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A biomechanical investigation of the surface stress of a synthetic femur using infrared thermography validated by strain gauge measurements

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A BIOMECHANICAL INVESTIGATION OF THE SURFACE STRESS OF A SYNTHETIC FEMUR USING INFRARED THERMOGRAPHY VALIDATED BY STRAIN GAUGE MEASUREMENTS

By

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in partial fulfilment of the

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Degree of Master of Applied Science

in the Program of

Mechanical Engineering

Toronto, Ontario, Canada, 2011

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Abstract

A BIOMECHANICAL INVESTIGATION OF THE SURFACE STRESS OF A SYNTHETIC FEMUR USING INFRARED THERMOGRAPHY VALIDATED BY STRAIN GAUGE MEASUREMENTS

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June 2011

As the North American population ages, there will be a massive increase in musculoskeletal impairments because these problems are most common in the elderly. A very common condition is osteoporosis, which can result in fractures. Therefore, the need for improved orthopaedic fracture repair implants is vital. Currently, the two main approaches in studying orthopaedic implants are strain gauge measurements and finite element modelling. This study introduces and validates a relatively new, non-destructive approach in analysing stress patterns in a biomechanics application. Lock-in infrared (IR) thermography calibrated with strain gauges was used to investigate the stress and strain patterns of a synthetic femur under dynamic loading. The femur was instrumented with strain gauges and tested using axial average forces of 1500N, 1800N, and 2100N at an adduction angle of 7 degrees to simulate the single-legged stance phase of walking. Three dimensional surface stress maps were obtained using an IR thermography camera. Results showed a good agreement of IR thermography versus strain gauge data with a Pearson correlation of $R^2 = 0.99$ and a slope ranging from 0.99 to 1.08, based on thermoelastic coefficient (K_m) ranging from $1.067 \times 10^{-5}/\text{MPa}$ to $1.16 \times 10^{-5}/\text{MPa}$, for the line of best fit. IR thermography detected bone peak stresses on the superior-posterior side of the femoral neck of 91.2MPa (at 1500 N), 96.0MPa (at 1800 N), and 103.5MPa (at 2100 N). There was strong correlation between IR measured stresses and force along the anterior ($R^2 = 0.87$ to 0.99), posterior ($R^2 = 0.81$ to 0.99) and lateral ($R^2 = 0.89$ to 0.99) surface. This is the first study to provide an experimentally validated three dimensional stress map of a synthetic femur using IR thermography.

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TABLE OF CONTENTS

Abstract.....	iii
TABLE OF CONTENTS.....	vi
LIST OF FIGURES	ix
LIST OF TABLES	xi
MEDICAL TERMINOLOGY	xii
LIST OF ABBREVIATIONS.....	xv
NOMENCLATURE	xvi
CHAPTER 1: INTRODUCTION.....	1
1.1 Background Motivation.....	1
1.2 Musculoskeletal Conditions Statistics in North America	2
1.2.1 The Cost of Musculoskeletal Diseases	2
1.2.2 Osteoporosis.....	3
1.2.3 Breakdown of Fractures Treated by Anatomical Site.....	4
1.2.4 Treatment of Hips	5
1.3 Research Question and Goals.....	6
1.4 Current Thesis Outline	6
CHAPTER 2: LITERATURE REVIEW	7
2.1 The Hip.....	7
2.1.1 Anatomy of the Hip	7
2.1.2 Acetabulum	8
2.1.3 Femoral Head.....	8
2.1.4 Femoral Neck.....	8
2.1.5 Gait Cycle	10
2.1.6 Motion of the Hip.....	12
2.1.7 Range of Forces on Hip Joints during Routine Activities	14
2.2 Testing Methods in Biomechanical Analysis.....	15
2.3 Infrared Thermography	18
2.3.1 Background	18
2.3.2 Theory of Thermoelasticity.....	20

2.3.3	Lock-in Processing Principle	21
CHAPTER 3: METHODS AND MATERIALS		24
3.1	General Approach	24
3.2	Specimen Selection	25
3.3	Specimen Preparation.....	26
3.4	Static Axial Loading.....	27
3.5	Dynamic Axial Loading Test	29
3.6	Stress and Strain Measurements.....	29
3.6.1	Lock-In Thermography	30
3.6.2	Strain Gauges	32
CHAPTER 4: RESULTS.....		37
4.1	Axial Stiffness Tests.....	37
4.2	IR Stress Maps	37
4.3	Maximum Stress Regions of the Femur.....	40
4.4	Strain Gauge Results	42
4.5	Comparison of IR Stress Results with Strain Gauge Results.....	43
4.6	Thermoelastic Coefficient Calculation using Experimental Strain Gauge Values and IR Temperatures.....	45
4.7	Correlation of Stress and Axial Force	46
4.7.1	Correlation of IR Stresses and Axial Force	47
4.7.2	Correlation of Strain Gauge Stresses and Axial Force	49
4.8	Femur fracture under Dynamic Loading.....	49
CHAPTER 5: DISCUSSION.....		51
5.1	General Findings from Current Study	51
5.2	Comparison of Current Results with Previous Studies	51
5.2.1	Axial Stiffness.....	51
5.2.2	Strain Gauge.....	52
5.2.3	IR Stress Maps	52
5.2.4	Thermoelastic Constant of Synthetic Femur.....	53
5.2.5	Femoral Fracture.....	54
5.3	Practical Implications	54

5.3.1	Biomechanical Testing Implications.....	54
5.3.2	Clinical Implications.....	56
5.4	Limitation/Sources of Error and Future Work.....	57
CHAPTER 6: CONCLUSION		60
APPENDICES		61
A.1	Static Axial Stiffness.....	61
A.2	Dynamic Axial Stiffness	62
A.3	Material Safety Data Sheet Black Dye.....	64
A.4	Technical Data Strain Gauge.....	67
A.5	Variation in Material Properties of Synthetic Bones Based on Generation.....	68
REFERENCES		69

LIST OF FIGURES

Figure 1.1: Breakdown of medical conditions in developed countries of people over 50 years	1
Figure 1.2: Productivity days lost due to medical conditions	2
Figure 1.3: Osteoporosis rates in men and women	3
Figure 1.4: Breakdown on types of fractures treated by anatomical site.....	4
Figure 1.5: Number of joint replacement procedures from 1991-2004	5
Figure 2.1: The hip joint	7
Figure 2.2: Conditions of femoral neck in the frontal plane.....	9
Figure 2.3: Conditions of femoral neck in the transverse plane	9
Figure 2.4: Medial and lateral trabeculae systems of the femoral neck.....	10
Figure 2.5: The human gait cycle	11
Figure 2.6: Sequence of the typical gait cycle	12
Figure 2.7: Typical motion ranges of the hip joint	13
Figure 2.8: Range of extension and flexion of the hip during the gait cycle.....	14
Figure 2.9: A bone under axial compression, torsion and 4-Point bending.....	16
Figure 2.10: Schematic overview of lock-in thermographic system	19
Figure 2.11: Representation of varying load and temperature under dynamic conditions	20
Figure 2.12: Signal analysis of lock-in thermography.....	22
Figure 2.13: Signal acquisition and processing in creation of thermal wave cycle.....	22
Figure 3.1: Axial loading of femur at 7 degrees of adduction	24
Figure 3.2: Fourth generation large left femur.....	25
Figure 3.3: Detailed geometry of large femur	26
Figure 3.4: Procedure for potting a femur	27
Figure 3.5: Instron 8874 mechanical loading system	28
Figure 3.6: Mechanical representation of the ball-socket joint system of the hip	28
Figure 3.7: Sinusoidal loading of the femur	29
Figure 3.8: FLIR SC5000 Silver 420 Camera	30
Figure 3.9: Lock-in thermography setup	31
Figure 3.10: Various sides of the proximal femur (a) posterior, (b) anterior, (c) lateral.....	32
Figure 3.11: Rectangular rosette strain gauge.....	33
Figure 3.12: Strain gauges positioned on posterior region of femur	33
Figure 3.13: Soldering of wire to gauge	34
Figure 3.14: DSub-15 connector.....	34
Figure 3.15: DSub connectors linked to data acquisition system	35
Figure 4.1: IR Stress image of the anterior, posterior and lateral sides of a femur under an average load of 1500N	38
Figure 4.2: IR Stress image of the anterior, posterior and lateral sides of a femur under an average load of 1800N	38

Figure 4.3: IR Stress image of the anterior, posterior and lateral sides of a femur under an average load of 2100N	39
Figure 4.4: Thermoelastic relationship between temperature and stress	40
Figure 4.5: High stress regions along the anterior side of the femur	41
Figure 4.6: High stress regions along the posterior side of the femur	42
Figure 4.7: Comparison of experimental and IR stress.....	44
Figure 4.8: Experimental Thermoelastic Coefficient.....	45
Figure 4.9: Stress comparison with IR stresses adjusted	46
Figure 4.10: Positions probed for stress on the posterior, anterior and lateral side of the femur.	46
Figure 4.11: Femoral neck fracture (anterior view).....	50
Figure A.1: Static axial stiffness test 1	61
Figure A.2: Static axial stiffness test 2	61
Figure A.3: Static axial stiffness test 3	62
Figure A.4: Dynamic stiffness for average load of 1500N	62
Figure A.5: Dynamic stiffness for average load of 1800N	63
Figure A.6: Dynamic stiffness for average load of 2100N	63

LIST OF TABLES

Table 2.1: Typical peak forces on the hip joint due to routine activities.....	15
Table 3.1: Material properties of femur.....	25
Table 3.2: Specification of the FLIR thermography system.....	30
Table 3.3: Strain gauge specifications	33
Table 4.1: Stiffness versus Loading Conditions	37
Table 4.2: Stress values from IR Images	39
Table 4.3: Temperature values from IR Images	40
Table 4.4: Experimental strain values from rosette strain gauges	43
Table 4.5: Experimental principal stresses	43
Table 4.6: Experimental versus IR stress.....	44
Table 4.7: Experimental stress (MPa) and respective IR temperature.....	45
Table 4.8: Correlation of posterior surface stresses from IR thermography versus average axial force	47
Table 4.9: Correlation of anterior surface stresses from IR thermography versus average axial force	48
Table 4.10: Correlation of lateral surfaces stresses from IR thermography versus average axial force	48
Table 4.11: Correlation of stresses from strain gauges versus average axial force	49

MEDICAL TERMINOLOGY

abduction	To draw away from the midline of the body or from an adjacent part or limb
acetabulum	The cup-shaped cavity at the base of the hipbone into which the ball-shaped head of the femur fits
adduction	The movement of a limb toward the midline or axis of the body
antegrade	Performed in the normal direction of flow
anterior	Pertaining to a surface or part situated toward the front or facing forward
anterversion	The angulation created in the transverse plane between the neck of the femur and shaft of the femur. The normal angle is between 15 and 20 degrees during stance
arthritis	Inflammation of a joint, usually accompanied by pain, swelling, and stiffness, resulting from infection, trauma, degenerative changes, metabolic disturbances, or other causes
articulation	The place of anatomical union, usually movable, between two or more bones
cancellous	Lattice like, porous, spongy. Cancellous tissue is normally present in the interior of many bones, where the spaces are usually filled with marrow
cartilage	A tough, elastic connective tissue found in the joints, outer ear, nose, larynx, and other parts of the body
celiac disease	Disease of the digestive system that damages the small intestine and interferes with the absorption of nutrients from food
condyle	A rounded projection at the end of a bone that anchors muscle ligaments and articulates with adjacent bones
contralateral	Affecting or originating in the opposite side of a point of reference, such as a point on a body

cortical	Pertaining to or emanating from a cortex, usually bone
coxa valga	Deformity of the hip with increase in the angle of inclination between the neck and shaft of the femur
coxa vara	Deformity of the hip with decrease in the angle of inclination between the neck and shaft of the femur
diaphysis	The shaft of a long bone
distal	Away from or the farthest from a point of origin or attachment
epidemiology	The study of the determinants of disease events in populations
epiphysis	The expanded articular end of a long bone
eversion	A turning outward or inside out, such as a turning of the foot outward at the ankle
extension	The act of straightening or extending a flexed limb
femur	The long bone of the thigh, and the longest and strongest bone in the human body, situated between the pelvis and the knee and articulating with the hipbone and with the tibia and patella
flexion	The act of bending or the condition of being bent
fracture	A break or rupture in a bone
gait	The manner or style of walking
in vitro	A procedure performed not in a living organism but in a controlled environment
in vivo	Experimentation using a whole, living organism
inversion	A turning inward, inside out, or other reversal of the normal relation of a part
lateral	Relating to or situated at or on the side
lunate	Shaped like a crescent

medial	Pertaining to, situated in, or oriented toward the midline of the body
morbidity	A diseased condition or state
mortality	The condition of being subject to death
musculoskeletal	Relating to or involving the muscles and the skeleton
osteoporosis	A disease characterized by decrease in bone mass and density
posterior	In the back part of a structure
proximal	Nearer to a point of reference such as an origin, a point of attachment, or the midline of the body
retrograde	Moving backward or against the usual direction of flow
retroversion	A turning or tilting backward
tibia	The inner and larger of the two bones of the lower leg, extending from the knee to the ankle, and articulating with the femur, fibula, and talus

LIST OF ABBREVIATIONS

AA	Antegrade Nailing
BW	Body Weight
FEA	Finite Element Analysis
IR	Infrared
IRT	Infrared Thermography
RA	Retrograde Nailing
TSA	Thermoelastic Stress Analysis

NOMENCLATURE

Q	Heat generated per unit volume during deformation of the material (Watt/m ²)
T_o	Ambient temperature (°Kelvin)
$\frac{\partial w}{\partial T}$	Mechanical work with respect to temperature (J/°K)
ρ	Density (kg/m ³)
C_p	Specific heat capacity (constant pressure) (J/kg.°K)
k	Conduction (W/m.°K)
r	External source supply (W/m ²)
S_{the}	Thermoelastic source (W/m ²)
dI	Intrinsic dissipation (W/m ²)
K_m	Thermoelastic coefficient (1/Pa)
α	Coefficient of thermal expansion (1/°K)
$\Delta(\sigma_1 + \sigma_2)$	Change in sum of principal stresses (Pa)
S	Image signal
Φ	Phase angle (°)
ε_p	Maximum principal strain (Pa)
ε_Q	Minimum principal strain (Pa)
σ_p	Maximum principal stress (Pa)
σ_Q	Minimum principal stress (Pa)
ν	Poisson ratio
E	Elastic modulus (Pa)

CHAPTER 1: INTRODUCTION

1.1 Background Motivation

Musculoskeletal conditions are considered to be one of the most costly and disabling conditions faced by North Americans. To provide national recognition to the fact that musculoskeletal disorders and diseases were the leading cause of physical disability in the United States of America, former president George W. Bush, in March 2002 proclaimed the years 2002-2011 as the United States Bone and Joint Decade [1, 2].

