Biomechanics of Acute Total Hip Arthroplasty after Acetabular Fracture: Plate vs Cable Fixation

by

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A thesis submitted in conformity with the requirements for the degree of Master of Science

Institute of Medical Science
University of Toronto

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2014

Abstract

The incidence of complex acetabular fractures is increasing worldwide, especially in the elderly. Fixation methods currently used surgically have no biomechanical justification. Therefore, this study compared the standard plate versus three cable techniques for anterior column plus posterior hemitransverse acetabular fractures in synthetic hemipelves. Phase I employed FEA validated by non-destructive mechanical testing. Phase II entailed destructive comparison of the superior cabling method from Phase I versus the standard plating procedure. Phase I identified the Mouhsine cable method as providing the best mechanical stability (i.e. based on lowest peak bone stress) among the cable techniques. Phase II demonstrated the Plate group’s superiority over the Mouhsine group for 2 outcomes (i.e. stiffness and failure force at clinical failure), its inferiority for 2 outcomes (i.e. failure displacement and gapping for posterior markers at mechanical failure), and statistical equivalence for the remaining outcomes. This investigation has implications for surgical repair of this injury.
Acknowledgements

I am extremely grateful to my supervisor, Dr. Emil Schemitsch, for giving me this opportunity of pursuing a Master’s program at the University of Toronto, Institute of Medical Science under his continuous guidance, mentorship and support. You were always there by my side to offer help, whenever needed. You made me feel deeply rooted into the big team of clinicians and researchers. Thank you very much for the great level of trust and responsibility you gave me, and for your obvious generosity over the past and most amazing two years of my life. I believe I will never find such a lovely, respectful and outstanding supervisor.

I regard forming excellent work and personal relationship with Dr. Radovan Zdero to be the greatest life-time gain I have achieved from this research. I am very happy to get to know such a wonderful, helpful and supportive character. Thank you very much for your tireless encouragement throughout my Master’s program, and for the countless hours you spent teaching me many technical and academic aspects of my research.

I owe much gratitude to all the exceptional members of my thesis committee, who provided much insight, encouragement and guidance throughout my studies, namely Drs. Albert Yee and Hans Kreder. I also thank Dr. Joanna Sale for officially supervising me at the Collaborative Musculoskeletal Program. I would also like to thank Dr. Omar Dessouki and Saeid Samiezadeh for their valuable assistance with my research project. I wish to acknowledge all the initiative offered by every staff member at the Institute of Medical Science, and also the grant provided by the Ontario Graduate Scholarship in Science and Technology – David E. Hastings at UofT, which made a big difference in accomplishing my research.

Most importantly, I owe my greatest appreciation to my 3 brothers and sisters-in-law from oldest to youngest: Mike and Janet, Beshoy and Maria, Ilbram and Angie, and particularly my mom, Nadia, who held my hand all the way through my hard times and difficulties. All of them were, are and, I am sure, will continue to be a tireless source of love, comfort and encouragement.

Above all, words cannot express my deepest gratitude to my lovely and loving Lord and Savior Jesus Christ, who is my ceaseless strength, victory, joy, and love. As the Bible says “I can do all things through Christ, who strengthens me” Philippians 4:13.
Contributions

Mina Aziz (author) was responsible for execution of the experiments, data analyses, computational analyses, and preparation of this thesis. The latter included the preparation of each section with all the integrated tables and figures, except for most of the figures in Chapter 2 (Literature Review), where every one of the figures is cited in its caption. The contributions made by others are officially and comprehensively listed below:

Dr. Emil H. Schemitsch (Supervisor and Program Advisory Committee (PAC) Member): Mentorship, laboratory resources, support, and guidance in planning and execution of this body of work.

Dr. Radovan Zdero (Research Director of the Martin Orthopaedic Biomechanics Laboratory): Laboratory facilities, guidance in planning, execution and analysis of the experimental and computational parts of this study.

Dr. Omar Dessouki: Assistance in preparing all the specimens for the experimental portion of the project.

Saeid Samiezadeh: Assistance in the preparation and analysis of the computational portion of the project.

Dr. Albert Yee (PAC Member): Guidance with the planning, execution, and analysis of the project.

Dr. Hans Kreder (PAC Member): Guidance with the planning, execution, and analysis of the project.

Dr. Joanna Sale (Supervisor for the Collaborative Program in Musculoskeletal Sciences): Mentorship and guidance with the various aspects of this specific program.

Dr. Habiba Bougherara: Assistance with some questions about the preparation and analysis of the computational part of the study.


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<td>UHMWPE</td>
<td>Ultra-High Molecular Weight Polyethylene</td>
</tr>
</tbody>
</table>
Chapter 1
Introduction

1.1 Overview

Acetabular fractures consist of two main categories: elementary and complex, each has five subcategories, giving the overall number of 10 distinctive types (Letournel and Judet, 1981). Acetabular fractures are complex, deep, and hard to manipulate and operate on. To achieve good clinical outcomes after an acetabular fracture, a perfect reduction (with a gap of less than 1mm) should be achieved (Boraiah et al., 2009; Letournel and Judet, 1993; Matta, 1996; Mears and Rubash 1986; Rowe and Lowell, 1961; Tile, 1994). This requires the application of the best surgical techniques performed by specialized trauma teams in highly specialized centres applying the best up-to-date evidence-based practice (Carroll et al., 2010; Pagenkopf et al., 2006). Additionally, acetabular fractures are increasing in numbers in the general population, and even more in the elderly because of rapidly aging western populations (Ferguson et al., 2010, Grazier et al., 1984; Jeanotte and Moore, 2007). People older than 60 years represent the most rapidly growing population subset to encounter acetabular fractures (Lonner and Koval, 1995; Mears, 1999), simply due to falls on the greater trochanter (Helfet et al., 1992; Mears and Velyvis, 2002).

By contrast, orthopaedic surgeons specialized in the management of acetabular and pelvic fractures are not numerous. Also, acetabular fractures are under-investigated from the clinical and more specifically the biomechanical point of view. Many techniques widely employed in the fixation of acetabular fractures are not biomechanically validated and their reliability and mechanical stability in fixing those kinds of fractures remain unknown. Not only that, but many kinds of complex acetabular fractures, which represent the more technically and mechanically difficult to manage, are never addressed in the literature with any biomechanical studies, and very few clinical and epidemiological studies. Hence, a focus on acetabular fractures should increase worldwide with more research being conducted, so that orthopaedic and trauma surgeons can rely on evidence-based results in their clinical and
surgical practices for the effective management of the very vulnerable group of elderly patients. Therefore, this thesis will mainly focus on analyzing and comparing the biomechanical aspects of two main classes utilized in repairing these complex fractures, namely the traditional “plate and screws” method versus the newly introduced “cable” method. This thesis will further determine which one of the three cabling techniques employed in acetabular fracture fixation and reported in the literature is preferable in the elderly in the acute setting. A complex acetabular fracture will be presented, which is the associated anterior column with a posterior hemitransverse (AHT) fracture. Moreover, the newly used surgical practice of an acute primary Total Hip Arthroplasty (THA) in the management of selected cases of acetabular fractures in the elderly will be introduced. The effect of the acetabular component not only on solving the hip joint pathology, but also on the biomechanical stability of the fracture fixation constructs will be estimated.

1.2 Thesis Outline

This thesis will adhere to the traditional format of thesis writing, where the introduction will be followed by literature review, research aims and hypotheses, a classical methods section, then a results section, discussion, conclusions, and future directions. Chapter 1 is the “Introduction”, where the reader will get an overview idea about acetabular fractures, current problems and practices, and what will be done during the thesis. This will be followed by Chapter 2, the “Literature Review”, where readers from diverse backgrounds will get to know the anatomy and biomechanics of the hip joint, the epidemiology and classification of acetabular fractures, the various pathways involved in the management of acetabular fractures, and previously conducted biomechanical studies on the fixation of some types of acetabular fractures. Chapter 3 is titled “Research Aims and Hypotheses”, where the research questions about the cutting-edge knowledge of the management of acetabular fractures are raised and the hypotheses are proposed. Chapter 4 “Methods” will follow, which is describes the methodology used in this thesis to make it fully reproducible by other researchers interested in the field in order to ensure that the results produced will be comparable to future results conducted by others. Chapter 5 “Results” presents Phase I data, which include both
experimental testing and FEA, and Phase II data, which compare the standard fixation method using plate/screw versus the best cable method until mechanical failure. Chapter 6 “Discussion” is the main section interpreting the results and discussing important findings on the fixation of complex acetabular fractures in the elderly. Chapter 7 “Conclusions” shortly explains the main outcomes of the current study in brief “take-home” messages. Chapter 8 “Future Directions” shows how the current study can be related to and followed by coming significant studies about the fixation of other complex acetabular fractures in the elderly employing currently applied surgical techniques and procedures.
2.1 Anatomy of the Hip

2.1.1 Descriptive Anatomy of the Hip

2.1.1.1 Hip Bone

The hip bone, also called Innominate bone, Pelvic bone, or Coxal bone (Gray, 2000) is formed of three separate bones, namely the Ilium, Ischium, and Pubis, that unite together over time until it is completely united in adulthood (Fig. 2.1 and 2.2). This union of the bones forms the Acetabulum in the middle of the hip bone. The two hip bones, specifically the Pubic bones, meet anteriorly in the midline of the body at the Symphysis Pubis and both bones, namely Iliac bones, unite posteriorly with the sacrum forming the pelvis region.

The Ilium is the biggest of the three bones and is the flat extended large superior bone. It consists of two parts; the body and the ala, which are distinguished from each other by the arcuate line on the inner surface and the margin of the acetabulum on the outer surface. The body forms the roof i.e. the superior portion of the acetabulum, and is partly articular being part of the Lunate surface of the acetabulum i.e. the horseshoe-shaped smooth region, and partly non-articular consisting part of the Cotyloid Fossa i.e. the central inverted U-shaped depressed region of the acetabulum. On the other hand, the ala consists of external and internal surfaces, anterior and posterior borders, the iliac crest on top, and the acetabular margin below. The characteristic features on the external surface are the three (anterior, posterior, and inferior) gluteal lines, while on the internal surface, the iliac fossa and arcuate line are the most important features. The iliac crest’s surface is broad and has outer and inner lips with an intermediate line. The crest starts anteriorly with the Anterior Superior Iliac Spine (ASIS) and ends posteriorly with the Posterior Superior Iliac Spine (PSIS), also it has the gluteus medius tubercle about 5 cm posterior to the ASIS, on top of the anterior pillar.
The Ischium is the hardest, lowermost, backmost bone of the three. It consists of a body, a superior ramus, and an inferior ramus. The body unites with the other two bones to form the acetabulum and participated into just more than 40% of the acetabulum; forming partly the lunate surface and partly the posterior portion of the Cotyloid fossa. From the body protrudes a thin elongated triangular projection, the Ischial spine, to which some muscles are attached. Above the ischial spine, a large notch is found, the Greater Sciatic Notch (GSN), which is converted into foramen by the attachment of the Sacrospinous ligament. The GSN transmits many important structures, such as the superior and inferior gluteal vessels and nerves, the Sciatic and posterior femoral cutaneous nerves, and the pyriformis. Below the spine, there is a smaller notch called the Lesser Sciatic Notch (LSN), which is transformed into a foramen by the Sacrotuberous and Sacrospinous ligaments. Through the LSN pass the internal pudendal vessels and nerve as well as the obturator internus tendon and nerve. The superior ramus originates from the body and is directed downwards and posteriorly until it ends as the inferior ischial ramus. The posterior surface of the superior ramus forms a large protrusion called the Ischial Tuberosity, which is divided into upper and lower parts by a transverse ridge. The lower part gives origin to the Adductor Magnus muscle and Sacrotuberous ligament, while the upper part gives origin to the Semimembranosus, long head of the Biceps Femoris, and Semitendinosus. The inferior ramus is thin and flat, extending from the superior ischial ramus to the inferior pubic ramus, where it joins with the latter forming a raised ridge.

The Pubis is the smallest and anteriormost bone of the three bones forming the hip. It consists of a body, a superior ramus, and an inferior ramus. The body, as in the previous two bones, attributes to the formation of the acetabulum representing 20% of it. The body is involved in both the lunate surface and the Cotyloid Fossa. The superior ramus extends anteriorly and medially from the body and is divided into two smaller parts, namely a medial flat portion, and a lateral prismatic portion. The medial portion has a large tubercle protruding anteriorly; the Pubic Tubercle, to which the inguinal ligament is medially attached. The superior surface of the lateral portion has a rough ridge, the Iliopectineal eminence, which represents the place of junction between the ilium and pubis.
Fig. 2.1: Lateral view of the hip bone showing all the characteristic features, which will be referred to during the whole thesis (Gray, 2000).

Fig. 2.2: Medial view of the hip bone demonstrating the important features, which will be referred to during the whole thesis (Gray, 2000).
2.1.1.2 Acetabulum

The acetabulum is a central hemispherical depression in the middle of the hip bone; formed by the union of the bodies of the three bones forming the hip bone, namely the ilium, ischium, and pubis (Fig. 2.3) (Gray, 2000). The body of the ilium forms just less than 40% of the acetabulum; the body of the ischium forms a little more than 40% of the acetabulum, while the pubic body constitutes the remaining 20% of the acetabulum. The acetabular surface has two important parts; the lunate surface and the acetabular (Cotyloid) fossa. The lunate surface is the articular region and it has a horseshoe-like shape, while the cotyloid fossa is a central inverted U-shaped depression, connected below to the acetabular notch. The acetabulum, being a hemisphere, has a rim all around except inferiorly, where the rim is missing and the acetabular notch is present. The rim is most prominent and strongest superiorly in the load-bearing region of the acetabulum and called the roof of the acetabulum. The roof is the area of articular surface that forms the arch of a 45-60° angle, extending from the Anterior Inferior Iliac Spine (AIIS) anteriorly to the ilio-ischial notch of the acetabular margin posteriorly. To the rim is attached the glenoidal labrum, which is a circular ligament, further deepening and narrowing the articular surface between the femoral head and acetabulum. The acetabular notch is converted into a foramen by the transverse ligament, through which nutrient vessels and nerves penetrate the joint.
Fig. 2.3: Anterior view of the pelvis in the anatomical orientation, where the acetabulum is facing anteriorly, laterally, and downwards.

2.1.1.3 Hip Joint

The hip is a deep seated joint formed by the articulation of the hemispherical cup-shaped acetabulum and the spherical head of the femur (Gray, 2000). The hip joint is considered an enarthrodial (ball-and-socket) joint. Both articular surfaces are covered entirely with thick articular cartilage to lubricate movements and prevent friction, except for a central point of the femoral head called fovea capitis femoris, where the ligamentum teres is attached, and the cotyloid fossa, which is filled with a mass of fat and covered by the synovial membrane. The hip joint has several ligamentous structures surrounding it, deepening the joint, narrowing the acetabular orifice, providing more support and stability to the joint, holding the joint in place with less muscle exhaustion, and preventing extreme joint movements. These ligaments are as follows: the articular capsule, the pubocapsular, the ischiocapsular, the iliofemoral, the glenoid labrum, the transverse ligament, and eventually the ligamentum teres femoris.

The hip joint is a deep-seated joint and not easily accessible, surrounded by a big number of muscles from all directions. Anteriorly, the psoas major and iliacus cover the joint. Posteriorly, the piriformis, obturator internus and externus, gemellus superior and inferior, as well as the quadrates femoris lie behind the joint. Medially, the joint is surrounded by the
obturator extenus and pectineus. Above and laterally, there are the long head of the rectus femoris and the gluteus minimus. The joint has vascular supply from the peri-acetabular circle formed by multiple nutrient vessels along the acetabular margin, namely the obturator artery, inferior gluteal artery, artery of the roof which is a branch of the superior gluteal artery, plus other local nutrient branches. The joint receive sensory nerve supply from the sacral plexus, sciatic, obturator, and accessory obturator nerves.

2.1.1.4 Hip Muscles and Movements

The hip joint is a multi-axial ball-and-socket joint, and therefore, movements along perpendicular planes occur over a wide arch of motion, namely flexion and extension, adduction and abduction, medial and lateral rotation, and circumduction (Gray, 2000). Muscles surrounding the hip are divided into groups; each is mainly, but not only, responsible for a certain movement of the hip. The main hip flexor is the psoas muscle, helped by the iliacus, but also other muscles assist in hip flexion. Extension is mainly performed by the gluteus maximus. Adduction is mainly carried out by the adductor group of muscles, such as the adductor brevis, longus, and magnus. Hip abduction is mainly exerted by the gluteus medius and minimus. Medial rotation usually occurs by the action of glutei medius and minimus, tensor fasciae latae and the adductors, while lateral rotation is primarily performed by the lateral rotator group, such as the obturators internus and externus, gemelli superior and inferior, and the quadratus femoris. Hip muscles coordinate together to achieve successful hip movement, and hold the position of the pelvic girdle during standing. Also, they integrate in action with neck, shoulder, trunk, and leg muscles to maintain position of the body and maximize stability.

2.1.2 Surgical Anatomy of the Hip

According to Letournel and Judet (1981), the acetabulum is looked at as being seated within the open arms of an inverted Y-structure (Fig. 2.4). This structure represents both the anterior and posterior columns of the acetabulum. The anterior column is much longer than the
posterior one and extends from the anterior portion of the iliac crest superiorly all the way down to the Symphysis Pubis (SP), while the posterior column travels from the lowermost end of the ischium to meet with the posterior aspect of the anterior column just above its mid-point, almost at the level of the upper corner of the GSN, creating a 60° angle. The acetabulum is enclosed within this angle and the apex is filled with condense cortical bone forming the roof for the acetabulum.

The posterior column, also called ilio-ischial column, is triangular in the cross section and is formed of an internal, posterior and antero-lateral surfaces. The internal surface forms the posterior portion of the quadrilateral plate. The posterior surface forms part of the posterior wall of the acetabulum and the subcotyloid groove for the tendon of the obturator externus above and the ischial tuberosity below. The antero-lateral surface represents the posterior portion of the articular surface of the acetabulum above, and the body of the ischium below. The posterior border of the posterior column consists of the ilium above, the GSN in the middle, and then the LSN below.

The Anterior column contains the iliac, acetabular, and pubic segments. The iliac segment extends from the iliac crest superiorly to just above the acetabular margin. Its main features are the anterior pillar, ASIS and Anterior Inferior Iliac Spine (AIIS), which is very close to the superior margin of the acetabulum. The acetabular segment forms the anterior articular surface of the acetabulum, part of the anterior wall of the acetabulum, and the anterior portion of the quadrilateral plate of the acetabulum. The pubic segment constitutes the anteriormost, and slimmest portion of the anterior column, and consists of the superior pubic ramus forming the roof to the obturator foramen. The iliopectineal line is a fundamental structure of the anterior column providing support to the anterio-superior aspect of the acetabulum and is the main guide for evaluating the continuity of the anterior column. Fractures of this line usually indicate fractures of the anterior column. Both columns are connected to the auricular surface of the Sacro-iliac (SI) joint by the sciatic buttress.
Fig. 2.4: Surgical anatomy of the hip bone. Red: Posterior column, White: Anterior column, Blue: Ischio-pubic ramus. A. Lateral view, B. Obturator-Oblique view, C. Iliac-Oblique view (Letournel and Judet, 1981).
2.2 Biomechanics of the Hip

2.2.1 Overview

Force can be transmitted from the head of the femur to the acetabulum causing acetabular fractures and/or dislocation of the femoral head, when the force is high enough to cause this and applied in any of four locations, namely the greater trochanter of the femur, the flexed knee, the foot (when the knee is extended), and posteriorly on the pelvis (Letournel and Judet, 1981). The force of interest in that situation is calculated at the centre of the femoral head, and so, the centre of the acetabulum as well. If a force ‘F’ is applied at point ‘A’ on the greater trochanter of the femur, the force ‘F’ can be analyzed into two perpendicular vectors ‘F’ and f” (Fig. 2.5). The resultant force ‘F’ will act through the centre of the femur ‘C’ and pushes the femoral head against the acetabulum at the impact point ‘I’. The force vector ‘f’ will cause the femoral shaft to rotate in its direction. If the surface of the hip joint was typically spherical and the material was homogeneous, it would be expected that the force would be distributed equally along all points of the surface. However, because this is not the case where bone is heterogeneous, and the presence of double layers of articular cartilage greatly alters the force transmission pathway. Therefore, the force is expected to be distributed in the immediate surrounding circle as vectors of the force ‘F’”, but in lesser quantities yielding an elliptical area of force magnitudes. The sum of these force vectors should amount to the force ‘F’".
2.2.2 Force Applied to the Greater Trochanter

In this condition, only the rotational (internal vs. external) and abductory/adductory movements of the hip have a big impact on the location of force impact on the acetabulum, and hence, on the fracture pattern. However, flexion/extension plays only a minimal role.

For rotational movements, because of the natural anatomical anteversion rotation of the femoral neck, the $F'$ force will be transmitted to the acetabulum in a more anterior position at the junction between the Cotyloid fossa and the anterior part of the lunate surface in the neutral abduction/adduction position. At 20° of internal rotation, the ‘I’ point moves to the centre of the acetabulum, and then, the more the internal rotation is, the more posterior the force is directed and vice versa. Letournel and Judet show sites of force impact at a full range of internal vs external rotations of the hip in a horizontal section, and therefore, types of acetabular fractures that can be predicted (Fig. 2.6).

For abduction/adduction movements, in the neutral position, the force is transmitted to the inner margin of the acetabular roof causing a transverse-type fracture, T-shaped, or both
column fractures. The more the adduction is, the more superior the force travels causing transverse fractures at the acetabular roof, and vice versa, the more the abduction is, the higher the possibility to encounter a fracture below the articular surface of the acetabular roof and is transversely placed.

