



**Effects of Scapular Notching and Bone Remodelling
on Long-Term Fixation of the Glenoid Component
in Reverse Shoulder Arthroplasty**

by

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ABSTRACT

Reverse shoulder arthroplasty (RSA) has been a proposed surgical treatment for severe rotator cuff deficiency associated with arthritis. Favourable clinical results for this type of prosthesis have been reported from short- and mid-term follow-up studies. However, the high revision rate (5% - 33%) at long-term follow up (i.e. greater than 6 years) is a concern. One of the principal factors leading to RSA failure is glenoid component loosening with an incidence of 5% - 38%. Therefore, one objective of this project is to investigate factors leading to long-term glenoid loosening. As various glenosphere positions have been proposed to minimize scapular notching, the other objective is to predict fixation strengths associated with these new surgical techniques.

Scapular notching is one of the most frequently reported complications for Delta RSA, due to the high postoperative incidence of 50% to 96%. In this thesis, the study of scapular notching showed negative effects on the inferior screw safety and safety of the bone close to the screw. The study of initial stability showed that scapular notching may not destroy the good environment for bony ingrowth.

Strain-induced bone remodelling has been an important factor for the bone loss after hip and knee joint prosthesis implantations. Effects of this factor on the bone loss after Delta RSA implantations were investigated. The results showed that bone resorption was considerable in the region close to the back of metaglene and the middle stem with a mean reduction of postoperative bone apparent density of approximately 63% at 8-year follow up. Thereby, postoperative bone loss could be caused by three factors: strain-induced bone resorption, scapular notching and osteolysis induced by the polyethylene wear particles.

In this study, prosthesis fixations in the case of inferior positioning and downward tilting of the glenosphere were assessed using two parameters: strain-induced bone resorption and initial stability. It was found that inferior positioning may lead to early bone resorption due to the inferior shift of postoperative glenohumeral force. The downward tilting may cause significant increase of bone-prosthesis micromotions and may result in poor initial stabilities of glenoid prosthesis.

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CHAPTER 1

INTRODUCTION

1.1 Background

Following knee and hip joints, the shoulder is the third most commonly replaced joint in the United States based on the PearlDiver patient record database, which contains orthopaedic patients' data collected from approximately 20 million patients annually and is provided by a consortium of insurance companies in the United States (Joshi, 2008). In another statistic report of orthopaedic joint associations in 2007, the number of shoulder joint operations also ranked the third according to the orthopaedic procedure of New Zealand (Rothwell et al., 2007). Additionally, the worldwide shoulder implants market increases dramatically with more than 15% annually (Joshi, 2008).

Shoulder joint arthroplasty is a common solution to relieve pain and restore normal shoulder function. It mainly includes three types: anatomical total shoulder arthroplasty (TSA), hemiprosthesis (HP) and reverse shoulder arthroplasty (RSA). The former two types of implants have similar constructions to the anatomical shoulder joint, while the reverse shoulder device has a reversed construction with a metal ball replacing the glenoid and a plastic socket replacing the humeral head (more implant descriptions in Chapter 3.1). This type of shoulder products have received increased worldwide attentions since the Grammont design, which showed good clinical results (Sirveaux et al., 2004; Boileau et al., 2006). Reverse shoulder prosthesis has played a significant role in the surgical treatment of severe rotator cuff deficiency and arthritis, as well as revision of total or hemi shoulder arthroplasties. In a statistic study of orthopaedic shoulder replacements from 2004 to 2007, the PearlDiver database revealed that approximately 47% of patients who were treated with a total shoulder replacement had either partial or full rotator cuff tears (Joshi, 2009). Reverse shoulder prosthesis was a common solution for the revision of the patients with massive rotator cuff (Joshi, 2009). Additionally, the reverse system products and treatments are now getting more worldwide recognitions and applications due to improved surgical techniques and prosthetic designs (Joshi, 2009). The PearlDiver database showed that requirements of RSA in the United States in 2008 was approximately 18 times of that in 2004 (Figure 1-1).

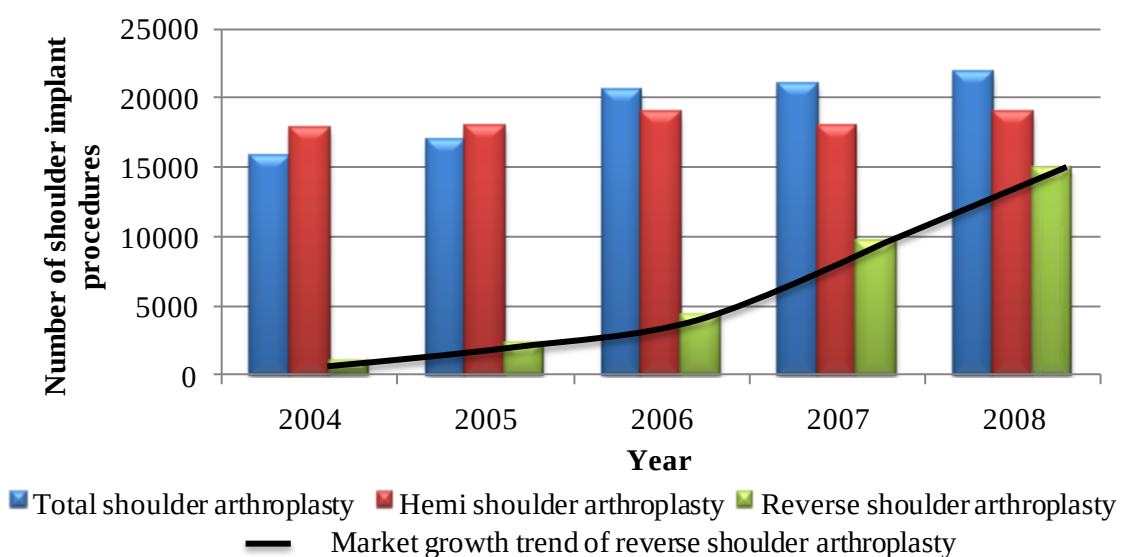


Figure 1-1 Number of shoulder arthroplasty procedures in the United States from 2004 to 2008 according to the PearlDiver patient record database (adapted from Joshi, 2009).

Though short- and mid-term studies of RSA showed favourable clinical results (Grassi et al., 2009), the high revision rate (5% - 33%) (Appendix A) of this type of prostheses at long-term follow up are concerning (Frankle, 2005; Hatzidakis, 2005; Lam, 2007). In a study of 80 Grammont RSAs in the treatment of glenohumeral osteoarthritis with severe rotator cuff deficiency, the survivorship-to-time curve when implant loosening was defined as the end point (Figure 1-2) showed a quick decrease after a mean follow up of approximately 80 months after implantations (Sirveaux et al., 2004). With a similar survivorship analysis for reverse devices for five to ten years (Guery et al., 2006), the survival rate of RSA dropped considerable after 120 months to 84% when loosening of the implant was defined as revision, and to only 58% when a constant score of smaller than 30% was considered to be implant failure. Guery et.al also pointed out that a progressive deterioration of shoulder function was observed after six-year follow up, and made a warning for the application of RSA for young patients (Guery et al., 2006). In a study of 225 RSA operations in Norway, the failure rate was 10% at 5-year follow up and 22% at 10-year follow up (Fevang et al., 2009).

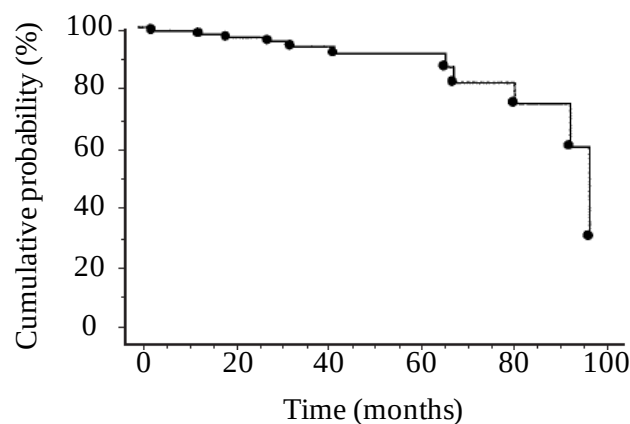


Figure 1-2 Kaplan-Meier survivorship curve for the Grammont RSA implant with loosening as the implant failure (adapted from Sirveaux et al. 2004).

Failure of RSA could associate with several factors, such as deep infection, dislocation, late acromial fracture, and glenoid loosening (Boileau et al., 2005; Naveed et al., 2011). Based on several failure studies of reverse shoulder devices, aseptic glenoid loosening which is associated with 10% to 100% of the postoperative implant reoperations (Appendix A), is of great concerned (Fevang et al., 2009; Grassi et al., 2009). In a study of 3 failed Grammont RSA at a mean follow up of 44 months, 2 of them were revised due to the glenoid loosening (Sirveaux et al., 2004). In another study of Delta III RSA

for 6 years, two implants were revised, and both of them were due to the aseptic glenoid loosening (Grassi et al., 2009). Another report about the revision of RSA showed that aseptic glenoid loosening was the cause in approximately 36% of cases (Fevang et al., 2009).

Factors leading to the long-term glenoid prosthesis loosening are unknown. Effects of postoperative bone loss after the RSA implantation is a cause of concern (Sirveaux et al., 2004; Boileau et al., 2005; Simovitch et al., 2007; Lévine et al., 2008; Lévine et al., 2010). Scapular notching (Boileau et al., 2005), which is initially caused by the impingement of the humeral cup against the scapular neck in a resting position and strain-induced adaptive bone resorption (Ahir and Walker, 2004) may be the causes of observed bone loss. Several approaches have been published to minimize the scapular erosions induced by the bone-prosthesis impingements (Nyffeler et al., 2005; Kelly II et al., 2008; Kontaxis, 2009). For instance, the glenoid component is positioned as inferior as possible, or inferiorly tilted (Nyffeler et al., 2005). However, no published articles quantitatively investigate the association between the scapular notching and the failure of RSA. For the study of postoperative strain-induced bone remodelling, only one published article reported variations of postoperative stress distributions around the glenoid after Bayley-Walker RSA implantations (Ahir and Walker, 2004). However, it was still unclear how stress shielding relates to the failure of glenoid prosthesis. This thesis is aimed to quantitatively predict and analyse the relationship between scapular notching, bone remodelling and the fixation of glenoid component in the Grammont RSA. In addition, the purpose of this study is also to predict the fixation of glenoid prostheses in the case of some recently proposed surgical techniques, such as various positions of the glenoid component.

1.2 Aims and Objectives

Despite the successful short- and middle-term applications of RSA, long-term survivorship of this type of implant is concerned, due to the high rate of aseptic loosening of glenoid prosthesis (Sirveaux et al., 2004; Guery et al., 2006; Fevang et al., 2009; Grassi et al., 2009). The primary objective of this project is to analyse factors relating to the fixation of glenoid component in the Grammont RSA. Delta CTA RSA

supplied by DePuy (DePuy Ltd.) was used in current study. DePuy is one of leaders in the global shoulder repair market, and Delta RSA is the main reverse system product in Depuy. Another objective of this study is to predict the fixation of Delta CTA RSA in the case of recently proposed modified implant positions with relative to the glenoid. The investigation of current study can be broken down into a number of smaller, incremental objectives:

- i) Analyses of bone osseointegration to predict the initial stability of glenoid component in Delta CTA RSA.
- ii) Investigation of the relationship between the scapular notching and the fixation of glenoid prosthesis.
- iii) Biomechanical analyses of postoperative adaptive bone remodelling and prediction of the association between this postoperative complication and the fixation of glenoid prosthesis.
- iv) Prediction and analyses of the fixation of glenoid prosthesis in the case of recently proposed surgical techniques—various positions of the glenoid component.

1.3 Outline of the Thesis

The entire thesis mainly includes four parts (Figure 1-3): introduction and research background, calibration and validation of finite element modelling, analyses of factors inducing the glenoid loosening of RSA, and finally, conclusions and future work. The structure of this thesis is organised as follows:

- **Chapter 1** introduces objectivities of this study and describes structures of the thesis.

- **Chapter 2** consists of the functional anatomy of shoulder and bone biomechanics. This chapter includes the terminology used in this thesis and a broad overview of the gross biomechanics of bone, especially the scapula.
- **Chapter 3** is an overview of shoulder pathologies and prostheses. In this chapter, a historical review of shoulder prostheses is presented. RSA is comprehensively introduced in several aspects: mechanical design of Grammont RSA, advantages of this type of design, complications and fixation methods.
- **Chapter 4** presents the finite element modelling which would be used for the studies in next a few chapters. It includes calibrations and sensitivity analyses of the finite element modelling.
- **Chapter 5** compares predicted finite element strains and micromotions with experimental measurements in order to validate finite element modelling.
- **Chapter 6** investigates associations between scapular notching and aseptic glenoid component loosening in RSA. In this chapter, the glenoid loosening was evaluated by three aspects: effects on inferior screw breakages, potential risk of postoperative bone fracture at the bone-to-screw interface, and the initial stability of glenoid prosthesis.
- **Chapter 7** predicts the postoperative adaptive process of bone remodelling after the implantation of RSA and analyzes the correlation between bone resorption and fixation of glenoid prosthesis.
- **Chapter 8** predicts effects of recently proposed implant positions on the fixation of glenoid prosthesis. Postoperative bone resorption and initial stability of glenoid prosthesis are presented.
- **Chapter 9** finally summarizes the main conclusions drawn from the present research, and suggests some recommendations for future work.

Thesis structure

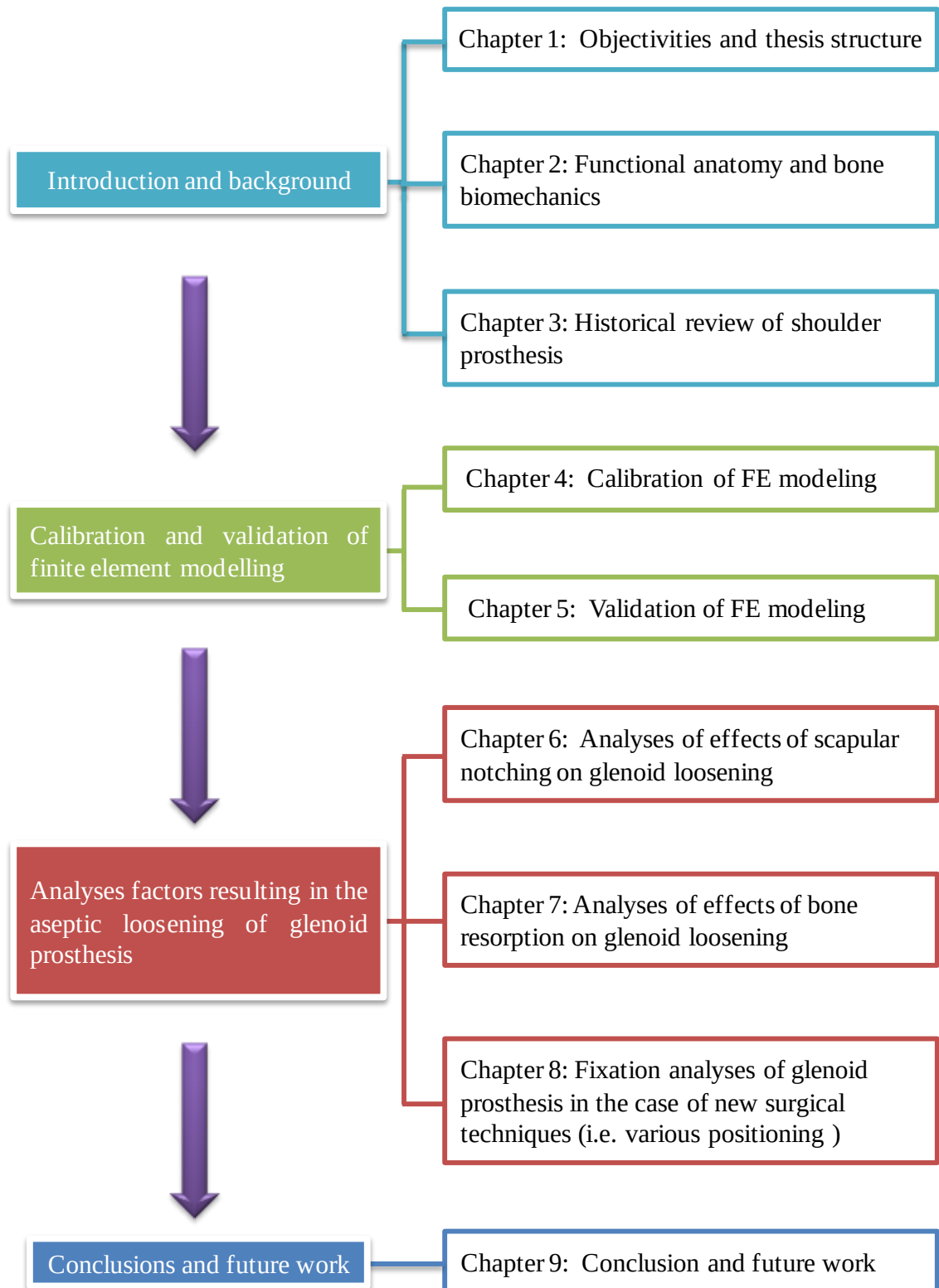


Figure 1-3 Thesis structure.

CHAPTER 2

MUSCULOSKELETAL ANATOMY AND BONE BIOMECHANICS

2.1 Musculoskeletal and Functional Anatomy of Shoulder

2.1.1 Clinical Terminology

2.1.1.1 Spatial Terminology

2.1.1.1.1 *Anatomical directional terms*

Directional terms describe a position of a body part or an external object with respect to the body. With a scapula being concerned, the anatomical positions are shown in Figure 2-1.

Superior or cranial: closer to the head.

Inferior or caudal: farther away from the head.

Anterior or ventral: toward the front of the body.

Posterior or dorsal: toward the back of the body.

Medial: toward the midline of the body.

Lateral: away from the midline of the body.

Proximal: toward or nearest the trunk or the point of origin of a part (for example, the knee is proximal to ankle).

Distal: away or farthest from the trunk or the point of origin of a part (for example, the wrist is distal to the elbow).

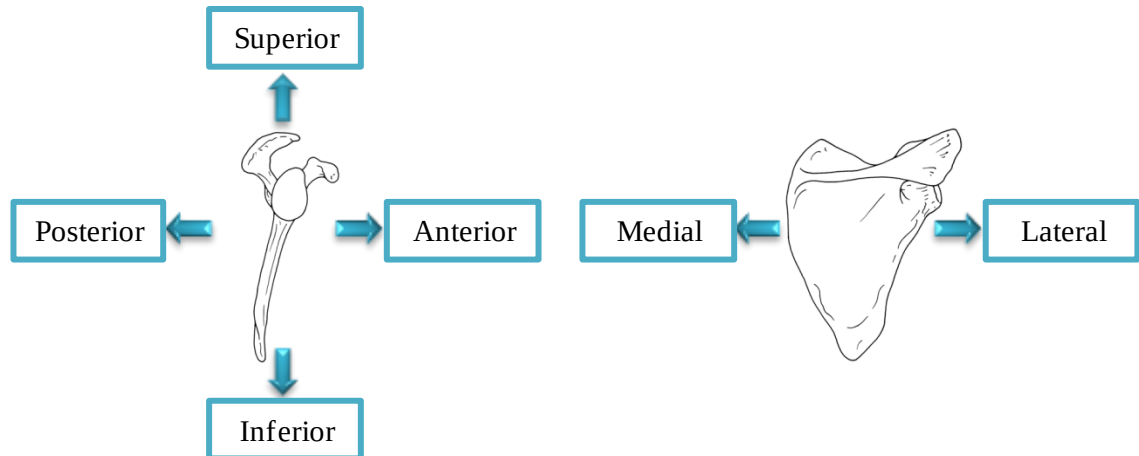


Figure 2-1 Anatomical position of a scapula.

2.1.1.1.2 *Anatomical reference planes and axes*

Anatomical reference planes consist of three imaginary two-dimensional surfaces (Figure 2-2): coronal plane, sagittal plane and transverse plane. With reference to these planes, the three main axes are mediolateral axis, anteroposterior axis and longitudinal axis.

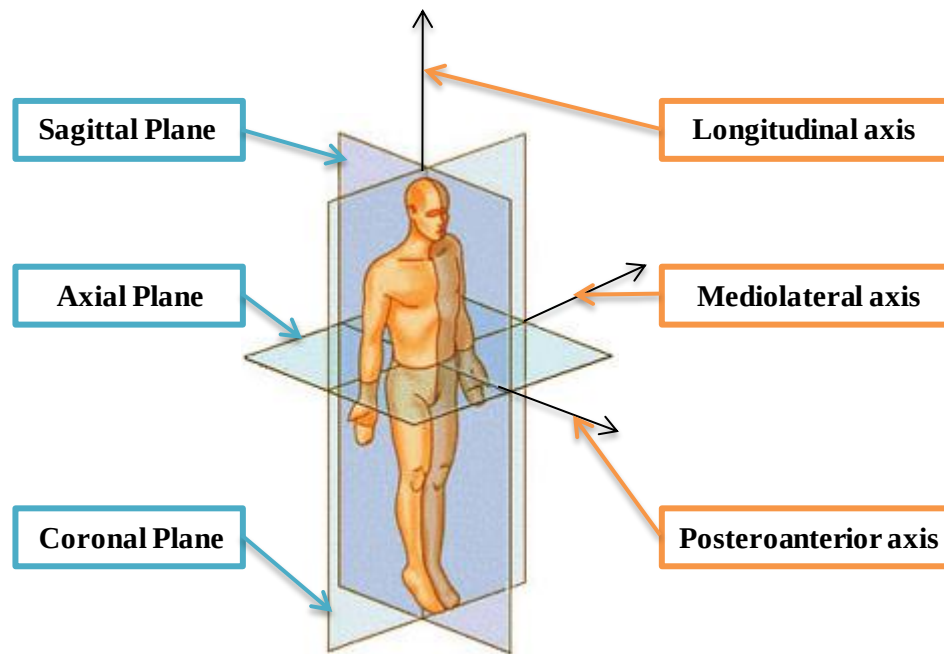


Figure 2-2 Anatomical reference planes and axes.

Coronal Plane (Frontal Plane): the vertical plane dividing the body or any of its parts into anterior and posterior portions.

Sagittal Plane (Lateral Plane): the vertical plane dividing the body or any of its parts into right and left sides.

Axial Plane (Transverse Plane): the horizontal plane dividing the body or any of its parts into upper and lower parts.

Mediolateral axis (frontal-horizontal axis): the horizontal axis that is perpendicular to the sagittal plane.

Anteroposterior axis (sagittal-horizontal axis): the horizontal axis that is perpendicular to the frontal plane.

Longitudinal axis (vertical axis): the vertical axis that is perpendicular to the transverse plane.

2.1.1.2 Motion Terminology

Gross movements of the upper arm are presented in Figure 2-3.

Abduction: the movement away from the trunk in the frontal plane.

Adduction: the movement toward the trunk in the frontal plane.

Flexion: anteriorly directed sagittal plane rotations of the arm.

Extension: posteriorly directed sagittal plane rotations of the arm.

Horizontal flexion: movement of the arms from the side horizontal position to the front horizontal position.

Horizontal extension: movement of the arms from the front horizontal position to the side horizontal position.

Internal rotation: the axial twist toward the body in the shoulder.

External rotation: the axial twist away from the body in the shoulder.

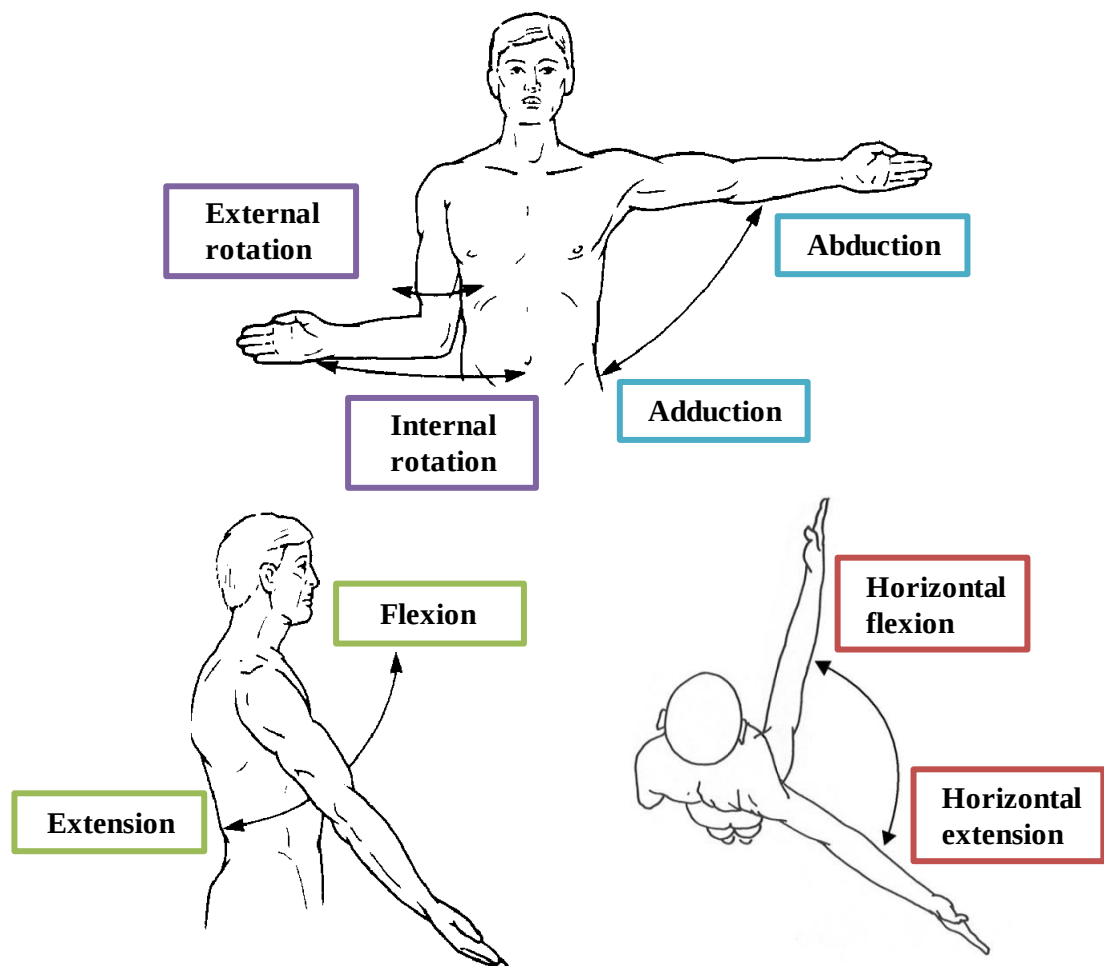


Figure 2-3 Motion of the upper arm.

2.1.2 Skeletal Structure of the Shoulder

The shoulder connects the upper limb to the trunk and the neck. The bone framework of the shoulder consists of the clavicle, the scapula and the proximal end of the humerus. The clavicle and the scapula form the pectoral girdle, which is also named as shoulder girdle.

2.1.2.1 Clavicle

The clavicle (collar bone) (Figure 2-4) is an S-shaped bone. It directly connects the axial skeleton to the upper limb. The lateral end of the clavicle is constrained to move with regards to the scapula and thus enables the forces to be transmitted from the trunk to the upper limb. The clavicle articulates with the superior and lateral border of the manubrium of the sternum at the sternoclavicular joint, and with the acromion of the scapula at the acromioclavicular joint.

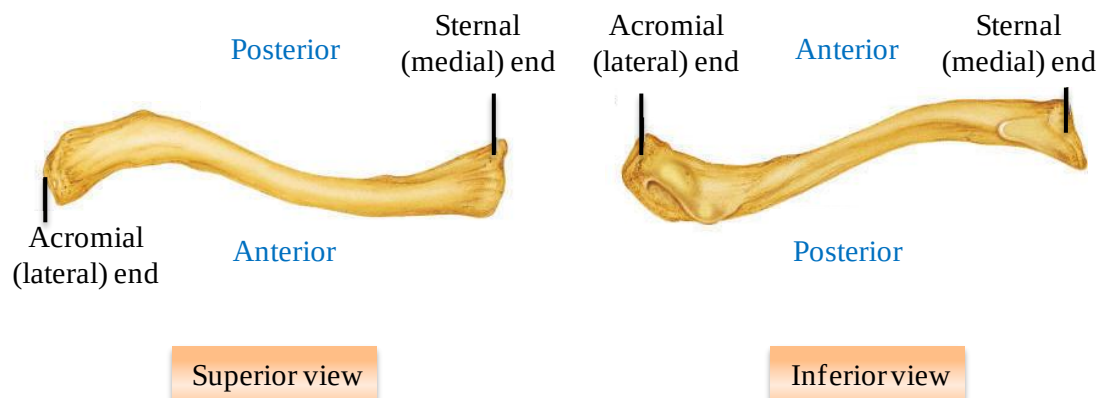


Figure 2-4 The clavicle from the superior and inferior views.

2.1.2.2 Scapula

The scapula (Figure 2-5) is a broad triangular blade lying on the posterior chest wall. It inclines approximately 30° to 40° anteriorly to the coronal plane from the transverse view (Saha, 1950). The anterior surface of the scapula presents a broad concavity, giving the origin to the subscapularis muscle. The posterior surface of the scapula is divided by the scapular spine into two fossae: the supraspinatus fossa and the infraspinatus fossa. These two fossae give origins to the supraspinatus and infraspinatus muscles, respectively. The three sides of the scapula are named superior, medial and

lateral borders with the corners being termed the superior, inferior and lateral angles. Adjacent to the lateral angle, the scapula thickens and forms the scapular neck, where scapular notching often occurs. The irregular pear-shaped shallow glenoid cavity is next to the scapular neck. It superiorly inclines 3° to 5° (Basmajian and Bazant, 1959), and retroverts 2.5° to 12.5° (Friedman et al., 1992).

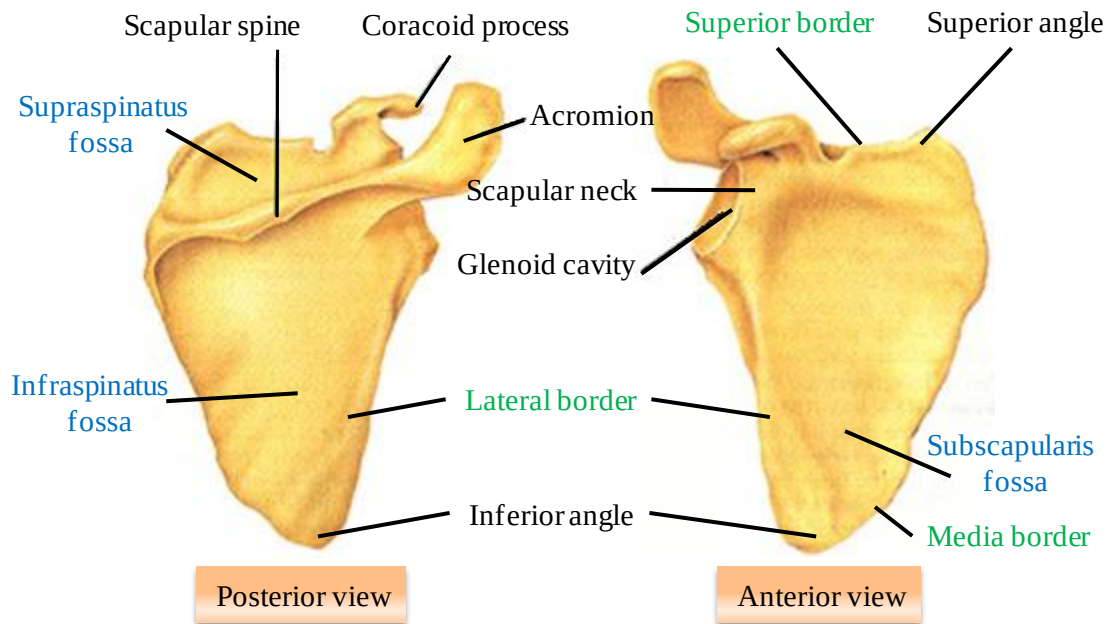


Figure 2-5 The scapula from the posterior and anterior views.

2.1.2.3 Humerus

The humerus is a long cylindrical bone, and articulates with the glenoid cavity at the glenohumeral joint (Figure 2-6). The proximal humeral head is elliptically-shaped and orients medially, superiorly, and posteriorly. It inclines 120° to 145° relative to the shaft (Iannotti et al., 1992) and retroverts within a range of 60° (Roberts et al., 1991). The greater tubercle is a projection locating on the lateral humeral head. It is separated from the lesser tubercle, which is a projection laying on the anterior and medial surface of the epiphysis, by the intertubercular groove (also termed bicipital groove). Both these tubercles are important sites for insertions of tendons, such as those of the supraspinatus, infraspinatus, teres minor and subscapularis muscles. The normal humeral arm is able to elevate 120° before the greater tuberosity impinges on the acromion (Roberts et al., 1991). Two necks in the humerus are the anatomical and surgical necks. The anatomical neck affords attachment to the articular capsule, and the surgical neck corresponds to the metaphysis of the growing bone.

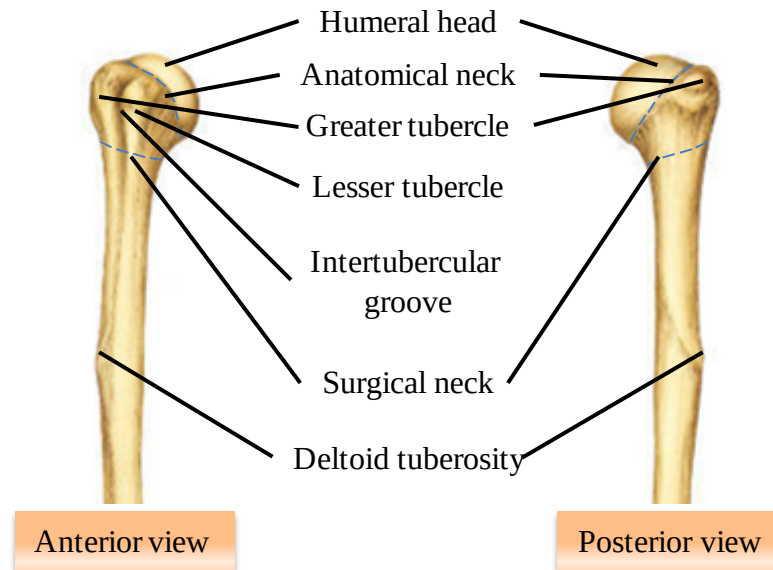


Figure 2-6 The proximal humerus from the anterior and posterior views.

2.1.3 Articulations of the Shoulder

The shoulder consists of the sternoclavicular joint, acromioclavicular joint, glenohumeral joint and scapulothoracic articulation (Figure 2-7). It provides a vast freedom of movements. The arm allows 180° of elevation, approximately 150° of internal and external rotations, and 170° of flexion and extension (Inman et al., 1944).

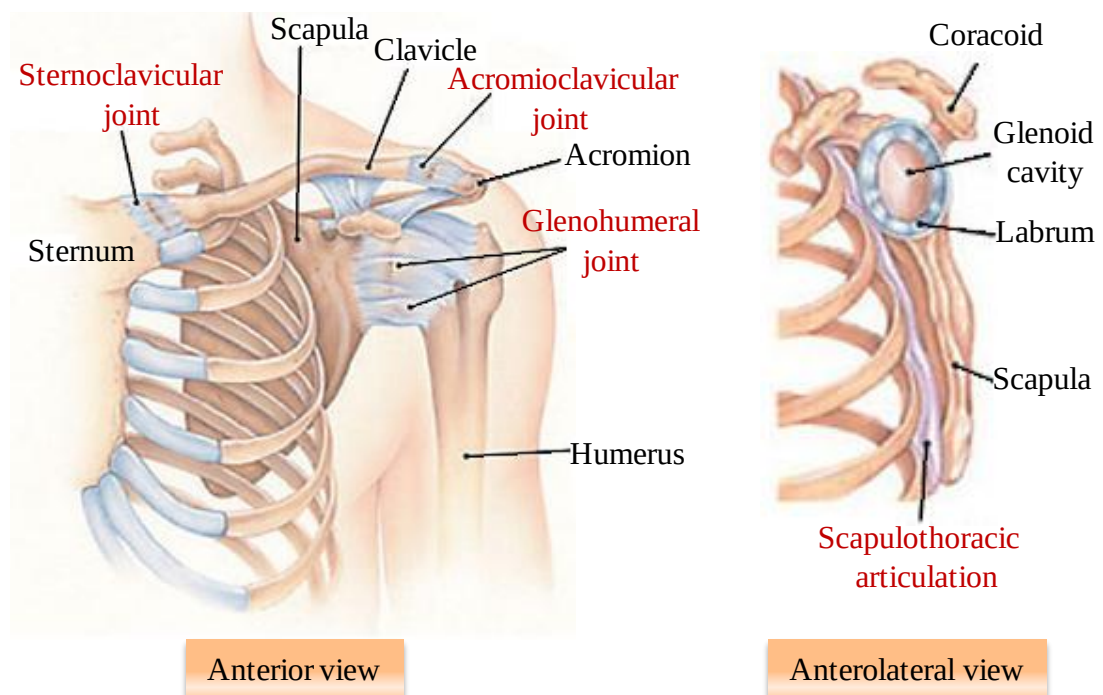


Figure 2-7 Schematic description of bony structures of the shoulder.

2.1.3.1 Sternoclavicular Joint

At the sternoclavicular (SC) joint (Figure 2-8), the medial end of the clavicle articulates with the posterolateral facet of the manubrium and with the cartilage of the first rib. The SC joint links the upper extremity directly to the thorax and allows up to 50° of axial rotation and 35° of elevation (Inman et al., 1944). The sternoclavicular joint itself has little intrinsic bony stability. The stability is primarily provided by the articular disc, anterior and posterior sternoclavicular ligaments, interclavicular ligament and costoclavicular ligament. Costoclavicular ligament links the undersurface of the medial end of the clavicle and the first rib. It is the major constraint to the SC joint motion.

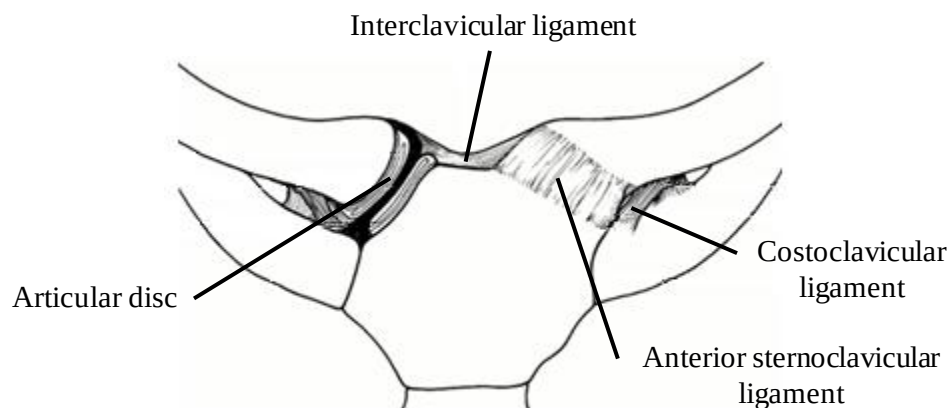


Figure 2-8 Schematic figure of sternoclavicular joint.

2.1.3.2 Acromioclavicular Joint

The acromioclavicular (AC) joint is a synovial joint. Unlike the SC joint which is composed of a ball and a socket, the AC joint has a planar configuration. The AC joint is composed of the lateral end of the clavicle and the acromion. It transmits forces from the chest musculature to the upper extremity. The AC joint is extremely stable and attributes to synchronous clavicular and scapular rotation (Anthony and Brian, 2006).

2.1.3.3 Glenohumeral Joint

Glenohumeral (GH) joint (Figure 2-9) plays a central role in the overall function of the shoulder complex. It is a ball-and-socket synovial joint and includes the head of the humerus and the glenoid cavity of the scapula. The shallow glenoid fossa covers only 25% to 30% of the humeral head (Anthony and Brian, 2006), and enables the humerus

to move across its surface. Thus, the GH joint is one of the most freely moving joint in the human body. Given such little inherent bony constraint, the stability of the GH joint is provided by the labrum, the articular capsule, and the surrounding muscles. The glenoid labrum, which is a fibrocartilaginous lip extending past the bony rim of the glenoid cavity, deepens the socket to improve the stability of the GH joint. The further static stability of the GH joint is provided by the coracohumeral ligament and the capsular-ligamentous complex (Lucas, 1973). The coracohumeral ligament arises from the lateral border of the coracoid process and inserts into the greater tubercle of the humerus. The capsular-ligamentous complex originates from the scapula and inserts into the anatomical neck of the humerus. With the help of associated tendons, the muscles which are referred as the rotator cuff are the most significant dynamic stabilizers during the shoulder motion. Function of the rotator cuff will be specified in Chapter 2.1.4. Due to the large mobility, there are a relatively large number of bursae to reduce the frictions between the muscles and tendons passing across the joint capsule.

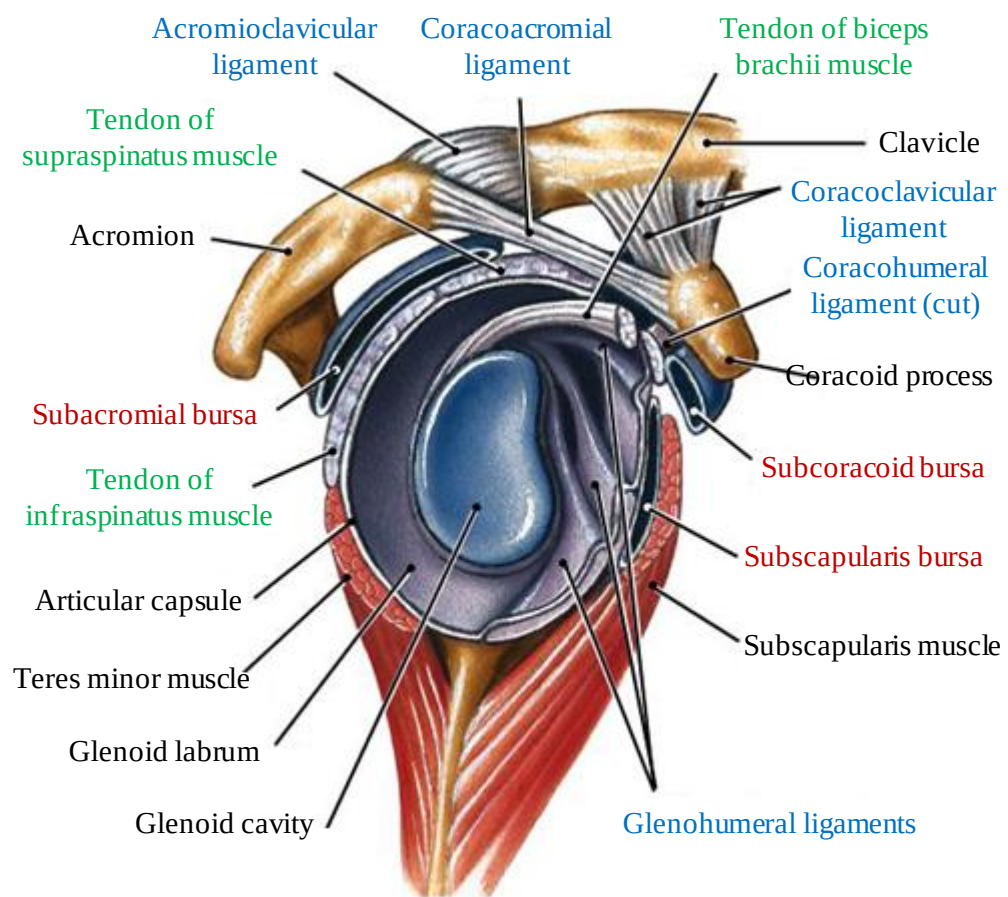


Figure 2-9 Lateral view of shoulder joint without the presence of humerus (adapted from Martini, F. H., 2005).

2.1.3.4 Scapulothoracic Articulation

Unlike the three synovial joints described in previous sections, scapulothoracic articulation is a bone-muscle-bone articulation between the scapula and thorax. This musculoskeletal connection allows the scapula to glide over the thorax during upper limb movement and other wide range of motion. Even though the contribution from scapulothoracic articulation and the glenohumeral joint varies with the arm position and the specific task being performed, the average ratio of these two articulation movements is 1:2 (Anthony and Brian, 2006).

2.1.4 Functional Muscles of the Shoulder

Muscles acting upon the shoulder complex provide both the mobility and dynamic stability. Muscle functions are highly associated with the origins and insertions. They are described from the anterior and posterior views next (Figure 2-10).

From the anterior view, the outer layer muscles (Figure 2-10 (a)) involves the deltoid and pectoralis major muscles. The deltoid muscle originates in three distinct sets of fibers (namely anterior, middle and posterior fibers), and inserts into the deltoid tuberosity of the humerus. The anterior fibers originate from the anterior border and upper surface of the lateral third of the clavicle, thus work as strong flexors and internal rotators of the humerus. The middle fibers originate from the lateral acromion and play a significant role in the arm abduction, especially beyond the initial 15° of arm rotation (Inman et al., 1944). The posterior fibers originate from spine of the scapula and function as extensors and external rotators. The deltoid muscle is utilized more efficiently in the reverse shoulder arthroplasty than in the anatomical shoulder prosthesis to power and position the arm. The pectoralis major muscle lies over the anterior chest wall and has two heads. The clavicular head originates from the side of the clavicle and the sternocostal head originates from the sternum, manubrium, and the upper costal cartilages. Both these heads insert into the intertubercular groove of the humerus and function as adductors and elevators of the humerus. In the inner layer, the pectoralis minor muscle (Figure 2-10 (b)) which lies deep to the pectoralis major functions as an important scapular stabilizer. The subclavius muscle lies inferior to the clavicle and assists in clavicular motions. It has a tendinous origin from the

anteromedial aspect of the first rib and inserts on the undersurface of the medial clavicle. The subscapularis muscle lies in the anterior scapular region and contributes to the arm rotation.

From the posterior view, the outer layer muscles include the trapezius, serratus anterior, latissimus dorsi and posterior deltoid muscles (Figure 2-10 (c) and (d)). The trapezius muscle originates along the midline of the neck and the back and inserts on the clavicles and the scapular spine. The trapezius muscle can contract independently, and thus has quite varied actions (Martini, 2001). It is a powerful elevator and a strong rotator of the shoulder. The serratus anterior muscle lies on the chest. It originates along the anterior surfaces of several ribs and inserts along the anterior margin of the medial border of the scapula. When the serratus anterior muscle contracts, it protracts the scapula and swings the shoulder anteriorly. The latissimus dorsi muscle originates along the thoracic vertebrae at the posterior midline and inserts onto the intertubercular groove of the humerus. It produces adduction and medial rotation of the humerus, collaborating with pectoralis major muscle.

Under the superficial muscles is the posterior inner layer muscles, which locate deep to the trapezius and deltoid muscles (Figure 2-10 (c) and (d)). The inner muscles include the supraspinatus, infraspinatus, teres minor and teres major muscles. The supraspinatus and infraspinatus muscles originate from two large fossae above and below the spine and form tendons inserting on the greater tubercle of the humerus. The supraspinatus produces abduction of the arm, especially at the initial 10° to 15° of elevation (Inman et al., 1944). Both the infraspinatus and teres minor muscles contribute to the arm rotation and the shoulder joint stabilisation. The teres minor muscle originates from a flattened area adjacent to the lateral border of scapula and inserts on the inferior facet of the greater tubercle of the humerus. The teres major muscle attaches along the inferior angle of the scapula on the posterior surface, and passes superiorly and laterally to the medial lip of the bicipital groove of the humerus. Thus teres major produces medial rotation and extension of the humerus.

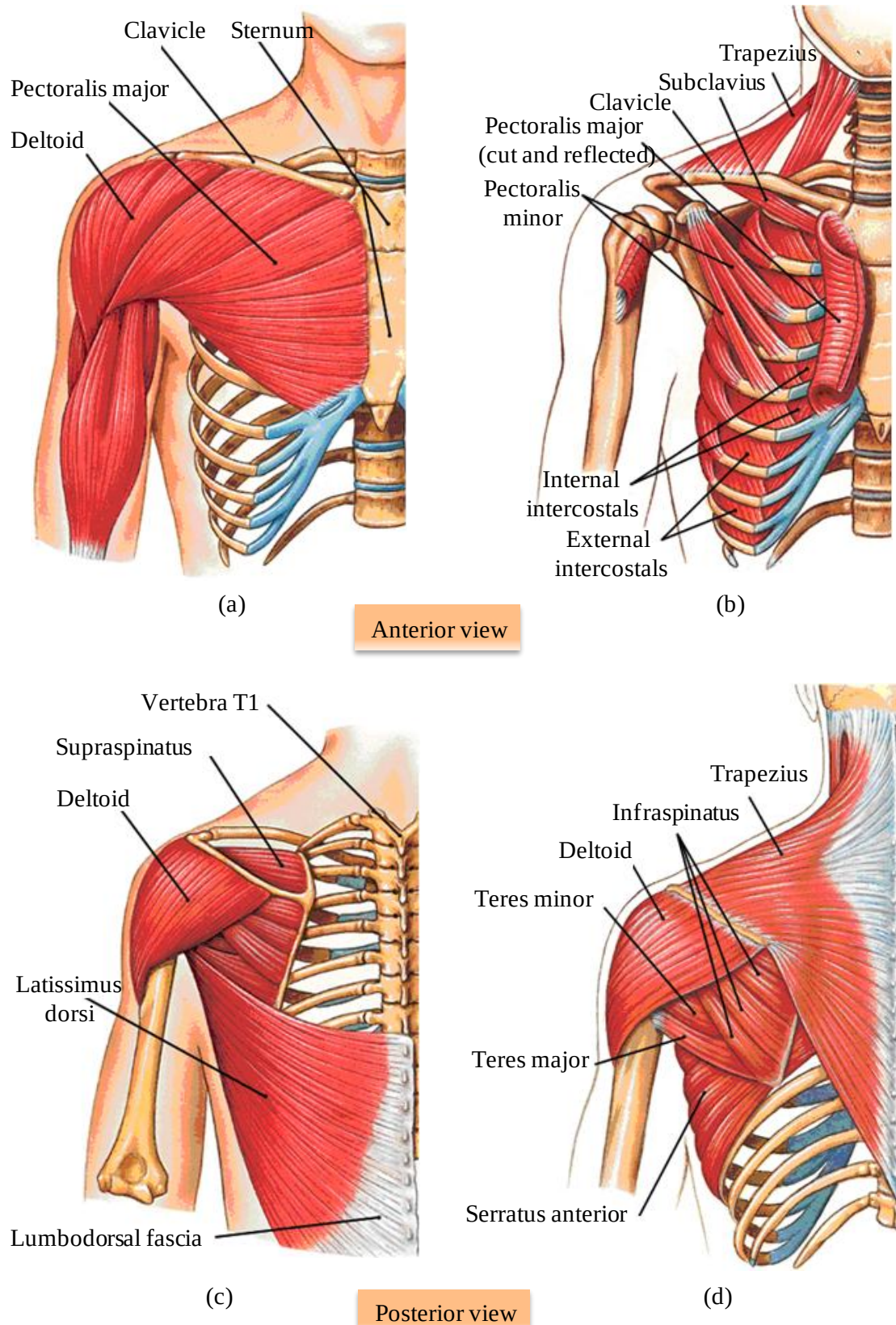


Figure 2-10 Muscles attaching on the shoulder complex: (a) and (b) from anterior view; (c) and (d) from posterior view (adapted from Martini, F. H., 2005).

The supraspinatus, infraspinatus, subscapularis, and teres minor muscles and their associated tendons form the rotator cuff (RC) (Figure 2-11). The RC muscles work as an abductor and rotator of the humerus and an important stabilizer of the glenohumeral joint by providing both passive muscle tension and dynamic contraction. In the case of large rotator cuff tears, especially the loss of supraspinatus, the resultant GH force will be altered from toward the glenoid fossa in the normal shoulder (Figure 2-12 (a)) to toward the superomedial direction (Figure 2-12 (b)), thus will lead to excessive superior translation. This translation may be up to 12 mm in some cases (Anthony and Brian, 2006) and may lead to subacromial impingement, thus result in erosions of the anterior acromion, the acromioclavicular joint, and the glenoid (more descriptions about severe rotator cuff deficiency are introduced in Chapter 3.2.2).

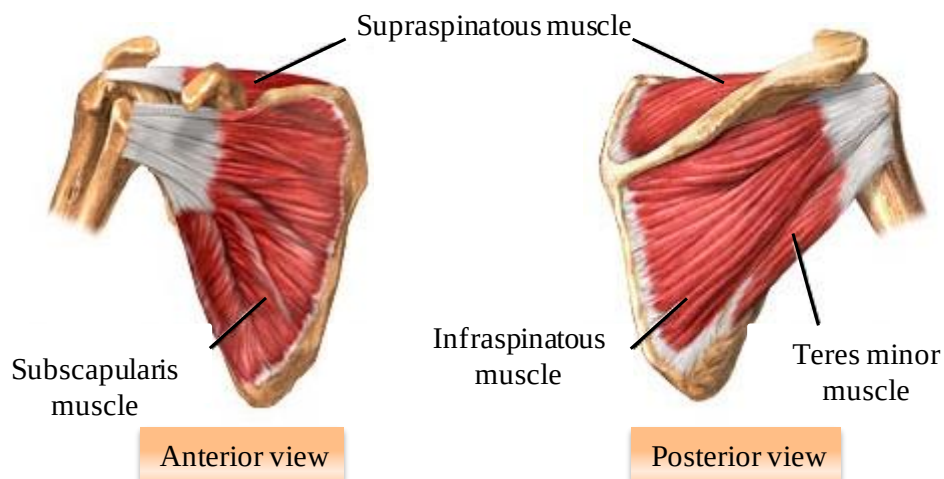


Figure 2-11 Rotator cuff muscles.

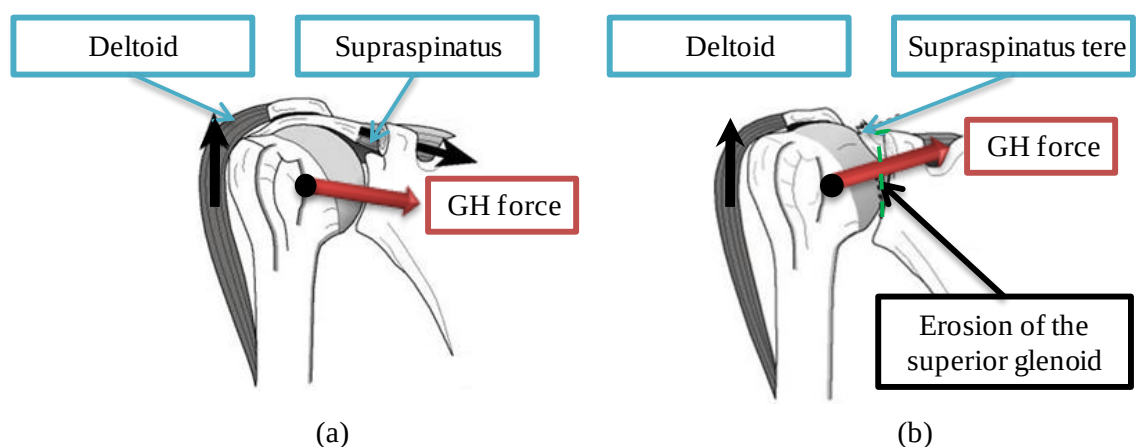


Figure 2-12 Schematic description of the glenohumeral (GH) force in (a) the normal shoulder and (b) the shoulder with severe rotator cuff deficiency.

2.2 Bone Biomechanics

2.2.1 Composition and Architectures of Bone

Bone, which consists of mineral, organic matrix, cells and water, is the main constituent of the skeletal system and differs from the connective tissues in rigidity and hardness. The strong and hard characteristics of bone are owing to its high calcium contents in the mineral phase, such as calcium carbonate and calcium phosphate. Bone is categorized into cortical bone and trabecular bone (also called cancellous or spongy bone) by the porosity, with trabecular bone being more porous. Both cortical and trabecular bones are basically organized by woven and lamellar bone, but in different structural arrangements, thus they represent different characteristics and are responsible for various functions. Dense and solid cortical bone covers approximately 80% of the mass of an adult human skeleton. It protects the soft tissues, supplies the framework for the bone marrow, and transmits forces of muscular contraction from one part of the body to another during movements. Trabecular bone covers the remaining 20% of bone mass, and lies in the inner part of bone. It is a three-dimensional, interconnected network of trabecular rods and plates which are known as trabeculae. The network of interconnected rods is termed as open cell structure and is more porous than the network of plates, which is termed as closed cell structure. Distributions of cortical and trabecular bone are subject dependent, and also relate to bone architecture, functions, ages and bone situations (i.e. bone fractures) (Cowin, 2001).

2.2.2 Mechanical Properties of Bone

Bone mechanical properties are mainly determined by the composition and structure (Keaveny and Hayes, 1993). On a length scale of the order of micrometers, bone tissue behaves like a fibre-reinforced composite. Its hydroxyapatite mineral content makes the bone stiff and its collagen fibre content improves the tensile strength and ductility to reduce bone brittle cracking. As a result of a longitudinal alignment of osteons, which are the fundamental functional units of cortical bone, bone presents greater stiffness and strength along the longitudinal direction than those in the transverse plane. In the honeycomb cell structure with mineralised connecting walls/struts (trabeculae), the

alignment of the trabeculae determines whether the structure transfers load by axial deformation (high alignment) or bending (low alignment).

The various porosity of trabecular bone can be quantified by apparent density which is the ratio of the wet mass of bone tissue divided by the tested specimen's bulk volume, which includes mineralized bone and marrow space. Due to the porous characteristics, apparent density of trabecular bone varies in a big range from 0.1 to 1.0 g/cm³ (Huiskes and van Rietbergen, 2005). The peak apparent density of solid cortical bone is approximately 1.8 g/cm³. The Young's modulus of cortical bone ranges from 11 GPa to 17 GPa (Reilly and Burstein, 1975). The ultimate tensile strength of cortical bone varies from 51 MPa in the transverse plane to approximately 133 MPa in the longitudinal direction. The ultimate compressive strength was between 133 MPa and 193 MPa (Reilly and Burstein, 1975). The strength-density relationship of trabecular bone remains an open question (Keaveny and Hayes, 1993). In a study of trabecular bone, Galante (1970) demonstrated that bone strength was proportional to apparent density. In another study of human vertebral trabecular bone, Kopperdahl (1998) reported that yield compressive and tensile stresses strongly correlated with the apparent density with the relationship shown in Equation 2-1 and Equation 2-2. He also pointed out that yield tensile strains were independent of the apparent density, but the yield compressive strains were proportional to the apparent density in the human bone.

$$\sigma_{cy} = 32.6\rho^{1.6} \quad \text{Equation 2-1}$$

$$\sigma_{ty} = -0.13 + 10.1\rho \quad \text{Equation 2-2}$$

where σ_{cy} is the compressive yield stress, σ_{ty} is the tensile yield stress, and ρ is the density.