As the North American population rapidly ages, there will be a massive increase in musculoskeletal impairments in the next quarter century because these problems are most common in the older segment of the population. It has been projected that the number of people over the age of 65 will double by the year 2030 [1]. These expected increases in musculoskeletal disorders will put severe financial pressure on health care services worldwide.

Currently, bone and joint disorders account for more than 50 percent of all chronic conditions in people of over 50 years of age in developed countries (Figure 1.1) [1, 3].

Figure 1.1: Breakdown of medical conditions in developed countries of people over 50 years [1]

The impact of musculoskeletal disorders does not only result in severe, long term pain and disability but also leads to loss of productivity. This loss in productivity could impact the patient's ability to work or perform routine daily activities. Musculoskeletal conditions also impact the quality of life, pain, discomfort, and disability of the patients, but it also affects relatives and friends. Direct costs of the load of musculoskeletal disease include inpatient, hospital emergency and outpatient services, physician outpatient services, other practitioner services, home health care, prescription drugs, nursing home cost, prepayment and administration and non-health sector costs. Indirect cost relates to morbidity and mortality, including the value of productivity losses due to premature death due to a disease and the value of lifetime earnings [1].

1.2 Musculoskeletal Conditions Statistics in North America

1.2.1 The Cost of Musculoskeletal Diseases

It was estimated that the average U.S. cost for treatment, during 2002 to 2004, for all patients diagnosed with a musculoskeletal disease and indirect lost wages was \$849 billion annually (Figure 1.2). About \$510 billion was attributed directly to the cost of treatment, while the remaining \$339 billion was estimated to be the indirect costs, expressed primarily as wage losses for persons aged 18 to 64 with a work history [1, 4].

Figure 1.2: Productivity days lost due to medical conditions [1]

1.2.2 Osteoporosis

One of the most common forms of musculoskeletal conditions is osteoporosis. Osteoporosis is a disease resulting in decreased bone mass and weakening of bone structure that increases the chances of a person suffering a fracture. Osteoporosis is often described as the “silent disease” or “silent thief” progressing without symptoms until an innocuous fall or minor activity fractures a bone [5]. Osteoporosis can occur without a known cause or be attributed to another secondary condition, such as hyperthyroidism or celiac disease, or to medication, such as steroids [6].

The epidemiology of osteoporosis has only been completely described in Caucasian women, therefore making it difficult to estimate the true number of persons suffering from osteoporosis. It has been confirmed recently that osteoporosis affects both men and women, and is not independent of a person’s ethnicity. The National Osteoporosis Foundation reported approximately 29.5 million women and 11.7 million men to be suffering from osteoporosis or reduced bone mass [1, 6]. According to the National Health and Nutrition Examination Survey, on average during 1999 to 2004, approximately 10.5 million women and men above the age of 65 were diagnosed with osteoporosis by their physicians (Figure 1.3).

Figure 1.3: Osteoporosis rates in men and women [1]

A rate of almost one in four women and one in twenty men above the age of 65 has osteoporosis [1, 7]. These rates were dramatically higher than those reported a decade earlier, likely due to more clinical testing of bone mass, aggressive awareness and educational efforts.

However, it is believed that osteoporosis is still extremely under-diagnosed. Approximately 16% of patients admitted with low energy fractures were diagnosed with osteoporosis. Studies have reported that falls are the leading cause of injuries in persons above the age of 65 years in the United States [1, 4, 8]. A fall resulting in a fracture is the primary cause of hospitalization or death. Osteoporosis is said to be the main root cause of low energy fractures after a fall. Almost 50% of women and 25% men above the age of 50 will have an osteoporosis-related fracture in her or his remaining lifetime.

1.2.3 Breakdown of Fractures Treated by Anatomical Site

It was reported in 2004, that almost 704,000 persons over the age of 45 sustained low energy fractures and subsequently discharged from hospital. More than 33% of the inpatient fractures were hip fractures (39% for males and 43% for females). In the same year, it was reported that almost 1.2 million fractures to persons in the same age bracket were treated in emergency rooms. Most of the emergency fractures treated were those to the wrist or hand as these are much simpler and faster to treat as compared to hip fractures (Figure 1.4) [1, 4, 8].

Figure 1.4: Breakdown on types of fractures treated by anatomical site [1]

1.2.4 Treatment of Hips

There are various methods in treating musculoskeletal conditions of the hip arising from conditions such as osteoporosis, osteoarthritis or fractures related to high impact forces caused by motor vehicle accidents. One of the most common modes of treatment is joint replacement. These are primarily used when there is a complete shattering of a joint due to a collisions or falls, or severely arthritic joints. Almost 66% of the patients admitted due to osteoarthritis undergo total hip replacement. Partial hip replacements are primarily carried out in cases with simple hip fractures. The numbers of joint replacements have steadily risen since the early 1990s, with total hip replacements showing the highest growth (Figure 1.5). Almost 90%, of the 1.07 million replacements in 2004, were either carried out on the hip or the knee and cost hospitals approximately \$30 billion. It is projected that by 2030 almost 570,000 primary total hip replacements will be performed annually [1, 4, 9, 10]. This number could rise if fractures not attributed to the aging population are taken into account. Therefore, it is imperative that more resources are diverted in researching musculoskeletal conditions to improve the longevity of implants and to reduce the significant burden and cost of affected joints on an active, aging population.

Figure 1.5: Number of joint replacement procedures from 1991-2004 [1]

1.3 Research Question and Goals

The aim of this study was to assess the potential of infrared thermography in biomechanical studies and to specifically establish a comprehensive methodology for analysing bones under dynamic loading regimes. The investigation was carried out using a large synthetic femur, undergoing dynamic axial loading conditions, and subjected to forces equivalent to those seen during normal walking.

By showing the viability of using infrared thermography as an alternative to strain gauge measurements and finite element modelling, this will allow a better understanding of the stress patterns along the surface of the bone and, in the long term, assist in improving current orthopaedic implants to speed up patient recovery times.

1.4 Current Thesis Outline

The thesis is comprised of six main chapters. The first chapter provides the motivation behind the need for this study and research goals the thesis sets out to accomplish.

Chapter 2 will introduce the reader to the field of thermography and what role thermography plays in current research. Also, a comprehensive literature review of orthopaedic biomechanical research will be presented.

Chapter 3 will focus on the methodology of the current study. A step by step approach to establish the validity of infrared thermography will be provided so that duplication of the methodology will be achievable by the reader.

Chapter 4 will provide the results of the research study. Chapter 5 will critically analyze the significance of these results by comparing them to previous studies and suggest clinical and practical implications.

Finally, the thesis will conclude with a summary chapter. This section will review the major findings of the thesis highlighting the potential of infrared thermography as an alternative to strain gauge measurements and finite element models. This chapter will also provide directions for future work.

CHAPTER 2: LITERATURE REVIEW

2.1 The Hip

The hip joint is one of the largest and most stable joints in the body. In comparison to the knee, the hip joint has intrinsic stability given by its rigid ball-and-socket system. It also has a large degree of mobility, allowing normal movement in carrying out daily activities. Therefore, any incongruency or impairment of the hip can result in altered stress distribution in the joint cartilage and bone, leading to chronic musculoskeletal conditions such as osteoarthritis. Such damage is further exacerbated by the large forces to which the joint is subjected to.

2.1.1 Anatomy of the Hip

The hip joint is made up of the head of the femur and the acetabulum of the pelvis as shown in Figure 2.1 [11, 12]. This articulation has a loose joint capsule and is encircled by muscles that are large and strong. The composition of this extremely stable joint enables a wide array of motion that are needed in carrying out daily actions such as walking, standing and running. Therefore, it is imperative that such a joint be accurately positioned and restricted.

Figure 2.1: The hip joint [11, 12]

2.1.2 Acetabulum

The acetabulum is a concave shaped part of the ball and socket arrangement of the hip joint. The surface of acetabular region is covered with articular cartilage found predominantly on the lateral side [13]. The cavity of the acetabulum faces outward, downward and obliquely forward. The acetabulum in the hip is deep into the socket to provide a high amount of static stability to the hip. The acetabulum when unloaded has a much smaller diameter than the femoral head but deforms readily around the femoral head when the joint is loaded [13, 14]. By having the ability to elastically deform, the acetabulum conforms to the femoral head and contact is made via the periphery of the anterior, superior, and posterior articular surface of the acetabulum [15].

2.1.3 Femoral Head

The proximal region of the femur is made up of the femoral head and neck. The femoral head is the convex component of the ball-and-socket joint configuration of the hip joint. Articular cartilage covers the femoral head pre-dominantly around the medial-central surface, and becomes thinner towards the periphery. Kempson et al., concludes that varying cartilage thickness results in various regions of the femoral head having different strength and stiffness [13]. Rydell proposed that loads were dominantly transmitted through the superior quadrant of the femoral head, while Von Eisenhart et al., demonstrated in his in-vitro study that loading patterns on the femoral head varied with magnitude of load [16-18]. At high loads, the load bearing area was concentrated at the center of the lunate, and at low loads the area was located at the anterior and posterior horns. It is still not clear as to how loads are distributed on the femoral head in-vivo but studies indicated that the majority of loading was transmitted by the anterior and the medial lunate regions during daily activities.

2.1.4 Femoral Neck

The femoral neck is vital to hip joint function based on its two angular relationships with the femoral shaft. Firstly, the angle of inclination of the neck to the shaft in the frontal plane (neck to shaft angle), is usually is 125° in adults, but can vary from 90° to 135° (Figure 2.2).

Conditions known as coxa vara (angles $\leq 125^\circ$) and coxa valga (angles $\geq 125^\circ$) alter the force relationships in hip joints [19].

Figure 2.2: Conditions of femoral neck in the frontal plane [19]

Secondly, is the angle of inclination in the transverse plane (angle of anteversion) is formed as a projection of the long axis and of the femoral head and the transverse axis of the femoral condyles, which usually is 12° in adults. Conditions of anteversion (angles $\geq 12^\circ$) and retroversion (angles $\leq 12^\circ$) force the femoral head to internally rotate and externally rotate respectively during gait (Figure 2.3) [20]. However, these conditions are fairly common in children and tend to disappear as they grow older.

Figure 2.3: Conditions of femoral neck in the transverse plane [20]

The structure of the femoral neck is internally composed of cancellous bone with trabeculae organized into medial and lateral systems as shown in Figure 2.4 [21].

Figure 2.4: Medial and lateral trabeculae systems of the femoral neck [21]

By showing the joint reaction force on the femoral head is parallel to the medial and lateral trabeculae systems; Frankel postulates the importance of this arrangement for supporting this force [22]. With aging, the femoral neck gradually degenerates, leading to thinning of cortical bone and resorption of the cancellous layer and degrading of the trabeculae system. This phenomenon is known as osteoporosis, which makes bones susceptible to fractures.