Fig. 2.6: A. Transverse section through the hip joint showing the effect of different degrees of internal/external rotation of the femur on locations of force transmission. B. Coronal section of the hip joint at 20° of internal rotation illustrating the effect of different abduction/adduction on the lines of force transmission (Letournel and Judet, 1981).

### 2.2.3 Force Applied to the Flexed Knee

When the hip is flexed at 90° and the knee is flexed at 90° as well, force ‘F’ applied to the knee will act on the centre of the femoral head with the magnitude of ‘F’ (Fig. 2.7). If the femoral neck is not fractured, an acetabular fracture may happen. The two hip movements which are relevant here are the flexion-extension, and abduction/adduction combinations, but not rotational movements.

At 90° of hip flexion, in a neutral abduction position, the force impact usually lies upon the posterior wall of the acetabulum leading to a fracture in that region. The more the abduction is, the more anteriorly the force vector shifts reaching at 50° of abduction to the postero-
medial region. The opposite is also correct i.e. the more adduction the hip is, the more posteriorly the force shifts leading to pure posterior wall or even posterior lip fractures and usually associated with posterior dislocation of the hip.

If the degree of abduction/adduction movement is fixed, the more flexed the hip is, the more the force is directed inferiorly leading to fractures of the posterior horn of the acetabulum. The less flexed the hip is the higher the fracture falls e.g. postero-superiorly of the posterior wall of the acetabulum.

Same principles apply to the two other points of application of force, namely on the foot with the knee fully extended e.g. a fall from the second floor vertically on the feet with the knees extended, or on the posterior aspect of the pelvis when the hip is flexed e.g. a stone falling on the back of a miner, while he was leaning forwards.

Fig. 2.7: Transverse section through the hip joint, hip is flexed at 90°, showing the effect of wide range of abduction/adduction on sites of force transmission. Force is applied through the knee (Letournel and Judet, 1981).
2.3 Acetabular Fractures

2.3.1 Epidemiology

2.3.1.1 Incidence and Prevalence

According to the “Fracture and Dislocation Classification Compendium 2007” executed by the “Orthopaedic Trauma Association – Classification, Database and Outcomes Committee” (Marsh et al., 2007), pelvic fractures (#6) are divided into pelvic ring (#61) and acetabular (#62) fractures. First, there are not much data on the exact epidemiology (incidence and prevalence) of pelvic fractures in general, and specifically pelvic ring and acetabular fractures. Furthermore, the literature is severely lacking enough information about the incidence and prevalence of specific types of acetabular fractures in different parts of the world. Second, the records and studies provided for pelvic fractures in Europe, especially the UK, are totally different from those provided for North America (Court-Brown et al., 2006, 2010), leading to the conclusion that the incidence and prevalence of pelvic fractures vary from one region and country to the other region and country, specifically between the UK and USA, where pelvic fractures take two totally different trends, distribution curves, incidences and prevalences.

According to Court-Brown et al. (2006, 2010), the authors chose the fracture population to be the Royal Infirmary of Edinburgh, Scotland because it is the only hospital that deals with orthopaedic trauma in a confined population. The overall prevalence of pelvic fractures represented about 2% of all types of fractures occurring in humans. In a comparison of their databases provided in the years 2000 vs 2007/8, the prevalence of pelvic fractures increased from 1.5% to 1.8% and the incidence increased by 46%. Also, acetabular fractures, unlike pelvic ring fractures, occur in a relatively younger age, with an average of 58 years, are more predominant in males than females 74/26, and follow a type G distribution curve (Fig. 2.8). Type G curve demonstrates a unimodal increase in older females and a bimodal increase in younger and older males with a flat low plateau in the middle age. A complete comparison between the two studies is shown (Tables 2.1, 2.2). The Scotland database shows that acetabular fractures constituted 7.7% in 2000 and almost doubled i.e. 13.4% in 2007/8 of all
pelvic fractures, with an obvious male predominance of 71% in 2000 and 76% in 2007/8 and with an average age of 59.1 in 2000 to 56.7 in 2007/8. This shows a two-fold tendency of acetabular fractures to increase: first, because of the overall increase in prevalence and incidence of pelvic fractures, and second, because of the big increase in their percentage with respect to pelvic fractures. The German Pelvic Study Group 2 (German Trauma Association – Association for Osteosynthesis) reported that acetabular fractures represented 14% of all pelvic fractures in the elderly population over 65 years (Culemann et al., 2010).

Table 2.1: Pelvic fracture characteristics, obtained from Royal Infirmary of Edinburgh, Scotland in years 2000 vs 2007/8 (Court-Brown et al., 2006, 2010).

<table>
<thead>
<tr>
<th></th>
<th>2000</th>
<th>2007-2008</th>
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<tbody>
<tr>
<td>Prevalence (%)</td>
<td>91 (1.5)</td>
<td>127 (1.8)</td>
</tr>
<tr>
<td>Incidence (n/10^5/yr)</td>
<td>17.0</td>
<td>24.8</td>
</tr>
<tr>
<td>Gender (M/F %)</td>
<td>30/70</td>
<td>39/61</td>
</tr>
<tr>
<td>Average age (yrs)</td>
<td>69.6</td>
<td>68.2</td>
</tr>
<tr>
<td>&gt;65 yrs %</td>
<td>72.5</td>
<td>66.1</td>
</tr>
<tr>
<td>&gt;75 yrs %</td>
<td>57.1</td>
<td>46.4</td>
</tr>
<tr>
<td>% of open fractures</td>
<td>1.1 (Gustilo III)</td>
<td>0.8 (Gustilo III)</td>
</tr>
<tr>
<td>Associated fractures #</td>
<td>37</td>
<td>78</td>
</tr>
<tr>
<td>Ratio of index/ass. Fractures</td>
<td>0.4</td>
<td>0.61</td>
</tr>
</tbody>
</table>
Table 2.2: Pelvic vs acetabular fracture characteristics, Royal Infirmary of Edinburgh, Scotland in years 2000 vs 2007/8 (Court-Brown et al., 2006, 2010).

<table>
<thead>
<tr>
<th></th>
<th>2000</th>
<th>2007-2008</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Pelvis</td>
<td>Acetabulum</td>
</tr>
<tr>
<td>%</td>
<td>92.3</td>
<td>7.7</td>
</tr>
<tr>
<td>Average age (yrs)</td>
<td>78.3</td>
<td>59.1</td>
</tr>
<tr>
<td>Gender (M/F %)</td>
<td>26/74</td>
<td>71/29</td>
</tr>
<tr>
<td>Distribution Curve</td>
<td>E</td>
<td>G</td>
</tr>
</tbody>
</table>

Fig. 2.8: Distribution curves for pelvic fractures (type E) vs acetabular fractures (type G) at the Royal Infirmary of Edinburgh, Scotland (Court-Brown et al., 2006).

A study performed by Laird and Keating (2005) on the same study population over 16 years from January 1988 and December 2003 reported that the overall incidence of acetabular fracture in Scotland remained the same of about 3 patients/10^5/year, with a significant decrease in the number of males suffering these fractures. Other studies even suggested a decline in the incidence of acetabular fractures after the introduction and strict application of seatbelt laws (Al-Qahtani and O’Connor, 1996; Blum et al., 1991)

On the other hand, Court-Brown et al. also reported that from the R. Adams Cowley Shock Trauma Center in Baltimore, Maryland, USA in 2007, pelvic fractures represented 11.3% of all fractures, ranking 2nd after spine fractures. Acetabular fractures contributed to more than
50% of all pelvic fractures, showed a male predominance of 70%, and followed a C-distribution curve (Fig. 2.9). C-curve means that there is a unimodal distribution affecting younger males and females. Table 2.3 provides more details regarding pelvic and acetabular fractures in Baltimore in 2007.

Table 2.3: Pelvic vs acetabular fractures, obtained from the R. Adams Cowley Shock Trauma Center in Baltimore, Maryland, USA in 2007 (Court-Brown et al., 2010).

<table>
<thead>
<tr>
<th></th>
<th>Acetabulum</th>
<th>Pelvis</th>
</tr>
</thead>
<tbody>
<tr>
<td>Prevalence (%)</td>
<td>5.7</td>
<td>5.6</td>
</tr>
<tr>
<td>Percentage of pelvic fractures</td>
<td>50.44%</td>
<td>49.56%</td>
</tr>
<tr>
<td>Average age (yr)</td>
<td>42</td>
<td>39</td>
</tr>
<tr>
<td>Gender ratio (male/female, %)</td>
<td>70/30</td>
<td>68/32</td>
</tr>
<tr>
<td>Open (%)</td>
<td>1.9</td>
<td>8.7</td>
</tr>
<tr>
<td>Average ISS</td>
<td>21</td>
<td>23</td>
</tr>
<tr>
<td>Mode of injury</td>
<td>MVA driver 43%</td>
<td>MVA driver 29%</td>
</tr>
<tr>
<td></td>
<td>Fall 17%</td>
<td>Fall 21%</td>
</tr>
<tr>
<td></td>
<td>Passenger 12%</td>
<td>Pedestrian 13%</td>
</tr>
<tr>
<td>Associated fractures</td>
<td>Lumbar spine 17%</td>
<td>Lumbar spine 32%</td>
</tr>
<tr>
<td></td>
<td>Tibial diaphysis 12%</td>
<td>Thoracic spine 13%</td>
</tr>
<tr>
<td></td>
<td>Femoral diaphysis 12%</td>
<td>Cervical spine 13%</td>
</tr>
</tbody>
</table>
Another study by Ferguson et al. (2010) having their database from the Greater Los Angeles area, the incidence of acetabular fractures in the elderly increased by 2.4 times over the 27 years period of the study, when both halves of the study were compared together. In the first half, elderly patients suffering from acetabular fractures formed only 10% of all acetabular fractures vs 24% in the second half of the study, with the mean age moving up from 38 to 45 years. These findings indicate that the incidence of acetabular fractures in patients >60 years is dramatically increasing. The main difference in acetabular fracture patterns between the elderly vs younger population was that displaced anterior column fractures were markedly more common in the elderly population. Therefore, the incidence of acetabular fractures tends to dramatically increase in the elderly because of the aging population (Grazier et al., 1984; Jeanotte and Moore, 2007). The elderly represent the fastest growing section of the population experiencing acetabular fractures (Lonner and Koval, 1995; Mears, 1999), mostly due to simple falls on the greater trochanter (Helfet et al., 1992; Mears and Velyvis, 2002).

### 2.3.1.2 Mechanism of Injury

The main mechanisms of acetabular fractures globally are falls and MVA (Ferguson et al., 2010; Laird and Keating, 2005). Falls include all kinds i.e. simple falls, falls down stairs and falls from heights, and MVA include both drivers and passengers. According to Ferguson et
al., a fall is the most important mechanism of injury in the elderly representing 49.8% of all causes, followed by MVA signifying 37.4%. While in the younger population, the number one cause of injury was MVA causing 66% of all fractures, followed by fall, which represented 17.7%. Laird and Keating also reported similar results showing that fall and MVA are the two most significant causes of acetabular injuries, where falls caused 40.2% and MVA contributed to 38.2% of fractures at all ages. Other causes of injury included bikes, pedestrian versus auto, gun-shot wound, crush injuries, and sports.

2.3.2 Classification

2.3.2.1 History of Acetabular Fracture Classification

Before 1951 and for a long time thereafter, the traditional classification of acetabular fractures was of two broad categories: either central dislocation, or posterior hip dislocation with an acetabular fracture (Letournel and Judet, 1981). Cauchoix and Truchet in 1951 used the above classification, but were not satisfied with it and added two more categories, namely posterior wall acetabular fractures associated with a central dislocation, and trans-acetabular fractures of the pelvis with posterior hip dislocation. This classification was insufficient to explain the mechanisms causing acetabular fractures because any force between the femoral ball and the acetabulum may cause posterior or central dislocation of the femoral head with countless number of different possibilities for acetabular fractures that cannot be accounted for. In 1961, Creyssel and Schnepf used the above-mentioned classification and tried to distinguish principal and accessory fracture lines, and they used the term “trans-acetabular” to describe transverse acetabular fractures. This classification system was rejected because any fracture passing through the acetabulum is of the same importance and cannot be considered “principal” or “accessory”. Also, the term “trans-acetabular” was not an accurate description of transverse fractures because any fracture crossing the acetabulum can literally be named “trans-acetabular”. Since 1960, Letournel and Judet created the current widely used classification of acetabular fractures. The authors kept modifying their classification since then to include all types of acetabular fractures, referring mainly to the essential
structure of the anterior and posterior columns holding the acetabulum in place, and ignoring the femoral head displacement in their classification of fractures.

2.3.2.2 Current Acetabular Fracture Classification

Elementary Acetabular Fractures

Elementary acetabular fractures include these kinds of fractures which affect in part or in total any of the two columns that support the acetabulum, namely fractures of the posterior wall of the acetabulum, fractures of the posterior column, fractures of the anterior wall of the acetabulum, and fractures of the anterior column (Letournel and Judet, 1981). The authors added to them the transverse fracture, being a single fracture passing transversely through both columns of the acetabulum, but not associated with any other fractures.

Associated Acetabular Fractures

The associated fractures mean that they have at least two of the above mentioned elementary fractures. There are five types of associated acetabular fractures: T-shaped fractures, fractures of the posterior column and posterior wall, transverse and posterior fractures, fractures of the anterior column or anterior wall with posterior hemitransverse fracture, and both-column fractures (Letournel and Judet, 1981).

2.4 Associated Anterior and Posterior Hemitransverse Fractures

2.4.1 Nomenclature

This associated fracture type was named so because it involves two components: an anterior component, which mainly includes the anterior column. But it may also affect only the
anterior wall, and a posterior component, which typically resembles the posterior half of a transverse fracture. This is the reason for the term “posterior hemitransverse fracture”.

### 2.4.2 Incidence

Not much data exist in the literature to determine the exact contribution of the anterior with posterior hemitransverse (AHT) to the overall incidence of acetabular fractures. However, only two detailed epidemiological studies about acetabular fractures including the detailed characterization into each of the 10 elementary and associated types, are available in the literature. One was performed by Laird and Keating in the UK, and the other one was done by Ferguson et al. in North America. The latter showed that the fracture type of interest was a common fragility fracture occurring in the elderly and represented 15% of all types of acetabular fractures and about 25% of associated acetabular fractures (Ferguson et al., 2010). However, Laird and Keating reported in their study that this fracture type caused only 6.7% of all acetabular fractures, and 15% of associated acetabular fractures (Laird and Keating, 2005). However, the two populations are totally different as noticed from other parameters mentioned before, and also Ferguson et al.’s study is much larger for two reasons; first, it was run over a period of 27 years vs only 16 years, and second, total number of patients involved was 1309 (all classified into the 10 different types except 14 cases (1.3%) were unknown) vs 351 (of them, only 163 were classified and 188 (53.6%) not classified). Another study stated that the anterior column combined with posterior hemitransverse fracture was particularly common in the elderly (Hessmann et al., 2002), and therefore, this kind of acetabular fractures is usually considered the typical acetabular fracture in the elderly (Culemann et al., 2010).
2.4.3 Morphology

2.4.3.1 Anterior Component

The anterior component of the AHT associated fracture can be either an anterior wall fracture (less common) or an anterior column fracture (more common) (Fig. 2.10) (Letournel and Judet, 1981). If it is an anterior wall fracture, it will have the typical features of anterior wall fractures, where the broken segment can be fully separated or remain in contact with the superior pubic ramus. Infrequently, there may be an associated fracture at the ischio-pubic ramus, or an elevation of the cortex from the quadrilateral plate displaced medially by the femoral head.

If it is an anterior column fracture, which is usually the case, it can be considered as an elementary anterior column fracture, and divided into four categories:

1) Low anterior column fracture, where the fracture line ruptures from the psoas groove just inferior to the AIIS.

2) Intermediate anterior column fracture, where the fracture line exits from the anterior border of the iliac wing somewhere in between the ASIS and AIIF.

3) High complete anterior column fracture, where the fracture ruptures through the iliac crest.

4) High incomplete anterior column fracture, where the fracture line does not go all the way up to the iliac crest, but is directed in that direction.

In either case, the displacement of the anterior component is always severe and associated with anterior dislocation of the femoral head. Also, the anterior component can be looked at as being a pure anterior fracture and considered to be independent of the posterior component.
2.4.3.2 Posterior Component

The posterior component can be regarded as a posterior half of the elementary transverse fracture (Fig. 2.10) (Letournel and Judet, 1981). The fracture line may cross the posterior column of the pelvis at any level. The posterior border of the acetabulum can be traversed by the acetabulum in its inferior quarter or below (most likely), in the middle, or through the upper part. Then, the fracture line crosses the anterior border of the GSN at any level from the superior border, to even splitting the ischial spine. The displacement of the posterior column is much less remarkable than that of the anterior counterpart. This component usually cuts the retro-acetabular portion of the posterior column obliquely from medial to lateral, then crosses perfectly transverse through the posterior column until it cuts the anterior component at a right angle. Occasionally, the posterior fracture may not split the posterior column completely and stop before the dense anterior border of the GSN. This kind of fracture is then regarded as an intermediary stage between pure anterior column fracture and the associated AHT.

Fig. 2.10: Schemes of associated anterior with posterior hemitransverse acetabular fracture. A. High anterior column acetabular fracture with posterior hemitransverse, lateral view, B. medial view, C. Middle anterior column fracture with posterior hemitransverse, lateral view, D. Medial view, E. Anterior wall fracture with posterior hemitransverse, lateral view, F. Medial view (Letournel and Judet, 1981).
2.5 Management of Acetabular Fractures

2.5.1 Overview

Early in the 20th century, almost all treatments offered to acetabular fractures were only conservative for the risk of operating on such difficult fracture types and also due to the risks related to anaesthesia and other pre-operative medical comorbidities (Letournel and Judet, 1981). It was not until Letournel and Judet started to realize that there was tendency to over-evaluate the outcomes of the conservative management and they decided to operatively manage any acutely displaced acetabular fracture with an ORIF, which if achieved anatomic reduction of the acetabular fracture, would lead to a big jump in the successful outcomes after acetabular fractures. The goal of operative treatment of any acetabular fractures is to achieve perfect anatomical reduction of the hip bone and the articular surface of the acetabulum (Boraiah et al., 2009; Matta, 1996). The anatomical reduction is considered the single most important factor for clinical success over the long-term follow-up (Letournel and Judet, 1993; Mears and Rubash 1986; Rowe and Lowell, 1961; Tile, 1994). Lately in the last 20 to a maximum of 30 years, teams of orthopaedic surgeons specialized in the fixation of acetabular fractures started performing the Combined Hip Procedure (CHP) which basically means the simultaneous fixation of the fracture with an ORIF to provide a good bone-stock and support to the acute primary THA performed at the same setting (Boraiah et al., 2009; Herscovici et al., 2010; Mears and Velyvis, 2000; Mears and Velyvis, 2002; Mears et al., 2003; Pagenkopf et al., 2006). Other options for the repair of acetabular fractures especially in the elderly include minimally invasive percutaneous fixation of the fracture, salvage/delayed primary THA after the conservative management failed yielding post-traumatic osteoarthritis, bone deformity, mal-union or non-union, or even after ORIF due to the development of post-traumatic osteoarthritis (Pagenkopf et al., 2003; Tidermark et al., 2003; Vanderschot, 2007). The main goal of all methods of acetabular fracture fixation is to provide a functional hip joint with or close to symptom-free with good to excellent clinical outcomes. Currently, most orthopaedic teams specialized in acetabular fractures are more or less in an agreement regarding the treatment of acetabular fractures in adults. However, in the elderly, there are no clear directions or guidelines to guide orthopaedic surgeons make the
proper decisions for fixing these fractures. After hard work fixing those fractures over many decades, employing various management plans, operative techniques, exposures and fixation methods, as well as reviewing, comparing and thoroughly considering the clinical and radiological outcomes after many years of follow-up for each single procedure, specialized teams lately started to formulate algorithms for the repair of acetabular fractures in the elderly (Carroll et al., 2010; Pagenkopf et al., 2006). These algorithms would indicate when to operate or conservatively manage acetabular fractures, and if to operate, which way to go; either ORIF alone or the CHP i.e. ORIF + acute primary THA. One algorithm was provided by DL Helfet and his team in 2006 (Pagenkopf et al., 2006), which was a bit modified later and published in 2010 (Carroll et al., 2010) (Fig. 2.11). This team designed this management algorithm for the elderly population taking into consideration three important factors that guide reaching a treatment decision, namely patient factors, injury factors, and treatment factors. Patient factors involve age, severity of osteoporosis, underlying medical conditions, preexisting osteoarthritis, and pre-injury activity level and mental status. Injury factors include mechanism of injury, fracture specifications, and associated injury. Treatment factors represent the selected management plan, time between the injury and management, and perioperative complications (Pagenkopf et al., 2006).
Fig. 2.11: Algorithm for the management of acetabular fractures in the elderly, developed by the specialized team of Helfet and co-workers (Carroll et al., 2010).