2.2.3 Glenoid Architecture and Mechanical Properties

As the porosity of cancellous bone is highly anatomic site dependent, it is very important to understand the architecture of glenoid when analysing the mechanical parameters of glenoid trabecular bone (Mansat et al., 1998; Anglin et al., 1999; Mimar et al., 2008). From the view of sagittal plane in the subchondral bone (Figure 2-13 (a)), struts interconnect in a honeycomb fashion in the centre and change into superior-inferior directed parallel trabeculae in the anterior and posterior regions with decreased

porosity (Frich et al., 1998). From the transverse view passing through the spine of the scapula (Figure 2-13 (b)), platelike trabeculae align and are perpendicular to the articular surface. The thin rods in the anterior and posterior regions directly connect the subchondral plate to the semicircular cortical base. In the middle portion, the bone presents high porosity (Frich et al., 1998; Mimar et al., 2008).

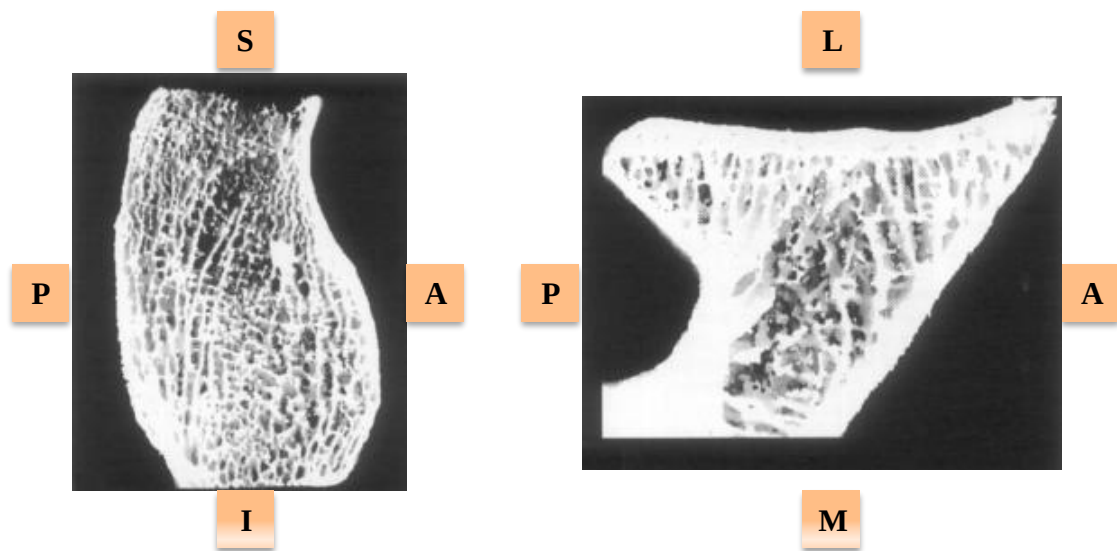


Figure 2-13 (a) The juxtaarticular subchondral area from the sagittal view; (b) glenoid bone from the transverse view; S/P/A/I/L/M: Superior/Posterior/Anterior/Inferior/Lateral/Medial (adapted from Frich et al., 1998).

Mechanical properties for the glenoid are highly associated with its architecture (Anglin et al., 1999; Lim et al., 2002; Lehtinen et al., 2004). As a result of high alignments of trabeculae in the peripheral regions in the glenoid, bone there presents great strength and high stiffness. Additionally, the peripheral glenoid bone tends to be more anisotropic than the bone in the middle portion (Frich et al., 1997; Mansat et al., 1998). From the lateral to medial view, the scapula bone is orderly the stiff subcondral bone, porous cancellous bone, and highly dense cortical bone. In an in-vitro study of strength and Young's modulus with depth in the glenoid, the strength reduced approximately 77% at a level of 3.5 mm beneath the subchondral layer, and further decreased to a deeper layer, then increased to the cortical bone layer (Anglin et al., 1999). The entire variation of strength and modulus is in a V-shaped pattern (Anglin et al., 1999). In the variously regional tests, the glenoid bone was the strongest in the posterosuperior

portion and the weakest in the anteroinferior portion (Frich et al., 1997; Mansat et al., 1998; Lehtinen et al., 2004; Mimar et al., 2008). In a study of the glenoid architecture, the isotropic bone at the centre has a capacity of absorbing and conveying heavy loads from the articular surface. The strong perapieral trabecular bone has a capability of absorbing the compressive stresses generated in general arm movements (Frich et al., 1998).

Although individual glenoids show a similar architecture (Frich et al., 1998; Mimar et al., 2008), Young's moduli and strength of individual cancellous bone vary in big ranges from 99 MPa to approximately 264 MPa and from 26 MPa to approximately 110 MPa, respectively (Frich et al., 1997; Anglin et al., 1999; Mimar et al., 2008). By investigating mechanical properties of ten glenoids, Anglin et al. (1999) pointed out that the mean Young's modulus ranged from 67 MPa to 171 MPa for the individual glenoids, with the average being 99 MPa. For the strength, they reported that the individual value varied from 6.7 to 17 MPa. Both Young's moduli and strength of cancellous bone are region dependent (Frich et al., 1997; Anglin et al., 1999). By investigating nine cadaveric glenoids (mean age, 82) with an indentation test, the regional Young's modulus was reported to range from 119 MPa to 234 MPa. In addition, the regional ultimate strength varied from 26 MPa to 67 MPa (Mimar et al., 2008). In another similar study, the Young's modulus was reported to vary from approximately 100 MPa at the glenoid bare area to 400 MPa at the superior part of the glenoid (Frich et al., 1997). From the view of glenoid depth, the mean ultimate strength reduced from approximately 13 MPa at the level beneath the subchondral plate to 7 MPa in the middle centre of the glenoid (Anglin et al., 1999). In another study, the strength decreased by 25% at the level of 1 mm underneath the subchondral plate and by 70% at 2 mm level. The modulus and strength also relate to bone ages, gender and healthy conditions (Keaveny and Hayes, 1993; Anglin et al., 1999; Lim et al., 2002; Kalouche et al., 2008).

In comparison with stresses, apparent yield strains depend weakly on density and modulus of trabecular bone in an anatomic site (Kopperdahl and Keaveny, 1998; Bayraktar, 2004). Due to the uniform characteristics within a single anatomic site (Morgan et al., 2001), strain is more proper to be used for judging trabecular bone failure (Kopperdahl and Keaveny, 1998; Stölken, 2003; Bayraktar, 2004; Morgan,

2004). As there have no published references about yield strains of the scapula and strains are anatomically site dependent (Morgan et al., 2001), strain criterion was not used to detect the failure of glenoid bone in this study. Yield stress criterion was used in current study instead.

2.2.4 Bone Remodelling Nature and Related Theories

2.2.4.1 Introduction

Living bone is a dynamic tissue. It is continually regulating its growth, maintenance, and loss in responds to external loading conditions. In the 17th century, the relationship between mechanical forces and bone morphology was inferred by Galileo (Treharne, 1981). Since then, describing the correlation between the bone adaptation and the function had been a considerable interest for many years. In 1892, Julius Wolff (Wolff, 1892), who was a German orthopaedic surgeon, first definitely expressed the adaptive characteristics of living bone and named this process as Wolff's law. In his theory, bone was hypothesized to adapt both its internal architecture and external conformation as a consequence of the alteration of strains or stresses that the bone was subjected. From the anatomic view, external strain changes may lead to variations of flow of the interstitial fluid through bone matrixes. In response to these changes of the flow, osteocyte—a type of cell embedded in the bone is stimulated and triggers actions of osteoblasts (bone forming cells) and osteoclasts (bone resorbing cells). The predominance of osteoblastic activity induces the formation of bone, thus improves the bone quality. This process of increases of the bone mass is associated with hypertrophy. While the predominance of osteoclastic activity results in the loss of bone, thus reduces the bone quality. This process is referred to atrophy. The balance of activities of these two types of bone cells (osteoblasts and osteoclasts) will not lead to any structural changes of bone. Processes of atrophy and hypertrophy, which lead to a significant alteration of bone morphology and properties, are referred to bone remodelling.

Well understanding and prediction of remodelling process in living bone is of a proper importance for the design and failure analyses of orthopaedic prostheses (Cowin and Hegedus, 1976). It has been observed that when an orthopaedic device is placed in a living bone, the normal stress patterns in the bone will change and these new local

functional requirements may induce variations of the local bone mass. Effects of bone adaptation on the fixation of an orthopaedic device are described next. When bone suffers higher postoperative stresses, it becomes stiff and strong. When bone engages in a process of postoperative degeneration, it may be weakened and endanger the long-term fixation of the orthopaedic device (Huiskes et al., 1987). Another concern related to bone resorption is that it will result in less bone to support the new implant and leads to complications in revision surgery, such as the procedures of stem extension and bone grafting. Thereby, a critical description of the postoperative bone adaptation is needed (Cowin and Hegedus, 1976; Carter, 1984; Huiskes et al., 1987).

2.2.4.2 Study of Functional Adaptation of Living Bone

Investigation of historical and physiological bone remodelling could be divided into three types: in-vitro testing (Chamay and Tschantz, 1972; Woo, 1981; Carter, 1984), in-vivo studies (Pressel et al., 2000) and computational modelling (Cowin and Hegedus, 1976; Carter, 1984; Huiskes et al., 1987; Cowin, 1993).

2.2.4.2.1 *Experimental study*

In-vitro testing is a direct way to study remodelling process of living bone. Up to date, the in-vitro testing was all animal experimental studies with the purpose of investigating stress-related bone remodelling and detecting the time scale of bone adaptation in response to the change of cyclic strain rates (Chamay and Tschantz, 1972; Woo, 1981; Carter, 1984). By resecting a small segment of radius of a group of dogs and then allowing them to walk on their weakened forepaws, Chamay and Tschantz (1972) reported that massive bone hypertrophy occurred over 9-month follow up. In another experimental study of adult canine during a short period (8 weeks), Carter et al. (1981) did not find apparent effects of cyclic overstrains on the radius adaptation. By investigating walking strains of the mature sheep with ulna removed for 50-week following up, the magnitude of strain on the cranial and caudal surfaces of the radius increased 20% and 8%, respectively (Lanyon et al., 1982). In a study of femoral cortical thickness of young pigs, Woo (1981) reported that there was a proportional relationship between the rate of bone deposition and the level of physiologic loading activities while the immobilization and the rate of bone loss may be not subject to linear correlation.

These in-vitro experiments provided valuable data-support for the simulation and prediction of mechanical adaption of bone (Cowin, 2001).

2.2.4.2.2 *In-vivo study*

Clinically, bone resorption can be observed by X-ray and usually quantitatively determined by the Dual-energy X-ray absorptiometry (DEXA) technique. The DEXA scan utilizes two X-ray beams with differing energy levels to measure the absorption by bone and thus to determine the regional bone mineral density (BMD, g/cm^2). The BMD is associated with the mechanical bone strength (Petersen et al., 1996). When the bone density decreases, the strength of bone will decrease and thus may lead to formation of weak bone zone, loss of prosthesis-bone support, component migration/loosening and bone fracture (Carlsson et al., 1983; Lonner et al., 2001; Davis et al., 2005).

2.2.4.2.3 *Theoretical and computational modelling*

The purpose of theoretical and computational modelling is to simulate the adaptive process of bone remodelling, to quantitatively describe the functional adaptation of bone, and finally to predict and analyse the fixation of orthopaedic prostheses. In most of these models, adaptive problem is assumed to comply the continuum, quasi-static and linear elastic theory. Thus effects of loading rates acting upon bone, viscoelasticity and inertial effects are not considered. Hart (2001) pointed out that linear elasticity equations are generally adequate for studying the strain-induced bone adaptation.

2.2.4.3 Computational Simulation of Bone Adaptation

In terms of Wolff's law, a general concept of bone remodelling is proposed by Hart (1984) (Figure 2-14). This feedback control process of bone adaptation is initially triggered by changes of external loading conditions. The variation of mechanical usage (strains in Figure 2-14) combined with non-mechanical factors, such as genetic, metabolic and hormonal factors, is converted into cellular responses (abnormal activities of osteoblasts and osteoclasts) by a transducer. As a result of the cellular responses, bone structural remodelling (bone formation and resorption) occurs. Consequently, the structural changes act as a feedback signal to trigger a new

circulation of bone adaptation. This process will continue until the structural changes are diminished.

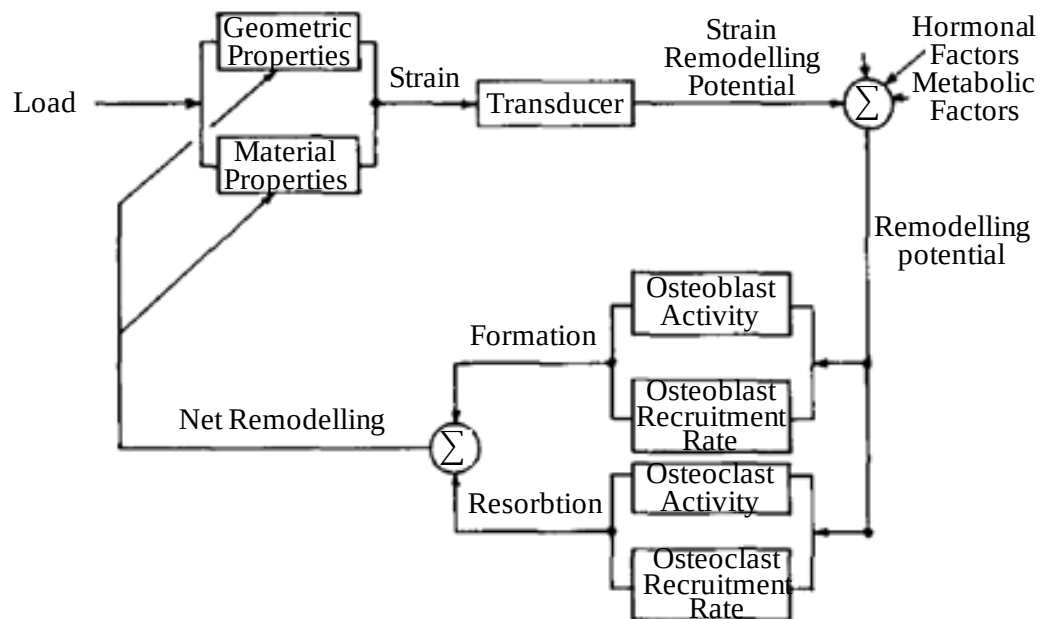


Figure 2-14 A schematic diagram of adaptive control process of bone (Hart et al., 1984).

In the computational simulation of bone remodelling, only mechanical adaptation is considered. In another word, non-mechanical factors, such as genetic, hormonal, metabolic factors, are ignored. Therefore, the computational study of bone remodelling can only simulate and predict the changes of shape and/or material properties induced by mechanical factors. In the concept of bone remodelling, it is clear that successful simulations of functional adaptation of bone is subject to effective and reasonable choices of mechanical usages and correct descriptions of the external and internal remodelling.

2.2.4.3.1 *Bone remodelling stimulus*

Bone remodelling stimulus is a measurable mechanical usage, which could activate bone responses (formation and resorption) to the loading histories. Mechanical stimuli used in most theoretical models of physiological and historical bone adaptation include strain (Cowin and Hegedus, 1976), stress (Cowin, 1993), strain energy (Fyhrie and Carter, 1986), strain rate (Lanyon, 1984), fatigue microdamage (Carter, 1984), strain

energy density (Huiskes et al., 1987), and strain energy density per apparent density (Weinans et al., 1992). Proper choices of mechanical stimuli are still arguable. Cowin (1984) highlighted that the stimulus should be strain and not stress, as strain was a direct measurable physical parameter whereas stress was computed indirectly. In a review of numerical theories and simulations subjecting to different mechanical stimuli, Hart (2001) pointed out that bone adaptation seemed not to be very picky about the use of specific mechanical stimuli.

2.2.4.3.2 *External and internal remodelling models*

Based on Wolff's law (Wolff, 1892), the remodelling process of living bone could be divided into two general classes: external (surface) remodelling, and internal remodelling (Frost, 1964). External remodelling refers to the deposition and resorption of bone tissues on the bone surface, while internal remodelling is associated to the increasing and decreasing volumetric porosity of living bone. In another word, internal remodelling relates to the rate of change of bone property or density. Bone remodelling rate for these two classes are different. In the external remodelling rate equation (Equation 2-3), the velocity (U) of bone morphology remodelling at a surface point (X) is specified as a function (f) of the change in the bone remodelling stimulus. Formation of the morphology is represented by the positive value of $U(X)$ while resorption is associated with the negative value of $U(X)$. In the internal remodelling rate equation (Equation 2-4), the time rate of change of bone density at a point (X) in the bone tissue is specified as a function (f) of density at a point (X) associating with the change of bone mechanical remodelling stimulus at that point. Given the inhomogeneous characteristics of bone, the equation is expressed at a point-by-point basis. These remodelling rate equations are subject to linear elasticity theory and a continuum mechanical basis (Cowin, 1993).

$$U(X) = f(\text{change in stimulus at the surface point } X) \quad \text{Equation 2-3}$$

$$\frac{\partial \rho}{\partial t}(X) = f(\rho(X) \quad \text{change in stimulus at point } X) \quad \text{Equation 2-4}$$

The internal and external remodelling rate equations would be diverse due to applications of different mechanical stimuli, anatomic sites and time. In the surface remodelling rate equation which employed strain as the mechanical stimulus, change of stimulus was out of proportion to the functional adaptation of bone (Cowin and Van Buskirk, 1979). While if using the strain energy density as the mechanical stimulus in the remodelling rate equation, the rate of change of bone structural features was linear to the rate of change of bone remodelling stimulus (Huiskes et al., 1987).

2.2.4.4 Bone Remodelling Finite Element algorithms

Numerous mathematical models have been proposed to explain and mimic the process of bone remodelling. A detailed introduction regarding to the theoretical models of functional bone adaptation can be found in the textbook edited by Cowin (2001). In 1982, Carter developed and published a conceptual framework to describe the adaptive remodelling response of cortical bone. In his framework (Figure 2-15), structural adaptation ($\Delta B/\Delta t$) was a function to the alteration of cyclic strains ($\Delta S/\Delta t$), which was associated with the cyclic strain range ($\Delta \epsilon$), the average of strains (ϵ_m) and the number of loading cycles (N). Bone was gained in severe repeated loading conditions, lost in the immobilization case, and relatively unresponsive to the changes of the cyclic strains in the normal physical activities.

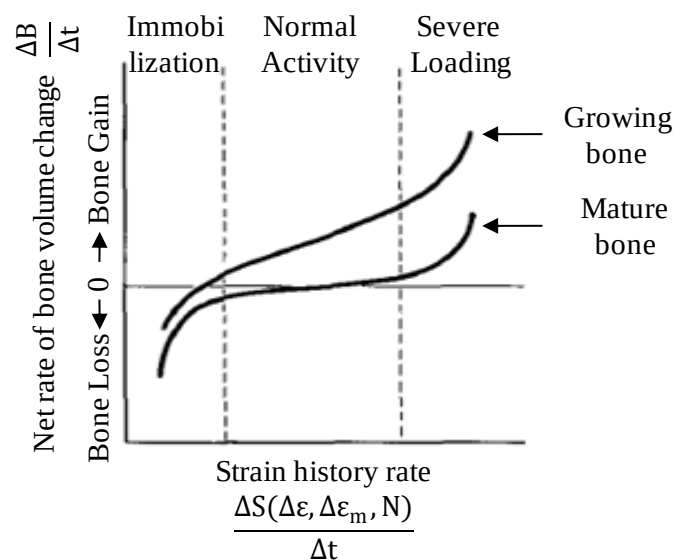


Figure 2-15 Diagram showing the relationship between bone strain history rate and structural adaptation for mature and growing animals (Carter, 1982).

Similar to the Cater's study (1982), Huiskes et al. (1987) used strain energy density (SED) (Equation 2-5) as the feedback control mechanical stimulus. The error between the actual SED, $U(t, x)$, and the site-dependent homeostatic equilibrium SED, U_n , was the driving force for bone adaptive activities. The relationship between the bone adaptation rate and the mechanical stimulus could be described with a trilinear diagram shown in Figure 2-16. Bone behaviour was inactive to the mechanical stimulus in the lazy zone $((1 - s)U_n \leq U \leq (1 + s)U_n)$, proportional increase in the hypertrophy region $(U > (1 + s)U_n)$, and linear decrease in the trophy region $(U < (1 - s)U_n)$. Formula expressions for the cortical and trabecular bone remodelling are shown in Equation 2-6 and Equation 2-7. In these two equations, X is the surface growth, E is the material property, C_x and C_e are the remodelling rate coefficients for cortical bone and trabecular bone respectively, $2s$ is the width of the lazy zone, U and U_n are the strain energy densities subjected to altered loading conditions and the homeostatic equilibrium strain energy density, respectively.

$$U = \frac{1}{2} \sigma_{ij} \varepsilon_{ij} \quad \text{Equation 2-5}$$

where U is the SED, σ_{ij} is the local stress tensor, and ε_{ij} is the local strain tensor.

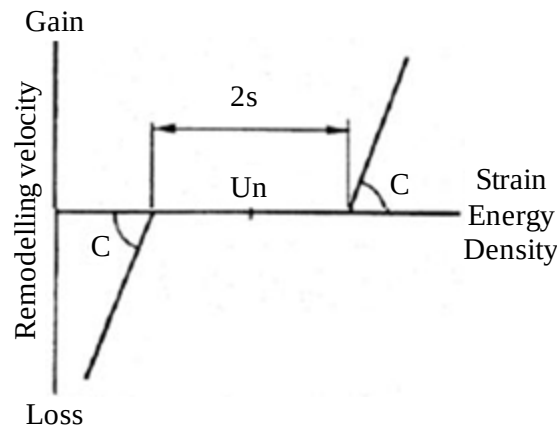


Figure 2-16 Schematic relationship between adaptive remodelling rate and changes of strain energy density (Huiskes et al., 1987).

$$\frac{dX}{dt} = \begin{cases} c_x[U - (1+s)U_n]; & U > (1+s)U_n \\ 0; & U_n < U < (1+s)U_n \\ c_x[U - (1-s)U_n]; & U < (1-s)U_n \end{cases} \quad \text{Equation 2-6}$$

$$\frac{dE}{dt} = \begin{cases} c_e[U - (1+s)U_n]; & U > (1+s)U_n \\ 0; & (1-s)U_n \leq U \leq (1+s)U_n \\ c_e[U - (1-s)U_n]; & U < (1-s)U_n \end{cases} \quad \text{Equation 2-7}$$

In combination with the self-optimization theory, in which apparent density is a function of strain energy density ($\rho = c'U$), Weinans et al. proposed a new feedback mechanical stimulus: strain energy per unit of bone mass (U/ρ). Thus the mathematical expression was changed to the relation between the rate of density change and the stimulus (Equation 2-8),

$$\frac{d\rho}{dt} = \begin{cases} A[S - S_n(1+s)]^2; & S > (1+s)S_n \\ 0; & (1-s)S_n \leq S \leq (1+s)S_n \\ A[S - S_n(1-s)]^3; & S < (1-s)S_n \end{cases} \quad \text{Equation 2-8}$$

$0 \text{ g/cm}^3 < \rho < 1.74 \text{ g/cm}^3$

where ρ is apparent density, which only varies between 0 and the value for cortical bone, A is a time constant and S_n is the reference stimulus. Exponent of the equation varies from 2 for bone apposition to 3 for bone resorption, indicating a more progressive activity of bone atrophy than hypertrophy. In the case of a few load cases, the stimulus S was the average of the stimuli in all the loading conditions (Equation 2-9). Figure 2-17 shows the relationship between the rate of change of apparent density and the mechanical remodelling stimulus.

$$S = \frac{1}{n} \sum_{i=1}^n \left(\frac{U_i}{\rho} \right) \quad \text{Equation 2-9}$$

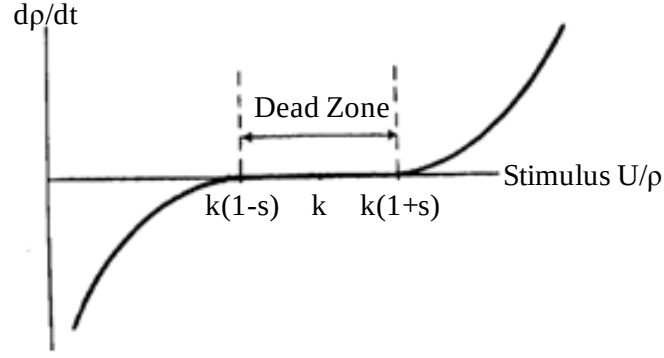


Figure 2-17 Schematic graph showing the relationship between the rate of change of apparent density and the mechanical remodelling stimulus (Weinans et al., 1992).

Implementation of the FE algorithm is elaborated in Equation 2-10.

$$\begin{aligned}\Delta \rho^i &= \tau \Delta t A(\rho^i) \{S^i - S_n^i(1+s)\} \Delta t, & S^i &\geq S_n^i(1+s) \\ &= \tau \Delta t A(\rho^i) \{S^i - S_n^i(1-s)\} \Delta t, & S^i &\leq S_n^i(1-s) \\ & & (0 \text{ g/cm}^3 < \rho < 1.8 \text{ g/cm}^3)\end{aligned} \quad \text{Equation 2-10}$$

where the rate of change of bone densities for each element i is associated with the time scale, τ , the free surface density at that element, $A(\rho)$, and the time step, Δt . τ is a relation between the simulated and the real time. $A(\rho)$ is a function of apparent density (Martin, 1984). Δt (expressed in Equation 2-11) is determined from a pre-defined maximum change of apparent density. The change of apparent density at each element of the bone model is solved by means of the forward-Euler integration method. At each iteration, the density change at each element will lead to the adaptation of elastic modulus at that position in terms of the pre-defined modulus-density relationship $E = 3790\rho^3$ (Carter and Hayes, 1977). New elastic modulus would be updated at the start of next iteration. The Poisson's ratio is assumed to remain unchanged throughout the whole adaptive bone remodelling process.

$$\tau \Delta t = \frac{\Delta \rho_{max}}{\{A(\rho)(S - (1+s)S_n)\}_{max}} \quad \text{Equation 2-11}$$

The depth of “dead zone”, $2s$, is proposed to vary between 0.2 and 1.6 for the simulation and analyses of bone adaptation (Kerner et al., 1999) (Huiskes 1992, van

lenthe 1997, barink 2003, Gulshan 2009). By validating the computational adaptive bone remodelling with clinical studies of femoral knee, $s = 0.75$ was reported to be able to generate realistic results (van Lenthe 1997). In a study of adaptive glenoid bone remodelling simulation, s was suggested to be 0.4 (Sharma et al., 2009).

CHAPTER 3

REVIEW OF PATHOLOGY AND SHOULDER REPLACEMENT

Artificial shoulder joint arthroplasty is used to relieve pain and to restore physiological natural shoulder functions. This chapter is aimed to introduce shoulder arthroplasty from a few aspects: (1) types of shoulder arthroplasties; (2) clinical indications for shoulder replacements; (3) a historical review of shoulder replacement designs, especially reverse shoulder arthroplasty; (4) some prosthetic knowledges, such as cementless fixation, stress shielding and glenohumeral joint force for reverse shoulder arthroplasty.

3.1 Types of Shoulder Replacements

There are a few ways to classify glenohumeral arthroplasties:

(1) Anatomical (Figure 3-1(a)) and reverse (Figure 3-1(b)) designs in terms of design configurations of shoulder prostheses. Anatomical shoulder device has the similar anatomical constructure to the natural shoulder joint. While, in reverse design, a modified ball replaces the scapular fossa, and a socket replaces the humeral head. In comparison with the anatomical shoulder arthroplasty, the reverse shoulder device is mainly used in the treatments of severe rotator cuff deficiency associated with arthritis (more comparisons between anatomical and reverse devices in treatments of rotator cuff arthropathy will be presented in Chapter 3.2), and revisions of total or hemi shoulder arthroplasties.

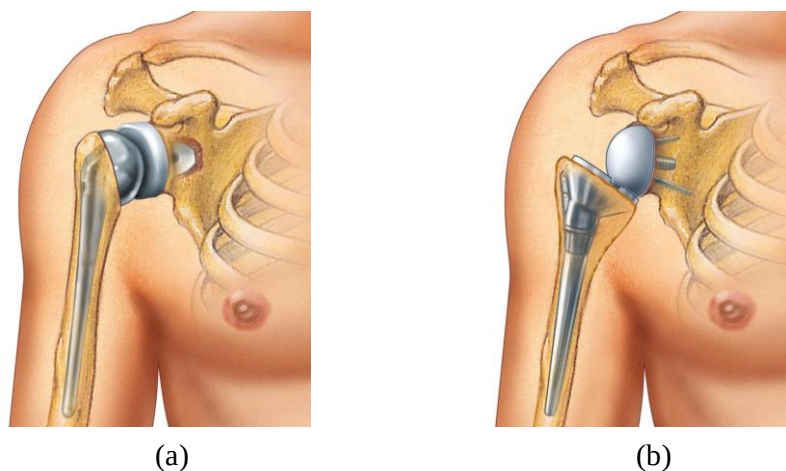


Figure 3-1 Schematic graph showing (a) anatomical total shoulder arthroplasty and (b) reverse shoulder arthroplasty (adapted from Roberts et al., 2007).

(2) Prosthetic humeral hemiarthroplasty (HP) (Figure 3-2 (a)) and total shoulder arthroplasty (TSA) (Figure 3-1 (a)). HP is used to replace only the proximal humerus, while TSA is used to reconstruct both the glenoid and the humerus. Thus HP surgical technique can preserve the entire glenoid and avoid some complications relating to the glenoid component in the TSA, such as glenoid loosening and polyethylene wear. In addition, the revision of a HP usually requires less complex surgery than that of a TSA. Hemiarthroplasty is usually considered for patients with intact cartilaginous surface of the glenoid. In the case of intrinsic cartilage abnormalities or severe glenoid erosion, HP does not provide comparable levels of pain relief to the TSA (Clayton et al., 1982; Bell and Gschwend, 1986; Boyd et al., 1990).

(3) Resurfacing (Figure 3-2 (b)) and stemmed shoulder replacements (Figure 3-2 (a)). Prosthesis resurfacing is used to replace the damaged surface of humeral head. This technique could minimize bone resection, thus preserve the native head-shaft angle, offset and inclination of the bone. In addition, resurfacing technique may also facilitate later conversion to a conventional TSA. Thus resurfacing prosthesis has been an attractive option for young, active or athletic patients (Levy and Copeland, 2004; Bailie et al., 2008b). Due to the metallic humeral resurfacing implant, the late glenoid arthrosis may require revision to a TSA (Bailie et al., 2008a)

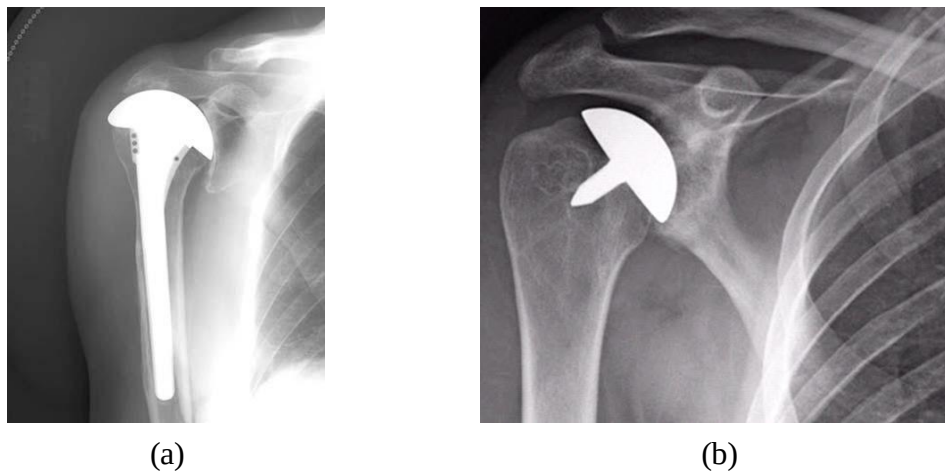


Figure 3-2 Radiographic hemiarthroplasties: (a) stemmed hemiarthroplasty (adapted from Wiater and Fabing, 2009); (b) resurfacing shoulder arthroplasty (adapted from Wiater and Fabing, 2009).

3.2 Clinical Indications for Shoulder Arthroplasty

Some indications of shoulder arthroplasty includes primary osteoarthritis (OA), cuff tear arthroplasty (CTA), rheumatoid arthritis, severe proximal humerus fractures, and failed shoulder arthroplasty (Wiater and Fabing, 2009).

3.2.1 Osteoarthritis

Osteoarthritis (Figure 3-3) is a disorder that results in progressive degradation of biomechanical properties of articular cartilages. In the early stages of osteoarthritis, the joint cartilage loses its smooth and glistening appearance. The surface becomes rough

and irregular. The cartilage wear continues until the two parts of the joint directly contact. Without the lubrication of articular cartilage, osteoarthritis is usually accompanied with thickening of the subchondral bone, formation of osteophytes, pain, swelling, and motion restriction (Susan, 2007). It is estimated that approximately 21 million Americans have symptoms relating to osteoarthritis. Most of them are older than 65 years of age (Muscle MagFitness. com). Nonsurgical options for symptomatic osteoarthritis include physical therapy, medications, and local injections (Kelley, 2006). For the moderate and severe osteoarthritis in conjunction with an intact rotator cuff, usual treatments are surgeries of HP or TSA (Edwards et al., 2003; Lo et al., 2005; Radnay et al., 2007).

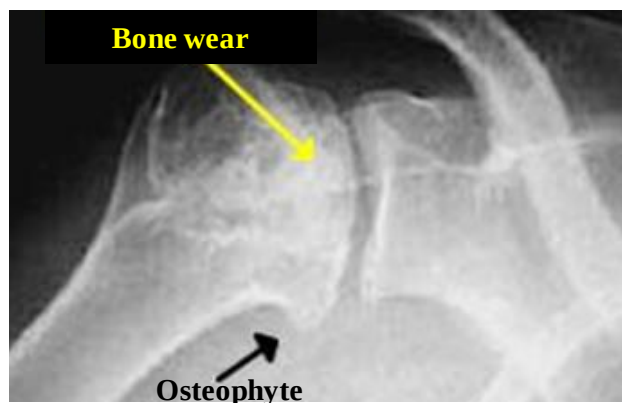


Figure 3-3 Radiograph showing osteoarthritis (adapted from shoulder surgeon .com).

3.2.2 Cuff Tear Arthropathy

Cuff tear arthropathy (CTA) (Figure 3-4) refers to the rotator cuff tear associated with shoulder arthritis (Neer et al., 1983). With severe CTA, the patients lose not only the function of the cartilage that normally covers the shoulder joint surface, but also the function of tendons of the rotator cuff. This shoulder disease is usually accompanied with the leakage of synovial fluid, atrophy of the glenohumeral articular cartilage and osteoporosis of the subchondral bone of the humeral head (Neer et al., 1983). As introduced in Chapter 2.1.5, rotator cuff (RC) is a significant dynamic stabilizer during the shoulder motion. Thus the loss of RC function may cause the superior dislocation of humeral head, the subacromial impingement, and consequently result in erosions of the

bone, noticeably the anterior portion of acromion, the acromioclavicular joint, and the superior glenoid. CTA may gradually develop and induce pain, pseudoparesis and loss of shoulder function. CTA is a common shoulder disease. In 2008, almost 2 million people in the United States went to see doctors due to the rotator cuff problem (American Academy of Orthopaedic Surgeons). In addition, CTA is also highly observed in the postoperative life. In terms of the PearlDiver database, approximately 47% of patients who were treated with a total shoulder replacement had either partial or full rotator cuff tears during the period from 2004 to 2007 (Joshi, 2009).

The normal surgical treatments for the CTA are hemiarthroplasty (HP) and reverse shoulder arthroplasty (RSA). If the CTA is mild, and the shoulder is stable, HP is used to smooth the surface of proximal humeral head and then allow the arthritic shoulders to regain some of the lost comfort and functions (Williams and Rockwood, 1996; Sanchez-Sotelo et al., 2001). However, if the shoulder is no longer stabilized with the humeral head locating in the glenoid socket, RSA is usually considered for the treatments. That is because the results of anatomical shoulder arthroplasty are less satisfactory (Werner et al., 2005; Wiater and Fabing, 2009). For instance, the postoperative active forward elevation is still smaller than 90°. In the absence of stabilization provided by the rotator cuff muscles, the floated centre of rotation of the arm in the anatomical shoulder will be moved superiorly (black arrow) under the deltoid muscle force (purple arrow), leading to indications of CTA (Figure 3-5 (a)). In this situation, the superior deltoid muscle force in the reverse configuration can be effectively used to rotate the arm around the fixed fulcrum (Figure 3-5 (b)). However, this force in the TSA cannot be stabilized and may lead to superior translation of the humeral head. In addition, this translation may result in eccentric loading on the glenoid component, thus induce the “rocking-house” phenomenon, and consequently cause glenoid prosthesis loosening (Figure 3-5 (c)). The RSA is currently suggested only for old people (age ≥ 65) with pseudoparalysis and pain (Wiater and Fabing, 2009), due to the high implant failure rate in long-term period (Sirveaux et al., 2004; Boileau et al., 2006; Guery et al., 2006).

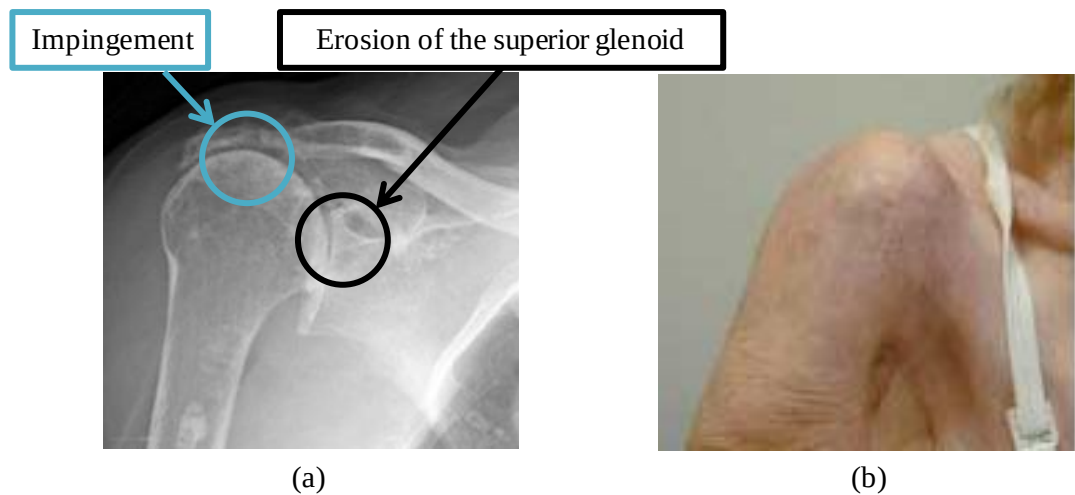


Figure 3-4 Rotator cuff arthropathy: (a) radiograph and (b) photograph of a patient with CTA.

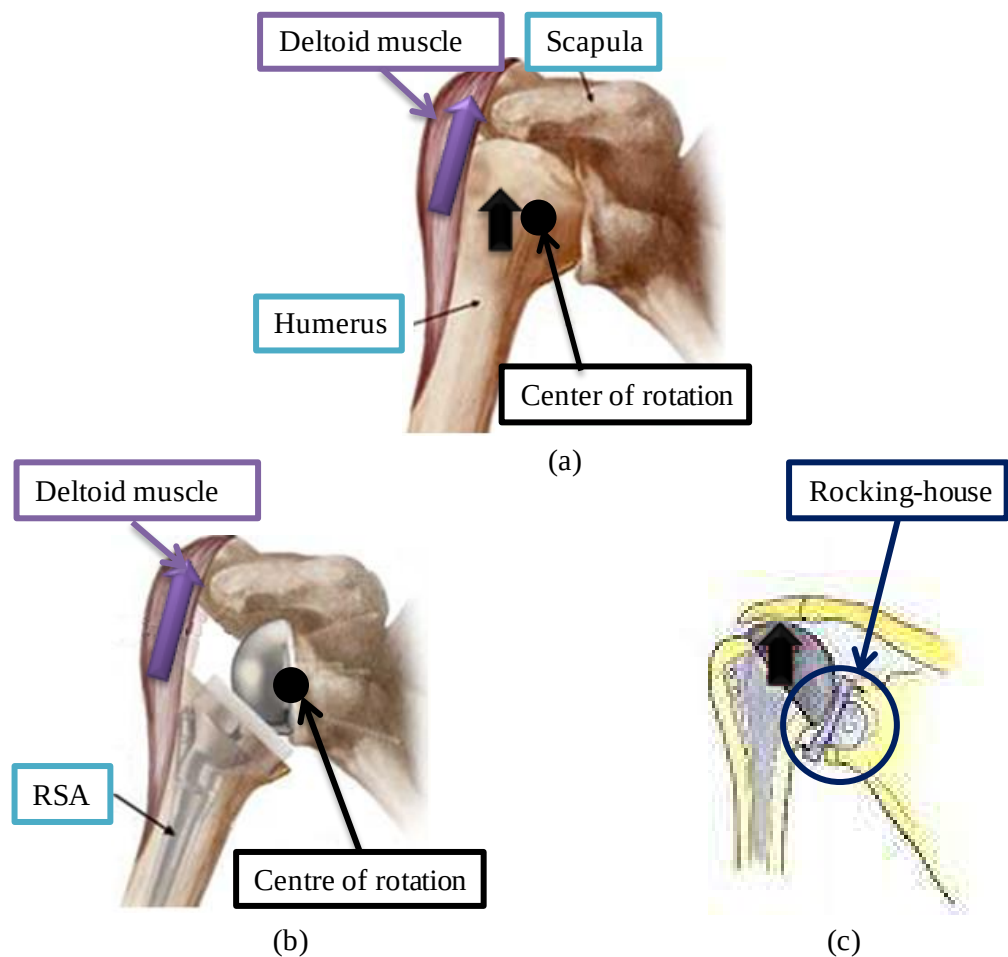


Figure 3-5 Schematic graph showing the CTA in (a) the anatomical shoulder; (b) the reverse shoulder arthroplasty and (c) the total shoulder arthroplasty.

3.2.3 Rheumatoid Arthritis

Rheumatoid arthritis is an autoimmune disorder. It is usually accompanied with inflammation and thickening of the synovial membranes and breakdown of the articular cartilage. With Rheumatoid arthritis, the patients usually have limited motion and ossification or fusing of the articulating bones. Surgical treatments of rheumatoid arthritis are rotator cuff arthropathy dependent. In the case of irreparable rotator cuff, reverse shoulder arthroplasty is usually suggested (Wiater and Fabing, 2009). In the case of intact rotator cuff, resurfacing hemiarthroplasty or total shoulder arthroplasty is considered, depending on whether the glenoid is worn (Hedtmann and Werner, 2007).

3.2.4 Acute Proximal Humerus Fractures

Proximal humerus fractures are identified by four types (Boileau et al., 2001) (Figure 3-6). Type 1 is the sequelae of impacted fractures with humeral head collapse or necrosis; Type 2 is the irreducible dislocations or fracture-dislocations; Type 3 is non-unions of the surgical neck; Type 4 is severe tuberosity malunions. Anatomical configuration prostheses are recommended for treatments of Types 1 and 2. Surgical treatments of Types 3 and 4 are cuff tear arthropathy dependent. In the case of intact rotator cuff, hemiarthroplasty with tuberosity reconstruction is recommended. For the situation of old patients with CTA (age ≥ 65), RSA is used (Wiater and Fabing, 2009).

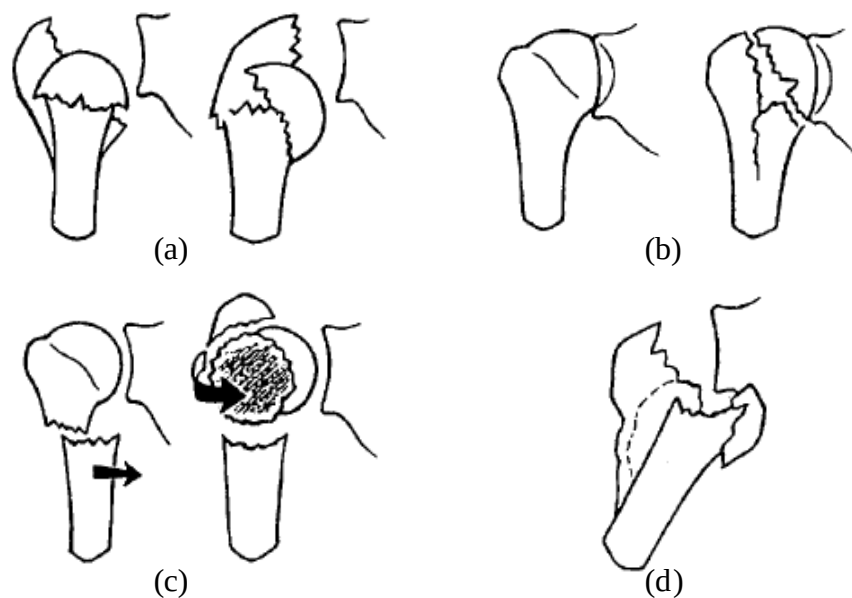


Figure 3-6 Proximal humeral fractures (adapted from Boileau et al., 2001).

3.2.5 Revision

Revision of shoulder arthroplasty is mainly subject to a few reasons: soft-tissue deficiency, osseous concerns (i.e. glenoid arthrosis, bone loss), and implant problems, such as malposition, improper sizing, wearing and loosening (Wiater and Fabing, 2009). Reverse shoulder arthroplasty has been a valuable option for the patients who have failed shoulder arthroplasty. In a study of 29 revisions of hemiarthroplasty, the use of RSA increased the ASES score of 30, and improved the forward elevation from 38.1° to 72.7° (Levy et al., 2007).

3.3 History of Shoulder Arthroplasty

The history of shoulder prosthesis could date from 1893, when a French surgeon Jules E. Pèan successfully performed the shoulder arthroplasty (Figure 3-7) which was designed by a Parisian dentist, J. Porter Michaels. This shoulder arthroplasty was made of platinum and leather, and used to treat tuberculous arthritis of a 37-year-old baker. The postoperative arm motion and strength was significantly improved, but the postoperative infection led to the prosthesis failure two years later. Since then, there was about half a century without any references to shoulder prostheses in the literature. In 1955, Neer pioneered shoulder replacement surgery. He successfully treated the humeral fracture dislocation with a vitallium humeral prosthesis (Neer I).

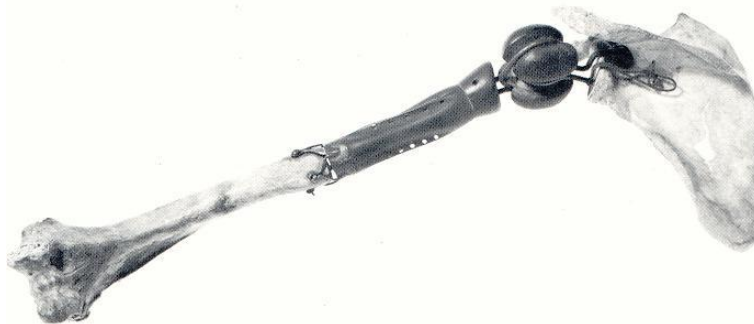


Figure 3-7 The first successfully performed shoulder arthroplasty (adapted from Lugli, 1978).

In 1974, a total non-constrained design, Neer II, was invented. In this implant, the glenoid was resurfaced with a cemented high density polyethylene (HDPE). However, TSA was hampered by the stability problem, particularly in the case of rotator cuff damage. To address this problem, a reverse ball-and-socket joint (Neer Mark III) (Figure 3-8) with a unique dual-compartment humeral component that allowed axial rotation of a metal stem within a polyethylene sleeve was tested. This implant failed due to the glenoid loosening. Since then, shoulder arthroplasties has developed by two lines: anatomical shoulder design and reverse shoulder configuration.

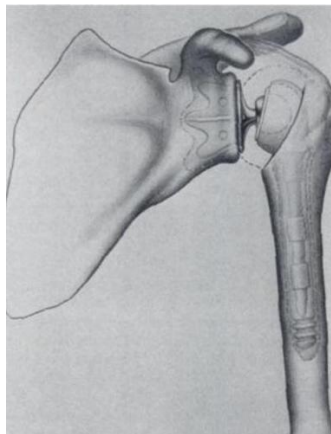
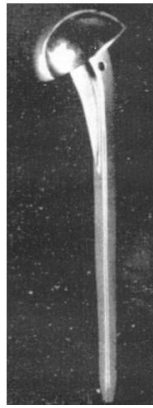


Figure 3-8 Neer Mark III fixed fulcrum prosthesis (adapted from Neer, 1990).

3.3.1 Improvements of the Anatomical Shoulder Design

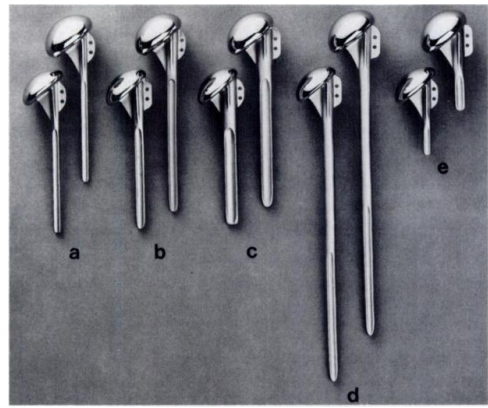
From the first generation of Neer II implant, the second generation introduced modular implants (Figure 3-9), which were to accommodate variations in humeral anatomy and space available for the joint. In the third generation, offset humeral heads in the medial and posterior directions was designed in order to produce a more anatomically aligned humeral component. The non-constrained glenoid design, which was first introduced by Neer et al. (1974), is the most commonly used in the TSA surgeries.



Original Neer design



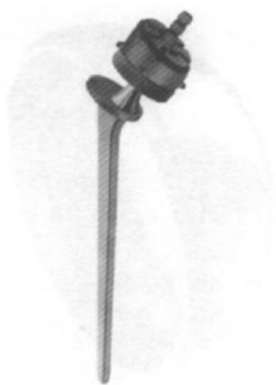
Neer II



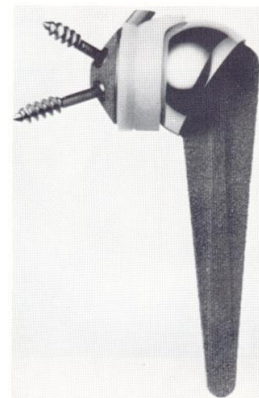
Modular humeral component designs

Figure 3-9 Different types of Neer shoulder arthroplasties (adapted from Neer, et al., 1953, Neer, 1974, Neer, et. al, 1982).

Opposite to the unconstrained shoulder design, constrained and semi-constrained implants (Figure 3-10) were proposed in attempt to improve the implant fixation and to compensate rotator cuff functions. One of the major problems of these implants is the limited shoulder motion due to the frequent impingements between the implanted components and the bones, especially the scapula.



Constrained design



Semi-Constrained design

Figure 3-10 Constrained and semi-constrained TSA.

Other pioneered anatomical shoulder arthroplasties include bipolar prostheses and resurfacing devices (Figure 3-11). The bipolar shoulder prosthesis was first designed by Swanson (1986). This large-head humeral implant yielded good results for pain relief but was inadequate for motion and durability (Swanson et al., 1989). The resurfacing device was used to replace only the articular surface of the humeral head. This

conservative design was first developed by Bateman (Bateman, 1978). Along the same line of Bateman's procedure was the cup arthroplasty (Figure 3-11) described by Jonsson (Jonsson et al., 1986). The most frequently reported problems about this type of prosthesis are cup loosening and central glenoid wear (Jonsson et al., 1988).

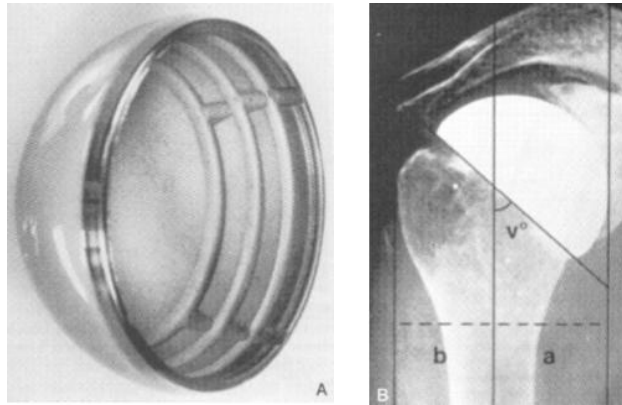


Figure 3-11 Cup arthroplasty introduced by Jonsson (adapted from Jonsson et al., 1986).

3.3.1.1 Indications for Total Shoulder Arthroplasty

Primary indications for TSA are moderate and severe osteoarthritis in conjunction with loss of articular cartilage, incongruent osseous surface, and rotator cuff tear (Wiater and Fabing, 2009). Other indications include inflammatory arthritis, advanced osteonecrosis with glenoid involvement, and posttraumatic degenerative joint disease with proximal humerus malunion (Wiater and Fabing, 2009).

3.3.1.2 Efficacy of Conventional Total Shoulder Arthroplasty

Since the pioneered Neer I, efficacy of conventional total shoulder arthroplasty has been presented by several authors (Neer, 1982; Cofield, 1984; Torchia et al., 1997). In a study of 194 shoulders treated with TSA at a minimum 2-year follow up, Neer (1982) reported that 98% of patients had benefited from the procedure. Cofield (1984) reviewed 73 TSAs at 2- to 6.5-year follow-up, and showed that the active abduction improved from 76° preoperatively to 122° postoperatively. In a long-term review of TSAs, the range of motion improved an average of 40°, and 75% of patients had mild or

no pain postoperatively (Torchia et al., 1997). In addition, this study showed that the probability of prosthesis survival was 93% at 10 year follow-up and 87% after 15 years.

3.3.2 Reverse Configuration and Improvements

Since the Mark III reverse ball-and-socket joint developed by Neer in 1974, several reverse designs have been introduced. The entire history of reverse shoulder arthroplasty includes two periods: Pre- and Post- Grammont designs.

3.3.2.1 Pre-Grammont Designs:

The Pre-Grammont designs refer to the reverse ball-and-socket implants introduced in 1970s and 1980s. Some of them are Kolbel prosthesis (Kölbel and Friedebold, 1973), Kessel prosthesis (Kessel and Bayley, 1982), the shoulder prosthesis introduced by Fenlin (Fenlin, 1975) and Liverpool total arthroplasty (Beddow and Elloy, 1982).

The same characteristics of these designs are that the humeral cup covers more than two thirds of the glenoid head. Kolbel prosthesis, Liverpool total arthroplasty, and Kessel prosthesis all have a small glenoid head. Differences between them are the implant fixation. In the Kolbel and Liverpool prosthetic designs, no stem components are included on the humerus side, while there is an extensive stem in the humerus component in the Kessel design. On the glenoid side, the Kolbel prosthesis is fixed with a central screw and a flange bolted to the base of the spine of the scapula. The Liverpool total arthroplasty is secured with a stem, which is inserted into the medullary cavity of the axillary border of the scapula at a depth of approximately 50 mm. In the Kessel design, the glenoid component is fixed by a large central screw. It is noticeable that Baylay-Walker prosthesis, an up to date product of Stanmore ImplantsTM, is a modification of Kessel design. In the Baylay-Walker prosthesis, the surface of the central screw is coated with hydroxyapatite in an attempt to promote the bone ingrowth and thus to reduce the rate of glenoid loosening. Unlike the reverse shoulder designs mentioned above, Fenlin (1975) developed a large glenosphere in an attempt to increase the lever arm of the deltoid muscles and increase the freedom of motion. In this design, the glenoid component is anchored by a peg through the scapular pillar. Most of the reverse shoulder prostheses introduced in 1970s and 1980s are not widely used. One of

the most common problems was glenoid prosthesis loosening (Broström et al., 1992; Friedman, 1994; Katz et al., 2007).

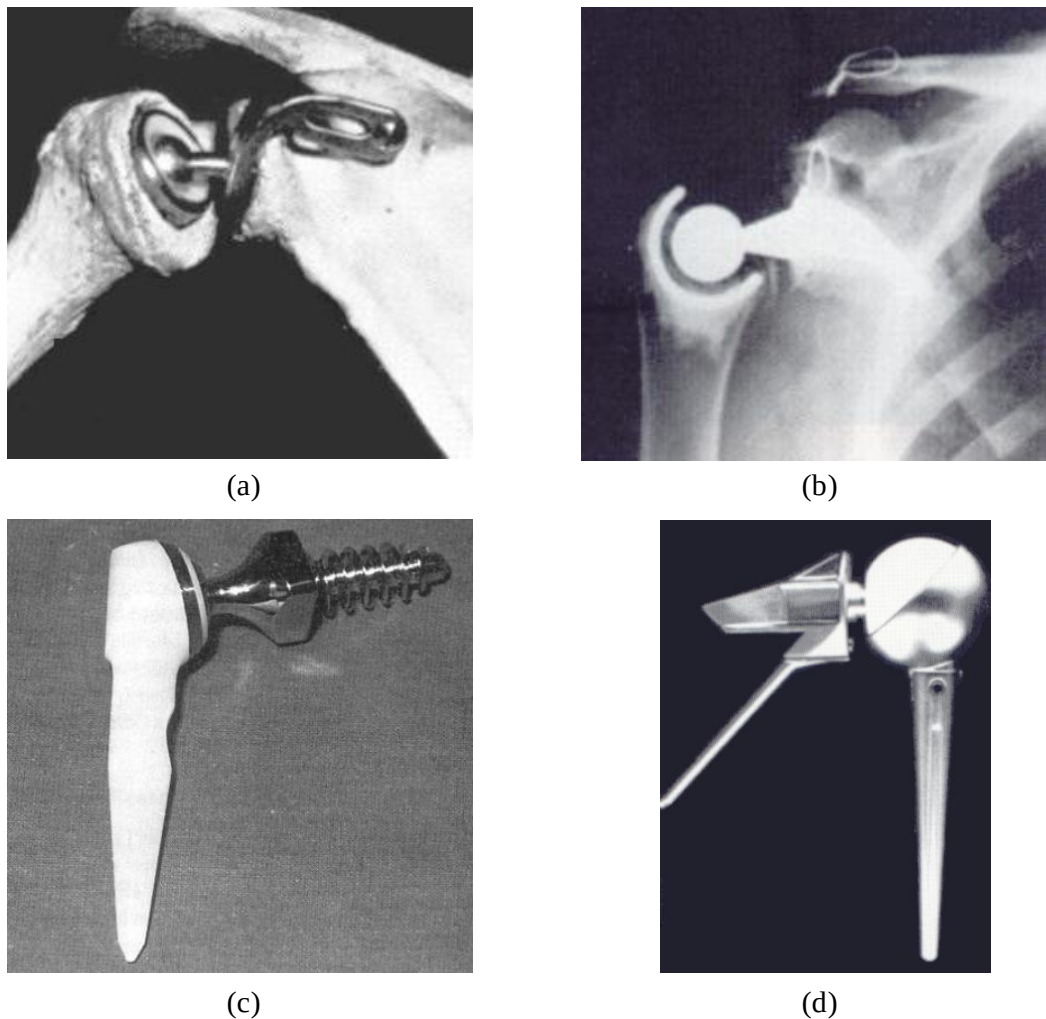


Figure 3-12 (a) The Kolbel design (adapted from Kölbel and Friedebold, 1973) (b) the Liverpool design (adapted from Beddow and Elloy, 1982) (c) The Kessel total shoulder (adapted from Kessel and Bayley, 1982) and (d) The prosthesis introduced by Fenlin (adapted from Fenlin, 1975).