2.1.5 Gait Cycle

Gait, also known as bipedal locomotion, is a process by which humans are able to move about on two limbs. Lower joints provide three unique functions for locomotion. Joints have the ability to bear weight. Joints provide means of locomotion and lastly equilibrium. Bipedal locomotion is a cyclic activity consisting of two phases for each limb, namely stance and swing phase (Figure 2.5) [23]

Figure 2.5: The human gait cycle [23]

The stance phase occupies 60% of the stride and has two periods where double limb support occurs (initial and terminal), when the contralateral foot is in contact with the ground, and an intermediate period of single limb support, when the contralateral limb is engaged in the swing phase (Figure 2.6) [24]. The stance phase can be broken down into six parts. Firstly, the initial contact or heel contact is defined as the time the foot makes contact with the floor. Next is the loading response, which is the interval during which the sole of the foot comes into contact with the floor and the weight of the body is borne by the supporting limb. The loading response matches with the end of the of the initial double limb support somewhere between 10 – 12% of the stride. Mid-stance refers to the period during which the tibia, the large bone between the knee and the ankle, rotates over the stationary foot in the direction of movement. The start of mid-stance matches the single limb support and lasts from 10 – 30% of the stride. Terminal stance refers to the point of the stride during which the body weight is passed from the hind and the mid-foot regions onto the forefoot. This occurs from 30 – 50% of the stride and coincides with the starting of the terminal double limb support. While the terminal double support occurs, so does the pre-swing phase, lasting from 50 – 60% of the stride. It is seen that during the pre-swing, weight is transferred to the contralateral limb in readiness for swing phase. The termination of pre-swing coincides with toe-off, the point at which the foot breaks contact with floor, thereby signaling the start of the swing phase.

Swing phase makes up 40% of the gait cycle and can be broken down into three periods (Figure 2.6).

Gait Cycle	Stance phase 0-60%	Loading response 0-10%	Initial contact to contralateral toe-off
		Midstance	Contralateral toe-off to when the body CG is directly above the reference foot; weight loading begins
		Terminal stance	CG directly above the reference foot to contralateral initial contact; weight loading ends and heel of the reference foot leaves ground at ~35%
		Preswing 50-60%	Contralateral initial contact to toe-off
	Swing phase 60-100%	Initial swing	Toe-off to maximum knee flexion
		Midswing	From maximum knee flexion to when the tibia is perpendicular to the ground
		Terminal swing	Tibia perpendicular to the ground until initial contact; knee reaches maximum extension just before initial contact

Figure 2.6: Sequence of the typical gait cycle [24]

The initial swing lasts from approximately 60 – 73% of the stride, which is about one-third of the swing phase, from toe off until the swinging foot is opposite the stance foot. Mid-swing terminates when the tibia of the swinging limb is oriented vertically and lasts from 73 – 87% of the stride. Terminal swing makes up the last period of the swing phase and lasts from 87 – 100% of the stride. Terminal swing terminates at the moment of initial contact.

2.1.6 Motion of the Hip

The hip joint motion during gait is tri-axial: flexion-extension occurs about a mediolateral axis in the sagittal plane; adduction-abduction occurs about an anteroposterior axis in the frontal (coronal) plane; and internal-external rotation occurs about a longitudinal axis in the transverse plane (Figure 2.7) [25].

Figure 2.7: Typical motion ranges of the hip joint [25]

Motion is the highest in the sagittal plane, where the range is $0 - 140^\circ$ for flexion and $0 - 15^\circ$ for extension. Abduction ranges from $0 - 30^\circ$; whereas adduction is smaller, from $0 - 25^\circ$. External rotation (eversion) ranges from $0 - 90^\circ$ and internal rotation (inversion) ranges from $0 - 70^\circ$ when the hip joint is flexed. Rotation is restricted during hip extension due to soft tissues.

Murray demonstrated that the joint was fully flexed during late swing phase, as the limb moved forward for heel strike [26]. The joint extended during the beginning of the stance phase as the body moved forward. At heel-off, maximum extension was achieved. There was a reversal into flexion by the joint during the swing phase and reached maximum flexion, $35 - 40^\circ$, prior to heel strike (Figure 2.8).

Similar studies on the motion in the frontal and transverse plane have been carried out [27, 28]. However, as people age, the range of motion becomes limited. Murray in his study observed older men had shorter strides and a decreased range of hip flexion and extension. Johnson and Schmidt studied the range of motion during common daily activities and concluded that at least 120° hip flexion and at least 20° of abduction and external rotation were required to do routine activities without any hindrance [26].

Figure 2.8: Range of extension and flexion of the hip during the gait cycle [26]

2.1.7 Range of Forces on Hip Joints during Routine Activities

Studies have shown that substantial forces act on joints during routine activities [16, 17, 29-37]. Tables 2.1 show range of typical peak forces on the hip during routine activities. The peak resultant forces for patients with prosthesis during gait ranged from 1.8 to 4.36 times the body (Table 2.1). Bergmann et al. have carried out thorough in vivo studies on forces during routine activities. In general it was observed that the force peaked initially at early stance and peaked again in late stance [29-31].

RANGE OF TYPICAL REPORTED PEAK JOINT FORCES FOR SELECTED STUDIES		
ACTIVITY	REPORTED PEAK FORCE (BW)	REFERENCE
Walking Ascending Stairs Descending Stairs	2.7-4.3 3.4-5.5 3.9-5.1	Bergmann et al., 1993, 1995
Walking	1.8-3.3	Rydell, 1966
Walking	4.9-7.0	Paul, 1967
Walking	4.5-7.5	Crowninshield et al., 1978
Walking	5.0-8.0	Rohrle et al., 1978
Walking	2.2-2.8	van den Bogert et al., 1999
Walking Slow Speed	2.7	English et al., 1979
Walking Normal to Fast Speeds Stair Climbing	2.7-3.6 2.6	Kotzar et al., 1991
Walking Slow Speed (Crutches) Ascending Stairs	2.6 2.6	Davy et al, 1988

Table 2.1: Typical peak forces on the hip joint due to routine activities. (BW = body weight) [16, 17, 26-34]

2.2 Testing Methods in Biomechanical Analysis

Biomechanical analysis incorporates two methods frequently. Firstly, the experimental approach is carried out in-vivo and in-vitro. In vivo experimental studies have been carried out on human

subjects and have provided an accurate picture of what is clinically relevant. However issues arise when working with live subjects such as accessibility to joints to gain relevant data. To that end, mechanical in-vitro tests have been carried out on human cadaveric bones and /or longbone-implants for almost a century [38]. However, researchers have observed physiologic loading to be an intricate interaction of hard and soft tissue and thus have found it difficult in replicating real-world physiological conditions experimentally. As such, researchers have implemented simple approaches to biomechanical testing by carrying out axial compression, lateral bending, torsion, 3-point bending and 4-point bending tests independent of each other [39-54]. Figure 2.9 shows a few illustrations of biomechanical tests.

Figure 2.9: A bone under axial compression, torsion and 4-Point bending [41]

Cadaveric bones have many drawbacks in experimental testing. Firstly, acquiring bones from cadavers require consent from family of the deceased as well as regulatory ethical approvals, which can be onerous. Secondly, studies lasting several weeks can change mechanical properties as seen in investigations by McConnell et al., where axial stiffness decreased by 30% over several months [53]. Embalming of cadaver bones seems to have succeeded in preventing mechanical properties from changing, but only after specimens have been embalmed for several months [55, 56]. The most important issue that faces both cadaver and embalmed bones is the

inter-specimen variability in material and geometric properties. Studies carried out by Papini et al., showed that axial and torsional stiffnesses for intact human femurs varied by 3.3 and 3.2 times, respectively. This specimen-to-specimen inconsistency leads to a large discrepancy in measured results. Therefore, it is difficult to compare against control results, which in turn causes difficulty in drawing any useful conclusions from investigations [49].

Synthetic longbones are becoming more popular due to advantages they provide over cadaveric specimens. Synthetic longbones are easy to manufacture and have consistent geometry that is 20 – 200 times more uniform than cadaveric bones. These bones do not degrade over time, are easy to store, they are non-toxic, they are available commercially and they are relatively cheap to acquire. Moreover, these bones have been shown to imitate human bone in axial stiffness, 4-point bending stiffness, torsional stiffness, cortical screw pullout strength, and cancellous pullout strength [49, 54, 57-61].

Regarding loading regimes, quasi-static loading regimes have also been implemented to assess stiffness and strength, as well as utilization of strain gauges in analyzing strain and stress patterns, of long bones or longbone-implant constructs [40-43, 45, 48, 49, 53, 54, 62-64].

Dynamic loading regimes have also been implemented [50, 65-67]. However, most dynamic loading tests have been carried out to investigate the material properties of the bone such as its viscoelastic properties [68-70].

To measure surface strain on bone and implant specimens, strain gauges have commonly been used by many researchers [64, 71-75]. Strain gauges are relatively easy to use, cheap to obtain and provide relatively stable results. However, there are issues related to placement which could affect accuracy and reproducibility of results. Gauges have to usually be positioned on flat and smooth areas, to ensure best possible planar strain detection. Curved surfaces tend to produce unstable results due to readings being averaged over an arc length. Areas of structural discontinuities should usually be avoided due steep strain gradients where results would be meaningless. Furthermore, a major drawback is found in the limited resolution obtained from strain gauges; therefore the ability to predict the overall stress behaviour of the longbone or implant becomes impossible.

Another common computer modeling method also implemented in biomechanical analysis is finite element analysis (FEA). FEA of bone has been carried out since the early 1970s. The first model of the human longbone incorporated two-dimensional geometries and implemented homogenous, isotropic, elastic properties [76]. Models used later on in the early 1980's incorporated three-dimensional models, but often simple tube geometries were used that were poor at imitating realistic anatomical structures or carrying out complex clinical investigations [77-83]. However, recent advances in computer hardware, improved finite element modelling software, has enabled the development of more accurate models that behave similarly to actual human bone [49, 64, 84-87].

Studies suggest FEA provides a good approximation for comparing performances of various implants but for absolute predictions, clinical conditions would still need to be replicated. FEA, even though may cut down on time and costs, is limited as models get more complicated such as introducing fracture fixation devices where modelling of forces become very difficult, requiring assumptions to be made which may not be valid. For instance, boundary conditions, in FEA, at the interfacial contact regions may affect how loads are distributed between bone and implant. FE models usually assume bonded contact between the two surfaces in order to achieve perfect osseointegration, i.e. bony ongrowth around the implant. In reality, this may not occur thereby affecting results [64, 81, 84, 85, 88, 89].

Numerous studies show that strain gauge measurement in tandem with finite element analysis can compensate for each respective method's drawbacks to evaluate problems in biomechanical research [64, 71, 88, 90, 91]. However, both approaches combined still require a lot of time in biomechanical investigations.

2.3 Infrared Thermography

2.3.1 Background

Infrared thermography (IRT) is a full-field non-contact, non-destructive technique to acquire temperature variations on the surface of an object. It has been used for almost half a century to measure temperature on inaccessible, moving and very small objects [92-97]. Moreover,

researchers and engineers have used IRT for observing very high temperature and minute temperature changes as well.

Developments on detector and data processing algorithms have enabled obtaining very small temperature variations at an extremely high accuracy. Thermographic Stress Analysis (TSA) is an extended application of IRT, where such small temperatures generated by dynamic loading of materials are measured. Kelvin, in 1885, theoretically explained the thermoelastic effect as the conversion between mechanical energy and heat [98]. However, Biot further expanded on this theory and provided the basic thermodynamics behind the thermoelastic effect [98]. The thermoelastic effect occurs when changes in stresses within a material changes its volume. Energy formed in the material element is converted to a local change of temperature. If the specific heat capacity of the material is high, this conversion is very minuscule in terms of temperature change. For steel, a change in stress of 1MPa results in a change of temperature of approximately 1mK. Conversion of this mechanical energy into heat only takes place during dynamic testing conditions. Under adiabatic conditions, the equation relating elastic and thermodynamic theory of material elements is valid for isotropic materials.

TSA is based on the principle of thermodynamic energy conversion. A material under dynamic loading undergoes alternate heating and cooling, where the differences in temperature is directly equivalent to the sum of principal stresses on the material and is synchronous with the loading frequency during dynamic testing. A typical TSA system includes an infrared camera, a link between the loading machine and the camera, and a lock-in processing module (Figure 2.10). The IR camera is a highly sensitive infrared focal plane array-based detector, which converts IR energy, emitted by the testing specimen, to electrical voltage.

Figure 2.10: Schematic overview of lock-in thermographic system

2.3.2 Theory of Thermoelasticity

Figure 2.11 shows the force applied on a material and resultant temperature change of the material [93].

Figure 2.11: Representation of varying load and temperature under dynamic conditions

The theory of thermoelasticity is based on the thermomechanical framework [99]. Under cyclic loading, a material undergoes alternate heating and cooling. Thermoelasticity can be expressed using Equation 2.1:

$$Q = -T_o \frac{\partial w}{\partial T} \quad (2.1)$$

where:

$$\begin{aligned} Q &= \text{heat generated per unit volume during deformation of the material;} \\ T_o &= \text{ambient temperature; and} \\ \frac{\partial w}{\partial T} &= \text{mechanical work with respect to temperature} \end{aligned}$$

Within such a thermomechanical framework, and incorporating the first and second principles of thermodynamics, the local heat changes can be described by Equation 2.2 [99]:

$$\rho C_p \dot{T} - k \Delta^2 T = r + s_{the} + d1 \quad (2.2)$$

where:

ρ	=	mass density
C_p	=	specific heat (at constant pressure)
k	=	conduction
r	=	external source supply
s_{the}	=	thermoelastic source
$d1$	=	intrinsic dissipation

The left-hand term in Equation 2.2 is a differential operator that describes the heat energy dissipated by a system, while the right-hand side of the equation groups energy inputs to the system [99]. As the left-hand side is a diffusion equation term, the right-hand side terms are commonly considered to be heat sources.