**ACETABULAR FRACTURE IN AN ELDERLY PATIENT**

**Medical Evaluation for Operative Risk Assessment**

- Deemed unacceptable risk for surgery
- Deemed acceptable for surgery

**Treatment Determinants:**

- **Patient and Injury Variables**
  - I. Non-ambulatory.
  - II. Has severe osteopenia.
  - III. Low Transverse fracture with intact roof arc angles (>45°).
  - IV. Posterior Wall fracture involving <25% of the posterior wall with stable hip joint.
  - V. Minimally or non-displaced fracture.
  - VI. Stable fracture.*
    - *As defined by: Tometta P, JAAOS, 2001

- A. Expectation of an acceptable reduction with ≤3-4 hours operative time through a single non-extensive approach.
  - a. Non-operative treatment
    1. Rapid mobilization, toe-touch weight bearing
    2. Secondary Total Hip Arthroplasty, as indicated

- A. Irreducible through single, non-extensive exposure.
  - a. Open Reduction & Internal Fixation

- B. Protracted surgery time expected (>3-4 hours).
  - a. Open Reduction and Internal Fixation with primary total hip arthroplasty

- C. Severe acetabular impaction / comminution
- D. Displaced femoral neck fracture.
- E. Significant femoral head fracture/impaction.
- F. Significant pre-existing arthrosis.
- G. Severe Osteoporosis
2.5.2 Conservative Treatment

2.5.2.1 Indications

Conservative management of acetabular fractures is mainly adopted in the following conditions (Fig. 2.11) (Boraiah et al., 2009; Carroll et al., 2010; Gross et al., 1993; Herscovici et al., 2010; Letournel and Judet, 1981; Mears and Rubash, 1986; Mears and Velyvis, 2000; Pagenkopf et al., 2006):

1) Medical co-morbidities that make the risks to operate much higher than the benefits the patient may achieve.
2) Immobility: because if the patient is immobile, even if the surgeon achieves perfect reduction and fixation of the acetabulum, there will be no much gain for the patient.
3) Undisplaced, minimally-displaced, or stable fractures.
4) Pre-existing osteoarthritis or severe osteoporosis in a rather non-operable patient rendering a successful reduction of the fracture hard to accomplish.
5) In some cases of both-column fractures with osteopenia, comminution, and displacement if secondary congruency can be achieved by traction.

2.5.2.2 Methods

Usually bed rest with passive movements is encouraged in the beginning that increases with time until walking with crutches can be accomplished at 6 weeks, followed by full weight-bearing by 10 weeks (Letournel and Judet, 1981). Traction either skeletal or skin traction is not really recommended in conservative management because it does not correct any displacement, which is usually rotational and not translational (Matt et al., 1986; Vanderschot, 2007). Also, using pins in the greater trochanter to apply traction usually fails because of slippage out of the bone and also predisposes to local infections. Longitudinal traction does not work as well. Manipulative reduction of a displaced femoral head is mandatory to prevent avascular necrosis (Letournel and Judet, 1981; Vanderschot, 2007).
2.5.2.3 Complications

The conservative treatment may cause residual pelvic deformities because it does not reduce the fracture. Therefore, occult or frank non-union or mal-union of the fracture may be the ultimate result. Also, this way of management may lead to loss of the bone-stock of the acetabulum, or the femoral head, resulting later in loosening of the implants installed in salvage (delayed) THA. The radiological loosening rates reached 30% for femoral components and 40-53% of acetabular components (Gross et al., 1993; Helfet et al., 1992; Herscovici et al., 2010; Mears and Rubash, 1986; Mears and Velyvis, 2000; Romness and Lewallen, 1990).

2.5.3 Surgical Treatment

2.5.3.1 Surgical Access

There are mainly four approaches for fixing acetabular fractures, depending on the nature and location of the fracture, namely the Kocher-Langenbeck, iliinguinal, extended iliofemoral, and combined approaches. The first two are the two mainly used approaches. The Kocher-Langenbeck is a posterior technique, and therefore, mainly used for posterior wall, posterior column, and associated posterior column and posterior wall fractures. It is also selected for most pure transverse, or transverse with posterior wall fractures, and also for some T-shaped fractures (Matta, 1996). The iliinguinal approach allows an anterior access of the hip bone, and therefore, is employed in anterior wall, anterior column, and anterior with posterior hemitransverse fractures. It is also used in most cases of both-column fractures and in selected cases of pure transverse fractures. The extended iliofemoral approach gives access to a larger area of the hip bone, and hence, utilized in more complex acetabular fractures, namely transverse with posterior wall, T-shaped, both-column fractures if difficulties with reduction through the traditional approach are anticipated (Matta, 1996). Finally, combined anterior and posterior approaches may be employed if a satisfactory reduction of the fracture cannot be achieved using only anterior or posterior approach, and therefore, joined by
another incision from the opposite side in order to achieve anatomical reduction which is the main goal of the whole surgical procedure.

2.5.3.2 Open Reduction and Internal Fixation (ORIF)

Indications

Currently, ORIF is the mainstay for all displaced acetabular fractures (Helfet et al., 1992; Kreder et al., 2006; Matta, 1996; Mayo, 1994; Murphy et al., 2003; Routt and Swiontkowski, 1990; Ruesch et al., 1994; Wright et al., 1994) because it usually allows the achievement of perfect anatomical reduction of the fracture, and therefore, reducing the risk of post-traumatic osteoarthritis (Matta, 1996; Matta et al., 1986). Even if the latter happens (De Ridder et al., 1994; Kreder et al., 2006; Matta, 1996; Mayo, 1994; Ruesch et al., 1994; Wright et al., 1994), having an anatomical reduction will help accomplishing better clinical and radiological outcomes of a belated THA, with much lesser rates of radiographic and symptomatic loosening (Helfet et al., 1992). ORIF may use any of the four surgical accesses discussed above depending on the type of fracture, and then, buttress or reconstruction plates alone, lag screws alone (not preferable), both, or lately cables can be applied to produce anatomical reduction (Kang and Min, 2002; Mears and Shirahama, 1993; Mouhsine et al., 2002; Mouhsine et al., 2004).

To sum up, if the patient is fit for surgery, any displaced acetabular fracture should be treated with ORIF. If the fracture was undisplaced or minimally displaced, a conservative management could be a valid option. In select cases with displaced fractures in the elderly, an acute THA with the ORIF might be tried, discussed in details below.

Complications

ORIF is considered the “gold standard” for fixing displaced acetabular fractures; however, it has some complications related to it (De Ridder et al., 1994; Jimenez et al., 1997; Kreder et al., 2006; Matta, 1996; Mayo, 1994; Ruesch et al., 1994; Wright et al., 1994), as follows:
1) Osteoarthritis
2) Avascular necrosis of the head of the femur
3) Heterotopic ossification
4) If a delayed THA is needed for any reason (either any of the complications, or any other reason), it will be very difficult procedure to perform due to the dense scar tissues, heterotopic ossification, deformed anatomy, obstructive hardware, and avascularity of the acetabulum and hip muscles.

2.5.3.3 Combined Hip Procedure (CHP)

Overview

There is an agreement that displaced acetabular fractures are best treated with ORIF providing good to excellent results (Matta et al., 1986; Matta and Mernt, 1988; Mayo, 1994; Helfet et al., 1992). However, many studies reported that there are inherent factors that oppose successful outcomes for this technique in the elderly (Kreder et al., 2006; Matta and Mernt, 1988; Moed et al., 2002; Murphy et al., 2003; Wolinsky et al., 1996), such as pre-existing osteoarthritis, osteoporosis, femoral head or neck fractures, or marginal impaction. Therefore, performing an acute THA simultaneously for these select elderly patients may greatly improve the outcomes through a single operation, provide high quality activity level to those patients, and avoid the “wait-and-see” attitude for a belated THA due to avascular necrosis of the femoral head or post-traumatic arthritis (Herscovici et al., 2010; Mears and Velyvis, 2002). The delayed THA would be also difficult as surgeons have to deal with much dense scar tissues, heterotopic ossification, distorted acetabular anatomy with mal- or non-unions, and atrophy or avascularity of the hip muscles (Harris, 1969; Mears and Shirahama, 1998; Mears and Velyvis, 2000). All these factors would lead to longer surgical times with much blood loss (Bellabarba et al., 2001; Mears and Velyvis, 2000; Mears and Ward, 1996; Weber et al., 1998).
**Indications**

The CHP is indicated in the following conditions (Fig. 2.11) (Herscovici et al., 2010; Joly and Mears, 1993; Mears and Velyvis, 2000; Mears and Velyvis, 2002):

1) Intra-articular comminution  
2) Full thickness abrasion of the articular cartilage  
3) Impaction of the femoral head  
4) Impaction of the acetabulum (occupying >40% of the joint surface and weight-bearing region)  
5) Pre-existing arthritis  
6) Hip dislocation  
7) Difficult non-reconstructable fractures  
8) Osteopenia

**Complications**

Some complications may still happen with CHP and are mainly due to its lengthy procedure leading to longer anaesthetic times, more blood loss occurs with higher transfusion rates. Also, this procedure may become very technically demanding especially it is mainly performed in elderly patients. Also, extensive pre-operative lab workups are needed, which may cause a delay in the time from injury to surgery (Herscovici et al., 2010).

**2.6 Biomechanical Studies For Acetabular Fracture**

**Fixation**

Biomechanical studies done on acetabular fractures so far are scarce and do cover neither all types of acetabular fractures nor all aspects of fixation methods currently used by specialized orthopaedic teams in big trauma centres worldwide. Few biomechanical studies in the literature are available to address the biomechanical stability of few different methods
applied for the fixation of some elementary acetabular fractures, namely pure transverse fracture (Chang et al., 2001; Gililland et al., 2013; Khajavi et al., 2010; Sawaguchi et al., 1984; Shazar et al., 1998) and posterior wall fracture (Goulet et al., 1994; Liu et al., 2010; Olson et al., 1995; Yuntong et al., 2013). Otherwise, there are just sporadic studies that hardly tackle any other fractures of the acetabulum, particularly the associated types. Only one biomechanical study was conducted to address some fixation methods for each of T-shaped (Simonian et al., 1995), anterior with posterior hemitransverse (Culemann et al., 2010), and both-column (Wu et al., 2013) fractures.

Relatively few previous biomechanical studies using both cadaveric and synthetic hemipelvises addressing various fixation techniques for mainly transverse and posterior wall acetabular fractures are currently available in the literature (Chang et al., 2001; Culemann et al., 2010; Gililland et al., 2013; Khajavi et al., 2010; Mehin et al., 2009; Sawaguchi et al., 1984; Shazar et al., 1998; Simonian et al., 1995; Wu et al., 2013). Due to different variables in the experimental preparations and setups, these studies showed a wide range of outcomes and values, with an overall stiffness (129.9 to 3130 N/mm), failure strength (848 to 1012N), failure displacement (0.036 mm to 5.8 mm), etc. However, no biomechanical studies in the literature have ever assessed cable fixation of acetabular fractures.

Acetabular fractures, as discussed before, represent more than 14% of all pelvic fractures (Court-Brown et al., 2010; Culemann et al., 2010). In the elderly, fractures involving the anterior wall/column are particularly prevalent (Ferguson et al., 2010; Hessmann et al., 2002). Of those, the AHT fracture is considered to be the typical fragility fracture in the elderly (Culemann et al., 2010; Hessmann et al., 2002). However, only Culemann et al. studied four different fixation methods for the AHT fracture. The authors, using synthetic and cadaver bones, compared the 3.5mm, 12-hole reconstruction plate mounted on the pelvic brim fixed with 1 or 3 periarticular long screws, the same plate with an H-plate underneath to support the quadrilateral plate, a locking 3.5mm, 12-hole reconstruction plate, and a locking titanium fixator with multidirectional interlocking screws mounted on the inner side of the pelvis i.e. the quadrilateral plate. The authors applied three preload cycles, followed by one load cycle for both types of bones. A measuring ultrasonic device with two pairs of sensors were installed on fixed pre-chosen locations on the specimens in reference to a fixed point on
the quadrilateral plate in order to measure the amount of rotational and translational displacement occurring at the quadrilateral plate, which was regarded as the medial displacement of the femoral head. Based on the results, the authors concluded that the standard reconstruction plate on the pelvic brim is the superior method of fixation as it yielded the shortest displacement of the quadrilateral plate. They also recommended the locking titanium fixator to be a good alternative as it produced a close result to the standard plate and can be inserted through the Stoppa approach, which is a less invasive incision than the ilioinguinal approach used for the standard plate. However, Culmann’s study has many limitations, which will be fully mentioned in the “Discussions” section with the main drawback being the employment of only one specimen for each fixation method. The latter may not be accurate because wrong steps may have been done during specimen preparation giving totally off results for this method, rendering it ineffective.
Chapter 3  
Research Aims and Hypotheses

3.1 Research Aims

This study was conducted to compare the biomechanical stability of four different widely-used surgical methods employed in the fixation of an anterior column with a posterior hemitransverse (AHT) associated acetabular fracture in the elderly as part of the combined hip procedure, which involves the open reduction and internal fixation (ORIF) of the acetabular fracture, followed by a total hip arthroplasty (THA) at the same surgical setting. The four surgical methods are: 1) a plate mounted to the posterior column accompanied with a lag screw inserted into the anterior column, which is considered the “standard” technique, 2) The Kang and Min (2002) cable fixation technique, 3) The Mears and Shirahama (1998) cable fixation technique, 4) The Mouhsine et al. (2002) cable fixation technique.

After a preliminary comparison of the four methods, the mechanical superiority of either the conventional plate/screw construct or any of the cable techniques can be established. Also, the best one of the three cable configurations can be decided. Finally, the most stable method out of the four for the optimal surgical fixation of this type of fracture will be identified, shedding light on the various indications for the application of the different repair techniques in the elderly.

A secondary goal of this thesis is to determine the biomechanical effect of the prevalent practice of inserting the acetabular component of a THA on the overall stiffness and stress distribution of the hip bone, as well as its contribution to the success of the fixation techniques employed in complex acetabular fractures experienced by the elderly population.
3.2 Speculated Hypotheses

The hypotheses proposed in this thesis are:

1) Cable fixation of complex acetabular fractures, particularly the AHT, in elderly patients is equivalent or even superior to the traditional plate / screw fixation of these fractures.

2) Cable fixation procedures employing two cables are not necessarily more mechanically stable than those applying just one cable.

3) The acetabular cup has a big impact on the overall stability and success of the fixation of complex acetabular fractures.
Chapter 4
Methods

4.1 Overview

In Phase I, non-destructive experimental and computational analyses were done on a series of six synthetic hemipelvis groups, which were created to simulate various repair methods for an acetabular column plus posterior hemitransverse (AHT) acetabular fracture. One group was the intact control group, while the other five groups were as follows: intact hemipelvis with an acetabular cup component of a total hip replacement (THR); fractured hemipelvis with an acetabular cup repaired using a posterior column plate and anterior column lag screw; fractured hemipelvis with an acetabular cup repaired using the Mears and Shirahama (1998) cable method; fractured hemipelvis with an acetabular cup repaired using the Kang and Min (2002) cable method; and fractured hemipelvis with an acetabular cup repaired using the Mouhsine et al. (2002) cable method. The six groups were experimentally and computationally mounted to fixed supports at both the sacro-iliac and symphysis pubis regions and exposed to an acetabular non-destructive compressive force of 200 N. For validation purposes, computational model stress maps were then directly compared to experimental results. Once validated, the computational model was then reanalyzed using a clinical-level force of 2207 N, i.e. 3 x body weight for 75 kg person, to simulate forces during regular daily activities such as walking, ascending/descending stairs, rising from a chair, etc. Stress maps were then analyzed using “highest stiffness” and “lowest peak bone stress” criteria to predict the most biomechanically stable cable repair method. In Phase II, the plate/screw method (which is a traditional “gold standard” approach) was then directly compared to the best cable method from Phase I using destructive experimental measurements of stiffness, failure displacement, failure force, failure energy, fracture “sliding”, and fracture “gapping”.
4.2 Phase I – Non-Destructive Mechanical Analysis

4.2.1 Experimental Methodology

4.2.1.1 Hemipelvis Properties

Eighteen large, left, fourth generation, composite synthetic hemipelvises were obtained (Model #3405-1, Sawbones, Vashon, WA, USA). This model is based on the pelvic anatomy of an actual human adult male (age, 45 years; height, 183 cm; weight, 91 kg) (Fig. 4.1). This hemipelvis model was composed of an outer cortical shell and an inner cancellous core in order to simulate normal bone quality. The cortical shell was made from a predetermined mixture of epoxy resin matrix with short glass fibers having an overall density of 1.64 g/cc, a compressive Young’s modulus of 16.7 GPa, a tensile Young’s modulus of 16 GPa, a compressive strength of 157 MPa, and a tensile strength of 106 MPa, and a Poisson’s ratio of 0.26. The cancellous matrix was made from polyurethane foam with various available densities ranging from 0.16 g/cc to 0.27 g/cc. Lower densities are not available since the cancellous material would be mechanically unstable within the cortical shell. Therefore, a “cellular-type” cancellous core density of 0.16 g/cc was utilized to simulate osteoporosis and replicate a weaker screw grip into the bone, which is the condition in real-life osteoporosis. The cancellous material had a Young’s modulus of 155 MPa, a compressive strength of 6 MPa, and a Poisson’s ratio of 0.3. Both cortical and cancellous are considered to be homogenous, isotropic, and linearly elastic.
Fig. 4.1: Synthetic hemipelvis (Model #3405-1, Sawbones, Vashon, WA, USA) demonstrating its dimensions; a) 235mm, b) Diameter of the acetabulum= 55mm, c) 140mm, and d) 175mm.

4.2.1.2 Specimen Preparation

Reaming

Each specimen in the four fixation groups was firmly mounted onto a lab bench using a large C-clamp and was reamed using a series of stainless steel acetabular reamer domes (Catalog #7136-2755/59, Smith and Nephew Inc., Memphis, TN, USA) connected to a reamer handle (Catalog #7136-2279, Smith and Nephew, Memphis, TN, USA). The latter was installed into a corded power drill. The reamer sizes that were used started from a diameter of 55 mm, increased sequentially by 1 mm all the way up to 59 mm (Fig. 4.2). The latter is 1 mm larger in diameter than the outer diameter of the applied acetabular cup to compensate for the 1 mm bone loss generated during fracture creation, i.e. jig sawing, so the inserted cups will be perfectly fit into the acetabulum without any gaps. The initial reamer was more vertically oriented to allow the reamer to be taken down to the medial wall; however, subsequent reamers were oriented 45 degrees superio-medially and neutral (i.e. no ante- or retro-version)
in the sagittal plane, which is the standard surgical practice for reaming and final acetabular cup alignment. Excess medialization of the cup was avoided to preserve the thin medial wall, and hence, provide good support to the acetabular shell. Reaming was the first step performed in because it would be impossible to precisely ream the hard cortical shell and cancellous core after creating the complex AHT fracture.

**Fracture Creation**

Each specimen that was firmly mounted using a C-clamp onto a lab bench was then had a simulated fracture created using a 120V, 1-1/8” corded reciprocating jigsaw (Model #RS7, Bosch, Home Depot, Canada). An anterior fracture was created first extending superiorly from ½ inch (12.7 mm) anterior to the gluteal line on the iliac crest, cutting the acetabular ring in the anterior third until reaching the Obturator Foramen, and exiting inferiorly close to the junction between the Inferior Pubic Ramus and Inferior Ischial Ramus. The posterior portion of the hemipelvis was then flipped so that the cut surface faced the lab bench and an oblique hemitransverse fracture was created being high medially at the tip of the Greater Sciatic Notch and low laterally at the dome of the acetabulum, thus making an angle of 40 degrees with the transverse plane. Anteriorly, the hemitransverse fracture met with the anterior fracture just above the Arcuate Line of the Ilium, creating a small triangular area with that line, which represents the typical clinical pattern of the AHT associated acetabular fractures (Fig. 4.2).

**Fracture Reduction**

Each specimen was divided into 3 fragments after creating the AHT fracture: an anterior fragment, a posterior-superior fragment, and a posterior-inferior fragment. To fix the fracture with the different methods, these 3 pieces needed to be anatomically reduced first. A reduction clamp and four 2.5 mm K-wires were used. To reduce the anterior fracture, the reduction clamp was installed first and connected two holes in the cortex across the fracture. The first hole was provided by the manufacturer at the base of the anterior inferior iliac spine (AIIS), while another superficial hole was created in the cortex 1.5” (38 mm) above the superior rim of the acetabulum using a 2.7 mm drill bit. These two holes allowed the reduction clamp to firmly grip the two fragments across the fracture. In addition, two parallel
K-wires were inserted across the anterior fracture at two different levels above the reduction clamp. For the reduction of the transverse fracture, other two K-wires were inserted divergently across the fracture line at two different levels (Fig. 4.2). All the reduction tools were kept in place until the specimen was accurately fixed and the acetabular component with the liner was inserted. This order of events would replicate the typical surgical technique and achieve the best possible reduction, and hence, biomechanical stability for each of the four different fixation methods.

