et al.*, 2003; Parvizi *et al.*, 2004). Moreover, the incidences of thigh pain were considerably reduced when compared with cylindrical stems.

Anatomic stems, as the name suggests, were designed to mimic the natural biomechanics of hip joint by facilitating biological load transfer. They incorporate an anteroposterior curve to match the natural bow of the patient's femur. It was hoped that the anatomic design would allow for enhanced physiological loading of the femur, thereby reducing stress-shielding and distal thigh pain. However, the results were found to be a tag discouraging, with studies indicating a higher frequency of thigh pain compared with other traditional designs, e.g., tapered or cylindrical (Campbell *et al.*, 1992; McAuley *et al.*, 1998). Nevertheless, some recent follow-up studies reported stable fixation and better durability for anatomic femoral stems (Butler *et al.*, 2005; Ferrel *et al.*, 2009; Nakamura *et al.*, 2012).

With increasing number of younger, healthier and more active patients undergoing hip replacement surgery, the method of bone preservation has become essential now-a-days (Mai *et al.*, 2010). Proximal fixation with less subsequent stress shielding has become the design focus. These implant designs typically include double-taper metaphyseal filling stems and single M-L (medial-lateral) taper stems. Each of these implants relies on metaphyseal fixation and bone ingrowth in the proximal, diaphyseal and subtrochanteric regions. A double taper design allows for 'fit-and-fill' of the metaphysis, theoretically allowing more rotational support (Howie *et al.*, 2007). Both these stem designs have an exceptional track record with minimal thigh pain and survivorship greater than 95% at 20 years (Lombardi *et al.*, 2009; Khanuja *et al.*, 2011). Apart from these, a lot of cementless stem designs have been reported to possess excellent long-term clinical and radiographic outcomes (Bojescul *et al.*, 2003; Capello *et al.*, 2003; 2006; Gul *et al.*, 2008; Guo *et al.*, 2009; Kim, 2005).

Short metaphyseal stems with minimal or no distal leg are also available. The design rationale behind these implants was to improve load transmission and to preserve femoral bone stock with no reaming and minimal broaching. These implants further address the problem of metaphyseal-diaphyseal diameter mismatch, a problem associated with some patients. However, there is a dearth in long-term follow-up

investigations on this design. In a 2-year follow-up study using short metaphyseal stem, Schmidutz *et al.* (2012) observed that the outcome was comparable to that of clinically proven conventional hip implants. They further suggested that the design might be an alternative for young patients, provided the results are confirmed by a more rigorous and longer term study.

Modern THAs are performed using femoral stems made of stainless steel, cobaltchrome alloy (Co-Cr), Ti-alloy, e.g., Ti-6Al-4V and, on a limited basis, composites having low modulus of elasticity. There is ample clinical evidence to support Ti-6Al-4V as the material of choice for cementless femoral stems (Bobyn *et al.*, 1992; Qureshi *et al.*, 2002; Kim, 2004; 2005). The primary advantages of Ti-6Al-4V over Co-Cr are relatively lower modulus of elasticity (E = 110 GPa) and greater biocompatibility. These factors result in decreased stress shielding and favourable bone ingrowth, respectively. Moreover, Ti-alloy is clinically found to be relatively inert in the physiologic environment. Currently, Ti-6Al-4V is extensively used as the cementless implant material in conjunction with hydroxyapatite coating (HAC).

1.4.7 Investigations on shape optimization of femoral implants

The shape (or geometry) of a femoral stem is known to influence the clinical outcome of a hip prosthesis (Huiskes and Boeklagen, 1988; Viceconti et al., 2001). This prompted researchers to carry out shape optimization studies on hip implants. The earliest ever study of shape optimization of femoral implants was conducted by Yoon et al. (1989), who minimized the stress concentration in the cement mantle. In their seminal work, Huiskes and Boeklagen (1989) optimized the shape of a cemented stem with the objective of minimizing the stresses in the cement layer. In this 2-D study, a numerical shape optimization (NSO) procedure was introduced to minimize strain energy density (SED) of the cement mantle at the bone-cement interface for various loading conditions. The geometric parameters were varied manually during the iterative process, using the values of the objective function. The optimal stem shape, thus obtained, was found to be narrower at the proximal side and distally tapered, with a relatively pronounced mid-stem area. The study culminated into commercialization of a novel design of cemented hip stem, called Biomet[®]. However simplistic be the model, it was the first step towards optimally designed stem providing valuable insights into future scope of study.