3.3.2.2 Grammont Design

In 1985, the reverse shoulder arthroplasty (RSA) designed by Paul Grammont showed successful clinical results, and led to a new era of reverse shoulder designs. The first model of Grammont RSA (Figure 3-13) is composed of two elements: glenoid and humeral components. The glenoid prosthesis is a two-thirds metallic or ceramic ball. The humeral component is made of polyethylene and corresponds to one-third of the

sphere. Both the glenoid and humerus components are fixed with cement. The published preliminary clinical results of this implant (Grammont et al., 1987) showed that all the shoulders were pain-free, but the mobility was variable.

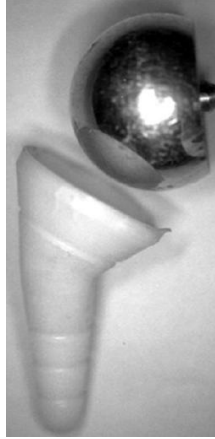


Figure 3-13 The first model of reverse shoulder prosthesis designed by Grammont (adapted from Boileau et al., 2005).

The second model designed by Grammont was named Delta III reverse shoulder arthroplasty (RSA) (Depuy International Ltd). The sound follows the pronunciation of Deltoid, which is an important motor for Delta RSA. This shoulder implant is composed of 5 parts: metaglene, glenosphere, polyethylene humeral cup, epiphysis, and diaphysis (Figure 3-14). In comparison with the design of the first model, changes in the Delta III RSA are described next.

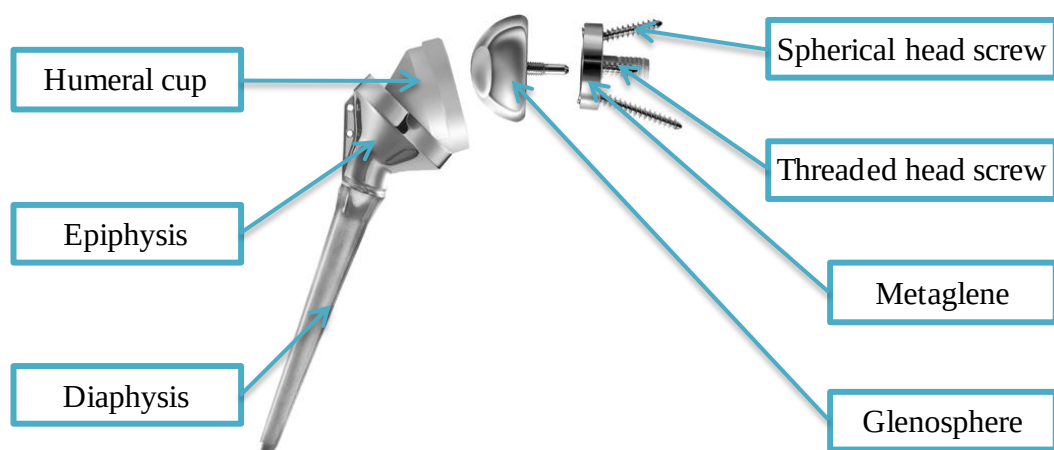


Figure 3-14 Delta III reverse shoulder arthroplasty.

One of the modifications in the Delta III design is the fixed and medialized centre of rotation. This change allows the glenohumeral joint forces to pass through the glenoid prosthesis attachment site, such that the bending at the prosthesis-bone interface is reduced and the implant stability is improved (Figure 3-15 (a)). In comparison with the normal shoulder, the medialization of the center of rotation increases the lever arm ($L_2 > L_1$) of the deltoid muscle, and thus reduces the requirements of deltoid muscle forces (Figure 3-15 (b)). In addition, the medialized centre of rotation enables the implant to recruit more anterior and posterior deltoid muscle fibers during the process of abduction (Figure 3-15 (c)). As shown in Figure 3-15 (c), in comparison with the normal shoulder, in which only muscle fibers in region II and III can be recruited for the abduction, Delta III RSA can recruit more muscle fibers in region I and IV.

Enlarged glenosphere is another modification in the design of Delta III RSA. There are two operations for the sizes of glenoid component: 36 mm and 42 mm diameters. This modification not only increases the adduction range before the prosthesis-bone impingements, but also increases the tension of deltoid muscle by lowering the position of the humerus.

The Delta III reverse shoulder arthroplasty employs a small humeral cup, which is less than half of the glenosphere. This modification increases the potential motion arc, but may compromise the implant stability. The humeral cup inclines 155° relative to the diaphysis in an attempt to reduce scapular notching.

In comparison with the first model of Grammont RSA, cementless method is used for the fixation of glenoid component in Delta III RSA. The back of metaglene is coated with hydroxyapatite to promote the bone ingrowth and thus to improve prosthesis initial stabilities. The humerus component is fixed with either cemented or cementless method.

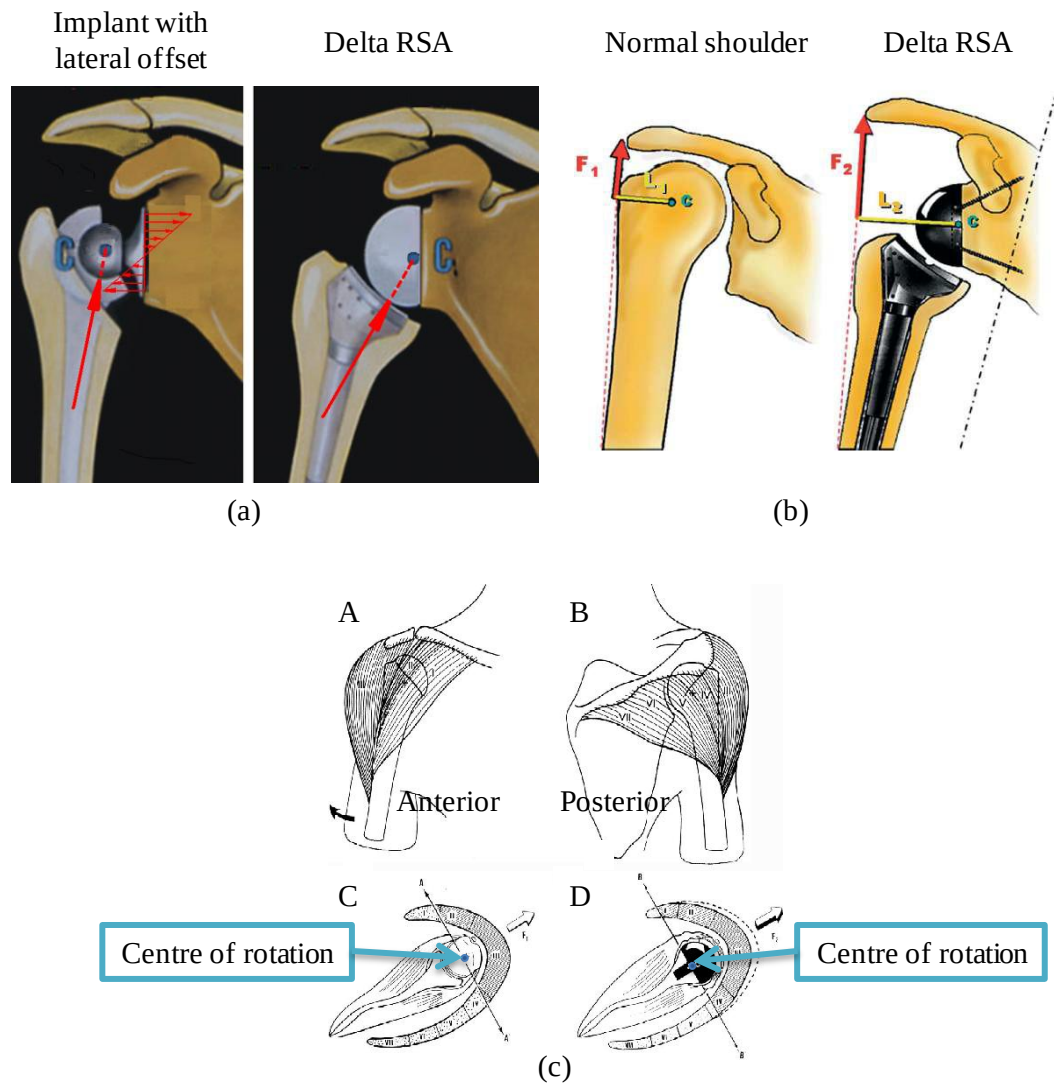


Figure 3-15 Advantages of medialized centre of rotation: (a) prosthesis-bone interface bending between other implant and Delta RSA; (b) moment arms of the deltoid muscle between normal shoulder and Delta RSA; (c) muscle fibers between normal shoulder and Delta RSA; A, B, C shows the deltoid muscles in the normal shoulder from the anterior and posterior views and on the transverse plane, respectively; D shows the deltoid muscle in the Delta RSA on the transverse plane.

3.3.2.3 Current Grammont Designs in the Market

Except for Delta system from DePuy, other leading prostheses following Grammont design include Aequalis[®] Reversed Shoulder Prosthesis from Tornier, Anatomical[™] Shoulder Inverse/reverse System from Zimmer[®] and Equinox[®] Reverse Shoulder from Exactech. A list of current reverse shoulder arthroplasties are shown in Appendix B. Most of these designs simply replicate the basic geometrical characteristics of Delta

RSA, but introduce some alterations in an effort to minimize the complications of Delta prostheses, or to improve the fixation of implant, or to simplify the revision.

Scapular notching, which is initially caused by mechanical impingements of the humeral cup against the glenoid, is one of the most frequently reported postoperative complications for the Delta RSA (more descriptions of Scapular notching is presented in Chapter 3.3.2.5.1). To address this problem, an offset (δ) of the centre of rotation lateral to the fixation fulcrum is introduced in Encore[®] reverse shoulder prosthesis (DJO Surgical) (Figure 3-16 (a)). This modification may compromise the fixation of glenoid prosthesis. Additionally, a recent study of the Encore[®] reverse shoulder prosthesis with a centre of rotation 2 mm lateral to the glenoid showed that inferior scapular notching still occurred in this type of prostheses, but may develop later than the devices with the center of rotation at the glenoid (Levy and Blum, 2009). ARROW reverse shoulder prosthesis (FH orthopaedics) (Figure 3-16 (b)), is another product in effort to minimize the scapular notching. In this design, a notching cup is introduced in the polyethylene humeral component.

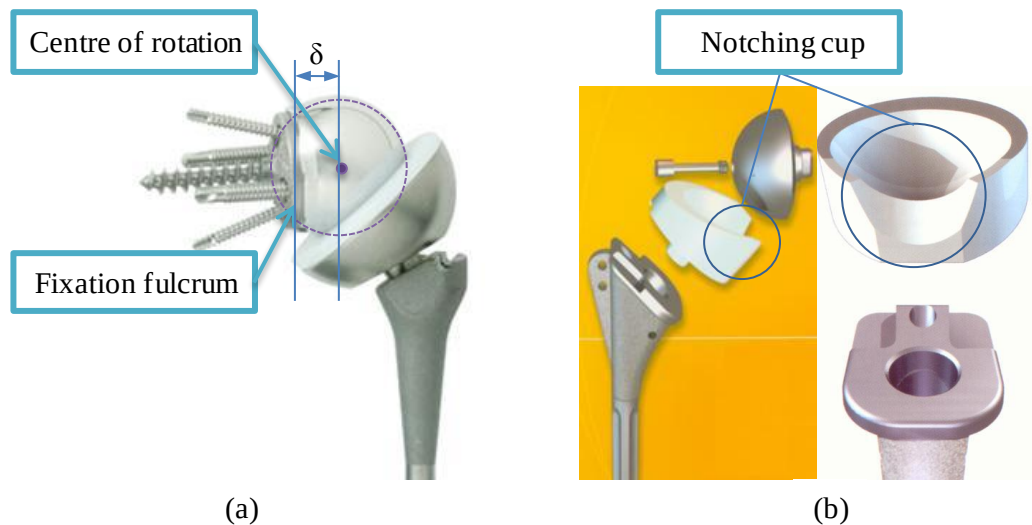


Figure 3-16 (a) Encore[®] reverse shoulder prosthesis (DJO Surgical); (b) ARROW reverse shoulder prosthesis (FH Orthopaedics).

In the Delta design, the superior and inferior screws are in a fixed angle to the back of metaglene, thus the inferior transfixing screw would protrude outside the scapula in the case of a patient with degenerated scapular neck, such as squared-off scapular neck

(Figure 3-17). To address this problem, variable-angle locking screws is introduced in Aequalis[®] Reversed Shoulder Prosthesis (Tornier).

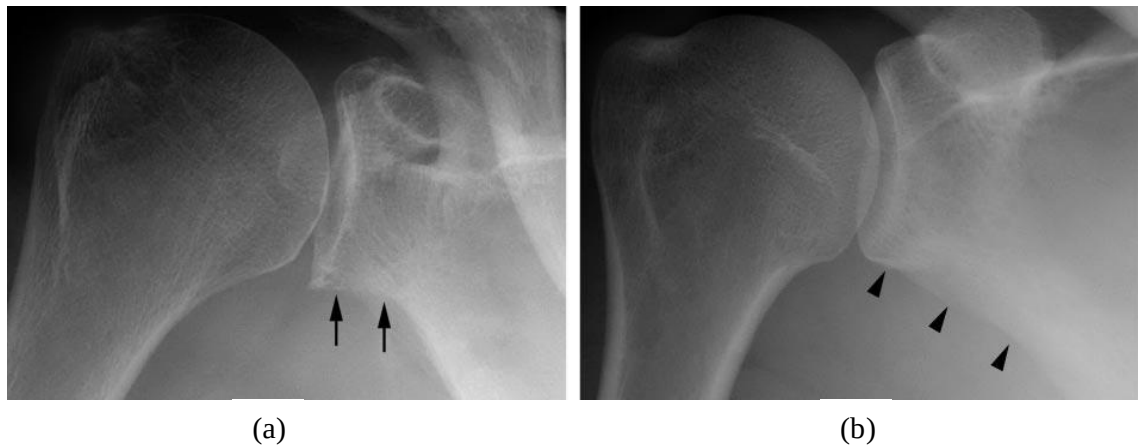


Figure 3-17 A radiographic comparison between (a) a squared-off axillary scapular border (arrows) and (b) a normal scapula (arrowheads) (adapted from Catherine 2007).

A concept of adaptable reverse from anatomical prosthesis has been applied in Zimmer shoulder system (Zimmer). In this design, the reverse and anatomical shoulder arthroplasty share the universal humeral stem and the glenoid fixation plate. The humeral head/cup and glenoid cup/sphere can be interchangeable. Thus, the humeral stem and glenoid fixation plate do not need to be removed when they are stable. Therefore, this design could protect the bone and improve the efficiency of revision.



Figure 3-18 Zimmer shoulder system which provides universal humeral stem and glenoid fixation plates.

3.3.2.4 Range of Motion and Function

Clinical experiences with the Delta RSA indicate a significant improvement in the active shoulder elevation, with a postoperative increase between 59° and 64° (Valenti et al., 2001; Sirveaux, 2004; Frankle, 2005; Werner et al., 2005; Boileau et al., 2006; Grassi et al., 2009; Naveed et al., 2011) (See Appendix A). In a review of 45 Delta shoulder prostheses for a mean follow-up of 40 months, the active shoulder elevation increased 61° after the prosthesis implantation (Boileau et al., 2005). In another similar study of 58 patients treated with Delta RSA, the postoperative active forward flexion improved 58° and the active abduction increased 47° (Werner et al., 2005). The clinical results with an increase of 50° of the forward flexion, and 60° of the active abduction in the study of Frankle (2005) also confirm the significant improvements of shoulder functions after the Delta RSA implantation.

Though the range of the postoperative arm elevation is increased considerably by using the Delta RSA, active external and internal rotations are not significantly improved (Boileau et al., 2005). In a study of 45 Delta reverse shoulder prostheses, the averaged postoperative external and internal rotations were 7° and 11°, respectively (Boileau et al., 2005). Limitations of external rotation after Delta RSA implantation could be due to the medialization of the centre of rotation. As introduced in Chapter 3.3.2.2 and Figure 3-15 (c), more anterior and posterior Deltoid muscle fibers are recruited for arm elevation in the Delta RSA. Thereby fewer posterior muscle fibers are available for external rotations. Additionally, deficiencies of teres minor muscle were reported to result in damaged effects on the range of active external rotation (Boileau et al., 2005). Deficient or absent internal rotation could be due to the lost function of subscapularis muscle. Moreover, lowering the humerus could make the remaining subscapularis muscle less efficient (Boileau et al., 2005).

Shoulder function after the Delta RSA implantation is significantly improved. The mean constant score, which is used to systematically evaluate the joint function, pain relief, and patients satisfaction, improved from 17-29 preoperatively to 56-66 postoperatively (Valenti et al., 2001; Sirveaux, 2004; Frankle, 2005; Werner et al., 2005; Boileau et al., 2006; Grassi et al., 2009; Naveed et al., 2011). In the study of Boileau et al. (2006), 78% of patients satisfied with the postoperative implant results, and 67% had no or

slight pain. In the same review, it was found that the implantation of Delta RSA was more effective for the patients with only massive cuff tear arthritis than the revision prostheses group.

3.3.2.5 Clinical Complications Associated with RSA

Reverse shoulder arthroplasty, which has been used for patients with severe rotator cuff deficiency associated with arthritis, has yielded satisfied pain relief and restoration of shoulder function (Sirveaux et al., 2004; Boileau et al., 2005; Werner et al., 2005). However, complications for this type of shoulder prosthesis are common (McFarland, 2006). A study of 457 RSA in a large French multicentre showed that the complication rate of these prostheses was 25.6% (Walch et al., 2006). In addition, this study also revealed that the complication rate of the revision arthroplasty (65%) was much higher than that of the primary arthroplasty (15%). In another study of 80 patients treated with RSA, the complication rate was 14% (Sirveaux et al., 2004). By reviewing 58 patients with the mean age of 68 year old, Werner et al. (2005) reported an incidence of complications of 50%.

One of the primary complications accompanied with Delta RSA is scapular notching, with an incidence rate of 50% to 90% (Sirveaux et al., 2004; Boileau et al., 2005; Lévine et al., 2008). Some other complications include glenoid component loosening (5% to 38%), which has been introduced in Chapter 1, polyethylene wear, and breakage of the inferior screw (Rittmeister and Kerschbaumer, 2001; Valenti et al., 2001; Sirveaux et al., 2004; Werner et al., 2005).

3.3.2.5.1 *Scapular notching*

Scapular notching is caused by the repetitive mechanical contact of the medial border of the humeral component against the scapula neck. It is an important factor for the formation of scapular erosion (De Wilde et al., 2001; Nyffeler et al., 2004; Sirveaux et al., 2004). Postoperative scapular notching mostly occurs in the inferior, posterior and anterior glenoid (Simovitch et al., 2007).

In comparison with inferior scapular notching, notches in the anterior and posterior glenoid are comparatively rare. In a radiographical study of 77 shoulders treated with Delta III RSA, anterior notch was observed in only 8% of cases (Simovitch et al., 2007). Posterior scapular notching (Figure 3-19 (a)) is associated with great external rotation in the resting position. It occurred more frequently than anterior scapular notching, but did not relate to clinical outcomes (Boileau et al., 2005; Simovitch et al., 2007).



Figure 3-19 Radiograph of a shoulder with the Delta III RSA: (a) posterior notch from the superoinferior view; (b) inferior notch from the anteroposterior view.

Inferior scapular notching was first reported in the thesis of Sirveaux in 1997 and has been one of the most frequently reported complications for RSA, due to the high incidence of 50% to 96% (De Wilde et al., 2004; Nyffeler et al., 2004; Sirveaux et al., 2004; Lévine et al., 2008; Levy and Blum, 2009; Lévine et al., 2010). In a 47-month follow-up clinical study, inferior scapular notching was observed in 62% of shoulders after the operation of Delta III RSA (Lévine et al., 2008). The incidence of inferior scapular notching was 50% in the study of Vanhove (2004), 63.6% in the study of Sirveaux (2004), and 60% in the study of Valenti (2000).

Inferior scapular notching could extend beyond the inferior screw, where the medial rim of the polyethylene humeral cup cannot reach. Reasons for the progression of severe inferior scapular notching are unknown. A general explanation is that wear particles of the polyethylene humeral cup trigger the formation of local osteolysis, leading to massive glenoid erosion (Boileau et al., 2005). The inferior scapular notching is classified into five grades based on the notch sizes in the system of Nerot (Valenti, 2000) (Figure 3-20) (Table 3-1). Grade 1 and 2 notches are safe for the stability of

glenoid prosthesis (Kalouche et al., 2010). However, effects of grade 3 and 4 notches (Figure 3-21) on the potential risk of screw fracture and secondary loosening (>7 years) have been concerned (Nyffeler et al., 2005; Simovitch et al., 2007; Kalouche et al., 2010). Additionally, scapular notching may associate with low mean relative Constant score (Constant and Murley, 1987; Sirveaux et al., 2004; Simovitch et al., 2007; Kalouche et al., 2010), subjective shoulder value (Kalouche et al., 2009) and limited range of motion (Simovitch et al., 2007).

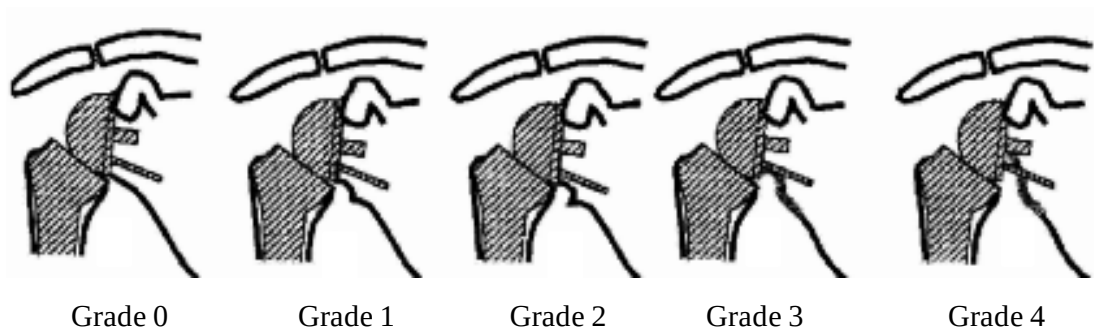
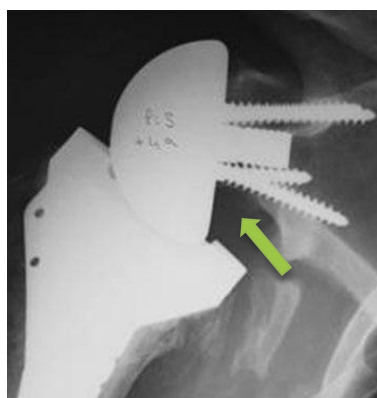


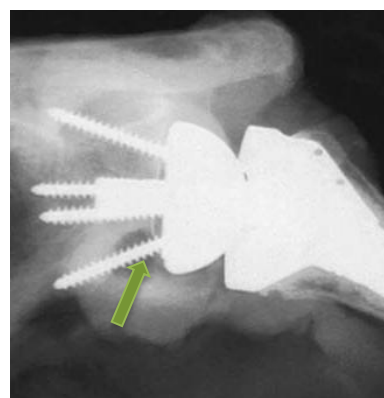
Figure 3-20 Classification of inferior scapular notching in the system of Nerot (Valenti, 2000).

Table 3-1 Classification of inferior scapular notching in the system of Nerot (Valenti, 2000).

Classification	Description
Grade 0	No notch
Grade 1	Small notch
Grade 2	Notch contacting with the inferior screw
Grade 3	Erosion over the inferior screw
Grade 4	Extension to the central peg of the metaglene



Grade 3



Grade 4

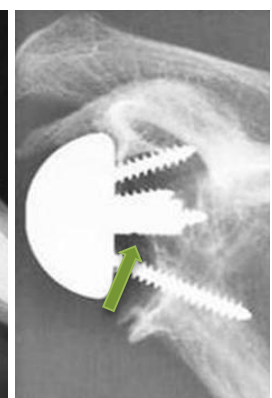


Figure 3-21 Severe erosions of scapular neck.

In recent published literatures, it was reported that early scapular notching could occur approximately 6 weeks after the operation (Sirveaux et al., 2004; L  vigne et al., 2008). There is a hot discussion about the progression of scapular notch with time. Sirveaux reported that erosions of inferior scapular neck ceased at an average of 18-month follow up. A postoperative study of RSA also revealed that 79% of the notches did not show any extension after 1 year (Werner et al., 2005). However, with a study of 337 shoulders treated with Delta RSA, L  vigne et al. (2008) pointed out that the percentage of severe notches (Nerot grade 3 and 4) significantly increased from 12% to 44% from 2- to 10-year follow up (Figure 3-22). A recent 6-year follow-up study of Delta III RSA also confirmed the development of scapular notching in both quantity and severity with time (Grassi et al., 2009).

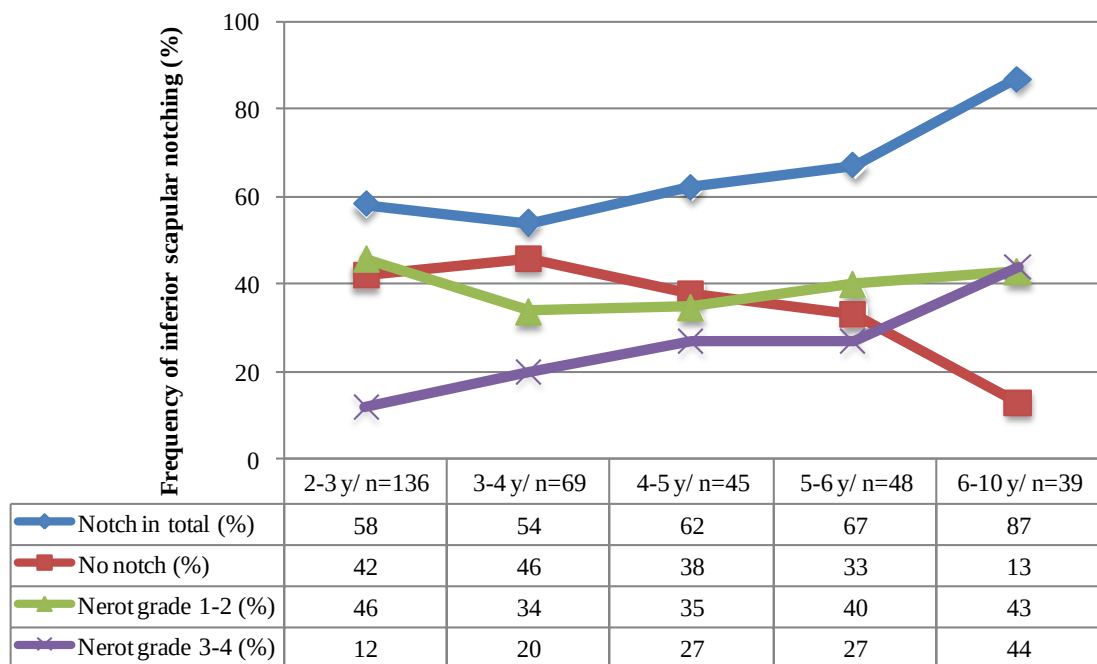


Figure 3-22 Frequency of inferior scapular notching and repartition of each grade in the Nerot system relative to the time; “y” is year, “n” is the number of scapular notching (adapted from L  vigne et al., 2008).

3.3.2.5.2 Polyethylene wear

Polyethylene wear (Figure 3-23) is a result of repetitive impingements of the medial aspects of the polyethylene humeral component against the scapular neck. Particles generated during the process of bone-prosthesis impingement could be the trigger of

osteolysis which is possibly an important factor for the formation of grade 3 and 4 inferior scapular notching (Boileau et al., 2005). With the high incidence of inferior scapular notching after orthopaedic implantation, the polyethylene wear has been concerned.

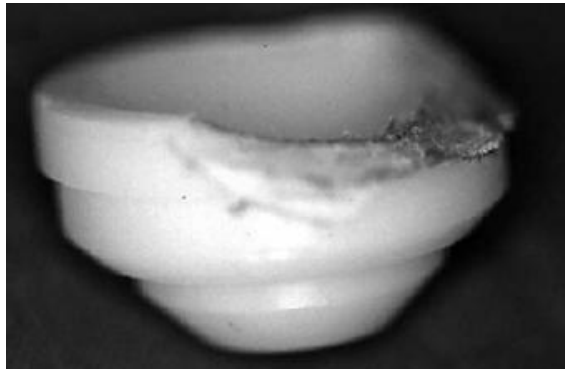


Figure 3-23 Polyethylene wear on a retrieved humeral cup

3.3.2.5.3 *Bone loss*

Bone loss (Figure 3-24) is a common radiographic finding in patients treated with the reverse shoulder prosthesis (Guery et al., 2006; Grassi et al., 2009). The glenoid erosion not only exists in the inferior region, but also in the superior, and even at the end of the middle peg. Scapular notching and wear particle-induced bone osteolysis are two common explanations for this postoperative complication of the RSA. In addition, bone resorption due to stress shielding could also be a factor for the glenoid erosion, as it is one of major factors for the bone loss in knee and hip joints (Vince et al., 1989; Whiteside, 1993). There are two major concerns relating to bone loss after the RSA. Firstly, the weakened bone foundation will lead to potential failure due to the component loosening and bone fracture. Secondly, it leads to complications in revision surgery.

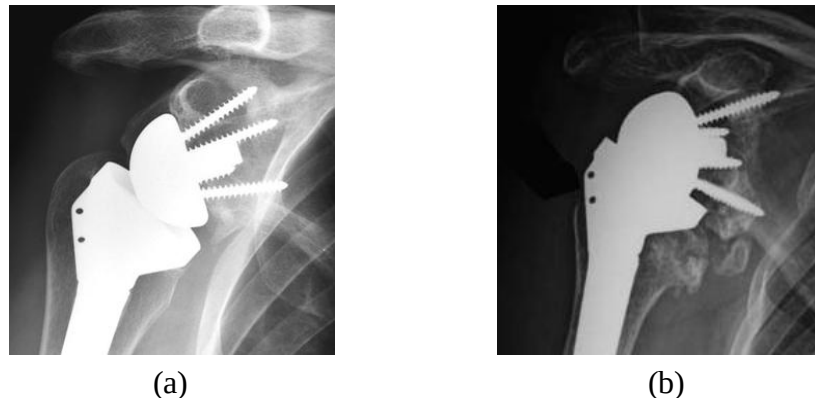


Figure 3-24 Radiograph showing the loss of bone stock on the glenoid side: (a) adapted from Guery et al. 2006; (b) adapted from Grassi et al. 2009.

3.3.2.5.4 *Breakage of screws*

Breakage of screws (Figure 3-25), particularly inferior screw, has been reported recently (Sirveaux et al., 2004; Grassi et al., 2009). Reasons of screw fracture are unknown. A general explanation for this postoperative phenomenon is inferior scapular notching (Grassi et al., 2009). A recent radiographic study showed that the breakage of inferior screw was also possibly the result of repetitive impingements between the humeral cup and the screw (Figure 3-26) (Roberts et al., 2007). In the author's opinion, glenoid prosthesis dislocation could be another factor leading to the screw fracture (Figure 3-25). In the absence of the inferior screw, fixation of the glenoid component in RSA has been concerned (Sirveaux et al., 2004). In a 6-year follow-up study of Delta III RSA, inferior screw fracture was observed in 2 patients. Additionally, inferior scapular notching and glenoid loosening were also observed in these two cases (Grassi et al., 2009).



Figure 3-25 Breakage of screws in the RSA (adapted from Juvenspan et al. 2005).

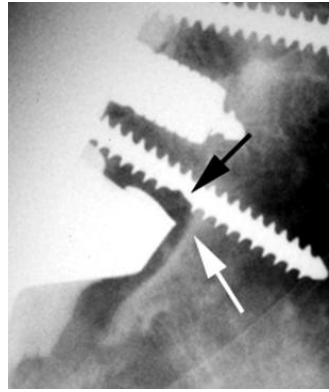


Figure 3-26 Impingement between the inferior rim of humeral cup and the inferior screw (adapted from Roberts et al. 2007).

3.4 Osseointegration

Two common types of fixations used to hold the prostheses in place are cemented and cementless fixations. Cemented fixation uses fast-curing bone cement (Polymethylmethacrylate (PMMA)) to anchor the prostheses. Cementless fixation relies on osseointegration to hold the prostheses in place. For the Grammont-style RSA, cementless technique is used for the fixation of glenoid component, as bone cement could not provide strong enough initial stability (Grammont, 1993). The humerus component is held with either cement or cementless fixation.

Osseointegration is “direct structural and functional connection between the living bone and the surface of load-bearing orthopaedic implant” (Branemark, 1983). Osseointegration requires minimum postoperative bone-prosthesis interface motion in order to avoid formation of soft tissues. Additionally, it requires a balance of sufficient loading to avoid stress shielding and bone remodelling.

Bone ingrowth refers to the process of new bone formation directly into the porous structure of an implant (Sumner and Galante, 1990). The prerequisites for bone ingrowth could be composed of three parts: (1) appropriate porous structure of the prosthesis attaching on the bone, (2) intimate contact between the prosthesis and the bone and (3) minimum postoperative bone-prosthesis interface motion (Cameron et al., 1976; Harris and Jasty, 1985; Haddad et al., 1987). A minimum pore size of approximately 50 μm is required for bone ingrowth (Klawitter et al., 1976; Robertson et

al., 1976; Bobyn et al., 1980). A pore size of 150 μm to 400 μm is optimal for fixation strength (Bobyn et al., 1980). The pore size larger than 400 μm will reduce the fixation strength (Bobyn et al., 1980). A bone-prosthesis gap between 1.5 mm and 2.0 mm could be bridged by living bone (Cameron et al., 1976; Bobyn et al., 1981). Additionally, the gap lower than 0.5 mm could enhance the rate and degree of bone ingrowth (Cameron et al., 1976). The peak bone-prosthesis interface micromotion allowing bone ingrowth varies between 20 μm and 50 μm (Pilliar et al., 1986; Jasty et al., 1997; Burke et al., 2006). The micromotion higher than 150 μm may induce the formation of fibrous tissues (Pilliar et al., 1986).

3.5 Background of Stress Shielding

Stress shielding is identified by stress transmissions from the bone to the metallic joint prosthesis. A detailed description of stress-shielding and load transmission mechanism in a knee joint is shown next (Vasu et al., 1986). In an intact tibia subjected to a bicondylar load, stresses are transmitted from the subchondral bone through the cancellous bone to the metaphyseal shell. When a short stemmed prosthesis is implanted, the stresses are directed towards the central stem, leading to low stresses underneath the tray and high stresses at the stem tip and closer to the metaphyseal cortex. With the similar structure (a wide tray and a short stem), stress shielding in the Delta RSA is studied in this thesis.

The relationship between stress shielding and bone resorption was initially found in the internal fracture fixation (Tonino et al., 1976). It was later frequently reported in the clinical studies of fixation of hip, knee, and anatomical total shoulder joint arthroplasties (Orr et al., 1988; Levitz et al., 1995; Petersen et al., 1995; Stone et al., 1999; Gupta and van der Helm, 2004; Abu-Rajab et al., 2006; Mulhall et al., 2006). Bone adaptation may yield more important effects on the cementless fixation, which is used in the reverse shoulder arthroplasty, as the loss of bone stock may hinder bone ingrowth (Lord and Bancel, 1983; Brown and Ring, 1985). Time dependence of strain-induced bone adaptation has not been well-defined. Bone changes were reported to be stabilized after about one year postoperatively (Seitz et al., 1987), but progressive late reactions have also been found (Brown and Ring, 1985).

3.6 Forces for the Delta Reverse Shoulder Arthroplasty

As introduced in Chapter 3.3.2.2, the centre of rotation of arm is medialized in the Delta RSA. This modification leads to approximately 46.3% to 78.6% increase of the moment arm of deltoid muscle (van der Helm, 1998; De Wilde et al., 2004; Terrier et al., 2008; Kontaxis, 2010). Additionally, in the study of Kontaxis (2010), the required deltoid muscle force does not change considerably after the implantation of RSA, even though the rotator cuff function is lost. Given the lack of compressive forces of rotator cuff and the moderate activation of the deltoid muscle, there is a reduction in the glenohumeral (GH) joint contact force after the implantation of Delta RSA (Terrier et al., 2008; Kontaxis, 2009). In a study of muscle forces for the Aequalis RSA (Tornier), Terrier et al. predicted that the required peak GH joint force in the case of full rotator cuff deficiency reduced approximately 50% in comparison with that for the anatomical shoulder. In addition, they also pointed out that the postoperative required force reduced only 30% when only supraspinatus muscle function was lost. By calculating required GH joint forces for more physiological daily activities (Table 3-2), Kontaxis (2010) predicted that there was a mean reduction of the peak GH force of 41.6% after the implantation of Delta III RSA. They also reported that the reduction of GH force was activity dependent, varying from 10% to 52.5%. This significant decrease of the GH force is beneficial for releasing the burden of deltoid muscle; however, it raises concerns about the postoperative stress shielding and bone remodelling.

Table 3-2 Peak GH force magnitudes for reverse shoulder arthroplasty and changes from those for the normal shoulder (adapted Kontaxis, 2010).

Task	Activities	Peak GH force magnitude ($\times$ BW)	Change from the peak GH force for normal shoulder (%)
T1	Reaching opposite axilla	0.33	23.26
T2	Reaching the opposite side of neck	0.36	10.00
T3	Reaching the back of head	0.38	22.45
T4	Eating with hand to mouth	0.28	47.17
T5	Drinking from mug (weight of mug and water: 0.4 kg)	0.29	51.67
T6	Eating with plastic spoon	0.44	12.00
T7	Answering telephone (telephone weight: 0.11 kg)	0.39	23.53
T8	Combing hair (contralateral side) (hair brush: 0.12 kg)	0.37	30.19
T9	Lifting 0.5 kg block to head height	0.65	31.58
T10	Lifting 0.5 kg block to shoulder height	0.72	17.24
T11	Reaching lower back	0.19	52.50
T12	Standing up from an armchair	1.47	34.96

CHAPTER 4

FINITE ELEMENT MODELLING OF AN IMPLANTED SCAPULA

4.1 Introduction

The finite element (FE) method was developed in early 1940s, and originally employed to solve the problem of complex elasticity and structural analyses in civil and aeronautical engineering. In 1972, this computational technique began to be employed in the study of orthopaedic biomechanics (Brekelmans et al., 1972). Since then, finite element analysis (FEA) is widely used in many biomechanical research areas, such as analyses and predictions of bone mechanical properties (i.e. stresses & strains), studies of musculoskeletal systems and investigations of artificial joint prostheses (Huiskes et al., 1987; Weinans et al., 1992; Kerner et al., 1999; Hopkins et al., 2006; Hopkins et al., 2008). The improved precision of imaging techniques and improvements of

computational capabilities make it possible to generate three-dimensional (3D) finite element models (FEMs) of bone from patient-specific computed tomography (CT) data and to assign material properties of the bone on an element-by-element basis. Thereby, 3D FE modelling of bone can provide realistic and comprehensive analyses, and has become a significant tool for the validation of clinical hypotheses (Ahir and Walker, 2004; Hopkins et al., 2005; Sharma et al., 2009).

The purpose of this thesis is to utilize FE modelling to analyse effects of scapular notching and strain-induced bone remodelling on the long-term fixation of glenoid component in the RSA. This chapter is aimed to introduce the process of FE modelling of a scapula merged with a Delta CTA RSA. Assessments of the quality of this created FEM will be introduced in Chapter 5. This chapter is composed of five parts: generation of the patient-specific geometry of a scapula associated with Delta CTA RSA, definition of bone material properties, modelling of interface conditions for the cementless fixation, set up of boundary conditions and sensitivity study.

4.2 Creating Patient-specific Geometry of an Implanted Scapula

Geometry of scapula is very irregular and patient-dependent. In this study, the patient-specific geometry of a scapula was obtained from CT data of this bone. Detailed procedures of generating the geometry of an implanted scapula are described in Figure 4-1. One cadaveric shoulder (Figure 4-1 (a)) was CT scanned in the Radiology Department at St. Mary Hospital. The CT images (Figure 4-1 (b)) were loaded in Avizo 5 (Mercury Computer Systems), which is the visualisation software allowing semi-automated medical image segmentation. Irregular and complex geometry of the scapula was generated in terms of the CT images (Figure 4-1 (c)). After that, the geometry of a Delta CTA RSA was virtually inserted into the scapula, and its position was verified by an experienced orthopaedic shoulder surgeon (Figure 4-1 (d)). With the prosthesis in place, it was followed by importing the 3D geometry of the scapula and the Delta CTA RSA into MSC. Marc Mentat (MSC Software Corporation). In this FE software, a virtual resected surface was created in the scapula (Figure 4-1 (e)). Consequently, the

bone associated with the prosthesis was meshed with finite elements. An FEM of the scapula merged with the Delta CTA RSA is shown in Figure 4-1 (f).

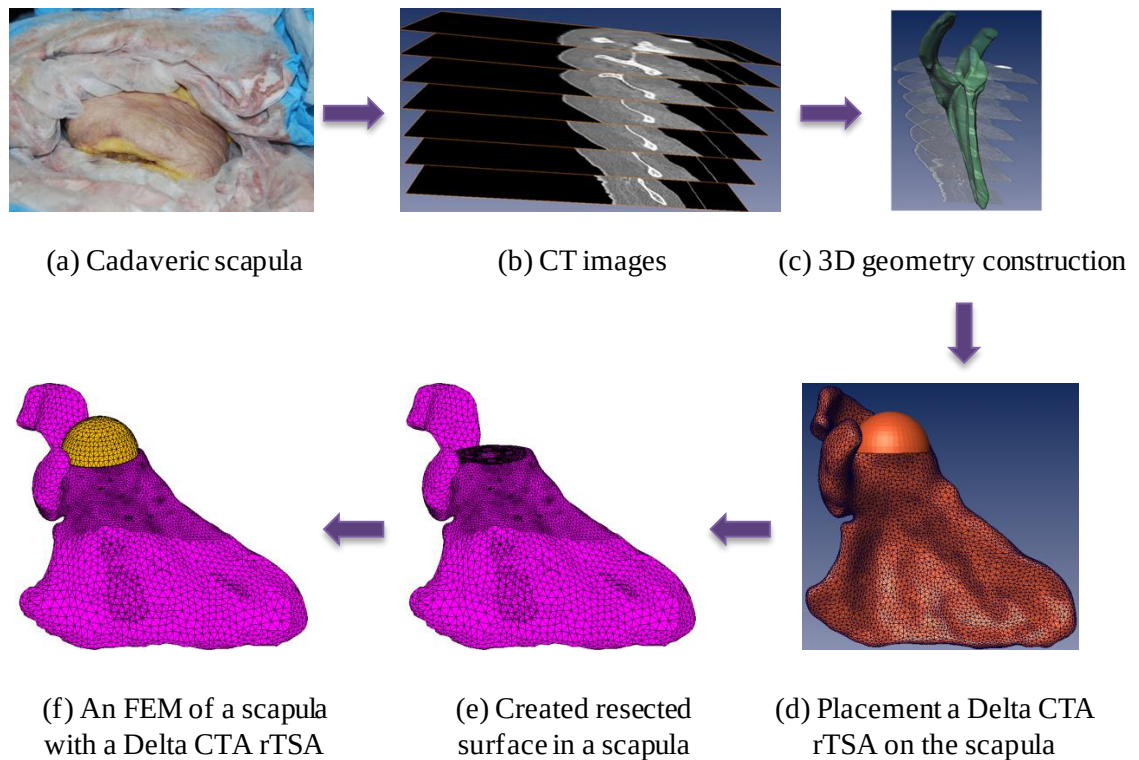


Figure 4-1 Process of creating the geometry of a scapula merged with a Delta CTA RSA.

Six cadaveric scapulae without any prior shoulder surgery or disease records (mean age 71 years) were used in this study, and would be used for the further studies of validations of FEM with in-vitro testing in Chapter 5. Some of them will be also used in the studies of effects of scapular notching and bone remodelling on the fixation of glenoid prostheses in Chapters 6 - 8. To produce mesh-independent mechanical results without prohibitive demands on computational recourses, the mesh size of lateral scapula was set as 1.5 mm, and that of the medial scapula was 3.0 mm (see mesh convergence analyses in Chapter 4.6). The information of patient-specific FEMs of the six scapulae is illustrated in Table 4-1.

Table 4-1 Information of cadaveric scapula models in MSC. Marc Mentat (MSC Software Corporation).

Subject No.	Number of tetrahedral elements	Number of nodes
1	86663	19169
2	102098	22678
3	94813	20408
4	83864	18480
5	73050	16522
6	80830	18098

4.3 Definition of Bone Material Properties

A proper definition of bone material properties is very crucial to generate an accurate FEM. To represent the site-dependent and heterogeneous characteristics of human bone, elastic moduli of bone were obtained from the CT values and assigned on an element-by-element basis. This study includes three parts: CT-apparent density correlation, modulus-density relationship and assignment of bone material properties. To assess how extent that the quantitative material properties could represent the reality, both the CT-density and density-modulus correlations are calibrated with published microstructures and Young's moduli of glenoid.

4.3.1 CT-apparent Density Correlation

A CT value, also called Hounsfield Unit (HU), is a linear transformation from the linear attenuation coefficient of a material to a radiodensity standard. In this standard, radiodensities of distilled water and air at standard temperature and pressure (STP) are defined as 0 HU and -1000 HU, respectively. Thus, the CT value of a material can be given by:

$$HU = \frac{\mu_m - \mu_{\text{water}}}{\mu_{\text{water}} - \mu_{\text{air}}} \times 1000 \quad \text{Equation 4-1}$$

where, the μ_{water} , μ_{air} and μ_m are linear attenuation coefficients of water, air and target material, respectively. As the linear attenuation coefficient of air is nearly zero, CT scanners are usually calibrated with linear attenuation coefficient of distilled water.

With this standard calibration, bone apparent density proportionally corresponds to CT values (Cann and Genant, 1980; Ciarelli et al., 1991). To create the CT value-to-density linear relationship, the maximum density of cortical bone and mean density of trabecular bone in a glenoid as well as their corresponding CT values were used. There are few references about apparent densities of glenoid. By dividing dry tissue weight by volume, Frich (1997) pointed out that the averaged apparent density of subchondral bone in the glenoid was 0.35 g/cm^{-3} . In the trabecular bone region, apparent density was reported to vary from 0.27 to 0.30 g/cm^{-3} in the study of Mansat et al. (1998), and to range from 0.17 g/cm^{-3} to 0.29 g/cm^{-3} in the study of Lehtinen (2004). Due to the dense characteristics, the maximum density of cortical bone was comparatively higher in comparison with that of cancellous bone, ranging from 1.75 g/cm^{-3} to 1.80 g/cm^{-3} (Carter and Hayes, 1977; Zannoni et al., 1998).

CT information for the six cadaveric scapulae is illustrated in Table 4-2. The mean maximum CT values of the scapulae (mean 71 year old) was 1843 HU, and corresponded to the peak apparent density of cortical bone (1.80 g/cm^{-3}) proposed by Carter and Hayes (1977). This CT-density correlation is comparable to that in the study of Zannoni et al., in which the peak radiographic CT value of human femur of 1840 HU was corresponded to the maximum apparent density (1.73 g/cm^{-3}). The mean CT value of cancellous bone in the six glenoids in this study was 203 HU. This value is comparable to that (212 HU) in the study of Mansat et al. (1998). The mean CT value of glenoid cancellous bone in this study was matched with the published averaged apparent density of glenoid cancellous bone (0.29 g/cm^{-3}) (Mansat et al., 1998). In terms of these two pairs of CT values and the corresponding densities, the linear relationship between the CT value and the apparent density was created and shown in Equation 4-2. This CT-density relationship was used in this thesis for all the FE modelling.

Table 4-2 CT information for the cadaveric scapulae of interest.

Subject No.	Peak CT value (HU)	Average CT value of trabecular bone in the glenoid (HU)	Number of voxels	Voxel sizes (mm)
1	1913	188	512×512×507	0.56×0.56×0.33
2	1783	217	512×512×788	0.85×0.85×0.33
3	1845	206	512×512×533	0.48×0.48×0.33
4	1834	208	512×512×457	0.43×0.43×0.33
5	1845	207	512×512×300	0.45×0.45×0.33
6	1838	192	512×512×402	0.48×0.48×0.33
Average	1843	203		

$$D_{\text{apparent}} = 0.0009H + 0.1072$$

Equation 4-2

In the FE modelling, marrow, of which the Hounsfield Unit in the scapula could be as low as -100 HU, was assumed to match with a CT value of 0 HU and the apparent density was defined by the linear CT value-apparent density relationship (Equation 4-2). As the elastic modulus is proportional to the cubic power of apparent density (Carter and Hayes, 1977), the differences caused by these assumption has negligible structural significance.

4.3.1.1 Calibration of Distributions of Apparent Densities in the Glenoid Region

For the purpose of verifying the CT value-apparent density relationship used in the FE modelling, distributions of FE predicted densities on the transverse plane passing through the scapula spine were compared with the regional variation of bone architecture at the same position (Frich et al., 1998) (Figure 4-2). There was a generally similar trend in the distributions of predicted densities of the six scapulae. In addition, the density distributions in the FEMs were in a good agreement with the regional variations of the glenoid bone architecture (Frich et al., 1998). In the area where cortical bone and heavy platelike trabeculae bone exist (Figure 4-2 (a)), the predicted densities (Figure 4-2 (b)) were relatively high. The highly porous cancellous bone in the deep

central region of the glenoid illustrated in Figure 4-2 (a) was able to map with the low densities shown in Figure 4-2 (b). Moreover, the predicted patient-specific geometries in the FEMs of scapulae were globally consistent to that illustrated in Figure 4-2 (a). They all displayed high degrees of curvature in the posterior glenoids in comparison with those anteriorly.

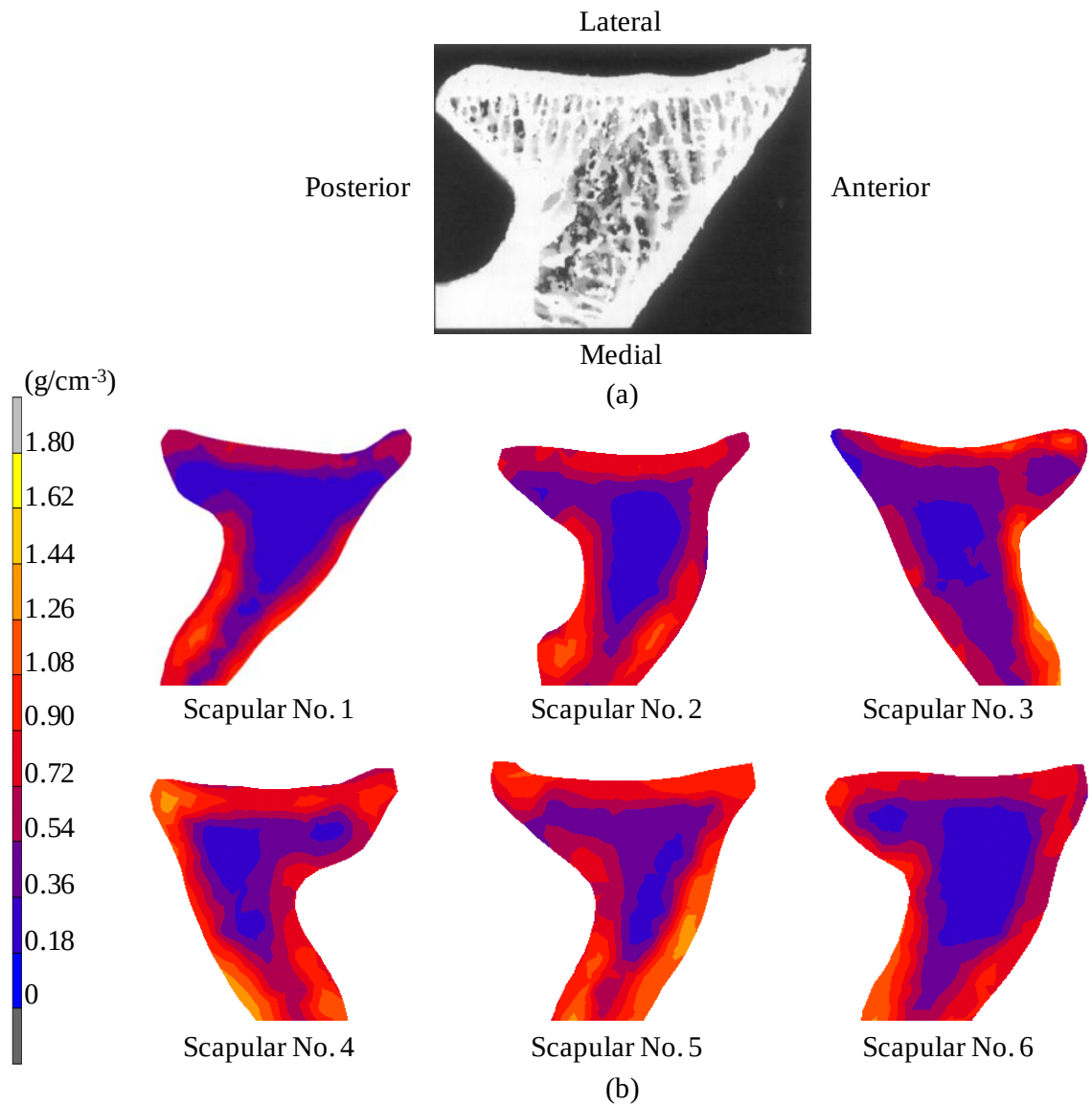


Figure 4-2 Comparison between (a) the microstructure of glenoid (adapted from Frich et al. 1998) and (b) density distributions of six cadaveric scapulae used in this thesis on the transverse plane.

4.3.2 Density-modulus Relationship

A quantitative density-modulus relationship is of significance for creating a reliable FEM of scapula. This relationship has been reported by many authors (Carter and Hayes, 1977; Lotz et al., 1990, 1991; Morgan et al., 2003) based on studies of various anatomic bones. A general equation for the relationship between apparent density and elastic modulus could be expressed by Equation 4-3, in which b varies between 1 and 3 (Carter and Hayes, 1977; Rice et al., 1988; Lotz et al., 1990; Morgan et al., 2003).

$$E = a\rho^b + c \quad \text{Equation 4-3}$$

where a , b and c are constants. E is Young's modulus, and ρ is apparent density (Currey, 1986).

By testing 100 human tibia plateau and 24 bovine cylindrical trabecular bone specimens, Carter and Hayes (1977) proposed a power constant of 3 to relate the elastic modulus to apparent density. In a specific study of trabecular bone, Rice (1988) reported that power 2 was more proper to describe the density-modulus relationship. In another study of trabecular bone, the elastic modulus was reported to be proportional to the square of apparent density for an open cell structure, and to the cube of that for a closed cell structure (Gibson, 1985). By comparing the human bone in different anatomic sites, Morgan (Morgan et al., 2003) pointed out that the power of density-modulus relationship of trabecular bone was 1.56 for vertebrae, 1.93 for proximal tibia, and 1.49 for femoral neck. In a study of scapular bone, Gupta et al. (2002) related the Young's modulus of bone to the apparent density with Equation 4-4. Variations of the published power constant of the density-modulus relationship could relate to many factors. Such as (1) differences of anatomic sites (Morgan et al., 2003), (2) anisotropy of material properties (Frich et al., 1997), and (3) various testing protocols (Carter and Hayes, 1977; Rice et al., 1988; Lotz et al., 1990; Morgan et al., 2003).

$$\begin{cases} E = 1049.25 \cdot 10^{-6} \rho^2 & \text{for } \rho \leq 350 \text{ kg} \cdot \text{m}^{-3} \\ E = 3.00 \cdot 10^{-6} \rho^3 & \text{for } 350 \leq \rho \leq 350 \text{ kg} \cdot \text{m}^{-3} \end{cases} \quad \text{Equation 4-4}$$

Although Gupta et al. (2002) reported a good agreement between the FE predicted strains of scapula and the experimental values, the density-modulus relationship

(Equation 4-4) in their study was created based on dry bones. Thus, this relationship is not proper to our FE modelling where all the cadaveric scapulae were fresh. Up to date, there have no published density-modulus relationships for fresh scapulae. In this study, the density-modulus relationship (Equation 2-1) proposed by Carter and Hayes (1977) was used. Predicted Young's moduli of the trabecular bone for individual glenoids are illustrated in Table 4-3. The distribution of Young's modulus based on this relationship will be calibrated next. The predicted strains calculated in terms of this relationship will be validated with experimental measurements in Chapter 5.

$$E = 3790\rho^3 \quad \text{Equation 4-5}$$

where E is Young's modulus (MPa), and ρ is apparent density (g/cm^{-3}).

Table 4-3 Glenoid cancellous modulus.

Scapula Number	Mean Young's modulus (MPa)
1	156
2	200
3	177
4	180
5	188
6	159
All	177 ± 22

4.3.2.1 Calibration of Predicted Young's Moduli in the FEMs of Glenoids

For the purpose of assessing reliability of predicted material properties in the FEMs of glenoids, predicted Young's moduli of glenoid trabecular bone were compared with the published values in other studies (Frich et al., 1997; Mansat et al., 1998; Anglin et al., 1999; Mimar et al., 2008).