Equation 2.3 is derived for linear, isotropic, homogenous materials loaded so that when adiabatic conditions prevail, the change in temperature is proportional to the change of the sum of the principal stresses [99]. Moreover, under sinusoidal loading, the relationship between temperature and the sum of principal stress is valid for the peak-to-peak value as well.

$$\Delta T = \frac{-\alpha}{\rho C_p} T_o \Delta \sigma = K_m T_o \Delta \sigma \quad (2.3)$$

where:

K_m	=	thermoelastic coefficient of the tested material
α	=	coefficient of thermal expansion of the material
T_o	=	ambient temperature
ρ	=	density of the material
C_p	=	specific heat capacity of the material
$\Delta \sigma$	=	change in sum of principal stresses in the material

2.3.3 Lock-in Processing Principle

As shown in Figure 2.10, the IR camera extracts a 0 – 10V signal from the loading machine, which corresponds to the load and frequency a material is subjected to during dynamic testing. This signal is then transmitted to the IR camera, where it is used as a lock-in reference signal to perform the frequency domain processing of the IR data.

When acquiring data, the image recording is synchronized with the modulation frequency, i.e., the reference signal from the loading machine. The system takes a number of sample images over a period of cycles as shown in Figure 2.12, for every pixel and constructs its respective thermal wave based on a mathematical algorithm similar to the mean squares method [92-94, 97, 99-101].

Figure 2.12: Signal analysis of lock-in thermography

Due to sinusoidal modulation, the sampling images are then consolidated into four basic equidistant thermograms S_1 , S_2 , S_3 and S_4 for every pixel during one cycle (Figure 2.13).

Figure 2.13: Signal acquisition and processing in creation of thermal wave cycle

Equations 2.4 – 2.5 are then used to obtain the respective local magnitude and phase of the temperature modulation of each pixel [99].

$$\mathbf{Amplitude(pixel)} = \sqrt{(s_3 - s_1)^2 + (s_4 - s_2)^2} \quad (2.4)$$

$$\varphi(pixel) = \tan^{-1} \left(\frac{s_3 - s_1}{s_4 - s_2} \right) \quad (2.5)$$

The lock-in procedure is able to combine thermographic data for each pixel and provide an overall thermogram of the material and remove the effects of inhomogeneous illumination, emissivity and reflected ambient radiation. The thermography image of the material corresponding to the DC part, that is, the average temperature is given by Equation 2.6 [99]:

$$\mathbf{T} = \left(\frac{s_1 + s_2 + s_3 + s_4}{4} \right) \quad (2.6)$$

CHAPTER 3: METHODS AND MATERIALS

3.1 General Approach

In order to validate the use of infrared lock-in thermography for use in biomechanical applications, firstly, a large left femur was mounted in a mechanical tester, oriented in 7 degrees of adduction in the coronal plane and aligned vertically in the sagittal plane in order to mimic contralateral toe-off during mid-stance phase of gait cycle and where maximum load bearing occurs, as shown in Figure 3.1 [29, 31, 44, 102]. The femur was then subjected to a dynamic axial load with average forces of 1500N, 1800N and 2100N to represent 3 times body weight (BW) for 3 different patient weight classes, i.e. 50 kg, 60 kg and 70 kg [64, 103]. The synthetic femur was cycled between a range of 2 BW and 4 BW to ensure the bone remained within the elastic range during the loading cycle. The temperature distributions on the femur due to the varying loading regimes were recorded using an infra-red lock in thermographic camera. The resultant changes in the sum of principal stresses were obtained, which were then compared to the average dynamic principal surface stress values obtained from the rosette strain gauges positioned on the posterior surface of the femur.

Figure 3.1: Axial loading of femur at 7 degrees of adduction

3.2 Specimen Selection

One large, left fourth-generation composite femur (model 3406, Pacific Research Laboratories, Vashon, Washington, USA) manufactured with a specially injected black dye (Appendix: A3) shown in Figure 3.2, was utilized [104]. The black dye is to ensure perfect infrared emission [105].

Figure 3.2: Fourth generation large left femur

The fourth-generation composite femur modeled natural cortical bone using a mixture of glass fibers and epoxy resin pressure injected around a foam core. The midshaft area had an intramedullary canal of 16mm diameter. The cancellous core material comprised of cellular rigid polyurethane foam. The material mechanical properties were obtained directly from the manufacturer and are summarized in Table 3.1 [104].

			Tensile		Compressive	
Layer	Material	Density (g/cm ³)	Strength (MPa)	Modulus (GPa)	Strength (MPa)	Modulus (GPa)
Cortical	Short glass fibre filled epoxy	1.64	106	16	157	16.7
Cancellous	Cellular rigid polyurethane	0.2	–	–	3.9	0.0475

Table 3.1: Material properties of femur

The detailed geometry of the femur is shown in Figure 3.3. These synthetic femurs have been used often in prior biomechanical studies in intact and instrumented conditions to assess a wide variety of orthopaedic injury conditions [49, 52, 57, 60, 61, 64, 106-110].

Figure 3.3: Detailed geometry of large femur

3.3 Specimen Preparation

The specimen was anchored distally to facilitate its incorporation into the mechanical testing system. As shown in Figure 3.4, the condyles at the distal end were augmented using an industrial band saw, so as to allow a perfect fit within a cube-like steel chamber of dimensions 88mm X 88mm X 75mm [51, 58]. Femoral shafts were mounted onto a chemistry stand using an adjustable multi-axial clamp and oriented vertically using leveling gauges in the coronal and sagittal planes. The distal condyles were then potted by insertion into the cube-like steel chambers filled with commercially available anchoring cement (Flow-Stone, King Packaged Materials Company, Burlington, ON, Canada) [111]. The chamber was 75mm deep, thus making the final working length of the femur to be 380mm. Distally, the potted femur was fixed firmly in an industrial vice, and then oriented in 7 degrees of adduction in the coronal plane and aligned vertically in the sagittal plane in order to mimic anatomical one-legged stance during the gait cycle [45, 102].

Figure 3.4: Procedure for potting a femur

3.4 Static Axial Loading

Experiments were done on an Instron 8874 machine (Instron, Canton, MA, USA) shown in Figure 3.5. Load cell characteristics included a capacity of ± 25 kN, a resolution of 0.1 N, and an accuracy of ± 0.5 percent [112]. The Instron's loading frame had an axial stiffness of 260 kN/mm, which is about 200 and 340 times stiffer than intact synthetic and human cadaveric femurs, respectively [49]. Thus, no compensation was necessary for Instron tester compliance. Similar test regimes to those described above have been employed in prior investigations on synthetic and human femurs [39, 49, 50, 53, 64, 113]. The proximal end of the femur was inserted into a 60mm diameter cup, which was cut out of a stainless steel cylindrical block, and meant to simulate the human acetabulum. The femoral head was able to rotate inside the cup to simulate the ball-socket system of the hip as shown in Figure 3.6 [51].

Compressive vertical loading rate and maximum load were inputted through the FastTrack™ 8800 servohydraulic controller unit (Instron, Illinois Tool Works, Norwood, MA, USA) in conjunction with a desktop computer running the interface software FastTrack™ 2 [114]. The software is also able to display a feedback on the actuator displacement at regular intervals.

Figure 3.5: Instron 8874 mechanical loading system

Figure 3.6: Mechanical representation of the ball-socket joint system of the hip

Initially, a pre-load of 100N was applied to provide complete contact with the femoral head and to minimize any slippage. A vertical load was applied at the apex of the head using displacement control (waveform = linear ramp-up/-down, maximum deflection = 2mm, preload = 100N). The slope of the force–displacement curve was used to calculate axial stiffness. Stiffness for the bone was obtained from an average of three re-tests.

3.5 Dynamic Axial Loading Test

Similar to the axial stiffness test, the femur was pre-loaded to 100N. Using load control, the femur was loaded up to the minimum dynamic load for each specific weight class, that is, twice the body weight of a 50kg, 60kg and 70kg human (1000N, 1200N and 1400N, respectively).

Figure 3.7: Sinusoidal loading of the femur

Using a displacement feedback loop system, a sinusoidal cyclic load of Amplitude = 1BW and frequency = 5Hz for 1000 cycles was applied as shown in Figure 3.7.

3.6 Stress and Strain Measurements

Two experimental techniques were implemented. Firstly, the lock-In thermographic measurement technique was applied to the dynamic loading of the femur. Secondly, strain gauge measurements were used for analyzing local strains during dynamic axial loading test regimes.

3.6.1 Lock-In Thermography

The thermographic investigation was carried out using the SC5000 Series Silver 420 Camera, shown in Figure 3.8 (Flir Systems, Oregon, USA) at an ambient temperature of 26.3⁰C. Specification of the thermography system is shown in Table 3.2 [115].

Parameters	Values
Camera	Silver 420
Detector Materials	InSb
Spectral Response	3.6 μ m - 5.1 μ m
Sub Windowing	160 x 120 pixels / 80 x 60 pixels
Frame Rate	5Hz to 170Hz full frame
Image Capture	Snapshot Integrate then Read mode (ITR)
Number of Pixels Pitch	320 x 256 pixels
Pitch	30 μ m - 30 μ m
Cooling Type	Integral stirling cooler
Cooling Time	<7mn @ 25°C ambient
Frame Rate Resolution	TTL (<300ns Jitter)
Integration Time	10 μ sn to 20000 μ s programmable, 1 μ s Step
NETD	<30mK (25mK Typical)
Power Supply	12 VDC / 5A
Power Consumption	50W in cool down mode, 30W in steady state mode
Digital Video	USB / Cam LINK
Analog Video	PAL (50Hz) or NTSC (60Hz)
Remote Control	USB / Cam LINK
Overall dim, (mm)	310 x 141 x 159
Weight (W/O lens)	3.8 kg
Water and Humidity	IP54
Operational and temperature range	- 20°C to +55°C
Temperature resolution	1mk

Table 3.2: Specification of the FLIR thermography system

Figure 3.8: FLIR SC5000 Silver 420 Camera

As mentioned in Chapter 2, thermodynamic analysis of a reversible adiabatic behavior of stressed, elastic element produces Equation 2.3. To maintain the adiabatic conditions, frequency of loading should exceed 3Hz [93, 116]. A 5Hz frequency was used in our investigation. The thermoelastic coefficient $K_m = 1.16 \times 10^{-5}/\text{MPa}$ for the synthetic cortical bone was used based on the cortical density $\rho = 1640 \text{ kg/m}^3$ [117, 118]. The room temperature = 299.3°K. Altair LI software (Cedip Infrared Systems) was used to process the temperature variations. Thermographic data was acquired after 500 cycles so as to ensure that the body had reached ‘quasi’ steady state conditions, that is, dynamic loading temperatures had stabilized [94, 119].

Figure 3.9, illustrates the lock-in thermography setup that was used for the experiment. A 0 – 5V signal, corresponding to the load and frequency, was extracted from the loading machine, filtered and amplified through the signal generator, and input to the IR camera. Altair Li was set up to capture 5000 images under E-mode setting directly from the camera at a frame rate of 50Hz. The signal from the loading machine was used as the lock-in reference signal to perform the frequency domain processing of the data. The Altair Li uses an algorithm similar to that of the least mean square method to extract the AC signal, that is, the average surface stress, from correlating the dynamic thermal signal and the dynamic reference signal. The DC value of the temperature which corresponds to non-harmonic heating is discarded.

Figure 3.9: Lock-in thermography setup

Images of the posterior, anterior and lateral side respectively of the proximal femur were captured using the lock-in setup as shown in Figure 3.10.

Figure 3.10: Various sides of the proximal femur (a) posterior, (b) anterior, (c) lateral

3.6.2 Strain Gauges

Strain Gauge Selection

Appropriate strain gauge selection depends on the types of forces the surface of the test material will be subjected to as well as conditions like amount of area available for gauge mounting. Some of the more important factors are temperature sensitivity, high strain sensitivity, and electrical resistivity of the foil. According to Szivek and Gharpuray, bone material (including simulated bone), has poor conductivity. It is therefore recommended that high resistance gauges (ideally 350 Ω) be implemented (Appendix: A4) [73].

Since interest was on multiple modes of loading in the experimental study, mainly axial and bending, rosette gauges were considered sufficient. As such, Vishay® 350-Ohms general-purpose rectangular rosette gauges (062UR, model CEA-06-062UR-350, Vishay Micro-Measurements & SR-4, Raleigh, NC, USA) are employed in this study, shown in Figure 3.11 [120]. Grid numbers 1 to 3 represent a uni-axial strain gauge aligned in a particular direction to complete the rosette strain gauge.