**Cup Insertion**

After creating the fracture, reducing and fixing it, an acetabular component of a THA was inserted (Fig. 4.3) to simulate the clinical situation of the simultaneous fixation of complex associated acetabular fractures and performing a THA at the same setting in elderly patients with osteoporosis, osteopenia, impaction or comminution of the acetabulum or femoral head. The fixation is performed first in a trial to provide good support for the acetabular cup and minimize medial and superior displacement. For this purpose, a size G with an outer diameter of 58mm, multi-hole acetabular shell (Reflection, Model# 71335158, Smith and Nephew Inc., Memphis, TN, USA) was installed. The shell was made of titanium alloy (Ti-6Al-4V), which would provide reasonable stiffness and at the same time is light in weight. The outer surface of the shell was porous to stimulate bone growth and entanglement of bone fibers and cells around these small pores at a micro-level leading to better mechanical stability. The acetabular shell was connected to a straight acetabular shell impactor (R3 straight shell impactor, catalog# 7136-4450, Smith and Nephew Inc., Memphis, TN, USA) and an X-Bar (Catalog# MT-2201, Smith and Nephew Inc., Memphis, TN, USA) was mounted on top to allow proper alignment of the cup with 45 degrees of abduction and neutral version of the cup. A mallet (Reflection mallet, Catalog# 7136-2106, Smith and Nephew Inc., Memphis, TN, USA) was used to tap on the impactor for proper insertion of the shell providing good contact with the medial wall of the acetabulum. The X-Bar was removed and the impactor was then unscrewed from the shell. Four pilot holes were drilled using a 3.5mm, 50mm long flexible screw drill (Catalog# 7136-2950, Smith and Nephew Inc., Memphis, TN, USA). The orientation of the pilot holes and screws exactly matched the real-life surgical practice. A depth gauge was used to measure the depth of the holes for choosing the right length screws.
Four screws were, therefore, utilized to keep the shell in place and help in the fixation of the complex nature of AHT fractures. Two 6.5mm, 25mm cancellous screws (Universal Cancellous, Model# 7133-6525, Smith and Nephew Inc., Memphis, TN, USA) were inserted into the anterior column of the acetabulum. One 6.5mm, 40mm cancellous screw (Model# 7133-6540) was placed into the dome of the acetabulum. The fourth cup screw was 6.5mm and 25mm long, which was inserted into the Ischium. All cup screws were made of titanium alloy (Ti-6Al-4V). After securing the shell, an UHMWPE size G liner, with an outer diameter of 58mm, an inner diameter of 32mm, and 20 degrees of overhang (Reflecion XLPE, Model# 7133-3336, Smith and Nephew Inc., Memphis, TN, USA) was placed in the correct location using a liner impactor (Fig. 4.3). The tabs on the outer surface of the liner were perfectly fitted into the indentions on the inner edge of the shell. The liner was then checked to ensure appropriate placement. All the procedures required for acetabular cup insertion were performed by an orthopaedic surgeon.

Fig. 4.2: Reaming of the acetabulum, creating the AHT fracture, and reducing it.
Fracture Groups

**Intact:** The “Intact” hemipelvis group. specimens were not fractured nor repaired and acted as a baseline or control group.

**Cup:** The “Cup” group was composed of intact hemipelvises equipped with an acetabular metal cup and polyethylene liner, in order to detect the influence of a THR’s acetabular component on the stiffness and stress distribution.

**Plate:** The “Plate” group was composed of hemipelvises with an AHT acetabular fracture repaired with a posterior column plate and anterior column lag screw (Fig. 4.4). This is considered the “gold standard” of care applied for complex acetabular fractures and is the most commonly used technique performed by the majority of orthopaedic surgeons specializing in trauma and pelvic fractures (Knight and Smith, 1958; Letournel and Judet, 1993; Levine, 1943; Matta, 1996). For each specimen of the plate group, a 3.5, 6-hole, 70mm long stainless steel reconstruction plate (Model# 71809516, Peri-Loc, Smith and Nephew
Inc., Canada) was employed (Fig. 4.4). The plate was bent by an orthopaedic surgeon using plate benders (Small Fragment tray, Smith and Nephew Inc., Canada) to perfectly conform the convex contour of the posterior column of the acetabulum and the upper end of the Ischial Tuberosity. Each plate was fixed using four stainless steel cortical screws (Small Fragment tray, Smith and Nephew Inc., Canada), two above and two below the transverse fracture line. Four pilot holes were first drilled using a 2.7mm drill bit, a hole for each screw was prepared, and a depth gauge was then used to choose the right length screw for each hole. All the screws used for plate fixation were 3.5mm stainless steel cortical screws. The lowermost screw was 60mm long placed all the way into the Ischium. The 2nd lower screw was 44mm long and inserted obliquely into the Ischium, where its tip can be seen at the postero-superior corner of the Obturator Foramen. The uppermost screw was 46mm long and directed anteriorly and medially into the Ilium. The 2nd upper screw was directed more medially into the Ilium. Careful attention was paid during the insertion of the upper plate screws to avoid the joint i.e. reamed acetabulum. A 4.0mm, 100mm long fully threaded stainless steel cancellous lag screw was introduced cautiously into the anterior column of the acetabulum. The lag screw extended obliquely from a point on top of the Inferior Gluteal Line 1.25” above the superior acetabular rim and 0.5” behind the anterior fracture line, into the anterior column of the acetabulum, Superior Pubic Ramus, and ended in the Pubic Body. A larger 6.5mm 90mm stainless steel cancellous lag screw was tried before the 4mm one, but unfortunately, did not work out because of its bigger diameter, while the anterior column of the pelvis was thinned out more by reaming of the acetabulum. As a result, the 6.5mm screw bulged into the roof of the acetabulum and could not be placed correctly. All plate screws as well as lag screws were precisely oriented, placed and securely tightened by an orthopaedic surgeon based on his experience and “subjective feel”.

Mears: The “Mears” group was composed of hemipelvises with the same AHT fracture repaired with the Mears and Shirahama (1998) cable method (Fig. 4.5). This technique was initially demonstrated by Mears and Shirahama (1998). For each specimen, a drill bit 2.7mm was employed to create a thru-hole at the base of the AIIS, which was directed from postero-lateral to antero-medial. Through this hole, a 2.0mm braided stainless steel cable with a clamp (Accord, Model#71340008, Smith and Nephew Inc., Memphis, TN, USA) was passed from out in (Fig. 4.5). The cable was then pulled along the inner table of the hip bone,
crossing the point of meeting of both fractures and the Arcuate Line of the Ilium, proceeding along the quadrilateral plate of the acetabulum to the Lesser Sciatic Notch (LSN), where it curves around from medial to lateral. The cable then crosses the lateral surface of the Ischium to the postero-superior end of the Obturator foramen, where it curves medially again, going up along the quadrilateral plate, where it makes a cross sign with the previous path of the cable on the inner side of the hip bone, till it reaches the Greater Sciatic Notch (GSN). At the GSN, it changes path again from in out, where it continues anteriorly to meet the beginning of the cable above the superior rim of the acetabulum. The end of the cable was then run through the clamp at the head of the cable creating a loop. The cable was then held in place till the free end of the cable was passed through the tip of the Accord cable tensioner (Catalog# 7136-0020, Smith and Nephew Inc., Memphis, TN, USA). Before using the tensioner, the “Reset Switch” was turned clockwise to reset the tensioner, while the “Locking Handle” was open. The free end of the cable was then passed all the way in through the tensioner until the tensioner was in direct contact with the clamp with no gaps. The “Locking Handle” was then turned clockwise to hold the cable firmly and prevent any slippage of the cable under tension. The “Lever” on the tensioner was then pumped until the “Tension Gauge” reached 100Kg. The clamp was checked to ensure that the cable ran through it properly and the clamp screw was strongly tightened until clicking sound was heard using a fitting surgical screwdriver. The cable was then loosened from the tensioner by turning the “Locking Handle” counterclockwise and the tensioner was removed.

**Kang**: The “Kang” group was composed of hemipelvises with the same AHT fracture repaired with the Kang and Min (2002) cable method (Fig. 4.6). Two 2.0mm braided stainless steel cables (Accord, Model#71340008, Smith and Nephew Inc., Memphis, TN, USA) were used for this fixation method (Fig. 4.6), as originally discussed by Kang and Min (2002). The first cable was wrapped around the hip bone from just above the AIIS, turning around the anterior border of the bone from lateral to medial, then going posteriorly on the inner table of the Ilium crossing the Arcuate line to the GSN, where it curves around it from in out running on the outer surface of the Ilium, until it passed through the clamp forming a full loop. The clamp rested on the Iliac bone about 1” above the superior rim of the acetabulum. Then, the same tensioning technique was applied as described above in the “Mears fixation” section. This cable mainly provided fixation across the anterior fracture
component of the AHT fracture pattern. The other cable was also run just above the base of the AIIS, going around the anterior border of the hemipelvis from lateral to medial, but this time the cable was passed posteriorly and inferiorly on the medial side of the pelvis, crossing the Iliac Fossa, the arcuate line, and across the anterior fracture. The cable was then passed across the quadrilateral plate providing support to this important region necessary for the success of fixation, and turned around the LSN to meet the other end and pass through the clamp about 0.5” above the superior acetabular rim. The same tensioning technique was employed as discussed in “Mears fixation” section. This cable provided compression across both components of the AHT associated acetabular fracture as well as supported the quadrilateral plate of the acetabulum.

**Mouhsine:** The “Mouhsine” group was composed of hemipelvises with the same AHT fracture repaired with the Mouhsine et al. (2002) cable method (Fig. 4.7). This method required just one 2.0mm braided stainless steel cable (Accord, Model#71340008, Smith and Nephew Inc., Memphis, TN, USA) as originally described by Mouhsine et al. (2002) (Fig. 4.7). In this fixation method, the cable was passed posteriorly and inferiorly above the superior acetabular rim towards the LSN, where the cable turns around it going up through another clamp held on the quadrilateral plate, then curved down again to pass through the postero-superior end of the obturator foramen. The cable was then directed posteriorly and superiorly in the Groove for the Obturator Externus tendon until it reached the GSN, where it curved down on the quadrilateral plate again to run through the clamp held there and exit from its anterior end, crossing the rest of the quadrilateral plate, arcuate line until it reached a point just above the AIIS, where it turned around to run through the clamp at the beginning of the cable forming a closed loop. The exact same tensioning procedure was used for this cable as the one mentioned in “Mears Fixation” section. This cable provided reasonable fixation to both components of the complex AHT acetabular fracture as well as good support to the quadrilateral plate in a trial to prevent medialization of the acetabular cup under the influence of force generated during daily life activities.
Fig. 4.4: The “Plate” fixation method. A posterior column plate with an anterior column lag screw. Left: lateral view. Right: medial view. An acetabular cup was inserted.

Fig. 4.5: The “Mears” fixation method. A 2.0 mm braided cable with clamp was employed. Left: lateral view. Right: medial view. An acetabular cup was inserted.
Fig. 4.6: The “Kang” fixation method. Two 2.0 mm braided cables were utilized. Left: lateral view. Right: medial view. An acetabular cup was inserted.

Fig. 4.7: The “Mouhsine” fixation method. One 2.0 mm braided cable was used. Left: lateral view. Right: medial view. An acetabular cup was installed.
4.2.1.3 Mechanical Test Jig

For the proper fixation and alignment of the specimens, a unique setup was designed and manufactured. The setup provided proper anatomical orientation of the acetabulum during femoral head force application during normal daily physiological activities. The setup also provided anatomical fixation of the hemipelvis at the Sacro-iliac (SI) and Symphysis Pubis (SP) regions as would occur physiologically. To this end, two hollow steel cubes (3.46” wide X 3.46” long X 2.95” high) were employed. Two thru-holes were drilled at specific locations to bolt the cubes together at 30 degree angle to one another. One of the surfaces of the top cube provided a good support to the pelvis at the SI area by using three #10, 2” fully threaded stainless steel socket head cup screws with washers and fixed in place by one nut at the end of each screw. A steel block (1.25” long X 1” wide X 9” high) was manufactured and connected to one side of the bottom cube by two threaded screws into threaded holes. This side block had four threaded holes on top, where it supported the SP region by the use of 1¼” fully threaded stainless steel socket head cup screw (Fig. 4.8 and 4.9). The setup was rigidly mounted on a vise (4” wide X 1½” deep jaws) with an adjustable angle ranging from 0 to 90 degrees. The setup allowed anatomical orientation of the acetabulum, where both the ASIS and the front of the top of the SP were aligned in the same coronal plane producing a force vector of 45 degrees superomedially and 20 degrees posteriorly with the mechanical tester in the sagittal plane, which approximates the position of the femoral head in the standing position (Bergmann et al., 1993; Chang et al., 2001; Khajavi et al., 2010; Letournel and Judet, 1993; Shazar et al., 1998; Witte et al., 1997). To mount each specimen, four holes were drilled: one at the SP area using an 11/64” drill bit for a ¼” metal screw, and three at the SI region in a supero-inferior line using a 9/64 inch drill bit for #10 metal screws. Drilling the right holes for mounting each specimen on the setup was done after reaming, fracturing, reducing, and repairing the hemipelvises to prevent pre-stressing the bone or the repair constructs. To simulate the femoral component of a THR for a size G acetabular cup and liner, an aluminum alloy ball was used which had an outer diameter of 1.26” and a short neck (0.59” diameter X 0.39” long). The metal ball was mounted to the mechanical tester and load cell using a stainless steel rod (1” diameter X 3” length). The lower surface of the rod had an
indentation to perfectly accommodate the ball’s neck. The upper surface of the rod had a threaded ½” diameter X 1” deep hole connected to a similar hole at the

Fig. 4.8: Mechanical test setup consisting of 2 metal cubes bolted to each other, a side plate, and a spacer on the side. The front face of the upper cube was used for mounting the SI region and the top surface of the side block was used for mounting the SP region.

Fig. 4.9: The setup with an intact with cup specimen mounted on it. Left: Lateral view of the setup showing the SP mounting screw. Right: Anterior view of the setup showing the three SI mounting screws.
centre of an aluminum alloy rectangular plate (5.51” long X 3.54” wide X 0.5” thick) with a ½” diameter X 1¼” long stainless steel set screws. The plate was then attached to the load cell of the mechanical tester using four bolts.

4.2.1.4 Strain Gauges

Gauge Specifications

To accommodate the complex 3D specimen geometry, rosette strain gauges (Model# CEA-06-062UR-350, Vishay Measurements Group, Raleigh, NC, USA) were selected to be the most suitable (Vishay Measurements Group, 2010). These rosettes have an overall grid resistance of 350.0 Ω ±0.4%. Each rosette has 3 grids at 45° angles in respect to each other so that it would detect strains on the surface of an object in 2 perpendicular directions and at 45° in between (Fig. 4.10). Each grip has a certain gauge factor (GF) provided by the manufacturer, which was used later for accurate calculation of strains. The GF for grid 1, 2, and 3, respectively, were 2.085 ± 0.5%, 2.105 ± 0.5%, and 2.085 ± 0.5%, with an overall GF for all three grids of 2.10 ± 1.5%.

Gauge Locations

Gauges were physically mounted according to the manufacturer’s instructions to ensure consistent and accurate readings (Vishay Measurements Group, 2011). Anatomical markers were used as references for consistent gauge placement. Locations were chosen based on geometry and to cover all the three bone fragments, medial and lateral surfaces of the hemipelvis, above and below the transverse fracture line, and on either side of the anterior fracture line. For the Intact group, five rosettes were instrumented on each specimen at the following five locations, while for the other five groups, only four rosettes were used at locations 1-4 (Fig. 4.11).

**Location 1:** The lower left corner of the rosette gauge was mounted on the outer surface of the Ilium, at a distance of 87 mm from the ASIS and 86 mm from the PIIS. For all four
fixation groups because of the presence of the fracture, this location was positioned in reference to 2 points on the same bone fragment. Therefore, this location was 87 mm away from the ASIS and 50 mm from a hole provided by the manufacturer at the intermediate line of the Iliac Crest, just posterior to the tubercle on the external lip of Iliac crest.

**Location 2:** The right lower corner of the gauge was placed at a location on the Superior Pubic Ramus at a distance of 45 mm from the pubic eminence and 23 mm from the lower edge of the acetabulum. For reamed and fractured specimens, a 59 mm distance from a hole at the base of AIIS provided by the manufacture was used as a reference point.

**Location 3:** The right lower corner of the gauge was mounted at a location on the anterior portion of the Iliac Fossa, 25 mm postero-inferior to the ASIS and 36 mm postero-superior to the AIIS.

**Location 4:** The left lower corner of the gauge was placed at a location on the internal surface of the Ischium at 42 mm away from the tip of Ischial Spine and 17 mm from the postero-lower corner of the Obturator Foramen.

**Location 5:** The left lower corner of the gauge was placed at a location on the quadrilateral plate at a 30 mm distance from the tip of the Ischial Spine and 46 mm from the tip of the PIIS. This location was used only in the Intact specimens because it was disturbed by the various cable and plate configurations.

**Gauge Wiring**

After mounting the gauges, wiring of the six available electrodes per one rectangular rosette was performed (Fig. 4.10 and 4.11) (Vishay Measurements Group, 2012). A three-conductor, ¼ mm wide, tinned stranded copper flat Vinyl-insulated cable (330 DFV, Vishay Measurements Group, Raleigh, NC, USA) was employed to connect each grid to an eight-channel Cronos-PL data acquisition system (IMC Mess-Systeme GmbH, Berlin, Germany). The three conductors of the wire were color coded (Black, White, and Red). The conductors were separated from each other by 1 inch, and uncovered using a cable stripper. The black and white conductors were twisted together and connected to one electrode of the grid, while
the red one was connected to the other electrode. Using three conductors, instead of only two, notably increased the strain gauge measurement sensitivity and decreased lead-wire temperature fluctuations (Vishay Measurements Group, 2010). The other end of each cable was then connected to two alligator clips (Nexxttech, 60cm Mini Alligator Clips, The Source, Canada), which were in turn connected to one channel of the two-channel DSub-15-pin connector (Acc/DSub-Uni2, IMC Mess-Systeme GmbH, Berlin, Germany). The DSub connector was subsequently connected to the Cronos system through a Uni2-8 amplifier (Fig. 4.10 and 4.11). The quarter bridge Wheatstone configuration was followed. Therefore, the four gauges with 12 grids required six DSubs i.e. 12 channels; however, only eight channels were available for the Cronos system. Thus, two rosettes were tested together at a time for three repetitive experimental tests, followed by the other two rosettes to complete testing each specimen.

Fig. 4.10: A) A DSub showing its internal electric circuit and connections, B) The eight-channel amplifier of the Cronos-PL system, C) A rectangular rosette strain gauge demonstrating its three grids numbered 1, 2, and 3 by the manufacturer.
Fig. 4.11: A) Lateral view of the hemipelvis showing SG1 and 2, B) Medial view of the hemipelvis illustrating SG3, 4, and 5, C) The Cronos connected to the DSubs, which in turn are connected to alligator clips. These clips were connected to the wires coming out of the SG. SG= Strain Gauge.

**Gauge Data Acquisition**

The Cronos system was connected through a LAN to an Acer TravelMate 15.6” Laptop (Intel Core i5-3320M / 500GB HDD / 8GB RAM /Windows7 Professional), where a data acquisition program (imcDevices v2.6) was installed for data acquisition and storage. All the manufacturer’s parameters, e.g. gauge factor for each grid, were inputted into this program. Another laptop with a Famos V5.0 software (IMC Mess-Systeme GmbH, Berlin, Germany) was used for data analysis and to provide the final average strain value for each grid ($\varepsilon_1$, $\varepsilon_2$, $\varepsilon_3$) during the middle 30s period of data acquisition. All experimental tests were repeated three times. The maximum principal, minimum principal, and Von Mises strains and stresses for each strain gauge were calculated for each test, and then averaged for all three tests. Using the following equations, employed by previous biomechanical studies (Davis et al., 2008; Pal et al., 2010; Papini et al., 2010), the maximum ($\varepsilon_P$) and minimum ($\varepsilon_Q$) principal strains for each strain gauge were obtained:
\[ \varepsilon_{P,Q} = \frac{\varepsilon_1 + \varepsilon_3}{2} \pm \frac{1}{\sqrt{2}} \sqrt{(\varepsilon_1 - \varepsilon_2)^2 + (\varepsilon_2 - \varepsilon_3)^2} \]

Then, the Von Mises (Equivalent) strain (\(\varepsilon_{\text{v}}\)) was calculated using the following equation:

\[ \varepsilon_{\text{v}} = \sqrt{\varepsilon_P^2 - \varepsilon_P \varepsilon_Q + \varepsilon_Q^2} \]

Given the Young’s Elastic Modulus (E) of the cortical bone = 16.7 GPa, and the Poisson’s Ratio (\(\nu\)) = 0.26, the maximum (\(\sigma_P\)) and minimum (\(\sigma_Q\)) principal stresses were calculated, as shown in the equation below:

\[ \sigma_{P,Q} = \frac{E}{2} \left( \frac{\varepsilon_1 + \varepsilon_3}{1 - \nu} \pm \frac{\sqrt{2}}{1 + \nu} \sqrt{(\varepsilon_1 - \varepsilon_2)^2 + (\varepsilon_2 - \varepsilon_3)^2} \right) \]

Von Mises stress (\(\sigma_{\text{v}}\)) was then obtained from the following equation:

\[ \sigma_{\text{v}} = \sqrt{\sigma_P^2 - \sigma_P \sigma_Q + \sigma_Q^2} \]

### 4.2.1.5 Mechanical Tests

All mechanical tests were performed on a computer controlled Instron machine (Instron 8874, Instron, Norwood, MA, USA) (Fig. 4.12), which had a capacity of ±25 kN, a resolution of 0.1 N, and an accuracy of ± 0.5%. After each specimen was mounted on the setup, an angle of 45° superomedially and 20° posteriorly in the sagittal plane was generated between the acetabulum/acetabular cup and the rod with the metal ball, representing the femoral component of the THR, as described above. This orientation simulated the position of the femoral neck in the standing position, and hence, the force vector applied during regular daily activities (Bergmann et al., 1993; Chang et al., 2001; Khajavi et al., 2010; Letournel and Judet, 1993; Shazar et al., 1998; Witte et al., 1997). Distally, the angle vise
was mounted on the base of the Instron machine using two metal bars and bolts. Proximally, the metal ball fit perfectly into the acetabular liner, which was free to rotate inside the liner. A 50 N axial preload was applied to the acetabular cup to overcome any mechanical slack, i.e. hysteresis. An axial load of 200 N was applied using load control at 37.5 N/s, the load was sustained for 90 s, and the load was then released at 50 N/s. Strain gauge readings were continuously recorded during the whole 90 s test period, however, only the middle 30 s “steady state” interval was analyzed to avoid any load rampup or rampdown effects. Each test was repeated 3 times for each strain gauge. Testing of individual hemipelvises was performed randomly between groups to avoid any order effects or bias.