The study by Katoozian and Davy (2000) implemented a numerical procedure for the 3-D shape optimization of the femoral component in THA for both cemented and cementless prostheses. A parametric scheme was developed for characterizing the component geometry in terms of longitudinal and cross-sectional shape variables. The 3-D design model was integrated with a 3-D FE analysis and a numerical optimization procedure. An idealized femoral geometry and perfectly bonded interfaces were used in the study. The design objective was to uniformly distribute stresses along the bonestem interface. The effects of two different musculoskeletal loading conditions and several different objective functions were examined. The general trend in all optimization outcomes was to produce a somewhat bulky implant with a rectangular cross-section. The outcome was, however, found to be more strongly affected by loading condition, rather than the choice of objective function. The study, conducted by Chang *et al.* (2001), was based on minimizing the difference between strain energy density of the intact and the implanted femur bone using a reduced mid-stem implant design. A sensitivity analysis was performed in the study to determine influential design and environmental factors. The reduced mid-stem geometry with a short stabilizing distal tip was found to minimize the bone remodelling signal while maintaining satisfactory stability. Hip joint force orientation was found to be more influential than the effect of the controllable design variables on bone remodelling, whereas the cancellous bone Elastic modulus was found to have predominant influence on the relative implant-bone displacement.

The post-operative short-term failure scenarios associated with cementless hip stem were addressed in the study proposed by Kowalczyk (2001). The failure type considered in the study was the post-operative factors leading to improper bone ingrowth into the porous-coated implant surface, which results in formation of a permanent gap filled with fibrous tissue. The reason for this is believed to be either initial improper fitting of the implant into the medullary canal or excessive micromotions occurring at the bone-implant interface. Such micromotions can prevent osseointegration and a permanent layer of fibrous tissue is formed instead of a desired level of implant-bone bonding. Such an interfacial layer, even if mechanically stable, may lead to inflammatory reactions and bone resorption in the long run (Kowalczyk, 2001). Hence, the initial stability was taken as the objective function for the study. The study assumed that the implant fitted the medullary canal perfectly. Thus, the investigation was carried out towards minimizing the micromotions between the bone internal surface and the porous-coated part of the implant surface. Furthermore, perfect bonding condition was assumed in the porous-coated part of implant surface. A gradient-based optimization scheme was employed for the study. The optimum shape appeared to be relatively long and proximally porous-coated on about half of the stem length.

Thus far, all the shape optimization models were single-objective in nature. However, it was evident that the assessment of femoral implant design needs to account for more than one failure objective. Therefore, a multi-criteria optimization model appeared to be a better design approach. Fernandes *et al.* (2006), in the first ever multi-objective study of hip prosthesis, proposed a 2-D FE model for shape optimization of cementless hip stems by minimizing relative displacement and stress on bone-stem interface. In order to incorporate several daily life activities, multiple loads were considered in the study. Geometric parameters at some selected cross sections were identified as design variables. The parameters were subjected to geometric constraints to ensure a clinically admissible shape. The implant-bone construct was considered a structure in equilibrium and contact condition at the interface was assumed. Different lengths of porous coating were analysed. The optimization problem was solved numerically using a steepest descent algorithm.

In a more sophisticated study on shape optimization of cementless stem, Ruben *et al.* (2012) developed a 3-D shape optimization procedure to design prostheses with a better initial stability and less proximal bone loss. The proposed model used a multicriteria formulation that allowed the simultaneous minimization of three important criteria for uncemented implants: relative tangential displacement, contact stress and proximal bone loss. The optimization method considered bone and interface conditions immediately after the surgery. However, a concurrent model for bone remodelling and osseointegration was also used to study the long-term behaviour of the optimized stem shapes. Stem shapes, thus obtained, were found to be contradictory. The minimization of displacement criterion led to stems with wedge-design and rectangular sections to improve axial and rotational stability, respectively (Min *et al.*, 2008). The minimization of contact stress criterion led to small stem tips to avoid direct contact with the cortical bone (Romagnoli, 2002). With the minimization of remodelling criterion 'minimally invasive stems' were obtained (Niinimaki *et al.*, 2001). The non-dominated solutions had a combination of geometric characteristics depending on the weight coefficients imposed on the objectives. The trade-off designs were found to predict better performance compared to the initial prosthesis geometry. The long-term analysis was in agreement with some clinical observations; totally coated stems were found to have more stability, whilst the partially coated designs yielded less bone loss (Sluimer *et al.*, 2006).