Model	CEA-06-062UR-350
Description	Universal general purpose strain gauges
Resistance	350.0 $\Omega \pm 0.4\%$
Overall length and width	(5.64×10.67 mm)
Strain range	$\pm 3\%$
Temperature range	-75°C to 175°C
Gauge factor, GF (at 24°C)	GRID 1: 2.070 $\pm 0.5\%$ GRID 2: 2.090 $\pm 0.5\%$ GRID 3: 2.070 $\pm 0.5\%$
GF sensitivity	(1.2 ± 0.2) % /100°C
Transverse sensitivity	GRID 1: (1 ± 0.2)% GRID 2: (0.3 ± 0.2)% GRID 3: (1 ± 0.2)%

Figure 3.11: Rectangular rosette strain gauge **Table 3.3: Strain gauge specifications**

Strain gauge setup

The bone test surface preparation requires a strict protocol to ensure proper gauge bonding [121]. Firstly, surface preparation is required to ensure a chemically clean surface with an ideal roughness for strain gauge application, as well as have a neutral pH of approximately 7. Following the manufacturer's protocol, the rosette strain gauges were placed at positions 1 and 2 as shown in Figure 3.12 [121]. Position 1 (15" from top of the steel potting chamber) was chosen to investigate the stress along the neck, whereas position 2 (12.5" from the top of the steel potting chamber) was chosen to investigate the stress along the shaft. Both strain gauges were positioned on the posterior region of the femur to investigate both axial and bending forces on the femur under axial loading conditions.

Figure 3.12: Strain gauges positioned on posterior region of femur

Strain gauge data acquisition

Insulated, three-conductor, stranded tinned-copper lead wiring was used to connect each Grid of the rosette gauge to the CRONOS-PL data acquisition unit (imc Meßsysteme GmbH, Berlin, Germany). The three conductors of the wiring were colour coded (black, white and red). The conductors were separated, and each was stripped of at least half an inch of insulation. As shown in Figure 3.13, the black and white lead wires were entwined and soldered onto one strain gauge terminal, and the red lead wire was soldered to the other terminal [122].

Figure 3.13: Soldering of wire to gauge

The three-wire attachment to the strain gauge is implemented to significantly minimize effects of lead-wire temperature fluctuations, as well as increased strain gauge measurement sensitivity as compared to a two-wire attachment [123]. The wiring of the two strain gauges was then connected to one DSub-15-pin connector (ACC/DSUB-UNI2, imc Meßsysteme GmbH, Berlin, Germany) shown in Figure 3.14.

Figure 3.14: DSub-15 connector

A DSub-15 connector can only provide two channels, which was then connected to the CRONOS-PL data acquisition system (imc Meßsysteme GmbH, Berlin, Germany) through a UNI2-8 eight channel amplifier as shown in Figure 3.15. This completed the Wheatstone quarter bridge circuit for strain measurements. The wiring procedure to the DSub-15 connector was carried out for the 3 grids of each rosette strain gauge. A total of three connectors were required.

Figure 3.15: DSub connectors linked to data acquisition system

The CRONOS-PL data acquisition system was connected via a LAN network to a laptop running the data acquisition software imcDevices v2.6 (imc Meßsysteme GmbH, Berlin, Germany) which stored the strain data. The strain gauge channels were configured according to the manufacturer's specifications for each strain grid (Table 3.3) in the imcDevices v2.6. Using the conditions described for dynamic axial loading test, strain data were captured starting at 500 cycles for duration of 90 seconds. The signal analysis software FAMOS v5.0 (imc Meßsysteme GmbH, Berlin, Germany) was used to find the average strain of the grids (ϵ_1 , ϵ_2 and ϵ_3) of the rosette strain gauge at positions 1 and 2.

Stress Calculations

As shown in Figure 3.11, the grids for a rectangular rosette were numbered from 1 to 3 in a counter clockwise direction. The equations for calculating principal strains from three rosette strain measurements are derived from what is known as a “strain-transformation” relationship.

As shown in Equation 3.1, the maximum and minimum principal strains, ϵ_P and ϵ_Q respectively were calculated from the grid strains ϵ_1 , ϵ_2 and ϵ_3 respectively.

$$\epsilon_{P,Q} = \frac{\epsilon_1 + \epsilon_2}{2} \pm \frac{1}{\sqrt{2}} \sqrt{(\epsilon_1 - \epsilon_2)^2 + (\epsilon_2 - \epsilon_3)^2} \quad (3.1)$$

Using the composite femur’s elastic modulus ‘E’ = 16.7 GPa and Poisson’s ratio ‘v’ = 0.3, the principal stresses were then calculated from the derived principal strains using the biaxial form of Hooke’s Law, as shown in Equation 3.2, or directly from the grid strains using Equation 3.3 [44, 49, 124].

$$\begin{aligned} \sigma_P &= \frac{E}{1 - \nu^2} (\epsilon_P + \nu \epsilon_Q) \\ \sigma_Q &= \frac{E}{1 - \nu^2} (\epsilon_Q + \nu \epsilon_P) \end{aligned} \quad (3.2)$$

$$\sigma_{P,Q} = \frac{E}{2} \left[\frac{\epsilon_1 + \epsilon_3}{1 - \nu} \pm \frac{\sqrt{2}}{1 + \nu} \sqrt{(\epsilon_1 - \epsilon_2)^2 + (\epsilon_2 - \epsilon_3)^2} \right] \quad (3.3)$$

CHAPTER 4: RESULTS

4.1 Axial Stiffness Tests

Results of the axial stiffness are provided in Table 4.1 below. Stiffness of the femur under static conditions was 1347.9 N/mm (Appendix: A1). Stiffness under average dynamic loads of 1500N, 1800N and 2100N were found to be 2008.9N/mm, 2087.6 N/mm and 2051.5 N/mm respectively (Appendix: A2). All specimens were kept within the linear elastic range to try to avoid permanent damage, as indicated by the linearity coefficients, R^2 , obtained from the force-versus-displacement graphs, namely, static test ($R^2 = 0.99$), dynamic test at 50 kg ($R^2 = 0.99$), dynamic test at 60 kg ($R^2 = 0.99$), and dynamic test at 70 kg ($R^2 = 0.99$) (see Appendix A1 and A2). Using Equation 4.1, the difference between the static axial stiffness and the dynamic axial stiffness of 50kg, 60kg and 70kg loads was 49%, 55% and 52%, respectively and the average difference between the three values was 52%.

$$\% \text{ Difference} = \left(\frac{\text{Static Stiffness} - \text{Dynamic Stiffness}}{\text{Static Stiffness}} \right) \times 100 \quad (4.1)$$

Specimen Configuration Versus Stiffness (N/mm)	
Loading Conditions	Stiffness (N/mm)
Static	1347.9
Dynamic Average Load of 1500N	2008.9
Dynamic Average Load of 1800N	2087.6
Dynamic Average Load of 2100N	2051.6

Table 4.1: Stiffness versus Loading Conditions

4.2 IR Stress Maps

Stress maps for the average dynamic loads of 1500N, 1800N and 2100N are shown in Figure 4.3, Figure 4.4 and Figure 4.5, respectively. Average stress maps for the anterior, posterior and the lateral sides of the femur were captured and it was observed that images for each load were similar irrespective of the side. The positive and negative values of the scale represent the tensile and compressive stress respectively. As the tensile stresses increase, the colour goes from

green to dark pink, while the rising compressive stresses are shown by the colours going from green to dark blue.

Figure 4.1: IR Stress image of the anterior, posterior and lateral sides of a femur under an average load of 1500N

Figure 4.2: IR Stress image of the anterior, posterior and lateral sides of a femur under an average load of 1800N

Figure 4.3: IR Stress image of the anterior, posterior and lateral sides of a femur under an average load of 2100N

Using thermographic stress and temperature maps, the average surface stresses and temperatures for positions 1 and 2 were extracted to verify how Altair Li Software incorporated the basic thermoelastic effect equation. Stresses at positions 1 and 2 in the posterior region were extracted by taking the average area of the IR stress map represented by the rosette strain gauges. Table 4.2, shows the stresses at each respective load for positions 1 and 2. It is seen that position 1 undergoes compression while position 2 undergoes tension.

Stress (MPa) from IR Camera				
Position	1500N	1800N	2100N	Load Experienced
1	-35.4	-38.3	-42.9	Compression
2	4.9	4.8	5.3	Tension

Table 4.2: Stress values from IR Images

Table 4.3, shows the temperatures at each respective load. Position 1 shows a positive temperature change signifying the region is undergoing compression, whereas position 2 shows a negative temperature change suggesting that region experiences tension.

Temperature (^o K) from IR Camera				
Position	1500N	1800N	2100N	Load Experienced
1	0.1226	0.1334	0.1488	Compression
2	-0.0172	-0.0167	-0.0184	Tension

Table 4.3: Temperature values from IR Images

Based on temperatures and stresses from the infrared camera, it can be shown, according to Figure 4.4, that the relationship between the two is highly linear and well correlated ($R^2 = 0.99$) and that the slope of the line ($m = -0.0035^{\circ}\text{K}/\text{MPa}$) = $-T \cdot K_m$ where $K_m = 1.16 \cdot 10^{-5}/\text{MPa}$ and $T = 299.3^{\circ}\text{K}$. The resolution of the camera is 0.001°K ; therefore, the distinguishable stress resolution of the camera for composite bones is 0.286MPa .

Figure 4.4: Thermoelastic relationship between temperature and stress

4.3 Maximum Stress Regions of the Femur

Figure 4.5 and 4.6 show the regions of high stress along the femur. Most of the stresses are concentrated along the posterior and anterior region of the femur and localized around the neck. On the anterior side, the highest stresses are seen at points A and B, corresponding to the medial and lateral sides of the femur. The range of stresses due to increasing axial force at point A is between 51.9MPa to 52.6MPa (compressive stress), while at point B is between 42.3MPa to 46.5MPa (tensile stress).

AVERAGE AXIAL FORCE (N)	ABSOLUTE STRESS (MPa)	
	A	B
1500	51.9	46.5
1800	53.4	42.4
2100	52.6	42.3

Figure 4.5: High stress regions along the anterior side of the femur

The posterior region of the neck experiences mostly compressive stress and experiences the highest stress as compared to the anterior and lateral sides. The range of stresses due to increasing axial force at point A is between 83.3MPa to 94.5MPa (compressive stress). This is a stress ratio of 1.8 ($= 91.2\text{MPa} / 51.9\text{MPa}$) to 2 ($= 103.5\text{MPa} / 52.6\text{MPa}$) as compared to the maximum compressive stresses on the anterior side.

AVERAGE AXIAL FORCE (N)	ABSOLUTE STRESS (MPa)
	A
1500	91.2
1800	96.0
2100	103.5

Figure 4.6: High stress regions along the posterior side of the femur

4.4 Strain Gauge Results

Table 4.4, shows the strain values for each respective grid of the rosette strain gauge for positions 1 and 2. All grids for position 1 are in compression and range from $-39\mu\text{strain}$ to $-1257\mu\text{strain}$ and grid 1 registers the highest strain ranging between $-899\mu\text{strain}$ and $-1257\mu\text{strain}$. All grid strains of position 2 are in tension and range from $48\mu\text{strain}$ to $134\mu\text{strain}$, while grid 2 registers the highest strains ranging from $107\mu\text{strain}$ to $134\mu\text{strain}$. However, grid 3 at position 2 registers almost similar strain as compared to grid 2.

Experimental strains (μ strain) from rosette strain gauge					
Position	Grid	1500N	1800N	2100N	Load Experienced
1	ϵ_1	-899	-1142	-1257	Compression
	ϵ_2	-81	-80	-39	
	ϵ_3	-645	-814	-873	
2	ϵ_1	48	55	73	Tension
	ϵ_2	107	126	134	
	ϵ_3	103	128	121	

Table 4.4: Experimental strain values from rosette strain gauges

Table 4.5, represents the principal stresses at positions 1 and 2. At position 1 the maximum principal stress ranges from -9MPa to -11.5MPa while the minimum principal stress ranges from -26.3MPa to -37.2MPa. At position 2 the maximum principal stress ranges from 2.3MPa to 2.9MPa while the minimum principal stress ranges from 1.3MPa to 1.7MPa.

Principal Stresses (MPa) from strain gauge					
Position	Grid	1500N	1800N	2100N	Load Experienced
1	σ_P	-9.0	-11.1	-11.5	Compression
	σ_Q	-26.3	-33.5	-37.2	
	$\sigma_P + \sigma_Q$	-35.3	-44.6	-48.7	
2	σ_P	2.3	2.8	2.9	Tension
	σ_Q	1.3	1.5	1.7	
	$\sigma_P + \sigma_Q$	3.6	4.3	4.6	

Table 4.5: Experimental principal stresses

4.5 Comparison of IR Stress Results with Strain Gauge Results

A comparison of experimental strain gauge stress values with IR stresses is provided (Table 4.6). Experimental stresses ranged from -35.3MPa to -48.7 MPa and 3.6 MPa to 4.6MPa for positions 1 and 2 respectively, while IR stresses ranged from -35.4 MPa to -42.9 MPa and 4.8MPa to 5.3 MPa for positions 1 and 2, respectively. All strain gauge and IR stresses were plotted together onto one graph to illustrate the degree of correlation between the two measurement methods (Figure 4.7). Excellent correlation was noted between experiment and IR analysis with a slope =

1.08 (where the ideal = 1) and a Pearson correlation of $R^2 = 0.99$. Strain gauge stresses exceeded IR stresses by 8% based on the slope value (Figure 4.7).

