PERFORMANCE CHARACTERISTICS OF
CUSTOM-DESIGNED POSTERIOR-STEP GLENOID PROSTHESES
IN TOTAL SHOULDER ARTHROPLASTY

A Dissertation in
Kinesiology
by
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ABSTRACT

Total shoulder arthroplasty (TSA) is the treatment of choice for advanced gleno-humeral osteoarthritis. Among different complications of the procedure, loosening of the glenoid component (the socket) is most common. The risk for glenoid component loosening is greatly increased with pre-existing bone erosion in the posterior glenoid, for which the treatment options are currently limited. The goal of the present project was to employ a cadaver model to evaluate the mechanical efficacy of two Posterior-step glenoid prostheses (Poly-step and Ti-step), two custom implants designed to compensate for a typical posterior glenoid defect. Our first hypothesis was that TSA using a standard (STD) glenoid prosthesis, in the absence of a defect, will not significantly alter joint stability, as measured by gleno-humeral translations, but will increase loading of the glenoid, as measured by the peri-glenoid principal bone strains. The second hypothesis was that TSA using custom Poly-step and Ti-step prostheses, to compensate for a biconcave posterior glenoid defect, will reverse the alterations in peri-glenoid strains and joint translations as caused by the defect. Furthermore, we also hypothesized that consequences of implantation of custom Poly-step and Ti-step prostheses in the presence of a defect will be similar to those following STD prosthesis implantation in the absence of such a defect.

Fifteen non-embalmed, fresh frozen human shoulder specimens were tested in a custom-built loading apparatus after fixing three triaxial strain gauge rosettes around the glenoid, and two marker-clusters on the acromion and the humerus. The arm was placed in 90° of abduction in neutral, in 30° of horizontal flexion and 30° of horizontal extension, while applying static forces to the tendons of the rotator cuff muscles, and axial forces to the distal humerus using a linear actuator. Peri-glenoid bone strains and marker-motion were recorded using analogue-to-digital converter and custom software, and a four-camera motion capture system respectively. Principal strains and joint translations were derived from the strains and marker-motion respectively, and were compared across conditions, arm positions and implant types. In five out of fifteen specimens, TSA was performed using a standard glenoid prosthesis while in the remaining ten specimens, a defect (20° biconcave type) was surgically created in the posterior glenoid, and custom Polyethylene-step and Titanium-step glenoid prostheses were implanted in five specimens each.

We observed that implantation of a STD glenoid prosthesis, in the absence of posterior glenoid defect, had no significant effect on glenoid bone loading or joint stability (p > 0.05). Creation of a 20° biconcave defect in the posterior glenoid caused some significant alterations in glenoid strains but not in joint translations, as compared to those in the native joints (p < 0.05). Implantation of a Polyethylene-step glenoid prosthesis in the presence of a defect, successfully reversed the mechanical alterations caused by the defect, while implantation of a Titanium-step prosthesis did not. Significantly greater anterior compressive strains and posterior tensile strains were observed following TSA using Ti-step prosthesis, and humeral head translations were also more anteriorly directed, as compared to the STD and Poly-step prostheses. Therefore, we concluded that Polyethylene-step glenoid prosthesis may be a viable option for treating posterior glenoid defects, and warrants further mechanical evaluation in the form of fatigue loading and testing for loosening etc. If successful, this prosthesis may provide an important treatment avenue for surgeons performing TSA on arthritic shoulders with posterior glenoid deficiency, and potentially reduce the incidence of glenoid component loosening and failure.
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<th>Abbreviation</th>
<th>Description</th>
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<tbody>
<tr>
<td>ACS</td>
<td>Anatomical Coordinate System</td>
</tr>
<tr>
<td>AP</td>
<td>Antero-posterior</td>
</tr>
<tr>
<td>CCS</td>
<td>Cluster Coordinate System</td>
</tr>
<tr>
<td>CT</td>
<td>Computerized Tomography</td>
</tr>
<tr>
<td>$\varepsilon_{\text{max}}$</td>
<td>Maximum (Tensile) Principal Strains</td>
</tr>
<tr>
<td>$\varepsilon_{\text{min}}$</td>
<td>Minimum (Compressive) Principal Strains</td>
</tr>
<tr>
<td>GCS</td>
<td>Global Coordinate System</td>
</tr>
<tr>
<td>GD</td>
<td>Glenoid Defect Condition</td>
</tr>
<tr>
<td>IS</td>
<td>Intact Specimen Condition</td>
</tr>
<tr>
<td>JR</td>
<td>Joint Replaced Condition</td>
</tr>
<tr>
<td>ML</td>
<td>Medio-lateral</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic Resonance Imaging</td>
</tr>
<tr>
<td>OA</td>
<td>Osteoarthritis</td>
</tr>
<tr>
<td>Poly-step</td>
<td>Polyethylene-step Glenoid Prosthesis</td>
</tr>
<tr>
<td>SI</td>
<td>Supero-inferior</td>
</tr>
<tr>
<td>STD</td>
<td>Standard or Conventional Glenoid Prosthesis</td>
</tr>
<tr>
<td>Ti-step</td>
<td>Titanium-step Glenoid Prosthesis</td>
</tr>
<tr>
<td>Angle $\theta$</td>
<td>Principal Strain Direction</td>
</tr>
<tr>
<td>TSA</td>
<td>Total Shoulder Arthroplasty</td>
</tr>
<tr>
<td>X ray</td>
<td>Roentgenogram</td>
</tr>
<tr>
<td>US</td>
<td>United States</td>
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CHAPTER 1
INTRODUCTION

1.1 Social Impact of Arthritis

In a National Hospital Discharge survey in 2004\textsuperscript{128}, orthopedic ailments resulted in 130 million annual patient visits in the United States (US), out of which 44.2 million were for arthritis and related conditions. Arthritis, being the most common cause of disability, has a tremendous impact on health care resources across the globe. Although typically associated with older age, arthritis is not uncommon even in younger population. While half of the US population over 65 has some form of arthritis, two thirds of all patients with doctor-diagnosed arthritis are younger than sixty five. In fact, arthritis has been found to be highly prevalent in the working age population in surveys conducted by the Centers for Disease Control and Prevention during 2003-2005. About 8.3\% (17.4 million) of patients in the working age group (18 to 64) reported significant limitation of activities in these surveys. It has been estimated that about 46.4 million adults over the age of 18 (roughly 1 in 5) are presently suffering from some form of arthritis in this country alone\textsuperscript{127}.

1.2 Osteoarthritis & Joint Replacement Surgeries

Also known as the degenerative or “wear-and-tear” type of arthritis, osteoarthritis (OA) is one of the most prevalent chronic ailments around the world. In a 2004 report by the National Center for Chronic Disease Prevention and Heath Promotion\textsuperscript{127}, about 27 million people in the US were affected with OA in one or more joints. OA is an irreversible and gradually progressive condition, characterized by cartilage degeneration, joint pain and restriction of movement. Etiology of idiopathic or primary OA is unclear; but genetic traits, advanced age, male gender, and obesity have been identified as risk factors. In contrast, secondary OA is often associated
with some form of trauma to the joint, which may be an acute injury such as an intra-articular fracture, or an aftermath of a disease process. Several arthritic conditions such as septic arthritis, gouty arthritis, neurogenic arthropathies, rheumatoid arthritis and other inflammatory arthropathies are known to subsequently lead to secondary OA. In both primary and secondary OA, the affected joint undergoes progressive and irreversible destruction that is mechanically driven and biochemically mediated.

Non-operative treatment including analgesic medications and physiotherapy is effective in early stages of OA; but with more advanced degeneration, joint replacement surgeries become the treatment of choice. Depending on the involvement, one or both of the articulating joint surfaces are replaced, and the procedures are called, \textit{“hemiarthroplasty”} and \textit{“total joint arthroplasty”} respectively. As reported by the US Bone and Joint Decade organization\cite{171}, the total joint replacement procedures performed in the US in 2004 included 454,652 knee, 232,857 hip and 41,934 shoulder replacements, and additional 12,055 procedures involving other joints. Among the different conditions necessitating these procedures, osteoarthritis was the most prevalent cause.

1.3 Background & Significance

1.3.1 Shoulder Anatomy

The shoulder or gleno-humeral joint is formed by a synovial articulation between the near-spherical head of the humerus and the relatively flat, pear-shaped glenoid face of the scapula (Figure 1-1). Although classified as a \textit{“ball and socket”} type joint similar to the hip, the two joints are very different regarding the mechanics, stability and ranges of motion. The glenohumeral joint is one of the most mobile joints of the human body, and it moves rhythmically and in unison with the other joints of the pectoral girdle, namely the acromio-clavicular,
sternoclavicular, coracoclavicular joints, and scapulothoracic “pseudo” joint, in order to execute smooth and effective motions of the upper extremity.

**Figure 1-1: Schematic illustration of the bone anatomy of shoulder**

While allowing multi-axial range of motion, the shoulder joint is quite deficient in intrinsic stability, and depends on several surrounding structures for strength. The joint capsule, ligaments and the surrounding musculature help to contain the humeral head inside the glenoid, while a fibrocartilaginous ring on the glenoid rim, named “glenoid labrum”, helps to deepen the socket and increase the joint surface area. The coracoid process, acromion, and the coracoacromial ligament form the **coracoacromial arch**, which serves as the superior boundary of the joint, while the superior, middle and inferior glenohumeral ligaments strengthen the joint capsule all around its perimeter. Furthermore, the subscapularis, supraspinatus, infraspinatus, and teres minor muscles – the so-called “**rotator cuff**” – exert compressive forces at the joint, and are particularly responsible for dynamic stability during movements. In spite of these mechanisms, the shoulder is highly susceptible to injuries such as rotator cuff tears, instability, fractures and dislocations, and such traumatic events increase the likelihood of osteoarthritis later in life.

### 1.3.2 Osteoarthritis of the Glenohumeral Joint

Osteoarthritis (OA) of the glenohumeral joint is less prevalent than that of the hip or knee (Nakagawa et al., 1999), but nevertheless, is a common ailment that accounts for 2 to 5% of all
cases of persistent shoulder pain (Meislin et al., 2005). It leads to substantial morbidity in the form of pain and limitation of movement. Primary OA is more common in women aged 60 and above, while secondary OA is typically seen in younger population. Degenerative changes can occur on both the humeral head and the glenoid to a varying degree. An early clinical sign of joint space reduction on radiographic images indicates cartilage wear, and as the condition progresses, the subchondral bone may undergo sclerosis, cyst and osteophyte formation, and erosion.

Opinions differ regarding the causal relationship between dislocations and osteoarthritis, and therefore the validity of the term *dislocation arthropathy*. However, more recent studies (Hovelius L and Saeboe, 2009; Marx et al., 2002; and Brophy et al., 2005) support the notion that repeated chondral damage due to chronic instability and recurrent dislocations increase the likelihood of OA. Furthermore, *capsulorrhaphy arthropathy*, an iatrogenic arthrosis, is believed to occur as a consequence of the imbalance of forces subsequent to excessive tightening on one side of the joint during surgical intervention for instability (Matsoukis et al., 2003; Hawkins et al., 1985; and Brems JJ., 1998). Likewise, imbalance of forces is also blamed in *cuff-tear arthropathy* subsequent to a large tear in a rotator cuff muscle, most commonly the supraspinatus. Abnormal joint loading occurs as the humeral head migrates superiorly through the torn supraspinatus tendon (Franklin et al., 1988). These diverse patho-mechanisms of OA are further discussed in the Literature Review section 2.2.

### 1.3.3 Shoulder Arthroplasty

Glenohumeral OA is estimated to be the cause for 71% of total shoulder replacement procedures (Jain et al., 2006), and the number of these procedures is rapidly increasing each year. In cases where degenerative changes manifest primarily on the humeral head with little or no glenoid involvement, *hemiarthroplasty*, a procedure in which the articular portion of the humeral
head is replaced with a metallic prosthesis, is the treatment of choice. However, in cases that involve considerable degeneration on the glenoid articular surface, total shoulder arthroplasty (TSA; Figure 1-2), a procedure involving replacement of the glenoid articular surface in addition to that of the humeral head, becomes necessary. Overall, TSA has shown to yield significantly better success rates as compared to hemiarthroplasty (Bryant et al., 2005; Edwards et al., 2003 and Radnay et al., 2007).

![Total Shoulder Arthroplasty System](image)

Figure 1-2: A modular unconstrained total shoulder arthroplasty system (Bigliani/Flatow®, Zimmer Inc.)

Major improvements in the operative techniques of TSA and implant designs have taken place over the last decade. Unconstrained TSA systems, based on the original design of Charles Neer (1974), have demonstrated the most successful long term results. Newer modular systems offer a wide range of sizes of different components, which allow the surgeon to customize a patient-specific appropriate combination at the operating table, and hence are preferred by many. Today, the most popular systems are of the unconstrained modular type, and consist of humeral ball and stem components, made of a biocompatible metal such as titanium or cobalt chrome alloy (vitallium), and a polyethylene glenoid component. The popularity of these systems has been justified by extensive clinical (Torchia et al., 1997; and Neer et al., 1982) as well as experimental evidence (Anglin et al., 2001; and Karduna et al., 1998).
1.3.4 Cost of Arthroplasty

Presently, the total cost of primary TSA to the patient is between $10,000 to 14,000 on average, while revision surgeries, warranted in the event of a complication of the primary procedure, are significantly more expensive. Preventing surgical complications has a huge impact on health care costs, for the patients on a personal level and for the community. Improvements in surgical techniques, combined with advances in metallurgy, implant manufacturing processes, and mechanical and other engineering sciences are simultaneously required to achieve superior functional outcomes and patient satisfaction.

1.3.5 Complications of Shoulder Arthroplasty

In spite of better clinical outcomes as compared to hemiarthroplasty, the TSA procedure is still young relative to total hip and total knee arthroplasty procedures. Rates of successful outcomes and complications of TSA vary widely across surgeons (Hammond et al., 2003) as well as across hospitals (Jain et al., 2006). Among a number of complications, failure of the glenoid component in the form of loosening and instability is most commonly reported (Cofield et al., 1990; Hasan et al., 2002; Deshmukh et al., 2005 and Franta et al., 2007). Indeed, the glenoid component fails more often than the humeral component.

Several factors play a critical role in implant survivorship, including the type and extent of the underlying disease, quality of the bone and rotator cuff, degree of any pre-existing bone loss, surgeon’s operative skill, the choice of implant and post-operative complications. Furthermore, technical issues such as the implant design, seating, alignment, and overall quality of implant fixation dictate the functional outcome (Iannotti et al, 2003 and Matsen et al., 2008). Of particular relevance to this work, is the pre-existing glenoid bone loss, which is believed to cause a threefold increase in the likelihood of component loosening (Haines et al., 2006).
1.3.6 Glenoid Bone Loss

Several patterns of glenoid erosion have been observed in different types of arthritis. While erosion is typically seen in the central region of the glenoid in rheumatoid arthritis, and in the superior region in cuff-tear arthropathy, it tends to occur in the posterior aspect of the glenoid in osteoarthritis. In addition to the usual degenerative findings such as cartilage wear, joint space narrowing, subchondral sclerosis, cysts and osteophytes, significant posterior glenoid bone loss was observed in about 60% and 48% of primary and secondary OA cases respectively (Levine et al., 1997). A prognostic classification proposed by Walch et al., (1998) describes different types of glenoid morphology as seen on computerized tomography (CT) imaging in primary glenohumeral OA (Figure 1-3).

<table>
<thead>
<tr>
<th>Type A: Centered humeral head</th>
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<tr>
<td>Minor concentric symmetrical erosion [43%]</td>
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<table>
<thead>
<tr>
<th>Type B: Posteriorly subluxated humeral head</th>
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<tr>
<td>Posterior joint space narrowing &amp; sclerosis [17%]</td>
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<table>
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<tr>
<th>Type C: Dysplastic glenoid with extreme (&gt; 25°) retroversion [9%]</th>
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Figure 1-3: Classification of the glenoid defects proposed by Walch et al. The incidence rates of the different types are indicated in brackets.

Type B2 defect appears biconcave on axial radiographic images, but is actually ellipsoidal or oblique wedge-shaped in three dimensions (Habermeyer et al., 2006). Glenoid bone loss causes joint instability, often with frank posterior subluxation (Bigliani et al., 1995). Glenoid
bone loss also gives rise to technical problems during TSA. Most notably, accurate assessment of the center and orientation of the glenoid articular surface (glenoid version) becomes difficult, and this may lead to anterior cortex perforation, component mal-alignment and mal-positioning (Neer et al., 1982). Improperly implanted prostheses are believed to result in asymmetrical joint loading, which may culminate into early component failure (Iannotti et al., 2005). Specific issues relating to glenoid bone loss and prosthesis loosening are discussed further in the Literature Review.

1.3.7 Critical Threshold of Posterior Glenoid Bone Loss

In a prior cadaver study in our laboratory, the significance of biconcave (type B2) posterior glenoid bone loss was quantified in terms of joint stability. Five cadaver shoulders were dynamically loaded in a custom built apparatus to simulate physiological in vivo joint loading. Humeral head translations with respect to the glenoid center in the antero-posterior (AP), supero-inferior (SI), and medio-lateral (ML) directions were used as dependent variables. Biconcave defects were created in the posterior glenoid in 5° increments up to total of 35°, and the specimens were repeatedly loaded with the arm in three different functional positions in the horizontal plane.

The shoulder joint became significantly unstable in the presence of 10° and 15° biconcave posterior glenoid defects with the arm in 90° of horizontal flexion and 90° of abduction respectively (Figure 1-4). Therefore, defects ≥ 10°, as measured from the glenoid center, were considered as critical threshold regarding the necessity for treatment intervention (section 1.3.7; Bryce et al., 2008).
Figure 1-4: Critical threshold of glenoid bone loss: Humeral head translations with respect to the glenoid center in the antero-posterior direction in 90° flexion. (Taken with permission from Bryce et al.)

1.3.8 Treatment Options for Glenoid Bone Loss

Guidelines for the treatment of glenoid bone loss have not been clearly established. Many surgeons prefer to neglect mild to moderate posterior glenoid bone loss during TSA, and accept a non-anatomical orientation of the glenoid component. A previously attempted method of buttressing the defect with bone cement has been discarded due to high incidence of cement fractures and failures (Rodosky et al., 1995; and Collins et al., 1992). Glenoid version can be restored to some degree by asymmetric reaming of the “high” anterior side (Neer and Morrison, 1988); however, normal healthy bone is sacrificed from the already bone-deficient glenoid, which may further compromise implant fixation. The method of exaggerating humeral component retroversion remains controversial (Spencer et al., 2005; Cofield and Edgerton, 1990; and Neer and Morrison, 1988). For moderate and severe bone loss, augmentation with bone grafts with or without internal fixation has had some success (Neer and Morrison, 1988). Nevertheless, it remains a technically challenging and time consuming method, and the success rates vary widely across surgeons (Steinmann and Cofield, 2000; Hill et al., 2001). Furthermore, graft resorption remains a concerning problem (Scalise and Iannotti, 2008). Lastly, custom-modified wedge-augmented prostheses have failed to show substantial benefits (Neer et al., 1982 and Rice et al., 2008). Different treatment options are discussed further in the Literature Review section 2.6.
1.4 Past Experimental Evidence

Normal and pathological shoulder mechanics have been explored by numerous researchers in the past, by means of animal, cadaver and computer models. The loading environment of the shoulder has been studied in cadaver specimens, by measuring bone strains under mechanical loading using strain gauges (Diop et al., 2005; Maurel et al., 2002; and Karduna et al., 1998) and photoelastic resin films (Pelletier et al., 2008). More recently, advanced finite element models have also been employed for this purpose (Farron et al., 2006; and Gupta et al., 2005). Furthermore, glenohumeral joint contact forces have been measured in cadaver experiments (Soslowsky et al., 1992) and in vivo by means of instrumented humeral stem prosthesis (Bergmann et al., 2007). Likewise, stability of the glenohumeral joint has been examined in terms of humeral head translations during passive movements (Harryman et al., 1990), and following arthroplasty in cadaver specimens (Karduna et al., 1997). In vivo joint translations have also been studied using an open Magnetic Resonance Imaging technique (Graichen et al., 2000). Some of these past experiments are discussed in more detail in the Literature Review.

1.5 Research Purpose

Experience has shown that a more complete understanding of in vivo joint mechanics is crucial to developing newer surgical techniques and more durable implants, and achieving better functional outcomes. The goal of arthroplasty is to restore normal joint mechanics as much as possible and thus, comparing joint kinetics and kinematics before and after TSA assumes critical importance. In many of the past cadaver studies, the soft tissues around the shoulder were removed and the humeral head was often simulated with a metallic ball. The current project was designed with two objectives in mind. The first objective was to develop a more physiological cadaver model, with intact capsule and rotator cuff muscles, in order to better understand the
loading environment of the native joint and following a conventional TSA. The second objective was to employ the model to evaluate performance characteristics of two custom designed Posterior-step glenoid prostheses, and assess whether they are a biomechanically and technically sound option for treating posterior glenoid bone loss. The prototypes of these prostheses were developed with a premise was that such glenoid components will reliably restore joint alignment while obliterating the need for eccentric reaming, and thus, preserving anterior glenoid bone in patients with posterior glenoid bone loss. This could possibly lead to improved clinical outcomes by reducing the incidence of component loosening, the need for revision procedures, and the associated health care costs.

1.5.1 Posterior-step Glenoid Prosthesis

The novelty of the Posterior-step design is the posterior step block intended to compensate for a 20° biconcave defect in the posterior glenoid. The design concept is similar to the augmentation blocks used in total knee arthroplasty for asymmetric tibial defects (Bartel et al., 1982; Brand et al., 1989; and Chen and Krackow, 1994). As compared to the posterior-augmented prosthesis of Rice et al., the proposed prosthesis does not have a sloping wedge, but rather includes a step block with a flat stable base in the posterior half, which may be a biomechanically superior construct that is less vulnerable to shear forces imposed at the shoulder joint. Two prototypes of the Posterior-step prostheses, namely the Polyethylene-step (Poly-step) and Titanium-step (Ti-step), were fabricated by attaching blocks of polyethylene and titanium onto the back of commercially available glenoid implants (Figure 1-5a). Both Poly-step and Ti-step types were manufactured in the commonly used sizes of 46 and 52mm (Figure 1-5b and c respectively). The step blocks were manufactured in accordance with the digitized dimensions of conventional glenoid prostheses, and dimensions of a 20° posterior glenoid defect. The designing process is described in further detail in the Methods section 3.9.
1.6 Hypotheses & Specific Aims

The present project consisted of three inter-related phases as described below:

**Phase I:** The objective of the first phase was to employ a physiological cadaver model, with intact joint capsule and rotator cuff muscles, to investigate shoulder loading and stability in native joints and following traditional TSA. Hypotheses were formulated after taking into consideration the results of past studies, whereby higher peri-glenoid strains were observed following conventional TSA (Maurel et al., 2002) while joint translations were unaltered (Karduna et al., 1997).

**Hypothesis 1:** In the absence of a posterior glenoid defect, TSA using a standard (STD) prosthesis will have no significant effect on joint stability, as measured by humeral head position and translations, but will increase loading at most locations of the glenoid bone, as measured by peri-glenoid principal strains.