Fig. 4.12: A Plate specimen mounted on the Instron. Left: Anterior view, right: lateral view.
4.2.2 Finite Element Analysis (FEA)

4.2.2.1 CAD Models

Hemipelvis

The 3D CAD files in SolidWorks format (SolidWorks Corp., Dassault Systèmes, Concord, MA, USA) for the same left, large, fourth generation, composite hemipelvis described earlier were supplied by the manufacturer (Model #3405, Sawbones, Vashon, WA, USA) (Fig. 4.13). The geometry consisted of two bodies, i.e. a solid cortex and a solid cancellous core, which was oriented in the anatomical position. Each body consisted of a certain number of faces. Before any processing, the cortex and cancellous core were originally made of 581 and 294 faces, respectively.

Implants

The femoral metal ball, side block, steel cube, and fixation screws for the setup were designed using SolidWorks 2014 x 64 software based on measurements of the physical setup obtained using a digital caliper with a resolution of 0.01 mm. The size G multi-hole acetabular shell and polyethylene liner (Smith and Nephew Inc., Memphis, TN, USA) were modeled according to the parameters provided by the manufacturer plus digital caliper measurements. Plates were designed to conform to the complex surface of the posterior column of the hemipelvis with dimension obtained using the digital caliper. Employing the “Spline” and “3 point Arc” features in SolidWorks, the curved plate was sketched to simulate the exact plate course. Then, using the “Sweep” function, the plate was generated and refined using the “Extruded Cut”, “Fillet”, and “Curve Driven Pattern” functions. Also, all cup screws, plate screws, and lag screws were designed according to manufacturer specifications plus digital caliper measurements for the diameters and shapes of screw heads. All screw threads were neglected because a previous study showed no significant difference for threaded vs. non-threaded screws if the contact at the bone/screw interface is assigned a “Bonded” contact, as done currently (Zdero et al., 2008a). For cables, a 3D drawing was
utilized to allow the precise replication of the path of the cable. Many points were added on the cortex along the cable’s path, then the “Spline” utility was used to connect these points together, creating a closed loop that simulated the exact orientation of the cable. The “Sweep” utility was then employed to create the 3D cable body. The points along the path were occasionally used to refine the orientation of the cable.

### 4.2.2.2 Component Assembly

For each of the six FE groups, the relevant parts were assembled using SolidWorks 2014 x 64 Edition Assembly (SolidWorks Corp., Dassault Systèmes, Concord, MA, USA).

For the Intact group (Fig. 4.14), the four fixation screws at the SI and SP regions were passed through cortical and cancellous bodies with the same orientation used for the experimental setup. The steel cube and side block were placed and oriented properly with respect to the pelvis, thereby touching the cortex to support the bone in the SI and SP areas, as done experimentally. The femoral metal ball was placed in contact with the acetabulum. A pressure-sensitive Fujifilm was inserted between the ball and the acetabulum to determine the exact shape and size of the contact area. The ball was oriented at 45° superomedially and 20° posteriorly in the sagittal plane to simulate the anatomical position of the femoral neck in the standing position, and therefore, apply physiological loads along the correct force vector yielded in human hips during routine daily activities.

For the Cup group (Fig. 4.14), a hemi-spherical body with a diameter of 59 mm was introduced into the acetabulum to simulate the largest size of the used reamers. Another 59 mm circular body was introduced on top of the hemispherical body to overlap with any protruding bone along the acetabular rim. Also, the acetabular shell and liner were put in place in relation to the acetabulum. Four cup screws were inserted: two in the anterior column, one in the posterior column, and the last one in the Ischium, replicating the correct surgical placement of these screws. The model was eventually shown to an orthopaedic surgeon for approval.
Finally, the four fixation groups (Fig. 4.15 and 4.16) were then assembled based on the Cup group with the addition of the corresponding fixation implants.

4.2.2.3 Assembly Processing

The six previous assemblies for the six FE models were saved as “STEP AP214” files, which were imported and read by the latest version of ANSYS Workbench R15.0 DesignModeler (ANSYS Inc., Canonsburg, PA, USA). The latter was used for further processing of the six assembly models as follows:

For the Intact group (Fig. 4.14), because the rectangular sketches resembling strain gauges were projected onto the cortex, each drawing did not form a separate entity. Therefore, the “Merge” function was used to merge the involved faces together forming that rectangular new face with the exact same location and size as experimental strain gauges. This task was performed for each of the strain gauges for each of the six models. Also, necessary Boolean operations were performed between the assembled bodies such as between the cortex and core on one hand and the fixation screws and femoral ball on the other hand.

For the Cup groups (Fig. 4.14), the previous tasks were performed along with the “Delete – Body” function, which was used to delete the hemispherical and circular bodies used to simulate reaming of the acetabulum.

For the four fixation groups (Fig. 4.15 and 4.16), the above functions were used, plus planes were created along which the “Slice” feature was utilized to create the AHT fracture. This fracture pattern typically matched the one created in experimental specimens.

4.2.2.4 Boundary Conditions

For the Intact model, the fixation screws at both SI and SP regions were “bonded” with no slipping at all cortical/cancellous bones, steel cube, and side block because screws were tightly threaded through the bone and nuts were used at the SI region, while threaded into the side block. The contact between the side block and steel cube to the cortex was assigned to
be “no separation” allowing slight sliding or micro-motion along that interface. The cortex and cancellous core were “bonded” together. Last, the metal ball was in “no separation” with the acetabulum allowing minimal sliding during axial loading, however, without any gapping.

For the Cup model, all contacts assigned before were employed. In addition, all bone/screw interfaces were “bonded” to simulate the strong screw bites over the long distance travelled into the bone, especially at the cortices. The interface between the bone/cup and the cup/screws were assigned “no separation” contact allowing some micro-motion as the screws were not locked into the cup, but just compression was applied at their interface. Cup/liner contact was “bonded” because tabs on the outer surface of the liner were fully seated into the indentions on the inner edge of the cup with no degree of freedom. Finally, the liner/ball interface was assigned “no separation” permitting minimal sliding of the smooth surface of the ball over the very smooth polished inner surface of the liner.

For the four fixation models, all the previous contacts were re-applied. Additionally, plate screws/bone, lag screw/bone, and cable/bone interfaces were “bonded” to replicate, respectively, material interdigitation around screw threads, and the notable compression of cables on bone, which was measured at 100 kg of tension force and the textured contact between braided cables and bone. However, the plate/bone and plate/plate screws interfaces were assigned “no separation” because these contacts mainly depend on compression of the screws on the plate or of the plate on the bone. Contact between cortical and cancellous bone fragments across the fracture were assigned “no separation” to permit sliding with two degrees-of-freedom without separating, simulating cable compression all around the bone, or the plate and lag screw pulling the two bone fragments together.

Finally, “fixed support” was applied to two of the setup screw holes of both the steel cube and side plate, which replicated the exact configuration of the experimental setup. Force was applied to the flat top surface of the metal ball to simulate the long rod connected to the load cell of the Instron machine in experiments. The force was applied along a vector of 45° superomedially and 20° posteriorly in respect to the sagittal plane, simulating the real situation performed in experimental testing.
Fig. 4.13: A 3D CAD Model of the hemipelvis. Left: lateral view, right: medial view.

Fig. 4.14: Assembly of component parts in the same orientation as experiments. Left: Intact group, right: Cup group.
Fig. 4.15: Assembly of component parts oriented exactly as experiments. Left: Plate group, right: Mears group.

Fig. 4.16: Assembly of component parts mounted on the setup and oriented as in experiments. Left: Kang group, right: Mouhsine group.
4.2.2.5 Material properties

All material properties are summarized (Table 4.1). In the ANSYS Workbench R15.0 “Engineering Data” section of the “Static Structural” analysis system, the material properties for each item were incorporated into the system to replicate the experimental conditions exactly. Cortical and cancellous bone properties were supplied by the manufacturer (Sawbones, Vashon, WA, USA) and were assumed to be homogeneous, isotropic, and linearly elastic. For all other components used as implants and/or for the test jig, standard values were obtained from the ANSYS Workbench R15.0. Specifically, the acetabular cup was defined as titanium alloy. Test jig screws, repair plates, repair plate screws, lag screws, and cables were stainless steel. The liner for the acetabular cup was made of UHMWPE. The femoral metal ball was assigned to be aluminum alloy. The steel cube and side block were assigned as structural steel.

Table 4.1: Mechanical properties of the materials assigned in the six FE models.

<table>
<thead>
<tr>
<th>Materials</th>
<th>Young’s Modulus (GPa)</th>
<th>Poisson’s Ratio</th>
<th>Tensile Strength (MPa)</th>
<th>Compressive Strength (MPa)</th>
<th>Density (g/cc)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical</td>
<td>16.7</td>
<td>0.26</td>
<td>106</td>
<td>157</td>
<td>1.64</td>
</tr>
<tr>
<td>Cancellous</td>
<td>0.155</td>
<td>0.30</td>
<td>6</td>
<td>6</td>
<td>0.16</td>
</tr>
<tr>
<td>Titanium Alloy</td>
<td>96</td>
<td>0.36</td>
<td>1070</td>
<td>1070</td>
<td>4.62</td>
</tr>
<tr>
<td>Stainless steel</td>
<td>193</td>
<td>0.31</td>
<td>586</td>
<td>586</td>
<td>7.75</td>
</tr>
<tr>
<td>UHMWPE</td>
<td>0.9</td>
<td>0.4</td>
<td>460</td>
<td>460</td>
<td>0.9392</td>
</tr>
<tr>
<td>Aluminum Alloy</td>
<td>71</td>
<td>0.33</td>
<td>310</td>
<td>310</td>
<td>2.77</td>
</tr>
<tr>
<td>Structural Steel</td>
<td>200</td>
<td>0.3</td>
<td>460</td>
<td>460</td>
<td>7.85</td>
</tr>
</tbody>
</table>

4.2.2.6 Meshing

Using the ANSYS Workbench R15.0 “Mechanical”, a mesh was generated over the complex geometry of both cortical and cancellous bodies, as well as the acetabular cup, all the fixation implants, and test jig. A fine tetrahedron patch-conforming mesh was used for all structures, with element sizes ranging from 2-4 mm depending on areas of interest (Fig. 4.17). A mesh
“relevance” of 80% was utilized to generate a fine mesh. Usually, mesh “relevance” utility ranges from 0-100%, where 0% indicates the coarsest mesh, while 100% indicates the finest mesh for the same assigned element sizes. A mesh relevance of 80% was selected because preliminary relevance tests were performed on the Cup model and showed good “Force Convergence” and “Solver Output” results with a change of less than 1% vs. the maximum mesh relevance of 100%. The overall number of nodes and elements for each of the six models were as shown (Table 4.2).

Table 4.2: Number of nodes and elements for each mesh per each of the six FE models.

<table>
<thead>
<tr>
<th>Model</th>
<th>Nodes</th>
<th>Elements</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intact</td>
<td>275394</td>
<td>172329</td>
</tr>
<tr>
<td>Cup</td>
<td>358501</td>
<td>225069</td>
</tr>
<tr>
<td>Plate</td>
<td>403488</td>
<td>249385</td>
</tr>
<tr>
<td>Mears</td>
<td>375093</td>
<td>233227</td>
</tr>
<tr>
<td>Kang</td>
<td>425391</td>
<td>265944</td>
</tr>
<tr>
<td>Mouhsine</td>
<td>381146</td>
<td>235436</td>
</tr>
</tbody>
</table>

Frequently, the “Sizing” feature was used to determine the element sized for certain faces or bodies to enable meshing of the overall body or model. Also, the “Sphere of Influence” utility was used for both cortical and cancellous bodies at the acetabular region to allow for proper force transmission and calculations between the acetabular cup, femoral ball, and hemipelvis. The “Sphere Radius” was determined to be 50 mm, where the centre of the sphere was almost the centre of the acetabulum. The “element size” for the sphere was 3 mm.

4.2.2.7 FE Analysis

Using the ANSYS Workbench R15.0 “Mechanical” feature, the FE models were analyzed to obtain stress and strain maps for all six models. First, a subclinical load of 200 N was applied through the femoral metal ball to validate the FE model against the maximum Von Mises stress values obtained from experiments. Second, then a clinical-level 2207 N compressive load was applied to simulate 3 x body weight for a 75 kg person. The resulting stress maps were adjusted by refining the meshes at areas of peak bone stress employing the “Sphere of
Influence” utility using a sphere diameter of 15 mm. This compensated for any artifacts in stresses due to geometric complexities or discontinuities. Finally, overall specimen stiffness was calculated by dividing the applied force over the resultant displacement in the same vertical direction of the force i.e. X axis at the tip of the metal ball.

![Meshing of the FE models. A Plate model was presented here as an example. Left: showing the whole mesh, right: a magnified view of the acetabular cup, acetabulum, plate and screws, as well as setup device.](image)

**Fig. 4.17:** Meshing of the FE models. A Plate model was presented here as an example. Left: showing the whole mesh, right: a magnified view of the acetabular cup, acetabulum, plate and screws, as well as setup device.

### 4.3 Phase II – Destructive Mechanical Testing

#### 4.3.1 Selection of the Two Best Repair Methods

Based on Phase I results, two out of the four methods for fracture fixation were chosen, namely the posterior column reconstruction plate with an anterior column lag screw and the Mouhsine et al. cable fixation method. The Plate method was selected because it is the “gold standard” and most widely used repair method by most orthopaedic surgeons specialized in fixing acetabular fractures. The Mouhsine et al. cable method was chosen because it provided the lowest “peak bone stress”, which means it would be the last cable method to experience structural bone failure (i.e. ultimate tensile strength = 106 MPa) if a compressive hip joint
load were applied indefinitely to all cable methods. 10 specimens were assigned per group, which was thought to be enough to detect all statistical differences between both groups based on expert’s opinion and previous biomechanical studies employing synthetic pelvises (Gililland et al., 2013; Khajavi et al., 2010; Shazar et al., 1998).

4.3.2 Marker Placement

Markers were placed on the specimen to detect the overall displacement occurring between fracture fragments along the fracture line during the entire MF event. The displacements measured were parallel “sliding” of the bone fragments along the fracture line and perpendicular “gapping” separation of the two fragments away from each other. Due to the complex geometry and potential complex movement of the fragments, relative “rotation” of fragments was not considered. Specifically, four circular coloured sticky markers of 6 mm diameter (Avery® Round Assorted Colour-Coding Labels ¼”, Staples, Canada) were placed along the posterior hemitransverse fracture line (Fig. 4.18 and 4.19). Bright green and yellow markers were employed, so that they would easily be contrasted against the brown-coloured background of the hemipelvis. Because the smallest commercially available markers were relatively large, i.e. 6 mm in diameter, a small black dot in the centre of each label was drawn using a black permanent-ink pen. Two markers were placed on each side of the fracture, all of which were 11 mm apart at the corner positions. Transparent Scotch® tape was applied overtop of the markers to keep them in place. A similar approach was applied in other biomechanical studies on acetabular fractures (Khajavi et al., 2010; Shazar et al., 1998).
Fig. 4.18: Three pictures extracted from video records of the Plate group showing the position of the markers along the posterior hemitransverse fracture at the starting position without any displacement (left), at clinical failure point (middle), and at mechanical failure point (right).

Fig. 4.19: Three pictures of the Mouhsine group extracted from video records, illustrating the position of the markers along the transverse fracture at the starting position without any displacement (left), at clinical failure point (middle), and at mechanical failure point (right).

4.3.3 Mechanical testing

The mechanical tester (Model# 8874, Instron, Norwood, MA, USA) used for Phase I was also employed for Phase II. The setup with the specimen mounted on it was fixed rigidly to the base of the Instron machine using two metal bars with two bolts for each bar. Specimens were tested in an alternative order, e.g. starting with a Plate specimen, then two Mouhsine specimens, then two Plate specimens, then two Mouhsine specimens, etc., finally ending with
a Plate specimen. This order was followed to avoid order effects that occur when one group is tested completely first followed by the other group, and to avoid a specimen from one group always being tested after a specimen from the other group. Specifically, each specimen was non-destructively “pre-conditioned” for three cycles before being mechanically tested to failure, exactly as in Phase I. Finally, a destructive test was done using “displacement control” at a rate of 10 mm/min for a maximum displacement of 25 mm to avoid accidental impaction of the specimens against the baseplate of the Instron. To record the displacement of the coloured markers placed along the transverse fracture line, the entire failure testing period was digitally recorded for each specimen using a Sony Cyber-Shot 12.1 megapixel 5X optical zoom camera with a 30 frame/second recording frequency (Model #DSC-W290, Digital Still Camera, Sony Corp.) (Fig. 4.18 and 4.19).

4.3.4 Data Analysis

Stiffness was defined as the slope of the linear portion of the force-vs-displacement curve for the 0 to 2207 N range, i.e. up to 3 x body weight for a 75 kg person. Failure force was defined as the force at which the construct failed, such that “clinical failure” (CF) was considered to occur at the clinically realistic value of 5 mm of displacement, whilst “mechanical failure” (MF) was the absolute peak force for complete structural failure. Failure displacement was identified as the displacement corresponding to the force during the MF event. Failure energy was defined as the total area under the force-vs-displacement curve from the start of the test until the failure force point was reached for both the “clinical failure” and “mechanical failure” events. Note that no previous studies in the literature defined “clinical failure” for acetabular or pelvic fracture repair, but prior investigators have used a 10 mm displacement criterion as “clinical failure” for femoral head experiments (McConnell et al., 2008; Zdero et al., 2008b). Consequently, it was felt that patients would likely present with complaints of pain and impaired function long before 10 mm of displacement was achieved, thus 5 mm was chosen as a conservative value for CF. Additionally, the videos showing marker displacement were analyzed at three points, namely the starting point, the CF point, and the MF point for sliding and gapping. The four markers in each specimen were divided into two anterior markers (AM) across the fracture line closer
to the acetabular rim, and two posterior markers (PM) across the fracture closer to the GSN. Displacements were measured at both levels: AM and PM at both CF and MF points of the test for each of sliding and gapping in order to detect the progression and trend of displacement taking place across the transverse component of AHT fracture e.g. sliding displacements happening between the AM at CF for the Plate group.

4.3.5 Statistical Analysis

Student’s t-tests were done for stiffness, as well as failure displacement during MF events, since there was only one possible paired comparison, i.e. Plate vs. Mouhsine. Results were first assessed using the F-test (two sample for variances) to determine if the variances between the two groups were different, i.e. a two-tailed P-value < 0.05, and hence, decide which specific t-test to employ. Conversely, one-way Analyses of Variance (ANOVA) were done for failure force, failure energy, sliding, and gapping comparisons, where all data from all groups were grouped into a single data array using IBM SPSS Statistics V21 software (SPSS Inc., Chicago, IL, USA). This was done since there were multiple possible intra- and inter-group comparisons, as data were collected for both CF and MF events. When ANOVA yielded P ≥ 0.05, this value was reported since no statistical difference was detected. However, when ANOVA yielded P<0.05, a statistical significance was noted somewhere. When the compared groups had the same number of data points, the Tukey’s Honestly Significant Difference (HSD) post hoc analysis was run to determine which specific pairwise comparison caused the statistical difference; however, when compared groups had a different number of data points, the Tamhane’s T2 post hoc analysis was employed. After all analysis was done, a two-tailed post hoc power analysis between each pairwise comparison for all the above-mentioned statistical tests was performed at α=0.05 in order to identify if the number of specimens for both Plate and Mouhsine groups was enough to detect all statistical differences that may have been present, i.e. avoid Type II error. Finally, irrelevant pairwise statistical comparisons were ignored for this analysis and are not reported, e.g. failure force at CF for the Plate group vs. failure force at MF for the Mouhsine group.
Chapter 5
Results

5.1 Phase I: Non-Destructive Analysis of Hemipelvises

5.1.1 Experimental Validation of FEA

For the non-destructive experiments at 200 N, results of the overall average experimental results for each strain gauge per each group are summarized (Table 5.1). The corresponding FEA results for Von Mises stress values are also provided (Table 5.2).

Table 5.1: Experimental Von Mises Stress Values in MPa. SG = strain gauge.

<table>
<thead>
<tr>
<th>Strain Gauge</th>
<th>Intact</th>
<th>Cup</th>
<th>Plate</th>
<th>Mears</th>
<th>Kang</th>
<th>Mouhsine</th>
</tr>
</thead>
<tbody>
<tr>
<td>SG1</td>
<td>0.279</td>
<td>0.250</td>
<td>0.333</td>
<td>0.179</td>
<td>0.243</td>
<td>0.374</td>
</tr>
<tr>
<td>SG2</td>
<td>1.738</td>
<td>0.826</td>
<td>2.942</td>
<td>0.857</td>
<td>1.503</td>
<td>0.931</td>
</tr>
<tr>
<td>SG3</td>
<td>0.291</td>
<td>0.441</td>
<td>0.197</td>
<td>0.385</td>
<td>0.415</td>
<td>0.361</td>
</tr>
<tr>
<td>SG4</td>
<td>0.451</td>
<td>0.284</td>
<td>0.129</td>
<td>0.205</td>
<td>0.159</td>
<td>0.411</td>
</tr>
<tr>
<td>SG5</td>
<td>0.953</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 5.2: FEA Von Mises Stress Values in MPa at each Strain Gauge Location. SG = strain gauge.