The genetic algorithm (GA) was first used by Ishida *et al.* (2011) to solve shape optimization problem for a cemented prosthesis. The multi-objective analysis considered the largest maximum principal stress of the cement mantle, proximal and distal, as two separate objective functions to be minimized. Two boundary conditions, walking and stairs climbing were considered separately. Hence, a total of four objective functions were evaluated. A total of 10 design parameters were used for the parametric 3-D model of the hip implant. The parameters were subjected to geometric constraints. A cement mantle of uniform thickness (2 mm) was added around the stem. The neighbourhood cultivation genetic algorithm (NCGA) was introduced to solve the multi-objective optimization problem. The objective functions were found to be mutually contradictory, for both walking and stairs climbing conditions. Five dominant stem shapes were chosen as the Pareto-optimal solutions. The optimization method was proposed for minimization of the chances of cement mantle fracture.

1.5 Evolutionary computing and genetic algorithms

Earlier computer scientists were as much interested in developing computers as in understanding the biological processes of evolution. In their quest for finding means to imbue computer programs with artificial intelligence, several scientists independently studied evolutionary systems with a vision that evolution could be used as an effective optimization tool for engineering problems. The primary objective of all these systems was to evolve a population of candidate solutions to a given problem using genetically inspired operators. Rechenberg (1965; 1973) introduced a method called 'evolution strategies' to optimize real-valued parameters for devices such as airfoils. This idea was further developed by Schwefel (1977). Fogel *et al.* (1966) implemented a technique called 'evolutionary programming', in which candidate solutions were represented as finite-state machines, which were evolved through random mutation and natural selection. In the 1950s and the 1960s, several other investigators developed algorithms inspired by natural evolution for optimization and machine learning (Box, 1957; Friedman, 1959; Bledsoe, 1961; Bremermann, 1962;

Reed *et al.*, 1967). In addition, a significant number of studies involved use of computers to simulate evolution for the purpose of controlled experiments (Fraser, 1957a; 1957b; Martin and Cockerham, 1960; Baricelli, 1962).

Genetic algorithms (GAs) were invented by John Holland in the 1960s, and later further developments were carried out by his group at the University of Michigan. GAs are heuristic search techniques developed on the basis of natural selection and are routinely used as an optimization tool for a variety of engineering problems, as well as problems in other fields. In a GA-based optimization, a population of candidate solutions (or individuals) is evolved towards better solutions while passing through a series of operations. The evolution starts from a population of randomly generated individuals, with the population in each iteration called a 'generation'. The fitness of every individual in the population is evaluated in each generation, while the fitness is usually the value of the objective function in the optimization problem to be solved. The fitter individuals are preferred from the current population based on suitable selection scheme, and each individual's genome is modified using genetic operators, cross-over and mutation, to form a new generation. The new generation of candidate solutions is then used in the next iteration (or generation). The algorithm is set to terminate when either a maximum number of generations has been attained, or a satisfactory fitness level has been reached for the population. The solutions are typically encoded in binary as strings of 0s and 1s; however, other types of encoding are also used.

Whilst a combination of evolution strategies, evolutionary programming and genetic algorithms form the backbone of the field of evolutionary computation, genetic algorithms, in particular, garnered extensive popularity over the years owing to its vast applicability. Many modern-day engineering problems demand searching through a large number of possibilities for optimum solutions. Moreover, there is further requirement for an algorithm to be adaptive for consistent performance under changing environment. Biological evolution has been an appealing source of inspiration for addressing these problems. In this natural process, fittest individuals or organisms, that are able to survive and reproduce in their environment, are chosen as desired 'solutions' from an enormous set of genetic sequences. Seen in this light, the evolution can also be conceived as a method for designing innovative solutions to complex problems by searching through a constantly changing set of possibilities. GA

search mechanisms. The inherently adaptive nature of GA, thus, makes it perform more consistently compared to the traditional optimization tools.