In this study, individual Young's moduli of glenoid cancellous bone in the six scapulae were predicted to range from 156 MPa to 200 MPa, with an average of 177 MPa. This mean value is comparable with the averaged Young's modulus (166 MPa) of glenoid cancellous bone reported by Mimar et al. (2008). As introduced in Chapter 2.2.3, elastic modulus of glenoid cancellous bone varied from 99 to 264 MPa (Frich et al., 1997;

Mansat et al., 1998; Anglin et al., 1999; Mimar et al., 2008). The variation of our results from the founding of other researchers was possibly due to the various age, gender, and bone qualities of specimens. Young's moduli generally reduce with age (Keaveny and Hayes, 1993). They are higher for the old male than female (Weaver and Chalmers, 1966). Moreover, bone qualities also highly relate to healthy conditions (Keaveny and Hayes, 1993).

For the purpose of further assessing the reliability of predicted Young's modulus, FE regional values with glenoid depth (Figure 4-3) were compared with the experimental measurements proposed by Anglin et al. (1999) (Figure 4-4). In their in-situ indentation testing with ten glenoids, Young's modulus at each 3.5 mm layer underneath the subchondral bone until a depth of 31.5 mm or the cortical edge was measured. In this study, Young's moduli in FEMs of the six scapulae at the same layers with that applied in the study of Anglin et al. (1999) were recorded (Figure 4-3). The results showed that moduli of cancellous bone were patient-specific, but displayed the same tendency in the variations with glenoid depth. Young's modulus reduced first from the area beneath the subchondral layer to the honeycomb middle portion and then increased in deeper layers. By comparing this tendency with the findings of Anglin et al. (1999) (Figure 4-4), a good agreement was observed. Variations of the predicted Young's moduli from their results are possibly due to different ages of specimens. In the study of Anglin et al. (1999), the specimens were averagely 10 years older than those in this study. Moreover, the scapulae used in this study were all small and medial sizes, thus cortical bone layer appeared approximately 14.0 mm under the subchondral bone.

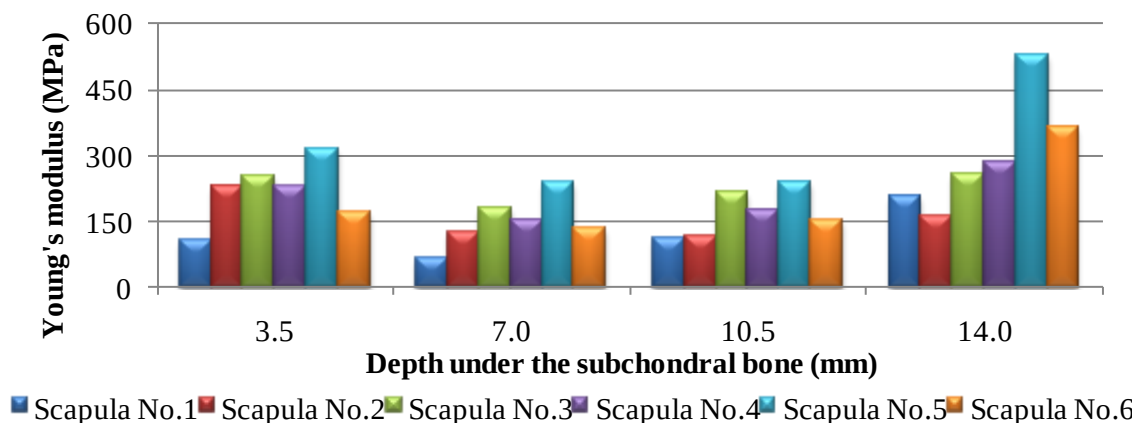


Figure 4-3 Predicted Young's moduli for the six individual glenoids at every 3.5 mm layer under the subchondral bone.

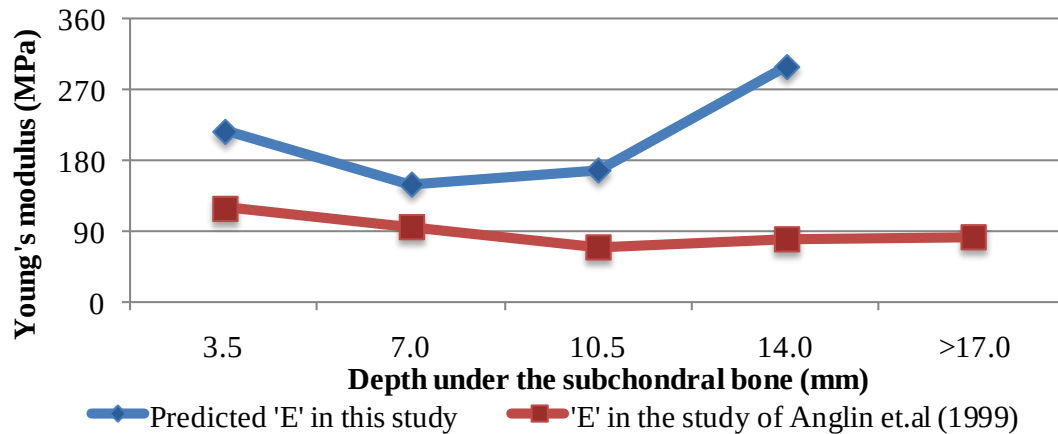


Figure 4-4 Tendency of mean Young's modulus variations by depth under the subchondral bone.

4.3.3 Assigning Material Properties to FEM

Material properties of bone are anisotropic, heterogeneous and anatomically site dependent. For the purpose of FE modelling, proper definition and simplified representation of these characteristics of bone is significant for the reliability of an FEM. Material properties have been assumed as either isotropic (Hopkins et al., 2005), or orthotropic (Maurel et al., 2005; Roche, 2009). In a study of validation of FE modelling with in-vitro testing, Gupta (2004) reported that assignment of bone as isotropic in the FEM of scapula was able to generate good predictions of strains. Isotropic bone material properties assignment has also generated good predictions in the bone remodelling (Kerner et al., 1999; Bitsakos, 2005), and micromotion studies (Hopkins et al., 2005). In comparison with the isotropic bone material property assignment, orthotropic property assignment produced small differences in von Mises stress and nodal displacement for a femur FE model (Peng et al., 2006). Regardless of anisotropy, correct heterogeneity representation in an FEM was reported to play more important role in the prediction of bone behaviour (Linde et al., 1990; Ciarelli et al., 1991). A possible reason was that the maximal change of elastic modulus along different axes was only within approximately 5 times (Frich et al., 1997), and this value could be negligible in comparison with the variation of 2-3 power when transforming from apparent density to Young's modulus (Linde et al., 1990; Ciarelli et al., 1991). Linear elastic property of bone was reported to be valid in Pugh's studies (1973). Thus, the material property of bone was assumed to be linear elastic, isotropic, and heterogeneous with a poisson's ratio of 0.26 (Frich et al., 1998) in the entire thesis.

The heterogeneous characteristics of bone were expressed by assigning material properties to an FEM at an element-by-element basis. The Young's modulus in one element was kept constant. By comparing the spatially varied material properties with the constant value within one element, Helgason (2008) demonstrated that there were no pronounced differences between these two methods in the analyses of mechanical properties. Moreover, keeping consistent Young's modulus within one element has been reported to be able to generate good predictions of strains (Gupta et al., 2004). The assignment of material properties of bone was performed with the in-house software (Biomesh) in this thesis.

4.4 Modelling of the Interface Conditions for Cementless Fixation

Cementless method was used for the fixation of glenoid component in Delta CTA RSA in attempt to promote the prosthesis initial stability and to improve the long-term fixation. It relies on osseointegration and bone ingrowth to provide biological attachments and thus functionally and structurally connect the bone with the porous-coated prosthesis. As introduced in Chapter 3, one of the prerequisites of bone ingrowth is minimal relative movements at the bone-prosthesis interface. Thus micromotion, which is defined as the relative movement between the adjacent pairs of nodes at the interface between the living bone and the load-bearing orthopaedic prosthesis, is introduced to predict the probability of bone ingrowth and thus to assess the initial stability of glenoid prosthesis. As micromotion occurred at the bone-prosthesis interface, proper setup of contact condition in the FEM is of significance.

Three bone-prosthesis interface modelling techniques have been reported. They are gap elements (node-to-node contact elements), interface elements (node-to-face or face-to-face contact elements) and direct contact (Hefzy and Singh, 1997; Abdul-Kadir et al., 2008; Hopkins et al., 2008). Both gap element and interface element methods employ FE elements to bridge the displacement discontinuity and to transmit forces between the adjacent pairs of nodes at the bone-prosthesis interface (Abdul-Kadir et al., 2008). For the direct contact method, no structural elements at the bone-prosthesis interface are employed. This method relies on the constraint of no or limited penetration to ensure the

two surfaces to keep contact. In comparison with gap elements, the use of direct contact was more proper to predict shear motion and slipping (Zachariah and Sanders, 2000). The use of gap elements for the bone-prosthesis interface modelling predicted higher micromotions (Hefzy and Singh, 1997). Moreover, Viceconti (2000) reported that gap elements could predict less accurate bone-prosthesis interface micromotions than interface elements. With direct contact technique, Hopkins (2008) obtained consistent predicted micromotions with the experimental measurements reported by Harman (2005). Due to the successful experience using direct contact technique in the analyses of prosthetic fixation (Hopkins et al., 2008; Chong, 2009), this method was used to model the bone-prosthesis interface in this study.

In the FEM of a scapula merged with Delta CTA RSA, Coulomb's friction was used to simulate the friction at the bone-prosthesis interface. Friction coefficient highly depends on the porous surface (Rancourt et al., 1990). In a study of comparing predicted micromotions with the experimental values proposed by Harman et al. (2005), a friction coefficient of 0.4 was reported to be able to generate consistent FE results (Hopkins et al., 2008). Thus 0.4 was used for the friction coefficient in this study. Influences of friction coefficient on the predictions of initial prosthesis stability will be introduced in Chapter 4.6.2. The peripheral screws in Delta CTA RSA were assumed to be secured firmly, and the contact surfaces between the bone and the screws were assumed to be bonded. Micromotions at the bone-prosthesis interface upon loading were calculated with an algorithm developed by Dr Andrew Hopkins (Post-Doctoral Fellow, Biomechanics Section).

4.5 Boundary Conditions

As most glenohumeral (GH) joint force data were obtained from Dr. Kontaxis' study (2010), coordinate system of FEM of intact scapula used in his study was set up in this study. His coordinate system was created based on the scapular frame in the ISG protocol developed by Van der Helm (1996). In the ISG protocol (Figure 4-5), AA, TS, and AI are the posterior point of acromion, the root of scapular spine and the inferior angle, respectively. Origin of the FEM is set at point AA. X axis passes through TS and AA, and points to the lateral side. Z axis is perpendicular to both $\overline{TS - AA}$ and $\overline{AI - TS}$,

pointing to the posterior. Y axis is perpendicular to X and Z axes and point to the superior.

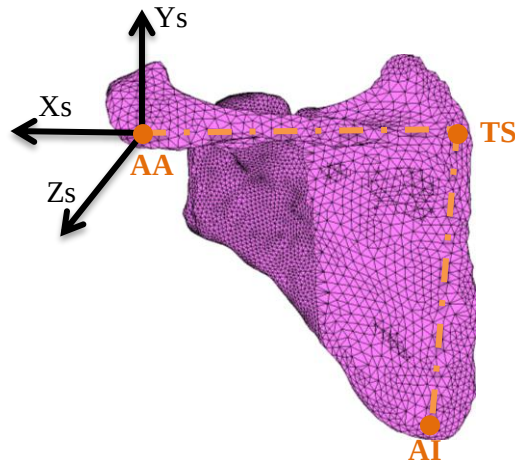


Figure 4-5 Coordinate system of scapular frame based on the bony landmark AA, TS, and AI (van der Helm, 1996).

The coordinate system of implanted scapula in this study was different from that of the intact bone. It was set based on the resected surface and the centre of rotation of the arm (Figure 4-6). The centre of rotation of the arm was the origin. X axis was perpendicular to the resected surface and faced to the lateral. Z axis was parallel with the line connecting the anterior and posterior rims of the glenoid, and pointed to the posterior. Y axis was perpendicular to both X and Z axes.

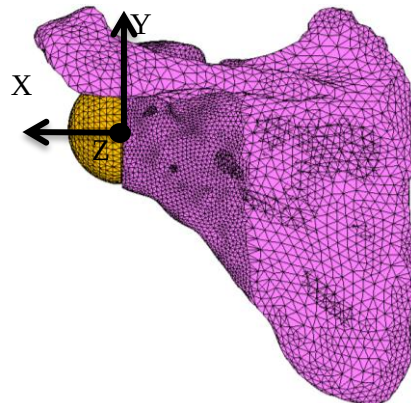


Figure 4-6 Setup of the coordinate system of an implanted scapula.

For the purpose of predicting initial stability and long-term fixation of the glenoid component of Delta RSA, physiological daily activities were interested. Peak forces for each activity were recorded in attempt to obtain the maximum stresses and bone-prosthesis interface micromotions. As the shape of scapula is complicated (wide in the

glenoid head and narrow in the scapular blade), effects of moments of component forces on the predicted stresses and micromotions are concerned. Given the non-symmetric scapula in the anterior and posterior regions, moment arms on both sides were considered. In this thesis, peak forces in 5 physiological daily activities proposed by Dr. Kontaxis (2010) were used. Two maximum forces (T8 and T11) yield the largest moment arms of shear forces in the anterior and posterior regions, respectively (Table 4-4). Two maximum forces (T12 and T9) generate the largest moment arms of compressive forces in the anterior and posterior regions, respectively. The activity of T10 was selected in this thesis due to the small required anteroposterior shear force. Based on the coordinate system of implanted scapula (Figure 4-6), force magnitudes and loading positions of peak forces for the 5 activities in cases of the Delta RSA are illustrated in Figure 4-7 (b) and Figure 4-7 (d). GH forces in the anatomical shoulder in the corresponding activities are shown in Figure 4-7 (a) and Figure 4-7 (c). In Figure 4-7 (d), the loading points mostly locate in the lower half sphere. This is possibly because the ratio between scapulothoracic and glenohumeral joint motions is 1:2 and the range of mean postoperative active arm elevation for Delta RSA ranges from 100° to 138° . In this study, 1 bodyweight (BW) is assumed to be 750 N.

Table 4-4 Loading cases used in this thesis.

Task No.	Task	Description
T8	Combing hair	The largest moment arm of shear forces in the anterior region.
T9	Lifting block to head height	The largest moment arm of compressive force in the posterior region.
T10	Lifting block to shoulder height	Very small anteroposterior shear force.
T11	Reaching lower back	The largest moment arm of shear forces in the posterior region.
T12	Standing up from an armchair	The largest moment arm of compressive force in the anterior region. The maximum force.

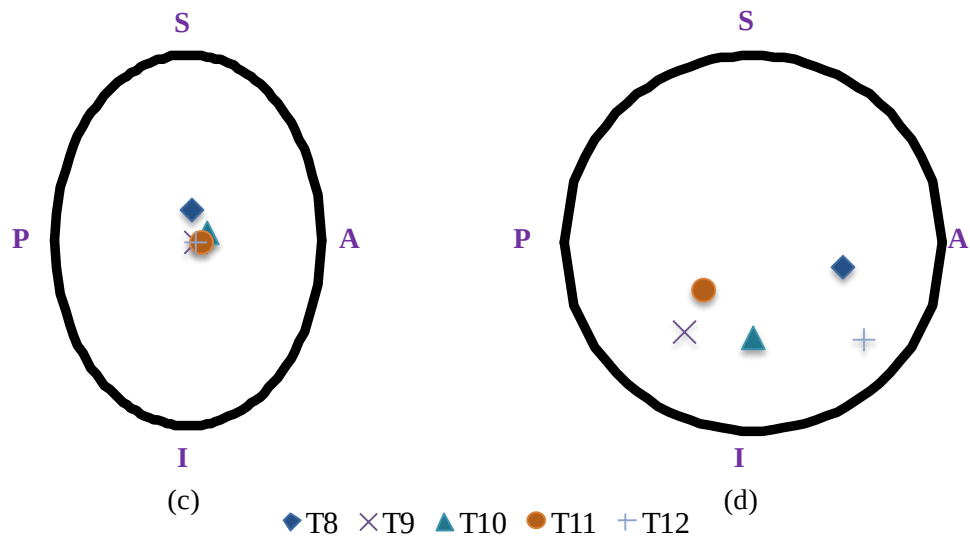
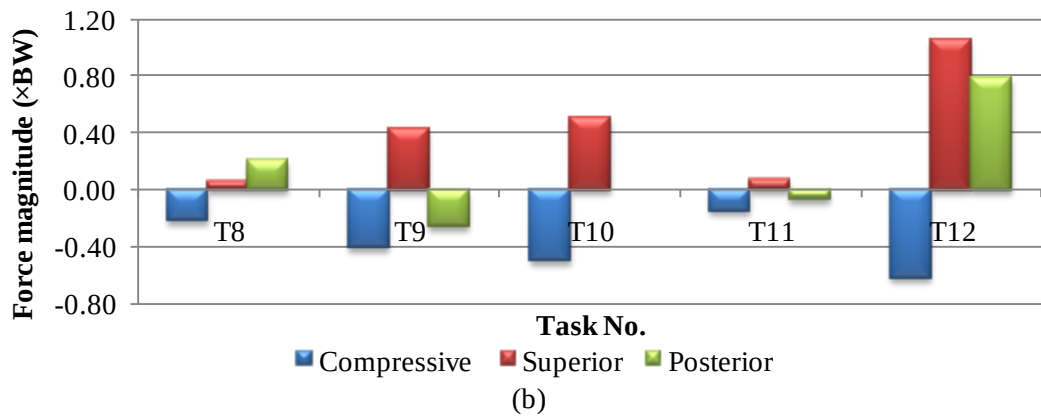
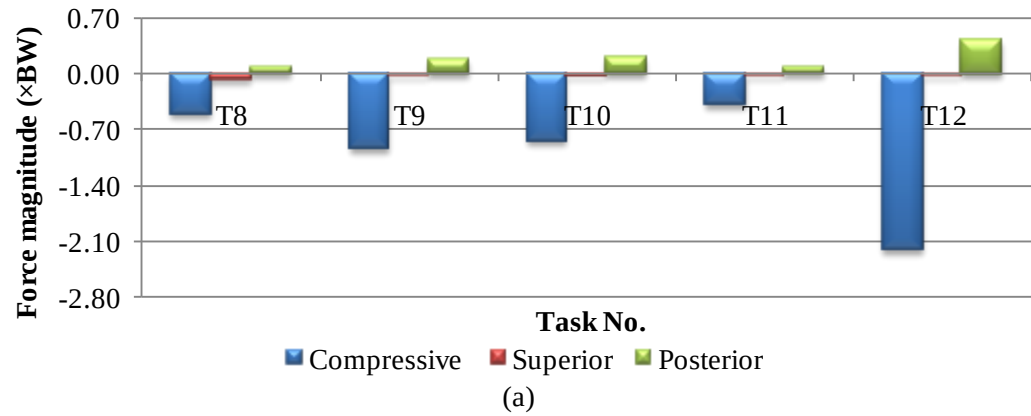


Figure 4-7 GH Forces for normal shoulder and reverse shoulder arthroplasty in different physiological daily activities: (a) GH force magnitudes for normal shoulder; (b) GH force magnitudes for Delta RSA; (c) Loading positions of GH force in the normal shoulder; (d) Loading positions of GH force in the Delta RSA (Adapted from Masjedi, 2009 and Kontaxis, 2010).

4.6 Mesh Convergence Study and Effects of Friction Coefficient on the Predicted Bone-prosthesis Interface Micromotions

4.6.1 Mesh Convergence Study

Mesh convergence study was performed to search a proper mesh size to produce mesh-independent mechanical results without prohibitive demands on computational resources. As the parameters of interest in this study are stress and micromotion, effects of mesh convergence on these two mechanical parameters were studied.

4.6.1.1 Materials and Methods

4.6.1.1.1 *Mesh-induced effects on the predicted stresses*

The FE modelling introduced in Chapter 4.2, Chapter 4.3 and Chapter 4.4 was used in this study. Geometric surface meshes in the FEM were modified with a function called Patran Surface Mesher in MSC.Marc Mentat (MSC Software Corporation) into three interested element sizes: approximately 1.5 mm, 2.0 mm, and 3.0 mm, creating three FEMs (Figure 4-8). Another function called Automatic Solid Meshing in MSC.Marc Mentat was used to convert the new surface meshes into solid tetrahedral elements. Information of these three FEMs is shown in Table 4-5. Glenohumeral joint forces for random physiological daily activities can be used in this mesh convergence study. The peak GH force when lifting a block to the shoulder height (T9) (Figure 4-7) was applied in the FEMs. One-third of the bone from the medial side was fixed. Principal stresses in the three FEMs were calculated and compared.

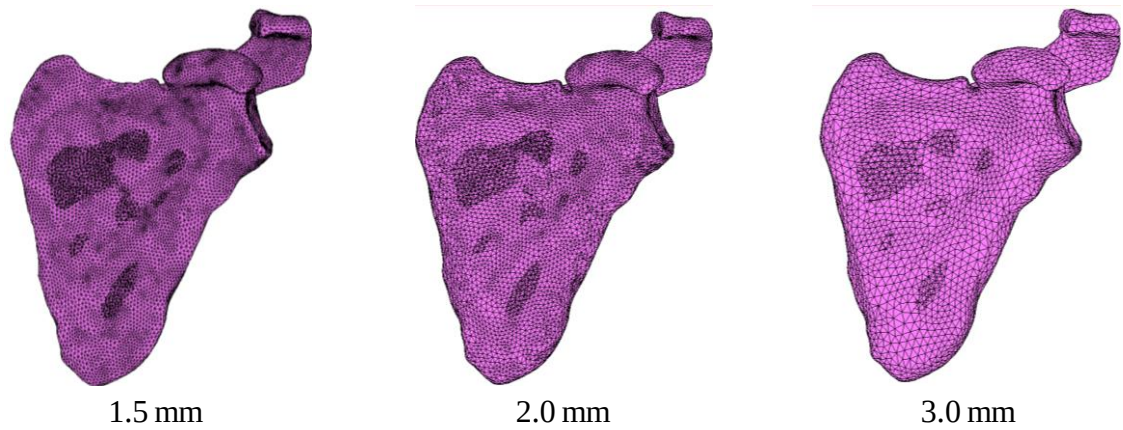


Figure 4-8 Intact scapula created with different size of meshes.

Table 4-5 Information of FEMs with various mesh sizes.

Mesh size	Number of nodes	Number of solid elements
1.5 mm	37274	179636
2.0 mm	17323	77762
3.0 mm	6257	25673

4.6.1.1.2 Mesh-induced effects on the predicted bone-prosthesis interface micromotions

FEM of the Delta CTA RSA was virtually inserted into the FEM of a scapula according to the surgical technique of this type of prosthesis (Detailed procedure of glenoid prosthesis implantation would be introduced in Chapter 5.3.2) and under the guidance of an experienced orthopaedic shoulder surgeon. Geometric surface meshes at the bone-prosthesis interface were modified with two element sizes: approximately 1.0 mm and 1.5 mm (Figure 4-9). Definition of bone material properties introduced in Chapter 4.3 was used in this study. As both the glenosphere and metaglene are made of cobalt-chromium alloy, the Young's modulus for them was set as 220 GPa. The Young's modulus of titanium peripheral screws was set as 110 GPa. GH forces for random physiological daily activities can be used in this mesh convergence study. The peak GH force for the Delta RSA when lifting a block to the head height (T9) (Figure 4-7) (Kontaxis, 2010) was applied in the FEMs. Micromotions between the bone and the prosthesis in the two FEMs with different mesh sizes were calculated and compared.

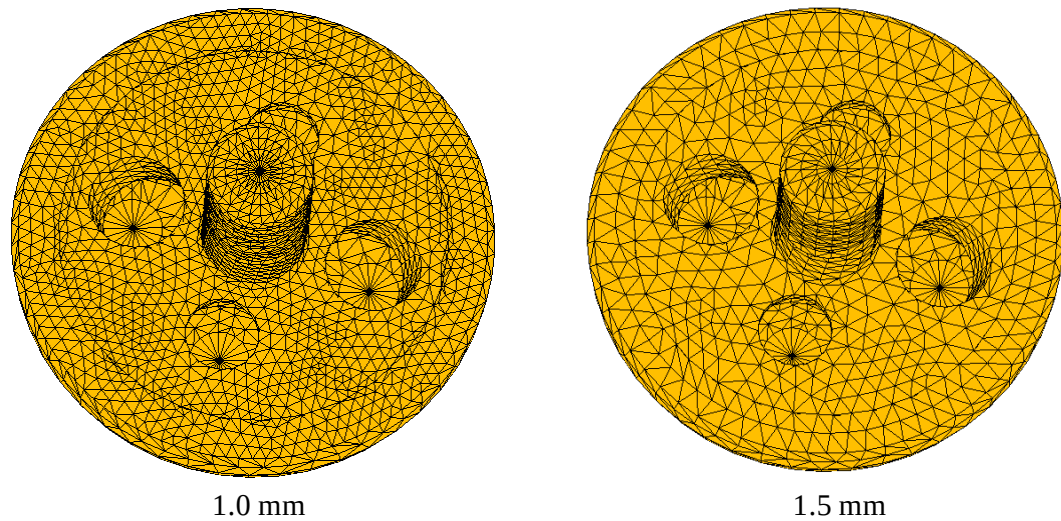


Figure 4-9 Bone-prosthesis interface created with different sizes of meshes (approximately 1.0 mm and approximately 1.5 mm).

4.6.1.2 Results

4.6.1.2.1 *Mesh-induced effects on the predicted stresses*

Three FEMs of intact scapulae were created with increasing mesh sizes (1.5 mm, 2.0 mm, 3.0 mm). Differences of predicted mechanical parameters between these three models were used to assess effects of mesh convergence on the results. As maximum principal stresses will be used to assess the bone fracture in Chapter 6, distributions of this mechanical parameter in the three FEMs are displayed in Figure 4-10. There were no significant differences in the stress distributions between the FEMs with different mesh sizes; although the predicted stress values in the three FEMs were different.

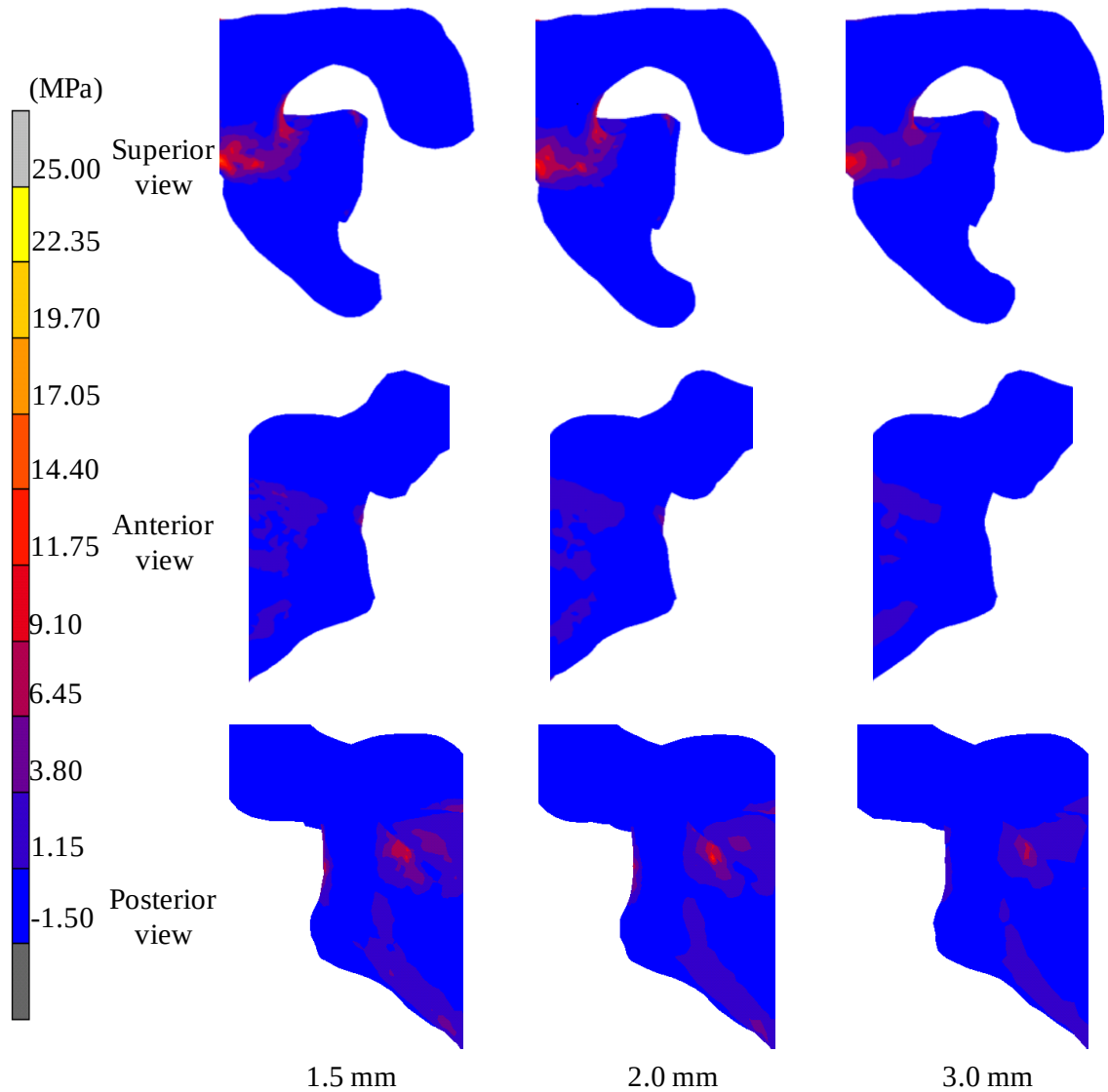


Figure 4-10 Maximum principal stress distributions around the glenoid with varied mesh sizes.

Figure 4-11 illustrates quantified comparisons of the maximum principal stresses at twelve random nodes locating around the glenoid in the three FEMs. These nodes were located at the exactly same places in the three models in order to allow proper and exact comparisons. The results showed that predicted levels of stresses generally decreased with the increase of the element sizes. The model with the mesh size of 2.0 mm and 3.0 mm converged with a maximum variation of 13.2% and 22.2% relative to the model with the finest meshes.

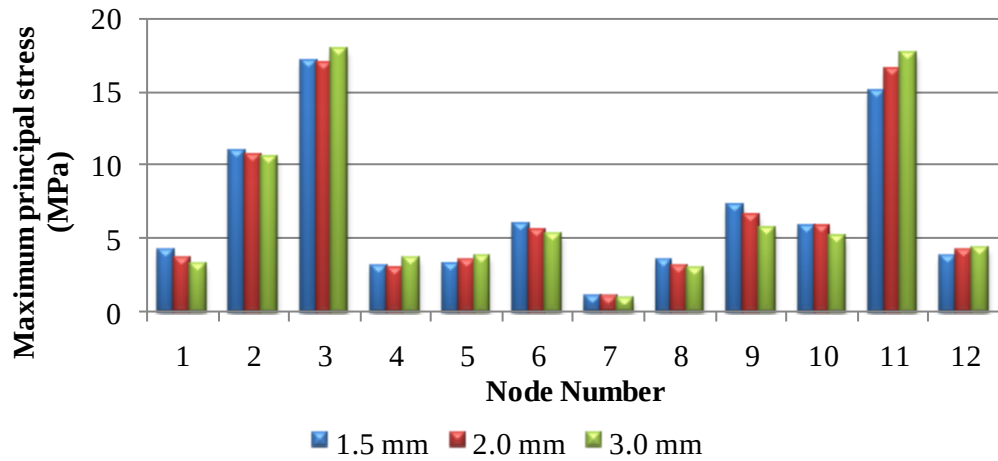


Figure 4-11 Maximum principal stresses at 12 random nodes around the glenoid in the FEMs with various mesh sizes.

4.6.1.2.2 *Mesh-induced effects on bone-prosthesis interface micromotions*

Distributions of predicted bone-prosthesis micromotions calculated from the two FEMs with different element sizes (approximately 1.0 mm and 1.5 mm) are shown in Figure 4-12. The results showed that the FEM with comparatively rough meshes displayed similar micromotion distributions to that with fine meshes. Both models predicted high micromotions at the tip of the middle peg and in the superior and inferior rims. In the further study of micromotion variations between the two FEMs, predicted values at 12 random nodes at the bone-prosthesis interface were compared (Figure 4-13). These nodes located at the exactly same positions in the two FEMs in order to allow proper and exact comparisons. With a reference to the results in the FEM with fine meshes, the maximum variation of bone-prosthesis interface micromotions in the FEM with rough meshes was 9.8%. For the purpose of investigating initial stability of glenoid prosthesis, maximum micromotions obtained from the two FEMs were interested. The predicted peak micromotion in the rough FEM (2 mm meshes) was only 1.0 μm smaller than the value (25.4 μm) in the fine FEM.

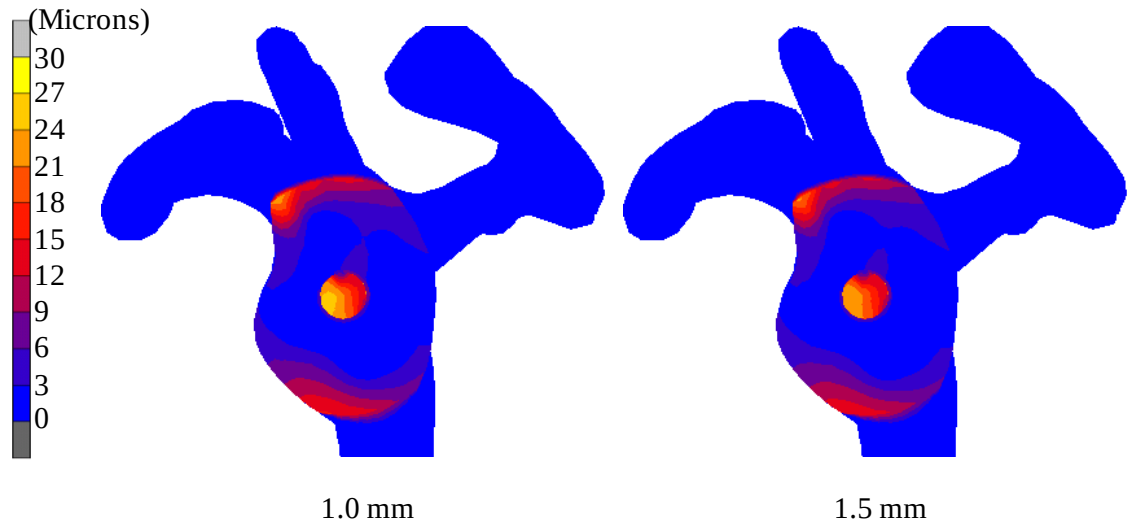


Figure 4-12 Predicted micromotions at the bone-prosthesis interface with different element sizes.

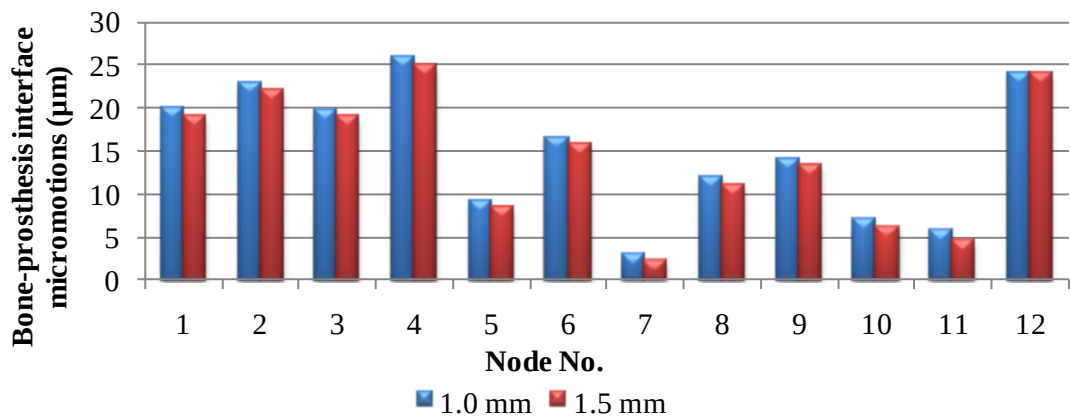


Figure 4-13 Predicted micromotions at 12 random nodes at the bone-prosthesis interfaces in the two FEMs with different sizes (1.0 mm and 1.5 mm) of elements.

4.6.1.3 Discussion and Conclusion

The purpose of this study is to investigate mesh-induced effects on the predicted stresses and bone-prosthesis interface micromotions. Three element sizes (1.5 mm, 2.0 mm, and 3.0 mm) were used to create the FEMs of scapula. Predicted stress distributions calculated from these three FEMs (Figure 4-10) showed that there was a general agreement between them. The further comparisons of maximum principal stresses at 12 random nodes around the glenoid in the three FEMs (Figure 4-11) showed that stresses in the FEMs with 2.0 mm and 3.0 mm elements maximally varied at 13.2%

and 22.2% from those in the FEM with 1.5 mm meshes. The required number of elements in these three FEMs with different size of elements (Table 4-5) showed that the FEM with 2-mm and 3-mm elements required almost half and 1/8 of the number of elements in the FEM with finest meshes. As the purpose of this study is to predict factors leading to long-term glenoid prosthesis loosening, postoperative stress changes in the glenoid region were of the most interests. In order to protect the accuracy of predicted stresses in the glenoid region, the element size in this region was set as 1.5 mm. 3.0 mm meshes were used to create the remaining bone to reduce the number of elements of the entire scapula.

In the study of mesh-induced effects on the interface micromotions, 1.0 mm and 1.5 mm element sizes were used to create FE bone-prosthesis interfaces, respectively. The results displayed that the FEM with low element density led to about 9.8% variation from the fine FEM. The predicted peak value in the FEM with 1.5 mm was only 1.0 μm smaller than that in the FEM with fine meshes. In addition, both these two FEMs displayed generally consistent micromotion distributions at the bone-prosthesis interface. Therefore, 1.5 mm elements were used to model the interface between the bone and the glenoid prosthesis in the entire thesis.

4.6.2 Effects of Friction Coefficients on Predicted Micromotions

4.6.2.1 Introduction

Cementless fixation, which relies on osseointegration to hold the prostheses in place, is used to secure the glenoid component in the Delta RSA. Bone-prosthesis interface micromotions were used to predict the possibility of bony ingrowth and then to analyse the risk of aseptic glenoid loosening. In the FEM of implanted scapula, unbonded bone-prosthesis interface was modelled by defining it as contact. Effects of prosthesis surface coating on the predicted interface micromotions have been expressed by friction coefficients (Biegler et al., 1995; Kuiper and Huiskes, 1996; Ando et al., 1999; Hopkins et al., 2008). This section is aimed to investigate the effects of friction coefficient on the predicted bone-prosthesis micromotions.

4.6.2.2 Materials and Methods

As the coefficient of friction between 0.2 and 0.9 has been used to model the prosthetic surface coating from smooth surface to cast mesh coating, this range of friction coefficient was interested. Five FEMs of scapulae merged with the Delta CTA RSA were created. Coefficients of friction at the bone-prosthesis interface in these models were different ranging from 0.1 to 0.9 with an increment of 0.2. The method, with which FEM of an implanted scapula was created in Chapter 4.6.1.1.2, was used in this study. Interface micromotions in these five FEMs were calculated.

4.6.2.3 Results

A sensitivity study of effects of friction coefficients on the predicted bone-prosthesis interface micromotions was performed. As the peak micromotion was used to be compared with the required threshold of bone ingrowth, this mechanical parameter was used to assess influences of friction coefficient on the prosthesis initial stability. Predicted peak micromotions for the FEMs with various friction coefficients ranging from 0.1 to 0.9 are illustrated in Figure 4-14. There were marginal changes in the micromotion distributions for the five models. The predicted peak micromotion in the FEM with a coefficient of friction of 0.9 was 0.05 μm lower than that with a friction coefficient of 0.2.

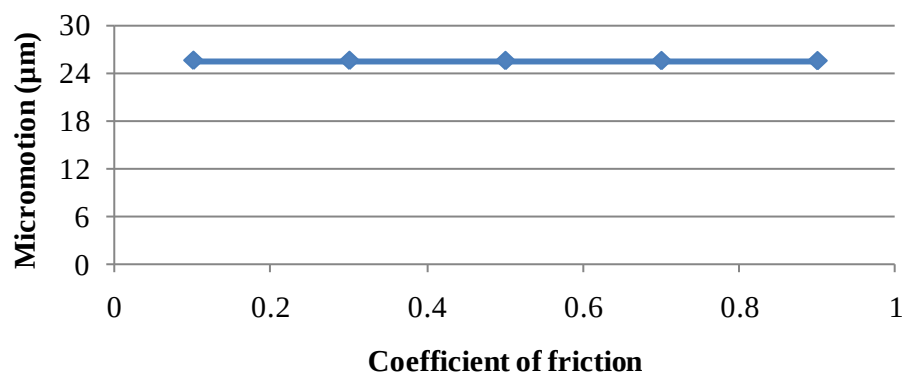


Figure 4-14 Predicted peak micromotions in the case of various friction coefficients at the bone-prosthesis interface.

4.6.2.4 Discussion and Conclusion

Effects of friction coefficients on the predicted bone-prosthesis interface micromotions were studied. Peak micromotions in the five FE models with various coefficients of friction ranging from 0.1 to 0.9 with an increment of 0.2 were calculated (Figure 4-14). The results showed that there were no significant influences of coefficient of friction on the predicted micromotions. This phenomenon could be explained as follows: firstly, except for the bone ingrowth, fixations of the four peripheral screws (two locked and two non-locked screws) also play a significant role in promoting the prosthetic initial stability. Thereby it could prohibit influences of bone-prosthesis interface friction coefficients on the micromotions. Additionally, glenoid component in the Delta CTA RSA is composed of a tray and a middle stem. The shape of the glenoid prosthesis looks like a capital letter “T”. With this structure, shear movements of the prosthesis relative to the bone were restrained by the hole where the middle stem was embedded securely. The downward movement of the prosthesis was constrained by the bone under the implant. As almost the whole resected surface of the bone was covered by the glenoid component, the majority bone at the resected surface could restrain the downward movements of the implant upon the compressive loading. In comparison with the constraint caused by the T-shape structure of glenoid prosthesis, coefficient of friction may play less significant roles in the influences on bone-prosthesis interface micromotions. In a sensitivity study for the coefficient of friction, Chong (2009) also reported a small frictional influence on the predicted bone-prosthesis interface micromotions.

CHAPTER 5

VALIDATION OF THREE-DIMENSIONAL FINITE ELEMENT MODELLING

5.1 Introduction

Finite element (FE) modelling was used as an important tool to investigate factors influencing the fixation of reverse shoulder arthroplasty (RSA) in this thesis. However, the results predicted by this method rely significantly on the quality of FE modelling. Hence, an important part of this thesis was dedicated to provide confidence for the FE analyses. Bone remodelling could be triggered by changes of strain, and the initial stability of an implant is evaluated with micromotions, thereby validating FE predictions of these two parameters with parallel in-vitro testing results is the purpose of this chapter.

In-vitro testing has been used as a common tool to validate FE modelling of a cadaveric scapula (Maurel et al., 2002; Gupta et al., 2004). The principle is shown in Figure 5-1. Bone strains in the positions of interest on a scapula upon loading were measured. In parallel, finite element models (FEMs) of the bone were created from CT data of the cadaveric scapula used in the experiments. Boundary conditions in the FEMs were set to represent the in-vitro loading and supports. Finally, predicted FE results were compared with the experimental strain measurements in the corresponding locations. Experimental-FE differences were used to assess accuracy of the FE modelling.

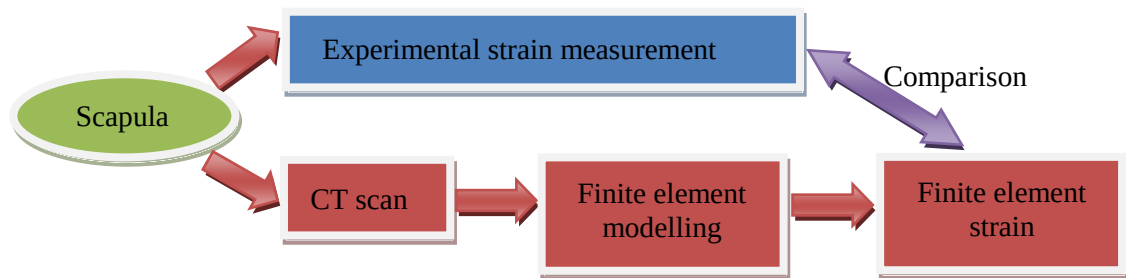


Figure 5-1 The procedure of validating finite element modelling with in-vitro testing.

Electric resistance strain gauging is one of the most commonly used strain measurement techniques (Miles and Tanner, 1992). The fundamental principle is that deformation of a specimen upon loading triggers changes of foil geometries of the strain gauges on the specimen, and thus induces variations of electric resistance of the gauge (An, 1999). The strain value varies proportionally to the change of the gauge electric resistance, as shown in Equation 5-1.

$$\frac{\Delta R}{R} = G \frac{\Delta L}{L} \quad \text{Equation 5-1}$$

where G is the gauge factor, R is the electric strain gauge resistance, L is the gauge length. One of the advantages of electric resistance strain gauging is that gauges are very sensitive and can be directly used for the strain measurements (Miles and Tanner, 1992). This strain gauging technique has been successfully used for the study of strain and stress distributions in cadaveric bone and in the validation of computational simulations (Maurel et al., 2002; Gupta et al., 2004).

5.2 Validation of the Bone Remodelling Signal (Strain)

5.2.1 Introduction

Experimental validation of three dimensional (3D) FE modelling of scapula has been reported in a few published papers (Gupta et al., 2004; Maurel et al., 2005). In the study of Maurel et al. (2005), the spongy bone was assumed to be anisotropic and partly homogeneous. One loading case was used to study the correlation between experimental and FE strains. In another study of the validation of FE modelling, CT data were used to obtain material properties of scapula, and Young's modulus was assigned on an element-by-element basis. The high correlation between experimental and FE results indicated a believable FEM (Gupta et al., 2004). However, in these studies, either only one load case was used, or FEM of only one scapula was validated. An FE modelling methodology for more subject-specific scapulae and in more loading conditions is needed.

In this thesis, heterogeneous characteristics of bone were represented by assigning material properties on an element-by-element basis. Eight positions around each glenoid were used to validate the reliability of FE modelling. In this study, to avoid effects of tedious boundary conditions and complicate implantations, models of six intact scapulae bearing compressive forces were used for the validation.

5.2.2 Materials and Methods

5.2.2.1 In-vitro Study

Six frozen cadaveric shoulders (mean age 71 years) (Table 5-1) with no prior shoulder surgery or diseases records, were obtained from Science Care (Science Care, USA), which is a professional organization providing donated human tissues for medical researches. The shoulders were stored at -20 °C, and thawed only one night before the in-vitro testing in order to keep the samples fresh.

Table 5-1: Cadaveric scapula information.

Subject Number	Gender	Age	Side
1	Female	70	Right
2	Male	66	Right
3	Female	71	Left
4	Female	73	Left
5	Female	71	Right
6	Female	72	Right

5.2.2.1.1 *Preparations of the tested scapulae*

Scapulae were first separated from the shoulders and soft tissues on the scapula surfaces were removed. After that, labrum was carefully cut off to ensure that the coordinate system in the experiment was set in terms of the glenoid bone, and thus to be consistent to that in the FEM. Before testing, each scapula was first fixed in a rectangular container. The initial position of scapula was rotated 90° anticlockwise relative to that used in the study of Terrier et al. (2008). In their approach, the scapular plane was positioned parallel to the vertical axis and the glenoid centre line was located horizontally. Approximately one-third of the scapula on the medial border side was rigidly fixed in the bone container with Stryker Simplex[®] polymethylmethacrylate (PMMA) bone cement.

5.2.2.1.2 *Strain gauge position in a scapula and strain measurements*

Before mounting strain gauges, careful preparations of the scapular surface, where strain gauges would be bonded were performed. In this study, the procedure indicated in the study of Miles and Tanner (1992) was executed, as it was found to be able to not only generate a strong enough bond but be beneficial for the bone to restore to its saline environment. Details of the procedure are described below. Firstly, the specific location where a gauge would be bonded was marked, and then was cleared of all the periosteum. Next, these positions were abraded with a piece of fine silicon-carbide paper (400 grit), and then cleaned. To make the target surface dried and degreased as suggested by Wright (1979), a CSM-2 degreaser (Vishay Micro-Measurements Group,

UK Ltd) was used. Afterwards, the scapular surface of interest was covered by a thin layer of M-Bond catalyst (Vishay Micro-Measurements Group, UK Ltd), and then was sealed with a thin layer of M-Bond 200 adhesive (Vishay Micro-Measurements Group, UK Ltd) prior to bonding a strain gauge (Miles and Tanner, 1992). Finally, one drop of the same adhesive for sealing was placed on the prepared surface and one gauge was pressed against the surface for approximately one minute with the finger pressure.

Due to the complicated surface of a scapula, uniaxial strain gauges with a small gauge length of 2 mm (FLA-2-11, Tokyo Sokki Kenkyujo Co., Ltd.) were used in the in-vitro testing. Eight strain gauges in total were bonded on the surface of each scapula. As vertical lengths of the middle peg and peripheral screws in the Delta RSA are 16 mm and between 24 mm and 48 mm, respectively, strain gauges were located around the glenoid at two levels: roughly 10 mm and 25 mm under the glenoid fossa. These positions were chosen in attempt to investigate postoperative strain shielding in the middle of the peg and in the middle of the screw after the prosthesis implantation (stress shielding study is introduced in Chapter 3.5). Due to the irregular and complex scapular surface, strain gauge locations on the six scapulae were not exactly the same for the purpose of finding comparatively flat surfaces. Except for the two inferior strain gauges which were mounted parallel to the lateral border, all other gauges were orientated approximately vertically on the scapular surface (Figure 5-2).

All the strain gauges were connected to a Model P3 strain indicator and recorder (Vishay Micro-measurements Inc. USA). For the purpose of temperature compensation, a half-bridge circuit configuration was chosen. The dummy strain gauges were fixed on a fresh scapula which was not under loading. The instrument setup was calibrated with a standard three-point bending test.

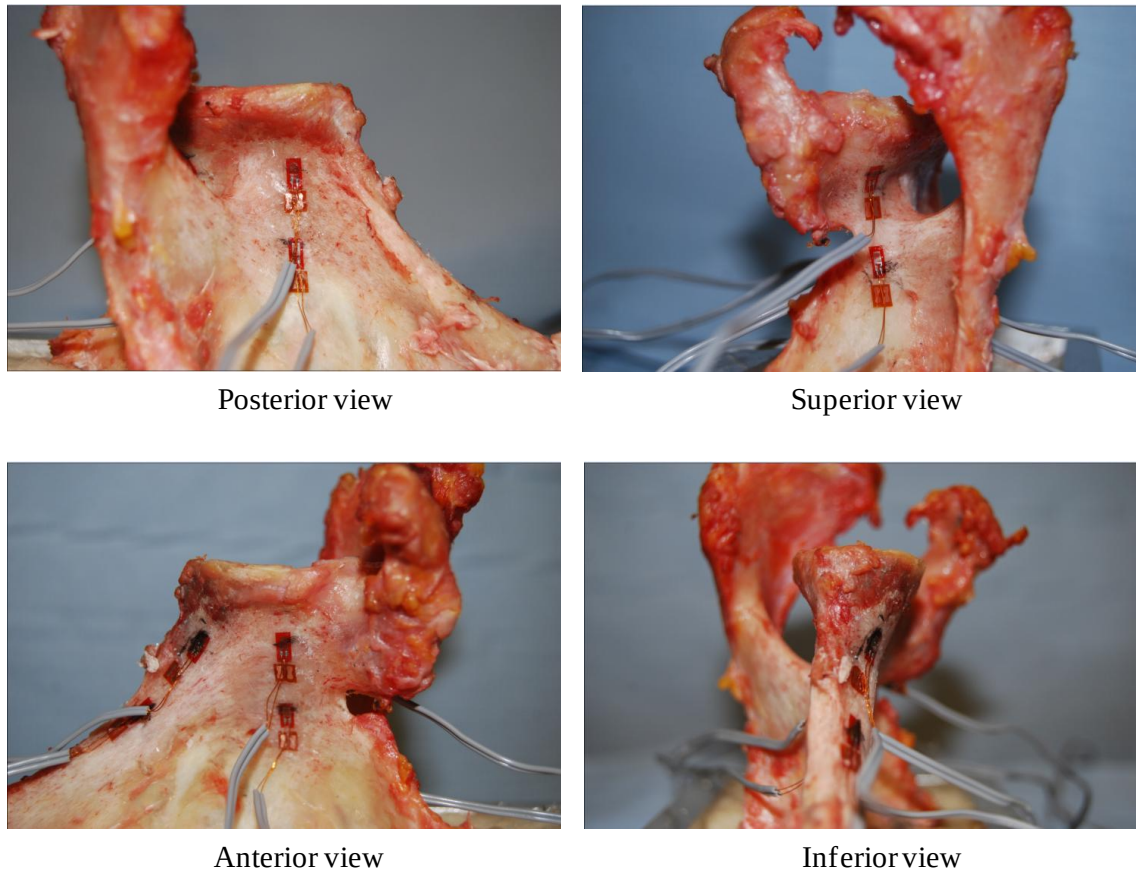


Figure 5-2 Positions of strain gauges on the cadaveric scapula surface.

5.2.2.1.3 *Loading condition*

Figure 5-3 shows the experimental setup of an intact cadaveric scapula. The tested scapula and its bone container were firmly fixed on the testing platform in an Instron machine 5565 (Instron Limited, High Wycombe, UK). A 36 mm-diameter half sphere humeral head was used to apply forces on the glenoid. As the glenoid had lost muscles and soft tissues which contribute to the dynamic stabilization of the shoulder, there were severe dislocations of the humeral head under external shear forces. Thereby, only compressive force was performed in this testing. The load magnitude was set by the Instron machine. To make the humeral head naturally contact with the glenoid surface, the head was allowed to be rotated in a very small range. Four vertical load conditions, which were 300 N, 450 N, 600 N and 750 N, respectively, were tested. Strains bonded around the glenoid under each load case were recorded. To reduce effects of viscoelastic properties of bone, tested scapulae were allowed to restore for approximately five minutes before next loading began. In each load case, the test was repeated three times.

The average was used to represent the strain in that loading condition. When studying the subject-specific characteristics of glenoid strains, t-test was used. A p value of less than 0.05 was considered to be statistically significant.

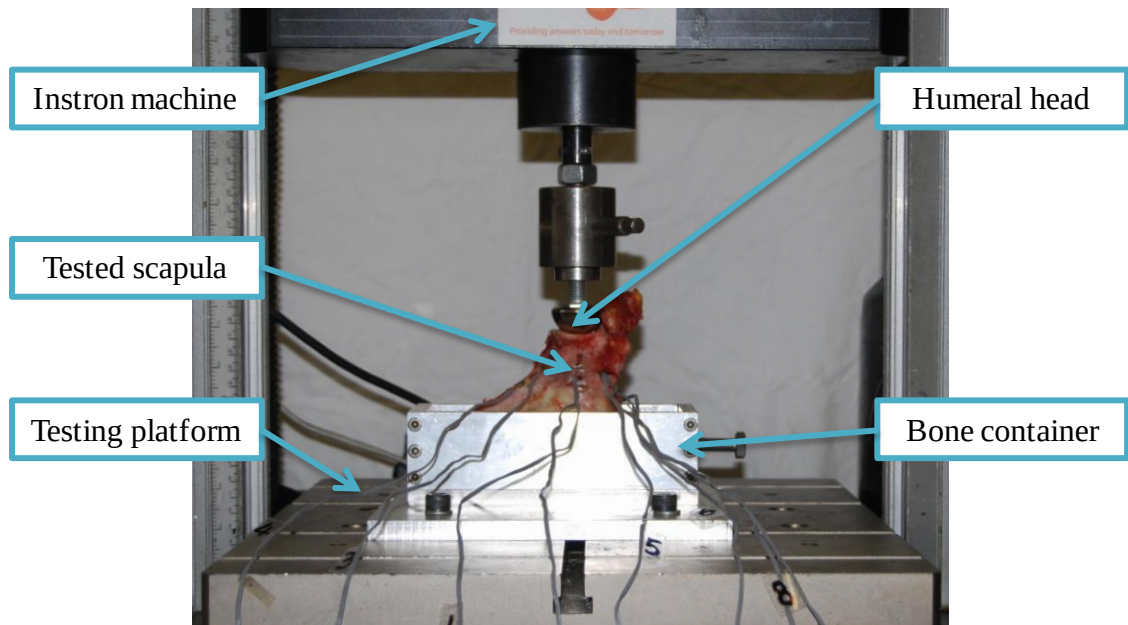


Figure 5-3 Experimental setup for the strain study of the intact scapula.

5.2.2.2 Finite Element Modelling of Intact Scapulae

Three-dimensional FE models of the tested cadaveric scapulae were built based on their patient-specific CT images. Except for the boundary condition, setup of the FE modelling of an intact scapula in this study was similar to the description in Chapter 4.

Boundary conditions in the FEM were set up to be the same to that in the in-vitro testing. Middle point of the glenoid articular surface was set as the origin. The glenoid center line was set as Y axis. The line which was parallel to the scapular plane and passed through the origin was set as Z axis. X axis was perpendicular to both Y and Z axes. The Cobalt-Chromium humeral head used in the experiment was about ten times stiffer than the cortical bone, thus deformation of the humeral head was negligible in comparison with bone. Therefore, the humeral head was created as a 36 mm-diameter rigid half sphere in the FEM, and contacted with the glenoid surface in the origin point. To be consistent with the fixation position in the in-vitro test, approximately one-third of the scapula from the medial border side was fixed. Compressive force parallel to the

glenoid centre line was applied on the half sphere (Figure 5-4). Four force cases which were 300 N to 750 N with an increment of 150 N were analyzed, respectively.