**Specific Aim 1:** Record peri-glenoid bone strains and humeral head translations under physiological loading before and after TSA using a conventional prosthesis in five cadaver specimens.
Phase II: The objective of the second phase was to employ the model developed in phase I to assess the ability of the custom Posterior-step (Poly-step and Ti-step) glenoid components to restore normal joint mechanics in the presence of a biconcave glenoid defect.

Hypothesis 2: TSA using a custom Posterior-step component to compensate for a biconcave posterior glenoid defect will reverse the alterations in peri-glenoid strains and joint translations as caused by the defect.

Specific Aim 2: Record glenoid bone strains and joint translations under physiological loading in intact shoulder, and following implantation of a custom Poly-step implant in five specimens and a custom Ti-step implant in the remaining five specimens, in all of which a 20° biconcave posterior glenoid defect is surgically created prior to TSA.

Phase III: The objective of the third phase was to compare performance characteristics of the custom Poly-step and Ti-step glenoid prostheses in the presence of a biconcave glenoid defect with those of the conventional glenoid prosthesis in the absence of a defect.

Hypothesis 3: TSA using a Posterior-step component done in the presence of a biconcave glenoid defect will have similar effects as those after TSA using a conventional prosthesis in the absence of a defect.

Specific Aim 3: Calculate the glenoid bone strains and joint translations as a consequence of TSA in all fifteen specimens, relative to those for the respective native joint, in order to compare the mechanical performance of the three types of implants (STD, Poly-step and TS).
CHAPTER 2
LITERATURE REVIEW

2.1 Biomechanical Aspects of Shoulder Anatomy

2.1.1 Salient Features of the Humeral Head

The “ball and socket” type of articulation between the humeral head and the glenoid is often likened to a golf ball on a tee, mainly because the humeral head is round and the glenoid is flat and much smaller. While the central 80% of the humeral head is perfectly spherical, the peripheral 20% is non-spherical. The humeral head is only partly covered with hyaline cartilage (Iannotti, 1992). The articular surface of the head of the humerus, as calculated by trigonometry, constitutes an arc of 150°, its height (HH in Figure 2-1a) being equal to three fourth of its radius (RC) (Pearl and Volk, 1995).

![Figure 2-1: a) Illustration of the geometry of the proximal humerus, showing the center of rotation (O), radius (RC), head height (HH), and the head inclination angle (X) b) Superior view of the glenohumeral joint showing the glenoid and humeral retroversion (Modified from Shoulder and Elbow Arthroplasty Williams et al., Lippincott 2005)](image)

In the coronal plane, the head of the humerus forms an angle of 120° to 135° with the longitudinal axis of the humeral shaft, which is known as the head inclination angle, and is
measured on the AP view (X in Figure 2-1a). In the transverse plane, the humeral head is rotated backward relative to the inter-condylar axis of the elbow joint through about 25° to 45°. This is described as humeral retroversion or retrotorsion (Figure 2-1b), and has been measured in vivo and in cadaver bones using different methodologies (Randelli and Gambriolli, 1986; and Hertel et al., 2002). Furthermore, the center of rotation of the humeral head has medial and posterior offsets with respect to the central axis of the humeral shaft, in the coronal and transverse planes respectively (Robertson et al., 2000). Furthermore, natural variations in the population with high, standard and low offset morphotypes have been described (Hertel et al., 2002). Thus, large variability is observed in the three dimensional geometry of the proximal humerus across different individuals, and often even between the two sides of the body of the same individual.

2.1.2 Salient Features of the Glenoid

2.1.2.1 Geometry and Structure of the Articular Surface

The glenoid is a trumpet-shaped bony process that forms the medial angle of the scapula. The glenoid vault is comprised of a cortical shell with cancellous bone tissue inside. The articular surface of the glenoid is pear-shaped, measuring 39 mm and 29 mm in the SI and AP directions on average respectively (Iannotti, 1992). Anglin et al. measured the mean bone strength of the glenoid using in situ indentation methods to be 10.3 MPa (range, 6.7 to 17 MPa), while the mean elastic modulus was 99 MPa (range, 67 to 171 MPa). They observed the postero-superior glenoid region to be the strongest and the central column to be the weakest, and that the bone strength and modulus decreased significantly below the subchondral bone layer (Anglin et al., 1999). Frich et al. mapped the topographic strength distribution of the glenoid using an osteopenetrometer, and calculated the average strength of the proximal subchondral bone to be 66.9 MPa, which decreased by 25% and 70% at 1 mm and 2 mm depths respectively (Frich et al., 1997). These findings elucidate the importance of preserving the subchondral bone tissue during total shoulder
arthroplasty (TSA). Frich et al. observed strong anisotropy across the glenoid, and reported much higher elastic moduli (100 to 400 MPa) as compared to those reported by Anglin et al. Recently, Mimer et al. used combined indentation and micro-CT methods and reported the average modulus of the glenoid subchondral bone to be 119 to 234 MPa and mean strength to be 26 to 67 MPa (Mimer et al., 2008).

2.1.2.2 Orientation of the Glenoid Articular Surface

The glenoid articular surface projects laterally with a superior tilt, which is also called the “inclination”; at an angle 5° on average (range, -7 to 15.8; Churchill et al., 2001; Mallon et al., 1992). In the transverse plane, it is oriented slightly posteriorly, i.e. retroverted, relative to the plane of the scapula (Figure 2-1b). The glenoid version has been measured in anatomical specimens of dry scapulae, and in live subjects using axillary X-rays and CT scans. Significant variability naturally exists in glenoid version across different individuals. In addition, the different techniques have revealed inconsistent measurements. In vivo measurement of retroversion is typically done on axillary X-rays or transverse CT slices as follows (Figure 2-1b): A line is drawn from the midpoint of the glenoid fossa along the scapular plane to meet the medial border, and another line is drawn perpendicular to this line. The second line defines the neutral glenoid version, and thus, the version angle is measured between the “neutral version” line and the glenoid rim.

The glenoid version measurements by different authors are tabulated below (Table 1), with anteversion denoted positive and retroversion, negative. In a couple of anthropometric studies on dry scapulae, the mean glenoid version angles have been found to be -1.1° and -1.23°, (range -12° to 9.5°) (Das et al., 1966; and Churchill et al., 2001). Measurements using axillary X-rays of healthy subjects have ranged from -7.4° to +2° (Saha AK, 1971; and Cyprien et al., 1983). In contrast, version measurements based on CT images have ranged from -12° to +14°
The mean values of version in these studies were +2° and -3° respectively. However, in both these studies, the measurements were carried out on chest CT scans which were done for other reasons, and the scapular rotation was not controlled.

To add to the complexity, different regions of the glenoid have dissimilar version angles (Deutsch et al., 1985; Randelli and Gambrioli, 1986; Mullaji et al., 1994; and Schlemmer et al., 2002). For the upper, middle and lower parts of the glenoid, average version of -5°, -2° and -7° has been reported respectively (Randelli and Gambrioli, 1986), while in another study, -12.8° and -3.1° have been measured for the superior and inferior parts respectively (Schlemmer et al., 2002).

Table 1: Glenoid version angle measurements by different authors

<table>
<thead>
<tr>
<th>First Author</th>
<th>n</th>
<th>Method</th>
<th>Mean ±SD</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Das (1966)</td>
<td>50</td>
<td>Direct</td>
<td>-1.1°</td>
<td>-12° to +10°</td>
</tr>
<tr>
<td>Churchill (2001)</td>
<td>344</td>
<td>Direct</td>
<td>-1.23°</td>
<td>---</td>
</tr>
<tr>
<td>Saha (1971)</td>
<td>50</td>
<td>X-ray</td>
<td>-7.4°</td>
<td>---</td>
</tr>
<tr>
<td>Cyprien (1983)</td>
<td>50</td>
<td>X-ray</td>
<td>-8° ± 5.01°</td>
<td>---</td>
</tr>
<tr>
<td>Mallon (1992)</td>
<td>28</td>
<td>X-ray</td>
<td>-6° ± 4.2°</td>
<td>-12° to +7°</td>
</tr>
<tr>
<td>Mallon (1992)</td>
<td>32</td>
<td>CT</td>
<td>-6° ± 4.2°</td>
<td>-13 to +2°</td>
</tr>
<tr>
<td>Randelli (1986)</td>
<td>50</td>
<td>CT</td>
<td>superior part</td>
<td>-5°</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>middle part</td>
<td>-2°</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>inferior part</td>
<td>-7°</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>-2° to +15°</td>
<td></td>
</tr>
<tr>
<td>Mallon (1992)</td>
<td>63</td>
<td>CT</td>
<td>+2° ± 5°</td>
<td>-12° to +14°</td>
</tr>
<tr>
<td>Mullaji (1994)</td>
<td>19</td>
<td>CT</td>
<td>-3° ± 4.2°</td>
<td>-4.5° to +2.2</td>
</tr>
<tr>
<td>Schlemmer (2002)</td>
<td>30</td>
<td>CT</td>
<td>Upper part</td>
<td>-12.8° ± 6.4°</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Lower part</td>
<td>-3.1° ± 4.4°</td>
</tr>
<tr>
<td>Kwon (2005)</td>
<td>12</td>
<td>3D-CT Direct</td>
<td>3D-CT</td>
<td>-1.0 ± 5.4°</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Direct</td>
<td>-1.6 ± 5.5°</td>
</tr>
<tr>
<td>Scalise (2008)</td>
<td>14</td>
<td>2D-CT Direct</td>
<td>3D-CT</td>
<td>-7.0 ± 0.7°</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>3D CT</td>
<td>-7.1 ± 0.7°</td>
</tr>
<tr>
<td>Kwon (2005)</td>
<td>12</td>
<td>3D-CT Direct</td>
<td>3D-CT</td>
<td>0° to -14°</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>3D CT</td>
<td>-1° to -15°</td>
</tr>
<tr>
<td>Couteau (2000)</td>
<td>--</td>
<td>CT/FEA</td>
<td>-17°</td>
<td>-12° to -22°</td>
</tr>
</tbody>
</table>

Nyffeler et al. (2003) compared the accuracy of measurement using X-rays and CT scans in twenty five patients, and concluded that despite standardized techniques, axillary X-rays are
inadequate to accurately determine the glenoid version in patients, either preoperatively or postoperatively. They observed that slight variations in patient positioning or beam orientation caused significant errors on X-rays, while the measurements on CT scans were more accurate (Nyffeler et al., 2003). Three dimensional CT reconstructions have been found to yield even more accurate results than the conventional two dimensional CT scans, and are believed to be especially useful for preoperative evaluation in patients with glenoid bone loss (Scalise et al., 2008; and Kwon et al.; 2005).

Because of the peculiar glenohumeral geometry and orientations, the glenoid normally covers only 60% of the humeral head in the coronal plane, and 46% in the axial plane, and thus, comes in contact with about 28% of the humeral articular surface area (Jobe and Iannotti, 1995). The anatomical characteristics of the glenohumeral relationship have great implications on the patterns of bone erosion in arthritis, and on implant designs and techniques of TSA. The extreme inter-person variability observed in these anatomical characteristics increases the complexity of the issues related to TSA.

2.1.3 Shoulder Joint Stability

2.1.3.1 Static and Dynamic Stability Mechanisms

In spite of extreme anatomical variability, the center of rotation of the humeral head has been shown to remain fairly constant relative to the glenoid during majority of the normal shoulder motion (Poppen and Walker, 1976), except during extreme extension and external rotation (Howell et al., 1988). Several anatomical structures are known to contribute to the static and the dynamic stability of the shoulder; these include the joint capsule, superior, middle and inferior glenohumeral ligaments, the glenoid labrum, and the rotator cuff muscles. During the middle ranges of motion, all the ligaments are lax and the glenoid labrum is largely responsible for maintaining joint stability by resisting humeral head translation on the glenoid. Stability is
achieved as the glenoid cavity is deepened and the joint surface area is increased by the labrum; this mechanism is known as *concavity compression* (Matsen et al., 1991). The bony concavity measures approximately 2.5 mm without the labrum and 5.0 mm with intact labrum.

At the extremes of shoulder motion, the capsulo-ligamentous structures become tight and stabilize the joint; the mechanism is called *capsular constraint*. Between the superior, middle and inferior glenohumeral ligaments, the inferior ligament is the primary stabilizer. However, unlike the concavity compression, this mechanism of stability results in obligatory translation of the humeral head in the opposite direction.

Another important source of dynamic stability is the rotator cuff, comprised of the subscapularis, supraspinatus, infraspinatus, and teres minor muscles. These muscles are active during the entire range of motion, and are believed to contribute to both the dynamic and static stability in a number of ways (Seok-Beom Lee et al., 2000). They add to the concavity compression mechanism by exerting compressive forces between the humeral head and the glenoid. They also create a strap effect in certain arm positions, and directly restrain humeral head translation (Sharkey et al., 1995; Debski et al, 1999). Furthermore, they control most of the shoulder rotations, and by creating a force couple, enable efficient actions of other larger muscles, such as the deltoid and latissimus dorsi (Perry J., 1983; Jobe et al., 1983; and Sharkey et al., 1994). The ratio of total compressive forces to the shear forces has been considered to be critical. A number of other secondary factors are also believed to contribute to shoulder stability, but they are of less relevance to the present study.

### 2.1.3.2 Gleno-humeral Translations

The humeral head normally undergoes translations in AP and SI directions during terminal shoulder movements. Howell et al. observed that the humeral head translated about four millimeters when the arm was in a cocked up position of maximum extension and external
rotation (Howell et al., 1988). Anterior and posterior translations of the humeral head have also been observed with passive flexion and extension of normal cadaveric glenohumeral joints respectively (Harryman et al., 1990). Within normal limits, these translations indicate that the glenohumeral kinematics are more complex than pure ball-and-socket mechanism, while excessive translations would be indicative of joint instability. The mean AP and SI translations have been measured in natural cadaver joints to be 1.5 and 1.1 mm respectively (Karduna et al., 1997). In vivo measurements of shoulder translations using three dimensional post-processing of magnetic resonance images (MRI) have yielded similar results (Graichen et al., 2000). During passive elevation from 30° to 150°, the humeral head translated 1.22 mm inferiorly and 1.62 mm posteriorly, and was more centered in the AP direction between 60° and 120° of abduction, and in the SI direction, between 90° and 120°. These observations reinforce the role of capsular constraint during terminal motions, when the tight capsulo-ligamentous structures push against the proximal humerus causing it to translate towards the opposite side.

Glenohumeral kinematics are significantly altered by injury or disease (Poppen and Walker, 1976; McGlynn and Caspari, 1984; and Howell et al., 1988). Abnormal and excessive translations of the humeral head’s instant center have been observed in the presence of a large rotator cuff tear, which often leads to “cuff tear arthropathy”. In this condition, the humeral head migrates superiorly through the tear, and the stabilizing effect of the rotator cuff on the glenohumeral joint is greatly reduced (Neer et al., 1983). Interestingly, increased glenoid inclination has been associated with higher superior translation of the humeral head, and greater risk for cuff-tears (Hughes et al., 2003; Wong et al., 2003; and Konrad et al., 2006). In cases of anterior instability, greater anterior translations have been observed (Howell et al., 1988), and likewise greater posterior translations have been observed in the scenario of posterior instability and glenoid bone loss (Shapiro et al., 2007; Nyffeler et al., 2006; and Bryce et al., 2007).
2.1.3.3 Balance Stability Angle and Ratio

To better explain the issues related to shoulder stability, a balance stability angle has been defined as the maximum angle that the resultant vector of net joint reaction force (JRF) can make with the glenoid center line before dislocation occurs (θ in Figure 2-2a). The stability angle varies around the glenoid, being smaller in the AP direction than in the SI direction (Figure 2-1b). Shoulder stability at any given time is greatly dependent on the ratio of the displacing and compressive forces acting on it; this is known as the stability ratio. These variables are important with regards to prosthesis designing, and evaluation of their mechanical performance.

Figure 2-2: Illustrations showing a) balance stability angle and ratio calculations and b) balance stability angle around the glenoid. Taken from Matsen FA III Lippitt SB, Sidles JA, Harryman DT II: Practical Evaluation and Management of the Shoulder, Philadelphia WB Saunders, 1994

2.1.3.4 Conformity and Constraint

In the context of glenohumeral relationship, conformity and constraint are two important concepts. Conformity refers to the difference between the radii of the humeral head and the glenoid, while constraint refers to the threshold to dislocation (Figure 2-3).
The radius of glenoid curvature in the coronal plane has been measured to be 2.3 mm greater than that of the humeral head (Iannotti, 1992). McPherson et al. calculated the mean conformity index of 0.72 in the coronal and 0.63 in the sagittal plane using high resolution roentgenograms in ninety three cadaveric shoulders. They observed that only 16% of the specimens had a conformity index greater than 0.9, while vast majority had a more curved humeral head and flatter glenoid. In contrast, Soslowsky et al. used stereophotogrammetry and reported that the conformity of native shoulder joint was within 3 mm of congruence, with deviations from sphericity of less than 1%. They opined that perceptions of glenohumeral incongruity obtained from roentgenographic measurements where the glenoid appears flatter than the corresponding humeral head may be explained by cartilage thickness, and that the actual articulating surfaces do conform well when the articular cartilage surfaces are analyzed (Soslowsky et al., 1992).

Theoretically, greater conformity should increase the contact area and stability, and decrease contact pressures, leading to reduced implant wear and loosening. Upon testing different prosthetic systems for subluxation resistance and fixation strength of the glenoid component, Fukuda et al., observed that subluxation strength was positively correlated with the glenoid curvature (Fukuda et al., 1988). Thus, higher conformity indeed leads to superior
stability; nevertheless, totally conforming designs may restrain translations to a greater degree than the natural joint. Karduna et al. tested different combinations of radial mismatch between the glenohumeral components in a cadaver shoulder model under simulated physiological loading. They observed the ranges of joint translations to be inversely proportional to joint conformity. The lack of coupled physiological translations, and increased stresses at the implant-bone interface in the presence of greater conformity have been blamed for eccentric loading at the glenoid rim (Harryman et al., 1995; Karduna et al., 1997 and Walch et al., 2002). Thus, excessive conformity is believed to be associated with glenoid component loosening. On the other hand, active translations of natural joints were best reproduced by the reconstructed joints with less conforming articulations. A radial mismatch of 4 mm was found to produce 1 to 2 mm translations, similar to those of native joints (Karduna et al., 1997). Furthermore in postoperative radiographic comparisons, a radial mismatch of 6 and 10 mm had fewer incidences of radiolucent lines, which are indicative of component loosening (Walch et al., 2002).

Recent studies on retrieved implants have also supported the notion that the conforming systems are more prone for loosening. Conforming implants exhibited more abrasion and delamination, while the nonconforming systems showed more damage from burnishing and a propensity for edge deformation in the posterior quadrant (Nho et al., 2008). Furthermore, impingement at the anterior and inferior part of the rim was seen with conforming glenoids, whereas a see-saw effect was exhibited by nonconforming glenoids that were implanted in retroversion. Both rim impingement and retroversion are believed to cause eccentric loading and loosening.

In summary, there seems to be a trade-off between optimal joint motion and stability, and between uniform stress distribution and eccentric loading. Patterns of translation seem to influence implant stability, wear and longevity. In this context, conformity index (humeral head radius/ glenoid radius) and constraint index (arc of enclosure/ 360) are important variables in
prosthesis design (McPherson et al., 1997). Present commercial prosthetic systems exhibit several variations of combinations of glenohumeral radial mismatch ranging from 0 to 10 mm; however, a 2 to 3 mm mismatch is most commonly preferred. Recently, a novel glenoid design with different radii of curvature for the center and the periphery has been introduced to optimize conformity and constraint, and reduce rim loading and glenoid wear (Wang et al., 2005; and Barwood et al., 2004).

2.1.4 Shoulder Joint Loading

2.1.4.1 Glenohumeral Contact Area

As mentioned earlier, the glenoid normally covers only 60% of the humeral head in the coronal plane, and 46% in the axial plane. Thus, only about 28% of the humeral articular surface comes in contact with the glenoid at any given position (Jobe and Iannotti, 1995). In contrast, the contact is uniformly distributed over almost the entire articular surface of the glenoid. The contact areas, as determined by stereophotogrammetry in loaded cadaver specimens, increased steadily during abduction to an average of 5.07 cm² at 120°, while slight internal and external rotations caused significant alterations (Soslowsky et al., 1992). In 2007, Shapiro et al. recorded contact areas on a pressure sensitive film in cadaver specimens, as loaded under simulated rotator cuff and humero-thoracic muscle activity. They reported that contact area decreased following TSA as compared to the natural joint regardless of the orientation of the glenoid component, but was smaller for the 15° retroverted component than those implanted in neutral version (Shapiro et al., 2007).

2.1.4.2 Kinetics of the Native Shoulder Joint

It is important to fully understand the loading environment of the native shoulder joint in order to comprehend the effects of arthroplasty. In spite of being non-weight-bearing, the shoulder joint encounters considerable forces during daily activities. Normal physiological JRFs
are a consequence of the activity of the shoulder muscles for counter-balancing the mass moment of the upper extremity (Apreleva et al., 2000). Therefore, the joint forces are significantly affected by the magnitude of contraction and the moment arm length of individual muscles, as well as the orientation of the upper extremity in space. The average weight of the upper extremity is only 5% of the body weight, i.e. 3.5 kg in a 70 kg individual (Dempster et al., 1955), but substantial torques are generated because of a long lever arm. Consequently, the muscles have to exert large amounts of forces as their respective moment arms are much shorter.

2.1.4.3 Peri-glenoid Bone Strains

Bone strains are a good measure of the loading environment of a joint and force redistribution following arthroplasty. Past strain measurements in the glenoid cortical bone in cadaver specimens have improved our understanding of shoulder loading patterns. Higher peripheral cortical bone strains have been measured following arthroplasty as compared to the native shoulder joints. The measurements were done using strain gauge rosettes placed around the glenoid in cadaver specimens (Maurel et al., 2002; Diop et al., 2005), and photoelastic resin films (Pelletier et al., 2008). Maximum strains were recorded at the anterior and antero-superior aspects of the glenoid between 60° and 120° of abduction following TSA (Maurel et al., 2002). These findings corroborate with the significantly increased contact forces following TSA (Shapiro et al., 2007).

Eccentric loading is believed to cause glenoid component loosen ing, and in cadaver experiments simulating eccentric loading, cyclic patterns of strains have been observed at the implant-keel (Karduna et al., 1998). In the same context, the shear stresses at the component anchorage are thought to cause implant toggling and loosening (Cofield R.H., 1984; and Karduna et al., 1998). Furthermore, alterations in the stress/strain environment of the bone surrounding an implant are believed to stimulate bone remodeling, a phenomenon known as “stress-shielding and
Abnormal bone remodeling often results in bone loss which is in turn blamed for component loosening and failure (Pelletier et al., 2008; and Nagels et al., 2003).

2.1.4.4 Glenohumeral Joint Forces

As mentioned earlier, muscle contraction generates significant forces at the shoulder in order to counterbalance the mass moment of the upper limb (Apreleva et al., 2000). Integrity of the rotator cuff has been shown to significantly affect joint contact forces. Specifically, a transverse couple is generated by the action of subscapularis on the anterior side and the infraspinatus and teres minor on the posterior side of the joint (Parsons et al., 2002). Glenohumeral contact forces have been calculated using analytical methods on two and three dimensional musculoskeletal models in the past. The peak resultant forces have been estimated to be 89% to 100% of body weight at 90° of abduction, while assuming the muscle forces to be directly proportional to the muscle area times the integrated electromyographic (EMG) signal. The maximum shear forces have been estimated to be about 42% body weight at 60° of abduction (Poppen and Walker, 1978). In another study, the joint forces were calculated to be 130-430% BW while lifting a 10 kg suitcase on the side, and 170% BW while walking with a cane (Anglin et al., 2000).