<table>
<thead>
<tr>
<th>Strain Gauge Location</th>
<th>Intact</th>
<th>Cup</th>
<th>Plate</th>
<th>Mears</th>
<th>Kang</th>
<th>Mouhsine</th>
</tr>
</thead>
<tbody>
<tr>
<td>SG1</td>
<td>0.151</td>
<td>0.211</td>
<td>0.296</td>
<td>0.308</td>
<td>0.231</td>
<td>0.306</td>
</tr>
<tr>
<td>SG2</td>
<td>1.154</td>
<td>1.321</td>
<td>1.306</td>
<td>1.527</td>
<td>1.437</td>
<td>1.439</td>
</tr>
<tr>
<td>SG3</td>
<td>0.110</td>
<td>0.109</td>
<td>0.032</td>
<td>0.079</td>
<td>0.059</td>
<td>0.078</td>
</tr>
<tr>
<td>SG4</td>
<td>0.519</td>
<td>0.288</td>
<td>0.069</td>
<td>0.081</td>
<td>0.066</td>
<td>0.054</td>
</tr>
<tr>
<td>SG5</td>
<td>0.437</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

All data points for Von Mises stress obtained from experiments and FEA were plotted against each other to detect the correlation, since they FE models are essentially variations of one and the same model (Fig. 5.1). This approach was adopted by many biomechanical
studies in the literature (Bougherara et al., 2010b, 2011b; Dubov et al., 2011; Ebrahimi et al., 2012; Shah et al., 2011).

One data point was excluded from the comparison, which was for SG2 of the Plate group, and therefore, the corresponding point of the FEA as well. This point was excluded because the anterior column lag screw was inserted directly under the cortex of the anterior column just under the surface strain gauge due to the very narrow cross section of the reamed acetabular anterior column and the relatively large diameter screw i.e. 4.0mm. The screw was clearly seen through the cortical shell by the naked eye by just looking at the specimen, and hence, the gauge wrongly read higher stress values because of the close proximity of the anterior column screw to the surface strain gauge. The lowest value for experimental Von Mises stress was 0.129 MPa, and the highest was 1.738 MPa, while for the FEA, the lowest was 0.032 MPa and the highest 1.527 MPa. The graph (Fig. 5.1) showed a good Pearson Correlation Coefficient (R=0.82), with a Slope of 1. These parameters are close to the ideal ones, which should be R=1 and Slope=1. Thus the FE model was valid, despite some expected data scatter which is typical of biomechanical experimentation.
Fig. 5.1: FEA Von Mises Stress Validation Graph. All data points, except one, from all six experimental and FEA groups were plotted together since strain gauge positions were exactly the same. Ideal correlation would result in Slope=1 and R=1.

5.1.2 FEA Stress Maps at 3 x Body Weight

5.1.2.1 Bone Stress Maps

After validation of the FE model, load levels were extended to include 2207 N, namely, 3 times body weight for a 75 kg person. For the Intact group (Fig. 5.3), the maximum value was 195 MPa, and located at the posterio-superior corner of the acetabulum, mainly in line with the direction of force application. For the Cup group (Fig. 5.3), the maximum Von Mises stress value reached 114 MPa at the lower end of SI region. For the Plate group (Fig. 5.4), the maximum was 255 MPa, positioned at the lower anterior end of the curved articular surface i.e. Lunate Surface of the acetabulum. For the Mears group (Fig. 5.4), a peak value of 205 MPa was achieved at the corner of the meeting of both fractures, namely anterior and posterior hemitransverse. For the Kang group (Fig. 5.5), the maximum was 250 MPa, at the
same position as the Plate group. For the Mouhsine group (Fig. 5.5), the peak value reached 181MPa and located close to the corner of both fractures as in Mears, along the cable at the meeting of both the transverse fracture with the ILiopectineal line. A graph for the Von Mises stress results of the six FE models is also provided (Fig. 5.2).

![Graph showing FEA Peak Bone Von Mises Stress in MPa for all Six FE Groups.](image)

**Fig. 5.2: FEA Peak Bone Von Mises Stress in MPa for all Six FE Groups.**
Fig. 5.3: FEA Von Mises Stress Maps at three times body weight of an average 75Kg person i.e. Force= 2207N. Peak bone stress values and locations are demonstrated in the pictures. Left = Intact Model, Right = Cup Model.

Fig. 5.4: FEA Von Mises Stress Maps at three times body weight of an average 75Kg person i.e. Force= 2207N. Peak bone stress values and locations are demonstrated in the pictures. Left = Plate Model, Right = Mears Model.
Fig. 5.5: FEA Von Mises Stress Maps at three times body weight of an average 75Kg person i.e. Force= 2207N. Peak bone stress values and locations are demonstrated in the pictures. Left = Kang Model, Right = Mouhsine Model.

5.1.2.2 Stress Maps for Acetabular Cups and Implants

Additionally, 3D stress maps were calculated for all the implants included in the five groups, excluding the Intact group as it did not have any implants. For the Cup group, the peak stress reached 262MPa and occurred at the contact between the Ischial screw of the cup and the orifice of the acetabular cup. For the Plate group, the peak stress of all the implants, namely the plate, plate screws, lag screw, acetabular cup and cup screws, was 427MPa and happened at the superior edge of the lower second hole, where the plate is bent to conform the contour of the posterior column of the acetabulum changing direction towards the Ischium. Also, at that point, the plate is squeezed by the head of the plate screw in that hole. However, the maximum cup stress achieved 235MPa, at the edge of one of the two anterior column screws with the corresponding hole of the cup. For the Mears group, the peak stress was 409MPa and was attained at the intersection of the cable with the iliopectineal line at the meeting of both fracture lines, while the maximum cup stress was 314MPa at the contact between the posterior column screw with the corresponding cup hole. For the Kang group, the maximum stress was 270MPa at the posterior edge of the cable going around the GSN, while for the
cup, the maximum was 205MPa at the area of contact between one of the two anterior column screws with the edge of the corresponding cup hole. This location is the same as for the cup of the Plate group. For Mouhsine group, the peak stress was 373MPa, at the intersection of the cable with the Iliopectineal line at the upper edge of the transverse fracture close to the meeting of both fractures, while for the cup, the maximum stress reached was 336MPa at the edge of the posterior column screw with the cup hole, same location as in the Mears group.

5.1.3 FEA Stiffnesses at 3 x body weight

The stiffness for each construct was calculated, which equals the slope of the force-vs-displacement curve, and at the same time equals force divided by displacement at any force value because the FEA assumed absolute linearity of all the models. The FEA displacements, stiffnesses and percentages of increase in stiffness in respect to the Intact group are presented in Table 5.3. A graph for FEA stiffness of all six models is also provided (Fig. 5.6).

Table 5.3: FEA Displacement, Stiffness, and Percentage Increase in Stiffness in Respect to the Intact Group at a Force of 2207N i.e. 3 x body weight.

<table>
<thead>
<tr>
<th></th>
<th>Intact</th>
<th>Cup</th>
<th>Plate</th>
<th>Mears</th>
<th>Kang</th>
<th>Mouhsine</th>
</tr>
</thead>
<tbody>
<tr>
<td>Displacement</td>
<td>0.471</td>
<td>0.271</td>
<td>0.311</td>
<td>0.390</td>
<td>0.274</td>
<td>0.394</td>
</tr>
<tr>
<td>Stiffness</td>
<td>4691</td>
<td>8156</td>
<td>7099</td>
<td>5661</td>
<td>8062</td>
<td>5607</td>
</tr>
<tr>
<td>Percentage of stiffness increase</td>
<td>0</td>
<td>74</td>
<td>51</td>
<td>21</td>
<td>72</td>
<td>20</td>
</tr>
</tbody>
</table>
5.2 Phase II: Destructive Analysis of Hemipelvises

5.2.1 Raw Data

Force-versus-displacement graphs obtained from the raw data for a representative specimen of each of the Plate and Mouhsine groups are presented below (Fig. 5.7 and 5.8). Each graph illustrates both points of clinical failure (CF) and mechanical failure (MF). It also demonstrates the elastic versus plastic regions for both groups, indicating that the Mouhsine construct is more ductile than the Plate group.
Fig. 5.7: A raw data graph for one specimen representative of the Plate group showing the points of MF and CF as well as elastic and plastic regions of the force-versus-displacement curve. CF= Clinical Failure, defined as absolute 5mm of axial displacement by the testing machine. MF= Mechanical Failure, defined as the point of maximum force peak generated in the same curve, after which a major deflection occurred.

Fig. 5.8: A raw data graph for one specimen representative of the Mouhsine group showing the points of MF and CF as well as elastic and plastic regions of the force-vs-displacement curve. CF= Clinical Failure, defined as absolute 5mm of axial displacement by the testing machine. MF= Mechanical Failure, defined as the point of maximum force peak generated in the same curve, after which a major deflection occurred.
5.2.2 Stiffness

For stiffness, to ensure linearity during the non-destructive aspect of the test, $R^2$ were calculated for all the slopes of the force vs displacement graphs, i.e. stiffness, both the Plate and Mouhsine groups and were of 0.986 and 0.987, respectively. The stiffness values are graphically compared showing the superiority of the plate group ($P= 0.001$) (Fig. 5.9). Stiffness for the Plate group was 72% higher than the other group.

![PHASE II - STIFFNESS](image)

**Fig. 5.9**: Experimental Stiffness of Both the Plate and Mouhsine Groups, Estimated from the Linear Portion of the Force-vs-Displacement Graph (between 0-2207N). Error bars are one standard deviation.

5.2.3 Failure Displacement

The overall distance each specimen travelled from the starting point till it mechanically failed was called “Failure Displacement”. For failure displacement at mechanical failure (MF), the results showed that the Mouhsine group was able to accommodate more motion than the Plate group ($P <0.005$) (Fig. 5.10). Failure displacement for the Plate group was 55% of that of the Mouhsine group.
5.2.4 Failure Force

For failure force, there was a statistical difference for the Plate group during CF vs MF events (P= 0.000), for the Mouhsine group during CF vs MF events (P= 0.000), and also during the CF event for Plate vs Mouhsine groups (P= 0.009) (Fig. 5.11). No statistical difference was found between MF for Plate vs Mouhsine groups (P=0.798). P-value for statistically different comparisons ranged between .000-.009. For the Plate group, the failure force at CF represented 67.5% of that at MF. For the Mouhsine group, the CF failure force attained 46.4% of the MF one.
Fig. 5.11: A Comparison of Phase II Failure Force between Both Plate and Mouhsine Groups at MF and CF. Symbols (#, *, □) indicate statistical differences (P< 0.05) between both the Plate and Mouhsine groups. No other statistical differences were detected (P=0.798). Error bars are one standard deviation.

5.2.5 Failure Energy

For failure energy, comparisons between failure energy at CF vs MF events for each of the Plate and Mouhsine groups were statistically different, with respective P-values of 0.022 and 0.000 (Fig. 5.12). However, the comparisons for either CF or MF between both groups were not significant with P-values of 0.991 and 0.055 respectively. The CF/MF for the Plate group was 24.2%, while for the Mouhsine group was 10.2%.
Fig. 5.12: A Phase II – Failure Energy Graph, Illustrating the Four Different Groups of both the Plate and Mouhsine Groups at Both Clinical and Mechanical Failures. Symbols (#, *) indicate statistical differences (P< 0.05) between both the Plate and Mouhsine groups. No other statistical differences were detected (0.055 ≤ P ≤ 0.798). Error bars are one standard deviation.

5.2.6 Sliding

Sliding displacement was defined as the amount of displacement occurring between the two bony fragments along the transverse fracture line. For each specimen of the two groups, sliding displacement was measured at both the anterior markers (AM) and posterior markers (PM) at both MF and CF points. All relevant comparisons between the various eight groups are clearly shown in Fig. 5.13. There was no statistical difference between any of the groups, with the P-value ranging from 0.086 to 1.
Fig. 5.13: A Graph Demonstrating a Comparison of all Phase II Sliding Displacement Groups. AM: Anterior Markers, PM: Posterior Markers, CF: Clinical Failure, MF: Mechanical Failure, P: Plate Group, M: Mouhsine Group, e.g. AMCFP means Anterior Markers at Clinical Failure for the Plate group. No statistical differences between groups were identified ($0.072 \leq P \leq 1$). Error bars represent one standard deviation.

5.2.7 Gapping

Same techniques employed for sliding displacement were then utilized to calculate the amount of gapping i.e. amount of displacement happening between the two bone fragments in a perpendicular direction to the transverse fracture line. The same eight groups were generated and an average and a SD for each group were estimated. Only two comparisons, namely the PM at MF between the Plate and Mouhsine groups, and the PM for the Mouhsine group between CF and MF, were statistically significant with P-values of .000 for both comparisons. All other comparisons were not statistically different with P-values varying from 0.271 to 1. A graph showing all the relevant comparisons between both groups can be seen below (Fig. 5.14).
Fig. 5.14: A Graph Illustrating Gapping Displacement between all Eight Comparison Groups, at Both Anterior Markers (AM) and Posterior Markers (PM) at Both Clinical Failure (CF) and Mechanical Failure (MF) for Both the Plate (P) and Mouhsine (M) Groups. Symbols (#, *) show statistical differences (P < 0.05) between groups. Other comparisons were not statistically significant (0.271 ≤ P ≤ 1). Error bars are one standard deviation.

5.2.8 Failure Modes

Analysis of the failure videos taken from two different angles for each specimen, namely lateral and posterior views, showed that eventually each specimen failed at two main sites (Fig. 5.15): the sacro-iliac joint, where the specimens were mounted to the steel cube resembling the sacrum, and/or the anterior column, where the cup was fixed in place to the anterior column using two cup screws inserted divergently at two different directions as exactly done in real-life events. The Plate group tended to fail earlier at the anterior column location than the SI area, while the Mouhsine group mainly cracked first at the SI location, followed by the anterior column, but eventually most specimens failed at both locations. In addition, the anterior fracture opened up earlier i.e. at less displacement for the Plate group than Mouhsine’s group. One of the specimens in the Mouhsine group failed because the cable’s clamp unlocked causing the cable to slip from above the AIIS. In another cable specimen, the specimen cracked in the anterior column and at the roof of the acetabulum causing the cable to slip as well. Few specimens in the Plate group cracked in the posterior
column around the plate screws. Three specimens from both groups broke around the SP fixation. The quadrilateral plate was significantly displaced in both groups. The amount of gapping and sliding occurring across/along the posterior hemitransverse fracture was captured and reported above. The degree of rotation taking place between the two bony fragments of the posterior fracture was noticeably larger for the Mouhsine group over the Plate group. All the specimens failed from the first wave of the 25mm displacement under the Instron machine except only two specimens from the cable group, which need another session of displacement. The latter continued from the point of stoppage of the first displacement under the same amount of applied force until both specimens eventually failed.

For the Plate group, the FEA peak bone stress and experimental fracture happened at the exact same location, which is the anterior column of the pelvis where there were 2 cup screws and 1 lag screw creating multiple stress risers. Additionally, it is known from the raw data graph for the Plate (Fig. 5.7) that it had a long linear elastic region and short plastic region making the FEA a better and closer predictor of the failure mode.

For the Mouhsine’s cable group, the FEA peak bone stress occurred at the medial aspect of the hip bone at the junction between the cable and iliopectineal line but experimental failure occurred at the SI region across the screw holes, so it is close, but not as accurate as the plate group. This is mainly suggested to be due to the long plastic region as shown in the raw data graph of the cable group (Fig. 5.8) which is not predictable by FEA which is linearly elastic.
Fig. 5.15: Failure Modes for both the Plate and Mouhsine groups showing broken anterior column starting at the anterior cup screws, and broken sacro-iliac region starting at the mounting screws.

5.2.9 Power Analysis

After all statistical analyses were run for Phase II, namely stiffness, failure displacement, failure force, failure energy, sliding, and gapping, the average power analyses for all relevant comparisons were calculated to detect if there were enough specimens per group in order to identify all statistical differences between groups, i.e. avoid type II error. The latter means that a failure to reject a false null hypothesis. The previous statement means that because there are no enough specimens, the experiments fail to show any statistical difference, when there is actually one. A table with all one-tailed and two-tailed power analyses is provided (Table 5.4). If there was more than one comparison per factor, an average was calculated for each factor. For example, for strength and failure energy analyses, there were four relevant
comparisons for each. For sliding and gapping displacements, there were 12 relevant comparisons. While for stiffness and failure displacement, only one comparison was performed for each.

Table 5.4: Averages of the One- and Two-tailed Post-Hoc Power Analysis for all Phase II Outcomes.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>One-Tailed Power (%)</th>
<th>Two-Tailed Power (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stiffness</td>
<td>98.7</td>
<td>97.2</td>
</tr>
<tr>
<td>Failure Displacement</td>
<td>96.2</td>
<td>92.8</td>
</tr>
<tr>
<td>Failure Force</td>
<td>79.5</td>
<td>77.8</td>
</tr>
<tr>
<td>Failure Energy</td>
<td>89.2</td>
<td>85.4</td>
</tr>
<tr>
<td>Sliding</td>
<td>57.6</td>
<td>51.7</td>
</tr>
<tr>
<td>Gapping</td>
<td>60.6</td>
<td>53.3</td>
</tr>
</tbody>
</table>
Chapter 6
Discussion

6.1 General Findings

This study is the first to assess the biomechanical effectiveness of different surgical fixation methods for the AHT associated acetabular fracture, which is considered the typical fragility acetabular fracture pattern seen in the elderly based on experts’ opinion and other epidemiological studies (Culemann et al., 2010; Ferguson et al., 2010; Hessmann et al., 2002).

This analysis provided valuable data about the biomechanical stability of employing three widely used cabling techniques described by Mears and Shirahama (1998), Kang and Min (2002), and Mouhsine et al. (2002) in fixing complex acetabular fractures. No previous biomechanical studies ever assessed the usage of cables in fixing any type of acetabular fractures, making this study a good foundational investigation.

At present, Phase I non-destructive evaluations identified the Mouhsine method as providing the best mechanical stability (i.e. based on lowest peak bone stress) among the cable techniques. Phase II destructive assessments demonstrated the Plate group’s superiority over the Mouhsine group for 2 outcome measurements (i.e. stiffness and failure force at clinical failure), failed earlier for 2 outcomes (i.e. failure displacement and gapping for posterior markers at mechanical failure), and had statistical equivalence for the remaining outcomes.

6.2 Comparison to Prior Biomechanical Studies

It is important to place present results into a broader context by comparison to other studies, where possible (Table 6.1). However, it is difficult to do so because of the large divergence among biomechanical studies on fractured pelvises. First, different types of specimens were used for each study, e.g. various cadavers with a wide range of ages and BMDs (Chang et al., 2001; Culemann et al., 2010; Mehin et al., 2009; Sawaguchi et al., 1984; Simonian et al.,
1995; Wu et al., 2013), and also variable synthetic bones (Culemann et al., 2010; Gililland et al., 2013; Khajavi et al., 2010; Shazar et al., 1998). Second, various force application techniques were used; some used direct force application on the acetabulum simulating the femoral head (Chang et al., 2001; Gililland et al., 2013; Khajavi et al., 2010; Mehin et al., 2009; Shazar et al., 1998), while others applied force indirectly through the fourth or fifth vertebral body transmitted through the sacrum to the pelvis, resembling reactionary joint forces (Culemann et al., 2010; Sawaguchi et al., 1984; Simonian et al., 1995; Wu et al., 2013). Third, specimen preparation differed from one study to another, especially when cadavers were employed, e.g. some studies used a full pelvis with both sides attached with anatomical ligaments, including proximal 1/4th or 1/3th of the femurs, meanwhile others just used hemipelvises covering the full range of different possibilities in between. Fourth, specimen orientations, experimental setups, and testing machines utilized were variable. For some studies, the specimens were oriented 45° superomedially and 15° posteriorly in the sagittal section simulating the position of the femoral neck in the standing position (Chang et al., 2001; Shazar et al., 1998), others used 25° of posterior orientation instead (Khajavi et al., 2010), some applied force directly perpendicular to the acetabulum (Mehin et al., 2009), and others simulated normal walking (6.5° flexion, 7.2° abduction) and descending stairs (9.2° extension, 9.5° abduction) modes (Gililland et al., 2013). Fifth, a huge spectrum of loads was applied on the specimens to detect stiffnesses or displacements ranging from 150 N to 2000 N. Also, some authors used quasi-static loading, while others used cyclical loading with different frequencies ranging from 0.25 Hz to 1 Hz. Sixth, various outcomes were sought from each study varying from axial stiffnesses, yield strengths, maximum strengths, overall displacements, directional displacements, e.g. gapping, sliding, and rotational displacements. Seventh, different definitions were used for the same parameter, e.g. strength was considered by some to be the ultimate failure force (Chang et al., 2001; Gilliland et al., 2013; Shazar et al., 1998), while others considered CF to occur at 2mm gap of the fracture line and MF to be the end of the linear portion of the force-vs-displacement curve (Mehin et al., 2009). The latter was considered yield strength by others (Chang et al., 2001). Eighth, different fracture types and fixation methods, even for the same fracture type, were examined. Most were done on elementary types of acetabular fractures, either a pure transverse or a posterior wall fracture. Finally, no prior studies used the conventional posterior column plate with an
anterior column lag screw to fix AHT fractures, and no biomechanical studies assessed the biomechanical stability of cable fixation for any type of acetabular fractures, a surgical method that is well known and practiced worldwide.