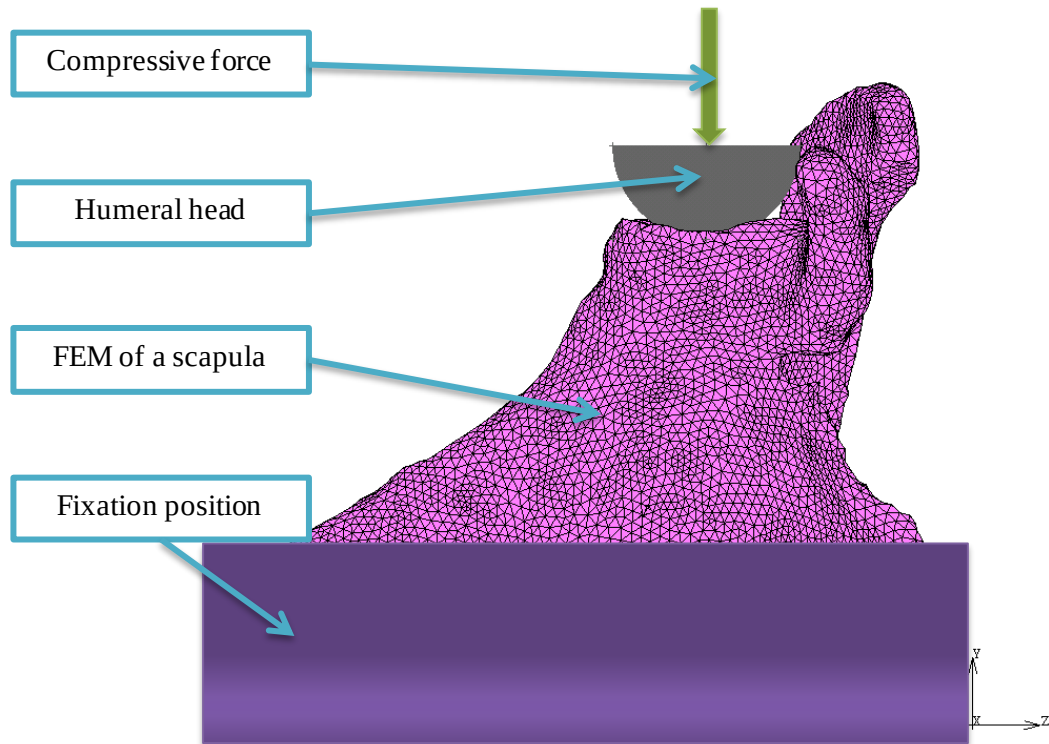


Figure 5-4 3D FE model of a scapula.

5.2.3 Results

Experimental strain measurements at two levels which were 10 mm and 25 mm below the glenoid fossa are illustrated in Figure 5-5 and Figure 5-6. Statistically significant difference in the magnitudes of measured individual surface strains in the same positions and under the same loading conditions were observed (t-test, $p < 0.05$). Another observation from Figure 5-5 and Figure 5-6 is that there was a global upward trend in the surface strain measurements around the glenoid with the increase of external compressive forces. At the level of 10 mm below the glenoid fossa, the mean posterior strains of the six tested scapulae were about 3 to 5 times of the mean anterior values for various compressive loading conditions. These strain differences reduced to approximately 1 time at the level of 25 mm below the glenoid fossa.

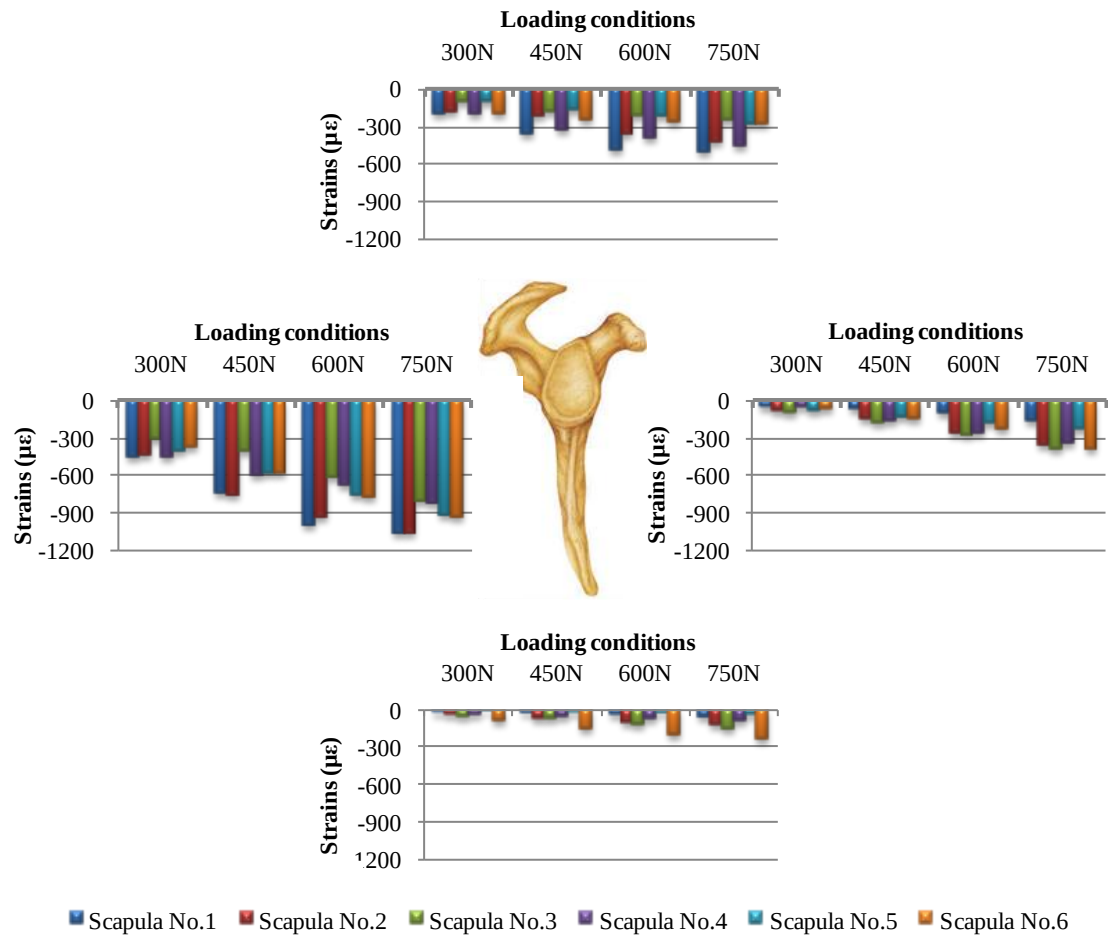


Figure 5-5 Strains around the glenoid at a rough 10 mm level under the glenoid fossa in different compressive loading tests.

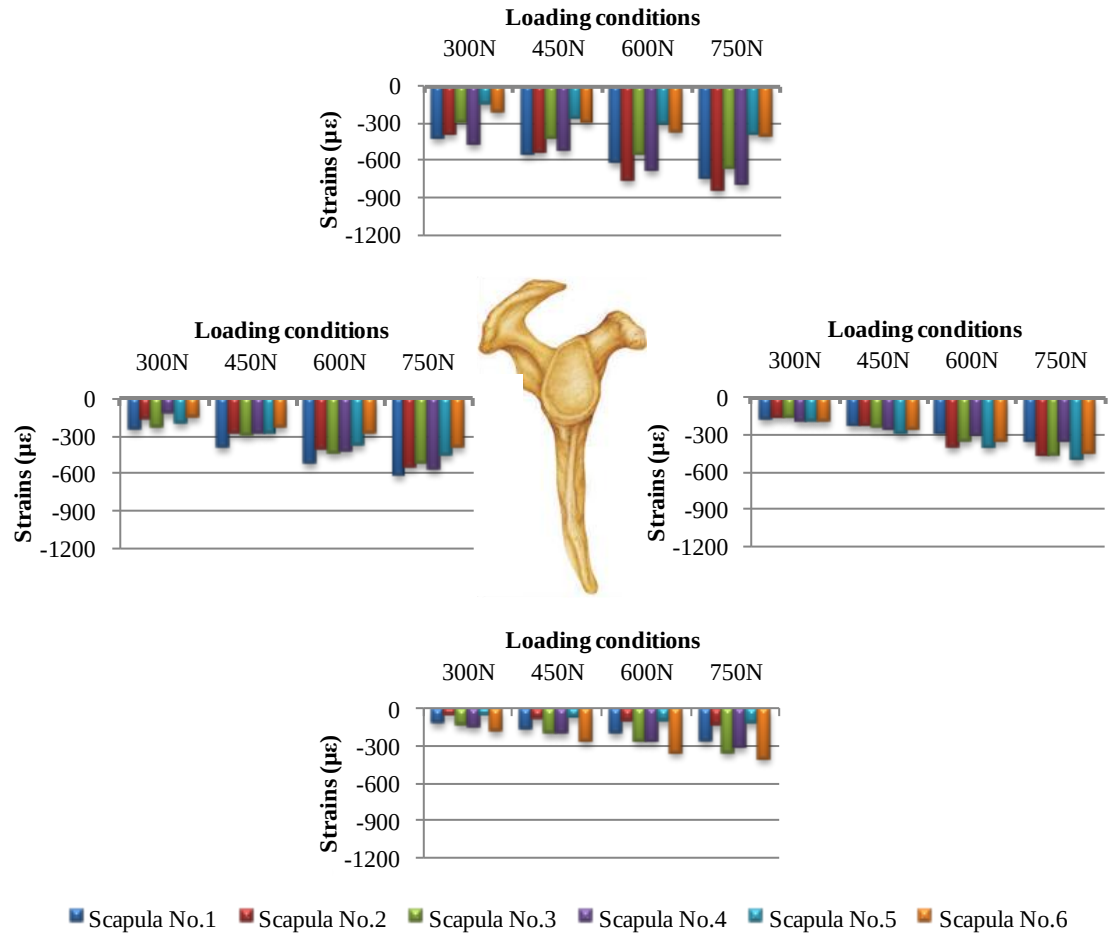


Figure 5-6 Strains around the glenoid at a rough 25 mm level under the glenoid fossa in different compressive loading tests.

Comparisons between experimental measurements of cadaveric scapulae and the corresponding patient-specific FE results under the same loading conditions are shown in Figure 5-7. An ideal line (the dark line), for which experimental and FE magnitudes were the same, was also included in each figure. The results indicated that strains predicted from the FEMs were consistent to the measured values. To quantitatively assess the reliability of the FE results, individual regression studies including slope of linear regression, correlation coefficient of the raw data to the linear regression, the maximal and mean percentage errors of FE predictions (Equation 5-2) were performed (Table 5-2). As strain box was zeroed before each loading condition and the value of strain in the FEM was zero under zero loads, an intercept of zero was used for the creation of linear regression. It is observed that the slope of linear regression with zero intercept for all the analyzed FE models ranged from 0.95 to 1.02, and, thereby, displayed a good correlation between the experimental and FE strains. Except for the

unsatisfied percentage error of scapula No. 5, all other individual maximal percentage deviations of FE strains from the experimental data were within 16%. In the No. 5 scapula model, the big maximal percentage error (65.83%) was caused by the small magnitudes of experimental and FE strains (-11.1 $\mu\epsilon$ to -17.3 $\mu\epsilon$) in the lateral inferior position under a 300 N compressive load. Except for this big value, the maximal percentage error in scapula No. 5 was 14.84%

Table 5-2 Individual regression studies of experimental and FE strains.

Scapula No.	Slope of linear regression, a (Y=aX)	Correlation coefficient ®	Maximal Percentage error (%)	Mean percentage error (%)
1	0.92	0.99	14.06	8.99
2	0.95	0.99	15.24	8.01
3	0.95	0.99	15.64	7.73
4	1.02	0.99	14.34	7.00
5	0.95	0.99	65.83*/14.84	10.40
6	0.99	0.99	13.46	6.43

* the percentage error induced by small experimental and FE strains

$$\text{Percentage error} = \frac{|FE \text{ strain} - \text{experimental strain}|}{\text{experimental strain}} \times 100\% \quad \text{Equation 5-2}$$

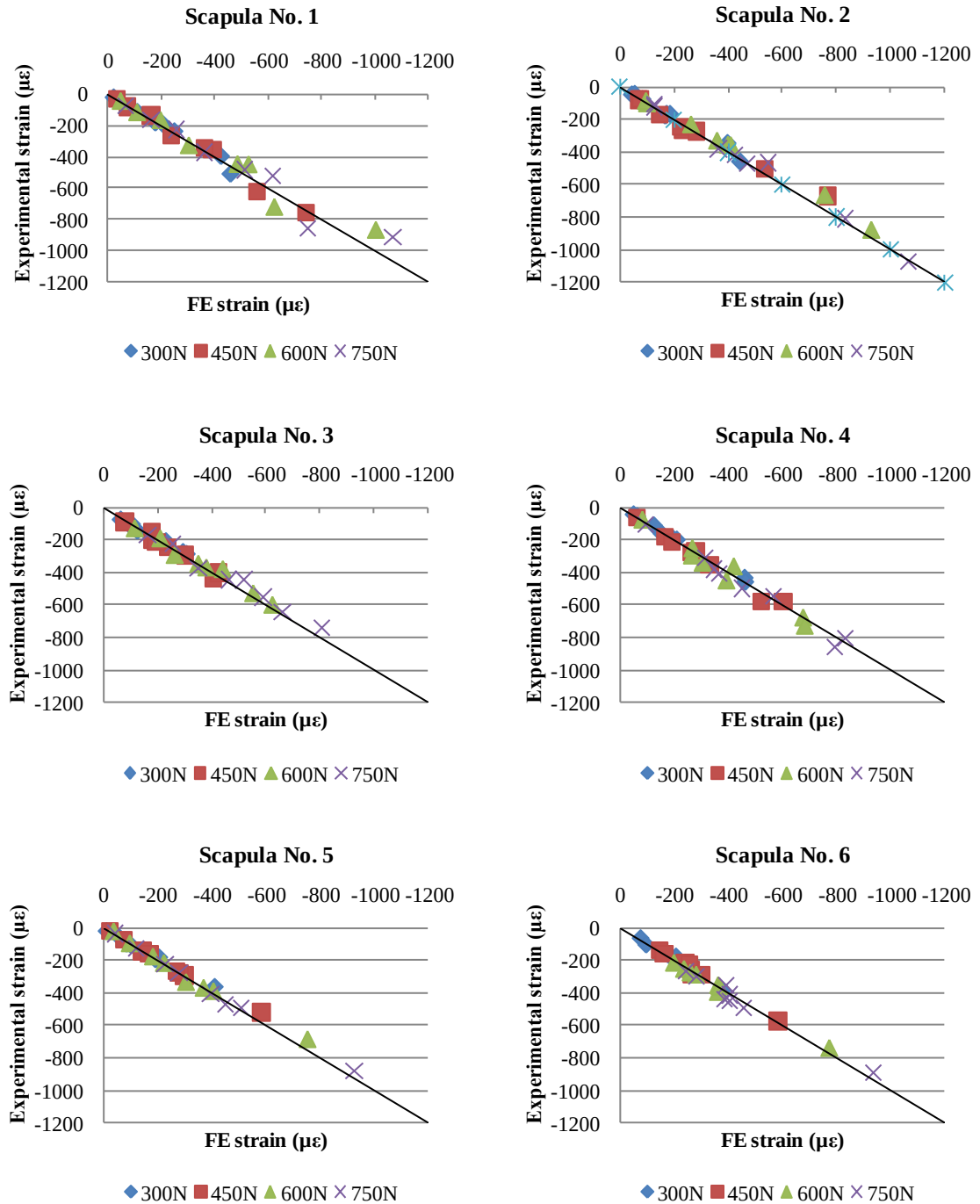


Figure 5-7 Individual correlation between the experimental and FE strains in four load cases.

5.2.4 Discussion

As strains would be used to evaluate influences of scapular notching and bone remodelling on the glenoid component loosening of RSA (introduced in Chapter 6 and 7), reliability of the predicted strains in the FEM of scapula was assessed with in-vitro

testing results in this study. Six cadaveric scapulae were tested in four loading conditions, meanwhile, the same bone and boundary conditions were FE modelled by using the patient-specific CT data of the bone. Strains in eight positions around the glenoid were measured and compared with the numerical predicted values on the corresponding positions. Experimental measurements and the comparison between measured and predicted strains were used to understand the strain distribution around the glenoid and to assess the accuracy of FE modelling.

Based on the results in Figure 5-5, it is found that the strain in the lateral posterior portion of the scapula was higher than that in the lateral anterior portion in a particular loading condition. This result is possibly caused by the irregular and complex geometry of scapula. As seen in Figure 5-2, the lateral posterior surface is very concave, and the degree of curvature of lateral posterior region in a scapula is much bigger than that on the other side. Therefore, the bone in this region deformed significantly and experienced high strains in the experiments. In the medial position shown in Figure 5-6, differences between the anterior and posterior surface strains reduced. It is possibly due to the decreased differences in degrees of curvature between the anterior and posterior regions. As superior and inferior strain gauges were bonded in different directions, no comparisons were performed. By comparing the results shown in Figure 5-5 and Figure 5-6, it can be found that anterior strains increased from lateral to medial regions while posterior strains decreased from lateral to medial portions. This lateral-medial changes of anterior and posterior surface strains indicated that the scapula fixation method didn't constrain the behaviour of glenoid. The negative strain values in all the tested positions showed that no bending occurred in the experiments. In anatomy, scapula is covered by muscles, so magnitudes of clinic strains may be different from our experimental measurements. However, measuring strains on a live bone is very difficult. Additionally, this study is aimed to validate reliability of FE modelling. As long as in-vitro testing and the finite element model were setup in the same conditions, in-vitro testing results could be used to validate the FEM.

Based on the results in Figure 5-5 and Figure 5-6, t-test ($p < 0.05$) was used to statistically analyse changes of individual strains on the same positions of scapula, and the results showed that strain values were very subject-dependent. There are a few possible reasons. Firstly, variations in scapular morphology (Churchill et al., 2001;

Torrens, 2009) could lead to differences of individual strains. The scapula position in the experiment was set up by using the glenoid center line and the scapula plane, therefore variations of glenoid surface, retroversion, size and inclination (Churchill et al., 2001) could lead to different strain positions on the surface of glenoids. Moreover, to avoid high degree of curvature of local surface, slight changes of strain locations were performed. These varied locations may cause the changes of site-dependent individual strains in the test. The variation of bone quality of the tested scapulae may be another reason leading to different individual strains. Bone quality varies with age, healthy condition and sex (Weaver and Chalmers, 1966; Keaveny and Hayes, 1993; Anglin et al., 1999; Lim et al., 2002; Kalouche et al., 2008). As the tested scapulae included male and female donators' bones and aged from 66 to 73, qualities of these bones may be varied. Given the deviation of individual strains, individualised evaluation is needed and very important for the study of reliability of FE modelling.

The correlation of measured and predicted strains is illustrated in Figure 5-7. The results showed that there was a good agreement between the experimental measurements and FE strain values, as most data in the graph closed to the ideal line (slope = 1). In the regression studies (Table 5-2) of experimental-FE strain correlation, the slope of linear regression in Figure 5-7 ranged from 0.95 to 1.02 and the maximal percentage errors was between 13.46% and 15.64%. The deviation shown in these results could be caused by a few reasons. One of them is possibly due to the isotropic assumption of material properties of bone in the FE modelling. In this FE study, the Young's modulus obtained from the average of CT values within one element volume was assigned to the element on the corresponding location in the FE model. It was reported that correct heterogeneity modelling played more crucial role than the anisotropic property assignment, as the elastic modulus of bone varies with density to a power factor of 2 to 3, whereas the maximum change in modulus along different axes is only within 5 times (Ciarelli et al., 1991; Frich et al., 1997). However, it is possible that the neglected different axial Young's modulus leads to the small mismatch between the experimental and FE strains. Another possible reason for the deviation of FE values is due to the accuracy of tools in the experiments. Exact distances of the strain gauges relative to the glenoid fossa were measured with a ruler, of which the accuracy is 1 mm and these distances were used to find the corresponding positions of the strains in the FE model. Due to the site-dependent characteristics of strains, uncertainties of ruler may lead to

differences between the experimental and FE values. Apart from the accuracy of ruler, the uncertainty of the strain recorder used in the experiments was $1\ \mu\epsilon$ and may influence the percentage errors when magnitudes of strains were only a few dozens.

5.2.5 Conclusion of the Validation of FE Strains

In this study, in-vitro testing was used to validate the reliability of FE predicted strains, which would be used to analyse influences of scapular notching and bone remodelling on the glenoid component loosening in RSA in Chapter 6 and 7. To understand the strain distribution and to evaluate the quality of FEM, eight strains around the bone on two depths, which were roughly half the length of the peg and periphery screws, respectively, were tested. Patient-specific CT data of the six cadaveric scapulae were used to assign the material properties on an element-by-element basis. Given the complicate evaluation, the relatively high correlation coefficient of the linear regression, and low mean percentage errors of the FE values with reference to the experimental measurements indicated that the FE modelling of scapula could predict the realistic changes.

5.3 Validation of the Signal of Osseointegration (Micromotions)

5.3.1 Introduction

As described in Chapter 3, interface micromotion was used as a tool to assess the likelihood of osseointegration and to predict the long-term fixation of shoulder prosthesis. Therefore, reliability of predicted FE micromotions is of significance. This study is aimed to validate the FE predicted micromotions with in-vitro testing results.

The first experimental assessment of micromotions in RSA was performed by Harman et al. (2005). In their study, the implant was secured on the polyurethane foam, a cyclic shear load was applied on the glenosphere surface, and the relative motion of glenoid component to the adjacent point on the polyurethane foam was measured with linear variable differential transducers (LVDTs). Two limitations of their study are that polyurethane foam was used to replace the real bone and non-physiological forces for

RSA were used. Micromotions were, later, measured by using other techniques such as high-resolution video camera (Codsì and Iannotti, 2008), to investigate effects of screw positions on the initial stability. However, artificial scapulae and non-physiological forces for rTSA were used in this study too.

In this study, six cadaveric scapulae and glenohumeral (GH) joint forces for RSA in physiological daily activities (Terrier et al., 2008) were performed in the experiments. Relative motion measurements in the in-vitro testing were compared with the FE predictions obtained from the FEM of the same scapulae upon the same loading conditions. The FE-experimental variations were used to verify the accuracy of FE predictions.

5.3.2 Materials and Methods

5.3.2.1 In-vitro Study

When the strain test was finished, the cadaveric scapulae were moved onto an experimental bench and then inserted with a Delta CTA RSA (Depuy) by an experienced professional orthopaedic shoulder surgeon, Dr Gregory, Thomas (Hospital Georges Pompidou, APHP, Paris, France) according to the Delta reverse shoulder system surgical technique (Depuy).

5.3.2.1.1 *Procedure of glenoid component implantation*

The glenoid preparation was started by choosing a position for the metaglene. According to the clinical and biomechanical studies, a good position of the implant can significantly affect the artificial joint performance and functionality (Nyffeler et al., 2005; Simovitch, 2007). The standard position of the metaglene is slightly posterior and inferior to the intersection of the glenoid axes as shown in Figure 5-8 (a) and this position was used in current study. Other prosthesis positions proposed by Nyffeler et al. (2005) will be analysed in Chapter 8.

Once the position of the implant was marked, a smooth platform devoid of cartilage was created to fix the metaglen and the glenosphere. To cut less bone, the handle of the

resurfacing reamer was always perpendicular to the glenoid surface during the whole process of reaming (Figure 5-8 (b)).

The metaglene was fixed on the bone with four peripheral screws. The threaded inferior screw was fixed in the direction following the inferior scapula border. Due to the fixed angles between superior and inferior screws (Figure 5-8 (c)), the position of threaded superior screw was fixed. As both posterior and anterior screw positions allow a variation of ± 20 degrees, these two spherical screws were set the most appropriate angles to ensure them to be hidden in the reliable bone stock (Figure 5-8 (d)).

Finally the glenosphere was secured on the metaglene (Figure 5-8 (e)). There were two available sizes of glenosphere: 36 mm and 42 mm diameters. Due to the small and medium sizes of cadaveric glenoids used in the experiments, 36 mm diameter glenosphere was selected for all the scapulae.

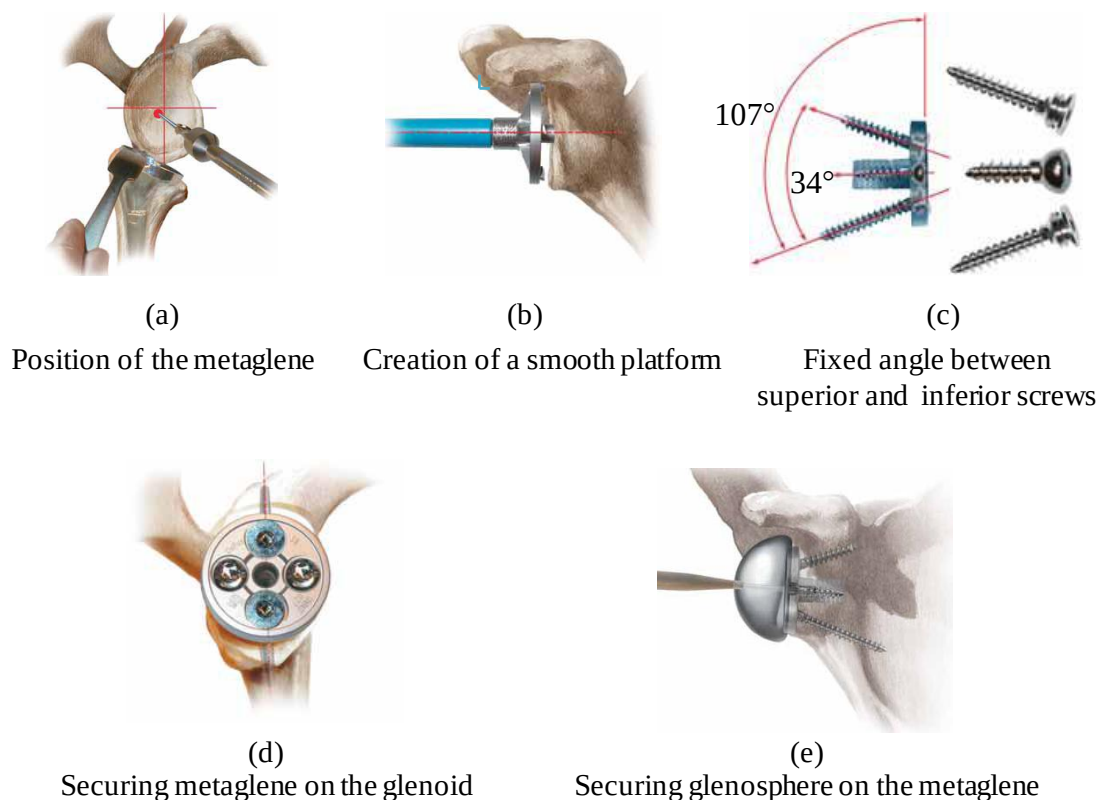


Figure 5-8 Surgical procedure of glenoid component in Delta CTA RSA (adapted from Delta reverse shoulder system surgical technique, Depuy).

One of the cadaveric scapulae associated with the Delta CTA RSA was shown in Figure 5-9. The scapular position and fixation method were the same with those in the in-vitro testing for the FE strain validation explained above. As the handle of resurfacing reamer was always perpendicular to the glenoid surface during the whole process of reaming, the glenosphere faced up after the prosthesis implantation (shown by the purple arrow in Figure 5-9).

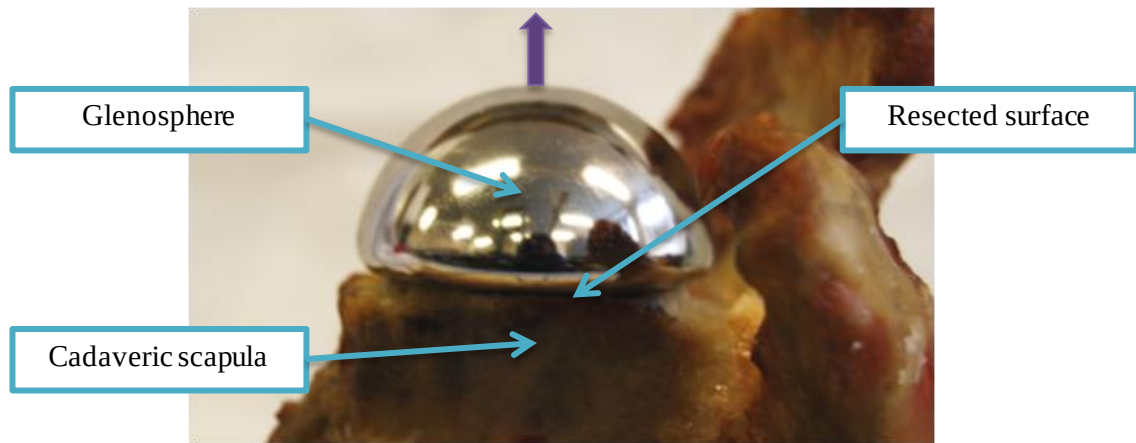


Figure 5-9 Picture of an implanted scapula.

5.3.2.1.2 *Setup of bone-prosthesis interface micromotion test*

As it is difficult to measure micromotions at the bone-prosthesis interface, the relative movement between the glenosphere rim and a point on the supporting bone were measured by using LVDT (Digital Probe DP/2/S, Solartron Metrology, UK) (Figure 5-10). The LVDT was carried by a holder, which was fixed on an external rob. This rob was firmly bonded on the nearest edge of the glenosphere. The probe of LVDT touched the adjacent bone and was able to be extended and compressed. The extension and compression of the probe corresponded to the relative single-direction movement between the point where the probe touched on the bone and the point where the external rob was fixed on the glenosphere (Figure 5-10). Four locations namely anterior, posterior, superior and inferior were measured. As the humeral head may impinge the external rob supporting the inferior LVDT in the load case of 30° and 60° abductions, inferior relative motions in this two load cases were not recorded.

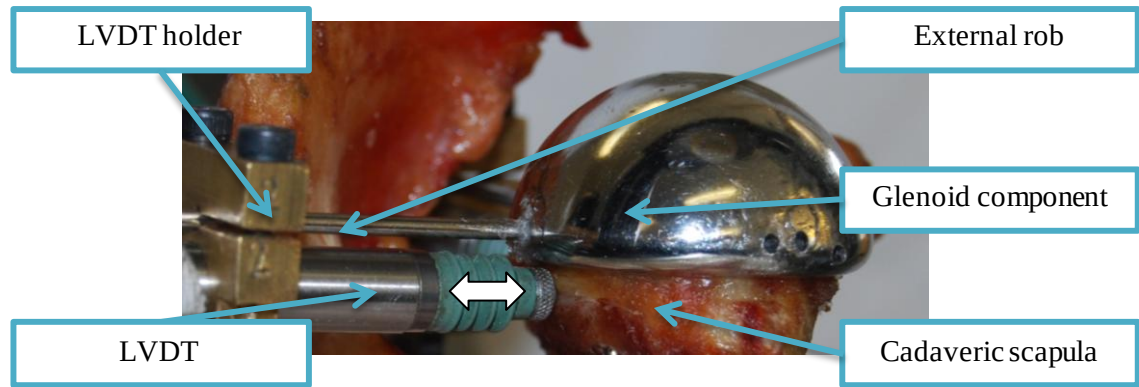


Figure 5-10 LVDT setup on the glenosphere rim and the adjacent supporting bone.

5.3.2.1.3 *Test method*

The cadaveric scapula merged with Delta CTA RSA was moved back to the Instron machine and orientated such that the superior-inferior direction of the scapula was parallel to the direction of horizontal load applied by a pneumatic cylinder and in line with the matching humeral cup (Depuy). The bone container was able to move in the superior-inferior direction of the scapula. The humeral cup which was fixed on a holder could rotate freely around the anterior-posterior direction axis. Three load cases, which were 30°, 60° and 90° arm abductions were simulated. If assuming that one body weight is equal to 750 N, glenohumeral joint contact force in the RSA could be obtained according to Terrier's study (2008) (Figure 5-11).

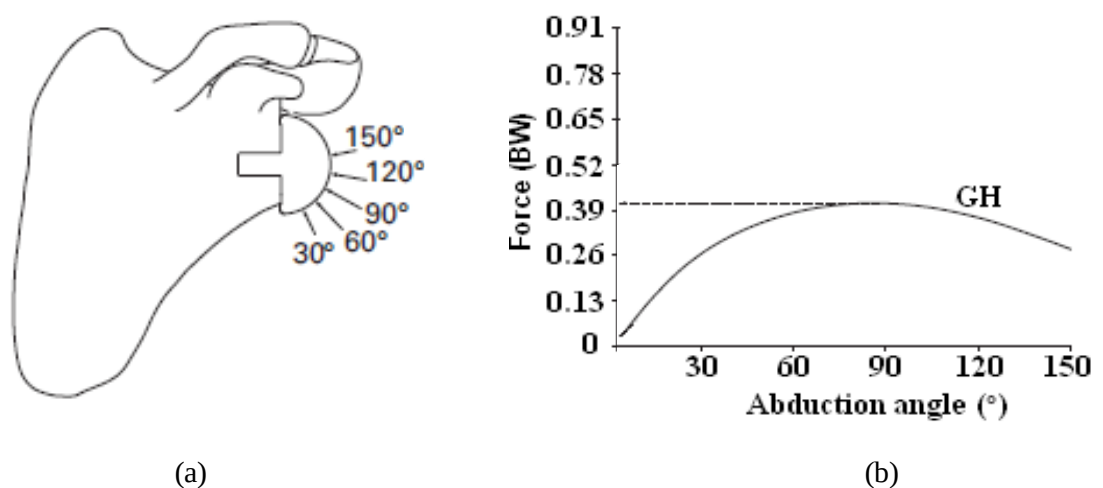


Figure 5-11 (a) Direction of glenohumeral joint force in RSA at particular abduction angles, (b) magnitude of glenohumeral joint force in RSA (adapted from Terrier et al. 2008).

For each loading condition, the GH force in the horizontal and vertical components were first calculated. The calculated vertical distance between the loading point and the resected surface was then set as the height of the humeral cup relative to the resected surface. After that, the horizontal component force was applied by a pneumatic cylinder. This force pushed the container and allowed it to move in the superior-inferior direction of the scapula. As this direction was in line with the humeral cup, the prosthesis is able to push the humeral cup to rotate until the whole system (the cup and the glenosphere) reached a balance. The vertical force measured by the load cell in the Instron machine was then checked with the calculated force. Slight operation to make the force reading from the Instron machine was consistent with the calculated vertical force was performed. When the entire system was balanced, micromotions were recorded. To reduce effects of viscoelastic properties of bone, tested scapulae were allowed to restore for approximately five minutes before next loading began. In each load case, the test was repeated three times. The average was used to represent the micromotion under that loading condition. In the study of force-dependent characteristics of bone-prosthesis interface micromotions, values were analysed with t-test. A p value of less than 0.05 was considered to be statistically significant.

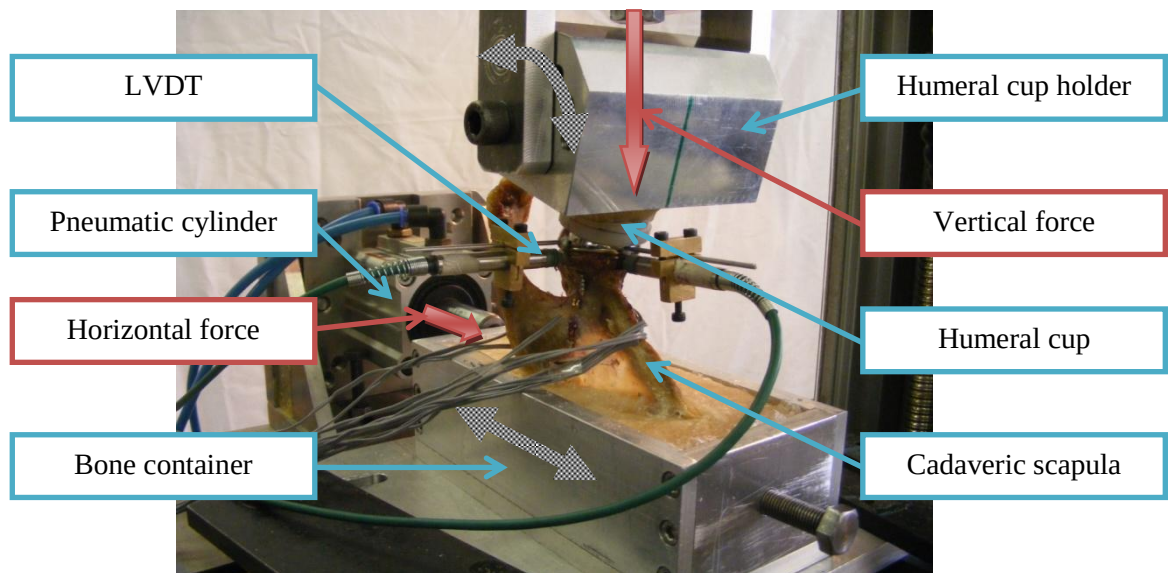


Figure 5-12 Experimental setup for the micromotion test.

5.3.2.2 FE Study

3D FEMs of the tested cadaveric scapulae were created from patient-specific CT images of the bone used in the in-vitro micromotion testing. Delta CTA RSA was also modelled

and virtually inserted into each scapula model, in terms of the implantation procedure introduced in Chapter 5.3.2.1.1. Except for the force conditions and coordinate system, FE modelling of an implanted scapula introduced in Chapter 4 was used in this study. Glenohumeral joint forces and the coordinate system used in the experiments were employed in this FE modelling. To be consistent with the positions where the micromotions were measured in the experiments, the fixation point of the external rob, which held the LVDT on the glenosphere, and the point where the LVDT touched on the bone were marked in the FEM. Bone-prosthesis relative moments between these two points in the FEM represent the micromotion in that region. Due to the single-direction characteristics of the LVDT, the anteroposterior FE micromotions were compared with the experimental measurements in the anterior and posterior regions. The FE micromotions in the superoinferior direction were compared with the experimental values in the superior and inferior regions.

5.3.3 Result

Experimental micromotions between the rim of glenoid component and the supporting bone were measured and are shown in Figure 5-13. The comparison between them and predicted values in the corresponding locations in the FEM (Figure 5-14) are illustrated in Figure 5-15. A, P, I, S in the figures represent the four locations around the glenoid, namely anterior, posterior, inferior and superior.

Figure 5-13 shows that the relative movements ($0.4\ \mu\text{m}$ to $9.8\ \mu\text{m}$) during the arm elevations were much smaller than the required threshold ($50\ \mu\text{m}$) (Pilliar et al., 1986) for bone ingrowth and demonstrated a stable initial stability of the glenoid prosthesis in the tested regions. Another observation was that regional micromotions under the same loading conditions were statistically different around the glenoid. The result also revealed that experimental bone-prosthesis relative motions were both load and subject dependent.

Figure 5-14 illustrates FE predicted micromotions around the glenoid in various loading conditions. The results showed a similar conclusion to that in the experiments. The predicted bone-prosthesis micromotions were very small compared with the required threshold ($50\ \mu\text{m}$) of bone ingrowth proposed by Pilliar et al. (1986). In comparison

with the mean experimental micromotions, variations of the averaged FE predictions (1.8 μm to 4.0 μm) (Figure 5-15) were comparatively lower. Another observations is that the experimental and FE results demonstrated the same tendency of micromotion variation with the changes of loading conditions.

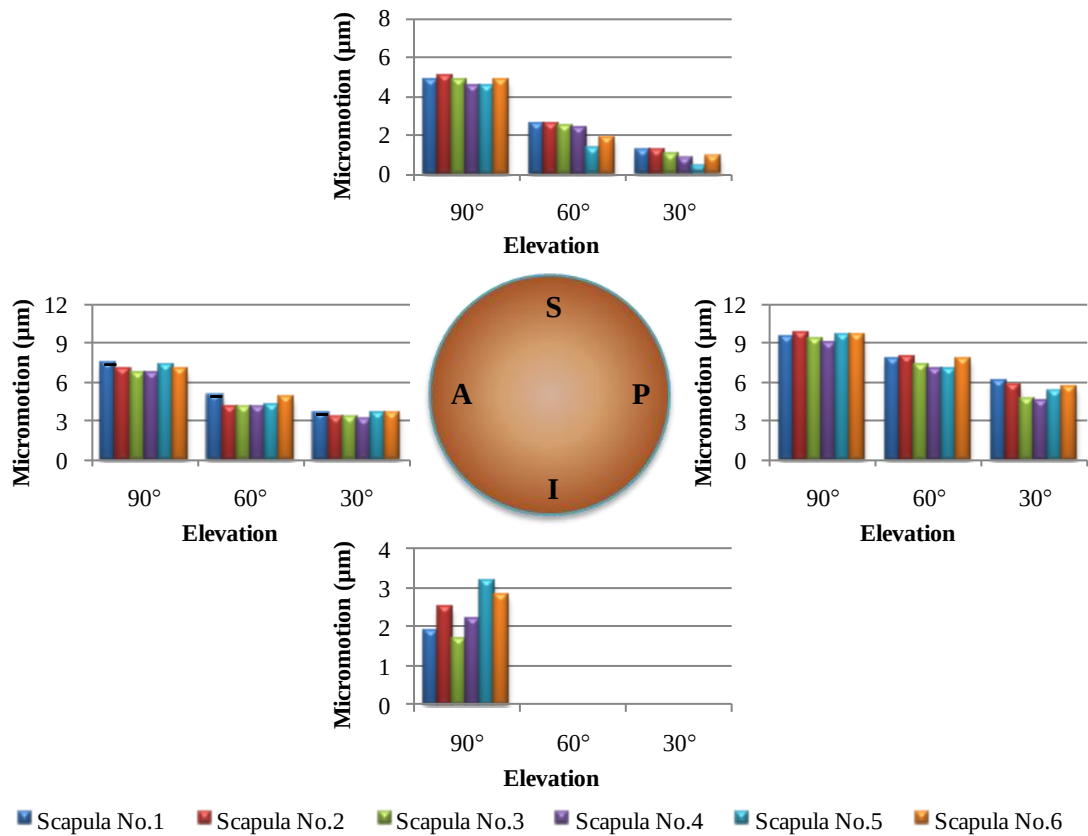


Figure 5-13 Experimental micromotions around the glenoid in various loading conditions.

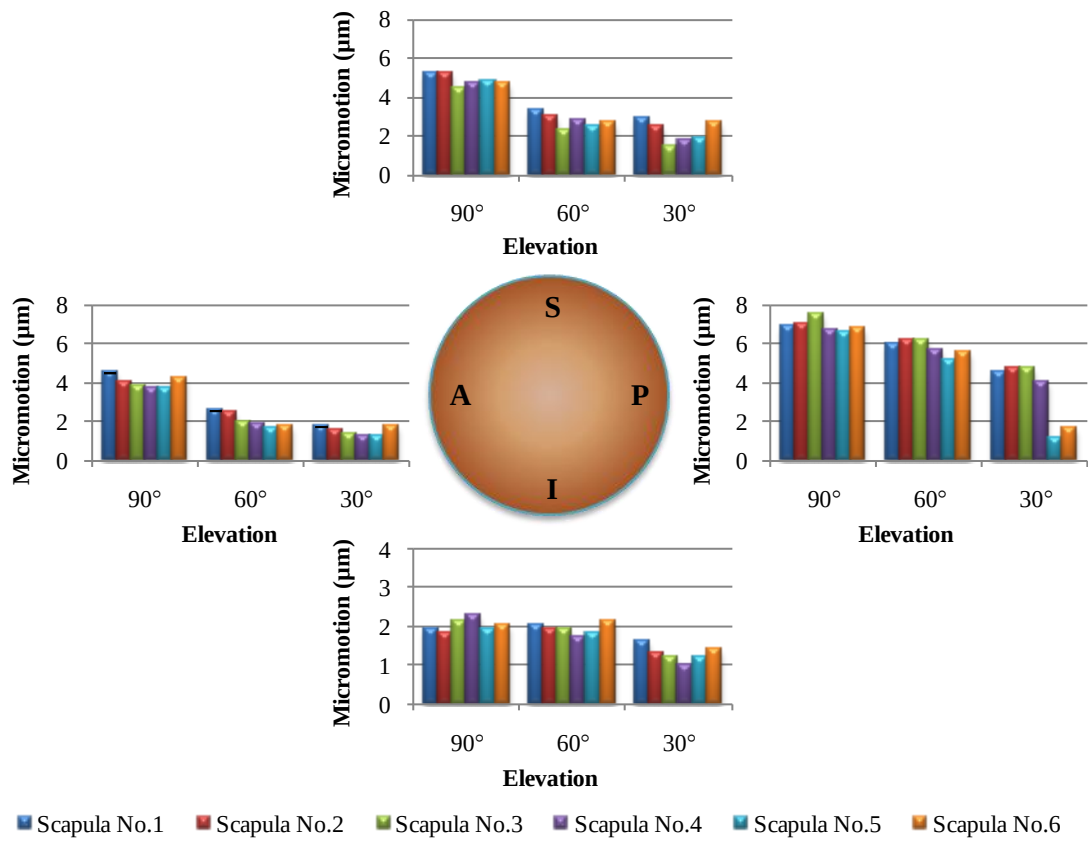


Figure 5-14 Individual FE micromotions around the glenoid in various loading conditions.

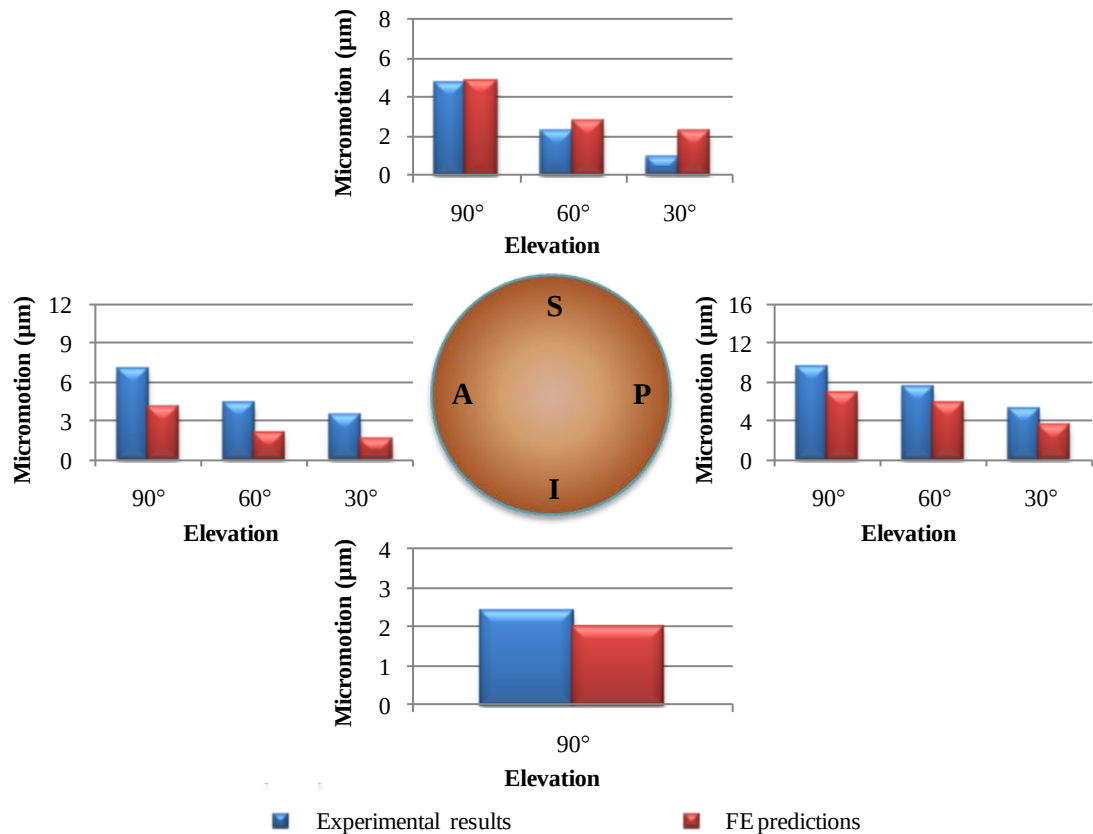


Figure 5-15 Comparison between the mean FE and experimental micromotions for different load cases.

5.3.4 Discussion

This study is aimed to assess the reliability of micromotions predicted from the FE modelling with in-vitro testing results. Micromotions at the edge of the glenoid component in the Delta RSA under three load conditions were measured in the experiments and the results were used to validate the micromotion predictions.

Micromotions obtained from both the in-vitro testing and the FE modelling showed much small values in comparison with the required threshold of bone ingrowth (50 µm) (Pilliar et al., 1986) and demonstrated a good initial stability of the glenoid prosthesis during the arm elevation in the tested regions. In comparison with the experimental results, the FE micromotions were a little smaller with a difference of 1.8 µm to 4.0 µm. The differences between the experimental and predicted results could be explained next. Firstly, In the FE modelling, all the four peripheral screws were ideally assumed to be firmly fixed. In reality, only the superior and inferior screws are locked screws in the

Delta RSA. The anterior and posterior screws are non-locked spherical head screws, thus fixation situation of these screws in the experiments were unknown. They were possibly loosened in the spongy cancellous bone under loading and thus led to high micromotions in the experiments. To address this problem, new FEMs of the same implanted scapulae were created. In these FEMs, the anterior and posterior screws were set loosening. It was found that the averaged micromotions in the four positions around the glenoid increased 3 μm to 6 μm in the new FEMs. Moreover, the bigger FE-experimental micromotion differences in the anterior and posterior regions than those in the superior and inferior glenoids are an evidence of the influences of the non-locked screws on the micromotions. Another possible reason for the differences between the predicted and measured micromotions is instrument uncertainties. The whole measuring system of the in-vitro testing included the load cell in the Instron machine, pneumatic cylinder, and LVDT. As instrument uncertainties propagated during the test and the value of measured micromotions were very small with the maximum of approximately 9.8 μm , the precision of these instruments could influence the micromotion measurements, particularly when the experimental value was only 0.9 μm in the superior region at the 30° of arm elevation. Finally, it should be noticed that the positions of glenoid component could be not the exactly same between in the in-vitro test and the FEM, as it is impossible to make two repeated operations exactly the same. In comparison with experimental measurements proposed by Harman et al (2005), underestimation of FE predicted bone-prosthesis interface micromotions in the same loading conditions was confirmed by Hopkins et al. (2008).

However, the underestimation of the FE micromotions with relative to the experimental results is limited, with a maximal value of 4 μm . This value is much smaller than the threshold for bone ingrowth (50 μm) (Pilliar et al., 1986). Given the same order of magnitudes between the current predicted FE bone-prosthesis micromotions and the experimental values, the FE predicted micromotions were reliable.

Although there were consistent FE-experimental micromotions, the limitation in this study should be mentioned. Bone-prosthesis interface micromotion is defined as the relative motion of adjacent nodes between the back of metaglene and the glenoid resected surface. As it is difficult to measure the relative movement at the bone-prosthesis interface with LVDT in the experiments, the relative motion between the

fixation point of external rod and the position where the probe of LVDT touched on the bone, as seen in Figure 5-10, were measured and used to be compared with the FE results. Thereby, the micromotions used in this study include partial bone deformation. However, the primary purpose of this study is to validate the predicted relative motions between the glenoid prosthesis and the bone. As long as the two points which were used to measure the relative motions between the bone and the prosthesis in the in-vitro testing are consistent with those in the FE modelling, the experimental micromotions can be used for the validation.

5.3.5 Conclusion of the Validation of FE Micromotions

In this study, in-vitro testing with six cadaveric scapulae was used to validate the reliability of FE predicted micromotions. There was a good agreement between the FE predicted micromotions and the experimental results, with the variations of 1.8 μm to 4.0 μm . Both the in-vitro testing and the FE modelling showed small micromotions in comparison with the required threshold of bone ingrowth (50 μm) (Pilliar et al., 1986) and demonstrated a good initial stability of the glenoid prosthesis during the arm elevation in the tested regions.

CHAPTER 6

EFFECTS OF SCAPULAR NOTCHING ON THE GLENOID PROSTHESIS FIXATION

6.1 Introduction

Scapular notching is a frequently reported complication in the study of RSA (Sirveaux et al., 2004; Boileau et al., 2005; Lévine et al., 2008; Lévine et al., 2010). It is believed to be initially induced by the direct impingement of the rim of polyethylene humeral cup against the scapular neck. Scapular notching occurs at a high rate (approximately 50% to 96%) after the RSA operations (Sirveaux et al., 2004; Boileau et al., 2005; Lévine et al., 2008) and the relationship between the incidence and the fixation of RSA has been a big concern (Sirveaux et al., 2004; Boileau et al., 2005). In a medial-term follow-up clinical and radiographic study of the Delta RSA, all the patients

with loosened glenoid component had severe scapular notching (Nerot grade 3 or 4) (Grassi et al., 2009). A biomechanical evaluation of effects of this frequently observed postoperative phenomenon on the fixation of RSA at long-term follow up is the aim of this chapter.

As introduced in Chapter 3, apart from the cementless fixation, Delta RSA is also secured with four peripheral screws, especially the two locked screws located in the superior and inferior regions. Thus the breakage of inferior screw shown in recent clinical studies (Sirveaux et al., 2004; Grassi et al., 2009) could be a direct factor for the failure of glenoid prosthesis. As the screw fracture is always accompanied with erosions of scapular neck (Grassi et al., 2009), relationship between these two parts was concerned in this chapter. Additionally, the possible increase of bone stresses at the bone-screw interface during the process of screw breaking could lead to bone fracture and thus result in glenoid loosening. Thereby, the change of bone stresses at the bone-screw interface was used as another factor to assess effects of scapular notching on the fixation of glenoid prosthesis in this study. Due to the cementless fixation, effects of scapular notching on the initial stability of glenoid prosthesis was also concerned in this chapter.

This chapter is mainly composed of two parts: (1) validation of predicted notch-induced variations of strains and micromotions; (2) analyses of effects of scapular notching on the fixation of glenoid prosthesis. As the most frequently reported erosion occur in the inferior glenoid neck (Boileau et al., 2006; John et al., 2009; Levy and Blum, 2009), only inferior scapular notching was focused in this chapter.

6.2 Validation of Predicted Variations of Strains and Micromotions due to Scapular Notching

6.2.1 Introduction

The purpose of this study is to validate the predicted notch-induced variations of strains and micromotions with in-vitro testing results. This study includes two aspects: in-vitro testing and FE modelling.

6.2.2 Materials and Methods

6.2.2.1 In-vitro Testing

Three cadaveric scapulae (mean age 69 years) were used. Each of them was merged with a Delta CTA RSA. The implantation procedure and preparation of tested scapulae in this study were the same to those described in Chapter 5. The in-vitro testing involved two stages: before and after the creation of inferior scapular notch. Before creating a notch, the process of measuring strains and micromotions in this study was the same to that described in Chapter 5. When all the measurements were finished, a hand-made artificial notch classified as grade 4 in the Nerot system was created in the scapular neck according to the study of Simovitch et al. (2007). In their article, the outline of a severe inferior scapular notch started from a point locating approximately 10 mm medial to the junction of the glenosphere and the most inferior and lateral part of glenoid rim. Photographs of an implanted cadaveric scapula before and after the creation of a notch were shown in Figure 6-1. Strain gauges used in the test without a notch were still used in the test with a big notch, in order to ensure the measured positions before and after the erosion of scapular neck were consistent. However, if the strain gauge was broken during the process of creating the notch, a new gauge would be attached on the same position. Three loading conditions which were arm elevation in the scapular plane at 30°, 60° and 90° (detailed descriptions of the forces were introduced in Chapter 5.3.2.1.3) were employed in both stages (with and without erosion). Figure 6-2 illustrates the experimental setup of an implanted scapula with a severe notch in the scapular neck. Strains and micromotions in each loading condition were recorded.

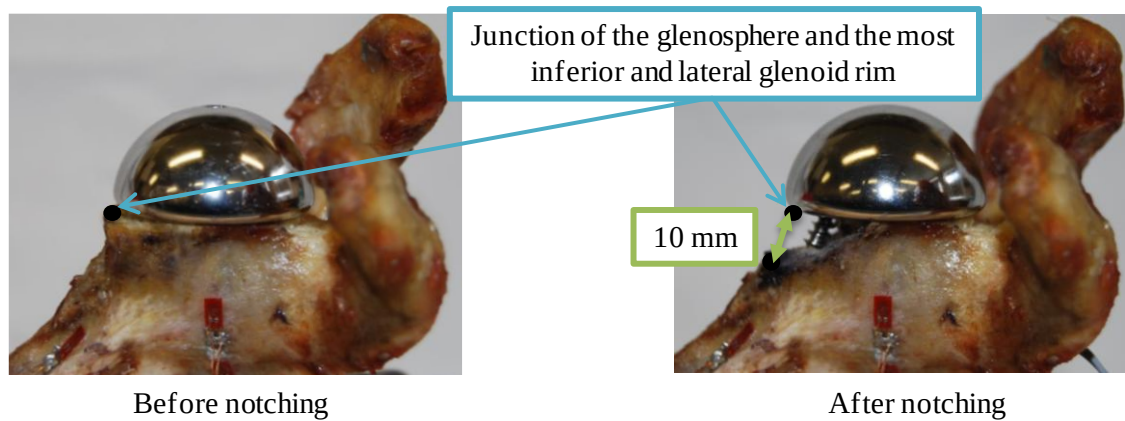


Figure 6-1 Implanted scapulae before and after the erosion of scapular neck.

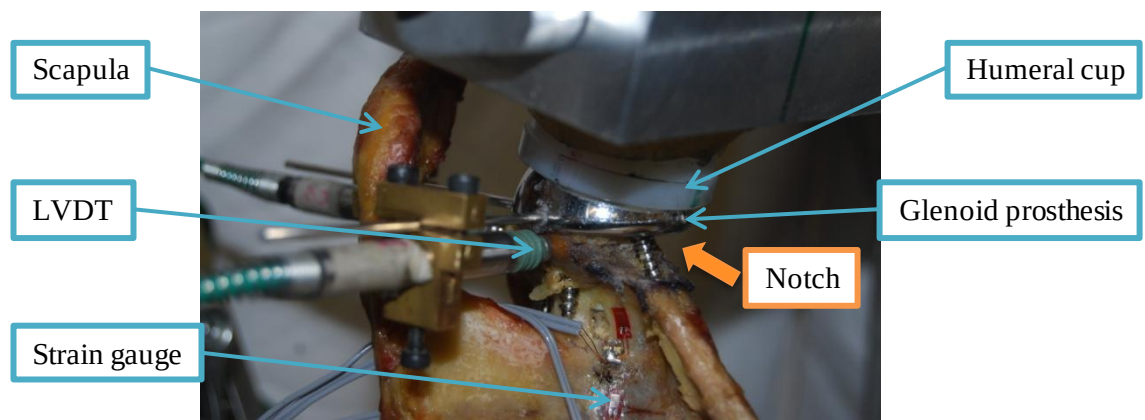


Figure 6-2 Experimental setup of an implanted scapula with a severe notch.