These calculations have been reinforced by direct measurements in dynamic cadaver experiments; the reaction forces increased throughout the motion, peaking at 90° of abduction (Apreleva et al., 2000). The joint forces have been shown to increase significantly following TSA in cadaver specimens (Shapiro et al., 2007). Furthermore, recent in vivo measurements of in patients, by means of instrumented humeral implants with telemetric data transmission capabilities, have yielded similar results. In these in vivo studies, the joint forces remained below 100% body weight for most activities of daily living, and approached 130% during terminal motions and 150% BW with maximal effort against resistance (Bergmann et al., 2007).
Abnormal and excessive joint forces are believed to play a critical role in causation of osteoarthritis (OA) in the native joints (Hawkins et al., 1990; Neer et al., 1982), as well as implant loosening following TSA (Franklin et al., 1988).

2.2 Glenohumeral Arthritis

Like any other synovial joint, the shoulder can be affected by several different types of arthritis including the mechanical types such as OA and cuff-ear arthropathy, and the non-mechanical types such as avascular necrosis, septic arthritis, metabolic diseases, rheumatoid arthritis and other inflammatory arthropathies. Even though these are common conditions, and often lead to secondary OA, they are outside the scope of this document, and only those with more mechanical origin are discussed in the following sections.

2.2.1 Primary Osteoarthritis

Idiopathic or primary glenohumeral OA accounts for 4.6% of all shoulder conditions, and is more common in women aged 60 and above (Nakagawa et al., 1999). The etiology of primary OA being unknown, it is believed to be multifactorial, and several theories involving genetic, mechanical and environmental factors have been put forth. Mechanical factors, such as imbalance of muscle forces, excessive or eccentric joint loading, and inadequate physiological response to normal loading as a result of age-related morphological changes in the musculoskeletal tissues, are believed to play a major role in causation of primary OA. The mechanical factors bear particular importance in case of shoulder because of its peculiar characteristics, such as the articular geometry, large range of motion, limited inherent stability, and reliance on several surrounding structures for stability.

The diagnosis of primary glenohumeral OA is made based on clinical signs and symptoms, including persistent shoulder pain and restricted motion, which occur in the absence of
a history of previous shoulder injury or disease. The symptoms often worsen with activity and improve after rest – a sign that indicates a mechanical etiology. The affected joint undergoes progressive and irreversible destruction over a prolonged period of time. Degenerative changes including cartilage wear, subchondral sclerosis, cyst and osteophyte formation, and bone erosion can occur on both the humeral head and the glenoid to varying degrees.

2.2.2 Secondary Osteoarthritis

As mentioned earlier, secondary or post-traumatic arthrosis follows identifiable traumatic events, such as intra-articular fractures, recurrent dislocations, chronic instability, or residual damage from non-traumatic events such as infections, vascular disruptions, inflammatory arthropathies and so forth.

2.2.2.1 Intra-articular Fractures

Three and four-part humeral head-splitting fractures and fracture-dislocations are notorious for causing glenohumeral OA. More than half of the patients with four-part humeral head fractures develop secondary glenohumeral OA. The articular damage may result from a combination of factors including direct cartilage injury, fracture malunion, subsequent capsular contracture, vascular disruption of the adjacent bone, and resultant joint instability.

2.2.2.2 Capsulorrhaphy Arthropathy

Opinions differ regarding the cause-and-effect relationship between dislocations and OA. In 1982, Neer et al. described a subset of patients with a history of prior surgical repair for instability, and coined the term capsulorrhaphy arthropathy for the iatrogenic arthrosis (Neer et al., 1982). Samilson and Prieto (1983) used the term dislocation arthropathy, but found no clear relationship between the number of dislocations and the presence or the severity of arthropathy. They blamed the delay in diagnosis and treatment for the higher risk of subsequent OA observed following posterior dislocations in their series (Samilson and Prieto, 1983). Zukermann et al.
reported on arthrosis following instability repair due to protruding staples and pins (Zukermann et al., 1984). Subsequently, excessive tightening of the capsulo-ligamentous structures on one side of the joint during instability repair has been blamed for the post-capsulorraphy shoulder arthrosis. Excessive capsular tightening is believed to cause obligate translations of the humeral head towards the less tight side, leading to an imbalance of joint forces. (Hawkins et al., 1990 and Brems JJ., 1998). This may occur after any of the commonly performed open or arthroscopic procedures including Bankart’s, Putti-Platt, Magnuson-stack and Bristow-Latarjet. Excessive anterior capsular tightening is shown to limit external rotation of the shoulder (Black et al., 1997; Gerber et al., 2003) and increase obligatory posterior humeral head translations, which may cause abnormal posterior joint loading and recurrent postero-inferior subluxations, often leading to secondary OA (Werner et al., 2004; and Ahmad et al., 2005). Particular attention to subtle subluxations, hyperlaxity and various instability characteristics, such as the nature, direction and extent, during preoperative evaluations are considered important in order to minimize the risk for arthropathy (Brems J.J., 1998).

2.2.2.3 Dislocation Arthropathy

It is a known fact that significant forces are involved in traumatic shoulder dislocations. In 1997, Taylor and Arciero reported their observations in patients who were arthroscopically evaluated within ten days of their first-time traumatic dislocation. Significant chondral and osteochondral lesions involving the humeral head were noted in the majority (fifty seven out of sixty three patients), often associated with casulo-labral disruptions and rotator cuff injuries (Taylor and Arciero, 1997, and Norlin R., 1993). In a retrospective case series of one hundred and sixty shoulders with chronic anterior instability, impaction fractures were identified in 73.1%, and osseous glenoid lesions in 78.8% of cases using radiographic images (Edwards et al., 2003). The risk of developing subsequent OA was observed to be ten to twenty times greater following a
shoulder dislocation in a case control study (Marx et al., 2002). These findings have been corroborated in a recent multi-center prospective study in which degenerative changes were observed in 9 to 29% of the total two hundred and fifty seven shoulders with previous dislocation. Specifically, 39% of patients with recurrence developed glenohumeral OA even in the absence of any surgical intervention (Hovelius L and Saeboe, 2009). Furthermore, the arthritis following non-operatively treated shoulder instability has been found to be indistinguishable from capsulorrhaphy arthropathy (Matsoukis et al., 2003). Specifically in younger adults, static posterior subluxations of the humeral head are believed to often go unrecognized, which may also increase the chances of subsequent arthritis (Walch et al., 2002). These recent studies support the notion that repeated chondral damage due to chronic instability and recurrent dislocations may be directly responsible for the ensuing arthropathy (Hovelius L and Saeboe, 2009; Marx et al., 2002; and Brems J.J., 1998).

2.2.3 Cuff-tear Arthropathy

In 1981, Neer et al. described advanced arthritic changes and rapid deterioration of the glenohumeral joint function in association with massive rotator cuff tears. The authors claimed the condition to be distinct from OA, and named it cuff-tear arthropathy (Neer et al., 1981 and 1983). They outlined a cascade of pathologic changes, starting with a large rotator cuff tear and followed by leaking of the synovial fluid, subacromial impingement, shoulder joint disuse, instability, and superior displacement of the humeral head through the torn supraspinatus tendon (Neer et al., 1982 and 1983). Superior migration of the humeral head leads to erosion of the acromion, acromio-clavicular joint, and the superior aspect of the glenoid. The erosion further deepens the glenoid socket which comes to resemble the acetabulum; and hence this process is also referred to as “acetabulization” of the glenoid. Likewise, the proximal end of the humerus erodes to resemble the femur, the so called “femorization” of the humerus. Increased glenoid
inclusion has been associated with more superior translation of the humeral head, and greater risk for cuff-tears (Hughes et al., 2003; Wong et al., 2003; and Konrad et al., 2006). Likewise, greater anteversion and retroversion are believed to be predictive of posterior and anterior cuff injuries respectively (Tetreault et al., 2004). The pathologic changes in cuff-tear arthropathy make any reconstruction extremely difficult, and thus, the failure rates of treatment are unusually high (Collins and Harryman, 1997).

2.3 Indications and Complications of Total Shoulder Arthroplasty

2.3.1 Indications and Success of TSA

As mentioned earlier, surgical treatment for OA is considered in case of inadequate response to analgesic medications and physiotherapy. In select younger patients with early stage arthritis, procedures such as arthroscopic debridement, synovectomy, capsular release and glenoidplasty may be appropriate. However, with progressive joint deterioration and disability, especially in older patients, either hemiarthroplasty or TSA may be indicated. Satisfactory short-term results of hemiarthroplasty have been reported in 86% of patients with a concentric glenoid, 63% for those with a non-concentric glenoid, and overall, in 74% of patients. However, these numbers declined on longer follow up, often necessitating conversion to TSA (Levin et al., 1997; Carroll et al., 2004; and Radnay et al., 2007). Several researchers have compared results of hemiarthroplasty and TSA based on the findings of physical examination, Radiographic evaluation, patient satisfaction and functional outcomes, using different scoring schemes. In addition to the technical parameters of success, some authors have suggested the level of patient satisfaction as an important indicator for interpreting the functional outcome (Franta et al., 2007; and Hasan et al, 2002), although this could be a very subjective measurement. Edwards et al. reported significantly better Constant scores for TSA as compared to hemiarthroplasty for pain relief, activity, mobility and strength. They observed 86% excellent or good results for
hemiarthroplasty, and 96% for TSA (Edwards et al., 2003). Specifically, the unconstrained type of TSA has consistently yielded significantly better functional outcomes as compared to hemiarthroplasty (Radnay et al., 2007; Bryant et al., 2005; and Edwards et al., 2003). In 1982, Neer et al. reported excellent to satisfactory results for unconstrained TSA in thirty nine out of forty patients with OA, and forty out of fifty with rheumatoid arthritis. In the review of one hundred and thirteen TSA procedures performed between 1975 and 1981 using the Neer-II system, Torchia et al. reported that complete pain relief was achieved in 86% of patients, while complications developed in fourteen patients requiring reoperation. They estimated the probability of survival of the unconstrained Neer prostheses to be 93% at ten years, and 87% at fifteen years (Torchia et al., 1997). On the other hand, hemiarthroplasty may be preferable in patients with irreparable rotator cuff tears, cuff-tear arthropathy, persistent instability, deficient glenoid bone stock, and osteonecrosis isolated to the humeral head (Franklin et al., 1988; Boyd et al., 1990; Rodosky et al., 1996; and Baumgarten et al., 2004). Today, the unconstrained TSA is routinely performed for different types of arthritis. The technical challenges, success rates and complications vary according to the indication.

2.3.2 Complications of TSA

Sophistication in the surgical procedures, aseptic techniques, implant designs and implant manufacturing in the recent years has drastically reduced the incidence of complications following joint replacement procedures. Nevertheless, the TSA procedure is not free from complications. The rates of successful outcomes and complications vary across surgeons (Martin et al., 2005; Torchia et al., 1997; Barrett et al., 1987; Cofield, 1984; and Neer et al., 1982); those performing higher numbers of TSA per year have lesser complication rates and shorter average hospital stays than surgeons who perform fewer procedures (Hammond et al., 2003). Likewise, fewer post-arthroplasty complications have been associated with the hospitals conducting greater
numbers of TSA procedures annually (Jain et al., 2006). In a meta-analysis study of all complications reported between 1995 and 2005, the overall incidence rate was found to be 14.7% (414 complicating events with 2810 procedures) (Bohsali et al., 2006). Most common complications include joint stiffness, instability, glenoid component loosening, periprosthetic fractures, rotator cuff tears, infection, nerve injury, deltoid muscle dysfunction, component malposition, and polyethylene wear (Bohsali et al., 2006; Hasan et al., 2002; Cofield R.H., 1994; Wirth and Rockwood, 1994). Certain preoperative factors have detrimental effects on the functional outcome of TSA, including humeral head subluxations, glenoid erosion and loss of passive range of motion, (Deutsch et al., 2007; Iannotti and Norris, 2003).

2.3.3 Failure of the Glenoid Component

Loosening and failure of the glenoid component is by far the most commonly reported complication of TSA, and it continues to diminish the overall success of the procedure. (Matsen et al., 2008; Deshmukh et al., 2005; Hasan et al., 2002; and Cofield R.H., 1990). The glenoid component fails more often than the humeral component, and its failure is observed in most cases of unsatisfactory functional outcomes (Franta et al., 2007). Revision surgery is warranted in about 7 to 12.5% of cases of symptomatic glenoid failure (Bohsali et al., 2006; and Wirth and Rockwood, 1996), and the same detrimental factors associated with TSA influence the outcomes of revision procedures as well, including glenoid bone loss, rotator cuff tears, instability and joint contractures (Cofield and Edgerton, 1990).

2.3.3.1 Incidence of Glenoid Component Loosening

In a study involving one hundred and thirty nine TSA patients who were unsatisfied with the procedure, Hasan et al. observed glenoid component loosening in 59% of the patients (Hasan et al., 2002). In another similar study, Franta et al. blamed glenoid loosening in 62.5% of their one hundred and thirty six unsatisfied patients. Patient dissatisfaction most commonly results
from pain and instability during movements. Clinical signs include gross displacement of the component and tiny radiolucent lines in the cement layer surrounding the component anchorage. The radiolucent lines can be seen on X-rays, CT scans (Yian et al., 2005), and also by roentgen stereo-photogrammetric analysis (Rahme et al., 2004; Nagels el al., 2002).

Radiolucent lines may be associated with symptomatic as well as asymptomatic glenoid component loosening, and these lines, if seen in the early postoperative period, are believed to result from interposition of fluid or clot, inadequate cement penetration, or lack of secure cement purchase in the surrounding cancellous bone, and thus are indicative of suboptimal cementing technique and component fixation (Nyffeler et al., 2006; Klepps et al., 2005; Mileti et al., 2004). Early radiolucent lines have shown a high probability of progression, often leading to component loosening (Boileau et al., 2002; Nagels et al., 2002). Newer pegged glenoid component designs, motorized glenoid reaming, vacuum cement mixing, and instrumented cement pressurization have significantly reduced the incidence of early radiolucent lines in the recent years (Barwood et al., 2008; Klepps et al., 2005; Norris and Lachiewicz, 1996).

\[2.3.3.2 \text{ Causes for Glenoid Component Failure}\]

Several researchers have been investigating the mechanisms of glenoid component failure. Common causes of prosthetic loosening include infection, inflammatory reaction to implant wear particles, inappropriate mechanical load, fatigue failure, implant motion, and hydrodynamic pressure (Bauer and Schils, 1999). The specific anatomical and mechanical characteristics of a given joint often dictate the failure mechanics of the respective implant. Investigations on retrieved glenoid components have indicated that component impingement and edge deformation secondary to eccentric forces applied on the glenoid rim are primarily responsible for the component failure (Nho et al., 2009). Polyethylene wear particles have been associated with osteolysis and aseptic loosening as well (Wirth et al., 1999). The glenoid component is thought to
fail due to an inability to replicate essential mechanical properties of the normal glenoid articular surface. It is challenging to achieve durable fixation that withstands repeated eccentric loading, while resisting wear and deformation (Matsen et al., 2008). Several factors such as the quality of glenoid bone, pre-existing glenoid bone loss, the integrity of the rotator cuff, eccentric joint loading, as well as a number of technical issues such as the surgeon’s operative skills, implant choice, prosthesis design, seating, alignment, adequacy of soft-tissue balancing, cementing technique, and the overall quality of fixation, appear to determine implant stability and survivorship (Collins et al., 1992; Matsen et al., 2008 and Iannotti et al, 2003).

Glenoid component failure can be better understood if segregated in terms of failure of the component itself, failure of component seating or fixation, failure of the glenoid bone, and failure to effectively manage eccentric loading (Matsen et al., 2008). Failure of the component may involve pitting, abrasion, surface injury producing diffuse wear particles, component fracture, or in case of a metal-backed implants, separation of polyethylene from the metal (Braman et al., 2006; Hertel and Ballmer, 2003; and Scarlat and Matsen, 2001). Incidentally, the metal-backed prostheses have been associated with higher rates of loosening than all-polyethylene prostheses (Martin et al., 2005; and Boileau et al., 2002).

Failure of the component seating may result from inadequate glenoid surface preparation (Anglin et al., 2001; and Collins et al., 1992). In this context, the method of surface preparation appears to have a significant effect, as motorized reaming has been found to result in lesser component displacement and deformation as compared to manual burring and curetting (Collins et al., 1992). Failure of the seating may also occur due to component malpositioning (Iannotti et al., 2003 and 2005; and Lazarus et al., 2002), and in this context, restoring appropriate glenoid version is believed to be critical to avoid glenoid component loosening and joint instability (Nyffeler et al., 2006; and Hopkins et al., 2004). On the other hand, no significant increases in component displacement and deformation have also been observed under eccentric loads with
substantial posterior bone deficiency (Collins et al., 1992). Lastly, failure of the component seating may result from bone deficiency due to erosion or dysplasia (Phipatanakul and Norris, 2006; Edwards et al., 2004; and Hill and Norris, 2001).

Failure of the primary fixation, as evidenced by early radiolucent lines around the prosthesis, may occur due to suboptimal cementing technique with interposition of fluid or clot and inadequate cement penetration into the surrounding cancellous bone (Boileau et al., 2002; Nagels et al., 2002).

Failure of the glenoid bone could result from bone resorption caused by micro-motion, infection, thermal injury induced by drilling, reaming and cementing, or osteolysis associated with immunological responses to polyethylene (Churchill et al., 2004; Berman et al., 1984).

Lastly, eccentric loading at the glenoid rim is believed to be an important mechanism of glenoid component failure. Eccentric loading of the glenoid component by the humeral prosthesis in the SI direction is likened to a rocking horse mechanism (Franklin et al., 1988). Rotator cuff insufficiency, malpositioning of the glenoid implant, and inadequate restoration of glenoid version are believed to play an important role in causing eccentric loading (Hopkins et al., 2004 and 2007; Collins et al., 1992; Iannotti and Norris, 2003). Thus, several factors seem to contribute to loosening of the glenoid component, and its etiology is not clearly understood.

2.4 Glenoid Bone in Osteoarthritis

2.4.1 Glenoid Version versus Bone Loss

There exists significant ambiguity regarding terms that define normal glenoid retroversion and pathological posterior glenoid bone loss; both are often referred to as “retroversion”. The posterior glenoid defects in OA have been referred to as “nonanatomic retroversion” (Pearl M, 2005; and Braman and Flatow, 2008) and “retroversion deformity” (Hebermeyer et al., 2006). In more classic terms, retroversion simply refers to the normal posterior angulation of the glenoid
articulate surface in the transverse plane, while excessive retroversion refers to a developmental deformity in adolescents and young adults (Brewer et al., 1986). Thus, it is important to note that the distinction between retroversion and bone loss is not always delineated in the literature, and both these terms are used interchangeably.

The angle of glenoid version (v in Figure 2-4a) is typically measured between the “neutral version” line and the glenoid rim as described earlier in section 2.1.2.2. However, the technique for measuring the degree of posterior bone loss or defect has not been clearly established. Hill and Norris extrapolated of the contour of the glenoid articular surface using X-rays or CT scans of the normal contralateral shoulder, and described the version of the defect separate from that of the glenoid articular surface (Hill and Norris, 2001). In this technique, a line is drawn connecting both edges of the defect in an axillary view X-ray or CT scan. The missing part of the glenoid fossa is recreated, and another line is drawn form the inside edge of the defect to the posterior glenoid rim. The angle between the two lines is called the angle of the defect (d) as depicted in Figure 2-4b. Although we did not use X-rays in the presented work, we defined the version of the defect in a manner similar to that proposed by Hill and Norris, and used a custom made jig with channels at different posterior angulations relative to the central axis of the glenoid. It should noted that this measurement technique may not work with other defects, and that in the case of biconcave (type B2) defects, the so-called “non-anatomic” retroversion angle (v in Figure 2-4a) will be different from the defect version angle (d in Figure 2-4b).

Figure 2-4: Axillary view of the glenohumeral joint showing a biconcave type B2 defect in the posterior glenoid, and techniques to measure a) the glenoid version angle. b) the defect version angle.
2.4.2 Glenoid Bone Loss in Osteoarthritis

Significant posterior glenoid bone loss has been observed in about 60% and 48% of primary and secondary OA cases respectively (Levine et al., 1997). As described in section 1.3.6, the morphological changes of the glenoid in primary OA have been classified based on the glenoid defects and location of the humeral head (Walch et al., 1998 and 1999). In this scheme, the osteoarthritic glenoids were classified into minor (A1) and major (A2) concentric glenoid erosions, posteriorly subluxated heads (B1), posteriorly subluxated heads and with biconcave defects (B2), and excessively retroverted dysplastic glenoids (C), with respective incidence rates being 42%, 16%, 17%, 15%, and 9%. Although, this scheme is a good reference for broad defect classification, it fails to provide practically useful treatment guidelines. Furthermore, only fair agreement was found among experienced shoulder surgeons when classifying arthritic shoulders using this classification scheme (Scalise et al., 2008).

Friedman et al. (1992) observed the mean glenoid version in patients suffering from severe glenohumeral arthritis, including OA, rheumatoid and gouty arthritis, to be -11° (range, +2° to -32°) as compared to 2° anteversion in their normal controls (range, +14° to -12°). In a similar study, Mullaji et al. (1994) reported osteoarthritic glenoids to be 5 to 8 mm wider in AP diameter and retroverted to -12.5° on average. The glenoid surface undergoes eccentric deformation in the AP and SI directions as a result of inferior glenoid wear and retroversion deformity in OA (Habermeyer et al., 2006).

Plain X-rays, two dimensional CT scans, and three dimensional CT reconstructions have been used to quantify glenoid defects. CT imaging has been recommended over X-rays for preoperative and postoperative assessment of glenoid version (Nyffeler et al., 2003; and Clavert et al., 2007). Furthermore, a reference point, known as the “bare spot” has been described to arthroscopically quantify the amount of glenoid bone loss (Burkhart et al., 2002). This point is
located roughly at the center of the spherical inferior half of the glenoid articular surface, where the cartilage is thinnest, and hence the name, *bare spot*. However, others have questioned usefulness of this reference point (*Aigner et al.*, 2004).

From the treatment standpoint, preoperative consideration of defect characteristics such as the location, containment, extent and depth is warranted. While the defect is in the central and the superior glenoid in rheumatoid arthritis and cuff-tear arthropathy respectively, it is seen in either the anterior or the posterior aspect of the glenoid in recurrent dislocations and capsulorrhaphy arthropathy, depending on the direction of the instability. In contrast, bone erosion is almost always seen in the posterior aspect of the glenoid in primary OA. *Contained* defects are surrounded by bone, and therefore cause little problems when packed with cancellous bone grafts during TSA. The *non-contained* or *peripheral* defects on the other hand are generally more difficult to treat. The extent of a defect refers to the area of the missing articular surface, while the depth refers to the greatest distance from the estimated articular surface to the base of the defect.