The only available biomechanical study on the fixation of the AHT fracture pattern was done by Culemann et al. (2010), which may be the most comparable to the present study. It provided a comparison of four different fixation methods; two non-locking implants and two locking-implants utilizing both synthetic and cadaveric specimens. The authors measured the main displacement occurring at a point located at the inner wall of the posterior column i.e. on the quadrilateral plate anterior to the anterior border of GSN, as their primary outcome comparing all non-locking implants together and also locking-implants. However, Culemann et al.’s study had several substantial limitations, which have been addressed by the present investigation. First and most importantly, they used only one specimen for each fixation method, thereby not allowing for proper statistical comparison between groups and potentially biasing the final results due to accidental superior surgical performance of one of the specimens. Second, they did not calculate or report any stiffness results for the four constructs, they only presented displacement as their ultimate result. Third, they wanted to see the effect of an extra posterior column long periarticular screw on the stability of the standard plate, so instead of starting with one screw and increasing the number to three screws (two in the anterior column and one in the posterior column), they inserted three screws and tested the specimen, then removed two screws (one from the anterior column and one from the posterior column) and then tested with only one screw. This means they had two wide and deep periarticular holes that were empty, which might introduce stress risers and significantly affect the results. Fourth, they did not report the results for some of their tests, e.g. when they tested the standard plate with only two periarticular screws, and also displacement at the posterior hemitransverse fracture. Fifth, the fracture they created is not a typical high anterior column with a posterior hemitransverse fracture pattern, as fully described by Letournel and Judet (1981). Sixth, the authors did not demonstrate how they mounted the specimen on the testing machine and how exactly they applied the load. Seventh, they did not provide pictures of the lateral view of the acetabulum for the reader to examine how accurate the fixation screws for the locking plates were inserted and if any of them penetrated the joint area. This omission is noteworthy since, for locking plates, screw
direction cannot be really controlled as it is predetermined by the orientation of the plate itself. Finally, only one loading cycle of subclinical loads was used, i.e. 750 N for synthetic and 375 N for cadaveric specimens.

Table 6.1: A general comparison of the outcomes of some biomechanical studies performed on acetabular fractures presenting the conventional methods used.

<table>
<thead>
<tr>
<th>Study</th>
<th>Strength (N)</th>
<th>Stiffness (N/mm)</th>
<th>Macro-Motion (mm)</th>
</tr>
</thead>
</table>
| **Current study (Plate method)** | CF= 3210±543 ‖  MF= 4753±574 | 662±180          | Sliding at CF: AM= 2.55±1.49  PM= 2.40±1.16  
   |                        |                  | Gapping at MF: AM= 5.70±3.59  PM= 4.95±2.03 |
| **Current Study (Mouhsine method)** | CF= 2060±475  MF= 4443±1202 | 386±137          | Sliding at CF: AM= 3.45±1.31  PM= 3.12±1.69  
<p>|                        |                  | Gapping at MF: AM= 10.02±3.82  PM= 13.58±5.09 |
| <strong>Chang et al., 2001</strong> | 0.95±0.36MPa ‖  Yield strength= 0.84±0.25MPa | 2.94±1.23 MPa  (until-failure loading) | using markers, values not reported |
| <strong>Shazar et al., 1998</strong> | --------          |                  |                   |
| <strong>Mehin et al., 2009</strong> | CF=848±805 †  MF=1012±557 † | 456±468 (cyclic loading 50-250N) | Max gap= 0.4±0.5mm  Rotation at CF=1.8°±0.6, at MF=3.4±1.8  using Infrared LED |
| <strong>Khajavi et al., 2010</strong> | --------          | 510±59 (cyclic loading from 0-2000N) | Combination of sliding and gapping (mostly sliding) AM= 2.1±0.6 |</p>
<table>
<thead>
<tr>
<th>Study</th>
<th>Methodology</th>
<th>Results</th>
<th>Notes</th>
</tr>
</thead>
</table>
| Gililland et al., 2013                 | SEM         | 769 with SEM 7
(unti...             | Average displacement
between cup-bone at posterior-inferior region=
0.252
(at base load=50N and peak load=1900N)
using Infrared LED |
| Sawaguchi et al., 1984                 | Infrared LED| Sliding: 0.036mm,
Gapping: 0.145mm (at a load=1334N)
using impedance transducers |
| Simonian et al., 1995                  | Impedance    | Along posterior fracture line
(at a load=150N):
First cycle= 0.13±0.13
Last cycle= 0.26±0.16
using strain gauges across fracture lines |
| Wu et al., 2013                        | Multifunctional digital dial gauge | Lateral displacement of inner wall of PC:
Standing: 1015.5±365.3
Sitting: 129.9±7.5
(at load= 600N) |
| Culemann et al., 2010                  | Multidirectional ultrasonic system | Main displacement of PC of quadrilateral plate (mostly medial displacement):
2.72±1.44 for synthetic,
1.76±0.88 for human specimens with conventional plate w/3 screws
5.8±0.16 for synthetic,
2.40±2.72 for human specimens with conventional plate w/1 screw
(at a load of 750N for synthetic and 375N for human specimens)
using multidirectional ultrasonic system |


¶ CF is defined as peak force at 5mm of vertical displacement of the actuator. MF is defined as the overall peak force in the force-vs-displacement curve.
All measures for strength and stiffness were in MPa (incorrect). Failure occurred most of the times around the two lower screws of the plate.

* Calculated stiffness for PCPLS from anterior displacement (giving higher values).

# Authors defined CF as failure happening at 2mm gap and MF as failure occurring at the end of the linear portion of force-vs-displacement curve.

⁻ Only two specimens were used. All specimens failed at the sacro-iliac mounting holes.

‖ Type of displacement was not identified whether it was gapping, sliding, or rotation.

| Mostly due to errors in the anterior displacements happening in the sitting position, which were not presented in the study. Displacements were not specified (gapping, sliding, or rotation).

6.3 Biomechanical Implications

Phase I results have several salient implications on the initial biomechanical stability of AHT fracture fixation prior to any significant weight bearing.

FEA showed that the acetabular cup increased the overall stiffness by 74% and decreased the peak bone stress by 42% in respect to the “Intact” control. This finding assists in understanding one of the many reasons for the great long-term success achieved by currently available acetabular cups. The latter has also an outer porous surface at the micro-level to stimulate the migration and activation of osteoblasts to its surface encouraging “contact osteogenesis” i.e. bone growth and interdigitation with the implant surface with no micro-gaps between the prosthesis and the newly-formed bone leading to better biomechanical stability (Davies, 2003)

FEA demonstrated that all fixation methods provided stiffnesses higher than the Intact group, with the Mouhsine group being the lowest offering 20% increase in stiffness than the Intact specimens, which means they are potentially acceptable ways for fixing acetabular fractures. It is also noticeable that the Plate and Kang groups achieved very similar values and locations of stiffness and peak bone stress results, meanwhile the stiffness and stress maps for the Mears and Mouhsine groups were alike, employing that each of these two methods can be theoretically applied interchangeably with accomplishing almost the same successful results. Additionally, it was shown that all the four methods provided good biomechanical stability for the immediate post-operative state handling stress applied on the bone from a
range of 1 ¼ body weights for the Plate group being the lowest to 1 ¾ times body weight for the Mouhsine group, when the peak bone stress attains 106MPa i.e. the maximum tolerable value for tensile strength before the cortical shell starts cracking. These values are by far higher than the actual amount of force needed to be applied in the first few weeks after surgery, which usually reaches only 10% of the body weight (Simonian et al., 1995) allowing for good fixation of bone fragments and giving them the chance to fully and properly heal together. Graded increase of load-bearing then follows till it reaches full weight bearing 12 weeks post-operatively.

“Stress Shielding” theory affirms that the stiffer the implant is, the higher the load the implant carries. This would lead to lesser load being transferred from the implant to the bone, producing less stress on the bone causing eventual bone resorption, implant loosening, and failure. The Plate technique, which is mainly dependent on bone/screw interaction for both the fixation of the plate and the lag screw, may be more affected by the stress shielding theory than the cabling technique, which mainly relies on the outer compression of bony fragments together rather than bone/implant interface. Consequently, even if bone loosening happens on the micro-scale, it would minimally affect the biomechanical stability offered by cables. Furthermore, as clearly seen from the FEA, the Mouhsine cabling technique generated better stress distribution of the bone, being wrapped all around the hip bone, as illustrated by a 30% decrease in the peak bone stress, yielding 181MPa vs 255MPa for the Plate group at a force of three times body weight i.e. 2207N. Even at lower force values, the ratio between peak bone stresses for both groups remains constant because the FEA assumes absolute linearity.

Phase II results also have several important implications on the biomechanical stability of AHT fracture fixation after significant weight bearing is introduced, which could lead to failure.

The Mouhsine cable method currently demonstrated statistical equivalence to the conventional Plate technique for a number of biomechanical outcomes, suggesting it is a feasible surgical approach. This is in agreement with Mouhsine et al.’s report on the clinical outcomes of this procedure in the presence of an acute THA in 18 older patients (average age
of 76 years) for a mean follow-up of three years. The report showed that the functional outcome of the patients was excellent in 65% of the cases and good for 35%. The mean hip motion was 108° of flexion and 12° of extension, 32° of internal rotation and 35° of external rotation, and 30° of abduction and 26° of adduction. Radiologically, 88% of the patients showed no radiolucency, while 12% had a less than 1mm central radiolucency. All fractures healed with 87% of the patients having an average of 2.3mm superior and 2mm medial migration of the acetabular cup, which happened in the first three month post-operatively and stabilized afterwards. Heterotopic ossification of Brooker grade I was noted in 35% of the patients. These results are comparable to the results reported in the literature for various fixation techniques in elderly patients with displaced acetabular fractures (Mears and Velyvis, 2002; Sarker et al., 2004; Tidermark et al., 2003).

The Plate group was much stiffer than the Mouhsine group presenting 72% increase in the overall construct stiffness, which would lead to better immediate post-operative stability. Accordingly, with the same force applied on both groups e.g. bedside activities or rehabilitation, more movement between the bone fragments will occur in the case of the Mouhsine group than the Plate group, which may endanger bone healing. Previous clinical and biomechanical studies reported that the higher the stiffness of the applied implant would lead to higher overall stiffness of the construct, which would lead to better fracture stability in the recent post-operative period i.e. before bone healing occurred (Dennis et al., 2001; Fulkerson et al., 2006; Lever et al., 2010; Peters et al., 2003; Schmotzer et al., 1996; Talbot et al., 2008; Wilson et al., 2005; Zdero et al., 2008b).

Regarding failure force results of both groups at the CF event, where the Plate group was 1.56 times stronger than the Mouhsine group, implies that more load needs to be applied on the Plate specimens in order to achieve the same amount of displacement taking place in the Mouhsine group. The same trend was also confirmed by investigating the MF results for both groups, where the Plate group achieved higher failure strength, though not statistically significant, with much lower failure displacement only constituting almost a half i.e. 55% of its counterpart at the Mouhsine group (P=0.005). This might give the impression that the Plate method is superior. However, this is not necessarily the case for elderly (osteoporotic) patients, who have fixation of an acetabular fracture accompanied with a THA and are not
altogether permitted to put any weight-bearing on their legs for at least the first six weeks post-operatively to allow for proper bone healing. The reason is that, in the immediate post-operative interval, all the loads applied are mainly passive movements performed by physiotherapists, occupational therapists, or nurses, plus any additional bedside movements, activities, or exercises the patients are asked to do. All these movements would only present 10% of the patient’s body weight (Simonian et al., 1995). As such, it can be observed that for the Mouhsine group, which is weaker, a load of 2060 N i.e. 2.8 times body weight of an average 75 kg person, was achieved. This means that a load with the magnitude of 28 times the force normally generated from the regular post-operative movements is required to clinically fail the Mouhsine fixation technique. Moreover, considering that there was no statistical significance in sliding and gapping at CF for either the AM or PM between both repair groups, it could be recommended that the Mouhsine fixation be applied. This is because at the very low load applied in the immediate post-surgical condition, a little bit more micromotion may occur between the bony fragments of the fracture, which might mechanically stimulate bone osteoblasts to start osteosynthesis earlier and cause better bone healing and union. The effect of slight micromotion between bone fragments on increasing the rate of bone healing was proved by some previous clinical in-vivo and in-vitro studies (Hak et al., 2010).

The Mouhsine group provided almost the same amount of failure force and higher failure energy (66% increase) at MF vs. the Plate group, even if statistically insignificant. This trend demonstrates the ductile nature of the cable/bone construct versus the more brittle nature of the plate/bone implant. Ductility by definition is “a solid material’s ability to deform under tensile stress”, which demonstrates the material’s capacity to stretch into a wire. Britteness is defined as “a material, when subjected to stress, it breaks without significant deformation”. To clarify, a material may be very strong, but brittle, as it does not absorb much energy before breaking e.g. ceramics and glass. To apply that on the two constructs, the Mouhsine group is less stiff, but it deforms substantially, i.e. displacement and absorbs much energy (long plastic region) before mechanically breaking down, even if it failed clinically much earlier, resulting in much higher failure energy than the other group. On the other hand, the Plate group is much stiffer, but not as ductile i.e. brittle, so it is stronger, but does not deform much (shorter plastic region), and therefore, does not absorb as much energy before failure.
Therefore, the amount of displacement at MF with the cable group is much larger, which consequently lead to more sliding, gapping and rotation. Also, because the plate is always in contact with the bone in a plane rather than just a line, it was able to restrict rotation along the posterior component of the fracture, which was not the case with cables because it holds the bone in just a slender line allowing the potential for rotation, especially under much displacement because of its ductile properties.

### 6.4 Surgical Implications

Several technical improvements can be suggested in the use of cables and screws for surgical practice and future studies based on the current author’s experience during specimen preparation.

First, two out of 10 cables employed in the Mouhsine fixation method slipped from above the AIIS at very high loads i.e. above 3500N. This happened in one case mainly because the clamp screw unlocked allowing lengthening and slippage of the cable. However in the second case, it was mainly because of breakage of the iliac fragment at the SI region and at the rim of the acetabulum at the roof causing slippage of the cable from around the AIIS. Despite that, it would be recommended to drill a small 2.5 or 2.7mm hole at the base of the AIIS directed from postero-lateral to antero-medial to pass the cable through it, as routinely done in Mears and Shirahama (1998), in order to secure the cable firmly and prevent any chances of slippage e.g. in case of a patient stumbling post-operatively and generating a high and sudden force magnitude at the hip joint.

Second, the Accord cabling system [Smith and Nephew Inc., Memphis, TN, USA] was employed. However, for both the Mears and Mouhsine techniques a longer course of the cable around the specimen is required vs. the Kang method, thus the remaining free end of the cable after passing through the clamp was too short to properly fit in the cable tensioner. The cable tensioner could not provide gripping of the cable when the “Locking Handle” was turned clockwise all the way. Therefore, the anterior nozzle (end) of the tensioner was unscrewed to save some length, and so, the free end of the cable can be inserted deeper into
the tensioner allowing a good grip of the tensioner on the cable. A wide washer was used at the front end of the tensioner to push against the clamp so that the latter does not sink inside the tensioner, and therefore, the clamp’s screw can be tightened using a surgical screwdriver. Therefore, it is proposed that either at least 10 cm longer cables (current cable length was 61.2 cm), or cable tensioners that act effectively with much less length (at least 10 cm) provided by the free end of the cable, should be designed for achieving effectiveness and comfort to operating surgeons. Furthermore, an Accord clamp taken from another new cable was used as a supplementary second clamp applied on the quadrilateral plate for the Mouhsine’s technique. However, this clamp was complex, difficult to pass the end of the cable twice across, and not meant to be used in the quadrilateral area for that purpose. Therefore, a simpler two-channel straight second clamp applied on the quadrilateral plate, as originally described by Mouhsine et al. (2002), needs to be designed for the easier passage of the cable.

Third, it was noted that for both the Plate and Mouhsine groups, all specimens failed first at either the sacro-iliac region, or the anterior column around the acetabular component fixation screws. For the SI region, it was noticed that cracks started at or propagated to the holes of the mounting screws causing eventually a complete fracture of that area along the line of fixation, as shown in Fig. 5.15 in the results section. This finding was also reported in another recent study performed on fixing pure transverse acetabular fractures applying three different constructs, where all specimens failed at the same region around the mounting holes as well (Gililland et al., 2013). For failing at the anterior column region, it was clear that the anterior column of those synthetic hemipelvises, and most likely also applies on human cadaveric specimens based on experts opinion, was really narrow and was further narrowed after reaming of the acetabulum to insert an acetabular shell as part of a THA. The diameter of the anterior column could not handle the passage of a 6.5mm lag screw in the standard orientation from the outer table of the ilium supra-acetabularly close to the AIIS through the acetabular component of the ilio-pubis to settle down in the superior pubic ramus close to the pubic body. Therefore, when this finding was detected in the pilot studies before the start of the current study, lag screws were changed into 4.0 mm ones instead. Thus, in osteoporotic patients with displaced acetabular fractures that need primary acute THA, it is advisable not to use two anterior column screws. One screw in the anterior column may be enough because
the anterior column is already weak, narrow, and osteoprotic, and the final lag screw to be inserted may create added stress risers and predispose the column to fail earlier. It may be intuitively argued that inserting more screws in the anterior column will make it stronger; however, because of the narrow area and poor bone quality, installing two large 6.5mm screws for stabilizing the acetabular cup is replacing a large amount of bone, which is already scarce, further weakening it. If surgeons are looking for better acetabular component stability, other large screws can be inserted in the dome of the cup, posteriorly towards the sciatic buttress, or even inferiorly in the ischium, which all were proven from this study to be extremely tight and strong and remained in place even after mechanical failure. Based on the current results, it is recommended that redistribution of the cup screws with just one anterior column screw and more dome, posterior column, or ischial screws. Otherwise, four 6.5mm cup screws were enough providing a 74% increase in the overall stiffness and 42% lower peak bone stress in comparison to the Intact group.

Fourth, the Mouhsine cable method is preferred among the various cable methods examined from a biomechanical point of view. However, the final decision is left to the operating surgeon because, from the surgical point of view, cables generally are technically much difficult to apply than the Plate method. Cabling requires much more stripping, splitting, and dissection of soft tissues and muscles in the vicinity of many vital structures, such as arteries, veins, and nerves. Moreover, the Mouhsine method in particular is even more technically demanding than the other two cable methods. Therefore, successful Mouhsine-type cabling would require substantial surgical training and practice by a specialized trauma surgical team to avoid vascular or neural injuries, minimize surrounding soft tissue trauma, and maintain good circulation in the area, i.e. periosteum, surrounding muscles, and soft tissues, for proper bone healing.

Fifth, the Mouhsine cabling method provides a great benefit over the Plate group, which is mechanically supporting the quadrilateral plate, having an extra clamp resting on the quadrilateral, with four cable ends going all the way across it. This advantage is considered very important for the success of any acute THA performed as a primary management for displaced acetabular fractures especially in the elderly, even if using procedures other than cables for fixing the fracture itself (Sarker et al., 2004). From the stability point of view, the
Mouhsine cable method is specifically superior because of the addition of an extra clamp i.e. sleeve on the quadrilateral plate, which serves as an extra origin for the cable that the more the tension is applied on the cable e.g. femoral head pushing against the acetabular component of a THA, the more the compression on the bone is applied. That is because when the clamp is pushed out by bone under the influence of the applied force, the clamp itself pulls all the four ends of the cable going through it generating more compression all the way around the bone compressing the bone fragments together and providing more support to the quadrilateral plate to keep its integrity and resist displacement i.e. sliding or gapping. That is referred to as “Tension band” theory.

Finally, From a clinical perspective, stiffness, CF force, and gapping are clinically relevant, where patient’s pain, discomfort, or functional impairment may start to occur. For stiffness, the higher the stiffness of the construct, the better the immediate post-operative stability. Also, a higher CF force means that more force was required to achieve the same amount of displacement i.e. 5mm. The latter indicates the strength of the construct, and therefore, the success of the fixation technique. For gapping, it has been previously shown that the better the fracture reduction with less gaps, the better the healing and clinical outcomes (Matta et al., 1986; Matta and Merritt, 1988; Matta, 1996).

### 6.5 Possible Limitations

The limitations of the current study are typical for in-vitro biomechanical studies, which may have influenced the absolute values of the reported outcomes but not the relative trends.

Synthetic, instead of cadaveric, hemipelvises, were used. Based on their manufacturing technique (i.e. glass fibers in epoxy resin are mixed thoroughly) and measured mechanical properties which fall within a relatively small range (i.e. longitudinal vs. transverse vs. compressive), the synthetic bones supplied by Sawbones are more isotropic and homogeneous than human bones whose microstructure follows lines of applied stress. Therefore, the outcomes of the present study need to be validated in future studies versus fresh frozen human cadaveric hemipelvises to achieve the absolute numbers of the
parameters addressed by this study and also confirm the same comparative trend between the various fixation groups. This limitation may affect the absolute values for the study’s outcomes causing more variation between the specimens and maybe decreased stiffness and strength, and increased failure displacement. However, the main reason for the use of synthetic hemipelvises was that inter-specimen variation of synthetic bones is much lower than that of cadavers (Cristofolini et al., 1996). Synthetic bones from Sawbones [Vashon, WA, USA] are also commonly used in many biomechanical studies on long bones such as femurs, tibias, and humeri (Aziz et al., 2014a,b,c; Bougherara et al., 2009; Ebrahimi et al., 2012; Shah et al., 2011; Tuncer et al., 2014) and compare reasonably well to human cadavers for a variety of mechanical properties (Aziz et al., 2014a,b,c; Dennis et al., 2000; Papini et al., 2007; Zdero et al., 2009) Moreover, the high financial cost, storage requirements, biotoxicity, and limited access to matched pairs of human hemipelvises made synthetic bones an attractive option for this study.