6.2.2.2 FE Modelling

CT data of the cadaveric scapulae, which were tested in the experiments, were used. For each scapula, two finite element models of bone were created. Both these models were virtually merged with a FEM of Delta CTA RSA. The implantation procedure introduced in Chapter 5.3.2.1.1 was used in the FE modelling. A big notch classified as grade 4 in the system of Nerot, was created in one of the two FE models. The notch was created in terms of the same geometry standard (Simovitch et al., 2007) with that used in the in-vitro testing. For the purpose of comparing the pre- and post-notching results at an element-by-element basis, the FEM with an intact scapula neck was composed of the FEM with an inferior scapular notch and a FEM of the remaining bone (Figure 6-3 (a)). There were no relative motions at the interface between these two partial models. Figure 6-3 also shows that the FE notch (Figure 6-3 (a)) created in this study is apparently similar to the real notches (Figure 6-3 (b) and (c)) shown in published articles

(Simovitch et al., 2007; Nyffeler et al., 2004). FE modelling of an implanted scapula, which was identified in Chapter 4, was used in this study. In this study, the same fixation setup and loading conditions, which were used in the in-vitro testing, were modelled. In the FEM, predicted strains on the positions where the strain gauges were bonded in the experiments were recorded. Differences between strains in the FEM with an intact scapular neck and those in the model with a notch were presented. Predicted micromotion variations due to the inferior notch were also validated with the experimental data.

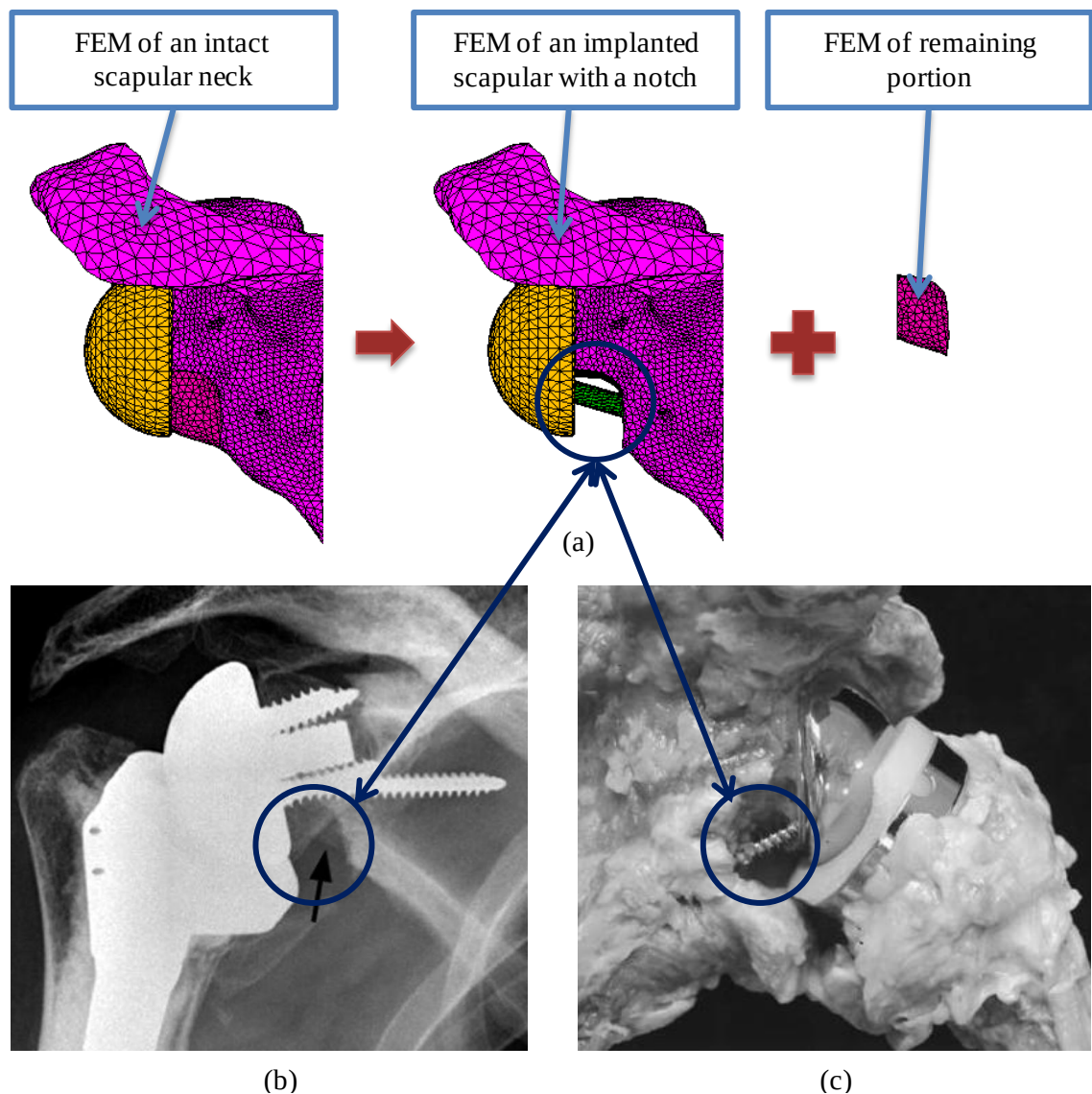


Figure 6-3 (a) FEM with an intact scapular neck; (b) FEM with an erosive scapular neck; (c) FEM of remaining bone; (d) Radiographic image of an implanted scapular with a severe inferior notch (adapted from Simovitch et.al 2007); (c) Retrieved DELTA III (adapted from Nyffeler et. al. 2004).

6.2.3 Results

For the purpose of validating predicted notch-induced variations of strains and micromotions with the in-vitro testing results, experimental and FE strains and micromotions before and after creating a notch are illustrated next.

6.2.3.1 In-vitro Testing

Figure 6-4 shows the averaged strain values of three subjects at a level of approximately 10 mm under the glenoid resected surface in various loading conditions. An increase in the experimental strain values due to the notch was observed in the posterior and anterior regions, and a decrease was found on the inferior and superior glenoid surfaces. Another observation is that strain changes due to the inferior glenoid notch were load-dependent. As a result of the predominant GH force during arm elevation being compressive and superior shear forces, most strains were negative, with exception of the inferior values which were positive in the entire test.

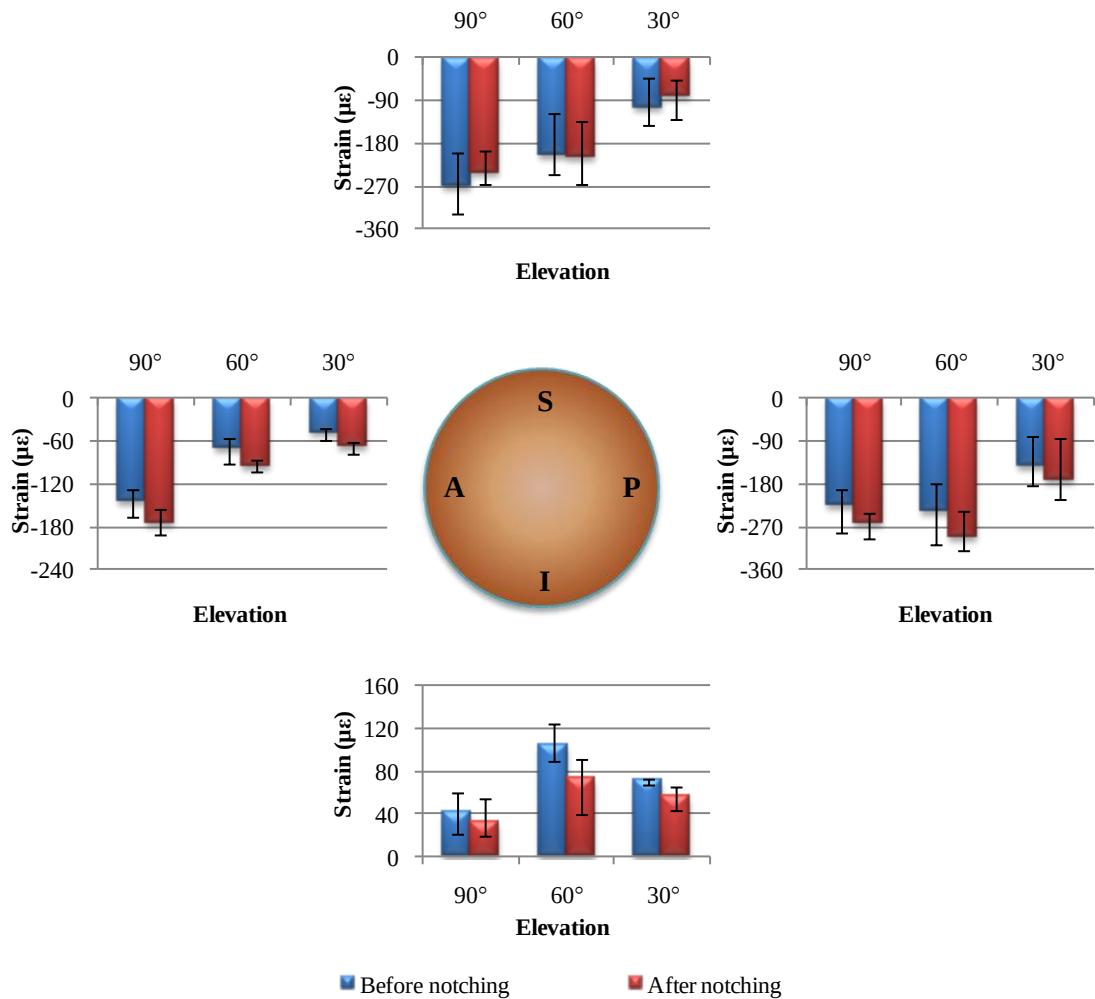


Figure 6-4 Strains on the glenoid surface at the lateral level (approximately 10 mm under the glenoid resected surface) before and after notching.

The experimental strains locating at a level of approximately 25 mm under the resected surface before and after creating the notch are illustrated in Figure 6-5. At this level, influences of the inferior scapular notch on strain variations were also load-dependent. For instance, the experimental strains in the medial superior region reduced 13 $\mu\epsilon$ and 11 $\mu\epsilon$ at 90° and 30° of arm elevation, respectively, but increased 24 $\mu\epsilon$ at 60° of arm elevation. Another observation is that notch-induced strain changes also depend on anatomical regions. For instance, at 90° of arm elevation, the strain measurements increased in both posterior and inferior glenoid after notching, while the values decreased in the anterior and superior glenoid.

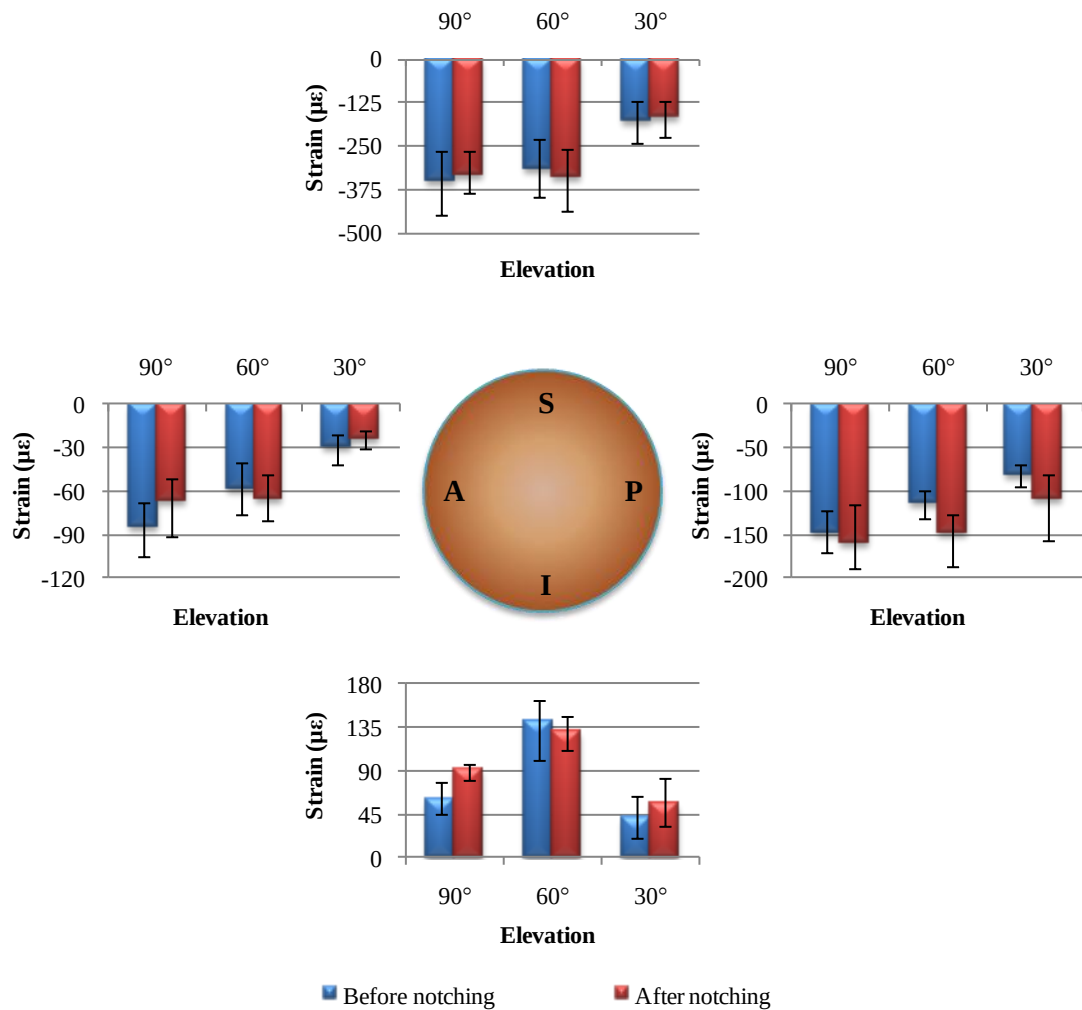


Figure 6-5 Strains on the glenoid surface at the medial level (approximately 25 mm under the glenoid resected surface) before and after notching.

Bone-prosthesis interface micromotions before and after creating a notch are illustrated in Figure 6-6. In cases of the erosive scapular neck, there was no interface between the inferior rim of the glenoid prosthesis and the adjacent bone, thus micromotions in the inferior region did not exist. Due to the single-direction characteristics of LVDT, the anterior and posterior measurements represented the relative motions in the anteroposterior direction, and the superior value displayed the micromotion in the superoinferior direction. In Figure 6-6, a notch-induced increase was observed in the anterior and posterior micromotions for all the three loading conditions. The trend of micromotion variations after creating an inferior notch in the superior region was complicated. The values declined at the elevation of 90° and 60°, increased at the elevation of 30°.

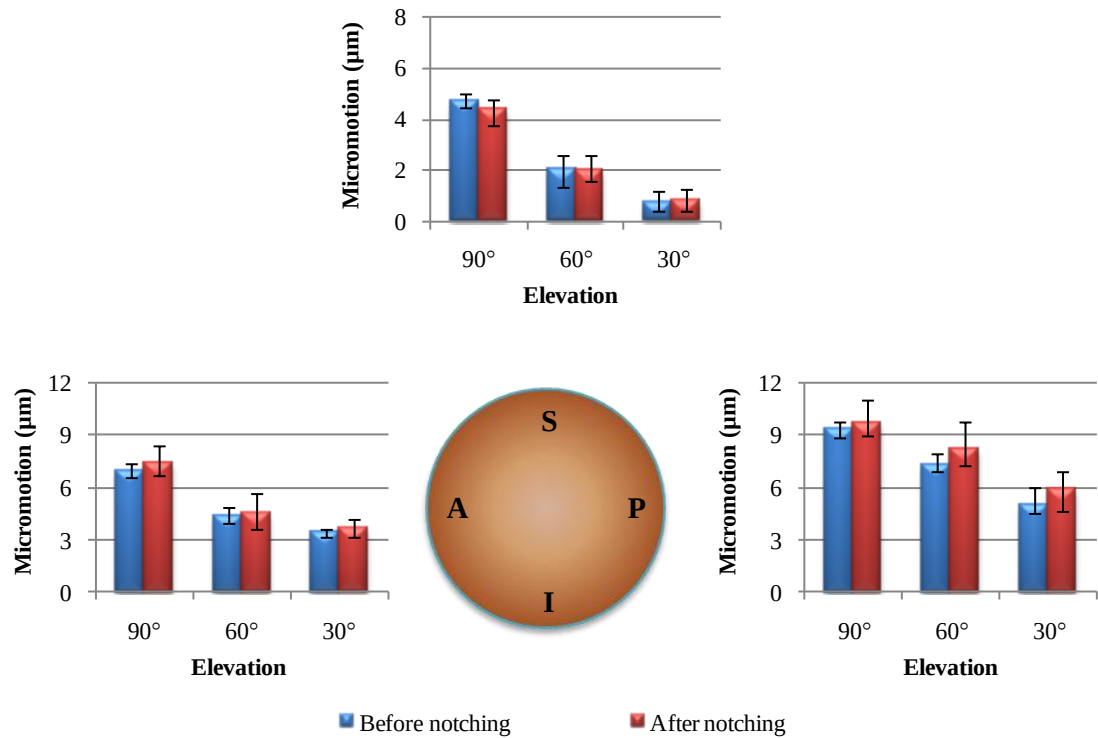


Figure 6-6 Bone-prosthesis interface micromotions around the glenoid in two situations of scapular neck (with and without erosion).

6.2.3.2 Experimental-FE Comparisons

In comparison with the experimental strain variations in the case of inferior scapular notch, the FE results generally presented a consistent trend (Figure 6-7 and Figure 6-8). The FE-experimental variation ranged from 0 $\mu\epsilon$ to 18 $\mu\epsilon$, although the percentage difference was big, varying from 0.38% to 76.5%. Another observation is that FE predicted strain variation was also load-dependent. Figure 6-9 presents the averaged strain variations around the glenoid. Both the experimental and FE results displayed an apparent reduction in strain variations from the lateral to medial level, with the significance in the anterior (15 $\mu\epsilon$ for the in-vitro testing and 19 $\mu\epsilon$ for the FE modelling) and posterior (16 $\mu\epsilon$ for the in-vitro testing and 17 $\mu\epsilon$ for the FE modelling) regions.

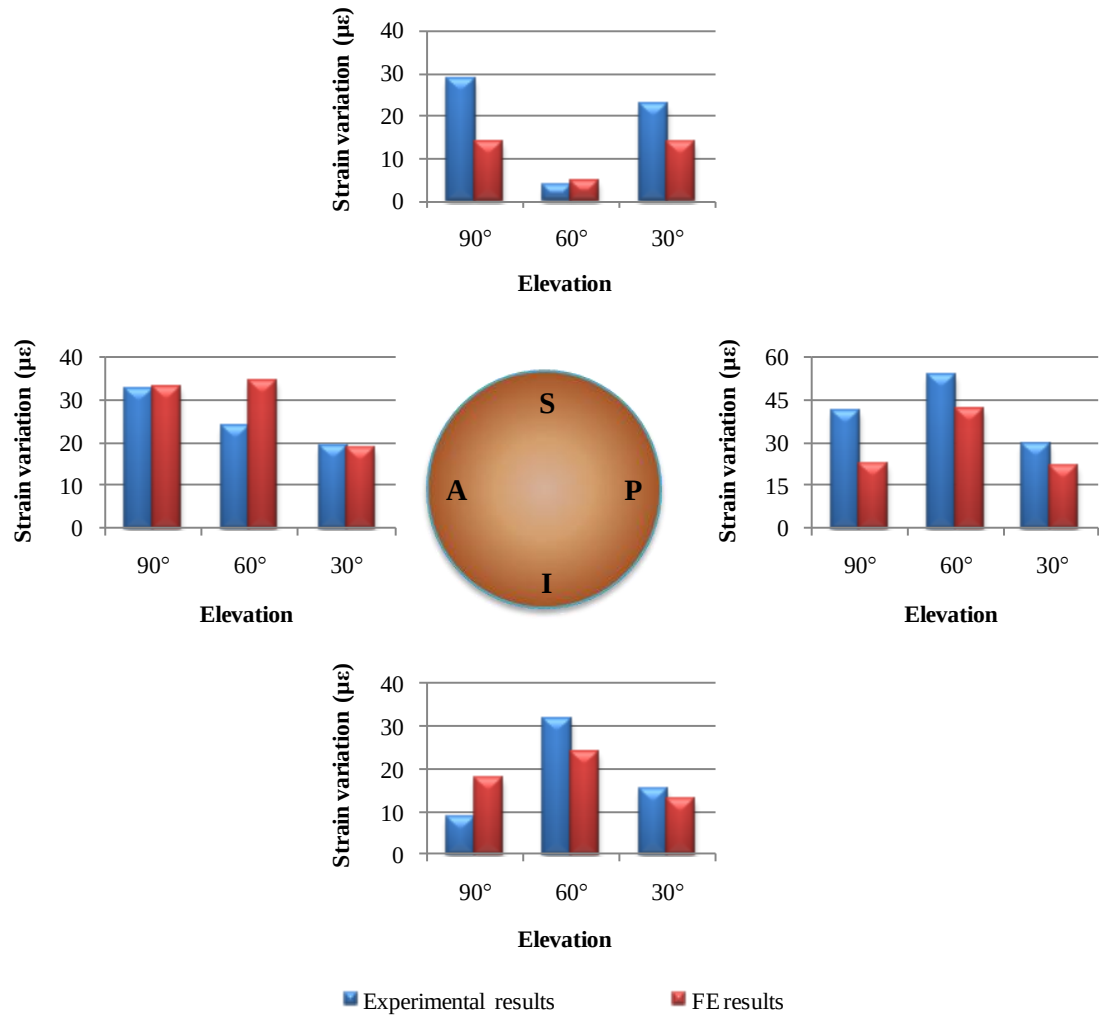


Figure 6-7 Experimental and predicted notch-induced strain variations at the level of approximately 10 mm under the glenoid resected surface for the three loading cases.

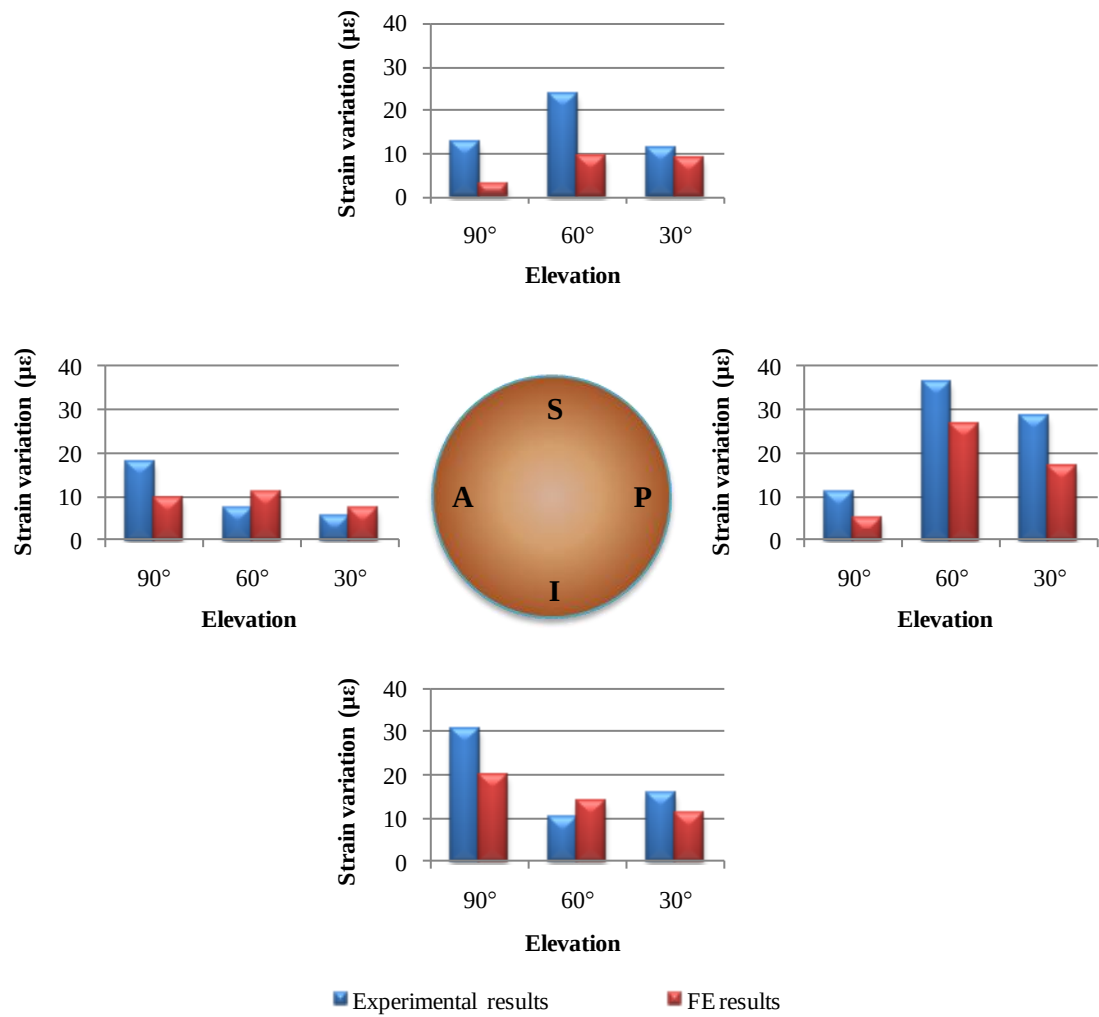


Figure 6-8 Experimental and predicted notch-induced strain variations at the level of approximately 25 mm under the glenoid resected surface for the three loading cases.

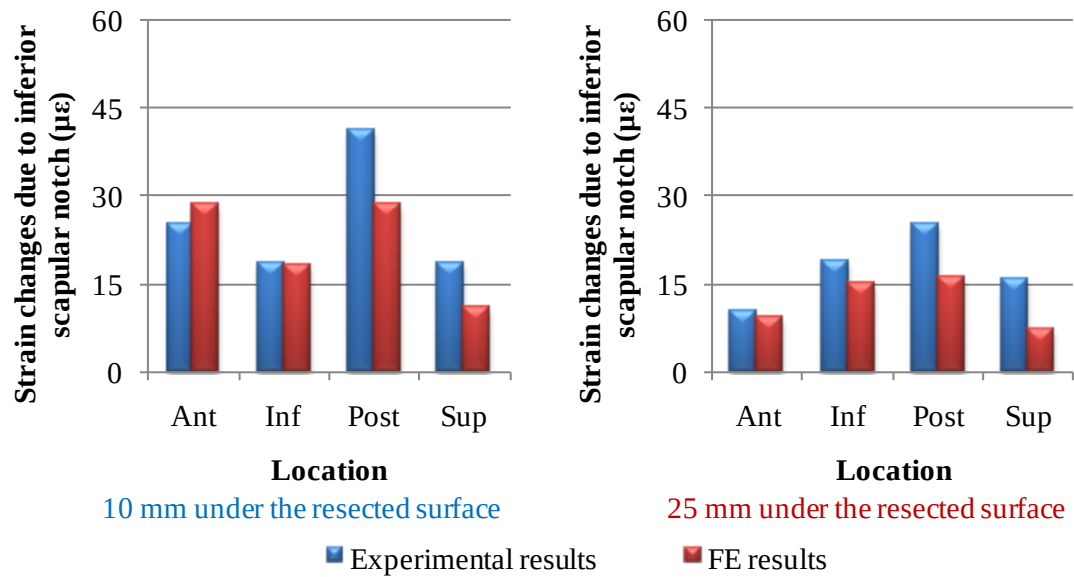


Figure 6-9 Comparison between the experimental and FE predicted strain variations in regions.

Figure 6-10 illustrates experimental and FE micromotion variations due to the inferior scapular notch. The results showed that differences between the FE and experimental micromotion variations vary from 0 to 0.5 μm . These values are much smaller than the required threshold (50 μm) for osseointegration in the Delta RSA (Pilliar et al., 1986), thus the FE predictions of notch-induced effects on the implant initial stability is believable.

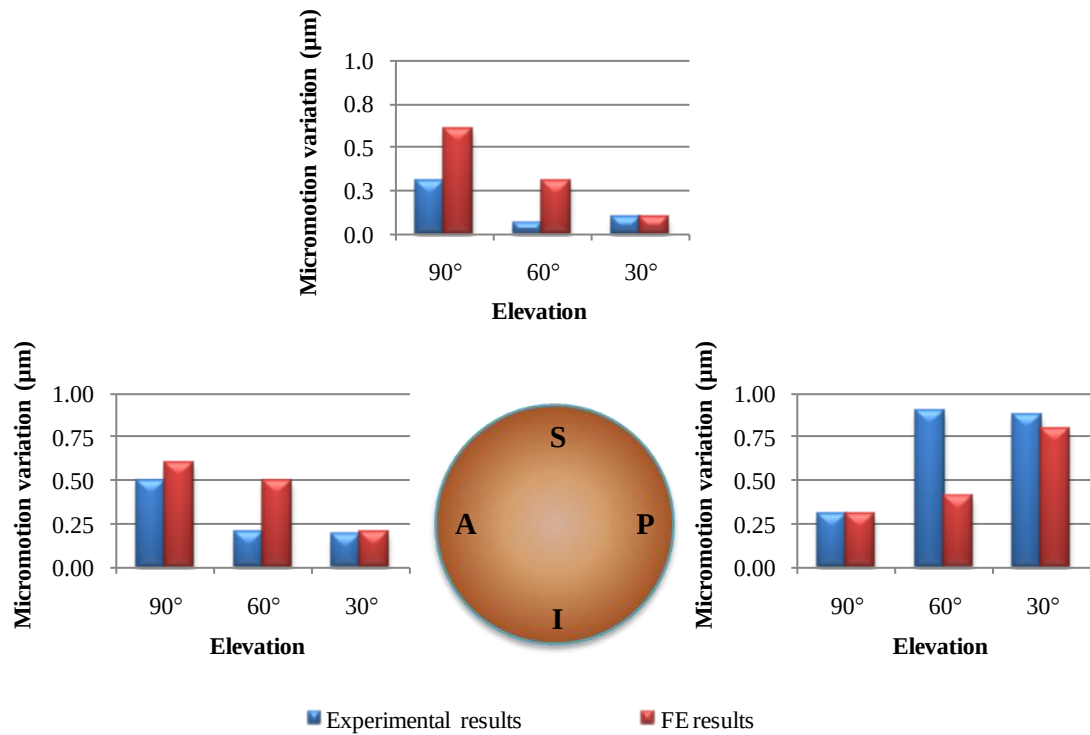


Figure 6-10 Experimental and FE predicted micromotion variations after inferior scapular notching.

6.2.4 Discussion

In this study, in-vitro methodology was used to validate the predicted notch-induced strain and micromotion variations. The results showed that differences between the FE and experimental strain variations after creating an inferior notch ranged from 0 to 18 $\mu\epsilon$. The notch-induced micromotion variations from the FEM and experiments were also different, varying from 0 to 0.5 μm . These differences between the FE and experimental values could be induced by the inconsistent glenoid prosthesis positions and notch geometries between the in-vitro testing and the FEM. In the experiment, the possible titling of glenoid prosthesis and the inconsistent hand-made notch relative to those in the FEM may lead to changes of force transmission from the implant to the bone, thus result in changes of notch-induced strain and micromotion variations. Another possible explanation for the FE-experimental differences is the contact condition between the bone and the non-locked anterior and posterior screws. In the FEM, these two screws were assumed secured firmly. While, the real fixation conditions of these two non-locked screws were complicated. They could be loosened during the process of creating the inferior notch in the experiment, and led to

experimental-FE stress and micromotion variations. However, given the low differences between the FE and experiment strains variations due to the notch, the FEM could predict effects of scapular notching on the bone strains. Additionally, the low differences between the FE and experimental micromotion variations after creating a notch demonstrated that the FEM of implanted scapula could predicted reliable initial stability.

Another observation in this study is that strain variations due to the notch at the lateral level were generally higher than those at the medial level. Possible reasons for this phenomenon are that the lateral strain gauges were closer to the notch, which was located in the inferior and lateral scapular neck, than the medial gauges, thus may be influenced more by the notch.

It is noticeable that only one simple activity (arm elevation from 30° to 90°) was tested in this study. The largest magnitude of GH joint contact force in this activity was only approximately 0.5×BW. This value is far smaller than the required peak GH force (approximately 1.5×BW) for the Delta RSA in physiological daily activities (Kontaxis, 2010). Thereby, results in this study are possibly not able to represent the notch-induced effects on the fixation of glenoid prosthesis. For this purpose, more complicated realistic daily activities should be studied.

6.2.5 Conclusion

This study is aimed to validate predicted notch-induced strain and micromotion variations with the results from the in-vitro testing. Experimental and FE strains and micromotions on the cadaveric implanted glenoid before and after creating an inferior scapular notch were recorded. There was a good agreement between the FE predictions and the experimental results. The FE-experimental difference in the notch-induced stress changes was within only 18 $\mu\epsilon$. This value was much lower than the yield strain of bone (at the level of 7000 $\mu\epsilon$) (Morgan, 2001). In addition, the FE-experimental variation in the notch-induced micromotion changes was no larger than 0.5 μm . This value is also quite low in comparison with the required threshold (50 μm) for the postoperative bony ingrowth (Pilliar et al., 1986). Therefore, FE modelling in this study

was able to predict believable strains and micromotions, and would be used for the investigation of notch-induced effects on the glenoid prosthesis fixation.

6.3 Investigation of Notch-induced Effects on the Glenoid Prosthesis Fixation

6.3.1 Introduction

FE methodology has been proved to be able to provide realistic and comprehensive analyses and has become a significant tool for validation of certain clinical hypothesis (Ahir and Walker, 2004; Hopkins et al., 2005; Sharma et al., 2009). In this chapter, FE methodology was employed to capture the strain and micromotion variations due to the inferior scapular notch and to predict notch-induced effects on the glenoid prosthesis fixation. This study is broken down into three aspects: (1) analysing the correlation between inferior scapular notching and breakage of inferior screw; (2) predicting the potential risk of notch-induced bone fracture on the surface of inferior screw hole and (3) investigating the initial stability of glenoid prosthesis in the Delta RSA after inferior scapular notching.

6.3.2 Materials and Methods

Except for the loading conditions, FE modelling of a scapula merged with a Delta CTA RSA before and after creating an inferior scapular notch in this study was the same with that introduced in Chapter 6.2.2.2. In this study, two physiological daily activities: (1) standing up from an armchair and (2) lifting a block to the head height were simulated. The first activity required the greatest glenohumeral contact force in the 12 tested daily activities for the Delta RSA in the study of Kontaxis (2010). The second activity also required comparatively high GH force (Kontaxis, 2010). Due to the loading-dependent characteristics of notch-induced strain and micromotion variations concluded from the in-vitro testing, these two activities were set in attempt to obtain the most severe results. Magnitudes and loading directions of required GH forces for the two activities were described in Figure 4-7. Maximum principal stresses on the surface of inferior screw and the screw hole were measured in the two FE models for each activity. Stress

changes were used to predict effects of inferior scapular notching on the breakage of inferior screw and on the potential risk of bone fracture on the surface of inferior screw hole. Micromotions in the two FEMs were also examined to investigate the notch-induced effects on the implant initial stability.

6.3.3 Results

Results in this study include three parts: (1) study of breakage of inferior screw, (2) predicted potential risk of bone fracture on the surface of inferior screw hole and (3) correlation between scapular notching and the initial stability of glenoid component of the Delta RSA.

6.3.3.1 Study of Breakage of the Inferior Screw

The maximum principal stress in the inferior screw was used as a failure parameter to assess the likelihood of screw breakage. Figure 6-11 illustrates maximum principal stress distributions on the surface of inferior screw when standing up from an arm chair. High stresses occurred in the corner of the screw and the metaglene for both the FEMs with and without an inferior notch. Additionally, notch-induced effects on the stresses were very significant, with an increase of 23 MPa when standing up from an arm chair and an increase of 7.8 MPa when lifting to the head height. The predicted maximum principal stresses reached 172 MPa and 72.5 MPa in the case of a grade 4 notch in these two activities.

Maximum principal stresses in each cross section from the conjunction between the inferior screw and the metaglene to the screw tip with a spacing of 2 mm were averaged. Variations of these mean stresses due to the inferior scapular notch are illustrated in Figure 6-11. There were similar trends in the notch-induced stress variations along the inferior screw for the two loading conditions of interest. They both displayed two peak maximum principal stress variations along the screw. One existed in the screw-to-metaglene conjunction and the other was in the intersection between the screw and the bone (transition area in Figure 6-12).

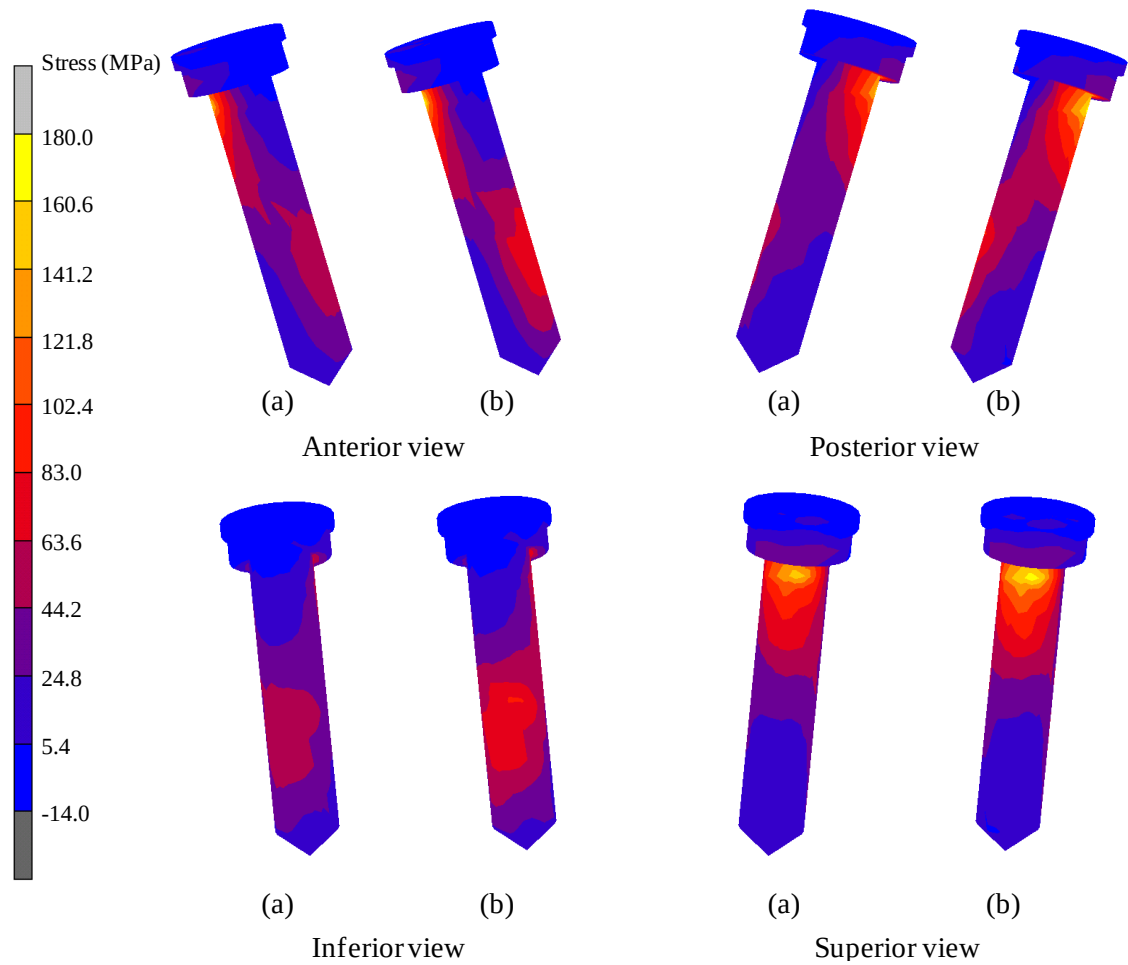


Figure 6-11 Distributions of maximum principal stresses on the surface of inferior screw: (a) before scapular notching, (b) after scapular notching.

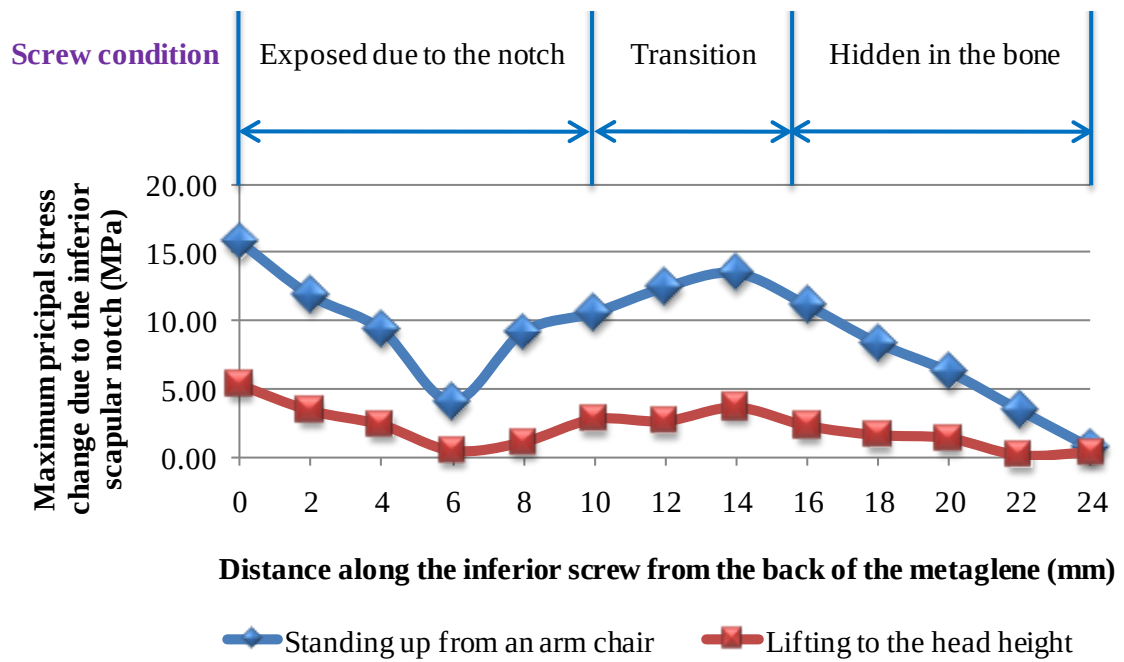


Figure 6-12 Maximum principal stress variations after the notching along the inferior screw from the back of metaglene to the screw tip.

6.3.3.2 Study of potential risk of bone fracture

Due to the frequent appearance of inferior screw fracture in recent postoperative radiographs for Delta RSA (Sirveaux et al., 2004; Grassi et al., 2009), bone safety at the bone-screw interface was concerned. The maximum principal stress in the cancellous bone was used as a parameter for the assessment of possible bone fracture and calculated under the two loading conditions of interest. Figure 6-13 illustrates the maximum principal stress distributions on the surface of inferior screw hole for the loading condition of standing up from an arm chair. The green line indicates the outline of the inferior screw hole. The orange line shows the surface of the inferior scapular notch. The results showed that the inferior scapular notch led to a mean increase of 0.98 MPa in the maximum principal stresses on the surface of the inferior screw hole. The peak notch-induced stress variation (3.7 MPa) was observed close to the notch surface. The predicted maximum stress in the cancellous bone reached 5.7 MPa after the notching.

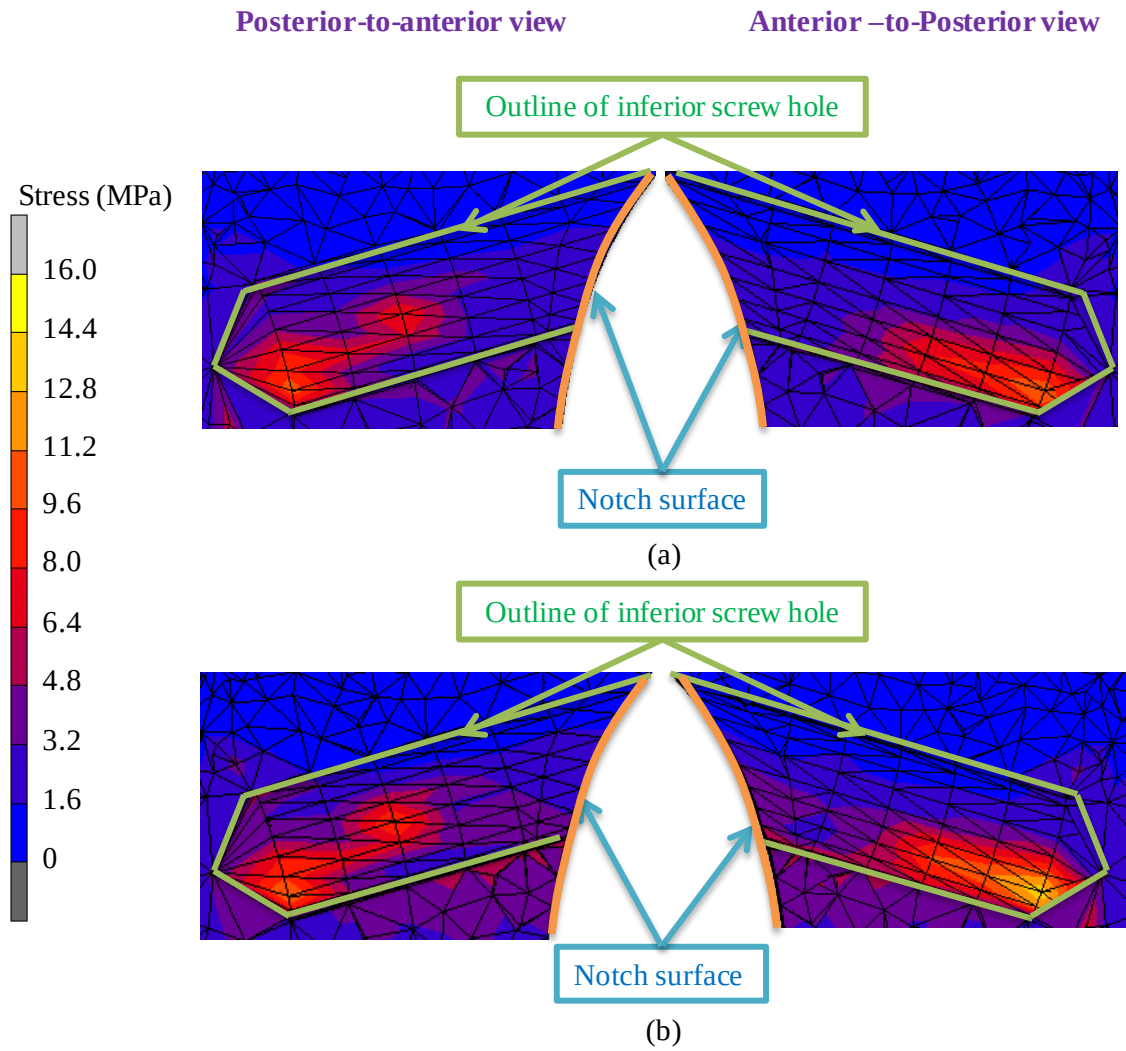


Figure 6-13 Maximum principal stress distributions on the surface of inferior screw hole and under the notch surface in cases of (a) intact and (b) erosive scapular necks.

6.3.3.3 Initial Stability of Glenoid Component in Delta RSA

Bone-prosthesis interface micromotions for the intact and erosive scapula neck conditions were calculated. Figure 6-14 illustrates bone-prosthesis interface micromotion distributions for the loading case of standing up from an arm chair. There were no significant changes of micromotions after the creation of a notch, with the mean value of about 0.3 μm for the standing up case and 0.1 μm for the lifting case. The peak micromotion increased 2.7 μm in the case of erosive scapular neck, and reached 59.6 μm .

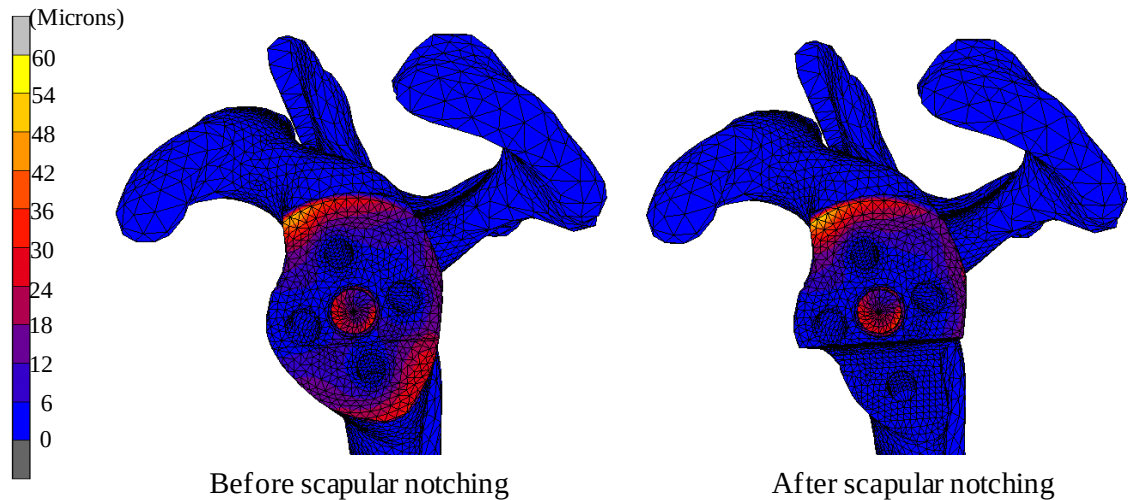


Figure 6-14 Predicted bone-prosthesis interface micromotions before and after inferior notching.

6.3.4 Discussion

In this study, FE methodology was employed to investigate effects of a grade 4 notch in the system of Nerot on the glenoid prosthesis fixation. Three factors were used to assess the notch-induced influences on the implant fixation: (1) the breakage of the inferior screw; (2) the potential risk of bone fracture and (3) the initial stability of the orthopaedic prosthesis.

6.3.4.1 Study of Breakage of the Inferior Screw

Due to the possible relation between the breakage of the inferior screw and the glenoid prosthesis loosening, notch-induced effects on the screw breakage were investigated. The results showed that notch-induced stress variations peaked at two positions: the screw-to-metaglene conjunction and the screw-to-bone intersection along the inferior screw. These two regions were consistent with the positions where inferior screw was broken in the postoperative radiographs of Delta RSA as shown in Figure 6-15 (Sirveaux et al., 2004; Grassi et al., 2009).

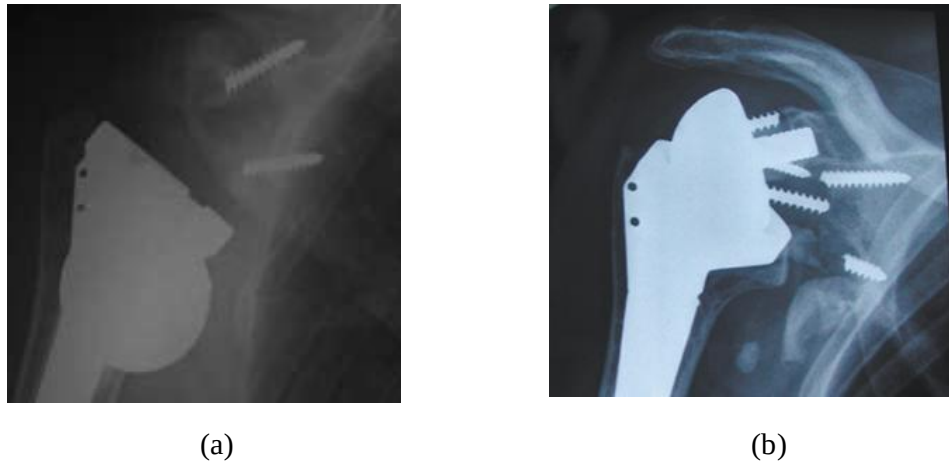


Figure 6-15 Screw breakage in the Delta RSA: (a) the screw was broken at the screw-to-metaglens conjunction (adapted from Grassi et al, 2009); (b) the screw was broken at the screw-to-bone intersection (adapted from Juvenstpan et al., 2005).

Another observation is that the predicted peak maximum principal stress of the inferior screw after creating an inferior scapular notch increased by 23 MPa, reaching 172 MPa. This value is much smaller than the yield stress of titanium material (815 MPa) (Biomechanics in Dentistry). Given the cyclic loading that the orthopaedic prosthesis suffers in daily life, the predicted peak maximum principal stress of the screw was almost half of the fatigue stress of titanium material (415 MPa) (Biomechanics in Dentistry). However, the stress concentration, which may occur in the sharp corner (i.e. thread) of the inferior screw, could substantially reduce fatigue life of the screw. Additionally, the damaged effects of inferior scapular notching may be aggravated when the prosthesis suffers high external loading, such as in the case of impingements between the screw and the inferior rim of polyethylene humeral cup (Figure 3-26), or in the case of some incorrect surgical techniques (i.e. overtensioning of deltoid muscle (Boileau et al., 2005)). Overtensioning of deltoid muscle may lead to higher glenohumeral joint contact force than that in the case of proper tension of deltoid muscle, and thus may result in higher stresses in the inferior screw than our predictions. Therefore, although the inferior screw could be safe after scapular notching, the increase of stresses in this scapular neck condition may increase the potential risk of inferior screw fracture.

6.3.4.2 Study of Potential Risk of Bone Fracture on the Surface of Inferior Screw Hole

Distributions of maximum principal stresses on the surface of inferior screw hole showed that the peak notch-induced stress variations occurred close to the intersection between the bone and the inferior screw. The predicted maximum stress variation due to the inferior notch was 1.34 MPa, and the peak stress of the cancellous bone after notching reached 3.32 MPa. This value was smaller than the failure stress (rang from 9 to 14.5 MPa) proposed by Anglin et al. (1999) and also lower than the regional ultimate strength (vary from approximately 13 MPa to about 110 MPa (Frich et al., 1997; Anglin et al., 1999; Mimar et al., 2008)). However, the bone experienced cyclic loading during the physiological daily activities. The fatigue stress of cancellous bone at the bone-to-screw interface during arm motions is unknown. For the 400 MPa epiphyseal cancellous bone, fatigue failure after 1 million cycles was reported at a stress level of 3.57 MPa (Michel et al., 1993). Moreover, the predicted significant factional increase of stresses (55%) due to the severe loss of supporting bone stock of scapular neck presented a potential risk of bone fracture for the elderly people whose bone quality may reduce with ages and a warning for the patients with osteoporosis, in which case the bone quality is lower than the normal. In addition, overtensioning of deltoid muscle could also aggravate the damaged effects of inferior scapular notching, as high GH joint force in this condition may generate higher stresses on the bone than those in the case of proper tension of deltoid muscle.

6.3.4.3 Initial Stability of Orthopaedic Prosthesis in Delta RSA

Bone-prosthesis interface micromotions were used to study the initial stability of glenoid component in Delta RSA. No significant changes of micromotions were observed after the creation of a notch, with the mean value of about 0.3 μm for the standing up case and of 0.1 μm for the lifting case. The FEM with the erosive scapular neck predicted a good bone-prosthesis environment for the postoperative bony ingrowth, as the peak micromotion increased only 2.7 μm . This conclusion was in agreement with the founding of Nyffeler and his associates (2004). In their study, a eight-month follow-up Delta III reverse shoulder prosthesis was retrieved post-mortem,

and found to be well supported by the bone biological attachments on the orthopaedic prosthesis, though there was a big notch (classified as grade 3 in the system of Nerot) at the inferior pole of the scapular neck.

6.3.5 Summary

This study is aimed to investigate effects of scapular notching on the fixation of glenoid component in the Delta RSA. Two FEMs of a scapula merged with the Delta CTA RSA were studied. In one of the FEM, an artificial Nerot grade 4 notch was created in the inferior pole of scapular neck. The relationship between inferior scapular notch and three factors relating to the implant loosening were analysed. Two physiological daily activities: standing up from an armchair and lifting a block to the head height were simulated.

The study of notch-induced effects on the inferior screw displayed that there were two peaks of the maximum principal stress variations along the inferior screw after creating a notch. They located in the screw-to-metaglene conjunction and the bone-to-screw intersection, respectively. These two positions of high stress variations were in agreement with the regions where inferior screw broke in the postoperative radiographs for the Delta RSA (Sirveaux et al., 2004; Grassi et al., 2009). The peak maximum principal stress in the case of an erosive scapular neck was observed in the load case of standing up from an armchair, being 172 MPa after the notching and 149 MPa before the notching. Both these maximal values were lower than the fatigue strength of titanium material (415 MPa). Thereby inferior scapular notching may be not the dominant factor for the breakage of inferior screw. However, it may aggravate the screw fracture in the case of impingements between the screw and the inferior rim of polyethylene humeral cup or in the case of some incorrect surgical techniques, such as overtensioning of deltoid muscles which may yield high glenohumeral joint contact force on the glenoid prosthesis.

Due to the frequently reported breakage of inferior screw, the safety of the bone at the screw-to-bone interface was studied. The predicted peak stress variation due to the inferior notch was 1.34 MPa, and the maximum stress of the cancellous bone reached 3.32 MPa. This value was lower than the failure stress (9 to 14.6 MPa) (Anglin et al.,

1999). However, the bone experiences cyclical loading during the physiological daily activities. In a failure study with bovine epiphyseal cancellous bone, the fatigue stress was reported at a stress level of 3.57 MPa for a modulus of 400 MPa bone (Michel et al., 1993). In addition, the high averaged fractional stress change (55%) due to the notch presented a high risk of bone fracture for the elderly people whose bone quality may reduce with ages and for the patients with osteoporosis.

In the bone-prosthesis interface micromotion study, the FEM with the erosive scapular neck predicted a good bone-prosthesis environment for the postoperative bony ingrowth. The notch-induced micromotion variation averagely increased 0.3 μm for the standing up case and 0.1 μm for the lifting case. The peak micromotion only increased 2.7 μm .

CHAPTER 7

PROSTHESIS LOOSENING INDUCED BY STRESS SHIELDING

7.1 Introduction

Bone loss is a common radiographic finding in patients treated with the reverse shoulder prosthesis (Guery et al., 2006; Grassi et al., 2009). However, the reason for this postoperative complication is still unknown. One of the most popular explanations is bone resorption due to stress shielding (detailed introduction was presented in Chapter 3.5). This phenomenon is identified by stress transmission from bone to metallic artificial joint replacements in some parts, leading to bone remodelling.

Finite element (FE) methodology has been a tool to assess strain-induced bone remodelling (Orr et al., 1988; Ahir et al., 2004). In the study of Orr et al. (1988),

stresses were found to be redistributed after the implantation of anatomical shoulder arthroplasty and the stress variations highly depended on the shoulder prosthesis designs. In 2004, Ahir et al. reported the postoperative stress variations after the implantation of Baylay-Walker reverse shoulder arthroplasty (RSA). This article is up-to-date the only published paper about the postoperative stress shielding in the RSA. Beyond these studies of stress shielding which occurred immediately after implantation, Sharma et al. (2009) used FE modelling to investigate the bone adaptation process after anatomical shoulder prosthesis implantation, and predicted (2010) the association between the bone remodelling and prosthesis designs. However, there have no published studies predicting strain-induced bone adaptation in the RSA, although bone loss of this type of prosthesis is frequently observed in the postoperative radiographs (Guery et al., 2006; Grassi et al., 2009).