### 2.5 Problems due to the Posterior Glenoid Bone Loss

#### 2.5.1 Problems in the Native Joint

Posterior glenoid bone loss in OA creates a three dimensional deformity, characterized by inferior wear and retroversion deformity that are independent of each other (*Habermeyer et al.*, 2006). The bone deficiency is often associated with joint instability and posterior subluxation of the humeral head (*Walch et al.*, 2002), which has also been confirmed in cadaver experiments. Significantly greater posterior translations of the humeral head were observed for \( \geq 10^\circ \) posterior glenoid defects in cadaver shoulders under simulated physiological loading with the arm in flexion (*Bryce et al.*, 2008). Repeated subluxations often lead to increased laxity of the posterior soft tissues and contracture of the anterior soft tissue.
2.5.2 Problems during Arthroplasty

Posterior glenoid bone loss creates a number of technical problems during TSA which have the potential to detrimentally affect the outcome of the surgery. The support offered by the deficient bone stock is often inadequate for secure prosthetic fixation. During glenoid surface preparation, it is often difficult to assess the glenoid center and the version angle. Therefore, the risk of anterior cortex perforation and component malalignment is substantially increased (Figure 2-5a; Neer and Morrison, 1988).

![Anterior cortex perforation, and posterior subluxation due to inaccurate component orientation](image)

Furthermore, posterior soft tissue laxity and anterior contractures, associated with posterior subluxations, make the soft tissue balancing difficult. This is a significant limitation as inadequately balanced soft tissues tend to adversely alter joint mechanics, often leading to eccentric loading and early component loosening.

2.5.3 Problems after Arthroplasty

Levine et al. reported significant correlation of the extent of posterior glenoid wear with the functional outcome of shoulder hemiarthroplasty, with excellent or good results in 86% of patients with a concentric glenoid, and in only 63% with posterior glenoid wear (Levine et al., 1997). The pre-existing bone loss is believed to have a detrimental effect on the functional outcome of TSA as well (Iannotti and Norris, 2003), causing a threefold increase in the likelihood
of glenoid component loosening (Haines et al., 2006). Investigations on the cement mantle stress and micromotion at the bone-cement interface using finite element analysis (FEA) techniques have revealed that that excessive retroversion exceeding 10° significantly increases the cement mantle stresses, and therefore the risk of component loosening (Hopkins et al., 2004; and Farron et al., 2006).

Excessive glenoid retroversion increases posterior displacement of the humeral head, and results in eccentric loading of the posterior glenoid, leading to postoperative joint instability, implant wear and loosening (Shapiro et al., 2007; and Nyffeler et al., 2006). In addition, the associated soft tissue contractures and laxity further contributes to the imbalance of joint forces and altered kinematics following surgery. Specifically, glenoid retroversion of 15° has been shown to decrease joint contact area and increase contact pressures significantly (Shapiro et al., 2007). Even small changes in glenoid version can provoke displacement of the humeral head, and alter the magnitude and direction of the JRF in the horizontal plane (Nyffeler et al., 2006). Similar observations have been made using FEA techniques (Farron et al., 2006).

Redistribution of forces between the cortical and cancellous subchondral bone alters the stress/strain environment of the bone surrounding the implant, and is believed to result in bone resorption and implant loosening. Eccentric glenoid loading has been classically described in association with large rotator cuff tears. In this so-called “rocking-horse phenomenon”, asymmetrical loading of the glenoid component by the superiorly migrated humeral head in the presence of a supraspinatus tear, is believed to cause superior tipping and loosening of the glenoid component (Figure 2-6; Franklin et al., 1988).
Shifts of the instant center of rotation and off-center rocking movements during shoulder motion are believed to stress the component anchorage (Cofield R.H., 1984; and Karduna et al., 1998). Therefore, dynamic rocking testing of glenoid prosthesis under cyclic loads have been recommended for evaluating the implant’s vulnerability to loosening (Anglin et al., 2000 and 2001). Although the rocking-horse effect has been described in the context of loading in the SI direction, eccentric loading in the AP direction as a consequence of posterior glenoid bone loss appears to contribute to glenoid loosening as well. Recent studies on retrieved glenoid implants have indicated increased rim deformation, especially in the postero-inferior quadrant, for the prostheses that were implanted in retroversion (Nho et al., 2009). In addition, increased anterior radileucent lines were seen with retroverted implants, which suggested a component “see-saw” pattern due to antero-posterior rocking mechanism.

Thus, glenoid retroversion leads to decreased glenohumeral contact areas, increased contact pressures, and posterior eccentric loading, all of which increase stresses on the cement mantle and increase the risk of edge deformation and component loosening (Farron et al., 2006; Matsen et al., 20008; and Shapiro et al., 2007), and restoration of appropriate version of the glenoid component is considered to be critical (Nyffeler et al., 2006; Shapiro et al., 2007).
2.6 Current Treatment Options for Posterior Glenoid Bone Loss

Successful restoration of glenoid version in both the transverse and coronal planes is believed to potentially reduce the risk of eccentric loading and component loosening (Nyffeler et al., 2006; and Habermeyer et al., 2006). In addition, the importance of accurate component positioning, preservation of the deltoid function and restoration of the muscle length and tension relationships, also known as soft tissue balancing, cannot be overstated. Post-operatively, physician-directed rehabilitation is also critical for achieving functional recovery, and restoring normal shoulder mechanics.

Though smaller amounts of glenoid bone loss are routinely neglected during TSA, it has been advised that glenoid retroversion (or bone loss) exceeding 15° should be corrected (Shapiro et al., 2007; and Nyffeler et al., 2006). In recent cadaver experiments, the critical threshold for posterior bone loss, beyond which glenohumeral joint stability is significantly compensated, has been found to be 10° (Bryce et al., 2007). Since this angle was the version of biconcave posterior defects, as measured at the glenoid center (Figure 2-4b), the corresponding glenoid retroversion angle could be even smaller. During TSA, the goal is to achieve a glenoid orientation in either neutral version or slight retroversion; however, the guidelines regarding how to treat different types and degrees of glenoid bone loss are currently limited. As mentioned earlier, smaller defects are often neglected at the discretion of the surgeon. In cases of a massive irreparable defect, some surgeons prefer to replace the humeral head, and only ream the glenoid without any prosthetic implantation, the so called “ream-and-run” technique (Weldon et al., 2004; and Matsen et al., 2007). Rarely, biologic resurfacing with interpositional arthroplasty may be considered for select young patients with end-stage arthritis. However, for the majority of significant but repairable posterior glenoid defects, the treatment options can be conceptually divided into two categories, namely realignment and reconstruction.
2.6.1 Realignment Methods

2.6.1.1 Eccentric Reaming

The “non-anatomical” or excessive glenoid retroversion due to bone loss can be partially corrected by asymmetric reaming of the “high” side. In this technique, the anterior glenoid rim is preferentially reamed to level the glenoid articular surface (Figure 2-7). Asymmetrical reaming can compensate for bone deficiency and correct the version angle to some extent; however, the procedure removes some of the normal and healthy bone stock which is essential for secure component fixation. This method may not work if the glenoid size is too small and if there is not adequate bone tissue on the anterior side. Another problem with eccentric reaming is that it may fail to achieve congruity between the prepared surface and the back of the prosthesis, which defeats the purpose of concentric circular reaming. This may lead to suboptimal non-concentric fit leading to early loosening (Collins et al., 1992; and Matsen et al., 1996). Furthermore, because of the trumpet-shaped morphology of the glenoid vault, the articular surface becomes narrower and shallower with medialization. Therefore, reaming of the anterior glenoid beyond 1 cm is not advised as further bone removal would only compromise secure prosthetic implantation (Steinmann and Cofield, 2000; Boileau et al., 2002).

![Figure 2-7: Asymmetric reaming of the anterior “high” side in an attempt to level out glenoid articular surface. (Taken from Matsen et al., 2007)](image)

In direct contrast to the proponents of eccentric reaming, Clavert et al. have reported that ≥ 15° glenoid retroversion cannot be satisfactorily corrected simply by eccentric reaming,
especially when using a glenoid component with peripheral pegs (Clavert et al., 2007). They observed that fracture of the glenoid and penetration of peripheral pegs through the glenoid vault were the limiting factors, and advised that the surgeon should consider bone grafting for retroversion angles of $\geq 15^\circ$. In another study, Gillespie et al. reported that correction of as little as $10^\circ$ retroversion deformity by preferential anterior glenoid reaming results in significant narrowing of the AP glenoid diameter. While asymmetrical reaming for $> 10^\circ$ retroversion defects resulted in peg penetration in some of the specimens in their study, downsizing of the glenoid component became necessary for others. The authors concluded that successful correction by anterior reaming was possible in only half of the specimens with a $15^\circ$ deformity, and less than a quarter of the specimens with a $20^\circ$ retroversion deformity (Gillespie et al., 2009).

2.6.1.2 Compensatory Humeral Anteversion

In most descriptions of surgical techniques for TSA, placement of the humeral component in 30 to 45$^\circ$ of retroversion is recommended. In cases of posterior glenoid bone loss, especially with chronic posterior instability and subluxations, several authors have recommended placing the humeral component in slight anteversion to compensate for the excessive glenoid retroversion (Wirth and Rockwood, 1994; Moeckel et al., 1993; Cofield and Edgerton, 1990; and Neer and Morrison, 1988). However alterations of humeral component version failed to improve joint stability in recent cadaver studies (Nyffeler et al., 2006; and Spencer et al., 2005). Increased glenoid retroversion leads to creation of an off-axis moment and posteriorly directed shear forces at the glenoid articular surface. Therefore, it is suspected that any alteration in the humeral component orientation may not affect the line of action of the rotator cuff muscles, and the joint reaction or shear forces on the glenoid (Spencer et al., 2005).
2.6.2 Reconstruction Methods

As originally suggested by Neer and Morrison, the lost glenoid bone could be reconstructed using bone grafts or a custom-modified prosthesis (Neer and Morrison, 1988). Cement buttressing is no longer recommended because of unacceptably high breakage and failure rates (Rodosky et al., 1996).

2.6.2.1 Bone Graft Augmentation

Contained defects can be easily packed with cancellous bone grafts from the resected humeral head, while peripheral noncontained defects are much harder to treat; and posterior biconcave (type B2) defects, often seen in OA are of such type. Typically, eccentric reaming is first attempted and the glenoid surface is reassessed. If the surface is still excessively retroverted, corticocancellous graft can be fashioned from the resected humeral head, especially in primary TSA procedures, and fixed using a countersunk screw. In cases where the native bone is of poor quality, such as occurs in osteoporosis or avascular necrosis, and in cases of revision surgery, an iliac crest autograft or an allograft may be preferable.

![Figure 2-8: Augmentation of posterior glenoid defect using a wedge-shaped bone graft, internally fixed with a cancellous screw underneath a conventional glenoid prosthesis. (Taken from Bell et al., 2000).](image)

Bone-grafting has been recommended if glenoid retroversion exceeds 15° as determined on computerized tomography (Friedman et al., 1992), and cannot be restored with less than 1 cm of eccentric reaming. Successful bone grafting in patients with glenoid bone loss has been
reported in the past (Neer and Morrison, 1988); however, the success rates have been inconsistent in the literature (Hill and Norris, 2004; Steinmann and Cofield, 2000). In 1988, Neer and Morrison reported excellent results in sixteen out of nineteen patients; there were no occurrences of nonunion or loosening at 4.4 year follow up. In 2000, Steinmann and Cofield reported good to excellent results in twenty three out of twenty eight patients in their series. On the other hand, Hill and Norris found satisfactory results in only nine out of their seventeen patients. They blamed metal-backed prostheses and preoperative instability for the failures.

Bone grafting has been described as technically challenging by many. Given the limited operating space and the small size of the glenoid, fixing an unstable wedge-shaped piece of bone onto the curved defect surface can be tricky and time consuming, and there is often a high risk of cortex perforation and fracture while drilling screw holes. Furthermore, oversized grafts may result in lateralization and overstuffing of the joint. If bone is insufficient or too small to accommodate a screw, the graft is wedged inside the defect and held in place by non-absorbable sutures, which does not yield very solid fixation. Additional problems include dissolution, fracture, nonunion and loss of fixation of the graft; all of which lead to unsatisfactory results (Phipatanakul and Norris, 2006).

2.6.2.2 Custom Modified Glenoid Prostheses

While most patients with glenoid bone loss in the series presented by Neer and Morrison in 1988 were successfully treated with bone grafts, two patients were treated using a posteriorly augmented prosthesis (Figure 2-9a); the outcome in these patients was reported as satisfactory. However, no additional information or statistical analysis was provided. Furthermore, the authors stated that they prefer the support provided by bone over polyethylene, and therefore, have stopped using the augmented prosthesis. They recommended a standard metal-backed glenoid prosthesis for most patients and a standard polyethylene prosthesis if a deficient glenoid precludes
insertion of the wider keel of the metal-backed prosthesis (Neer and Morrison, 1988). Incidentally in subsequent studies, the metal-backed prostheses have been associated with unique problems such as separation of the metal from the polyethylene part, and diffuse metal wear particles (Hertel and Ballmer, 2003; and Scarlat and Matsen, 2001), and with significantly higher loosening rates as compared to all-polyethylene implants, are considered to yield inferior results (Martin et al., 2005; and Boileau et al., 2002; and Anglin et al., 2000).

Figure 2-9: Custom modified, posterior-augmented glenoid prostheses of a) Neer and Morrison (1999) and b) Rice et al. (2008)

Recently, Rice et al. reported on the performance of a wedge-augmented polyethylene glenoid component (Figure 2-9b) in thirteen patients at mean follow up of five years (range, two to eight years) (Rice et al., 2008). Their evaluation criteria included postoperative pain, patient satisfaction, peri-prosthetic radiolucent lines on X-rays, and range of motion, as assessed according to Neer’s classification. The authors reported excellent results in five, satisfactory in seven and unsatisfactory in two cases. Three patients had moderate (20-50%) posterior subluxation, and one had severe (>50%) anterior subluxation; however, none of their patients required a revision or any other operative procedure. The authors reported that they had a high degree of success in correcting posterior glenoid wear and subluxation using a combination of techniques including a posteriorly augmented prosthesis, soft-tissue balancing and release, and modular humeral heads. However, they concluded that the contribution of the custom implant seemed marginal, and even though no negative outcomes were observed, they discontinued using
their wedge-augmented prosthesis. Interestingly, these authors also did not provide additional details or statistical analyses about the custom implant’s efficacy.

2.7 Implant Studies

2.7.1 Studies on Retrieved Implants

2.7.1.1 Common Wear Patterns

Implants retrieved from patients during revision surgery have revealed a lot of information regarding joint mechanics in the past, leading to significant modifications in implant designs and surgical techniques. Retrieved polyethylene glenoid components have more often revealed substantial damage as compared to humeral components. The damage commonly involves rim erosions, surface irregularities, fractures and diffuse wear (Scarlat and Matsen, 2001; and Swieszkowski et al., 2003). Different combinations of wear patterns have been observed (Braman et al., 2006). Gunther et al. described a wear classification system for the shoulder, as adapted from previous hip and knee damage models. They reported surface abrasion, pitting and delamination as the most prevalent damage modes for the glenoid prostheses, and that a combination of abrasive surface wear and subsurface fatigue failure was a likely mechanism of glenoid component failure (Gunther et al., 2000).

In a more recent study, scratching, pitting, burnishing, and inferior rim deformation were reported to be the most common and the most severe types of polyethylene wear. The damage was observed to be focused in the inferior glenoid quadrant, which suggested a propensity for glenoid impingement by the humeral component, and edge deformation secondary to eccentric loading (Hertel and Ballmer, 2003; and Nho et al., 2009). Furthermore, Radiographic assessment was observed to severely underestimate the presence of clinical glenoid loosening, when correlated with the wear characteristics and the clinical signs and symptoms (Nho et al., 2009).
2.7.1.2 Wear Particle Size

Enzymatic digestion of polyethylene wear particles from the membranes around three failed glenoid prostheses have revealed particles with a mean equivalent circle diameter of 1.04 ± 0.03 micrometers. These particles were larger and more fibrillar and had vastly different characteristics from those isolated from failed hip prostheses. Therefore, joint mechanics are thought to influence the wear mechanism and the nature of the resulting debris. The particles generated by wear of ultra-high polyethylene glenoid implants are believed to be associated with osteolysis, its pathogenesis being similar to that of aseptic loosening seen after total hip replacement (Wirth et al., 1999).

2.7.1.3 Conformity and Wear

Conforming systems have been found to have a higher (3.2 ± 2.0) loosening score, as compared to nonconforming systems (2.4 ± 1.2) (Nho et al., 2008). Conforming systems also had greater abrasion and delamination scores and revealed impingement at the anterior and inferior parts of the glenoid rim. In contrast, nonconforming systems were associated with a greater burnishing score and a propensity for edge deformation in the posterior quadrant. Furthermore, the nonconforming glenoid exhibited a see-saw effect when implanted in retroversion. Though polyethylene wear is more common, a case of metallosis following an uncemented Nottingham shoulder arthroplasty has also been reported, where titanium porous coating separated from the humeral stem, and became embedded in the polyethylene glenoid component (Khan et al., 2008).

2.7.2 Prosthesis Designing and Material Testing

In the endless pursuit of ideal implants, newer alloys, techniques and implant designs are developed with the goal of reproducing the natural kinetics and kinematics of the respective joint. Better understanding of the geometry and structural properties of the articulating bones, and joint mechanics is critical in implant designing. Cadaver testing has provided insights into natural joint
mechanics, and alterations due to arthroplasty. Information from cadaver models is very useful in testing and designing newer implants. In case of the shoulder for example, the shifts of the instant center of rotation and off-center rocking movements during shoulder motion stress the component anchorage (Cofield R.H., 1984; and Karduna et al., 1998). Therefore, dynamic rocking testing of glenoid prosthesis under cyclic loads have been recommended for evaluating the implant’s vulnerability to loosening (Anglin et al., 2000 and 2001). These methods have been promptly adopted by the American Society for Testing and Materials as the standard for testing implant designs in terms of the susceptibility for loosening.

Along with cadaver and animal model studies, newer techniques such as computer modeling and wear studies have greatly advanced our understanding of mechanics of different joints. Computer models based on CT images especially when cross-validated with cadaver measurements provide useful information (Bryce et al., 2008; and Couteau et al., 2001). Likewise FEA-based models of implanted glenoids have advanced our knowledge regarding the loading environment of the shoulder, and provided important clues regarding implant designing and even some of the operative techniques. Glenohumeral contact points, articular pressure, bone and cement stresses, and implant micromotion, and effects of conformity have been studied using finite element models in the past. Specifically, conformity had only a slight effect at 0 degrees of retroversion, whereas all quantities increased by more than 200% and exceeded critical values above 10 mm of mismatch at 15° of retroversion. Superior joint restraints incorporated with an intension to prevent subluxation were found to significantly increase component stresses as compared to unconstrained components, while altered fin geometry of the implant keel was observed to better stabilize the component (Orr et al., 1988). Regarding operative technique, cement thickness between 1.0 and 1.5 mm was found to optimally avoid excessive cement fatigue and interface failure by means of FEA techniques (Terrier et al., 2005), and implantation of glenoid components in retroversion was shown to significantly increased the stresses within the
cement mantle and glenoid bone, and the micromotion at the bone-cement interface (Farron et al., 2006). FEA techniques have also been utilized to compare mechanical performance of different implants. For example, the stress-shielding effect, i.e. preferential stress distribution on to cortical bone and off-loading of the subchondral trabecular bone, has been observed with non-cemented metal-backed components (Stone et al., 1999; and Orr et al., 1988).

2.7.2.1 Total Shoulder Arthroplasty Systems

Several different total shoulder systems including, constrained, semi-constrained, unconstrained, and reverse ball-and-socket type, have been invented, and currently, there is a wide range of choice available for the surgeon. Modular systems offer different combination of sizes which allows the surgeon to customize an optimal component combination for the patient. Among the different commercially available types, the unconstrained systems based on the original anatomic designs of Neer et al. have consistently yielded better outcomes, clinically as well as experimentally. Subluxation resistance and glenoid component fixation strength of several prostheses (Cofield, Gristina Neer II) have been found to be sufficient against normal shoulder joint forces. Higher subluxation strength has been associated with larger glenoid curvature, and increased conformity (Fukuda et al., 1988). However, a glenohumeral radial mismatch of 2 to 4 mm is preferred in order to reduce the risk of rim loading and loosening. Between the pegged and keeled glenoid prostheses, biomechanical and animal work has shown favorable results for the pegged glenoid prostheses in neutrally oriented glenoids (Gartsman et al., 2005; Hopkins et al., 2004; and Lazarus et al., 2002). However, implants with peripheral pegs are suspected to be unsuitable for cases with severe posterior glenoid defects and retroversion (Clavert et al., 2007). Upon repeated loading through 100,000 cycles to assess the implant vulnerability to loosening, the curve-backed polyethylene glenoid component with threaded pegs was found to perform better than other implant types (Anglin et al., 2000 and 2001; Fukuda et al.,
Nevertheless, no system is free from the risk of glenoid component loosening and fracture under excessive loads (Fukuda et al., 1988).

### 2.7.2.2 Material Properties

Newer implant materials and designs are tested according to a number of protocols, as established by The American Society of Materials Testing. Cobalt chrome, titanium, stainless steel, and related alloys are most commonly employed in orthopedic prostheses due to their specific material properties such as strength, modulus, ductility etc. Biologic inertness in the form of low allergic, carcinogenic and inflammatory potentials is also critical. Polyethylene has a modulus closely approximating that of normal subchondral bone, while titanium has a much higher modulus (Appendix B). Materials also differ in terms of bony ingrowth and bonding with the surrounding bone, which are important properties regarding the durability of implant fixation. Processes such as cement coating, porous-coating, sintering, casting, forging and hot isostatic pressing improve implant fixation. Newer materials such as trabecular metal, ceramics and highly cross-linked polyethylene hold promise for future use (Stiehl J.B., 2005). The trabecular metal has a high strength-to-weight ratio, and its compressive strength and modulus are comparable to other prosthetic materials (Bobyn et al., 1999). Furthermore, because of its trabecular nature it allows increased bone and vascular ingrowth. Highly cross-linked polyethylene has increased wear resistance properties, possibly allowing a better compromise between nonconformity and wear; however, it may present with an increased risk of fracture because of its reduced toughness and resistance to crack propagation (Nho et al., 2008 and 2009). Further studies on these newer materials are warranted.
CHAPTER 3
METHODS

3.1 Specimen Preparation

3.1.1 Specimen Procurement

Fifteen non-embalmed, fresh frozen shoulder girdles were procured from human donors. Donated extremities were examined physically and under fluoroscopic guidance prior to selection, with the major exclusion criteria being joint hyperlaxity, major bony abnormalities and rotator cuff tears. Entire pectoral girdles were dissected out of the surrounding musculature through standard delto-pectoral approach. Care was taken to preserve the integrity of the glenohumeral joint capsule and rotator cuff muscles (supraspinatus, infraspinatus, teres minor and subscapularis). The scapula was separated by disarticulation at the scapulo-thoracic and acromio-clavicular joints, while the humerus was osteotomized 15 cm distal to the center of the humeral head. After confirming normal anatomy, the origins of the rotator cuff muscles were elevated from the scapula using a periosteal elevator while preserving their insertions on the humerus.