Osteoporotic-like synthetic hemipelvises designed based on data obtained from a 45 year old male were initially planned for the study. There may be no differences in the human anatomy of the pelvis between younger and older populations, except for thinner cortices, lower BMD, and increased porosity, where bones of the elderly may become more osteopenic and sometimes, even osteoporotic.; however, there is very limited data addressing this issue (Brockstedt et al., 1993; Foldes et al., 1991; Vedi et al., 2011). In light of the current results, the author expect that using osteoporotic bone would be in favor of the cabling technique, which mainly depends on compressing the bony fragments together vs. the plating technique, which primarily employs the bone/screw interactions. However on the other hand, the cables may also cut through the osteoporotic specimens leading to an early failure. Therefore, the final conclusions would be based upon completion of a similar study on true osteoporotic fresh frozen human cadavers. To date, “fully osteoporotic” synthetic hemipelvises are not commercially available from any medical product manufacturer. Although Sawbones only produces one standard density and geometry for the cortical shell which simulates normal bone, the lowest possible density of 0.16 g/cc was chosen for the cancellous matrix to represent osteoporosis.
A quasi-static axial load, instead of cyclic load, was applied for mechanical testing. Cyclic loading is more physiological as it resembles daily-life activities, e.g. walking, running, climbing stairs, and playing sports. However, a quasi-static loading minimized visco-elastic effects, thus producing a more “conservative” result. Also, the long-term fatigue failure of the different constructs was not assessed, which usually often requires 1 million cycles. Under repetitive cyclical loading, the fixation implants develop more “mechanical slack”. The latter would affect the present mechanical results in the form of decreased stiffness and strength and increased gapping, sliding and rotational displacements. However, the relative trends between the groups would likely be maintained. For instance, the latter was done by Simonian et al. (1995), where the displacement across the posterior fracture line of a T-shaped fracture was 0.13±0.13 in the first cycle and 0.26±0.16mm at the last cycle.

Bone healing was not considered in the scope of this study as the author meant to look at the biomechanical stability in the immediate post-operative period before any substantial bone healing and partial weight bearing can take place. To simulate bone healing using FE models, the fracture gaps could be abolished replicating full bone union as previously done in other biomechanical studies (Bougherara et al., 2009; Ebrahimi et al., 2012). However, it is very difficult to simulate progressive degrees of bone healing in in-vitro biomechanical experimental studies (other than full union) such as those for this study. If bone healing is to occur, it will affect all the relevant biomechanical outcomes in the sense that it would probably increase the overall stiffness, load and energy to CF, and decrease the failure displacement i.e. sliding and gapping.

There are different groups of muscles that act on the hip joint in order to achieve all the wide spectrum of movements happening across that joint, namely flexion, extension, abduction, adduction, medial rotation, lateral rotation and circumduction. The main hip flexor is the psoas muscle, helped by the iliacus. Extension is mainly performed by the gluteus maximus. Adduction is mainly carried out by the adductor group of muscles. Hip abduction is mainly exerted by the gluteus medius and minimus. In the current study, the effect of muscles and surrounding soft tissues on the bone stiffness and stress distribution was not determined. Bitsakos et al. (2005) showed that the more muscle attachments included in the FE models, the more bone conserving effect around the THA would eventually result. Stolk et al. (2001)
showed that an FE model involving only the abductor muscle force vector can adequately imitate the mechanics of cemented THA during in vivo loading. Several other studies have further simplified the process by not simulating any of the muscle groups around the hip joint (Bougherara et al., 2010a,b, 2011a,b; Davis et al., 2008; Dennis et al., 2000, 2001; Dubov et al., 2011; Ebrahimi et al., 2012; Fulkerson et al., 2006; Lever et al., 2010; Peters et al., 2003; Schmotzer et al., 1996; Shah et al., 2011; Stevens et al., 1995; Talbot et al., 2008a,b; Wilson et al., 2005; Zdero et al., 2008b). Additionally, the FEA in the current thesis was designed to simulate the exact experimental setup (and not the real life clinical situation) in order for the experimental results to be adequate for validation of the FEA outcomes. If the abductor force was modeled in the current study, it would have neutralized some of the stresses applied on the hip bone decreasing the overall absolute values of Von Mises peak bone stresses; however, this would not affect the comparative relations between the various fixation methods. However, this is a typical simplification for in-vitro biomechanical studies, even for those which use human cadaveric specimens, because soft tissues and muscles are frequently stripped off the bone.

Almost all the specimens failed during the “mechanical failure” event at or around the mounting holes of the sacro-iliac region. Although this finding was also reported by another recent biomechanical study done on transverse acetabular fracture fixation (Gilliland et al., 2013), it cannot be firmly concluded whether this occurred mainly because the screw holes in the bone at that region were close to each other or it is a normal finding given the nature of the test setup.

The AHT fracture was created for the current study by drawing the fracture lines on the exact location every time and using a jigsaw, the fractures were consistently created in all the specimens. However, this way of making fractures does not replicate the real-life situations causing bone fractures. On the other hand, a small drill bit e.g. 1.5mm can be used to drill some partial-thickness holes along the desired fracture line 2mm apart. By using an osteotome and applying a force along that line, the fracture can be created in a more realistic and anatomic manner, but it may be less consistent way because of the increased number of steps (Simonian et al. 1995).
For Phase I a low axial load of 200 N was applied, which is much lower than physiological loading. A previous biomechanical study on a simple transverse acetabular fracture, dividing the bone in only two fragments instead of 3, applied a maximum load of 250 N (Mehin et al., 2009), and another study on a T-shaped acetabular fracture applied an even lower load of 150 N (Simonian et al., 1995). Therefore, low load was intentionally selected to prevent damaging the specimens during strain gauge measurements, which are meant to remain within the linear elastic range.

FEA assumes absolute elastic linearity regardless of the amount of load applied. Thus, when different fixation groups are compared together for relative performance between them, rather than absolute values reflective of “real life”, the exact same trend between groups is achieved whether FEA is run far below (i.e. technically correct) or far above (i.e. technically incorrect) the yield point. Consequently for the present goal of Phase 1, FEA analysis at low (or high) loads was appropriate and predictive of which repair method would fail earlier and which would fail later. However, the author did not use the FEA for modeling bone failure, but to achieve a preliminary idea of how the different fixation groups would behave in relation to each other until the yield point i.e. the elastic region of the stress-strain curve.

Experimental stiffnesses were lower than FEA stiffnesses since absolute relationships between parts are assumed for FEA, e.g. for the fixed support regions at SI and SP locations, FEA assumed no micro-motion in that area, “bonded” contact between SI screws with bone and steel block by definition have no movement, etc, that do not replicate the mechanical slack that possibly existed in the experimental setup.

For the validation graph, the correlation coefficient was only R=0.82 and not higher as previously reported by other biomechanical studies performed on long bones (Bougherara et al., 2010b; Bougherara et al., 2011b; Dubov et al., 2011; Ebrahimi et al., 2012; Shah et al., 2011). However, this data scatter was likely caused by a combination of the complex geometry of the pelvises in comparison to long bones and human error involved in orientating and mounting specimens. Even so, the slope of the validation graph reached a value of 1, which represented the accuracy of the FE model vs. experiments.
The test setup’s mechanical stiffness was not accounted for in the measured stiffnesses. This is because the Instron’s mechanical stiffness was 260 kN/mm, which is two orders of magnitude larger than the measured stiffnesses for all groups in Phase I and II, thereby eliminating this factor as a confounder.

Rosette strain gauges were chosen, instead of linear gauges, since rosettes are known to be more suitable for curved surfaces. Unfortunately, some variation in the rosette readings was noticed, possibly due to the highly complex 3D geometry of the specimens and the presence of plates, screws, and cables, which can introduce stress risers in the vicinity of the rosettes.

Specimens were over-reamed by 1 mm relative to the diameter of the cup, which would not be practiced in real-life surgery. However, this was performed because the thickness of the jigsaw used for cutting the specimens and creating the fracture is almost 1 mm with angled teeth, which eventually lead to a total loss of bone material of 1 mm. This over-reaming enabled perfect reduction of fractures even after the insertion of the acetabular shells.

In the Plate group, the lag screw inserted into the anterior column was oriented in the standard fashion, i.e. textbook version, and not the practical fashion, i.e. daily surgical practice. This also allowed for practical advantages during specimen preparation for mechanical testing.

One of the most important factors for the success of the CHP is to provide enough mechanical stability to allow for bone ingrowth around the acetabular shell, and therefore, micromotion at the cup bone interface is a very important biomechanical parameter to assess as done in a prior biomechanical study (Gililland et al., 2013). However, the current study did not evaluate this outcome for practical issues and absence of the proper equipment needed for the accurate measurement of the micromotion at the bone/implant interface. However, the overall stiffness, which represents the total displacement for a given force, as well as the failure energy at CF, which incorporates both the applied force and the concurrent displacement, can provide a rough idea about the overall mechanical stability of the fixation technique around the cup. Also, gapping and sliding along the posterior fracture component of the AHT can estimate the success of the fixation techniques as discussed before in some
clinical studies about displaced acetabular fractures (Matta et al., 1986; Matta and Merritt, 1988; Matta, 1996).

The orientation of the specimen in the setup under the mechanical testing machine did not allow much access to assess the exact amount of medial displacement of the quadrilateral plate, i.e. femoral head, which is regarded as the mainstay for the repair of acetabular fractures using acute primary THA, and is also looked at as a very important category in the long-term follow-up evaluation of the success of the fixation method performed. Also, this rigid support might have decreased the medial displacement of the acetabulum. However, similar rigid support was employed by almost all other biomechanical studies performed on fixation of pelvic fractures. Additionally, because the same setup and orientation was used for all the specimens, the relative performance between various groups was not affected.

A more basic measure was taken to assess the displacement occurring along the posterior fracture line using markers and videotaping the whole displacement process. This procedure enabled a calculation of sliding and gapping; however, rotation was seen but not measured. Rotation is deemed to be essential in acetabular fractures because most fractures are mainly rotated more than translated (Vanderschot, 2007). Other studies employed more sophisticated techniques in identifying all 3D displacements occurring along one or more areas of the acetabulum such as infrared LED (Gililland et al., 2013; Mehin et al., 2009), multidirectional digital dial gauge (Wu et al., 2013), and multidirectional ultrasonic system (Culemann et al., 2010), which were not readily available in the current study.

Post-hoc power analysis showed that sliding and gapping was below 80%, indicating that the study was under-powered for these parameters. Thus, not surprisingly, no statistical difference was noted for these outcomes. However, stiffness, failure displacement, failure force, and failure energy approached or achieved 80%.

Finally, the current study did not include other methods applied for fixing AHT fractures, such as the conventional pelvic brim 12-hole long reconstruction plate, locking plates, or other buttress plates used for supporting the quadrilateral plate, for instance as done by Culemann et al. (2010). Also, fixation of other complex acetabular fractures was not currently addressed. However, as acetabular fractures are becoming more common and rapidly
increasing especially in the elderly population, further clinical and biomechanical studies should expand upon the current methods and findings.
Chapter 7
Conclusions

To date, this is the first biomechanical study on the stability of the standard Plate method, as well as 3 different cable methods, commonly used for fixing complex acetabular fractures in the elderly as part of the combined total hip arthroplasty (THA) procedure, which includes both ORIF of the acetabular fracture with the simultaneous replacement of the hip joint.

Methods for Phase I involved non-destructive FEA and mechanical testing for an initial comparison of the repair methods (i.e. stiffness and peak bone stress), whilst methods for Phase II employed measurement of destructive mechanical testing parameters (i.e. stiffness, failure force, failure displacement, failure energy, sliding motion, and gapping motion). Additionally, an intact group acted as a “control” and another intact group had only an inserted acetabular component of a THA. The clinical situation of osteoporosis was simulated employing “osteoporotic-like” hemipelvises and an anterior column with a posterior hemitransverse (AHT) acetabular fracture.

Results for Phase I concluded that, among cable groups, the Mouhsine group provided the best biomechanical stability, yielding the lowest peak bone stress of 181 MPa at 3 x body weight for a 75 kg person. Results for Phase II showed the Plate group’s superiority over the Mouhsine group for 2 outcome measurements (i.e. stiffness and failure force at clinical failure), its inferiority for 2 outcomes (i.e. failure displacement and gapping for posterior markers at mechanical failure), and statistical equivalence for the remaining parameters.

The biomechanical and surgical implication is that the standard Plate technique was superior for the 2 critical outcomes of stiffness and failure force during clinical failure event. However, the Mouhsine cable technique can be a reasonably effective alternative to the standard Plate group for repairing complex acetabular fractures in the elderly, as clinically reported by others in the literature (Mouhsine et al., 2002, 2004). Moreover, there are technical aspects of surgical repair which this study highlighted, that may be considered by surgeons regardless of which of these two methods they choose.
Chapter 8  
Future Directions

The following is a grant submitted to the Orthopaedic Trauma Association (OTA), which is a potential future study related to the current thesis.

8.1 Biomechanical Optimization of Complex Acetabular Fracture Fixation using Computational Modeling and Human Cadaveric Testing

8.1.1 Definition of the Research Question

The incidence of acetabular fractures is expected to keep increasing worldwide, especially among the elderly population (Grazier et al., 1984; Jeanotte and Moore, 2007). A study of 1309 patients which covered the years 1980 to 2007 revealed a difference between the first and second half of this time period, namely, a 30% increase in the frequency of acetabular fractures and a 2.4-fold rise in the number of elderly patients over 60 years of age with this injury (Ferguson et al., 2010). As many as 80% of these fractures are complex (i.e. associated), having multiple fragments and/or fracture lines (Wheless, 2014) and about 83% of these injuries are caused by motor vehicle accidents or falls (Ferguson et al., 2010; Laird and Keating, 2005). Combining data from the only prior epidemiological studies shows that the two most common acetabular injuries comprise 43% of all acetabular fractures and are of the complex type, namely, transverse plus posterior wall (TPW) and both column (BC) (Fig.8.1) (Ferguson et al., 2010; Laird and Keating, 2005).

However, there are no previous biomechanical studies on surgical fixation of TPW fractures and only one biomechanical investigation on BC fractures (Wu et al., 2013). Specifically, Wu et al. (2013) fixed a simulated BC acetabular fracture using an anterior construct plate plus a 1/3 tubular buttress plate which provides quadrilateral area fixation, as well as a plate for the iliac crest. Yet, this study has some limitations because the authors used only a small number of human cadavers (n=6), they tested only one method of surgical repair, they
applied only a low sub-clinical load (600 N), and they did not measure any load-to-failure parameters. Therefore, a comprehensive biomechanical evaluation of surgical fixation for both TPW and BC acetabular fractures is clearly warranted, which uses a larger number of human cadaveric specimens, applies more physiological-type loads, and compares multiple surgical repair methods.

Therefore, the aim of this study is to find the optimal biomechanical fixation methods for TPW and BC acetabular fractures. This will be done using computational modeling, as well as mechanical testing on artificial and human cadaveric hemipelvises. It is hypothesized that one method will clearly be the most mechanically stable way to fix TPW fractures, as well as one method for BC fractures, by providing the highest construct stiffness, the highest construct strength, the lowest bone stress, and the lowest interfragmentary motion compared to other repair methods. If the null hypothesis is fulfilled by way of no statistical differences in the biomechanical properties between repair methods, then the optimal method will be identified as the one that is technically easiest to perform surgically and uses the fewest implants.

Fig. 8.1: Complex acetabular fractures to be examined. (a) transverse plus posterior wall (TPW), and (b) both-column (BC). Fixation methods are not shown.
8.1.2 Importance of the Research Question

The goal of this study is to investigate the biomechanical stability of a series of surgical methods for repairing TPW and BC acetabular fractures in order to determine an optimal repair technique for each injury type. This will have several important consequences:

First, this study will be the only one in the literature to comprehensively address the biomechanics of fixing TPW and BC acetabular fractures, which are the two most common types of acetabular injuries.

Second, as a consequence of such evidence, orthopaedic surgeons will more easily identify the best way to repair these fractures and adopt these methods into their practice based on objective scientific evidence, rather than simply personal preference or subjective experience.

Third, choosing the best surgical repair method will also allow quicker and better fracture union at the injury site, thus promoting healing, minimizing complications, and decreasing the risk for re-operation.

Fourth, this will then allow patients to have earlier mobility and weight bearing shortly after surgery, thereby facilitating a quicker recovery time, a regaining of their pre-injury functional status, a return to their usual daily routines, and an increased quality of life.

Fifth, quicker recovery times result in shorter hospital stays, thus reducing overall healthcare system financial costs.

Finally, this study will be the first step in validating the widely-used artificial hemipelvises from Sawbones (Vashon, WA, USA) against human fresh-frozen cadaveric hemipelvises, which are considered the “gold standard” for biomechanical testing. This will aid biomedical engineers to improve the characteristics of these artificial hemipelvises and upgrade the quality of future biomechanical investigations using these surrogate bones.
8.1.3 Research Design

8.1.3.1 General Strategy

Phase 1 involves experimentally-validated computational finite element (FE) analysis of 5 repair methods for TPW (and BC) fractures to identify the 2 most biomechanically stable methods. Phase 2 involves biomechanical experimentation on human cadaveric hemipelvises to directly compare these 2 repair methods. Prior computational and experimental analyses on acetabulae will serve as guidelines for the proposed study (Chang et al., 2001; Culemann et al., 2010; Khajavi et al., 2010; Mehin et al., 2009; Sawaguchi et al., 1984; Shazar et al., 1998; Shim et al., 2010, 2011; Simonian et al., 1995; Wu et al., 2013).

8.1.3.2 Phase 1 – Computational Modeling

CAD models will be created of an artificial hemipelvis (Model #3405, Sawbones, Vashon, WA, USA), as well as for fixation plates and bone screws. The FE model will use bone material properties provided by the manufacturer, stainless steel material properties for plates and screws, and simplifying assumptions typical of FE studies (Shim et al., 2010, 2011). TPW and BC fractures will be modeled and repaired with 5 fixation methods commonly used clinically (Table 8.1) (Letournel and Judet, 1993; Moed and Reilly, 2010; Wheeless, 2014). FE stiffness and stress analysis will be done (ANSYS Workbench 15.0, ANSYS, Inc., Canonsburg, PA, USA) by applying 250 N through a steel ball (i.e. femoral head) into the acetabulum. The force will act at 45 deg superomedially and 20 deg posteriorly in the sagittal plane to simulate force vector orientation during single-leg stance (Chang et al., 2001; Mehin et al., 2009; Shazar et al., 1998). Of the 5 repair methods for each fracture type, the 2 most mechanically stable methods by definition will have the highest stiffnesses and lowest peak bone stresses. The FE model will be validated by non-destructive mechanical tests (250 N, 10 mm/min, 3 rampup-rampdown cycles) on artificial hemipelvises equipped with strain gages and having TPW (and BC) fractures that are repaired, oriented, and loaded as in the FE model (Chang et al., 2001; Khajavi et al., 2010; Sawaguchi et al., 1984; Shazar et al., 1998) (Fig.8.2).
Table 8.1: Surgical repair methods to be biomechanically tested [4,16,17]. AC = anterior column, AIIS = anterior inferior iliac spine, IC = iliac crest, IIF = internal iliac fossa, PC = posterior column, PW = posterior wall.

<table>
<thead>
<tr>
<th>TPW repair methods</th>
<th>BC repair methods</th>
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<tbody>
<tr>
<td>1 PC plate, PW small plate</td>
<td>AC plate, PC lag screw towards ischial spine</td>
</tr>
<tr>
<td>2 Method 1 + AC lag screw</td>
<td>Method 1 + PC lag screw towards greater sciatic notch</td>
</tr>
<tr>
<td>3 PC plate, 2 PW screws</td>
<td>AC plate, PC lag screw, IC plate</td>
</tr>
<tr>
<td>4 Method 3 + AC lag screw</td>
<td>AC plate, PC lag screw, IIF plate towards AIIS</td>
</tr>
<tr>
<td>5 PC lag screw, PW small plate</td>
<td>AC plate, PC lag screw, IC plate, IIF plate towards AIIS</td>
</tr>
</tbody>
</table>

Fig.8.2: Experimental setup for Phase 1 and 2. Fixation methods are not shown. Dotted lines (····) are fractures. Dashed lines (---) are hidden structures.

8.1.3.3 Phase 2 – Human Cadaveric Testing

Ten matched pairs of human fresh-frozen cadaveric hemipelvises will be used for TPW fractures, as well as another 10 pairs for BC fractures. DEXA scans will provide bone mineral density and T-scores. After TPW (or BC) fracture creation, left hemipelvises will be repaired using one of the 2 best methods from Phase 1, and right hemipelvises will be repaired with the other approach. Adhesive markers will be attached on either side of the fracture line along the posterior column to measure interfragmentary motion (Khajavi et al., 2010; Mehin et al., 2009). Specimen orientation, mounting, and stiffness testing will be done as in Phase 1 (Fig.8.2). Then, a single continuous force rampup will be applied at 10 mm/min
until structural collapse (Chang et al., 2001; Khajavi et al., 2010; Sawaguchi et al., 1984; Shazar et al., 1998). Data analysis will be done for stiffness (i.e. slope of linear segment of the force-vs-displacement graph), failure force (i.e. peak force), failure displacement (i.e. displacement at peak force), failure energy (i.e. area under the curve up to the peak force), and relative interfragmentary motion (i.e. from image analysis at 50 N, 250 N, and failure). Statistical analysis will compare left-vs-right repair methods using paired student t tests with significance at p<0.05. The optimal repair method for TPW (and BC) fractures will statistically have the highest stiffness and failure parameters, and the lowest interfragmentary motion. If there is no statistical difference, then the surgical procedure that uses fewer implants and is technically easier to perform will be considered the optimal method.
References


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