In Delta RSA, the metaglene and glenosphere are made of cobalt-chrome alloy with the Young's modulus of approximately 220 MPa (Hopkins et al., 2008). The peripheral screws are made of titanium, with the Young's modulus of approximately 110 MPa (Hopkins et al., 2008). The considerable differences of material properties between the prosthesis and the bone (the maximum Young's modulus is approximately 20 MPa) could lead to stress shielding after the prosthesis implantation. In addition, the considerable reduction (range between 10% and 52.5% (Kontaxis, 2010)) of glenohumeral (GH) joint force for Delta RSA is possible another factor that result in the decrease of postoperative stresses in the bone and thus trigger adaptive bone remodelling. Given the bone loss observed in several postoperative radiographs for Delta RSA (Guery et al., 2006; Grassi et al., 2009) and the possibility of bone stress shielding for this type of prosthesis, adaptive process of bone remodelling in the case of Delta RSA is investigated.

The purpose of this chapter is to investigate the relationship between clinically observed postoperative bone loss and adaptive bone remodelling, and to predict effects of postoperative bone resorption on the long-term glenoid fixation and reduction of shoulder function. Before this study, validation of FE predicted stress variations due to the insertion of metallic prosthesis with in-vitro testing results was performed.

7.2 Validation of FE Postoperative Stress Variations with In-vitro Results

This study is aimed to investigate the phenomenon of stress shielding due to the insertion of metallic RSA, and to validate FE predicted stress variations with experimental results. The entire study includes two parts: in-vitro testing and FE modelling.

7.2.1 Materials and Methods

7.2.1.1 In-vitro Testing

Three cadaveric shoulders from three subjects without shoulder pathology were used in this study. The preparation of tested scapulae in this study was the same to those described in Chapter 5.2.2.1. The in-vitro testing involved two stages: before and after the prosthesis implantation.

7.2.1.1.1 *Before glenoid prosthesis implantation*

The intact scapula was secured in a rectangular bone container and attached with eight strain gauges. The method of bonding strain gauges and locations of these gauges were the same to those introduced in Chapter 5.2.2.1. As rotator cuff and other muscles and ligaments around the glenoid had been cut off, the humeral head may migrate on the smooth and shallow glenoid cavity under shear forces. Thereby, only axial compressive force was used in this study. A half sphere humeral head was used to apply compressive loading on the glenoid. Four load conditions, 300 N, 450 N, 600 N and 750 N, were tested for each scapula, respectively. Strains around the glenoid under each load condition were recorded. The setup of the test was shown in Figure 7-1 (a).

7.2.1.1.2 *After glenoid prosthesis implantation*

When the intact scapula test was finished, Delta CTA RSA (Depuy) was inserted in each bone by an experienced orthopaedic shoulder surgeon (Dr. Gregory) according to

the Delta CTA RSA surgical technique (Depuy). The procedure of glenoid component implantation was the same with those introduced in Chapter 5.3.2.1. To keep the locations of pre- and postoperative strain gauges completely constant, the gauges used preoperatively were still used postoperatively, unless some gauges were broken during the process of prosthesis implantation. For the purpose of avoiding complicated factors due to changed GH forces on the pre- and post-strain measurements, the force used in the intact bone strain test was used in the implanted scapula experiment. Setup of the implanted scapula test is shown in Figure 7-1 (b). Strains around the glenoid for each loading condition were recorded. Changes of strain values after the orthopaedic prosthesis implantation were used to evaluate postoperative stress shielding. T-test was used to investigate differences of strain variations due to varied loading conditions. The p value lower than 0.05 was considered to be statistically significant,

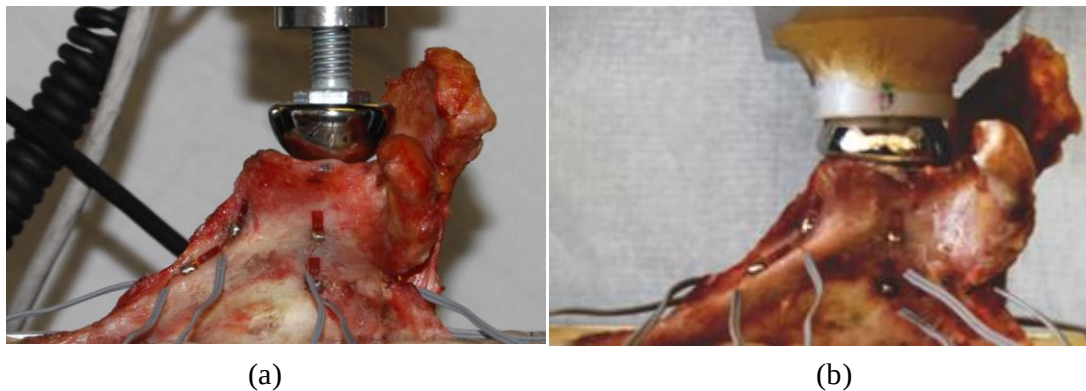


Figure 7-1 Photograph showing the setup of strain tests before and after orthopaedic prosthesis implantation: (a) intact scapula test; (b) implanted scapula test.

7.2.1.2 FE Modelling

Two FEMs were created based on the CT images of each cadaveric scapula. One of the FEMs was virtually merged with a FEM of Delta CTA RSA with the implantation procedure described in Chapter 5.3.2.1. For the purpose of investigating variations between pre- and postoperative strains or mechanical stimuli at an element-by-element basis, the FEM of intact scapula (Figure 7-2 (a)) was composed of the FEM of bone which was cut off (Figure 7-2 (c)) and the FEM of remaining bone (Figure 7-2 (b)).

There were no relative movements at the interface between the two FEMs of partial bone.

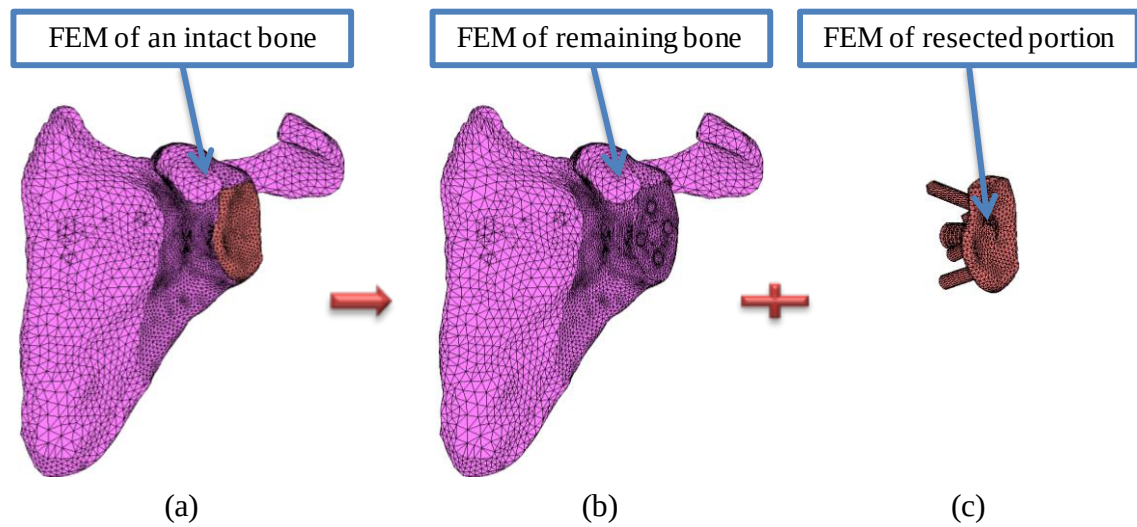


Figure 7-2 Schematic components of the FEM of an intact scapula.

Except for loading conditions, the FE modelling introduced in Chapter 4 was used to create the FEMs of both intact and implanted scapulae. Loading conditions, which were used in the in-vitro testing (Chapter 7.2.1.1), were used in this study (Figure 7-3). Predicted strains on the locations, where the strain gauges were bonded in the experiments, were recorded in the FEMs of both intact and implanted scapulae. Predicted strain variations due to the insertion of glenoid prosthesis were validated with the experimental measurements.

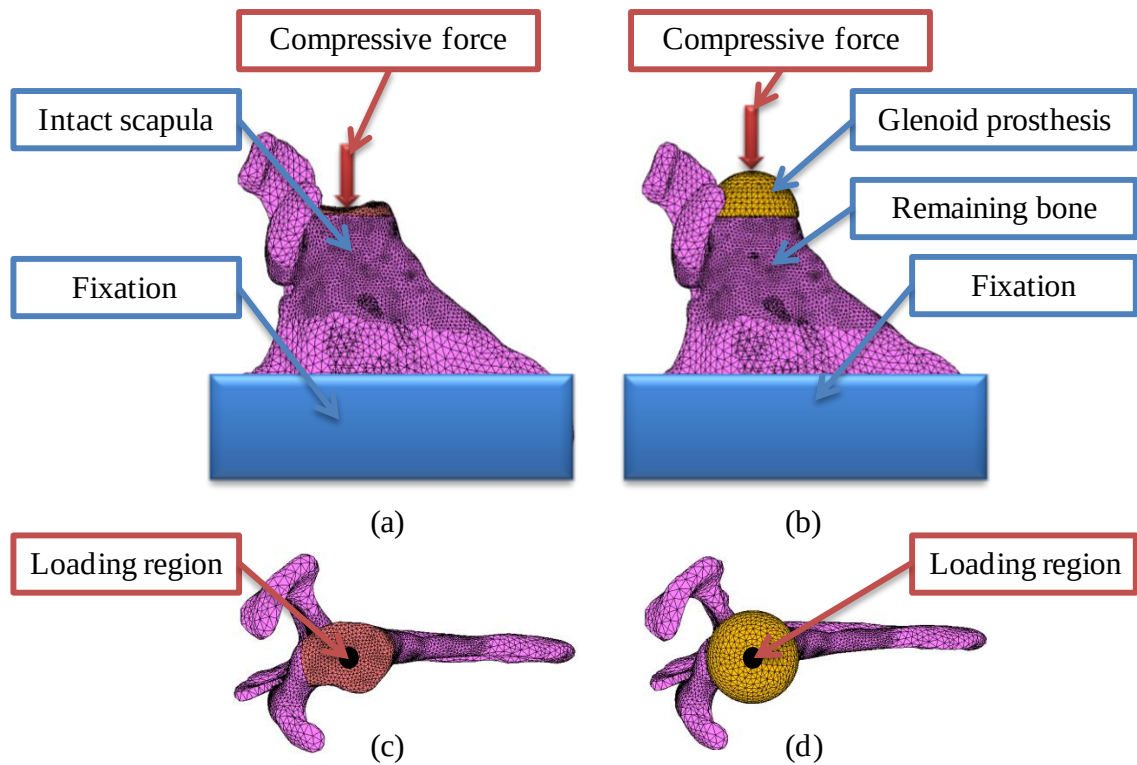


Figure 7-3 FE modelling of (a) an intact scapula and (b) the same scapula merged with the Delta CTA RSA, and the loading position on (c) the intact and (d) the implanted bone.

7.2.2 Results

7.2.2.1 In-vitro Testing

Mean pre- and postoperative strains for the three subjects at two levels (10 mm and 25 mm under the glenoid fossa, respectively) were shown in Figure 7-4 and Figure 7-5. Due to differences in individual bone quality, size and gender, the measured strain magnitudes for the three subjects varied in a large range. Another observation is that the averaged surface strains were reduced in most regions around the glenoid after the prosthesis implantation, with the exception of the values in the medial inferior and posterior regions (at the level of 25 mm under the resected surface), where postoperative strains increased in all the load cases. The mean strain reduction for all the three subjects in all the load cases due to the insertion of metallic prosthesis was 12.95%. The maximum strain reduction (136 μm) was observed in the lateral posterior region under 700 N compressive forces, followed by the lateral superior region (128 μm). For a particular position of interest, strain changes for various loading conditions were

statistically significant (paired t-test, $p < 0.05$). In comparison with the medial level, the mean postoperative strain reduction at the lateral level for all the load cases (52 $\mu\epsilon$) was high (Figure 7-6).

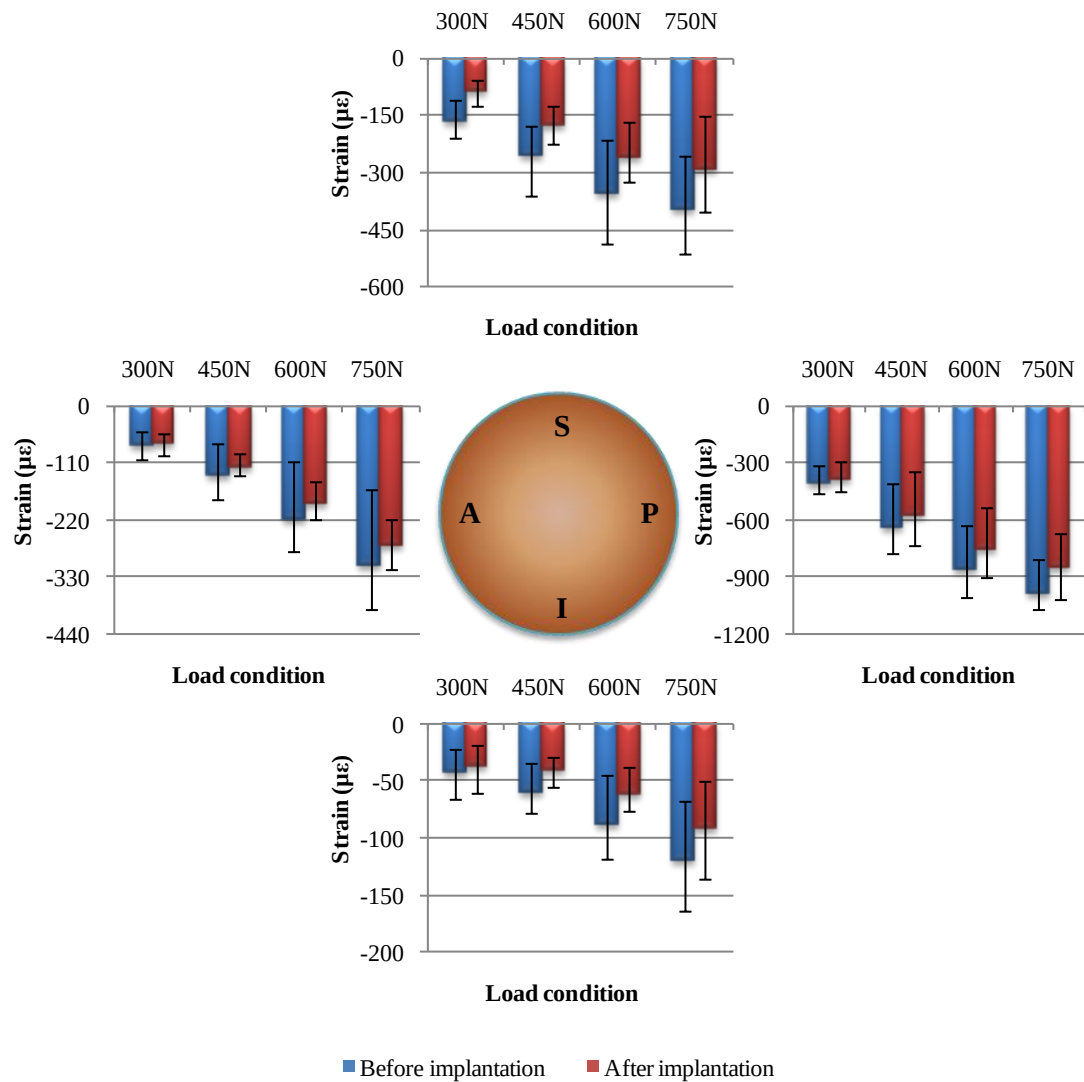


Figure 7-4 Mean compressive strains around the glenoid at a level of 10 mm under the glenoid resected surface before and after glenoid prosthesis implantation.

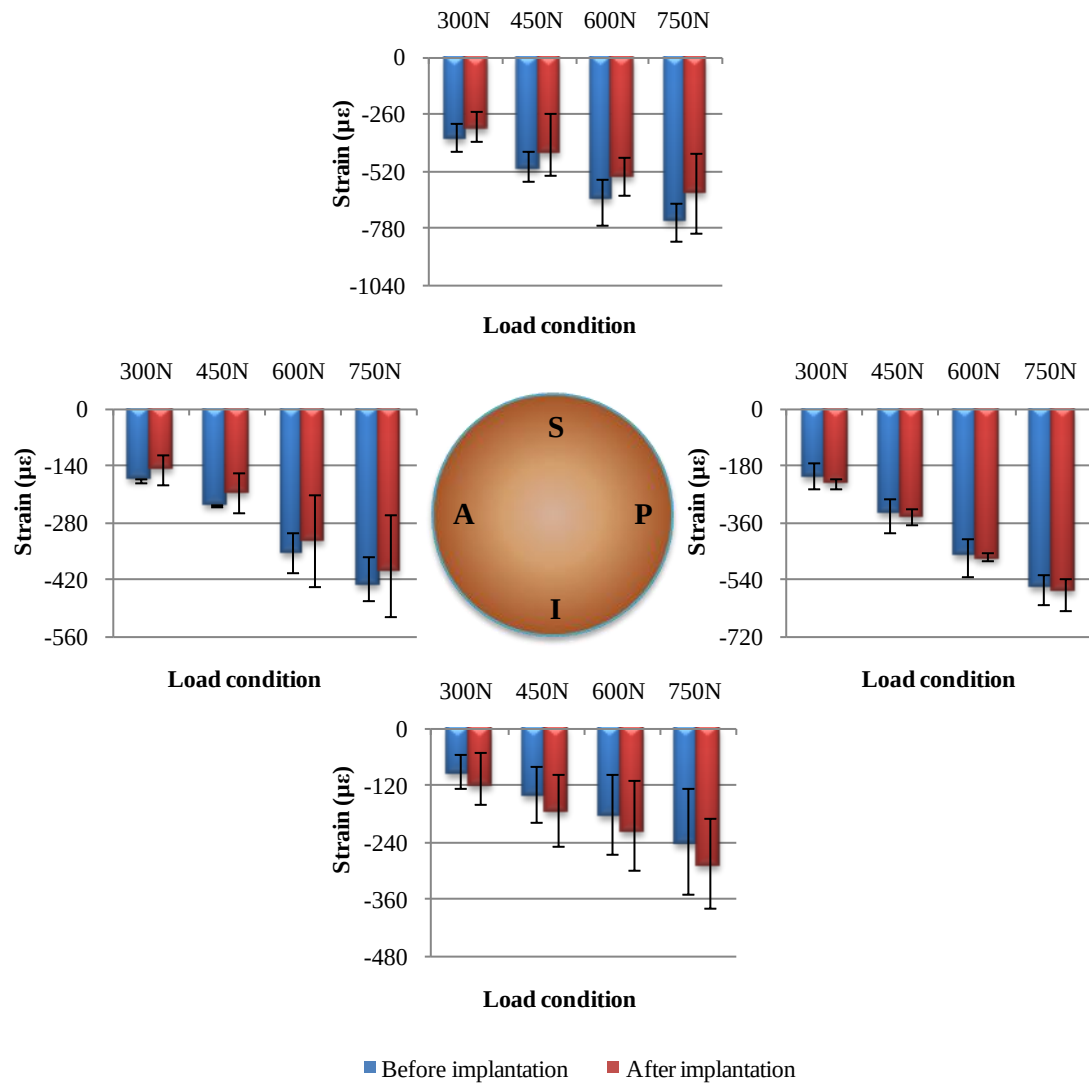


Figure 7-5 Mean compressive strains around the glenoid at a level of 25 mm under the glenoid resected surface before and after glenoid prosthesis implantation.

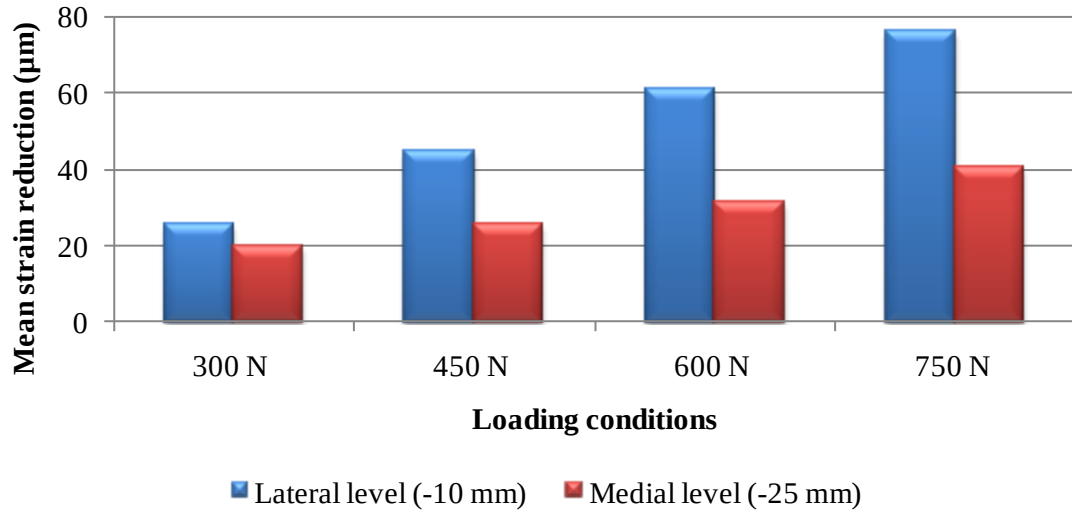


Figure 7-6 Postoperative strain reductions after glenoid prosthesis implantation in various loading conditions.

7.2.2.2 Comparison between FE and Experimental Results

Figure 7-7 shows the mean experimental and FE bone strain variations around the glenoid due to the insertion of metallic prosthesis. Predicted FE results showed an agreement in the distributions of strain variations around the glenoid with those measured in the experiments. Both the FE and experimental results displayed a predominant reduction of the strains after prosthesis implantation, with the exception of the medial inferior and posterior regions, where increases of postoperative strain were observed. In addition, both the results showed high strain variations after the prosthesis implantation in the superior and lateral posterior regions. Another observation was that strain variations at the lateral level (10 mm under the resected surface) in both the FEMs and the experiments were generally high, and predicted a high possibility of strain-adaptive bone remodelling in comparison with those values at the medial level.

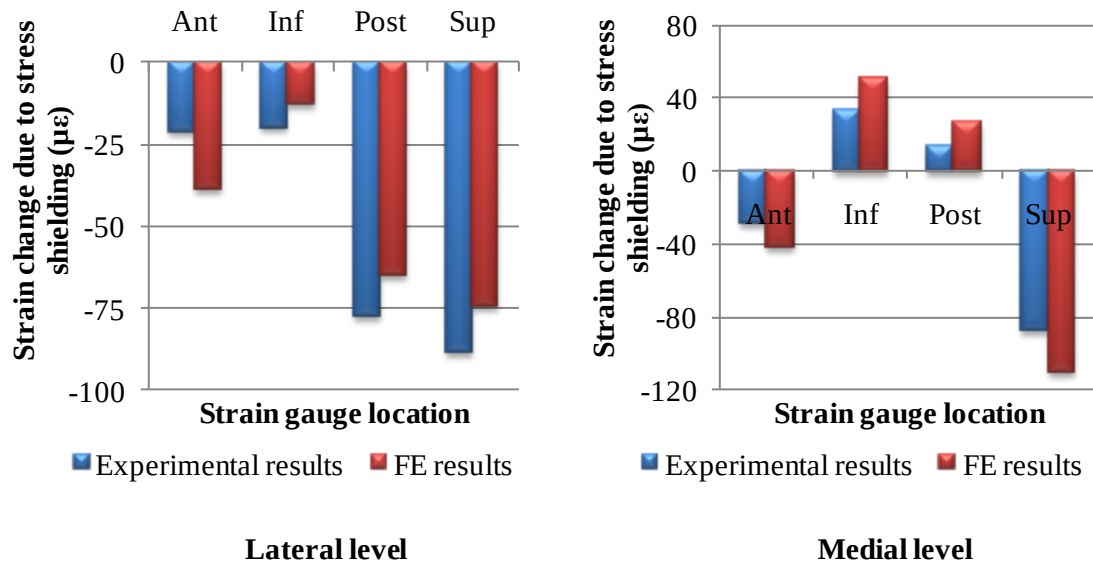


Figure 7-7 Experimental and FE predicted averaged strain changes in the regions of interest around the glenoid; Ant/Inf/Post/Sup: Anterior/Inferior/Posterior/Superior.

7.2.3 Discussion

In this study, stress shielding after the Delta CTA RSA implantation was investigated with in-vitro testing and FE modelling. Both the experimental and FE results showed high strain reductions at the lateral level, which was close to the metaglene. This finding is in agreement with the radiographic observations for Delta III RSA, in which significant bone loss occurred on the back of metaglene (Grassi et al., 2009). In comparison with the lateral level, strain reduced slackly in the medial anterior and superior, and increased pronouncedly in the medial posterior and inferior. This tendency of strain variations from the lateral to the medial is also consistent with the clinical finding in which bone atrophy occurred close to the tray and some bone hypertrophy existed at the stem tip postoperatively (Bertin et al., 1985).

The purpose of this study is to validate the FE predicted strain variations due to the insertion of metallic prosthesis with the experimental results. The FE strain variations postoperatively showed consistent trend with the experimental results. Strains decreased in most regions around the glenoid, with the exception of the medial inferior and posterior regions (at the level of 25 mm under the resected surface), where the values increased in all the load cases. In addition, FE predicted strain reduction due to the

insertion of metallic prosthesis (an average of 10.02% for all the three subjects in all the load cases) was comparable with that (12.95%) in the in-vitro testing. The difference between the FE and experimental results may be caused by a few factors explained next. Firstly, in the FEM, the loading point in the intact scapula study was the same to the location of glenoid prosthesis. However, it was difficult to make the prosthesis position in the experiment consistent to the loading point in the intact bone test. Thereby, the experimental variations between the pre- and postoperative strains were different from those predicted from the FE modelling. Secondly, in the FE modelling, the glenoid component was secured ideally parallel to the glenoid fossa in terms of the surgical technique. While, in the in-vitro testing, the glenoid prosthesis may incline slightly relative to the glenoid articular surface due to the hand-made reaming. The difference between the FE and experimental prosthesis position would lead to varied bending on the complicated-shape scapula, and consequently result in the FE-experimental variations. However, given the consistent FE and experimental tendencies of postoperative strain variations around the glenoid and comparable percentages of mean strain reduction, the FE model could predict believable stress shielding.

Another observation from the in-vitro testing is that there was a big variation (i.e. ranging from 49 $\mu\epsilon$ to 221 $\mu\epsilon$ for different subjects at 750 N compressive loading condition) in the subject strain changes after the glenoid prosthesis implantation. It could be explained by the subject-specific characteristics of human bone and variations of metaglene inclinations with respect to the glenoid in the experiments. For instance, the subject-specific retroversion of glenoid surface with respect to the scapula plane ranged from -7.4° to 10° (Saha, 1971, 1983)), thus the glenoid prosthesis inclination could be different with respect to the glenoid surface, and lead to different bending of the scapula.

There are two limitations in this study. Firstly, only axial compressive forces were employed, thereby, strain changes after prosthesis implantation may not represent the real stress shielding which is upon both compressive and shear loading. However, this study is aimed to evaluate the FE predicted strain variations after the insertion of metallic glenoid prosthesis, thus as long as the experiments and FEMs were subject to the same loading conditions, the experimental results can be used for the validation. Secondly, the same forces including magnitudes and directions were used in both the

intact and implanted bone tests in order to avoid the complicated factors induced by the changed GH forces postoperatively. In reality, the GH force reduced after the Delta RSA implantation (Terrier et al., 2008; Kontaxis, 2009), thus the experimental and FE stress shielding results in this study could not represent the reality. Forces for the Delta CTA RSA in physiological daily activities should be used for the purpose of investigating stress shielding and bone remodelling. This study will be introduced in Chapter 7.3.

7.2.4 Conclusion

In this study, in-vitro testing was used to validate the reliability of FE predicted strain variations due to insertion of metallic prosthesis. The FE modelling predicted consistent trend of postoperative strain variations with the experimental measurements. Results obtained from both the FEM and experiments showed strain decreases in most regions around the glenoid, with the exception of medial inferior and posterior regions, where the values increased in all the load cases. In addition, the mean FE predicted strain reduction after prosthesis implantation was 10.02%. This value was comparable with that (12.95%) obtained from the in-vitro testing.

7.3 Bone Remodelling Study after Glenoid Prosthesis Implantation

7.3.1 Introduction

Radiographs used as a common tool to measure bone adaptation is not very sensitive and accurate (Grassi et al., 2009). Moreover, clinic radiograph may not be able to quantitatively measure the extent of postoperative bone resorption, and to comprehensively evaluate the effect of bone remodelling on the glenoid prosthesis fixation. In comparison with the in-vitro testing, in which strains at a few points on the glenoid surface were obtained, FE methodology is able to calculate variations of mechanical parameters at each node or element and to predict postoperative density redistributions even inside of the bone. In this study, a computational bone remodelling

model was created to understand and quantitatively describe the functional adaptation of bone, to simulate postoperative bone adaptation and to analyse the effect of this adaptive phenomenon on the failure of glenoid prosthesis in Delta RSA.

In this chapter, bone adaptation in physiological activities was simulated, and the effects on the failure of RSA were analysed. Additionally, the correlation between bone resorption and scapular notching, which frequently occur after the Delta RSA implantation, was also investigated.

7.3.2 Materials and Methods

Except for the force conditions, FE modelling introduced in Chapter 7.2.1.2 was used in this study. For the purpose of mimicking the bone remodelling process in reality, five physiological daily activities shown in Table 4-4 were used in this study. Four of these activities yield the largest moment arms of compressive and shear component forces in the anterior and posterior glenoids, respectively. Another activity yields very small anteroposterior shear force. Force magnitudes and loading positions relative to the glenoid or the implant in the intact and implanted scapula FEMs were illustrated in Figure 4-7.

Although there have no clinical reports regarding the relationship between bone resorption and the time after RSA implantation, studies of knee and hip joints show that strain-induced bone loss occurs immediately after the operation, and develops pronouncedly within 1 to 2 years (Seitz et al., 1987; Li and Nilsson, 2000; Soininvaara et al., 2004). Bony ingrowth on the orthopaedic glenoid prosthesis base may occur as early as 1 to 3 months after the implantation (Boby et al., 1981; Soballe et al., 1992). Thereby, there could be two bone-prosthesis interface conditions in reality: no bony ingrowth and biological bony attachments covering the entire porous-coated glenoid prosthesis basing. In this study, both these two bone-prosthesis interface conditions were modelled and investigated. For the prosthesis fixation without bony ingrowth, the bone-prosthesis interface was unbounded and modelled by defining contact between the two surfaces with a Coulomb's friction coefficient of 0.4 (Hopkins et al., 2008). For the case of full bony ingrowth, the bone-prosthesis interface was bonded. Bonded screw-bone interface was set up for both cases.

In this study, the strain-adaptive bone remodelling algorithm described in Chapter 2.2.4.4 was performed. FE computational implementation of the bone remodelling theory was initially developed in Fortran at the Orthopaedic Research Laboratory, Institute of Orthopaedics, University of Nijmegen, Netherlands, and further improved by Jan Kerner, Charalampos Bitsakos and Desmond Chong. The detailed description of the bone remodelling theory integrated with FE codes and its clinical validation with the periprosthetic human bone adaptation were reported in Jan Kerner's (1999) and Charalampos Bitsakos' (2005) PhD dissertations in the Biomechanics Section, Mechanical Engineering Department, Imperial College London. The bone remodelling stimulus is the strain energy density per unit of mass. For the purpose of studying postoperative bone density changes on the glenoid side, only internal structure (bone density change) was remodelled and analysed. In other words, the change of outer shape of the glenoid remained static. FEMs of both the intact and implanted scapulae were used. The iterative computer simulation program is illustrated in Figure 7-8. The intact glenoid FEM provided the homeostatic reference stimulus (S_{ref}). FEM of the same scapula merged with a Delta CTA RSA supplied the postoperative stimulus (S_{op}). Five physiological daily activities were simulated, in order to obtain realistic bone behaviours. The average of mechanical stimuli in this five load conditions, S , ($S = \frac{1}{n} \sum_{i=1}^n (\frac{U_i}{\rho})$, where n is the number of loadings, U is the strain energy, and ρ is the apparent density) was used as the stimulus in one iteration and compared on an element-by-element basis. The threshold level of s was set as 0.75, which has been successfully validated with in-vitro tests by Jan Kerner (1999), Charalampos Bitsakos (2005) and Weinans (1993). The apparent density, ρ , defined as wet weight divided by total volume of specimen (Rho et al., 1995) varies from 0.01 g/cm³ (complete bone resorption) to 1.8 g/cm³ (maximum density of cortical tissue in glenoid). The minimum density was set to 0.01 g/cm³ but not 0 g/cm³ in order to avoid numerical problems due to element with zero Young's modulus.

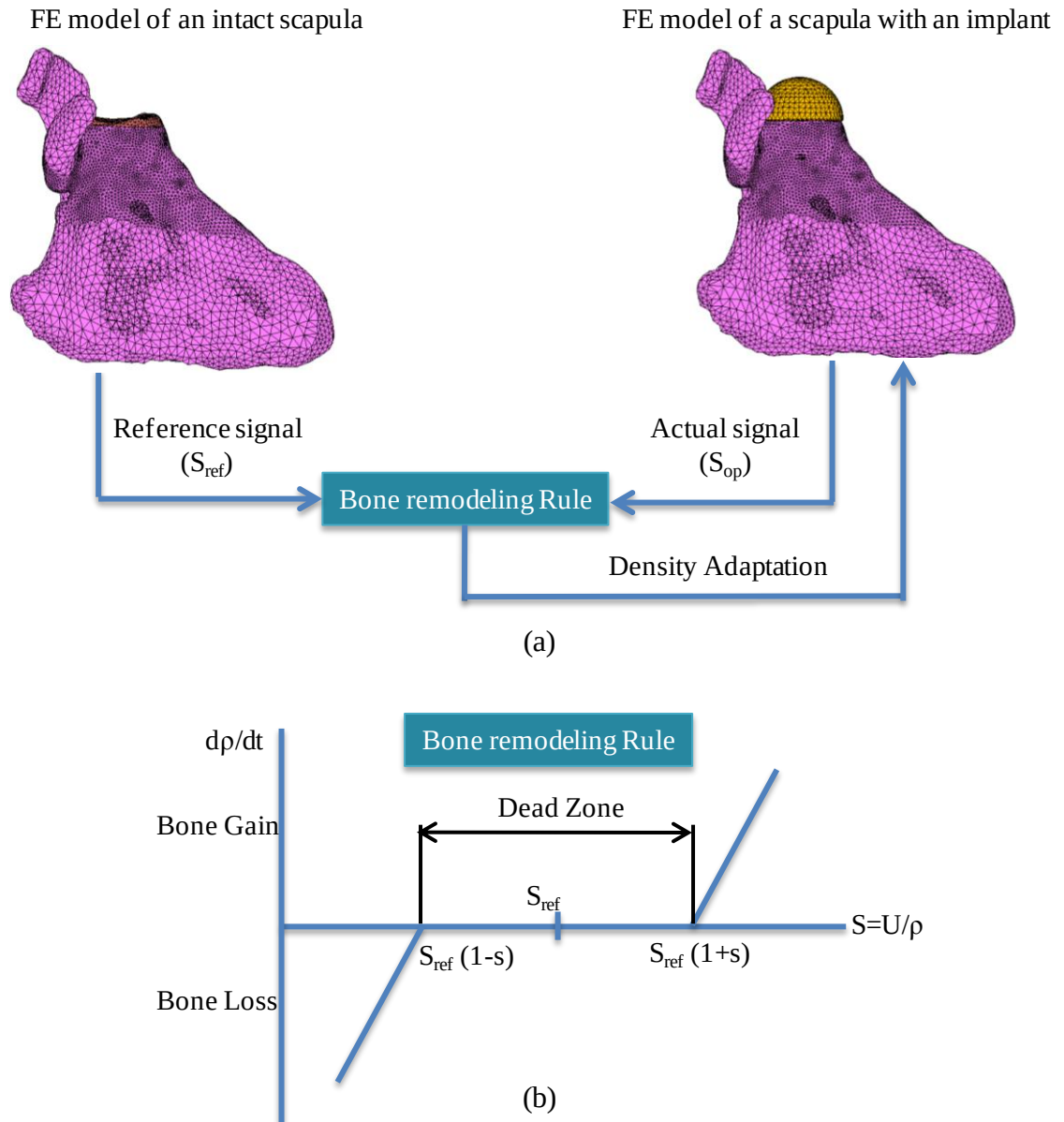


Figure 7-8 (a) Schematic representation of the bone remodelling algorithm implemented in FE simulations; (b) Strain-adaptive bone remodelling rule.

An 8-year follow-up anteroposterior radiograph of a patient treated with the Delta RSA obtained from European Hospital Georges Pompidou France (Figure 7-9) was used as a reference. The reason choosing this clinical radiograph is that this patient had significantly reduced shoulder function in the latest test (at 8-year follow-up) with a constant score of shoulder being only 45 points (full is 100 points). When the superior contour of predicted bone resorption was consistent with the pattern of radiographic bone loss, FE predicted bone densities at this time was assumed to be able to represent the realistic bone situations of this patient at the time when the radiograph was made.

The predicted bone density variations at that time were used to evaluate effects of bone adaptation on implant failure. Only the superior contour was used as a reference in this study, as factors for the generation of inferior bone erosion are very complicated, and may be associated with impingements of the humeral cup against the scapular neck and osteolysis which could be induced by polyethylene wear particles.

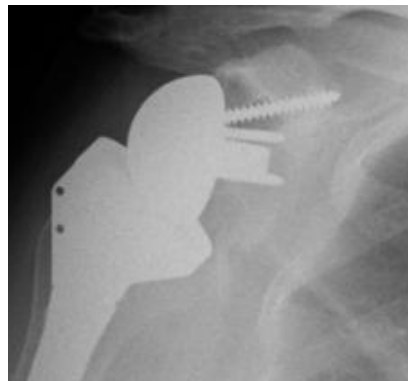


Figure 7-9 Radiograph of a patient treated with Delta RSA (obtained from European Hospital Georges Pompidou France).

7.3.3 Results

This study is aimed to simulate and predict effects of strain-induced adaptive bone remodelling on glenoid loosening. FEMs of intact scapula and the same scapula merged with the Delta CTA RSA were used to predict strain changes after orthopaedic implantation, and to investigate the correlation between adaptive bone remodelling and bone loss, which is frequently observed in the postoperative radiographs.

7.3.3.1 Correlation between Bone Adaptation and Bone Loss

Changes of bone apparent densities with time are illustrated in Figure 7-10. Bone resorption started from the middle peg and the resected surface, and then extended to peripheral regions. Additionally, it is noticed that bone resorption for the Delta RSA was observed around the end of the middle peg. With a reference to the bone loss pattern in the radiograph of a patient whose shoulder function reduced significantly (Figure 7-9), the superior contour of predicted bone resorption at iteration 17 (Figure 7-11) was consistent. The FE predicted bone densities at this iteration were use to

represent the realistic bone situations of the patient at the time when the radiography was made (8-year follow up).

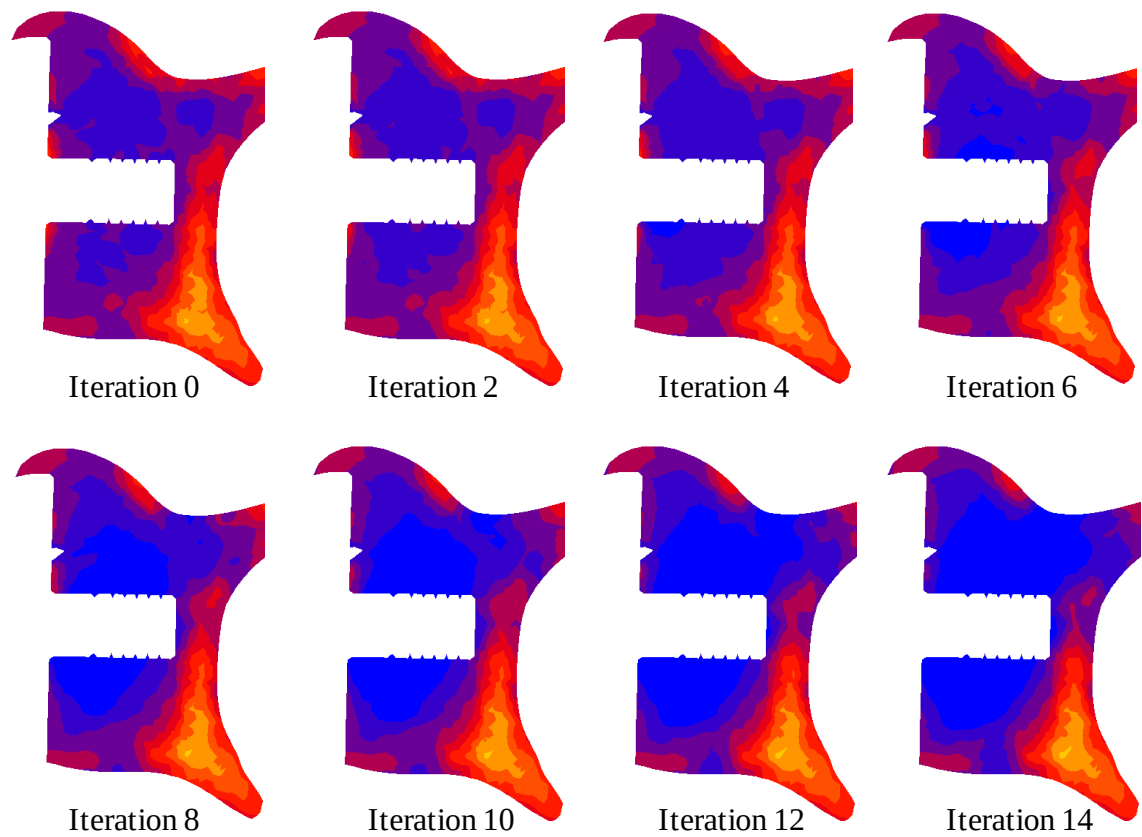


Figure 7-10 Progress of bone adaptation.

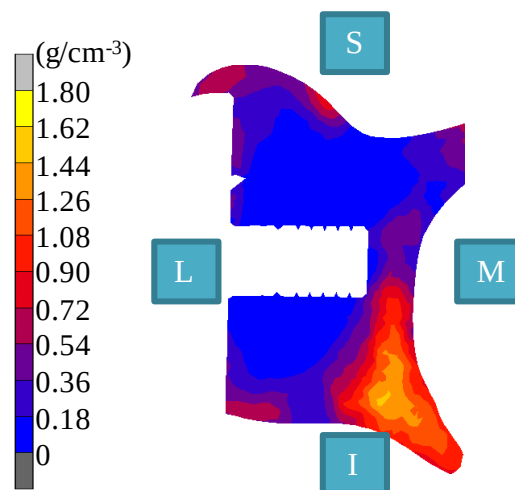


Figure 7-11 Predicted density distribution after bone adaptation at iteration 17.
(S/M/I/L stands for superior/medial/inferior/lateral).

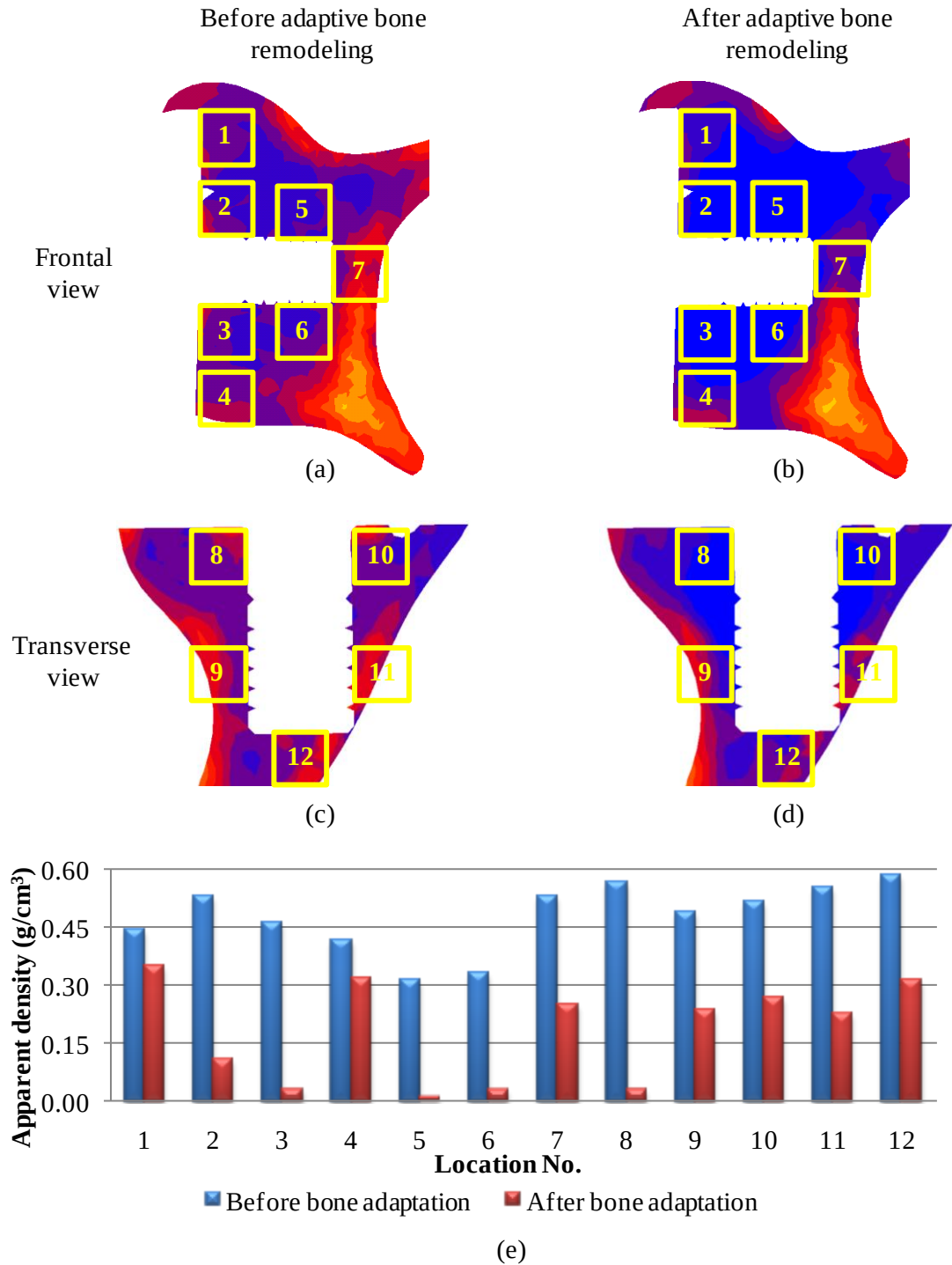


Figure 7-12 (a)-(d) 12 regions of interest on the frontal and transverse planes and (e) predicted apparent densities in these regions.

12 regions (4×4 mm squares), which were numbered from 1 to 12, were distributed on the frontal and transverse planes of the FEMs (Figure 7-12 (a)-(d)). Bone apparent densities on the nodes locating in each square were averaged and used to represent the density value in that region. Figure 7-12 (e) illustrates the predicted apparent densities

in each region of interest before and after the bone adaptation. The results showed that apparent densities close to the middle peg and resected surface (in the regions of 2, 3, 5, 6, 8, 10) reduced the most significantly at a percentage of 82.64% from iteration 0 to iteration 17. The mean density in these regions reached 0.08 g/cm.

Density variations at iteration 17 around the glenoid are illustrated in Figure 7-13. The results indicated that the most significant bone resorption occurred in the posterior glenoid, with the reduction of apparent densities of approximately 75%. Reductions of postoperative bone density in the superior and inferior regions were also very high at 63% and 69%. Due to the significant decrease of GH force loading upon the implant postoperatively (Terrier et al., 2008; Kontaxis, 2010), bone density at the peg tip also reduced after the prosthesis implantation, with a value of approximately 46% at iteration 17. The different amount of bone density reduction around the glenoid indicated that the force could transmit along the middle peg from the back of the metaglene to the peg tip and from the lateral posterior to anterior (arrow direction).

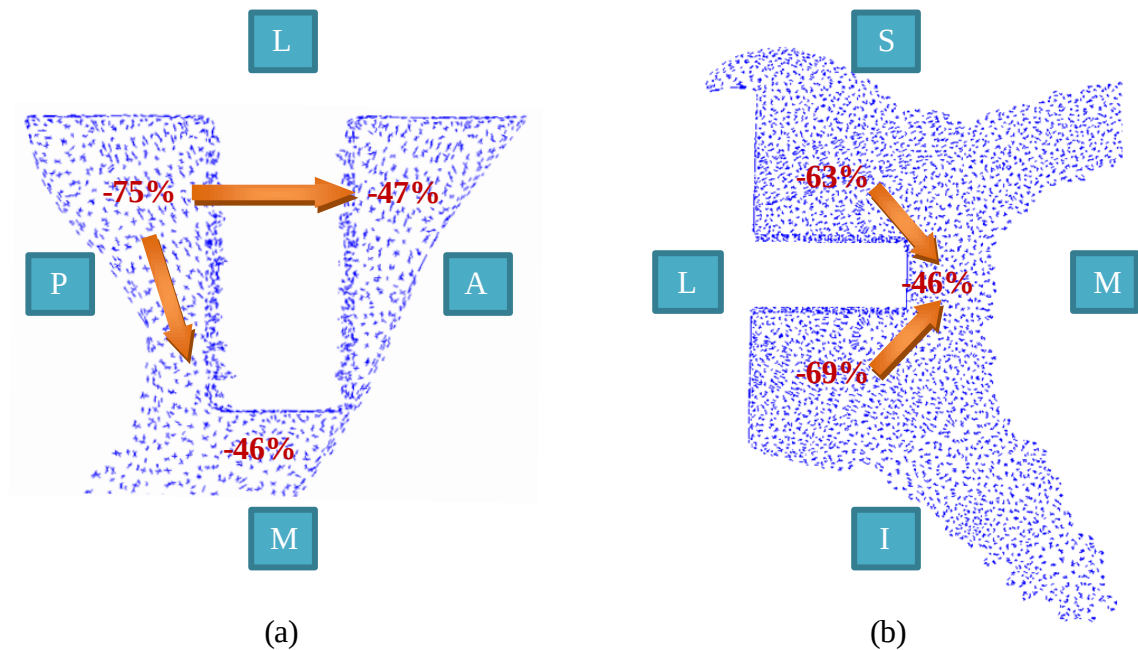


Figure 7-13 Reduction of bone apparent density in the regions of interest from iteration 0 to iteration 17.

7.3.3.2 Correlation between Predicted Bone Adaptation and Bone Ingrowth

Predicted bone density distributions in the case of two bone-prosthesis interface conditions: no bone ingrowth and full bone ingrowth at the same time (iteration 8) after bone adaptation were studied. Figure 7-14 illustrates the results at iteration 8. FEMs with these two fixation conditions showed consistent patterns of bone density distributions in the glenoid. High bone density reduction was observed in the region close to the middle stem and the resected surface. At a particular time after the orthopaedic operation, such as iteration 8, bone resorption for the full bone ingrowth case was 1.8% higher than that for the no ingrowth case, with the most significance being at the glenoid resected surface (3.6%).

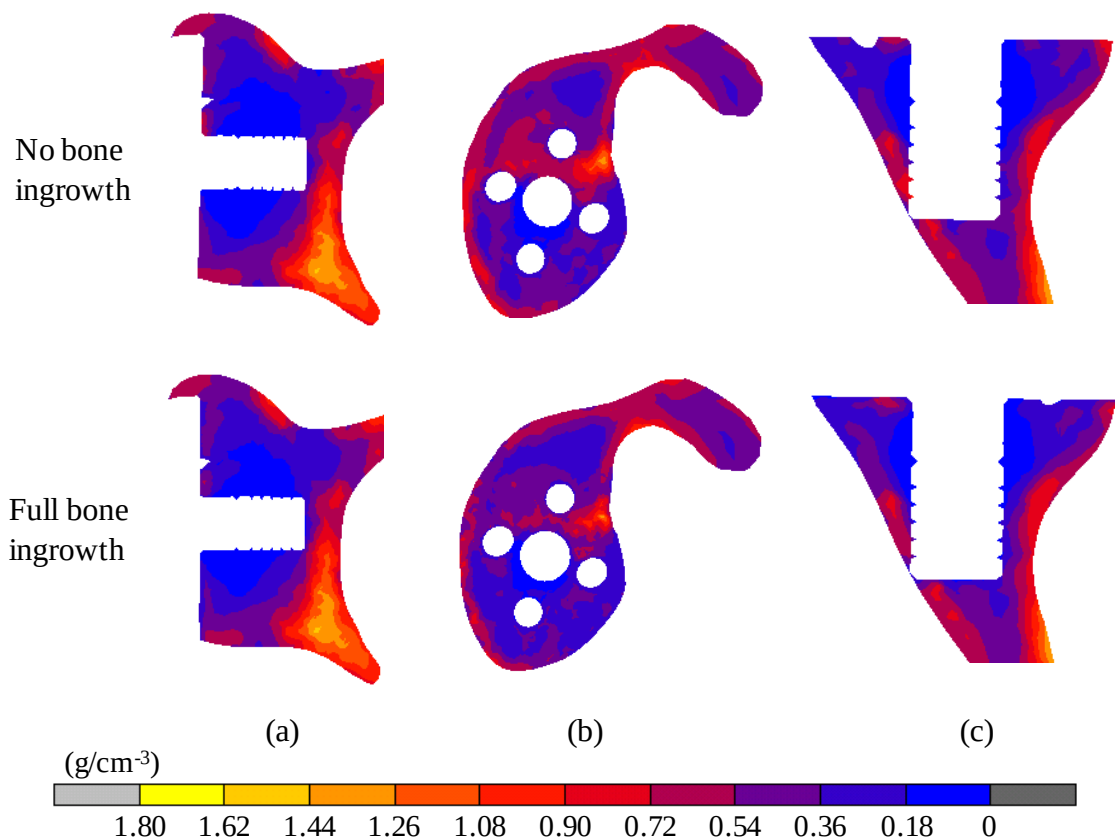


Figure 7-14 Bone density distributions at three surfaces: (a) frontal plane, (b) resected surface and (c) transverse plane in the case of no bone ingrowth and full bone ingrowth at iteration 8.

7.3.4 Discussion

The purpose of this study is to investigate stress shielding after glenoid prosthesis implantation and predict the effects of adaptive bone remodelling on the bone loss and the long-term fixation of glenoid prosthesis.

Bone apparent density variations with time were predicted in this study. Bone resorption occurred initially close to the middle peg and the resected surface, and then extended to peripheral regions. A radiograph (Figure 7-15 (b)) of a patient who was treated with Delta RSA was selected to evaluate the predicted bone resorption results (Figure 7-15 (a)). This radiograph was kindly supplied by Dr. Denny T. T. Lie from Singapore General Hospital. Figure 7-15 (b) clearly shows solid scapular neck contour as pointed by the orange arrow. The solid scapular neck contour demonstrates few impingements of the humeral cup against the neck, and indicates that inferior scapular notching and osteolysis induced by the polyethylene wear may less likely to occur. Dr. Denny T. T. Lie also mentioned that inferior scapular notching is not very often found in Singapore. In comparison with this postoperative radiograph for Delta CTA RSA (Figure 7-15 (b)), the predicted pattern of bone resorption (Figure 7-15 (a)) showed a good agreement. Both the FE and radiographic results displayed bone resorption/loss in the region close to the middle peg and the resected surface. In addition, it is noticeable that they both showed bone resorption at the end of tip. This finding was different from those observations from postoperative radiographs of knee joint prostheses, in which bone deposited close to the stem tip (Bertin et al., 1985). This phenomenon may be caused by the significant reduction of glenohumeral (GH) joint contact force after implantation of the Delta RSA (more explanations would be shown next). Both the predicted and radiographic results indicate that bone loss could occur even in the case of no apparent inferior scapular notch and osteolysis. The loss of supporting bone could come from bone resorption which was induced by the postoperative stress-shielding and reduction of postoperative GH forces. The consistence between the prediction and the radiograph indicates that the FEM of bone remodelling could provide believable results.

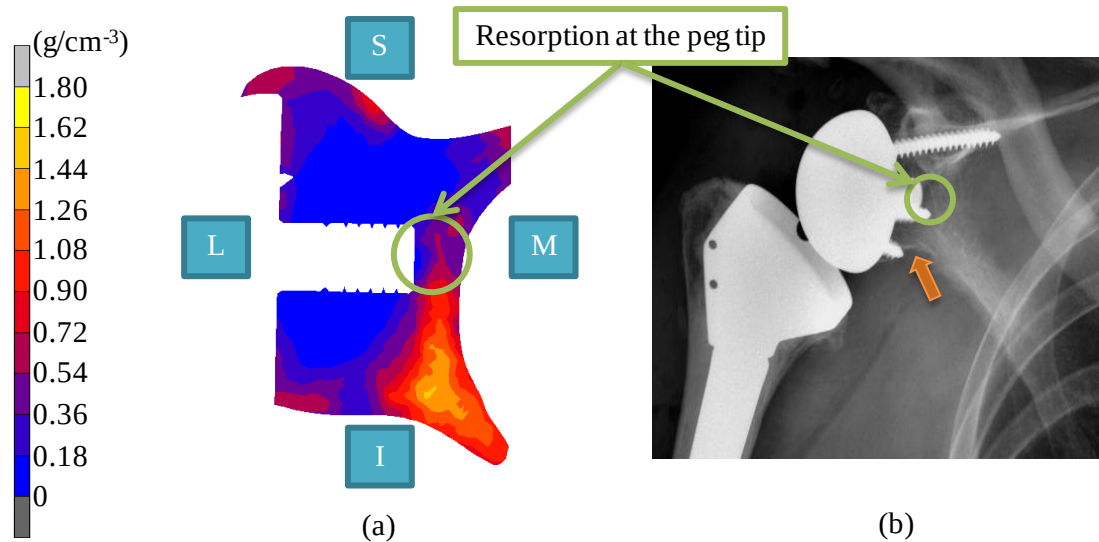


Figure 7-15 Comparison between (a) the pattern of predicted bone resorption and (b) the pattern of radiographic bone loss (obtained from Singapore General Hospital). The solid scapular neck contour was pointed by orange arrow.