3.1.2 Specimen Mounting and Alignment

Scapulae of the procured specimens were fixed inside custom mounting pots using Dynalite plastic filler (Dynatron/ Bondo Corporation, GA). A linkage assembly attached to the pot base was sequentially adjusted in all three axes under fluoroscopic guidance to achieve specimen orientation closely simulating the in vivo pose at 90° of abduction (Figure 3-1a; Poppen et al., 1976; Perry et al., 1983, Flatow et al., 1993 and Halder et al., 2000). During this process, the blade of the scapula was first aligned at 10° of anterior tilt (Karduna et al., 2000; and McClure et al., 2001) which is measured as the angle between the medial scapular border and the vertical axis using a goniometer. Next under fluoroscopic guidance, the glenoid face was oriented
parallel to the edge of the pot base as seen from the axial view, and inclined at 25° to the vertical axis as seen in the antero-posterior view (Poppen et al., 1976 and Sharkey et al., 1994). Orientation of the glenoid parallel to the pot base was assumed to achieve “neutral version” (Freidman et al., 1992), while inclination, i.e. the upward angulation of the glenoid articular surface, was measured as the angle between a line drawn from the superior most and inferior most points of the glenoid rim and the vertical axis (Figure 3-1b and c). The 25° inclination angle was chosen while taking into consideration the normal inclination of the glenoid articular surface with respect to the scapula, and the rotation of the entire scapula at 90° of abduction (Poppen et al., 1976). The scapula pot was firmly fixed in place after each rotational alignment. Subsequently, a short intramedullary rod was fixed into the distal end of the humerus with two interlocking pins. Clusters of four reflective spherical markers of 10 mm diameter were rigidly anchored on the humeral shaft and on the acromion of the scapula using screws. Prepared specimens were kept frozen at -20°C until the night prior to the experiment.

Figure 3-1: a) Specimen potted inside the pot b) Specimen alignment in progress under the guidance of a fluoroscope c) Image on the screen showing the edge of the glenoid aligned at 25° to vertical.

3.2 Shoulder Loading Apparatus

After proper alignment, potted shoulders were fixed to a custom built loading apparatus consisting of a solid aluminum base and a rotating side bar with a sliding mechanism, on which a
linear actuator was secured. Attached on the other end of the apparatus base, was a system of cables and pulleys (Figure 3-2).

![Figure 3-2: A shoulder specimen being tested in the loading apparatus](image)

### 3.2.1 Linear Actuator

A linear actuator with a brushless servo motor (N2-BK23-S205-A6-MS1-MT1-C6, IDC/Danaher Motion) was fixed on the sliding mechanism, which was on top of the rotating side bar of the apparatus. This provision allowed application of axial forces to the humerus with the arm in lateral abduction, horizontal flexion and horizontal extension (Figure 3-3).

![Figure 3-3: Schematic illustration depicting the simulated arm positions](image)
Axial forces were generated along the longitudinal axis of the humerus by the actuator as specified by the operator. These forces were considered to be mechanically similar to the forces resulting from contraction of the deltoid muscle with the arm elevated at 90°. The specimen was connected to the actuator using a custom coupling device in series with a force transducer, which allowed real-time monitoring of the applied axial forces. The actuator was operated at a speed of 12 mm per minute until 200 N (~ 45 lbs) forces were recorded by the force transducer.

3.2.2 Cryogenic Clamps, and Cable-Pulley System

Forces generated by the rotator cuff muscles were statically reproduced by applying tension to their muscle bellies through cryogenic (freeze) clamps (Figure 3-4); custom made devices with built in channels for passage of liquid nitrogen. Muscle fibers are grasped in the inter-digitations of the serrated inner surface while pointed corner posts pierce the muscle belly for additional anchorage. Frozen clamp-muscle assemblies are able to transmit supra-physiological forces without slippage (Sharkey et al., 1995). Three cryogenic clamps were used to secure the supraspinatus, subscapularis and combined infraspinatus and teres minor. Static loads were applied through a system of cables and pulleys, which ensured that the direction of pull was consistent with the normal action of the respective muscle.

Figure 3-4: Cryogenic (freeze) clamp with an inlet and outlet for liquid nitrogen, serrated plates for grasping muscle belly, and a hook for cable attachment.
3.2.3 Estimation of Muscle Forces

Optimal forces for the deltoid and each of the rotator cuff muscles were estimated based upon established facts regarding rotator cuff action and previous experimental work (Sharkey et al., 1994 and 1995; Maughen et al., 1983; and Akai et al., 1968). Selected force magnitudes were well within the contractile capacities of the respective muscles. Other muscles acting at the shoulder namely, teres major, coracobrachialis, long head of biceps, latissimus dorsi, and pectoralis major were not modeled.

3.2.4 Simulation of Rotator Cuff Activity

For the purpose of simulation, the infraspinatus and teres minor muscles were considered as a single functional unit due to their analogous origins, insertions, fiber directions, and lines of action, while supraspinatus and subscapularis were considered individually. 30N of supraspinatus muscle force was simulated by applying 6.75 lbs weight to its muscle belly through the cable-and-pulley system, while 50N of subscapularis force and 50N of combined infraspinatus and teres minor force were simulated using 11.24 lbs weights. The loading protocol is schematically depicted in Figure 3-5.

![Figure 3-5: Schematic representation of the loading protocol indicating the simulated muscle forces and their respective lines of actions](image-url)
3.3 Motion Capture System

A three-dimensional motion capture system (Eagle, Motion Analysis Corporation, CA) was used to track the relative motion between the marker clusters that were fixed on the humeral shaft and the acromion. A 4-camera configuration was established to capture a 1.5 m³ volume directly centered about the specimen. Cameras were connected through a signal conditioning box (Eagle Hub unit) to the host computer. The system was calibrated before data collection as discussed in section 3.6. Marker motions, and the axial forces as measured by the humeral force transducer, were collected by the computer at 100 Hz and stored in binary (C3D) format. The accuracy of the system was validated using a custom made device as described in section 0.

3.4 Strain Gauge System

Three 45° pre-wired triaxial strain gauges (model KFG-1-350-D17-11L3M3S, Kyowa, Japan; Figure 3-6), also known as “rosettes”, were used to record glenoid bone strains in the present study. Each rosette was comprised of three uniaxial strain gauges, stacked upon each other such that the middle gauge forms a 45° angle with the two gauges on either side.

![Figure 3-6: A triaxial strain gauge (rosette) consisting of three stacked uniaxial strain gauges at 45°](image)

The nine individual uniaxial strain gauges comprising the three rosettes were connected to nine separate channels of the signal conditioner box, while the force transducer was connected to a tenth channel. The signal conditioner box was in turn connected through an Analogue-to-Digital card to the host computer. Data were collected using custom written software in LabWindows, CVI (National Instruments Corp., TX), which provided a real-time display, and a
graphical user interface with control panels. Strains and axial forces were recorded as pre-tared voltages at 100 Hz.

3.5 Force Transducer

The humeral force transducer (S-beam LC101 model, Omega Engineering Inc., CT), provided real-time monitoring of the axial forces applied to the distal end of the humerus by the actuator. In addition to the real-time display unit, the transducer signals were also sent at 100 Hz to the two host computers recording marker motion and bone strains. Simultaneous collection enabled subsequent time-synchronization of the strain and motion data. The actuator was operated at the speed of 12 mm/minute until voltage corresponding to 200 N was reached.

3.6 Systems Calibration

The motion capture system and the force transducer were calibrated at the start of each experiment. Calibration of the motion capture system was performed by static and dynamic procedures according to the manufacturer recommended guidelines in order to establish the global coordinate system. A right-angled 4-marker frame was used for static calibration, and the wand technique for dynamic calibration. A typical reconstruction residual of 0.3 mm or less was considered acceptable. Data were recorded using the eVart software (Motion Analysis Corporation, CA). The force transducer was calibrated in compression and tension using known weights in order to calculate voltage-to-force conversion factor and determine the voltage corresponding to 200 N forces (Figure 3-7a and b respectively).
3.7 Specimen Set-up

Each specimen, having thawed overnight, was mounted in the apparatus following system calibration. The mounting pot, as previously aligned to reproduce the in vivo glenohumeral joint orientation, was rigidly fixed to the base of apparatus. An adjustable bracket provided additional support to the specimen mount. Next, the apparatus side bar and actuator assembly was aligned with the distal end of the humerus which was elevated into 90° of abduction (Figure 3-2). This was achieved by adjusting the proximal end of the side bar in a plumb line directly under the center of the humeral head, and the height of the actuator in line with the humerus. The intramedullary rod in the distal end of the humerus was connected to the linear actuator through a coupling device and the force transducer, as mentioned earlier. Finally, the rotator cuff muscles were locked inside the respective cryogenic clamps while ensuring adequate purchase of the muscle mass, with the clamp orientated in the line of action of the respective muscle.

Strain gauge rosettes were attached around the glenoid after all the other set-up adjustments were made, in order to avoid accidental damage to the gauges. Sites for rosette attachment, on the superior, anterior and posterior aspects of the glenoid, were approximately 10 mm proximal to the joint line to allow optimal gauge seating and adequate strain measurement, while causing the least amount of damage to the peri-glenoid capsular reflections during site preparation (Figure 3-8).
Furthermore in ten specimens, the gauges needed to be positioned so as to avoid disruption when reproducing the posterior glenoid bone loss. The anterior rosette was positioned midway between the superior and inferior glenoid tubercles, while the posterior rosette was placed in the spino-glenoid notch at the base of the scapular spine, and the superior rosette, at the mid-point on the superior glenoid rim. An effort was made to keep the attachment sites consistent across all specimens. Selected sites were cleaned and roughened using a periosteum elevator and sandpaper to promote firm bonding. Acetone was applied to dehydrate the site after which rosettes were affixed on the bone using a fast-acting cyanoacrylate bonding solution (Zap adhesive, Pacer Technology, CA; (Figure 3-8).

The central gauge of each rosette was positioned perpendicular to the glenoid articular surface. Thus owing to the pre-established glenoid orientation, the central gauge of the superior rosette was directed in the scapular plane, in line with the humerus in 90° of abduction, while the anterior and posterior rosettes were aligned at a 25° angle from the vertical axis. Attaching the superior rosette was especially difficult because of narrow access to the area. Gauge wires were secured to the underlying bone over 5 to 8 cm in order to avoid accidental pull-outs. Gauges and the non-insulated portions of wires were sealed with an adhesive tape, and a coat of commercial acrylic solution for electrical insulation and protection from moisture.
3.8 Loading Trials

Prior to commencement of the loading trials, the clamp-muscle assemblies were frozen using liquid nitrogen. With the arm in the desired position (abduction, horizontal flexion or horizontal extension), the side bar was locked in place to prevent further rotation. Static loads were applied to the rotator cuff to properly seat the humeral head while allowing translation of the actuator attached to the distal end of the humerus. After confirming proper head seating and alignment, the sliding mechanism was locked. This was considered to be the starting point of the loading trial, and no axial translation or rotations of the assembly were possible in the locked position, and thus, the actuator motion was directly imparted to the humerus. Loading by the linear actuator was executed at a rate of 12 mm per minute, while recording bone strains, marker motion, and axial forces at 100 Hz. Loading was continued up to 200N of compressive force, as measured by the corresponding voltage value in the real-time display of the force transducer; the process typically took about 4 to 5 seconds. Marker motion was recorded in the global coordinate system as explained in the following section. Three loading trials were executed under each condition with the arm in the three simulated arm positions. In case of joint subluxation, which was a rare event, the loading was discontinued and the trial repeated from the start.

3.9 Digitization Trials

Motion of the two marker clusters was recorded in a right-handed global coordinate system (GCS), which was established during calibration protocol as outlined in section 3.6. Following capsule release, global coordinates of specific anatomical landmarks of the native joint were collected using a 3-marker wand; some of these are marked with stars in Figure 3-9.
Digitized humeral points included the superior most ("north pole") and inferior most ("south pole") points on the humeral head, a couple of points on the head circumference ("equator center" and "periphery" respectively), and the distal end of the humeral intramedullary rod with the arm held in lateral abduction. Points on the scapula included the glenoid center, and a peripheral point on the glenoid rim towards the right side of the apparatus (i.e. on either the anterior or the posterior glenoid rim for the right and left sided specimens respectively). Digitized landmarks were subsequently used for constructing the local or anatomical coordinate system (ACS), as based on the glenoid center, in order to derive the motion of the humeral head relative to the glenoid. Global-to-anatomical coordinate system transformation is described later in section 3.13.2. The entire digitization process was repeated for the corresponding points on the humeral and glenoid implant surfaces following TSA to establish a new coordinate system for the implanted joint surfaces.
3.10 Experimental Design

3.10.1 Specimen Groups

Fifteen specimens were randomly selected to receive a Standard (STD), Polyethylene-step (Poly-step) and a Titanium-step (Ti-step) implant, and were segregated into groups as such (Figure 3-10). Each specimen was repeatedly loaded under simulated rotator cuff and deltoid activity in a consistent manner.

![Figure 3-10: Superior views of different glenoid prostheses: a) Conventional prosthesis b) Polyethylene-step prosthesis c) Titanium-step prosthesis (Insets in b and c show the respective step blocks)](image)

3.10.2 Arm Positions

Each shoulder was loaded with the arm in 90° of lateral abduction in the coronal plane of the body, horizontal flexion and horizontal extension (Figure 3-3). The latter two positions were achieved by rotating the side bar and the actuator assembly as a single unit about a vertical axis of rotation running through the glenohumeral joint. The rotations were made through 30° anterior and 30° posterior to the coronal plane respectively using pre-set markings on the apparatus to easily switch between positions during the experiment.

3.10.3 Experimental Conditions

While conventional humeral stem and head prostheses were used for all fifteen specimens, the treatment for the glenoid was different across the groups. A standard glenoid prosthesis was implanted in five specimens belonging to the STD group. Contrarily a custom modified glenoid prosthesis with a posterior polyethylene step-block was implanted in five Poly-step group
specimens, while a similar custom prosthesis with a titanium step block was implanted in five Ti-step group specimens. A 20° biconcave defect was surgically created in the posterior glenoid prior to TSA in specimens of the latter two groups, in order to replicate the bone loss pattern (type B2) that is commonly encountered in glenohumeral osteoarthritis.

Thus for five specimens that were designated to receive a STD prosthesis, the experimental conditions were Intact Specimen (IS), and Joint Replaced (JR), reflecting the status of the joint, while for the remaining ten specimens, they were Intact Specimen (IS), Glenoid Defect (GD) and Joint Replaced (JR). Repeated loading trials were run under each condition, and at least two trials were cleanly collected in each of the three arm positions. The humerus was disconnected from the actuator only between the conditions to perform the operative measures, and was reconnected without disturbing the rest of the settings. The clamp-muscle junctions were frozen as required prior to resuming the trials.

3.10.3.1 Condition #1: Intact Specimen (IS)

Specimens with intact joint capsule were tested before any surgical intervention. Periglenoid strains and humeral head translations recorded in this condition served as the baseline for comparisons across groups, conditions and positions. Following data collection in the intact specimen, the joint capsule was accessed through the rotator cuff interval on the anterior side, between the superior edge of subscapularis and the anterior edge of supraspinatus, by combination of sharp and blunt dissection. The subscapularis muscle belly was reflected, and the capsule was incised on the anterior and inferior aspects from twelve to seven o’clock locations. The intracapsular portion of the long biceps tendon and the glenoid labrum were resected. After opening the capsule, pre-determined points on the articular surfaces of the glenoid and the humeral head were digitized using a three-marker wand. The collected data were subsequently
used for construction of the shoulder model and global to anatomical coordinate system transformations as described in section 3.13.2.1.

3.10.3.2 Condition #2: Glenoid Defect (GD)

In ten specimens that were designated to receive either a Poly-step or Ti-step prosthesis, a biconcave (type B2) defect was created by reaming over a guide wire that was inserted at 20° posterior angle to the central glenoid axis (Figure 3-11a). Repeated loading trials were conducted as per the protocol. This condition was omitted for the five specimens of the STD group. The process of defect size determination and creation are described below in detail.

Figure 3-11: A 20° biconcave defect created in the posterior glenoid

Subsequent to labrum removal in the Poly-step and Ti-step group specimens, a K wire was inserted at the glenoid center, perpendicular to the articular surface and was considered representative of the central glenoid axis. Another K wire was driven at a 20° angle to the central glenoid axis using angle guide. The central K wire and angle guide were removed, a 52 mm cannulated reamer was inserted over the angled K wire, and eccentric reaming of the posterior half of the glenoid was performed (Figure 3-16a). Given that 20° angle defect (defect version angle) was created using a 52 mm circular reamer, as measured from the glenoid center, the defect depth could be calculated from the reamer dimensions (Figure 3-12).
Figure 3-12: Schematic axial view of the glenoid with the reamer creating a posterior biconcave defect

The center of the reamer dome was level with the surface at the glenoid center during the reaming process. Defect depth (line \( d_1 = 1.56 \text{ mm} \)) was calculated as the perpendicular distance of the farthest point (point Q) from the glenoid center that the reamer could reach at 20° posterior inclination. Given the tapering geometry of the glenoid vault, the length of the defect base in the supero-inferior direction was slightly less than that of the glenoid, while the width of the defect base was slightly less than half of the glenoid width. The defect base was later flattened by removing a sliver of bone to achieve the final defect depth (\( d_2 \)) in order to match the geometry of the modified implant.

3.10.3.3 Condition #3: Joint Replaced (JR)

Superficial reaming was performed following labrum removal in STD group specimens, and after creation of the glenoid defect in Poly-step and Ti-step group specimens. First the glenoid quality, size and version were evaluated with the humerus retracted posteriorly. The center of the glenoid articular surface was estimated and marked. A Kuntschner (K) wire was driven at the glenoid center, perpendicular to the surface, into the underlying subchondral bone.
A cannulated circular reamer of appropriate size was inserted over the K wire. The articular surface was superficially reamed until cancellous bone was clearly visible (Figure 3-11b).

A modular unconstrained TSA (Bigliani/Flatow®, Zimmer Inc., IN) was performed in a consistent fashion by a senior shoulder surgeon from Hershey Medical Center in all specimens. As mentioned earlier, all specimens received conventional humeral head and stem prostheses, but the glenoid prosthesis differed across the three group (Figure 3-13). The prosthetic implantation was executed as described below. Digitization process was repeated following prosthetic implantation.

![Figure 3-13: Post-TSA pictures showing STD, Poly-step and Ti–step implants](image)

### 3.11 Prosthetic Implantation

#### 3.11.1 Humeral Prosthesis

Following capsulotomy, the soft tissues attached on the humeral head were reflected over, and the humeral shaft was positioned in adduction, extension and external rotation, in order to clearly visualize the anatomical neck. The head of the humerus was cut freehand, along the anatomical neck to match the inclination of the implant, while maintaining accurate retroversion using the rotator cuff interval as the reference. The medullary canal was opened in the superolateral aspect of the cut humeral surface with a bone gauge. Next, the canal was reamed using 7, 8 and 9 mm humeral reamers sequentially. Given that a short-stemmed humeral prosthesis was to be used, reaming was required over a short distance. Sizing of the resected humeral head was done by direct comparison with the selected head component (Figure 3-14).
A standard short-stemmed humeral stem prosthesis (10 mm × 130 mm × 8 mm) was implanted using the inserter/extractor tool. Accurate retroversion, as established by the rotator cuff interval, was carefully maintained during the process. Accuracy of component size-fitting and orientation were confirmed during trial insertion. Definitive implantation of the humeral stem and head prostheses was done together with the glenoid prosthesis subsequently.

### 3.11.2 STD Glenoid Prosthesis

A cylindrical hole for the central component peg was drilled at the glenoid center and perpendicular to the superficially reamed articular surface, using centering guide and straight driver. Holes for the two peripheral component pegs were drilled using a pegged drill guide (Figure 3-15), which ensured accurate vertical alignment of the holes as well as precise matching with the corresponding component pegs. Correct component seating, with the implant flush on the articular surface without overhang, was confirmed by visual inspection.
3.11.3 Poly-step and Titanium-step Glenoid Prostheses

The initial steps for glenoid preparation were the same as with the STD group specimens. Nevertheless, a 20° posterior glenoid defect was created following labrum removal in specimens belonging to the Poly-step and Ti-step groups. Subsequently, a custom Poly-step glenoid implant was implanted in five specimens, and custom Ti-step implant, in the remaining five specimens.

3.11.3.1 Posterior-Step Modification of Defect

The ellipsoidal (or oblique wedge) defect was surgically modified into a stepped pattern with horizontal and vertical surfaces in order to provide a flat and stable foothold for the modified prosthesis. Conversion of the ellipsoidal defect base into a flat surface was achieved by removing 3 mm sliver of bone (gray area of d₂ depth in Figure 3-12) off the defect base by means of specially engineered cutting jig, burr, and rotary tool (Dremel Inc., WI; Figure 3-18). Thus, 1.44 mm deep bone sliver was removed in addition to the defect depth of 1.56 mm (d₁), and this was done to standardize corrective surgery in the face of variedly sized defects. As a consequence of defect shape modification (Figure 3-16 b and c), step blocks of only two sizes were adequate to treat the defect regardless of the anatomical variations in the glenoid in different specimens. Implantation of the custom manufactured Poly-step glenoid prosthesis in five specimens and custom Ti-step prosthesis in another five specimens (Figure 3-16 d and e respectively) was carried out in a manner identical to that done in the STD group specimens, except that a thin layer of additional cement was also applied on the base of the defect.

Figure 3-16: Prepared biconcave posterior glenoid defect
3.11.4 Definitive Fixation

Humeral and glenoid surfaces were prepared for definitive implant fixation by cleaning any loose bone particles and debris (Figure 3-17 a). Polymethylmethacrylate bone cement of medium viscosity (Endurance, Depuy Inc.) was carefully injected in the medullary canal of the humerus, and peg holes in the glenoid (Figure 3-17 b).

![Figure 3-17: a) Humeral and glenoid surfaces as prepared for implantation of prostheses](image)

The humeral stem prosthesis was inserted and impacted home using the inserter/extractor tool, while ensuring accurate retroversion. The selected glenoid component was likewise impacted onto the prepared surface. Superfluous cement that extruded out during implantation was removed. The selected head component was press-fitted on the humeral stem, and the joint was reduced. Manual pressure was maintained until the cement consolidated and cooled.

3.12 Design of Posterior-step Glenoid Components and Instruments

The first task of the designing process involved determining the geometry of the typical biconcave posterior glenoid defect. Step blocks of compensatory geometry were then manufactured, and attached on to the back of the standard pegged polyethylene glenoid prosthesis. Furthermore, some additional tools were also custom manufactured for the defect modification procedure; these included a 20° angle guide, a cutting jig, a burr and a connecting piece. Designing process is described below in further detail.
3.12.1 Design of Posterior-step Burr and Cutting Jig

Special tools were custom manufactured for defect modification, since they are not commercially available (Figure 3-18). The cutting jig was based on the digitized dimensions of a standard prosthesis, but cut into half. A commercially available sliding device of appropriate size was mounted on top of the cutting jig using two screws. Design of the burr was based on those used for total knee arthroplasty, and was made as small as possible, with a stem to fit in to the rotary tool. Thus, two units were assembled, the first one comprised of the cutting jig and the attached sliding device, and the second one comprised of the rotary tool and the burr. Lastly, a connecting piece was manufactured to connect these two assemblies, which enabled movement of the burr along the cutting jig that was impacted in the component peg holes. Actual designs are provided in the appendix.