Comparison of Experimental and IR Stress (MPa) % Difference = [(IR stress – strain stress) / IR stress] * 100						
Load	Position 1			Position 2		
	IR Camera (MPa)	Strain Gauge (MPa)	% Difference	IR Camera (MPa)	Strain Gauge (MPa)	% Difference
1500N	-35.4	-35.3	0.3	4.9	3.6	26.5
1800N	-38.3	-44.6	16.4	4.8	4.3	10.4
2100N	-42.9	-48.7	13.5	5.3	4.6	13.2

Table 4.6: Experimental versus IR stress

Figure 4.7: Comparison of experimental and IR stress

4.6 Thermoelastic Coefficient Calculation using Experimental Strain Gauge Values and IR Temperatures

The actual thermoelastic coefficient can be obtained using the experimental stresses obtained from positions 1 and 2 and compared to the temperature values obtained from the IR camera (Table 4.7).

Load	Position 1		Position 2	
	Temperature (°K)	Strain Gauge (MPa)	Temperature (°K)	Strain Gauge (MPa)
1500N	0.1226	-35.3	-0.0172	3.6
1800N	0.1334	-44.6	-0.0167	4.4
2100N	0.1488	-48.7	-0.0184	4.6

Table 4.7: Experimental stress (MPa) and respective IR temperature

Using the slope = $-0.0032^{\circ}\text{K}/\text{MPa}$ (Figure 4.8), and absolute temperature = 299.3°K , it is calculated that the thermoelastic coefficient (K_m) of the femur = $1.069 \times 10^{-5}/\text{MPa}$. If the stress maps are adjusted for the calculated thermoelastic coefficient, the analysis between IR stresses and experimental stresses show nearly perfect correlation ($R^2 = 0.99$) and slope = 0.99. This is a difference of approximately 1% between experimental and IR stresses based on the slope value (Figure 4.9).

Figure 4.8: Experimental Thermoelastic Coefficient

Figure 4.9: Stress comparison with IR stresses adjusted

4.7 Correlation of Stress and Axial Force

Six positions on the IR stress maps for the anterior, posterior and lateral side were probed and compared to their respective axial loads (Figure 4.10) using Figures 4.1 – 4.3. The two positions of the strain gauge on the posterior side of the femur were also compared to their respective axial loads. Positions on the posterior and anterior sides were chosen to coincide with the angle of inclination in the frontal plane and were equidistant from each other. Positions on the lateral side were chosen from the top most point on the greater trochanter region, midline to the antero-posterior sides. Each position had a space of 1cm between them.

Figure 4.10: Positions probed for stress on the posterior, anterior and lateral side of the femur. Each position was spaced 1cm apart. Position 6 on the shaft of the femur was approximately 10" from the top of the potting.

4.7.1 Correlation of IR Stresses and Axial Force

Posterior Side

Results of the posterior thermographic stresses versus axial force are shown in Table 4.8.

POSITION	IR STRESS (MPa)			PEARSON CORRELATION R ²
	AXIAL FORCE			
	1500N	1800N	2100N	
1	-50.6	-53.6	-61.0	0.94
2	-11.3	-13.6	-16.1	0.99
3	-7.9	-7.4	-6.6	0.96
4	3.9	4.2	4.4	0.99
5	6.0	6.1	7.3	0.81
6	5.3	5.4	5.6	0.96

Table 4.8: Correlation of posterior surface stresses from IR thermography versus average axial force

In the neck region of the femur (i.e. positions 1 to 6), stresses increased steadily with increasing axial force with a high degree of linearity ($R^2 = 0.81$ to 0.99), whether the stresses were compressive (positions 1 to 3) or in tension (positions 4 to 6). Moreover, for positions 1 to 6, the ratio of axial forces was comparable to the ratio of corresponding stresses. For instance, for a high-force-to-low-force ratio of $2100\text{ N} / 1500\text{ N} = 1.4$, position 1 (compressive stress) yielded a corresponding stress ratio of $61.0\text{MPa} / 50.6\text{MPa} = 1.21$, while position 5 (tensile stress) yielded a stress ratio of $7.3\text{MPa} / 6.0\text{MPa} = 1.22$.

Anterior Side

Results of the anterior thermographic stresses versus axial force are shown in Table 4.9. Positions 1 to 6 along the anterior side of the femur, stresses increased steadily with increasing axial force with a high degree of linearity ($R^2 = 0.87$ to 0.99), whether stress was compressive (position 3) or tensile (positions 1, 2, 4, 5, and 6). Moreover, for positions 1 to 6, the ratio of axial forces was comparable to the ratio of corresponding stresses.

POSITION	IR STRESS (MPa)			PEARSON CORRELATION R ²
	AXIAL FORCE			
	1500N	1800N	2100N	
1	16.8	18.0	18.9	0.99
2	0.1	0.2	1.0	0.99
3	-7.2	-9.4	-10.3	0.94
4	11.3	11.8	11.9	0.87
5	1.8	3.5	4.8	0.99
6	0.3	1.0	1.9	0.99

Table 4.9: Correlation of anterior surface stresses from IR thermography versus average axial force

For instance, for a high-force-to-low-force ratio of 2100 N / 1500 N = 1.4, position 1 (tensile stress) yielded a similar stress ratio of 18.9MPa / 16.8MPa = 1.13, while position 3 (compressive stress) yielded a stress ratio of 10.3MPa / 7.2 MPa = 1.43.

Lateral Side

Results of the lateral thermographic stresses versus axial force are shown in Table 4.10.

POSITION	IR STRESS (MPa)			PEARSON CORRELATION R ²
	AXIAL FORCE			
	1500N	1800N	2100N	
1	-1.6	-2.0	-2.1	0.92
2	1.1	1.2	1.4	0.90
3	7.0	7.1	7.6	0.89
4	8.5	9.3	9.6	0.95
5	6.6	6.7	7.1	0.89
6	7.2	7.6	7.8	0.99

Table 4.10: Correlation of lateral surfaces stresses from IR thermography versus average axial force

Positions 1 to 6 along the lateral side of the femur, stresses increased steadily with increasing axial force with a high degree of linearity ($R^2 = 0.89$ to 1.00), whether stress was compressive (position 1) or tensile (positions 2 to 6). Moreover, for positions 1 to 6, the ratio of axial forces was comparable to the ratio of corresponding stresses. For instance, for a high-force-to-low-force ratio of 2100 N / 1500 N = 1.4, position 1 (compressive stress) yielded a similar stress ratio

of 2.1MPa / 1.6MPa = 1.31, while position 2 (tensile stress) yielded a stress ratio of 1.4MPa / 1.1MPa = 1.27.

4.7.2 Correlation of Strain Gauge Stresses and Axial Force

Results of the strain gauge stresses versus axial force are shown in Table 4.11.

POSITION	STRAIN GAUGE STRESS (MPa)			PEARSON CORRELATION R ²
	AXIAL FORCE			
	1500N	1800N	2100N	
1	-35.3	-44.6	-48.7	0.95
2	3.6	4.4	4.6	0.92

Table 4.11: Correlation of stresses from strain gauges versus average axial force

For positions 1 and 2, stresses increased steadily with increasing axial force with a high degree of linearity ($R^2 = 0.92$ to 0.95). Moreover, for both locations, the ratio of axial forces was comparable to the ratio of corresponding stresses. For instance, for a high-force-to-low-force ratio of 2100 N / 1500 N = 1.4, position 1 (compressive stress) yielded a similar stress ratio of 48.7MPa / 35.3MPa = 1.38, while position 2 (tensile stress) yielded a stress ratio of 4.6MPa / 3.6 MPa = 1.28.

4.8 Femur fracture under Dynamic Loading

The synthetic femur underwent catastrophic failure during dynamic loading under an average force of 2100N. Figure 4.11, shows a subcapital neck fracture caused due to fatigue under dynamic loading conditions. It is seen that the fracture occurs at the superior surface of the femoral neck and travels down to the inferior surface of the femoral neck at the junction between the femoral head and the femoral neck.

Figure 4.11: Femoral neck fracture (anterior view)

CHAPTER 5: DISCUSSION

5.1 General Findings from Current Study

Lock-in infrared thermography was implemented for non-destructive assessment of a commercially available synthetic femur. The method allowed the generation of a comprehensive three-dimensional surface stress mapping of a synthetic femur under cyclic axial loading conditions. The results were validated using strain gauge measurements. The femur experienced high peak stresses along the neck region, which is a likely clinical site for potential fracture. Surface stresses increased in direct proportion to the axial cyclic loads, especially in the proximal region. This is the first study that tries to comprehensively map and evaluate the surface stress of an intact femur under various loads using lock-in thermography and then validates the method experimentally using strain gauge measurements.

5.2 Comparison of Current Results with Previous Studies

5.2.1 Axial Stiffness

The static axial stiffness of the potted femur was compared to the three different dynamic loads (average load of 1500N, 1800N and 2100N). The average static axial stiffness (1348 N/mm) compared reasonably to previous average data for synthetic femurs yielding stiffnesses of 1543 N/mm, 1540 N/mm, 1416 N/mm, 1333 N/mm, 1290 N/mm and 1230 N/mm, as well as human cadaveric femurs with stiffnesses of 1360 N/mm, while other human values 1070 N/mm, and 757 N/mm were much lower [39, 41, 45, 49, 113, 125]. Any differences in results are likely attributed to variations in experimental conditions, such as proximal load application method, distal fixation method, femur orientation, material properties, and so forth.

However, the dynamic axial stiffnesses was an average of 52% greater compared to the static axial stiffness. No prior studies seem to report the changes in static to dynamic stiffness using current loading conditions with synthetic femurs. However, Larsson et al. carried out static and dynamic tests on three independent planes of a cadaveric femur, showing the highest stiffness along the axis perpendicular to the femoral neck in the frontal plane. The dynamic stiffness = 2090 N/mm was about 14% higher than static stiffness = 1840 N/mm [126].

The difference in axial static and dynamic stiffnesses suggests the viscoelastic nature of synthetic femurs. The increasing loading rate from static to dynamic axial loading resulted in increased axial stiffness of 52%, thereby showing the strain rate dependency of the synthetic femur. This mimics the viscoelastic nature of actual human bone [24].

5.2.2 Strain Gauge

Results obtained at positions 1 and 2 showed stresses to be much higher at position 1 as compared to position 2 under axial loading conditions. The results tend to concur with various studies that suggest that high stresses are concentrated primarily around the proximal region on the femur at the neck [85, 127, 128]. Position 2 shows low stresses due to its positioning along shaft where stresses have shown to be much lower under axial loading [129].

Position 1 undergoes compressive strain whereas position 2 is under tensile strain. This could be attributed to the strain gauge positioning within the trabecular system as shown in Figure 2.4 and the dynamic force acting at an adduction angle of 7 degrees [21, 130, 131]. Studies carried out by Cristofolini et al., using cadaver femurs with rosette strain gauges in static loading at approximately 8 degrees suggested the maximum and minimum principal stresses on the posterior and anterior sides were equally distributed between tension and compression. However, the present work seems to suggest, this may not be the case in dynamic loading. The observations could also be attributed to the fact that the synthetic femur in this study was free to deform within the cup therefore generating high bending forces, whereas for the study by Cristofolini et al., the cadavers' motion of the femoral head was restricted horizontally with respect to the coronal and sagittal plane axes, as well as the posterior and anterior sides along diaphysis being equally influenced by the medial and lateral sides [72].

5.2.3 IR Stress Maps

Stress patterns for the various dynamic loads (Figure 4.1 – 4.3) seem to agree significantly with prior finite element models and strain gauge measurements along the medial and lateral areas of the proximal femur especially with respect to the femoral neck, where high stresses are seen [45, 73, 83, 132-134]. Studies reported stresses on the femur ranging from 34.2MPa – 53.3MPa [45, 134, 135].

However, in the current study, the highest stresses were seen in the posterior region of the femoral neck. This was almost 1.8 to 2 times higher than that of the reported peak stresses on the medial and lateral femoral neck region. This could be attributed to the fact that axial loading is not the only type of force present during dynamic load. There could be twisting and bending forces on the femoral head due to the ability of the head to rotate freely under the jig, which could produce high amounts of deformation on the posterior region of the femoral neck. As no prior studies suggest this phenomenon, further investigation would be required. Also, decreasing stresses with increasing loads in Figure 4.5, suggest signs of fatigue damage.