1.6 Motivation of the study: unsolved problems

Although several shape optimization models were studied and new femoral stem designs were proposed, there were several limitations in those studies (Huiskes and Boeklagen, 1989; Kowalczyk, 2001; Fernandes *et al.*, 2006; Ruben *et al.*, 2012). Most of these studies were based on idealised femoral bone model, consisting of one distinct layer for cortex and one homogeneous core of cancellous bone (Fernandes *et al.*, 2006; Ruben *et al.*, 2012; Ishida *et al.*, 2011). Such an idealised bone model does not represent the cancellous bone heterogeneity, since it has been observed that the bone material property varies considerably across the proximal part of the bone.

The biomechanical causes of failure may sometimes be mutually conflicting in nature (Kuiper and Huiskes, 1997). Consequently, solution of one problem has been reported to trigger another (Khanoki and Passini, 2012). Therefore, it is necessary to implement multi-objective shape optimization as a pre-clinical tool to tackle the design conflict, and subsequently attain trade-off implant designs by judiciously compromising on the design objectives. Nonetheless, most of the existing shape optimization studies on femoral implants were single-objective in nature (Yoon *et al.*, 1989; Katoozian and Davy, 2000; Kowalczyk, 2001).

The existing parameterization schemes for defining 3-D implant geometry (Katoozian and Davy, 2000; Ruben *et al.*, 2012; Ishida *et al.*, 2011) too have certain limitations with regard to exploring vast varieties of implant shapes. The search for an optimal geometry calls for extensive exploration of all possible and admissible shapes. In order to explore such vast possibilities, more number of design parameters need to be introduced, which would further enhance the design complexities. Such complexities may not be suitably handled by traditional optimization tools and requires more robust search technique to be implemented.

Shape optimization problems were further reported to be typically non-convex and characterized by multiple local optima (Fraternali *et al.*, 2011). Compared to a traditional optimization method, such as the steepest-descent method, a robust heuristic method, such as the GA, has the advantage of complex multi-variable analysis and its search for optimum values does not require a gradient function.

Therefore, the chance of a GA-solution being trapped into a local minimum is diminished. The single solution approach of Fernandes *et al.* (2006), while searching for optimization direction, was also criticized in a recent study by Ishida *et al.* (2011). The GA, on the other hand, presents a multi-solution approach to the optimization problem without requiring knowledge of the search environment. Thus, the GA seems to be a more appropriate optimization tool for solving such a multi-variable space optimization problem. The multi-criteria optimization study by Ruben *et al.* (2012) considered imposing weight coefficients for setting relative priorities on the objective functions. However, such approaches may fail to attain Pareto-optimality due lack of selection pressure on the solutions (Deb *et al.*, 2002). Multi-objective GAs, such as non-dominated sorting genetic algorithms-II (NSGA-II), have proven capabilities to address such issues.

The importance of primary stability of the implant-bone interface with regard to the hip stem design has already been discussed. Extensive preclinical investigations on various hip implant designs, considering primary stability as design objective, have scarcely been carried out. The design evaluation of thousands of new implant models involves remeshing on each implanted model; a task which is manually intensive in nature (Harrysson *et al.*, 2007; Bah *et al.*, 2011; Abdul-Kadir, 2014). Moreover, solving the non-linear finite element (FE) model to assess the corresponding implantbone relative micromotion is computationally expensive and time consuming. Therefore, the development of a predictive mathematical model, with no recourse to FE analyses, may be endeavoured in order to identify the relationship between the design parameters and post-operative micromotion of hip implant. Finally, a hybrid intelligent framework, comprising of both GA and neural network (NN), may be employed to build a multi-objective optimization scheme in order to gain further insights into the optimal design of hip stem. However, there is a scarcity of such investigations in the field of shape optimization of cementless THA.

1.7 Objectives and scope of the thesis

The primary goal of the study is to find optimally designed cementless femoral stems for better performance and durability, using a 3-D multi-objective shape optimization scheme, aided by a hybrid intelligent framework comprising of the GA and NN. The generic biomechanical failures related to cementless THA were addressed in an attempt to evolve improved design of femoral implant. The study was based on static analysis considering multiple musculoskeletal load cases, representing normal walking and stairs climbing. Final design outcomes were reassessed using an evolutionary interfacial condition by simulating the combined effect of bone remodelling and implant-bone interface adaptation due to bone ingrowth. The present investigation consists of the following objectives, which collectively contribute towards achieving the goal of the study.