![Custom designed tools for defect modification procedure](image)

Figure 3-18: Custom designed tools for defect modification procedure

3.12.2 Determination of the Step-block Geometry

The step block was designed to compensate for up to $20^\circ$ posterior glenoid defect, with its dimensions being complimentary to those of the undersurface of a standard glenoid implant (Figure 3-19).
Thus, the bottom of the step block was flat, corresponding to the base of the surgically modified defect, while its top was ellipsoidal matching the undersurface of the posterior half of conventional glenoid prosthesis. The geometry of standard glenoid prosthesis is quite complex with different antero-posterior and medio-lateral curvatures. Therefore, the inside height at the center was the same as the depth of the defect ($d_1$), but the heights at all other regions were different. Curvatures of the top surface of the step-block matched those of the back of a standard prosthesis. The supero-inferior length ($D_{SI}$) of the step block matched that of the glenoid component, while its antero-posterior length ($D_{AP}$) corresponded to the length of the posterior half of the glenoid component. Three half-slots for half component pegs were made on its inside length, while two through-and-through holes were drilled for attachment to the backside of a standard component.
3.12.3 Manufacturing and Attachment of the Step Blocks

Young’s moduli of different materials (Appendix B) and other technical issues were considered for selecting the material for the step-blocks. Polyethylene and titanium were selected, as these are commonly used materials in orthopedic implants. Furthermore polyethylene has a modulus closely approximating that of normal subchondral bone, while titanium has a much higher modulus. This dissimilarity was projected to modulate the magnitude and distribution of post-implantation bone strains. Step-blocks were made in two sizes to match the glenoid component sizes that were used (46 mm and 52 mm). Corresponding holes were drilled at the back of standard prostheses, just short of the articular surface, and the step-blocks were rigidly fixed with two self-tapping stainless steel screws (#0 size, 0.25 inches, sheet metal, and pan head screws) and a cyanoacrylate bonding solution (Figure 3-20; Zap adhesive, Pacer Technology, CA).

![Figure 3-20: Picture showing the polyethylene and titanium step blocks, and the manufactured Poly-step, and Ti-step prostheses](image)

3.13 Data Processing

Evart software was used for motion data collection and initial post-processing. Additional processing of the motion and strain data was done using custom written Matlab algorithms. Filtering of all the data was done using a Butterworth digital filter with a cut-off frequency of 1 Hz; this frequency was chosen based on Fourier analysis on sample data.
3.13.1 Synchronization of Motion and Strain Data

Load cell measurements of the applied axial forces were simultaneously recorded by the host computers of both motion capture and strain measurement systems at 100 Hz. Since no external trigger was used during data collection, the data sets from the two systems were time-synchronized based on the time frame of peak axial forces during post-processing.

3.13.2 Motion Data Processing

3.13.2.1 Construction of the Shoulder Model

A shoulder model was constructed based on the coordinates of specific glenoid and humeral points, as recorded in a right-handed global coordinate system (GCS = Xg-Yg-Zg) during digitization trials (Section 3.9). A line connecting the peripheral glenoid point (i.e. anterior peripheral point for right-side specimens, and posterior point for left-side specimens) to the glenoid center formed the X axis, while the line connecting the glenoid center and the distal humeral shaft point constituted the Z axis, and the cross product of the X and Z axes yielded the vertical Y axis. The center of the humeral head was calculated by fitting a sphere on the four digitized humeral head points and computing its radius using trigonometric formulae.

3.13.2.2 Global-to-Anatomical Coordinate System

Global to the Anatomical coordinate system (ACS) transformations were performed in order to ultimately derive the motion of the humeral head center with respect to the glenoid center. This was achieved by using established techniques of coordinate system transformation for moving body segments (Cappozzo et al., 1995; Figure 3-21a). In this process, dynamic intermediary local coordinate systems (CCS = Xc-Yc-Zc), as based on the humeral and scapular marker clusters, were computed for each time frame. The coordinates of the glenoid and humeral points (P), as recorded in the GCS (Equation 1), were transformed into the respective scapular
and humeral CCS using the transformation matrix $T_{CG}$ at each time frame (Equation 2), and the respective Global-to-anatomical coordinate system transformation matrices were calculated.

\[ [P_c] = [X_c, Y_c, Z_c] \] \hspace{1cm} \text{(Equation 1)}

\[ [P_c] = [T_{CG}]^*[P_G] \] \hspace{1cm} \text{(Equation 2)}

Subsequently, the coordinates of the humeral head center in the humeral CCS were transformed in the glenoid ACS using the transformation matrix $[T_{AC}]$ (Equation 3). These steps, executed for each time frame of the loading trials yielded the motion trajectory of the humeral head center with respect to the glenoid center in three dimensions. The exact same protocol was followed for the loading trials after TSA when the humeral head motion was calculated using new coordinate systems, as established for the altered joint surfaces.

\[ [P_d] = [T_{AC}]^*[P_c] \] \hspace{1cm} \text{(Equation 3)}

Figure 3-21: Schematic representations of a) the digitized anatomical landmarks in global coordinate system b) the local or anatomical coordinate system based on the glenoid center

Since the marker clusters were rigidly fixed to the respective bones, there was no motion of the bony landmarks with respect to the respective marker cluster, which allowed derivation of the coordinates of the anatomical landmarks in a consistent manner.
In addition, the absolute location of the humeral head center with respect to the glenoid center at 20N of pre-loads was also compared across different arm positions and experimental conditions, in order to detect premature humeral head subluxations. This was especially important in the presence of the simulated posterior glenoid defect.

3.13.2.3 Accuracy Validation

An experiment was conducted to verify the accuracy of the motion capture system, and to quantify the system’s error in determining marker locations. A device with two marker clusters was custom built to simulate our experimental model (Figure 3-22).

Figure 3-22: Experiment to validate accuracy of the motion capture system with the device in a) abduction b) flexion and c) extension

One marker cluster was fixed on a stationary base, another one was fixed on a translating piece in order to simulate the scapular and humeral marker clusters respectively. A central pin-joint allowed $30^\circ$ of rotation on both sides of neutral, without any distraction or compression, while a sliding piece on which the distal marker cluster was fixed allowed axial translations. The center of the top surfaces of the pin joint was digitized. The location of this point with respect to the local coordinate systems, as based on the centers of the humeral and scapular marker-clusters, was recorded in abduction, flexion and extension positions. Since, the pin-joint allowed no movement other than pure rotations, the difference of location of the center in the two local coordinate systems indicated the system’s error in estimating marker locations. Next, the same protocol was repeated with the distal marker piece was shifted by 51 mm. The average error was
1.55 mm when the humeral cluster was set 5 cm apart from the scapular cluster, and 1.65 mm when the inter-cluster distance was 10 cm; the overall average error was 1.6 mm.

### 3.13.3 Strain Data Processing

As mentioned earlier, strain data were recorded as tared voltages at 100 Hz during the loading trials. The maximum ($\varepsilon_{\text{max}}$) and minimum ($\varepsilon_{\text{min}}$) principal strains, and the direction of principal strain from the $\varepsilon_a$ axis (angle $\theta$) were derived from recorded strains using the manufacture-recommended formulae (Equation 4, 5 and 6). Since the maximum principal strains were mostly positive, and the minimum strains were mostly negative, they are hereafter described as tensile and compressive respectively for convenience. Data were baselined at 20N axial loading, interpolated, averaged and compared across positions, conditions and implant groups.

![Diagram showing the strain gauges arrangement in a rosette](image)

**Figure 3-23: Diagram showing the strain gauges arrangement in a rosette**

\[
\varepsilon_{\text{min}} = \frac{1}{2} \left[ \varepsilon_a + \varepsilon_c - \sqrt{2 \left( \varepsilon_a - \varepsilon_b \right)^2 + \left( \varepsilon_b - \varepsilon_c \right)^2} \right] 
\]  
(Equation 4)

\[
\varepsilon_{\text{max}} = \frac{1}{2} \left[ \varepsilon_a + \varepsilon_c + \sqrt{2 \left( \varepsilon_a - \varepsilon_b \right)^2 + \left( \varepsilon_b - \varepsilon_c \right)^2} \right] 
\]  
(Equation 5)

\[
\theta = \frac{1}{2} \tan^{-1} \left( \frac{2 \varepsilon_{b} - \varepsilon_{a} - \varepsilon_{c}}{\varepsilon_{a} - \varepsilon_{c}} \right) 
\]  
(Equation 6)

### 3.13.4 Data Interpolation and Averaging

Time synchronized strain and motion data were zeroed at 20N of axial loading, and were interpolated to 101 frames in order to enable comparisons across different conditions and
specimens. Out of three loading trials that were conducted under each scenario, data from the two cleanest trials were averaged. Increased noise was observed in strains during trials immediately after freezing of the clamps; hence, the first trial was often dropped. In other cases, minor technical issues warranted omission of a trial.

3.14 Statistical Analyses

Dependent variables included magnitudes and direction angles (θ) of the compressive ($\varepsilon_{\text{min}}$) and tensile ($\varepsilon_{\text{max}}$) principal strains, and humeral head translations at 200N loads with respect to those at 20N. The analysis was done in two phases using a general linear repeated measures ANOVA model followed by Tukey pairwise comparisons. SAS and Minitab statistical software were used for the analyses, and significance was set at 0.05 alpha.

The first phase involved evaluating the alterations in strains and translations as a consequence of TSA, which was done separately for the three different types of implants (STD, Poly-step and TS). The mechanical consequences of a glenoid defect were also analyzed for the Poly-step and Ti-step groups. Thus the variables were compared across the Intact Specimen (IS) and Joint Replaced (JR) conditions for all specimens. In addition, variables in the Glenoid Defect (GD) condition, for the five Poly-step and five Ti-step group specimens, were also compared with the IS and JR conditions. Independent variables included Specimen as a random factor, and Arm Position and Condition as fixed factors. The findings of the first phase of analysis served as the baseline for the second phase, which involved comparing the variables of interest across the three implant groups in order to assess and compare their mechanical performance under loading. The relative differences in the variables between the Intact Specimen and Joint Replaced conditions were compared across the implant groups. Thus, Specimen was a random factor, while Arm Position, Condition and Implant were fixed factors for the second phase.
CHAPTER 4
RESULTS

The results are presented in two sections. In the first section, data from the five specimens belonging to each of the three implant groups (STD, Poly-step, and TS) are presented separately, while in the second section, the alterations in the variables as a consequence of total shoulder arthroplasty (TSA) are compared across the implant types. The variables, as analyzed at peak axial loads of 200N, included the magnitudes and directions of the principal strains, and the humeral head translations. The maximum ($\varepsilon_{\text{max}}$) and minimum ($\varepsilon_{\text{min}}$) principal strains are described as tensile and compressive respectively here onwards.

4.1 Section I: Data for Individual Prosthesis Types

4.1.1 Standard (STD) Prosthesis

4.1.1.1 Compressive Principal Strains

The compressive principal strains in the intact specimens ($IS$ condition) and those following TSA (i.e. $Joint Replaced$ or $JR$ condition), as averaged across five specimens, are plotted against the axial forces imposed at the mid-humerus (simulated deltoid action) in Figure 4-1a. From left to right, the columns correspond to the three arm positions namely lateral abduction, horizontal flexion and horizontal extension. The compressive principal strains at peak loads of 200N following TSA using a STD glenoid prosthesis were statistically not different from those in intact specimens (Figure 4-1b; Table 2a). However, the anterior compressive strains were significantly higher in extension as compared to abduction ($p = 0.007$; Table 2b); specifically, those in extension following TSA were higher than those in abduction in intact specimens, which was considered as the neutral reference for loading scenarios ($p = 0.0121$).
Figure 4-1: Comparison of average compressive principal strains in intact specimen and after TSA using a standard glenoid prosthesis a) dynamic profiles from 20 to 200 N loading b) strains at 200 N loads. (*) indicates significant difference as compared to the respective values for intact specimens in abduction.

Table 2: P-values of compressive principal strains with STD prosthesis

a) P-values across the conditions Intact Specimen (IS) and Joint Replaced (JR)

<table>
<thead>
<tr>
<th>Variable</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Compressive Principal Strains</td>
<td>0.645</td>
</tr>
<tr>
<td>Anterior Compressive Principal Strains</td>
<td>0.173</td>
</tr>
<tr>
<td>Posterior Compressive Principal Strains</td>
<td>0.595</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Compressive Principal Strains</td>
<td>1.000</td>
<td>0.100</td>
<td>0.097</td>
</tr>
<tr>
<td>Anterior Compressive Principal Strains</td>
<td>0.099</td>
<td>0.007</td>
<td>0.417</td>
</tr>
<tr>
<td>Posterior Compressive Principal Strains</td>
<td>0.575</td>
<td>0.486</td>
<td>0.987</td>
</tr>
</tbody>
</table>
4.1.1.2 Tensile Principal Strains

The tensile principal strains under peak loading of 200N following TSA using a STD glenoid component were statistically not different from those in the intact specimens ($p > 0.05$; Figure 4-2b; Table 3a). The arm position had a significant effect on tensile strains; overall, the strains were different in flexion as compared to those in extension ($p = 0.023, 0.003$ and $0.005$ for the superior, anterior and posterior strains respectively; Table 3b). Analysis of interaction between Condition and Arm Position revealed that the anterior tensile strains in flexion following TSA were significantly greater than those in abduction in intact specimens and those in flexion following TSA ($p = 0.021$ and $0.041$ respectively). As mentioned earlier, abduction in intact specimens was regarded as the neutral loading reference.

Figure 4-2: Comparison of average tensile principal strains in intact specimen and after TSA using a standard glenoid prosthesis a) dynamic profiles from 20 to 200 N loading and b) strains at 200 N loads. (* indicates significant difference as compared to the respective values for intact specimens in abduction.)
Table 3: P-values of tensile principal strains with STD prosthesis

a) P-values across the conditions Intact Specimen (IS) and Joint Replaced (JR)

<table>
<thead>
<tr>
<th>Variable</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Tensile Principal Strains</td>
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</tr>
<tr>
<td>Anterior Tensile Principal Strains</td>
<td>0.452</td>
</tr>
<tr>
<td>Posterior Tensile Principal Strains</td>
<td>0.827</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Tensile Principal Strains</td>
<td>0.670</td>
<td>0.130</td>
<td>0.023</td>
</tr>
<tr>
<td>Anterior Tensile Principal Strains</td>
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<td>0.003</td>
</tr>
<tr>
<td>Posterior Tensile Principal Strains</td>
<td>0.949</td>
<td>0.002</td>
<td>0.005</td>
</tr>
</tbody>
</table>

4.1.1.3 Principal Strain Directions

The principal strain directions exhibited substantial variability across specimens, and were not different across conditions ($p > 0.5$; Figure 4-3)).

4.1.1.4 Glenohumeral Translations

The average humeral head translations, with respect to the glenoid, in the antero-posterior (AP), supero-inferior (SI) the lateral-medial (LM) directions during loading from 20 to 200 N are shown in Figure 4-4a. The translations occurring between a pre-load of 20N and a maximum
humeral load of 200N were small in the native joints as well as after arthroplasty (Table 4; Figure 4-4b). Statistically, the AP, SI and LM translations following TSA were not different from those in the intact specimens ($p > 0.05$; Table 5a). Arm position did not have any significant effect on translations ($p > 0.05$; Table 5b).

Table 4: Mean translations and ranges in specimens implanted with a Standard prosthesis

<table>
<thead>
<tr>
<th></th>
<th>Abduction</th>
<th>Flexion</th>
<th>Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>IS</td>
<td>JR</td>
<td>IS</td>
</tr>
<tr>
<td>Antero-Posterior</td>
<td>-1.5</td>
<td>0.3</td>
<td>0.0</td>
</tr>
<tr>
<td></td>
<td>(-3.7, +1.3)</td>
<td>(-1.7, +1.5)</td>
<td>(-3.6, +3.7)</td>
</tr>
<tr>
<td>Supero-Inferior</td>
<td>-1.2</td>
<td>-0.1</td>
<td>0.2</td>
</tr>
<tr>
<td></td>
<td>(-6.4, +0.7)</td>
<td>(-1.0, +0.7)</td>
<td>(-2.4, +3.2)</td>
</tr>
<tr>
<td>Lateral-Medial</td>
<td>-2.2</td>
<td>-0.4</td>
<td>-1.9</td>
</tr>
<tr>
<td></td>
<td>(-8.8, +1.1)</td>
<td>(-3.3, +1.7)</td>
<td>(-7.8, +3.1)</td>
</tr>
</tbody>
</table>

Figure 4-4: Comparison of average joint translations in intact specimen and after TSA using a standard glenoid prosthesis a) dynamic profiles from 20 to 200 N loading and b) translations at 200 N loads.
Table 5: P-values of the glenohumeral translations with STD prosthesis

a) P-values across the conditions Intact Specimen (IS) and Joint Replaced (JR)

<table>
<thead>
<tr>
<th>Variable</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Antero-posterior Translations</td>
<td>0.330</td>
</tr>
<tr>
<td>Supero-inferior Translations</td>
<td>0.820</td>
</tr>
<tr>
<td>Lateral-medial Translations</td>
<td>0.594</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Antero-posterior Translations</td>
<td>0.613</td>
<td>0.716</td>
<td>0.984</td>
</tr>
<tr>
<td>Supero-inferior Translations</td>
<td>0.799</td>
<td>0.963</td>
<td>0.645</td>
</tr>
<tr>
<td>Lateral-medial Translations</td>
<td>0.915</td>
<td>0.885</td>
<td>0.997</td>
</tr>
</tbody>
</table>
4.1.2 Polyethylene-Step (Poly-step) Prosthesis

4.1.2.1 Compressive Principal Strains

The superior compressive principal strains at peak loads, in the presence of a 20° posterior glenoid defect, were significantly lower than those in the Intact Specimen (IS) and Joint Replaced (JR) conditions ($p = 0.027$ and $0.003$ respectively; Figure 4-5; Table 6a). Overall, the arm position had greater effect on the anterior compressive strains; those in flexion were significantly higher than those in abduction and extension ($p = 0.002$ and $0.017$ respectively; Table 6b).

Figure 4-5: Comparison of average compressive principal strains in intact specimen and after TSA using a Poly-step glenoid prosthesis a) dynamic profiles from 20 to 200 N loading b) strains at 200 N loads

Table 6: P-values of compressive principal strains with Poly-step prosthesis

<table>
<thead>
<tr>
<th>Variable</th>
<th>GD vs. IS</th>
<th>GD vs. JR</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Compressive Principal Strains</td>
<td>0.027</td>
<td>0.003</td>
<td>0.652</td>
</tr>
<tr>
<td>Anterior Compressive Principal Strains</td>
<td>0.750</td>
<td>0.288</td>
<td>0.700</td>
</tr>
<tr>
<td>Posterior Compressive Principal Strains</td>
<td>0.209</td>
<td>0.852</td>
<td>0.466</td>
</tr>
</tbody>
</table>
b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Compressive Principal Strains</td>
<td>0.055</td>
<td>0.115</td>
<td>0.932</td>
</tr>
<tr>
<td>Anterior Compressive Principal Strains</td>
<td>0.002</td>
<td>0.662</td>
<td>0.017</td>
</tr>
<tr>
<td>Posterior Compressive Principal Strains</td>
<td>0.945</td>
<td>0.354</td>
<td>0.214</td>
</tr>
</tbody>
</table>

4.1.2.2 Tensile Principal Strains:

The tensile principal strains on the superior, anterior and posterior aspects of the glenoid in the native joints under 200 N loads were not significantly different from either those following defect creation or TSA ($p > 0.05$; Figure 4-6; Table 7a).

The arm position had significant effects on tensile strains all around the glenoid (b). In flexion, the anterior tensile strains were significantly greater than those in either abduction or extension ($p < 0.001$ in both cases). The posterior strains were statistically different across all arm positions ($p = 0.025, 0.002$ for abduction as compared to flexion and extension, and $< 0.001$...
for flexion as compared to extension). The superior tensile strains were significantly higher in extension than in abduction \((p = 0.006)\). Interaction between different arm positions and conditions revealed that the anterior tensile strains in flexion in intact specimens, and anterior and posterior strains following arthroplasty were higher than those in the reference scenario, i.e. in abduction in intact specimens \((p = 0.003, <0.001\) and 0.006 respectively; Figure 4-6).

Table 7: P-values of tensile principal strains with Poly-step prosthesis

<table>
<thead>
<tr>
<th>Variable</th>
<th>GD vs. IS</th>
<th>GD vs. JR</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Tensile Principal Strains</td>
<td>0.524</td>
<td>0.999</td>
<td>0.512</td>
</tr>
<tr>
<td>Anterior Tensile Principal Strains</td>
<td>0.926</td>
<td>0.287</td>
<td>0.483</td>
</tr>
<tr>
<td>Posterior Tensile Principal Strains</td>
<td>0.137</td>
<td>0.926</td>
<td>0.064</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Tensile Principal Strains</td>
<td>0.291</td>
<td>0.006</td>
<td>0.175</td>
</tr>
<tr>
<td>Anterior Tensile Principal Strains</td>
<td>0.000</td>
<td>0.905</td>
<td>0.000</td>
</tr>
<tr>
<td>Posterior Tensile Principal Strains</td>
<td>0.025</td>
<td>0.002</td>
<td>0.000</td>
</tr>
</tbody>
</table>

4.1.2.3 Principal Strain Directions

The principal strain angles exhibited substantial variability and were not different across conditions \((p > 0.05; \text{Figure 4-7}).

Figure 4-7: Comparison of average principal strain directions at 200 N loads in intact specimen and after TSA using a Poly-step glenoid prosthesis.
4.1.2.4 Glenohumeral Translations

The average AP, SI and LM translations at peak loads in intact specimens, and those following defect creation and joint replacement with a Poly-step prosthesis are shown in Table 8 and Figure 4-8. The shoulder translation ranges in the SI direction following defect creation were different as compared to those following TSA ($p = 0.034$); however, they were not different in all other scenarios ($p > 0.05$; Figure 4-8; Table 9).

Table 8: Mean translations and ranges in specimens which received a Poly-step prosthesis

<table>
<thead>
<tr>
<th></th>
<th>Abduction</th>
<th>Flexion</th>
<th>Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>IS</td>
<td>GD</td>
<td>JR</td>
</tr>
<tr>
<td>AP</td>
<td>-0.9 (-2.1,+0.6)</td>
<td>-0.7 (-2.7,+2.5)</td>
<td>-0.3 (-4.5,+2.7)</td>
</tr>
<tr>
<td>SI</td>
<td>-0.5 (-1.1,+0.5)</td>
<td>-1.3 (-3.6,+1.3)</td>
<td>1.8 (-0.7,+7.6)</td>
</tr>
<tr>
<td>LM</td>
<td>-1.2 (-2.0,+0.4)</td>
<td>-2.2 (-5.4,+0.5)</td>
<td>-1.9 (-5.8,+2.1)</td>
</tr>
</tbody>
</table>

Figure 4-8: Comparison of the average joint translations in the intact specimen and after TSA using a Poly-step glenoid prosthesis a) dynamic profiles from 20 to 200 N loading and b) translations at 200 N loads.
Table 9: P-values of the glenohumeral translations with Poly-step prosthesis

a) P-values across the conditions Intact Specimen (IS), Glenoid Defect (GD) and Joint Replaced (JR)

<table>
<thead>
<tr>
<th>Variable</th>
<th>GD vs. IS</th>
<th>GD vs. JR</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Antero-posterior Translations</td>
<td>0.997</td>
<td>0.830</td>
<td>0.789</td>
</tr>
<tr>
<td>Supero-inferior Translations</td>
<td>0.588</td>
<td>0.034</td>
<td>0.244</td>
</tr>
<tr>
<td>Lateral-medial Translations</td>
<td>0.936</td>
<td>0.499</td>
<td>0.712</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Antero-posterior Translations</td>
<td>0.313</td>
<td>0.639</td>
<td>0.058</td>
</tr>
<tr>
<td>Supero-inferior Translations</td>
<td>0.942</td>
<td>0.572</td>
<td>0.381</td>
</tr>
<tr>
<td>Lateral-medial Translations</td>
<td>0.878</td>
<td>0.167</td>
<td>0.064</td>
</tr>
</tbody>
</table>
4.1.3 Titanium-step (TS) Prosthesis

4.1.3.1 Compressive Principal Strains

In the presence of a defect, the superior compressive principal strains at 200N were lower than those in intact specimens \((p = 0.007; \text{ Figure 4-9, Table 10a})\), while the anterior strains in extension were significantly lower than those following TSA using a Ti-step prosthesis.