The stresses maps (Figure 4.1 – 4.3) and Table 4.10, suggest that, laterally, the femur at the epiphysis along the femoral neck and the diaphysis are in tension. However, the greater trochanter region at the epiphysis does not heat up at all during dynamic loading, but instead shows signs of compressive stress which could be attributed to the complex geometry of the femur. Also, there seems to be no direct relationship between stress and the position on the proximal diaphysis. The values seem to be relatively the same which confirms that most of the strain is absorbed by the femoral neck. However, it should be noted that there is a stress concentration around positions 3 and 4. This is due to the presence of the injection molding hole in the synthetic femur that results in a stress riser [136]. With respect to the posterior and the anterior sides (Table 4.8 – 4.10 and Figure 4.1 – 4.3), positions along the femoral neck vary between compression and tension due the complex geometry of the femur at this region. However; the diaphysis, which could be modelled as a cylinder, seems to be mostly in tension suggesting high bending forces act on these regions in dynamic loading.

Several prior studies have investigated the application of IR thermography in analysing stress patterns on bone. However, due to limited advances in thermographic technology, the quality of images produced was poor and inconsistent, while relevant stress data were not reported to make any useful observations with the current study [117, 137-140].

5.2.4 Thermoelastic Constant of Synthetic Femur

The present study showed the calibrated thermoelastic coefficient (K_m) to be $1.069 \times 10^{-5}/\text{MPa}$. Hyodo et al., in his investigation into thermoelastic stress analysis suggested a $K_m = 1.47 \times 10^{-5}/\text{MPa}$ based on experiments carried out on cortical plates from second

generation synthetic bone material (Appendix A: 5) [104, 117, 141]. The differences could be attributed to firstly, the differences in cortical and cancellous density between second generation composites used by Hyodo et al., as compared to the current study which implemented fourth generation composite femurs. Secondly, there was a variation in fibre glass filled epoxy between the consecutive design generations that could alter the thermoelastic properties significantly [117, 141]. The K_m of natural cortical bone = 1.1×10^{-6} /MPa, suggesting that testing with synthetic bones would provide an equal or greater stress resolution than natural bone [118, 137, 142].

5.2.5 Femoral Fracture

The subcapital fracture (Figure 4.11) shows similarities to fractures in studies carried out by Gardner et al., and Nicayenzi et al., where the load to failure under axial loading was analysed [125, 143, 144]. The femoral fracture for the dynamic conditions in the current study (cancellous density = 0.20g/cm^3) were identical to the study carried out by Nicayenzi et al. in those specimens whose cancellous densities were less than or equal to 0.24g/cm^3 [143]. The stress maps generated in the present study indicated the peak stresses to be concentrated in the superior region of the femoral neck especially in the posterior region. This suggests that under repeated cyclic loading, internal stresses generated seem to cause micro-damage along the femoral neck, which leads to the subcapital fracture [132]. Studies by Hodge et al., and Brown et al., seem to strengthen the assumption of high posterior stresses due to the maximum pressure being observed on the superior-posterior aspect of the lateral roof of the femoral head [145-147]. Another factor that could suggest high posterior neck surface stresses is the result of the injection molding process on synthetic femurs which may have created micro-defects in the cancellous layer at the neck.

5.3 Practical Implications

5.3.1 Biomechanical Testing Implications

Average stress values obtained using IR thermography was highly comparable with the average stress results from the surface strain gauges (Figure 4.7). The Pearson coefficient of determination (R^2) of 0.99 indicates a high degree of precision between the two sets of variables,

that is, the IR stress and strain gauge stress. However, the slope = 1.08 provides a measure of accuracy between the two sets of data. In this study, overall IR stresses were lower than strain gauge stresses by 8% suggesting IR stresses underestimated strain gauge stresses or overestimation on the part of the strain gauges. This difference could be attributed to the texture of the surface or curvature of the bone, which could affect both infrared emissions captured by the camera, and resistance signals measured by strain gauges.

IR thermography, due to its capability to provide highly accurate measurements, could be a viable replacement for strain gauge measurements in biomechanical testing of synthetic bones undergoing dynamic loading conditions. The infrared thermographic approach to stress analysis for orthopaedic applications raises several potential benefits over current methods [89].

Firstly, IR thermography has the ability to generate a complete three-dimensional stress map under actual experimental loading conditions which is not possible with strain gauges unless every point on the surface is covered. IR stress maps may not completely remove the requirement for FEA, but could provide a synergetic option with FEA for biomechanical researchers to refine current orthopaedic devices and improve upon them. Also, stress maps would allow researchers to select particular sites of interest where strain gauges could be positioned to avoid issues of high stress gradients. IR cameras are highly sensitive, having the ability to measure temperature variations as low as 1mK.

Secondly, IR thermography is a non-destructive technique and therefore does not damage the surface of the testing sample. Once the specimen is experimentally calibrated using strain gauges, subsequent specimens no longer require mounting of strain gauges which is usually the standard procedure in biomechanical investigations.

Thirdly, IR thermography requires minimal preparation of the testing setup and is a very simple “point and shoot” method requiring focusing on the specimen and taking a picture. This means testing times can be shortened tremendously.

Lastly, IR principles require testing specimens to be under cyclic loading conditions in order to generate temperature maps that are then converted into stress values. These loading conditions match real-life clinical conditions that bones experience.

With respect to synthetic femurs as compared to cadaveric femurs, it should be noted that there is usually a pin-hole present at the greater trochanter region of the lateral part of the synthetic femur. This is a manufacturing feature is a result of injection molding. As shown in Figures 4.1 – 4.3, the presence of this pin-hole results in a stress riser around this area which could affect experimental results. Therefore, biomechanical testing involving synthetic femurs should account for this. The manufacturer upon request may be able to fill this hole.

5.3.2 Clinical Implications

The IR stress maps suggest that stress fracture of intact femurs has a high probability of occurring along the femoral neck. This is known to be a common clinical site for injury, especially in elderly women suffering from osteoporosis who undergo a fall, and in young people with good bone quality who are involved in a motor vehicle accident [148, 149]. Therefore, it could be suggested that early testing procedures such as Bone Mass Density (BMD) tests, X-rays etc. on patients, especially the elderly, should be refined to detect damages along the neck. The indications of high stresses along the femur agree with common clinical recommendations given to older patients and heavier patients to minimize any activities that heavily load the joint.

The static and dynamic axial stiffness results show there is a drastic difference of 52% in stiffness between the two loading regimes. FEA studies have also similarly shown that there is about a 10% to 20% rise in loading on the hip joint when dynamic conditions are applied compared to static conditions [65]. Clinically, this could be vital with respect to orthopaedic applications as static axial stiffness relates to standing while dynamic axial stiffness refers to activities such as walking or running. Bougherara et al., compared the static axial stiffness of an intramedullary nail-bone construct for repairing mid-shaft femur fractures [64]. Based on the stiffness values of two techniques, antegrade nailing (AA) and retrograde nailing (RA), it was suggested antegrade nailing would be suited for young patients because the nail-bone stiffness of a health young patient without the nailing, matched the stiffness of AA, whereas the retrograde nailing technique would be suited for elderly patients because the RA stiffness matched the stiffness of an elderly patient without the nailing [49, 64]. However, Bougherara et al., along with many other similar studies, fail to examine the response of these femur and implant-femur constructs under dynamic loading, which could provide vital insights into the viability of these

fracture fixation devices and alter their clinical recommendations [39, 44, 45, 49, 52, 54, 64, 84, 113, 113, 134, 143, 150]. Therefore, biomechanical investigations using cyclic loading on instrumented femurs that mimic fracture fixation or total joint replacement may be more realistic in duplicating clinical conditions [46, 50].

5.4 Limitation/Sources of Error and Future Work

Despite the limitations given below, this is the first investigation to experimentally validate lock-thermography using a well established method, like strain gauges, and first to provide an experimental three dimensional stress map of the intact femur.

The camera was focused to capture the proximal region of the femur as this is the region of interest in biomechanical studies of the hip joint. However, the camera has the ability to be positioned further away to provide a complete map of the femur. In this study, sinusoidal loads were applied within the elastic region of the synthetic femur for a limited number of cycles. This was to ensure no permanent damage occurred, so that all three load levels with all surface views were captured. However, using other cyclic waveforms, such as triangular or square waves, at identical frequencies would not significantly affect results as long as adiabatic conditions are maintained.

However, with the current setup, the steel pelvic cup prevented the capture of the stress values on the medial surface of the femur directly. The three other views captured compensated for the missing view and provided a good viewpoint of the surface stress map on the medial side.

Displacement, rather than force control was used for applying the sinusoidal load. This could result in drifting of force levels with subsequent cycles as the specimen became less mechanically stiff with each subsequent cycle. Figures A.4 – A.6 show, however, there was minimal drifting as evidence by the linearity of the data ($R^2 = 0.99$); thus, loads were consistent at the average values of 1500N, 1800N and 2100N respectively. However, it should be noted that even though clinical loads are dynamic and cyclic, they are not perfectly sinusoidal during walking gait [29, 31]. Therefore, experimentally, loading cycles should reflect the actual gait cycle with respect to loads.

The synthetic femur was loaded without influence of the hip joint muscles, which is not a realistic case clinically. Muscle action accounts for a significant portion of generated joint load and it provides stabilization of the joint and, therefore, can alter the distribution of compressive and tensile stresses in the proximal femur [151-155].

Cyclic loading was carried out at 5Hz because a frequency greater than 3Hz is usually required to achieve adiabatic conditions and obtain high quality IR images. Normal human walking gait is often replicated in biomechanical studies using frequencies between 1 to 3Hz [24, 156, 157]. This increased cycling frequency may have exaggerated the magnitude of the stresses measured on the surface of the femur. This could be because at lower frequencies, impact forces and fatigue on bones would be much lesser; therefore, generating lower temperatures and hence resultant thermodynamic stresses would be much lesser, unlike stresses at higher frequencies.

If the emissivity of the specimen surface is poor due to surface texturing, this can result in poor image quality [105]. This also results in images that consist of a stress field that is superimposed on top of a “roughened” background. Such images can be difficult to interpret properly, since high stress points may be image artifacts due to surface texturing, rather than stress concentrations of clinical relevance. Similarly, if an object has a light colour, its emissivity will also be poor, so that, spraying or completely dyeing the specimen with black ink usually increases emission to 100% [105]. In the current study, these problems were avoided, since the femur surface was very smooth and it was injected with black dye during the manufacturing process.

The loading had to be done within the elastic region in order to carry out thermoelastic stress analysis based on surface temperatures. However, based on recent advances in lock-in thermographic equipment, a new feature based on dissipation energy could allow analysis of specimens at the microstructure level. Dissipation mode would not be limited to frequencies of 3Hz or greater, as well as allowing anelastic analysis and assessing fatigue limits of specimens [95, 96, 115, 119, 158].

FEA was not used as a validation technique as FEA tends to provide stresses based on static loading [64, 88-91, 128]. IR on the other hand is based on dynamic stresses and hence not easily comparable to FEA. FEA, also, is used in tandem with other techniques and is not usually

sufficient as a stand-alone technique for providing stress values unlike techniques such as strain gauge measurements.

There are various other techniques of measuring stresses under experimental loading conditions, such as brittle coating, photoelastic coating, ultrasound techniques and so on [159, 160]. However, all these techniques are cumbersome and difficult to implement unlike strain gauges. Strain gauges, due to their ease of use, have become the most commonly relied method for stress measurements in biomechanical studies [64, 71, 74, 75, 89, 161-164]. Strain gauges are the so called “gold-standard” of biomechanical stress analysis techniques. However, IR thermography should be compared to other experimental techniques that provide a full field stress map, as strain gauges can only provide local stress and strain.

CHAPTER 6: CONCLUSION

Lock-in IR thermography was used for non-destructive evaluation of a synthetic femur. It produced full three-dimensional stress maps of the femur under cyclic conditions reflecting the walking gait cycle of healthy human subjects.

It was seen that the synthetic femur stiffness values were significantly higher during dynamic axial loading as compared to the stiffness values under static axial loading (49 – 55% difference based on 50kg – 70kg loads). Clinically, static loading represents standing, whereas dynamic loading represents activities such as walking or running. It is therefore critical that biomechanical studies incorporate dynamic testing regimes in analysis of fracture and joint repair devices due to vast differences in stiffness values between the two loading regimes.

The synthetic femur, due to extensive cyclic loading, underwent catastrophic failure that occurred at the superior surface of the femoral neck and travelled down to the inferior surface of the femoral neck at the junction between the femoral head and the femoral neck.

The IR results were validated by strain gauge measurements and showed good correlation ($R^2=0.99$). The femur itself showed high peak stresses in the neck region, which is a site for potential fracture. Surface stresses rose in direct proportion to the amount of axial cyclic load applied. It was also observed that most of the strain was absorbed by the femoral neck and head, however; due to the complex geometry of the femoral head and neck, there was no clear pattern between compressive and tensile stresses. This is the first study in the literature which has experimentally validated lock-in IR thermography technology in assessing the stress patterns in the proximal region of the femur.

IR thermography not only has the potential for permitting an understanding of how physiological conditions affect movable human joints, but may potentially be used in orthopaedic biomechanics applications to improve upon current fracture fixation devices and artificial joints.