- Development and experimental validation of FE models of intact and implanted femur using digital image correlation (DIC).
- Multi-objective shape optimization of cementless femoral implant based on minimization of stress shielding and interface stresses.
- A combined neural network and genetic algorithm based approach for optimally designed femoral implant having improved primary stability.
- Effects of interfacial conditions on shape optimization of cementless femoral implant: a study based on hybrid intelligent framework.
- Assessment of optimally designed femoral implant based on evolutionary interfacial conditions.

1.8 Structure of the thesis

This study presents a novel custom-based multi-objective shape optimization scheme for cementless femoral implant that employs genetic algorithms (GA) as the optimization tool. Mechanically induced post-implantation adverse complications, such as excessive implant-bone interface stresses, bone resorption due to stress shielding and initial micromotion, were minimized in the optimally designed implants and important conclusions on the favourable geometric features of the implants were discussed. A general introduction, including literature review, motivation, and objectives of the study is presented in Chapter 1. The scope of other chapters of the thesis, which collectively contributes towards achieving the primary goal of this study, is presented in the following order.

In Chapter 2, FE predictions of surface strains in intact and implanted composite femurs were verified using DIC. Relationships were sought between post implantation strain states and clinically observed longer-term bone density changes. An elaborate description on the development of the 3-D FE models of the femurs (implanted and intact) is presented, with necessary details of geometry, meshing, material properties, interface conditions, and loading and boundary condition data. Once validated, the FE model generation technique was adopted for further numerical investigations in the subsequent chapters of the thesis.

Chapter 3 presents a customised approach to develop a fully automated 3-D multi-objective shape optimization scheme for cementless femoral implant design by integrating FE analysis and the GA. The FE model of the femur bone was developed and subsequently, heterogeneous bone material properties were assigned element-wise based on a subject-specific CT-scan dataset. Two biomechanical failure criteria associated with cementless THA, stress shielding induced proximal bone resorption and excessive interface stresses, were minimized simultaneously and optimal femoral stems were assessed, subsequently, based on adaptive bone remodelling algorithm.

Chapter 4 presents a predictive mathematical model based on back-propagation neural network (BPNN) to relate femoral stem design parameters to the post-operative implant-bone micromotion. The sample data set used for training the BPNN was obtained from multiple FE analyses of bone-implant constructs for a range of implant designs. Unlike bonded implant-bone interfacial condition used in Chapter 2, frictional contact was assumed for obtaining sliding micromotion data at the interfacial nodes. Once the BPNN was trained and validated, a single-objective mixed-integer GA-search was carried out to seek for the optimal stem geometry that would minimize micromotion.

The conflicting design outcomes arising from the previous two chapters were reconciled in a multi-objective shape optimization in Chapter 5, by introducing a novel hybrid intelligent system, comprising of the GA, NN and FE analysis. In this chapter, bonded interfacial condition was used to analyse the long-term failure objectives, stress shielding and interface stresses, whereas the BPNN developed in Chapter 3 was used to predict micromotion, based on the implant geometry. From the cluster of Pareto-optimal solutions, two dominant trade-off stem geometries were chosen for further analysis.

In Chapter 6, a trade-off implant model was assessed based on an evolutionary interfacial condition and the results were compared with a generic design of femoral implant. The entire simulation accounted for the combined effect of adaptive changes in bone density (bone remodelling) and bone ingrowth, which eventually influenced the evolutionary interfacial characteristics. Several rule-based criteria were adopted to predict bone ingrowth onto the cementless implant, assuming a fully-coated implant surface. The internal bone remodelling simulation was carried out based on the adaptive bone remodelling theory.

Finally, in Chapter 7, the significance and conclusions of the study, as a whole, are presented. Based on the results of each chapter, conclusions pertaining to favourable design outcomes of cementless femoral implants have been discussed in detail. Furthermore, a retrospective review and recommendations for future research on the shape optimization of cementless femoral implants have been presented.