After arthroplasty, the anterior strains were significantly higher than those in both the GD and IS conditions \((p = 0.006 \text{ and } 0.003 \text{ respectively})\). Overall, the anterior strains were higher in extension than in abduction \((p = 0.049; \text{ Table 10 b})\). Specifically, the anterior strains in extension following arthroplasty were greater than those in abduction and extension in intact specimens \((p = 0.001 \text{ and } 0.024 \text{ respectively})\). The superior strains in extension were also significantly greater than those in either flexion or abduction \((p = 0.001 \text{ and } 0.015 \text{ respectively})\), while the posterior strains were higher in flexion than in extension across different conditions \((p = 0.01)\).
Table 10: P-values of the compressive principal strains with Ti-step prosthesis

a) P-values across the conditions Intact Specimen (IS), Glenoid Defect (GD) and Joint Replaced (JR)

<table>
<thead>
<tr>
<th>Variable</th>
<th>GD vs. IS</th>
<th>GD vs. JR</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Compressive Principal Strains</td>
<td>0.007</td>
<td>0.371</td>
<td>0.143</td>
</tr>
<tr>
<td>Anterior Compressive Principal Strains</td>
<td>0.937</td>
<td>0.006</td>
<td>0.003</td>
</tr>
<tr>
<td>Posterior Compressive Principal Strains</td>
<td>0.979</td>
<td>0.649</td>
<td>0.768</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Compressive Principal Strains</td>
<td>0.477</td>
<td>0.015</td>
<td>0.001</td>
</tr>
<tr>
<td>Anterior Compressive Principal Strains</td>
<td>0.169</td>
<td>0.049</td>
<td>0.813</td>
</tr>
<tr>
<td>Posterior Compressive Principal Strains</td>
<td>0.090</td>
<td>0.127</td>
<td>0.001</td>
</tr>
</tbody>
</table>

4.1.3.2 Tensile Principal Strains

The superior and anterior tensile strains under peak loads were not different across the conditions, but the posterior tensile strains following TSA using the Ti-step prosthesis were statistically higher than those in the native joint (p = 0.001; Table 11a; Figure 4-10a and b). In extension, the superior tensile strains were greater than those in abduction and flexion (p = 0.020 and 0.006 respectively; b), and so were the posterior tensile strains (p = 0.007 and <0.001 respectively). In contrast, the anterior tensile strains were greater in abduction as compared to flexion and extension (p < 0.001 and 0.006 respectively). As compared to the neutral loading reference, the anterior as well as posterior tensile strains in extension following TSA using a Ti-step prosthesis were significantly higher (p = 0.038 and 0.003 respectively). Furthermore, the anterior strains in extension following TSA were also greater than the respective values for intact specimens (p = 0.036).
Figure 4-10: Comparison of average tensile principal strains in intact specimen and after TSA using a Ti-step glenoid prosthesis a) dynamic profiles from 20 to 200 N loading and b) strains at 200 N loads. (* indicates significant difference as compared to the respective values for intact specimens in abduction, while ○ indicates significant difference as compared to IS condition)

Table 11: P-values of tensile principal strains with Ti-step prosthesis

a) P-values across the conditions Intact Specimen (IS), Glenoid Defect (GD) and Joint Replaced (JR)

<table>
<thead>
<tr>
<th>Variable</th>
<th>GD vs. IS</th>
<th>GD vs. JR</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Tensile Principal Strains</td>
<td>0.472</td>
<td>0.190</td>
<td>0.818</td>
</tr>
<tr>
<td>Anterior Tensile Principal Strains</td>
<td>0.859</td>
<td>0.387</td>
<td>0.167</td>
</tr>
<tr>
<td>Posterior Tensile Principal Strains</td>
<td>0.201</td>
<td>0.091</td>
<td>0.001</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Superior Tensile Principal Strains</td>
<td>0.876</td>
<td>0.020</td>
<td>0.006</td>
</tr>
<tr>
<td>Anterior Tensile Principal Strains</td>
<td>0.000</td>
<td>0.006</td>
<td>0.584</td>
</tr>
<tr>
<td>Posterior Tensile Principal Strains</td>
<td>0.310</td>
<td>0.007</td>
<td>0.000</td>
</tr>
</tbody>
</table>
4.1.3.3 Principal Strain Directions

The principal strain angles were not different across conditions ($p > 0.05$; Figure 4-11).

![Diagram of average principal strain angles at peak loads in abduction, flexion, and extension]

Figure 4-11: Comparison of average principal strain directions at 200 N loads in intact specimens, after defect creation and after TSA using a Ti-step glenoid prosthesis.

4.1.3.4 Glenohumeral Translations

The average joint translations under peak loads (Table 12; Figure 4-12) tended to be higher in flexion and extension; but the differences were not significant across different arm positions and experimental conditions in most scenarios. The average AP translations following TSA were significantly higher and more anteriorly directed than those in the native joints ($p = 0.02$; Figure 4-12). Translation ranges were considerably higher in one of the specimens in flexion and extension, which was likely due to occasional subluxations. Analysis of the humeral head coordinates at baseline, i.e. under 20N pre-loads, after excluding that subluxating specimen did not reveal significant difference between the Intact Specimen and Glenoid Defect conditions.

Table 12: Mean translations and ranges in specimens which received a Ti-step prosthesis

<table>
<thead>
<tr>
<th></th>
<th>Abduction</th>
<th>Flexion</th>
<th>Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>IS</td>
<td>GD</td>
<td>JR</td>
</tr>
<tr>
<td><strong>Antero-Posterior</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Antero-Posterior</td>
<td>-1.3</td>
<td>-1.6</td>
<td>3.5</td>
</tr>
<tr>
<td></td>
<td>(-7.1,+0.6)</td>
<td>(-10.4,+3.4)</td>
<td>(-0.4,+15.3)</td>
</tr>
<tr>
<td>Supero-Inferior</td>
<td>0.3</td>
<td>1.6</td>
<td>-1.2</td>
</tr>
<tr>
<td></td>
<td>(-1.4,+3.1)</td>
<td>(-3.9,+12.5)</td>
<td>(-5.1,+1.5)</td>
</tr>
<tr>
<td>Lateral-Medial</td>
<td>1.1</td>
<td>1.5</td>
<td>0.4</td>
</tr>
<tr>
<td></td>
<td>(-1.7,+7.2)</td>
<td>(-3.0,+10.1)</td>
<td>(-1.5,+2.5)</td>
</tr>
</tbody>
</table>
Figure 4-12: Comparison of average joint translations in intact specimen and after TSA using a Ti-step glenoid prosthesis a) dynamic profiles from 20 to 200 N loading and b) translations at 200 N loads.

Table 13: P-values of glenohumeral translations with Ti-step prosthesis

a) P-values across the conditions Intact Specimen (IS), Glenoid Defect (GD) and Joint Replaced (JR)

<table>
<thead>
<tr>
<th>Variable</th>
<th>GD vs. IS</th>
<th>GD vs. JR</th>
<th>IS vs. JR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Antero-posterior Translations</td>
<td>0.567</td>
<td>0.181</td>
<td>0.021</td>
</tr>
<tr>
<td>Supero-inferior Translations</td>
<td>0.867</td>
<td>0.930</td>
<td>0.988</td>
</tr>
<tr>
<td>Lateral-medial Translations</td>
<td>0.743</td>
<td>0.922</td>
<td>0.934</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Antero-posterior Translations</td>
<td>0.678</td>
<td>0.982</td>
<td>0.566</td>
</tr>
<tr>
<td>Supero-inferior Translations</td>
<td>0.943</td>
<td>0.992</td>
<td>0.955</td>
</tr>
<tr>
<td>Lateral-medial Translations</td>
<td>0.610</td>
<td>1.00</td>
<td>0.604</td>
</tr>
</tbody>
</table>
4.2 Section II: Comparison of STD, Poly-step and Titanium-step Prostheses

In this section, the change in variables as a consequence of TSA is compared across the STD, Poly-step and Ti-step prostheses. The variable measurements in the intact specimens were subtracted from those following TSA, which also adjusted for the inter-specimen variability to some extent.

4.2.1 Compressive Principal Strains

Figure 4-13 shows the change in average compressive principal strains due to TSA during axial loading from 20 to 200N. The columns indicate different arm positions while the rows indicate different prostheses which were implanted in five specimens each.

Figure 4-13: Comparison of average compressive principal strains across STD, Poly-step and Ti-step prostheses in abduction, flexion and extension
In general, the alterations in compressive principal strains were smallest after TSA using a Standard prosthesis, and greatest for the Ti-step prosthesis. At peak loads, the change in superior as well as anterior strains with the Ti-step implant was significantly greater than those with the Standard and Poly-step implants ($p = 0.023$ and $0.024$ for superior, and $0.017$ and $0.014$ for anterior strains respectively; Figure 4-14). In contrast, no significant difference was observed in the posterior strains ($p > 0.05$; Table 14). Although arm position did not have an overall significant effect, specifically, the anterior strains in extension were higher for the Ti-step implant than for the Poly-step implant ($p = 0.021$).

![Figure 4-14: Comparison of average compressive principal strains at 200N across STD, Poly-step and Ti-step prostheses in abduction, flexion and extension (● indicates significant difference as compared to the respective values for the Poly-step implant).](image)

<table>
<thead>
<tr>
<th>Variable</th>
<th>STD vs Poly-step</th>
<th>STD vs TS</th>
<th>Poly-step vs. TS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Change in Superior Compressive Principal Strains</td>
<td>1.00</td>
<td>0.023</td>
<td>0.024</td>
</tr>
<tr>
<td>Change in Anterior Compressive Principal Strains</td>
<td>0.996</td>
<td>0.017</td>
<td>0.014</td>
</tr>
<tr>
<td>Change in Posterior Compressive Principal Strains</td>
<td>0.068</td>
<td>0.285</td>
<td>0.721</td>
</tr>
</tbody>
</table>

Table 14: *P*-values of GLM ANOVA on the alteration in the compressive principal strains due to TSA across the STD, Poly-step and Ti-step prostheses (n = 15)

(a) *P*-values across the STD, Poly-step and Ti-step prostheses

(b) *P*-values across the Abduction, Flexion and Extension arm positions
4.2.2 Tensile Principal Strains

The dynamic profiles of the change in tensile principal strains are shown below. The posterior tensile principal strains following TSA using a Ti-step implant appear to be increased across all arm positions. Significant increases were observed in the posterior tensile strains with the Ti-step implant as compared to those with the STD implant (Figure 4-15). The results in all the other scenarios were not statistically different across the arm positions and the implant types (Table 15). However, examination of effect of interaction between Arm Position and Condition, revealed that the anterior strains in extension for the Ti-step implant were also greater than those for the STD implant ($p = 0.011$).

Figure 4-15: Comparison of average tensile principal strains across STD, Poly-step and Ti-step prostheses in abduction, flexion and extension
Figure 4-16: Comparison of average tensile principal strains at 200N across the STD, Poly-step and Ti-step prostheses. (○ indicates significant difference as compared to the respective values for the standard implant).

Table 15: P-values of GLM ANOVA on the change in tensile principal strains across the prostheses (n = 15)

a) P-values across the STD, Poly-step and Ti-step prostheses

<table>
<thead>
<tr>
<th>Variable</th>
<th>STD vs. Poly-step</th>
<th>STD vs. TS</th>
<th>Poly-step vs. TS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Change in Superior Tensile Principal Strains</td>
<td>0.247</td>
<td>0.994</td>
<td>0.292</td>
</tr>
<tr>
<td>Change in Anterior Tensile Principal Strains</td>
<td>0.997</td>
<td>0.835</td>
<td>0.873</td>
</tr>
<tr>
<td>Change in Posterior Tensile Principal Strains</td>
<td>0.317</td>
<td>0.002</td>
<td>0.083</td>
</tr>
</tbody>
</table>

b) P-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
<tbody>
<tr>
<td>Change in Superior Tensile Principal Strains</td>
<td>0.618</td>
<td>0.905</td>
<td>0.370</td>
</tr>
<tr>
<td>Change in Anterior Tensile Principal Strains</td>
<td>0.875</td>
<td>0.936</td>
<td>0.681</td>
</tr>
<tr>
<td>Change in Posterior Tensile Principal Strains</td>
<td>0.864</td>
<td>0.612</td>
<td>0.901</td>
</tr>
</tbody>
</table>

4.2.3 Principal Strain Angles

Substantial variability was observed in the average change in principal strain angles at 200N loads. There were no significant differences across the implant types (p > 0.05).
4.2.4 Glenohumeral Translations

The change in translations in the AP, SI and LM directions following TSA appear to be in a small range. Statistically, the change in antero-posterior translations as a consequence of TSA using the Ti-step implant were greater than those with the STD and Poly-step implants, as assessed by the differences under peak loads ($p = 0.01$ and $0.04$ respectively; Figure 4-18; Table 16). However, the changes in translations in the supero-inferior and lateral-medial directions were not significantly different across the implant types ($p > 0.05$).

Figure 4-18: Comparison of average glenohumeral translations across STD, Poly-step and Ti-step prostheses in abduction, flexion and extension
Mean Change in Joint Translations at 200N in Abduction

Mean Change in Joint Translations at 200N in Flexion

Mean Change in Joint Translations at 200N in Extension

Figure 4-19: Comparison of average glenohumeral translations at 200N across STD, Poly-step and Ti-step prostheses in abduction, flexion and extension.

Table 16: *P*-values of GLM ANOVA on the alteration in the joint translations due to TSA across the STD, Poly-step and Ti-step prostheses (n = 15)

a) *P*-values across the STD, Poly-step and Ti-step prostheses

<table>
<thead>
<tr>
<th>Variable</th>
<th>STD vs Poly-step</th>
<th>STD vs TS</th>
<th>Poly-step vs. TS</th>
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</thead>
<tbody>
<tr>
<td>Change in Antero-posterior Translations</td>
<td>0.834</td>
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<td>0.012</td>
</tr>
<tr>
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<td>0.996</td>
<td>0.876</td>
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<td>Change in Lateral-medial Translations</td>
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<td>0.992</td>
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</table>

b) *P*-values across the Abduction, Flexion and Extension arm positions

<table>
<thead>
<tr>
<th>Variable</th>
<th>Abduction vs. Flexion</th>
<th>Abduction vs. Extension</th>
<th>Flexion vs. Extension</th>
</tr>
</thead>
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<tr>
<td>Change in Antero-posterior Translations</td>
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<td>Change in Supero-inferior Translations</td>
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<tr>
<td>Change in Lateral-medial Translations</td>
<td>0.894</td>
<td>0.623</td>
<td>0.359</td>
</tr>
</tbody>
</table>
4.3 Summary of Results

The magnitudes and angles of the compressive and tensile principal strains as well as glenohumeral translations at peak loads of 200N were not significantly different following TSA using a standard glenoid prosthesis as compared to the native joints. In the presence of a 20° biconcave osteoarthritic defect in the posterior glenoid, the superior and anterior compressive principal strains were significantly lower than those in intact specimens and those following TSA using a Poly-step and Ti-step implants. Most other variables were not different in the GD, IS and JR conditions. Significant variability in principal strain angles precluded any definitive conclusions about loading directionality. Following implantation of the Poly-step glenoid prosthesis, all the variables were reasonably similar to those measured in the native joints; however, following Ti-step prosthesis implantation, the anterior compressive and posterior tensile principal strains were statistically higher, and the AP translations were directed more anteriorly.

Comparison of the change in the research variables, as a consequence of arthroplasty, using the STD, Poly-step and Ti-step prostheses, revealed increased alterations in the superior and anterior compressive strains, and AP translations with the Ti-step prosthesis as compared to the STD and Poly-step prostheses. The results in the other categories were otherwise similar across the three implant types.
CHAPTER 5
DISCUSSION AND CONCLUSIONS

5.1 Mechanical Alterations by STD Prosthesis

5.1.1 Evaluation of Hypothesis # 1

“In the absence of a posterior glenoid defect, TSA using a conventional prosthesis will have no significant effect on joint stability, as measured by humeral head position and translations, but will increase loading at most locations of the glenoid bone, as measured by peri-glenoid principal strains.”

We observed that compressive principal strains ($\varepsilon_{\text{min}}$) on the superior and anterior aspects of the glenoid at peak loads, following TSA using a STD prosthesis, were slightly higher as compared to those in intact specimens; however the differences were not statistically significant. Likewise, the tensile principal strains ($\varepsilon_{\text{max}}$) and strain angles were also not altered as a consequence of arthroplasty. Thus, it appears that arthroplasty using a STD glenoid prosthesis did not significantly alter the loading environment of the joint. These findings are somewhat different than those reported in the past, whereby higher peri-glenoid strains were observed following conventional TSA (Maurel et al., 2002; and Pelletier et al., 2008). The anterior compressive strains were significantly higher in extension as compared to abduction. The arm position had even greater effect on tensile strains; the anterior tensile strains being greater in flexion while the posterior tensile strains being greater in extension.

Regarding joint translations across all three arm positions, we observed that the average AP, SI and LM translations ranged from -3.7 to 3.7 mm, -6.4 to 3.2 mm and -8.8 to 3.1 mm respectively for intact specimens. The respective values following TSA were -2.3 to 4.5, -8.2 to 3.8 mm and -7.3 to 6.2 mm, and were not statistically different from those for the native joints,
indicating that TSA using a STD prosthesis did not cause significant changes in joint stability, which concurs with the previous literature (Karduna et al., 1997).

5.2 Mechanical Alterations by Posterior-step Prostheses

5.2.1 Evaluation of Hypothesis # 2

“TSA using a custom Posterior-step component to compensate for a biconcave posterior glenoid defect will reverse the alterations in peri-glenoid strains and joint translations as caused by the defect.”

5.2.1.1 Polyethylene-step Prosthesis

In the presence of a 20° biconcave (type B2) posterior glenoid defect, the peak compressive principal strains on the superior aspect of the glenoid were significantly smaller than those in the intact specimens and those following TSA using a Poly-step prosthesis, while post-arthroplasty strains were not different from those in the native joints. In contrast, the tensile principal strains in the presence of a defect were not significantly different from those in either the intact specimens or following TSA using a Poly-step prosthesis. These results indicate that the glenoid defect altered the loading pattern of the joint to some extent. The smaller superior compressive strains in the presence of the defect may partly be due to the reduced contact area in the postero-superior aspect. Surprisingly, the posterior compressive strains were not significantly different across the conditions. Nevertheless, both the compressive and tensile principal strains following TSA with a Poly-step prosthesis were similar to those in the intact specimens, findings in agreement with hypothesis # 2. Overall, the arm position had a greater effect on tensile strains than compressive strains. While superior tensile strains were significantly higher in extension than in abduction, and the anterior tensile strains greater in flexion, posterior strains were
different across all arm positions. In contrast, the anterior compressive strains were highest in flexion.

The average AP, SI and LM glenohumeral translations across the arm positions ranged from -4.8 to 2.4 mm, -4.8 to 4.6 mm and -5.1 to 1.0 mm respectively for intact specimens, -6.6 to 3.2 mm, -4.2 to 1.6 mm and -7.7 to 1.8 mm in the presence of a defect, and -19.4 to 2.7 mm, -1.9 to 10.7 mm and -5.8 to 11.4 mm following TSA using a Poly-step implant respectively. The translations in the SI direction in the presence of a defect were statistically different as compared to those following TSA; however, no difference was observed in other directions or loading scenarios. These findings suggest that joint stability was not substantially altered across the conditions or arm positions in the presence of a defect. It is possible that the humeral head impingement at the edge of the defect may have caused the translations to remain low.

5.2.1.2 Titanium-step Prosthesis

In the presence of a posterior glenoid defect, in the five specimens that received the Ti-step implant, the superior compressive strains remained significantly lower than those in the intact specimens. The anterior strains in the presence of a defect were lower than those after TSA using a Ti-step prosthesis, especially in extension. On the other hand, the tensile strains in the presence of a defect were not different across the IS, GD and JR conditions. Thus, the glenoid defect caused some significant alterations in loading of the superior and anterior aspects of the glenoid. The post-arthroplasty anterior compressive strains and posterior tensile strains were statistically greater than those in the native joint, which indicates that the Ti-step prosthesis did not fully restore the native joint loading pattern. Regarding the effect of arm position, both the compressive and tensile strains on the superior, as well as anterior aspects of the glenoid were
highest in extension. While the posterior tensile strains were also highest in extension, the posterior compressive strains were highest in flexion.

The average respective AP, SI and LM glenohumeral translations across the arm positions ranged from -27.8 to 1.8 mm, -22.0 to 32.4 mm and -31.2 to 36.6 mm for intact specimens, -15.4 to 8.1 mm, -31.8 to 16.8 mm and -3.4 to 30.3 mm in the presence of a defect, and -3.9 to 17.5 mm, -17.6 to 33.0 mm and -8.6 to 38.3 mm following TSA using a Ti-step implant respectively. Statistically, the AP translations were different following TSA as compared to the intact specimens, while those following defect creation were not. Nevertheless, the translation ranges appear to be much larger with this group of specimens. Occasional subluxations of one of the specimens may have caused this effect. Specimen-wise inspection of the data confirmed this suspicion.

5.3 Comparison of Mechanical Performance of STD, Poly-step & Ti-step Implants

5.3.1 Evaluation of Hypothesis # 3

“TSA using a Posterior-step component done in the presence of a biconcave glenoid defect will have similar effects as those after TSA using a conventional prosthesis in the absence of a defect.”

In general, the alterations in the compressive and tensile principal strains, and glenohumeral translations were smallest for the Standard prosthesis, and greatest for the Ti-step prosthesis. Statistically, alterations in the research variables as a consequence of implantation of the STD and the Poly-step prostheses were similar, thus supporting hypothesis # 3. On the other hand, alterations in the superior and anterior compressive principal strains, posterior tensile strains, and AP translations for the Ti-step prosthesis were significantly greater than for the STD and Poly-step prostheses. Arm position did not have an overall significant effect for any of the alterations in the variables across implant types. However, the change in anterior compressive
and tensile strains in extension for the Ti-step implant was significantly higher than those with the Poly-step and STD implants respectively.