APPENDICES

A.1 Static Axial Stiffness

For Figures A.1 – A.3, the numerical values on the axis are based on the Instron testing machine's coordinate system. Thus, one loading cycle started at the top right of the graph (minimum), continued until peak force was reached at bottom left of the graph (maximum), and then returned to a zero force condition at the top right of the graph (minimum). The equation gives the slope and the intercept of the line of best fit, while R^2 indicated the degree of linearity and agreement between experiment and the line of best fit.

Figure A.1: Static axial stiffness test 1

Figure A.2: Static axial stiffness test 2

Figure A.3: Static axial stiffness test 3

A.2 Dynamic Axial Stiffness

For Figures A.4 – A.6, the numerical values on the axis are based on the Instron testing machine's coordinate system. Thus, one loading cycle started at the top right of the graph (minimum), continued until peak force was reached at bottom left of the graph (maximum), and then returned to a zero force condition at the top right of the graph (minimum). The equation gives the slope and the intercept of the line of best fit, while R^2 indicated the degree of linearity and agreement between experiment and the line of best fit.

Figure A.4: Dynamic stiffness for average load of 1500N

Figure A.5: Dynamic stiffness for average load of 1800N

Figure A.6: Dynamic stiffness for average load of 2100N

A.3 Material Safety Data Sheet Black Dye

6836 BLACK

Date of Preparation: 09/28/2004

Revision: 09/28/2004

Section 1 - Chemical Product and Company Identification

Product Name: 6836 BLACK

Product Class: Pigment

Chemical Type: Non-aqueous colorant

Manufacturer: BJB Enterprises, Inc., 14791 Franklin Avenue, Tustin, CA 92780, Phone (714) 734-8450, Fax (714) 734-8929, (M-Th: 8-4:30, F: 7:30-4), Emergency Phone: Chemtrec (800) 424-9300 or (703) 527-3887

Section 2 - Composition / Information on Ingredients

Ingredient Name	CASRN	% wt
1. Carbon Black	1333-86-4	28

Trace Impurities: N/A

Ingredient	OSHA PEL		ACGIH TLV		NIOSH REL		NIOSH IDLH
	TWA	STEL	TWA	STEL	TWA	STEL	
#1	3.5 mg/m ³	NE	NE	NE	NE	NE	NE

Section 3 - Hazards Identification

☆☆☆☆☆ Emergency Overview ☆☆☆☆☆

Appearance: Black liquid; Odor: Mild. May cause eye and skin irritation. Harmful if inhaled. Use in well-ventilated areas. Emits toxic fumes under fire conditions. This product contains one or more reported carcinogens or suspected carcinogens which are noted NTP, IARC, or OSHA-Z in the other limits recommended column. This product contains pigments which may become a dust nuisance when removed by abrasive blasting, sanding or grinding.

HMIS

H 1

F 1

R 0

PPE†

†Sec. 8

Potential Health Effects

Primary Entry Routes: Eye and skin contact; inhalation of vapors, accidental ingestion. No harmful effects by chronic exposure reported.

Acute Effects:

Inhalation: Heating can generate vapors that could cause headaches, nausea, dizziness, and respiratory irritation if inhaled.

Ingestion: Low oral toxicity.

Eye: May cause minor irritation but no corneal injury.

Skin: Exposure may cause mild skin irritation. Symptoms may include redness and burning.

Carcinogenicity: This product contains carbon black, a substance identified by the International Agency for Research on Cancer (IARC), as a Classification 2b (possibly carcinogenic to humans). Based on the International Agency for Research on Cancer (IARC) conclusion that there is sufficient evidence in experimental animals for the carcinogenicity of carbon black dust and inadequate evidence if carcinogenicity in humans, IARC's overall evaluation is that carbon black dust is possibly carcinogenic to humans (Group B). Consult IARC's Monograph Volume 65. The results of the working group were based on studies involving the inhalation of carbon black and other insoluble fine dust particles. Other routes of entry were not reviewed as part of this study. This dispersion contains carbon black in a "wet out" form and does not pose an inhalation hazard. Good hygiene practices should be followed to minimize exposures to any respirable dusts. The study findings produced results consistent with the massive accumulation of fine dust particles in the lung, which overwhelm the natural lung clearance mechanisms, known as the "lung overload" phenomenon, rather than a specific chemical effect of the dust particle in the lung. Carbon Black has not been listed as a carcinogen by the national Toxicology Program (NTP) or the Occupational Safety and Health Administration (OSHA). The National Institute of Occupational Safety and Health (NIOSH) criteria document on carbon black recommends that only carbon blacks with a PAH level greater than 0.1% be considered suspect carcinogens.

Medical Conditions Aggravated by Long-Term Exposure: None known

Section 4 - First Aid Measures

Inhalation: Not likely. Remove to fresh air environment.

Ingestion: If swallowed, call a physician immediately. Do NOT induce vomiting. Never give anything by mouth to an unconscious person.

Eye Contact: Flush eyes with clean, lukewarm water for 15 minutes. Flush under both upper and lower lids. Obtain medical attention if irritation develops.

Skin Contact: Remove contaminated clothing and wash affected areas well with soap and water. Launder contaminated clothing before use.

Note to Physicians: Treat any ill effects symptomatically.

Section 5 - Fire-Fighting Measures

Flash Point: 200°F (93°C)

Extinguishing Media: CO₂, foam, or dry chemical. Water may be used to keep fire-exposed containers cool until fire is out.

Unusual Fire or Explosion Hazards: Emits toxic fumes under fire conditions.

Fire-Fighting Instructions: Cool fire exposed containers with water spray. Remove containers from fire area if possible. Do not release runoff from fire control methods to sewers or waterways.

Fire-Fighting Equipment: Firefighters should wear positive pressure self-contained breathing apparatus (SCBA) and consider use of unmanned hose holders or monitor nozzles for fighting large fires.

Section 6 - Accidental Release Measures

Spill /Leak Procedures: Provide adequate ventilation and wear personal protective equipment. Eliminate all ignition sources. Prevent product spill from entering sewers, streams or drinking water supplies. Collect liquid or soak up with inert filler or an absorbent, such as dry earth, sand or oil absorbent (sweeping) compound. Collect material into suitable containers for disposal. Wash area with detergent and flush with copious amounts of water.

Containment: For large spills, dike ahead of liquid spill for later neutralization, absorption, clean up and disposal.

Section 7 - Handling and Storage

Handling Precautions: Avoid contact with eyes, skin and clothing. Containers of this material may be hazardous when emptied. Since emptied containers retain product residues (vapor, liquid, and/or solid), all hazard precautions given in the data sheet must be observed.

Storage Requirements: Store in closed, properly labeled containers away from heat, open flames and strong oxidizers.

Shelf life: 24 months from date of shipment under manufacturers recommended storage conditions.

Section 8 - Exposure Controls / Personal Protection

Eye Protection Requirements: Safety goggles or glasses are recommended. Plastic face shield should be worn for complete face protection.

Skin Protection Requirements: Impermeable gloves should be worn. Employees should wash their hands and face before eating, drinking or using tobacco products.

Ventilation/Respiratory Requirements: Exhaust ventilation recommended. An organic vapor cartridge or fresh air supplied respirator (NIOSH certified) may be necessary for certain applications. Consider the type of application, environmental concentrations, and other materials being used concurrently when determining respirator use and selection. Observe OSHA regulations for respirator use (29 CFR 1910.134).

Additional Protective Measures: Safety showers and eye wash stations should be easily accessible to the work area. Training is important. Follow all label precautions.

Section 9 - Physical and Chemical Properties

Physical State: Liquid

Appearance/Odor: Black/Mild

Vapor Pressure: 0.35

Specific Gravity (H₂O=1): 1.11

Water Solubility: NE

% Volatile: None

V.O.C. (ref EPA meth 24): None

Section 10 - Stability and Reactivity

Stability: Stable at room temperature in closed containers under normal storage and handling conditions.

Hazardous Polymerization: Product can undergo hazardous polymerization.

Chemical Incompatibilities: Strong oxidizers such as chlorates, bromates, and nitrates.

Conditions to Avoid: Excessive heat, flame and other possible ignition sources.

Hazardous Decomposition: Carbon monoxide, carbon dioxide, and various hydrocarbons.

Section 11- Toxicological Information

No Toxicological Information Available

Section 12 - Ecological Information

No Ecological Information Available

Section 13 - Disposal Considerations

Waste Disposal Method: In a licensed, permitted facility, incinerate or landfill as a solid after cementation or encapsulation. Do not discharge into waterways or sewer systems. Spilled material, empty containers and unused contents must be disposed of in accordance with federal, state or local environmental control regulations.

Section 14 - Transport Information**DOT**

Not regulated

IATA/ICAO

Not regulated

IMO/IMDG

Not regulated

Section 15 - Regulatory Information**U.S. Federal Regulations:****OSHA:**

This document has been prepared in accordance with the MSDS requirements of the OSHA Hazard Communication Standard.

SARA TITLE III:**Sections 311/312 Hazard Classification:**

None

Section 313: This product contains the following substances subject to the reporting requirements of EPCRA, Section 313 and 40 CFR Part 372:

None

TSCA: This product or its components are listed in or exempt from the TSCA inventory requirements.

This product contains the following substances subject to export notification under Section 12 (b) of TSCA:

None

Section 16 - Other Information

Reason for Issue: New Issue

Prepared By: M. Rose

Approval Date: 09/28/2004

Supersedes Date: N/A

Disclaimer: This information is furnished without warranty, expressed or implied, except that it is accurate to the best knowledge of BJB Enterprises, Inc. The data on this sheet relates only to the specific material designated herein. BJB Enterprises, Inc. assumes no legal responsibility for use or reliance upon these data.

A.4 Technical Data Strain Gauge

CEA Series

CEA-Series gages are the most widely accepted for use in general-purpose experimental stress analysis applications in the world today. These polyimide encapsulated constantan gages feature large, rugged, copper-coated tabs. This construction provides optimum capability for attaching leadwires directly to the tabs, eliminating the need for separate terminals. In most applications, the CEA Series is preferred over the EA Series with options such as E, L, LE and W. Nominal single-plane gage thickness is 0.0027 in [0.069 mm]; stacked rosettes, 0.0039 in [0.099 mm]. The extremely tough but flexible cast polyimide carrier can be contoured to almost any radius. Strain limits are approximately $\pm 5\%$ [50 000 $\mu\epsilon$] for gage lengths 1/8 in [3 mm] or greater, and $\pm 3\%$ for smaller sizes.

OPERATING TEMPERATURES FOR CEA-SERIES GAGES

The thermal output of constantan increases rapidly below -50°F [-45°C]. Static measurements become difficult if temperatures are varying.

Positive zero drift of constantan alloy increases significantly above $+150^{\circ}\text{F}$ [$+65^{\circ}\text{C}$] for single active gages in static measurements. Use half-bridge or full-bridge circuits when possible. Stacked rosettes are limited to a maximum operating temperature of $+150^{\circ}\text{F}$ [$+65^{\circ}\text{C}$].

GAGE PATTERN DATA

actual size

GAGE DESIGNATION

See Note 1

RESISTANCE (OHMS)

CEA-XX-062UR-120
CEA-XX-062UR-350

120 $\pm$ 0.4%
350 $\pm$ 0.4%

OPTIONS AVAILABLE

See Note 2

P2
P2

DESCRIPTION

Small 45° rectangular single-plane rosette in a compact geometry. Exposed solder tab area 0.07 x 0.04 in [1.8 x 1.0 mm]

GAGE DIMENSIONS

Legend:

ES = Each Section

S = Section (S1 = Sec 1)

CP = Complete Pattern

M = Matrix

inch

millimeter

Gage Length	Overall Length	Grid Width	Overall Width	Matrix Length	Matrix Width
0.062 ES	0.222 CP	0.062 ES	0.420 CP	0.32	0.48
1.57 ES	5.64 CP	1.57 ES	10.67 CP	8.1	12.2

GAGE SERIES DATA

See Gage Series data sheet for complete specifications.

Series	Description	Strain Range	Temperature Range
CEA	Universal general-purpose strain gages.	$\pm 3\%$	-100° to $+350^{\circ}\text{F}$ [-75° to $+175^{\circ}\text{C}$]

A.5 Variation in Material Properties of Synthetic Bones Based on Generation

AVERAGE MATERIAL PROPERTIES									
Simulated Cortical Bone (short fiber filled epoxy)									
	Density	Longitudinal Tensile			Compressive			Flexural	
		0.2% yield	strength	Modulus	0.2% yield	Strength	Modulus	Strength	Modulus
	g/cc	MPa	MPa	GPa	MPa	MPa	GPa	MPa	GPa
2nd generation	2.08		172	18.6				275	1.42
3rd Generation	1.70		90	12.4		120	7.6		
4th Generation	1.64	96	106	16.0	112	157	16.7		
Material property based on ASTM D-638 and D-695									
			Density	Transverse Tensile					
				0.2% yield	strength	Modulus			
			g/cc	MPa	MPa	GPa			
		3rd Generat	1.70		76.0	8.6			
		4th Generat	1.64	76	93.0	10.0			

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