5.4 Significance of Peri-glenoid Principal Strains

The glenohumeral joint encounters significant forces as a result of muscular contraction, the weight of the upper limb and the weight of the objects in hand (Apreleva et al., 2000; Anglin et al., 2000; and Poppen and Walker, 1978). As for arthroplasty of any other joint, the goal of TSA is to optimally restore normal joint mechanics. Therefore, comparing mechanical variables before and after arthroplasty is of particular importance. The glenohumeral contact forces are thought to significantly increase following TSA (Shapiro et al., 2007). These forces have been recorded \textit{in vivo} using instrumented implants (Bergmann et al., 2007); however, it is not possible to record the \textit{in vivo} forces in intact shoulder joints in live subjects. Experimentations using animal, cadaver and computer models are attempts to fill this gap.

The maximum and minimum principal strains are calculated for a unit element of a loaded body, as oriented along the principal strain direction where the shear strains are zero. Principal strains and angles are a reliable standardized measure of mechanical loading that help to simplify the stress/strain environment down to a few numbers, and enable comparison across different loading scenarios. It has been well established that mechanical loading dictates bone architecture. The trabeculae of cancellous bone in the body align themselves along the principal strain direction (Cowin SC, 1986). For example, the collagen fibers in the acetabulum show dense circumferential arrangement. \textit{Calcar femorale}, a dense triangular region of the femoral neck is another classic example (Venieratos et al., 1987). Likewise, higher bone density is also observed in regions of greater loading. Using imaging and FEA techniques, carrying a light load was found to make the greatest contribution to the trabecular architecture of the glenoid (Lim et al.,
Issues regarding bone stresses and strains are also important in understanding the alterations in the mechanical environment of bone following prosthetic implantation.

5.5 Implications of Alterations in Shoulder Loading

Higher peri-glenoid strains following arthroplasty, as measured in cadaver specimens in previous studies (Maurel et al., 2002; Diop et al., 2005; and Pelletier et al., 2008) indicated that the post-arthroplasty loading environment of the shoulder is probably not the same as the native joints. Similar to these studies, we observed a general trend of increased compressive and tensile strains following arthroplasty; however, the increases were not statistically significant. Alterations in joint loading are particularly important in the case of shoulder as eccentric loading following TSA is believed to cause glenoid component loosening, the most common complication of the procedure. Cyclic patterns of normal and shear stresses at the component anchorage site are thought to cause implant toggling and loosening (Cofield R.H., 1984; and Karduna et al., 1998).

Mechanical alterations due to arthroplasty are also important in terms of subsequent bone response to loading. Altered patterns of load transfer onto the cortical and trabecular bone tissues are believed to stimulate bone remodeling. Patho-mechanisms of the remodeling process are not clearly understood, but changes in mechanical loading are believed to stimulate osteocytes through altered patterns of fluid-flow in bone at a microscopic level. Changes in stress at the lacunar spaces, and in the electrical membrane potentials of osteocytes and bone lining cells are also thought to play a role. In particular, metal-backed prostheses are hypothesized to cause preferential load transfer on to the cortical shell while off-loading the subchondral trabecular bone tissue of the glenoid, a phenomenon known as “stress-shielding”. Consistent with the theory, we indeed observed significantly greater anterior compressive principal strains and posterior tensile principal strains following TSA using the Ti-step prostheses, as compared to Poly-step and
STD prostheses in this study. The different loading patterns with the Poly-step and Ti-step prostheses are largely due to the different material properties of bone, polyethylene and titanium. The modulus of elasticity of polyethylene is between 2.2 and 8 GPa which is closer to 0.099 to 0.4 Pa modulus of the normal glenoid subchondral bone (Mimer et al. 2008; Hin T.S, 2004; and Choi et al., 1990; Appendix B). On the other hand, the modulus of titanium is between 110 and 120 GPa. Therefore, it appears that the titanium step-block diverted more stresses on to the anterior-half of the glenoid, while providing off-loading or stress-shielding the posterior-half.

Our findings are also consistent with some finite element (FE) studies; several FE models have supported the notion that metal-backed prostheses transfer stresses differently than the polyethylene prostheses (Orr et al., 1988; and Stone et al., 1999). FE techniques have been employed to better understand bone mechanics and alterations following prosthetic implantation. These studies involve creating a model of the loaded body, dividing it finite elements in the form of a mesh and nodes, and defining boundary conditions, and material and structural properties of each individual element. Alterations in the stresses and strains could be accurately estimated using advanced three-dimensional models of human body parts and joint prostheses. However, the issues of stress-shielding are controversial, and interpretations of results differ. While Stone et al. concluded that retention of trabecular loading is more physiological, Orr et al. concluded that the preferential stress transfer to cortical bone is in fact more physiological as bone morphology is known to follow stress distribution patterns, and that greater strength of the cortical bone is a natural consequence of this preferential stress transfer. Consistent with prior observations and theory, we observed that the alterations in both the compressive and tensile strains were greater for the Ti-step implant, which were often significantly different as compared to the conventional prosthesis.
5.6 **Significance of Gleno-humeral Translations**

Superior mobility of the shoulder comes at the cost of its stability. The joint relies heavily on the reinforcement by the surrounding capsulo-ligamentous structures and musculature for stability; but nevertheless is particularly susceptible to instability, subluxations and dislocations. Therefore, the goal of shoulder replacement is to cause minimal damage to the soft tissues while optimally restoring normal joint mechanics. During normal shoulder motion, the center of rotation of the humeral head remains fairly constant relative to the glenoid (Poppen and Walker, 1976), except at terminal motion when it undergoes about 1 to 2 mm of obligate translations (Howell et al., 1988; Harryman et al., 1990; Karduna et al., 1997; and Graichen et al., 2000). Though small, these translations are considered an important aspect of glenohumeral kinematics, which are obviously more complex than pure ball-and-socket rotations. In our experimental set up, we observed larger glenohumeral translations in the intact joints as well as after TSA even in the absence of overt subluxations. Substantial inter-specimen variability was observed and differences were mostly not statistically significant. The translations were generally smaller for the STD prosthesis as compared to the Poly-step and Ti-step prostheses. In the presence of a glenoid defect, the humeral head translations with respect to the glenoid were not significantly different as compared to the intact specimens. This may mean that a 20° defect, as measured from the glenoid center did not cause significant joint instability.

5.7 **Implications of Alterations in Shoulder Stability**

Several technical nuances during TSA can potentially lead to significantly altered shoulder kinematics. Joint line medialization due to excessive reaming, lateralization due to large bone grafts and oversized implants, overstuffing of the joint, inadequate soft-tissue balancing, incorrect component positioning and orientation are some of the complications that can adversely affect the joint kinematics, and lead to instability and implant failure. Indeed, retrieved
glenoid components have revealed wear patterns consistent with impingement by the humeral component in the inferior quadrant, and edge deformation secondary to eccentric loading (Hertel and Ballmer, 2003; and Nho et al., 2009). Lastly, the conformity of the implant system, i.e. glenohumeral radial mismatch, plays an important role in determining postoperative joint kinematics. While less conforming joints are less stable, more conforming systems are believed to be susceptible to early loosening of the glenoid component because of the lack of coupled physiological translations, and increased stresses at the implant-bone interface (Harryman et al., 1995; Karduna et al., 1997 and Walch et al., 2002). The conforming implants are also thought to be more susceptible for surface abrasion, delamination, and rim impingement (Nho et al., 2009). Active translations of the natural joints are best reproduced by less conforming TSA systems, typically with a radial mismatch of 2 to 4 mm (Karduna et al., 1997).

5.8 TSA in the Presence of Excessive Glenoid Retroversion or Bone Loss

Noncontained peripheral glenoid deficiencies pose a significant challenge; the biconcave (type B2) defects in glenohumeral OA is one classic example. As mentioned earlier, posterior glenoid defects create several technical difficulties during TSA, and if inadequately treated, often lead to instability and early component failure. The topic has been extensively discussed throughout this document, but a few key points deserve additional emphasis. Accurate preoperative assessment of posterior glenoid defects using CT images, and understanding the three dimensional geometry of the defect are crucial steps in initial treatment planning. Secondly, defining version of the posterior defect separate from that of the glenoid articular surface, by superimposing images of the normal side, may be necessary in cases of biconcave defects. Accurate positioning and orientation of the component, and soft tissue tensioning, such that the loads across the glenohumeral interface are concentric and aligned along the glenoid centerline, seem to be even more critical in the presence of posterior deficiency (Matsen et al., 2004).
Recent studies on retrieved implants indicate increased susceptibility to toggle for glenoid components that were implanted in retroversion. With rim deformation seen in the postero-inferior aspect and increased anterior radiolucent lines, a *see-saw loading pattern* from antero-posterior rocking mechanism is suspected in cases with glenoid retroversion (Nho et al.). Retroversion leads to decreased glenohumeral contact areas, increased contact pressures, and posterior eccentric loading, all of which increase stresses on the cement mantle, and seem to increase the risk of edge deformation and component loosening (Farron et al., 2006; Matsen et al., 2008; and Shapiro et al., 2007).

Based on recommendations from several authors, eccentric reaming may be appropriate for smaller defects while bone grafts may be considered for larger defects. In addition, custom designed Polyethylene-step prosthesis maybe another viable option for treating shoulder OA with posterior glenoid bone loss and excessive retroversion, as mechanical alterations following its implantation in the presence of a posterior defect was comparable to the use of a conventional prosthesis in the absence of a defect. With the eccentric reaming procedure being suitable only for mild defects < 15°, and the grafting procedure being technically challenging and with inconsistent success rates, another treatment option is certainly warranted, and the idea of having an array of custom prostheses for treating posterior glenoid defects of different sizes and shapes is very appealing.

### 5.9 Designing Issues with Custom Glenoid Prostheses

Prosthetic augmentations, in the form of wedges and blocks, are routinely used in total knee and total hip replacement surgeries to compensate for bony deficiencies, but the concept of using custom designed glenoid prostheses for glenoid bone deficiency has received little attention. In 1988, the posteriorly augmented glenoid prosthesis design by Neer and Morrison had a sloping wedge shaped augmentation on the posterior aspect of the glenoid component.
Likewise, the design developed by Rice et al. (2008) also had a wedge shaped posterior augmentation. In contrast, the prototype design developed in this study has a flat posterior step, which is believed to provide better support for the underlying bone, and to be less vulnerable to the shear forces at the joint. It may be argued that the pointed corners of the step would potentially act as stress-risers, and that some variation in the current step-block design could prove to be biomechanically superior. Further studies involving additional mechanical testing and FEA could provide better insights.

5.10 Clinical Efficacy of Custom Glenoid Prostheses

5.10.1 Past Custom-designed Prostheses

In 1988, Neer and Morrison treated two patients with posterior glenoid deficiency using an augmented prosthesis with a satisfactory outcome. However, the authors discontinued using their augmented prosthesis thereafter, stating a preference for the support provided by a bone graft over polyethylene. No additional information or statistical analysis was provided.

In 2008, about two decades later, Rice et al. implanted a wedge-augmented polyethylene glenoid component (Figure 2-9b) in thirteen patients with a deficient posterior glenoid. Postoperative assessment at a mean follow up of five years (range, two to eight years), as based on pain, patient satisfaction, radiolucent lines, and range of motion, revealed excellent results in five, satisfactory in seven and unsatisfactory in two cases. Three patients had moderate (20-50%) posterior subluxation, and one had severe (>50%) anterior subluxation; however, none of their patients required a revision or any other operative procedure. The authors claimed their high degree of success in addressing posterior glenoid wear and the associated subluxation to combination of techniques, including a posteriorly augmented prosthesis, meticulous soft-tissue balancing and release, and modular humeral heads. Ironically, they concluded that the contribution of the custom implant seemed marginal, and discontinued using the wedge-
augmented prosthesis. No additional details or statistical analyses about the custom implant’s efficacy were provided.

Several critical points can be raised regarding these past studies. Posterior glenoid defect in osteoarthritis is a three-dimensional (3D) deformity (Habermeyer et al., 2006), and therefore, 3D reconstructions have been recommended for preoperative evaluation in patients with glenoid bone loss (Scalise et al., 2008; and Kwon et al.; 2005). 3D CT measurements of the defect dimensions were apparently not conducted in the above mentioned patient series. Furthermore, Rice et al. reported on several techniques that they used to account for posterior glenoid wear prior to implantation of their custom prosthesis. The techniques included eccentric reaming with 5° posterior inclination, and more posterior positioning and angulation of the centering hole. In the absence of an alignment jig of any sorts, it is possible that these techniques may have been inconsistent across the patients. Furthermore, posterior positioning and inclination during reaming and creation of the centering hole may have resulted in excessively retroverted seating of the implant, and therefore, adversely affected the subsequent joint mechanics. No references to biomechanical studies of the posteriorly augmented glenoid implants could be found in past literature. In the lack of detailed information, it is not clear if Neer and Morrison or Rice et al. conducted any mechanical testing of their respective custom prostheses, such as fatigue loading, loosening performance testing or cadaver instrumentation, prior to their implantation in patients.

5.10.2 Metal-backed Prostheses

In their 1988 study, Neer and Morrison recommended a standard metal-backed glenoid prosthesis for most patients, while a standard polyethylene prosthesis was recommended for only those shoulders in which a deficient glenoid precludes insertion of the wider keel of the metal-backed prosthesis (Neer and Morrison, 1988). Subsequent studies on retrieved implants have established that metal-backed prostheses have additional problems of their own, including
separation of the metal from the polyethylene part, and diffuse metal wear particles which are thought to cause osteolysis and bone resorption (Hertel and Ballmer, 2003; and Scarlat and Matsen, 2001). Metal-backed prostheses are believed to give rise to increased stress-shielding, and have been associated with higher rates of loosening than all-polyethylene prostheses in clinical studies (Martin et al., 2005; and Boileau et al., 2002), as well as experimental studies (Anglin et al., 2001). In the present project, we observed significantly increased alterations in the superior and anterior compressive principal strains, and AP translations for the Ti-step prosthesis as compared to both the STD and Poly-step prostheses, which is in concurrence with the previous studies.

5.11 Experimental Efficacy of Custom Glenoid Prostheses

Extensive biomechanical experiments are warranted to test implant efficacy in achieving the intended purpose. As already mentioned, no reference to any bench-top testing of posterior augmented implants was found in past literature. Cadaver studies involving several measurements before and after TSA, including peri-glenoid bone strains, implant strains, glenohumeral translations, joint contact areas and forces, have been executed in the past. However to our knowledge, the present project is the first study involving simultaneous measurements of peri-glenoid strains and glenohumeral translations in the presence of a biconcave defect, as seen in osteoarthritis, and following implantation of custom designed glenoid prostheses.

5.11.1 Future Direction

In the future, efficacy of the Posterior-step and other custom implants could be experimentally tested in several different ways. Further experimentations on cadaver model could utilize a more robust loading apparatus that is able to withstand greater loading forces. The
effects of errors in restoring normal glenoid version following implantation of a Posterior-step prosthesis could be examined using an experimental set up similar to that used by Nyffeler et al. The susceptibility of these implants to loosening could be assessed as per the guidelines of the American Society for Testing and Materials$^9$, as based on the recommendations by Anglin et al. The implant micromotions, and stresses within the glenoid bone, cement mantle and bone-cement interface of the Posterior-step prostheses could be modeled by FEA. Subsequently, animal models of glenohumeral OA with posterior glenoid deficiencies could be employed.

Additional variations in the prosthetic design could be considered. The step block with a flat base in our prototype design was intended to make the implant less vulnerable to shear forces imposed at the shoulder joint, as compared to the sloping wedge design of the past posterior-augmented designs. The technique used in the current project to flatten the defect base is not possible during live surgery on patients, due to lack of adequate access needed to slide the cutting assembly over the jig. Therefore, a different technique needs to be invented and a new set of tools need to be designed for the purpose.

5.12 Assumptions and Limitations

5.12.1 Assumptions

The experimental set up in the present study was assumed to adequately simulate in vivo loading environment of the shoulder in different arm positions. The actuator-driven axial loading was assumed to be equivalent to moderate deltoid action, while the static forces in the rotator cuff muscles, as generated through the application of static weights, were assumed to adequately model the normal cuff activity. The premise was that this set up is a good representation of moderate shoulder loading during normal activities of daily living.

The specimen pot orientation, with 10° of anterior scapular tilt and with the glenoid articular surface at 25° superior inclination and perpendicular to the coronal axis of the body, was
assumed to recreate the *in vivo* orientation of the glenoid articular surface. Wide variations are normally present in the relative gleno-humeral orientation in space, and the respective humeral head and glenoid versions in healthy population. Though the glenoid version existing in the specimen was not changed, rotation of the specimen pot, so as to align the glenoid articular surface parallel to the pot base under fluoroscopic guidance, was assumed to achieve “neutral version”, which is also different from the surgical practice of implanting the glenoid components in slight retroversion. This assumption may also be an oversimplification of the anatomy, but nonetheless was necessary to standardize the specimen orientation.

5.12.2 Limitations

Not all shoulder muscles were included in the model, which is one limitation of the study. It is also recognized that simulation of deltoid action by a single linear actuator is probably an oversimplification, and that the three parts of the deltoid muscle should be considered as three separate muscles from the functional standpoint. However, in several past cadaver experiments, the rotator cuff muscles and the joint capsule were removed, and the humeral head was often modeled as a metallic ball. Contrarily, the glenohumeral joint capsule was preserved intact for the initial testing in the present study (condition # IS), and with the inclusion of the deltoid and rotator cuff muscle forces, the present set up was considered an adequate model of *in vivo* shoulder loading.

The small number of cadavers in each experimental group was an obvious limitation of the study; additional cadaver specimens could have increased the statistical power and possibly revealed more definitive differences among conditions. Substantial inter-specimen variability was another obvious limitation; however, relative measurements and repeated measures ANOVA likely helped to dampen its effect. The increased noise, often observed during trials immediately following freezing the muscle-clamp junctions, may have been due to some effect of freezing on
strain gauges. Therefore, the initial trial was not included in the analysis whenever possible. The joint translations recorded under simulated in vivo loading may not be representative of the actual *in vivo* translations as the stabilizing role of soft tissues was compromised due to dissection. Only one arthroplasty system (Bigliani/Flatow®, Zimmer Inc., IN) was tested in this study.

### 5.13 Conclusion Summary

We successfully developed a physiological cadaver loading model for the shoulder joint, with the intact capsule and rotator cuff muscles. Implantation of a conventional glenoid prosthesis, in the absence of posterior glenoid deficiency, had no significant effect on joint stability or loading of the glenoid bone, as measured by humeral head translations, and peri-glenoid principal strains respectively under 200N loads. Creation of a 20° biconcave defect in the posterior glenoid, as seen in osteoarthritis, caused some significant alterations in peri-glenoid strains and joint translations. Implantation of a Poly-step glenoid prosthesis in the presence of a defect, restored the mechanical alterations due to the defect creation, while implantation of a Ti-step prosthesis did not. Following Ti-step prosthesis implantation, the anterior compressive strains and posterior tensile strains were statistically higher, and the humeral head translated more anteriorly as compared to the intact joints. Therefore, we conclude that the Poly-step glenoid prosthesis may be a viable option for treating some posterior glenoid deficiencies, and warrants further mechanical investigations, in the form of fatigue loading and testing for loosening performance etc, to evaluate this novel prosthesis. If successful, this prosthesis may provide an important treatment avenue for surgeons performing total shoulder arthroplasty on arthritic shoulder with posterior glenoid deficiency, and may also reduce the incidence of glenoid component loosening and failure.
APPENDICES

Appendix A: Sizes of Different Modular Components

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Group</th>
<th>Humeral Component</th>
<th>Glenoid Component</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>STD</td>
<td>52 × 21 × offset modular</td>
<td>STD* glenoid</td>
</tr>
<tr>
<td>2</td>
<td>STD</td>
<td>46 × 19 × offset modular</td>
<td>STD* glenoid</td>
</tr>
<tr>
<td>3</td>
<td>STD</td>
<td>52 × 21 × offset modular</td>
<td>STD* glenoid</td>
</tr>
<tr>
<td>4</td>
<td>STD</td>
<td>52 × 21 × offset modular</td>
<td>STD* glenoid</td>
</tr>
<tr>
<td>5</td>
<td>STD</td>
<td>46 × 21 × offset modular</td>
<td>STD* glenoid</td>
</tr>
<tr>
<td>6</td>
<td>Poly-step</td>
<td>52 × 21 × offset modular</td>
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</tr>
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<td>7</td>
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<td>52 × 21 × offset modular</td>
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</tr>
<tr>
<td>8</td>
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</tr>
<tr>
<td>9</td>
<td>Poly-step</td>
<td>52 × 21 × offset modular</td>
<td>Poly-step†</td>
</tr>
<tr>
<td>10</td>
<td>Poly-step</td>
<td>52 × 21 × offset modular</td>
<td>Poly-step†</td>
</tr>
<tr>
<td>11</td>
<td>Ti-step</td>
<td>52 × 21 × offset modular</td>
<td>Ti-step§</td>
</tr>
<tr>
<td>12</td>
<td>Ti-step</td>
<td>46 × 19 × offset modular</td>
<td>Ti-step§</td>
</tr>
<tr>
<td>13</td>
<td>Ti-step</td>
<td>52 × 21 × offset modular</td>
<td>Ti-step§</td>
</tr>
<tr>
<td>14</td>
<td>Ti-step</td>
<td>52 × 21 × offset modular</td>
<td>Ti-step§</td>
</tr>
<tr>
<td>15</td>
<td>Ti-step</td>
<td>52 × 21 × offset modular</td>
<td>Ti-step§</td>
</tr>
</tbody>
</table>

* Standard polyethylene glenoid component

† Posterior-step design glenoid component backed with a polyethylene step block

§ Posterior-step design glenoid component backed with a polyethylene step block
Appendix B: Moduli of bone and different relevant materials*

<table>
<thead>
<tr>
<th>Material</th>
<th>Modulus (GPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cobalt chrome</td>
<td>210-250</td>
</tr>
<tr>
<td>Stainless</td>
<td>190</td>
</tr>
<tr>
<td>Titanium</td>
<td>110-120</td>
</tr>
<tr>
<td>Trabecular titanium</td>
<td>3</td>
</tr>
<tr>
<td>Polyethylene</td>
<td>2.2-8</td>
</tr>
<tr>
<td>Bone cement (PMMA)</td>
<td>2-3</td>
</tr>
<tr>
<td>Tibial cortical bone</td>
<td>5.44</td>
</tr>
<tr>
<td>Tibial trabecular bone</td>
<td>4.59</td>
</tr>
<tr>
<td>Tibial subchondral bone</td>
<td>1.15</td>
</tr>
<tr>
<td>Glenoid subchondral bone</td>
<td>0.099 to 0.4</td>
</tr>
</tbody>
</table>

* Values taken from Mimer et al., 2008, Choi et al., 1990; and Hin TS, 2004
Appendix C: Designs of Custom Tools and Components

a. Design of the step block
b. Design of the cutting jib base
c. Design of the connecting piece
d. Design of the burr
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Curriculum Vitae

Education
2009  Ph.D. in Kinesiology, The Pennsylvania State University, University Park, USA
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2007  Congratulatory Mention, American Academy of Orthopedic Surgeons News Bulletin
2007  First Place, Penn State Department of Kinesiology Poster Exhibition
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1992  Elected as Student General Secretary, Nagpur Government Medical College

Refereed Journal Publications

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Other Invited Presentations
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