A radiograph (Figure 7-9) of a patient who had the Delta CTA RSA for 8 years and had shoulder dysfunction (constant score of shoulder was only 45 points) was used to investigate the relationship between the bone remodelling and the prosthesis failure. In comparison with this radiograph, the predicted bone density distribution at the iteration of 17 showed the consistent superior contour. Thus, densities at this iteration were assumed to be able to represent the realistic bone situations of the patient at the time when the anteroposterior radiograph was taken. In comparison with the densities at iteration 0, when bone adaptation didn't occur, the values at iteration 17 (Figure 7-13) reduced significantly in the superior (63%), inferior (69%) and posterior (75%) regions and moderately in the anterior glenoid (47%) and at the end of stem tip (46%). The distribution of these density variations in the glenoid demonstrated a force transmission along the middle peg from the resected surface to the tip, and from posterior to the anterior glenoid. This prediction is consistent to the finding in our in-vitro test (Chapter 7.2), in which higher stress shielding occurred at the lateral than the medial level. In addition, this finding is also consistent with the clinical observation from the studies of Bertin et al. (1985), in which the bone resorption was severe in the proximal tibia. The 46% of bone density reduction at the stem tip could be due to the significant reduction of glenohumeral joint contact force (averagely 29.7% of reduction in the physiological

daily activities for the Delta RSA), as stress shielding may not lead to bone loss at the end of the middle peg (Bertin et al., 1985).

Figure 7-16 shows different inferior erosion patterns between the radiographs and FE results for the Delta RSA. Figure 7-16 (a) and (b) illustrate the inferior erosion with solid inferior scapular neck contour, which demonstrated few bone-prosthesis impingements and osteolysis induced by the polyethylene wear particles. In these two images, the bone under the middle peg is the weakest and becomes dense gradually to the inferior rim of the scapular neck. Figure 7-16 (c) and (d) display the inferior bone loss with erosive scapular neck contour. In these two graphs, bone is dense under the peg and gradually weakens to the inferior periphery. This finding indicates that generation of bone boss in the inferior glenoid is very complicated and could be an integrated work of inferior scapular notching, bone resorption and osteolysis.

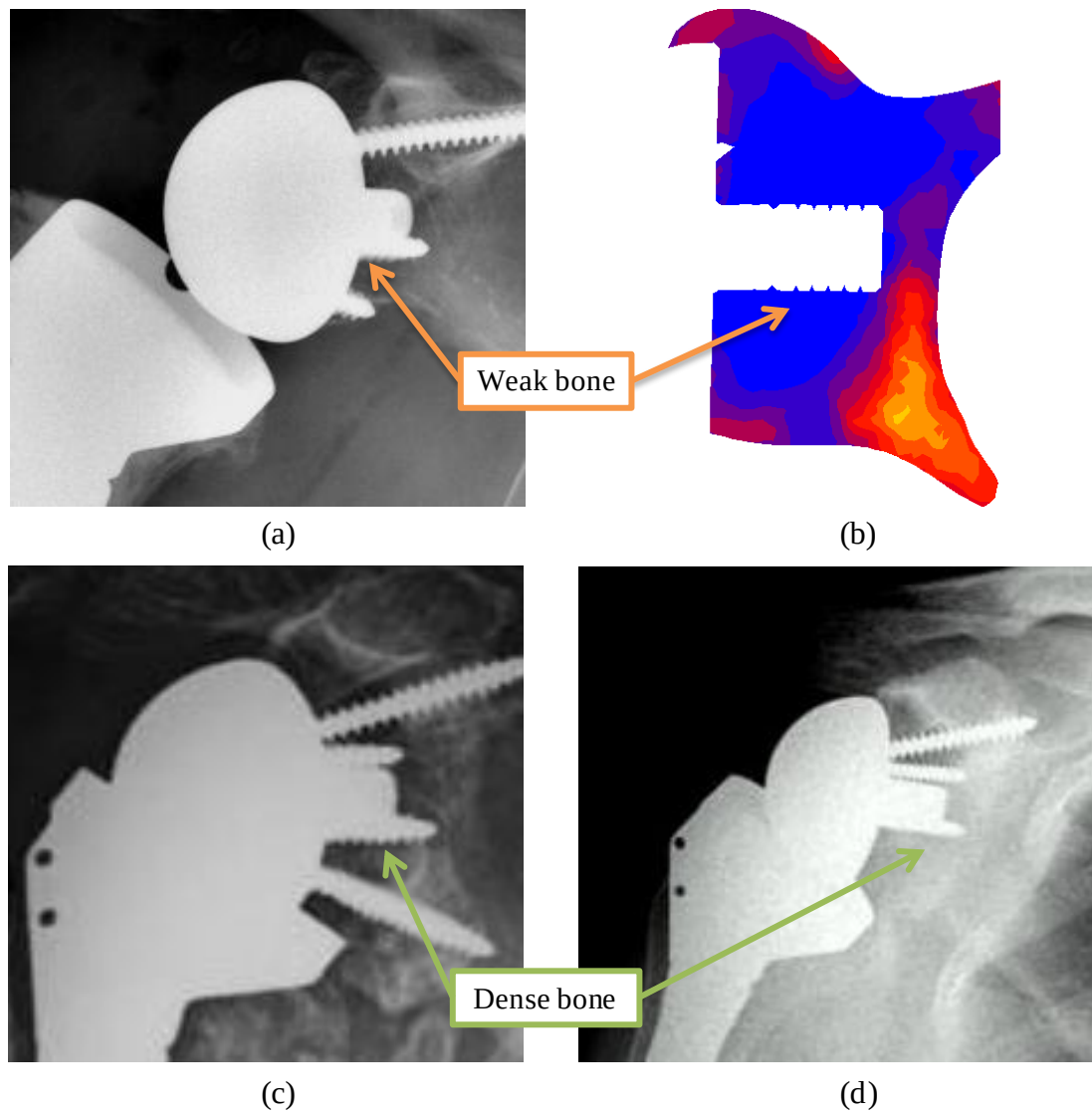


Figure 7-16 Comparison of different types of inferior bone loss: (a) a radiograph without notch (obtained from Singapore General Hospital); (b) FE modelling without notch in current study; (c) a radiograph with a severe inferior scapular notch (adapted from Grassi et al, 2009); (d) a radiograph with a big notch (obtained from European Hospital Georges Pompidou, France).

As found in our above results, bone resorption progresses from the back side of metaglene and the middle peg to the peripheral regions (Figure 7-17 (b)). The process of this type of bone loss may result in bone density reductions (Figure 7-17 (c)) and accelerate the generation of inferior scapular notching, as weak bone is more likely to break under the bone-prosthesis impingements. For the inferior scapular notching, it initiates from the inferior and lateral rim of the resected surface and developed to the medial and superior area (Figure 7-17 (a)). The process of this type of bone loss could

lead to increase of bone stresses/strains (See Chapter 6) and thus slow down bone resorption or even result in bone deposition.

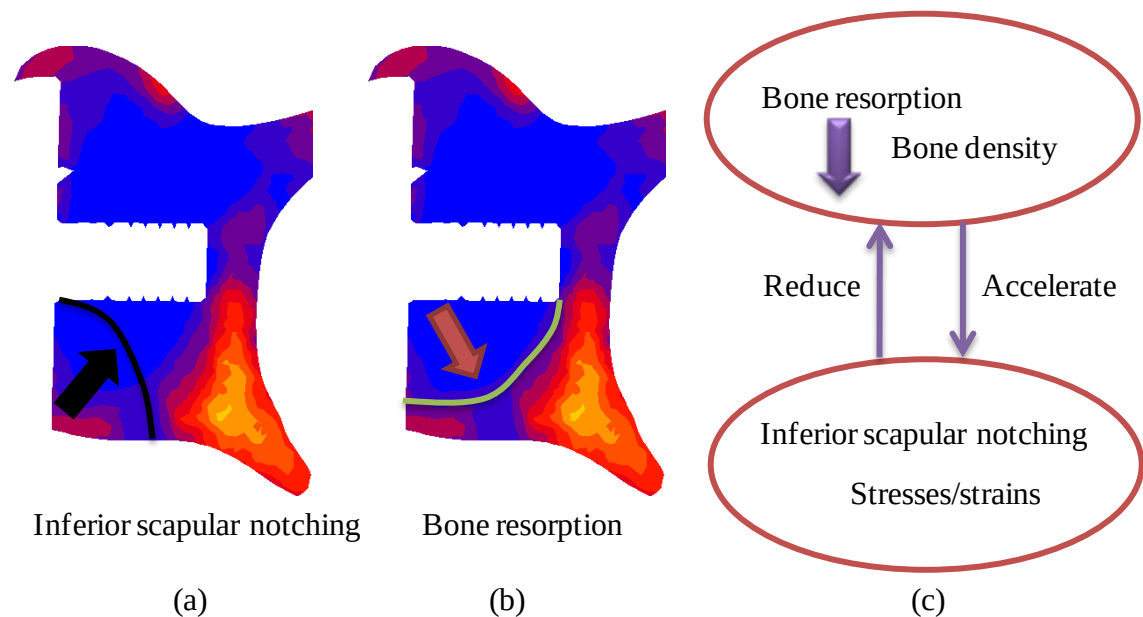


Figure 7-17 Developments and intercorrelation of inferior scapular notching and bone resorption: (a) development of inferior scapular notching; (b) development of bone resorption; (c) intercorrelation between inferior scapular notching and bone resorption.

Figure 7-18 shows the radiographic inferior bone loss in the case of severe inferior scapular notching. Three regions, marked as 1, 2 and 3, were chosen to analyse the factors which lead to the inferior glenoid bone loss. Bone loss in region 1, which is close to the origin of adaptive bone remodelling and far away from the origin of inferior scapular notching, may be predominantly caused by bone resorption and osteolysis. Bone loss in region 2, where is close to the origin of inferior scapular notching, is possibly predominantly subject to the impingement between the humeral cup and the scapular neck and osteolysis. Region 3 is far away from both origins of bone adaptation and scapular notching, thus bone loss in this region could be predominantly affected by the adaptive bone remodelling and osteolysis.

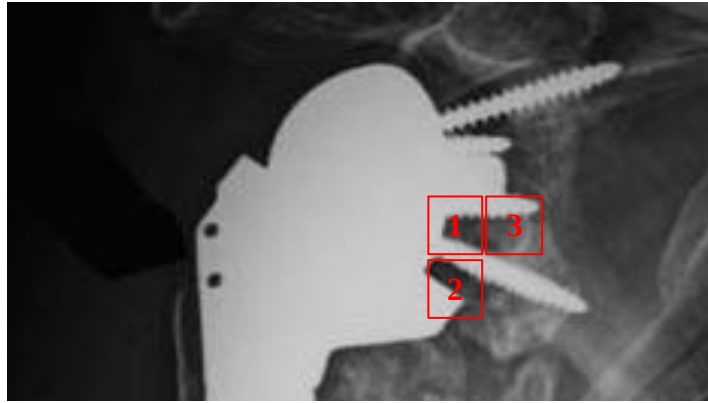


Figure 7-18 Location marks in a radiograph of patient with severe inferior scapular notching (adapted from Grassi et al, 2009).

Variations of the bone-prosthesis interface conditions (full bone ingrowth and no bone ingrowth) were found to influence the stress shielding and load transmission capability after the glenoid prosthesis implantation. Great amount of bone resorption was observed in the condition of bonded interface (full bone ingrowth). This may be because the direct connection at the bone-prosthesis interface could be able to transmit high GH joint load medially, and thus relieve the lateral bone stresses, consequently result in high lateral bone resorption. In the case of no bony ingrowth, the shear stresses transmitting across the contact surfaces were reduced by the frictions at the bone-prosthesis interface, and thus lead to comparatively small amount of bone resorption. The effect of interface conditions on the bone remodelling was also investigated in a femoral fixation study. It was found that bone resorption was less extensive when the interface was unbonded (van Lenthe et al. 1997; Barink et al. 2003).

Though the bone remodelling simulations after the Delta RSA implantation have shown good correlations with the radiograph of patients treated with this type of shoulder prosthesis, some specific limitations should be mentioned. The time process in the strain-adaptive bone remodelling algorithm, which was used in this study, has already been validated with the clinical DEXA scan results in the hip replacement studies (Kerner et al., 1999; Bitsakos, 2005). However, it has never been correlated to the patient-specific scans of scapula associated with shoulder joint replacements. As a result, the superior bone loss contour in a radiograph of a patient who was treated with Delta RSA and had shoulder dysfunction at 8-year follow up was used as a reference in this study. The time when the predicted superior contour of bone resorption was the

same to the radiographic pattern was selected, and bone density at this time was assumed to be able to represent the reality when the radiograph was taken. It would be valuable to create a time constant which could relate the simulated time to the physical time, based on in vivo and in vitro studies for shoulder joint. Additionally, it would be better to validate our FE bone adaptation predictions with DEXA scan, which is the “gold standard” for bone density quantification in clinical practice.

7.3.5 Conclusion

The purpose of this study was to investigate effects of bone resorption on postoperative bone loss which has been observed in many radiographs of patients treated with Delta RSA, and may play an important role in the long-term glenoid loosening. In comparison with the clinical radiograph in the case of no apparent inferior scapular notch, the contour of predicted bone resorption in terms of the adaptive bone remodelling algorithm was consistent. This agreement between the prediction and the radiograph indicates that the FEM of bone remodelling could provide believable results.

In comparison with the bone densities before bone adaptation, there was a significant reduction in the region close to the resected surface and the middle peg. This finding was consistent to the observation for the Delta RSA (Grassi et al., 2009), in which considerable bone loss occurs at the back side of the metaglene. In addition, the results also demonstrated that strain-induced adaptive bone adaptation is likely to occur after glenoid prosthesis implantation.

In this study, the radiograph of a patient who had dysfunction shoulder at an eight-year follow up was used as a reference to investigate the relationship between bone remodelling and the prosthesis failure. Predicted bone apparent density at iteration 17 showed a consistent contour of bone loss with that in the radiograph. Thus bone conditions at this time were assumed to be able to represent the real bone condition when the radiograph was taken. In comparison with the bone density before bone adaptation, the results at iteration 17 significantly reduced in the superior (63%), inferior (69%) and posterior (75%) regions and moderately decreased in the anterior glenoid (47%) and at the end of stem tip (46%). This density variation distribution in

the glenoid demonstrated a force transmission along the middle peg from the resected surface to the tip, and from posterior to the anterior glenoid.

This study also shows that there were differences in patterns of inferior bone loss with and without notching. In the case of less or no inferior notches, the bone under the middle peg is the weakest and becomes dense gradually to the inferior rim of the scapular neck. In the case of inferior notches, bone is dense under the peg and weakens to the inferior periphery. Factors for the inferior bone loss are complicated. They could be integrated effects of postoperative adaptive bone resorption, impingement of humeral cup against the scapular neck, and osteolysis induced by the polyethylene wear particles.

CHAPTER 8

ANALYSIS OF THE RELATIONSHIP BETWEEN GLENOID PROSTHESIS POSITIONS AND FIXATIONS

8.1 Introduction

Inferior scapular notching is one of the most frequently reported postoperative radiographic complications for Grammont reverse shoulder arthroplasties (RSA). It is initially caused by the impingement of inferior rim of polyethylene humeral cup against the scapular neck in a resting position or during arm adduction. Inferior scapular notching is associated with about 50% to 96% of the Delta prostheses (Sirveaux et al., 2004; Boileau et al., 2005; Lévine et al., 2008) and several postoperative clinic outcomes (Constant and Murley, 1987; Nyffeler et al., 2004; Sirveaux et al., 2004;

Boileau et al., 2005; Simovitch et al., 2007). Additionally, in Chapter 6, it was reported that inferior scapular notching resulted in negative effects on the fixation of glenoid prosthesis in Delta RSA. Therefore, minimizing the impingement between the humerus prosthesis and the scapular neck has been of particular interest (Frankle, 2005, 2006; Humphrey et al., 2008; Kelly II et al., 2008).

Alterations of prosthetic designs and various implant positions have been proposed to reduce postoperative scapular notching in RSA (Nyffeler et al., 2005; Humphrey et al., 2008; Kelly II et al., 2008; Kontaxis, 2009). It was found that lateralising the offset of the centre of rotation and varus neck-shaft angle were able to effectively decrease bone-prosthesis impingements and scapular notching (Gutiérrez et al., 2008; Kontaxis, 2009). In addition, Kontaxis (2009) predicted that 5 mm offset of the centre of rotation could averagely reduce approximately 50.8% of inferior contact between the humeral cup and the scapular neck. In a study of correlation between bone-prosthesis impingements and the position of glenosphere, Nyffeler et al. (2005) reported that inferior positioning could not only reduce the inferior and superior impingements, but also improve the arm motion range. In their study, four different fixation configurations of glenoid prosthesis shown in Figure 8-1 were tested: (a) glenosphere didn't cover the inferior glenoid rim; (b) glenosphere flushed with the inferior glenoid rim; (c) glenosphere extended beyond the inferior glenoid rim; and (d) glenosphere tilted downward 15°. The inferior overhang (Figure 8-1 (c)) and downward inclination (Figure 8-1 (d)) of the glenosphere could effectively improve the arm adduction for 24° and 16° respectively without abutment of the polyethylene cup against the scapula. However, downward inclination of the glenosphere resulted in a reduction of the glenoid pole, thus only inferior positioning (configuration 3) is recommended to minimize the bone-prosthesis impingements without altering the original prosthetic geometry (Nyffeler et al., 2005; Kontaxis, 2010).

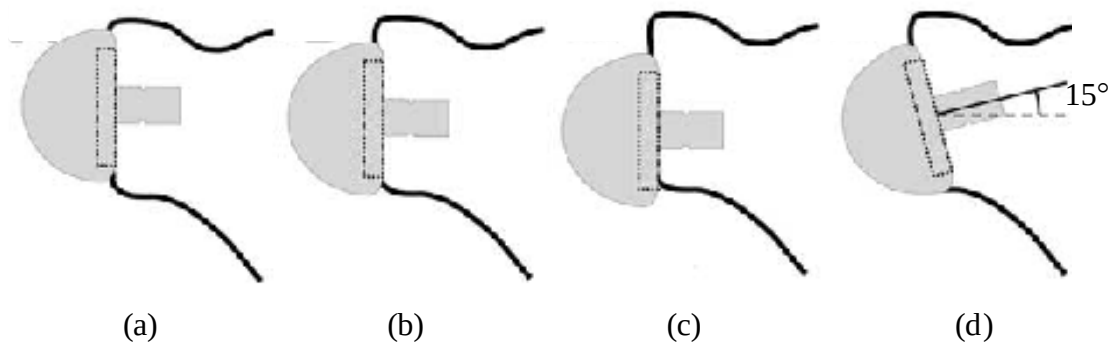


Figure 8-1 Four different fixation configurations of glenoid prosthesis (adaped from Nyffeler et al., 2005).

Since the study of Nyffeler et al. (2005), geometry of the overhang of glenoid prosthesis with relative to the inferior glenoid rim has been of great interest (Simovitch et al., 2007; Gutiérrez et al., 2008; Kelly II et al., 2008; Kontaxis, 2010). The recommended values range from 2 mm to 6 mm (Nyffeler et al., 2005; Kelly II et al., 2008). In a study of the Delta III RSA with 8 cadaveric human scapulae, Nyffeler et al. (2005) reported that placing the glenoid component 2 to 4 mm more inferior than the standard surgical technique could improve the range of arm motion and reduce the inferior scapular notching. In another similar study of Aequalis RSA (Tornier), the optimized overhang of the glenoid prosthesis was approximately 6 mm to the inferior glenoid rim (Kelly II et al., 2008).

Even though glenoid prosthesis repositioning proposed in the study of Nyffeler et al. (2005) may minimize the impingement problem of scapular neck and reduce the development of bone notches, effects of this modified surgical technique on the glenoid prosthesis fixation is still unclear. The stable long-term implant incorporation is associated with not only the scapular notching, but also many other factors, for instance the postoperative strain-induced adaptive bone remodelling (described in Chapter 7) and osseointegration.

The purpose of this chapter is to investigate the correlation between implant positions and the long-term fixation of glenoid prosthesis. Four glenoid prosthesis positions proposed in the study of Nyffeler et al. (2005) (Figure 8-1) were studied. The fixation of glenoid prosthesis for each position configuration was evaluated by two independent

factors: initial stability of the glenoid prosthesis and postoperative strain-induced adaptive bone remodelling. To simulate the real cases, five physiological daily activities were used in this study.

8.2 Materials and Methods

Four FEMs were created based on CT images of the same cadaveric scapula. Figure 8-2 shows the glenoid geometry used in current study. An FEM of the Delta CTA RSA was also modelled and virtually implanted into the four FEMs of scapula under the guidance of an experienced orthopaedic shoulder surgeon (G.T.) and according to the implantation procedure described in Chapter 5.3.2.1. Positions of the glenoid prosthesis relative to the inferior glenoid rim or the articular glenoid surface are illustrated in Figure 8-3. In the first scapula model (Figure 8-3 (a)), the glenoid prosthesis was positioned in terms of the conventional surgical technique, being in the middle of the glenoid articular surface. In the second model (Figure 8-3 (b)), the glenosphere just covered the inferior rim of the glenoid resected surface. In the third model (Figure 8-3 (c)), the glenosphere was moved as inferiorly as possible before the inferior screw graphically run out of the bone. It should be mentioned that the superior and inferior locked screws in the Delta CTA RSA form a fixed angle of 17° to the axis of the middle peg. In this FEM, the overhang of the glenosphere to the inferior glenoid rim was 3.9 mm. This value was in the recommended range (2 to 4 mm) of inferior positioning of the glenoid prosthesis in Delta RSA (Nyffeler et al., 2005). In the last scapula model (Figure 8-3 (d)), the glenoid component was tilted downward a recommended value proposed by Nyffeler et. al (2005), being approximately 15° . The inferior inclination of glenoid prosthesis was achieved by performing an oblique osteotomy with resection of the inferior glenoid pole.

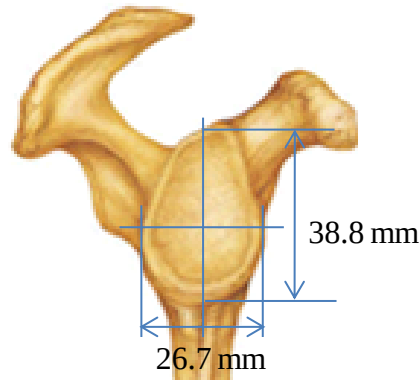


Figure 8-2 Geometry of the glenoid which was modelled in this study.

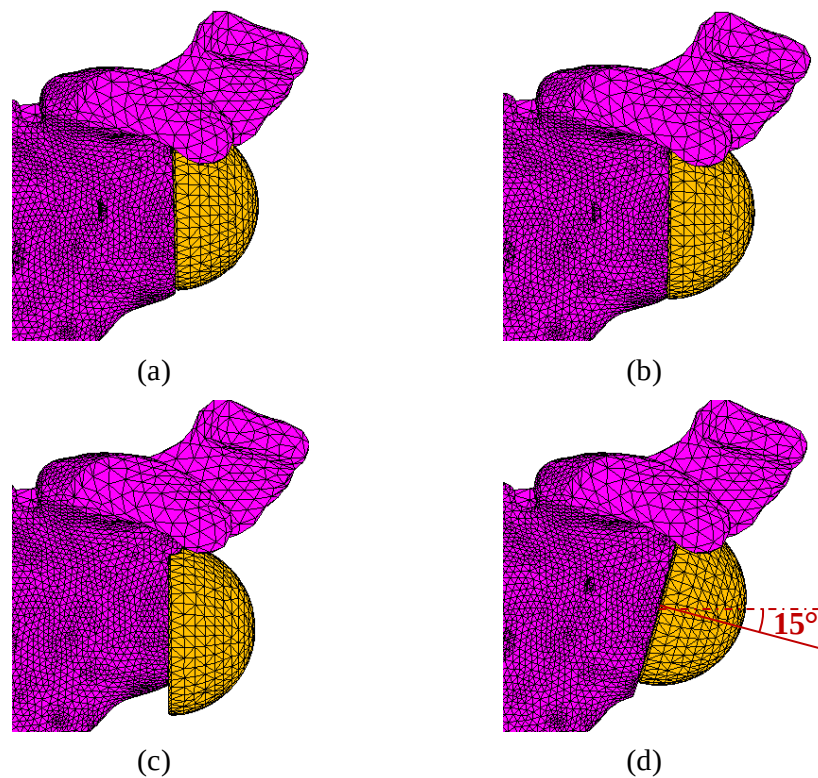


Figure 8-3 Four fixation configurations of glenoid prosthesis in FEMs.

The FEM of an implanted scapula was created with the same method introduced in Chapter 4. Loading conditions of five physiological daily activities (Table 4-4) were tested. Force magnitudes and vectors were the same irrespective of implant positions. It is noticeable that the coordinate system for the GH force in the case of fixation configuration 4 was created relative to the glenoid prosthesis without downward inclination (Figure 8-4). Bone-prosthesis micromotions and postoperative adaptive bone resorption were predicted in each fixation configuration case. In this study, the unbonded interface between the bone and the implant was modelled by defining the two

parts to be in contact with a Coulomb's friction of 0.4 (Hopkins et al., 2008). Peripheral titanium screws were assumed to be secured firmly and thus the bone-screw interface was modelled as rigidly bonded. In the study simulating the postoperative adaptive process of bone, the strain-induced bone remodelling algorithm used in Chapter 7 was employed.

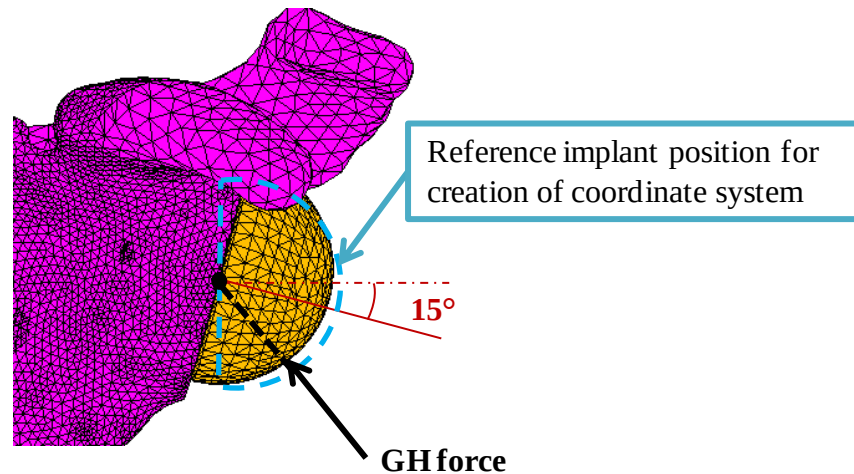


Figure 8-4 Reference position of the glenoid prosthesis for the creation of coordinate system in the case of fixation configuration 4.

8.3 Results

8.3.1 Bone Remodelling Predictions for Various Glenoid Prosthesis Positions

As a result of varied time steps used in the iterations in our strain-induced adaptive bone remodelling algorithm, postoperative bone apparent density distributions at the duplicated follow ups for the four glenoid prosthesis fixation cases were recorded. Figure 8-5 shows developments of bone resorption and densification in the frontal plane at five various periods after the orthopaedic operation. These five periods are named as T0 to T4. T0 represents the period before bone adaptation, and T1 to T4 stand for the periods during the adaptive bone remodelling process. The four fixation position cases shown in Figure 8-5 were named as FP 1 to 4. The results showed that great bone resorption was predicted to occur in the periprosthetic regions, close to the back of metaglene and the middle peg for all the implant fixation position cases. Differences

were that bone resorption developed almost evenly above and under the middle peg for the FP 1 and 4 cases, while it predominantly appeared above the middle peg for the FP 2 and 3 cases. Another finding was that apparent bone resorption for the FP 3 case was observed at T1, which was earlier than those for other fixation cases. Figure 8-6 illustrates the predicted percentage of bone resorption in the frontal plane of the glenoid for the four fixation cases during the bone adaptation process. Positioning the glenoid prosthesis the most distally led to the most amount of bone density reduction in comparison with other fixation cases at the same time in the adaptive process of bone remodelling. For instance, in the initial period after the bone remodelling, 4.5% of bone resorption in the frontal plane for FP 3 was observed, followed by the case of FP 2, being 2%. The percentages of bone resorption for both cases of FP 1 and 4 were very low, at 0.5% and 0.4%, respectively. Developments of postoperative bone resorption were observed for all the fixation cases, with the rapidest growth being in the case of FP 3.

Bone density variations on the resected surface and at the end of peg tip were calculated in order to investigate transmissions of bone densities for all the fixation cases. Since bone resorption increased during the process of bone adaptation, results at T1 (the initial period) and T4 (the period when severe bone resorption were observed for all the fixation cases), are shown in Figure 8-7. At the resected surface, both FP 3 (overhang of the glenoid prosthesis) and FP 4 (inferior tilting of the implant) led to large amount of postoperative bone density reduction during the adaptive process of bone remodelling (40.2% and 27.8% for the two fixation cases at T4). In the region close to the peg tip, the amount of bone resorption for FP 3 was the smallest in comparison with the values for all other fixation cases, being 0.2% at T1 and 5.2% at T4. Even though the percentage of bone density reduction for FP 1 was comparatively small at the resected surface, the value was the largest in the region close to the stem tip (1.05% at T1 and 13.2% at T4).

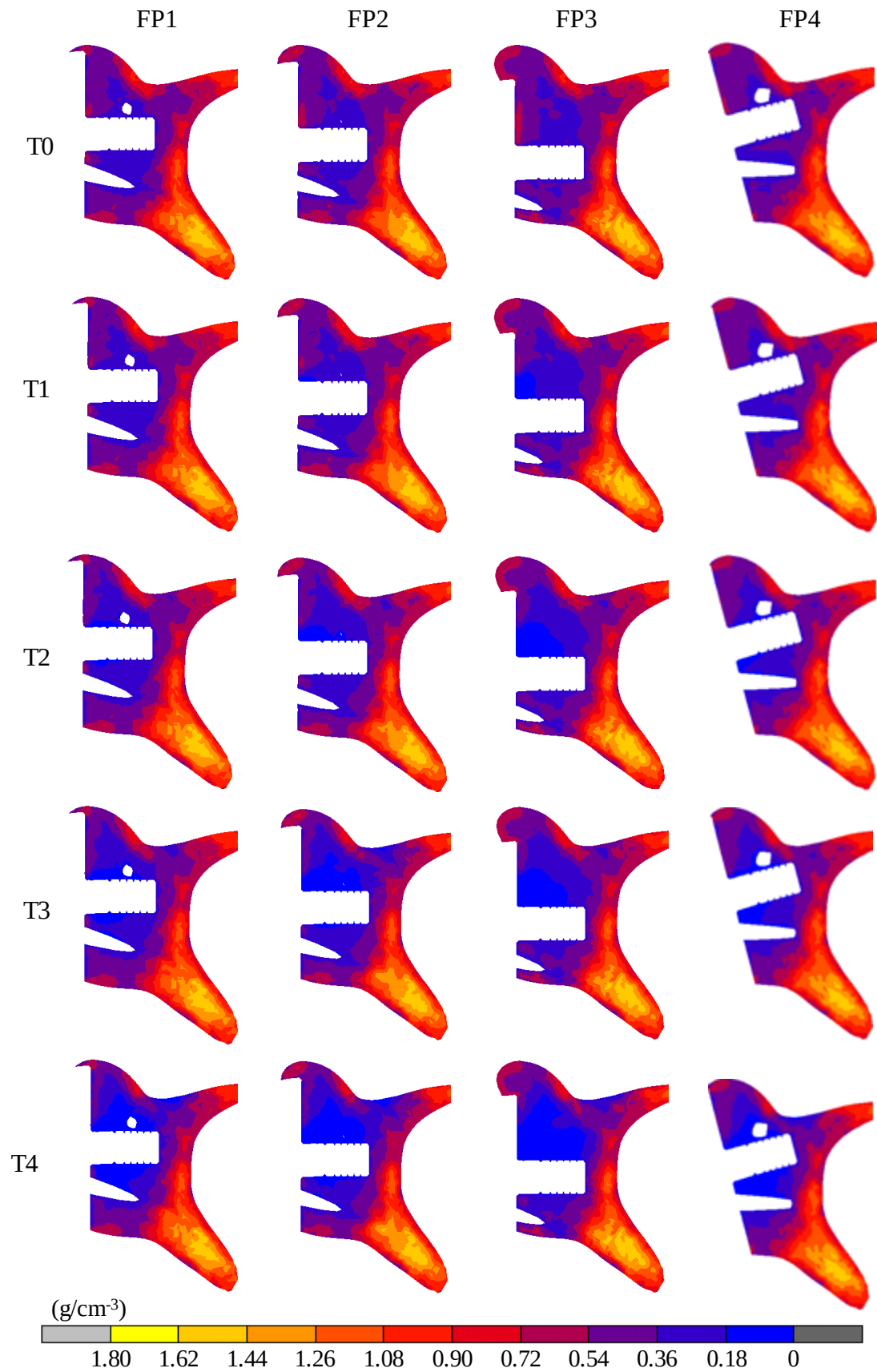


Figure 8-5 Bone remodelling process on the frontal plane of glenoid for various fixation configurations (FP: fixation position).

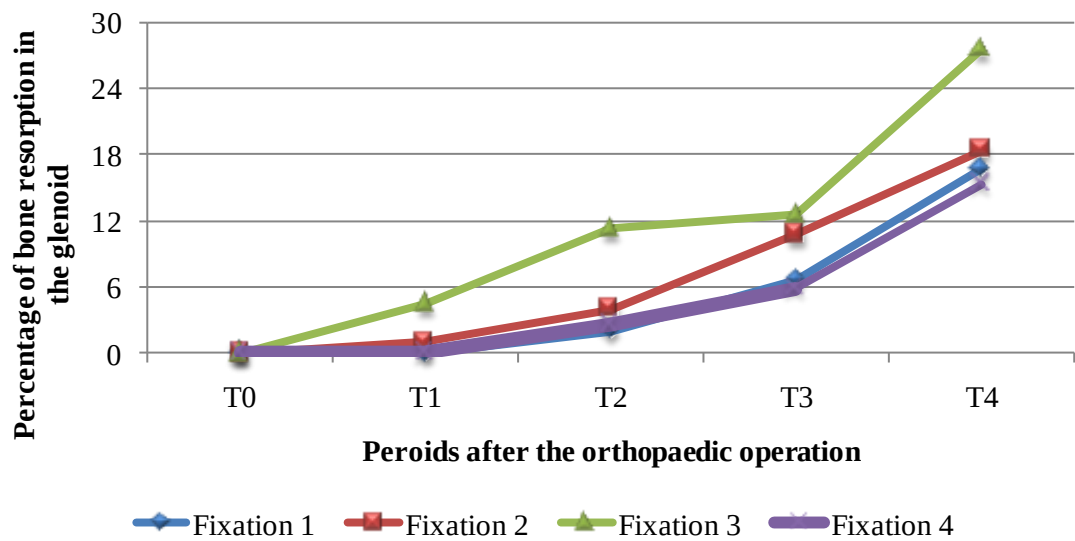


Figure 8-6 Predicted percentage of bone resorption in the glenoid for all the fixation cases during the period of bone remodelling.

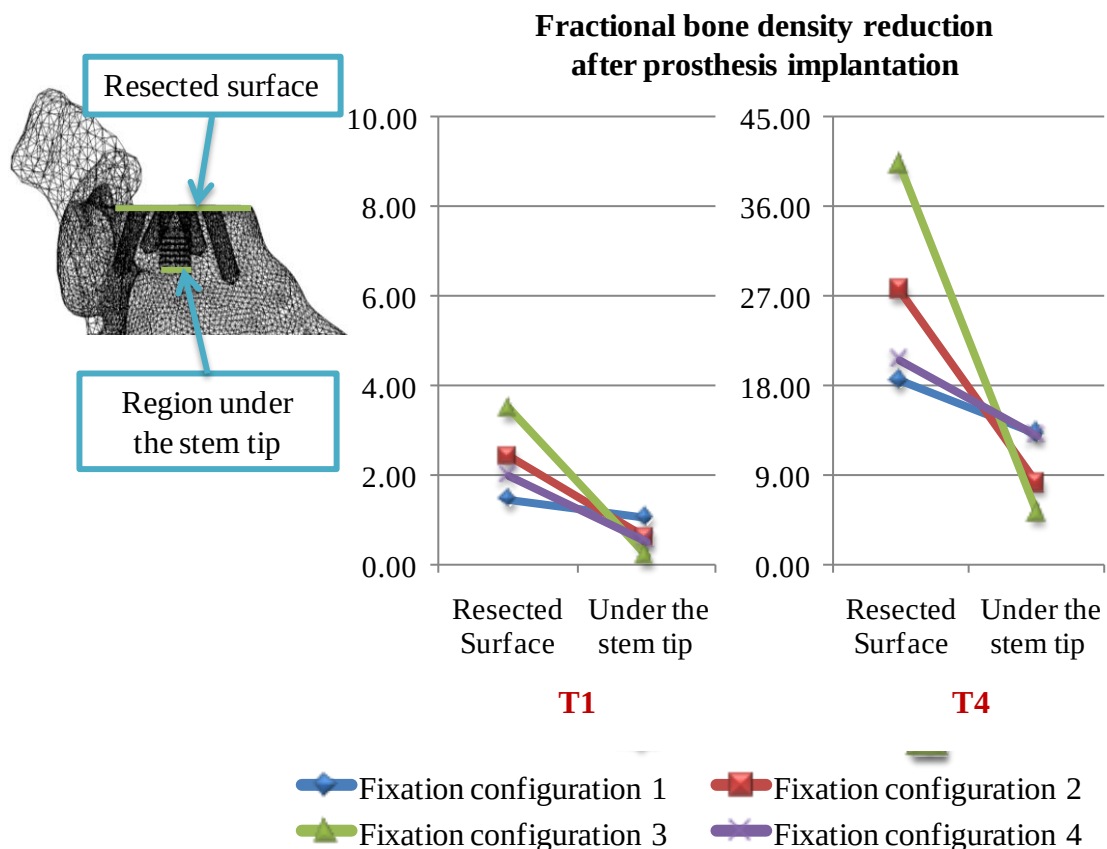


Figure 8-7 Predicted fractional bone density reduction at the resected surface and in the region under the middle stem tip for various fixation configurations at two follow ups of interest after the adaptive bone remodelling.

8.3.2 Initial Stability Prediction for Various Glenoid Prosthesis Positions

Predicted micromotions at the bone-prosthesis interface for the four glenoid prosthesis positions were studied. Results for the load case of T12 (standing up from an arm chair) are illustrated in Figure 8-8. The green circles represent contour of the glenoid component of Delta RSA. To show better the intact resected surface, the scapula in the case of glenosphere inclination (configuration 4) (Figure 8-8 (d)) was superiorly rotated 15°. The results showed that high micromotions at the interface between the bone and the load-bearing implant predominantly appeared at the end of the middle stem and the inferior and superior rims of glenoid resected surface. In comparison with the conventional surgical technique (Figure 8-8 (a)), the 3.9 mm inferior shift of glenoid prosthesis (Figure 8-8 (c)) led to slightly increase of micromotions at the superior rim of glenoid resected surface. The downward inclination of implant resulted in considerable increase of micromotions, with the significance in the inferior portion at the metaglene-to-bone interface.

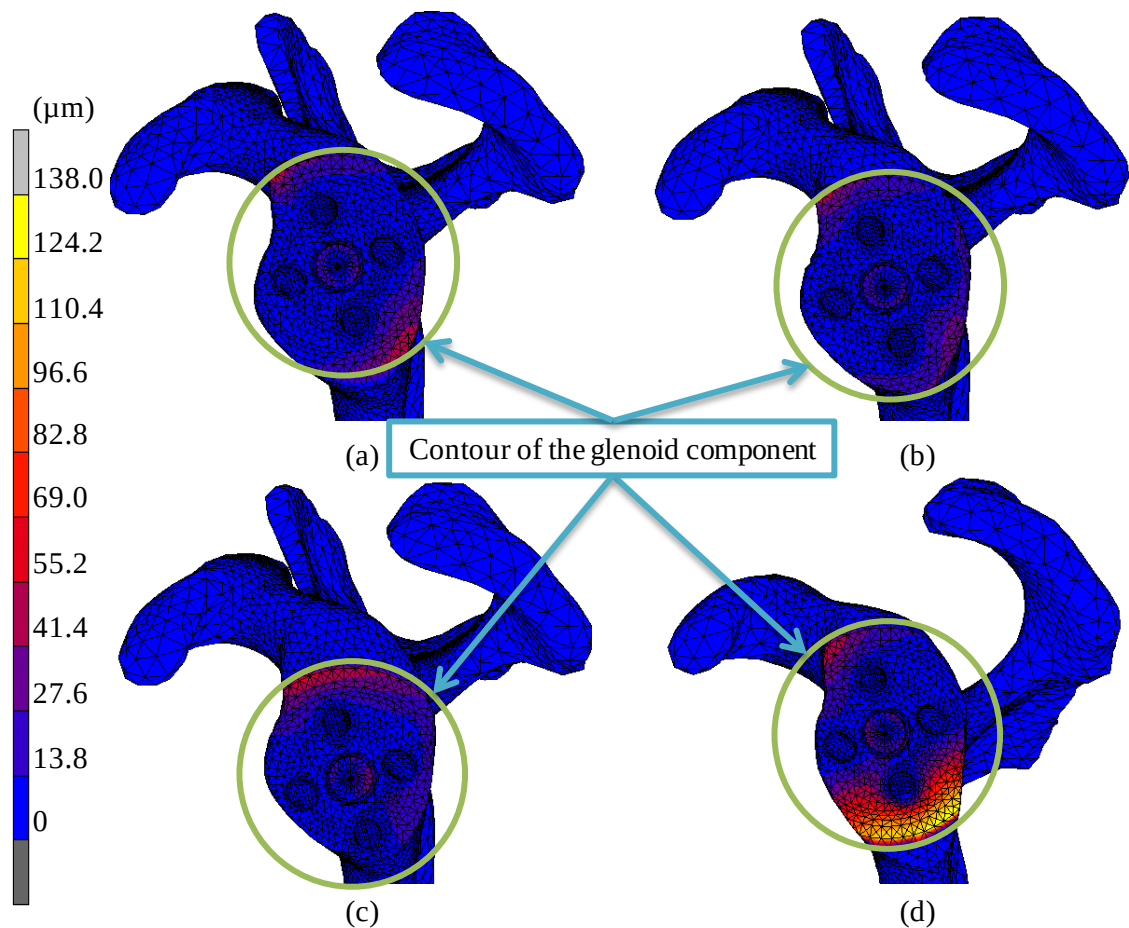


Figure 8-8 Micromotion distribution at the resected surface in the same load case (T12) for the four fixation configurations; (a)/(b)/(c)/(d): fixation configuration 1/2/3/4.

Figure 8-9 illustrates predicted peak bone-prosthesis interface micromotions for the four glenoid prosthesis position cases under the five force conditions (Table 4-4). The results showed that micromotions in the case of configuration 4 (downward inclination of the glenoid prosthesis) were higher (an average of 53.5 μm for the five load cases) than those for other implant position cases, noticeably in task 9 (82.5 μm) and task 12 (137.4 μm). This fixation position may lead to poor initial and second implant stability, when relating the predicted peak micromotions with the required threshold (50 μm) for osseointegration and bone ingrowth (Pilliar et al., 1986). For the case without any inclinations of implant (fixation configuration 1, 2 and 3), predicted micromotions were mostly lower than 50 μm , thus a good initial stability of the glenoid component is indicated. Another observation is that micromotions during physiological daily activities reduced with the inferior positioning of glenoid prosthesis for most load cases, with an average of all the load cases of 27.4 μm in configuration 1, and 25.2 μm in

configuration 2. While with the overhang of glenoid prosthesis, micromotions at the superior rim of glenoid resected surface increased, thus the predicted peak value increased slightly to 26.6 μm for the five load cases, noticeably in task 9 and 12. The results from Figure 8-9 also showed that inferior inclination of glenoid prosthesis may lead to big effects on the bone-prosthesis micromotions, with the significant increase of 55.1 μm in task 9 and of 72.2 μm in task 12. An exception was noticed in task 10, for which the predicted peak micromotions for configuration 1, 2 and 3 were exactly the same, being 21.3 μm , and the value in the case of fixation position 4 reduced 1.0 μm .

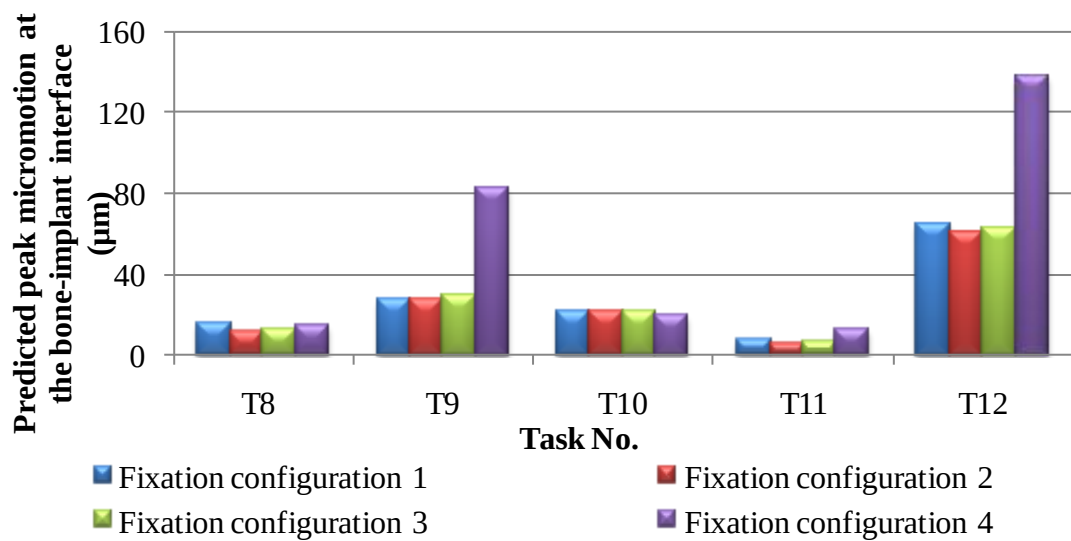


Figure 8-9 Predicted peak micromotions at the bone-prosthesis interface for the four fixation configurations under five loading conditions.

8.4 Discussion

Inferior positioning and downward inclination of glenoid prosthesis are proposed to minimize scapular notching, which is concerned due to the high postoperative radiographic incidence and the possible correlation with the glenoid loosening. However, since these altered implant positions were first proposed in 2005 (Nyffeler et al.), effects of these position modifications on the long-term implant fixation was still unknown. The purpose of this study is to predict the correlation between positions of the glenoid prosthesis in Delta RSA and the long-term implant fixation. In this study, the fixation of glenoid prosthesis was assessed by two factors: bone resorption caused by stress shielding and initial stability.

8.4.1 Bone Remodelling Predictions for Various Glenoid Prosthesis Positions

Bone density distributions in the frontal plane under various loading conditions (Figure 8-5) indicated that bone loss occurred and developed in the same tendency for all the four fixation positions cases. The loss of supporting bone starts from the region close to the resected surface and the middle peg and extends to the peripheral regions. The results also displayed that bone resorption in configuration 1 and 4 (the implant was located in the middle point of the glenoid articular surface) almost evenly distributed above and under the middle stem. While, the amount of bone resorption in the inferior positioning cases (configuration 2 and 3) was greater above than under the middle stem. This result could be induced by the inferior shift of glenohumeral (GH) joint contact force after the implantation of Delta RSA. As shown in Figure 8-10, the GH force loading around the glenoid fossa in the anatomical shoulder was shifted by a distance (being represented with “d”) inferiorly due to the inferior positioning of glenosphere. In this study, d was 1.8 mm for configuration 2 and 3.9 mm for configuration 3. Thereby for the cases of inferior positioning, strains in the middle part of glenoid reduced significantly after the prosthesis implantation, thus led to greater amount of stress shielding and subsequently caused bone resorption above the middle peg. Due to the significant inferior shift of GH force in the case of configuration 3, postoperative bone resorption for this implant position occurred earlier than those for other configurations.

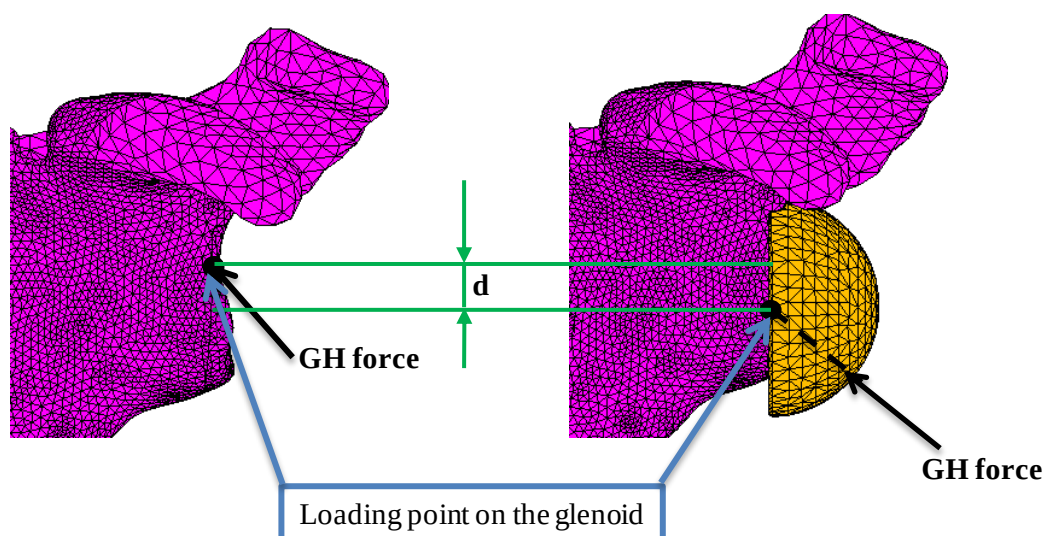


Figure 8-10 Schematic inferior shift of the GH force relative to the glenoid after orthopaedic operation in the cause of fixation configuration 3.

8.4.2 Initial Stability Predictions for Various Glenoid Prosthesis Positions

Postoperative micromotions at the bone-prosthesis interface for the four glenoid prosthesis positions of interest were studied. In comparison with the predicted maximum micromotion for the traditional surgical technique, the predicted value mostly reduced for the case of inferior positioning of glenoid prosthesis, and considerably increased in the case of glenosphere inclination, noticeably in task 9 and 12 (55.1 μm and 72.2 μm , respectively). Therefore, the inferior positioning may improved the postoperative prosthesis initial stability, while, the downward inclination of implant may lead to negative effects on the fixation of glenoid prosthesis. Reasons for the significant micromotion variations due to glenosphere downward tilt are explained next.

Since the GH resultant force was assumed to be irrespective of positions of glenoid component in Delta RSA, the inclination of glenoid prosthesis led to superior movement of loading point of the GH force on the glenosphere surface. In this case, component forces with a reference to the glenoid component attachment site were changed (increase in the compressive force, and decrease in the inferosuperior shear force). Thereby, the apparent micromotion variations in the case of inferior tilting of implant could be induced by the changed loading point with reference to the glenosphere surface, or the loss of inferior glenoid pole. To address the reasons, three FEMs in Figure 8-11 were compared. The first FEM (Figure 8-11 (a)) had normal position and was loaded by the GH force for task 12. In the second FEM (Figure 8-11 (b)), the glenoid prosthesis was inclined 15° and loaded by the GH force for task 12. In the third FEM (Figure 8-11 (c)), the glenoid prosthesis was also inferiorly titled 15° , but the loading point on the glenosphere was the same to that in the first FEM. Thus the first and the third FEMs were under the same shear force with relative to the resected surface, and the second and the third FEMs had the same implant positions. Figure 8-12 illustrates micromotion distributions and the predicted peak values in these three FEMs. In comparison with FEM 1, the loss of inferior glenoid pole in FEM 3 led to an increase of 92 μm in the predicted peak micromotions. In the same bone condition, the predicted peak micromotion in FEM 2 was lower than that in FEM 3. It could be due to the decrease of superior shear force with reference to the resected surface. The predicted

micromotion distributions (Figure 8-12 (a)) showed that great bone-prosthesis interface micromotions in the three FEMs came from the interface between the metaglene and the bone. In comparison with FEM 1, the significant increase of the peak micromotion in FEM 3 came from the inferior part of the resected surface. As introduced in Chapter 2.2.3, bone quality in the glenoid reduced from the subchondral bone layer to a deep layer and then increases. It is noticed that the loss of inferior scapular pole induced by the downward inclination of glenoid prosthesis led to more weak cancellous bone contacting with the implant. In a study of glenoid cancellous bone strength and modulus (Anglin et al., 1999), the Young's modulus reduced approximately 40% from the level of 3.5 mm to 10.5 mm under the glenoid fossa. It could be the weak bone supporting the implant that lead to less stable fixation and thus result in high component migration. With the same bone conditions, micromotion distributions in FEM 2 were similar to that in FEM 3. High values were presented at the inferior bone-metaglene interface.

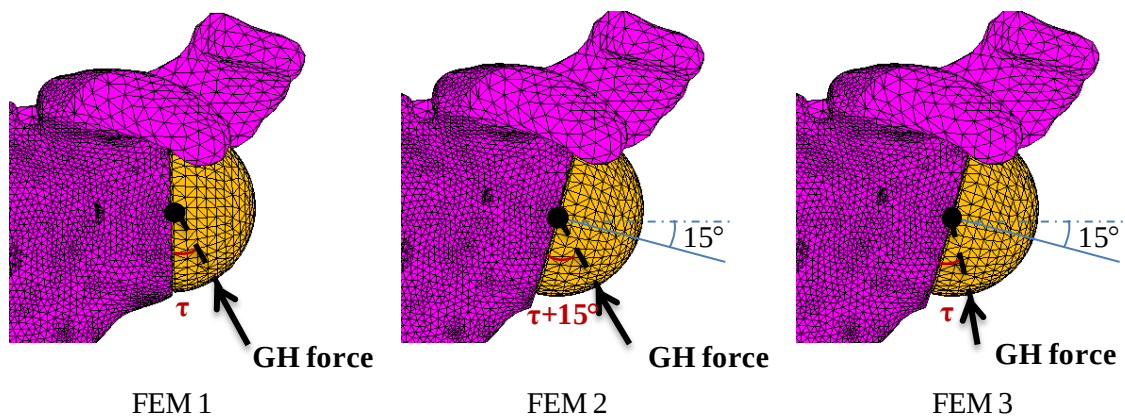


Figure 8-11 (a) FEM 1, normal implant position and GH force; (b) FEM 2, implant downward inclination and the same GH force with that in the normal position; (c) FEM 3, implant downward inclination and the same loading position relative to the glenosphere with that in the normal position.

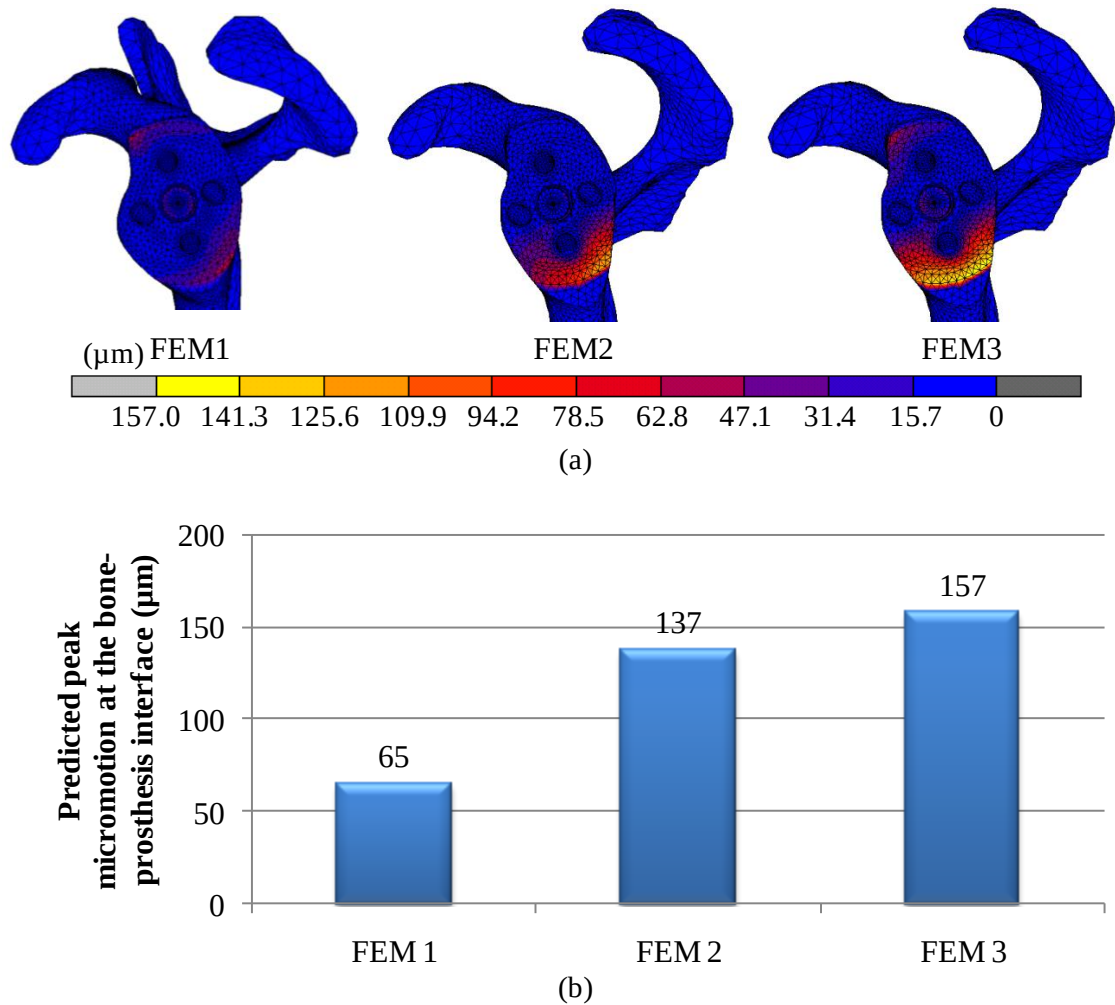


Figure 8-12 Predicted micromotions from three FEMs: (a) micromotion distributions at the bone-prosthesis interface; (b) peak micromotions.

It was noticed that the peak micromotion in task 10 did not vary with the inferior positioning of glenoid prosthesis, and reduced when the implant was inferiorly tilted. Figure 8-13 illustrates micromotion distributions for the four implant position cases under the same force for task 10. The results showed that peak values were presented in the region close to the stem tip for the case without glenosphere tilting (configuration 1, 2, and 3) and at the inferior bone-metaglene interface for the case with glenosphere inclination. Due to the dense bone (mean apparent density of 1.0 g/cm^3) at the end of the middle stem, micromotions there may not vary with the change of implant positions. For the case of configuration 4, the peak micromotion in the region close to the stem tip was reduced due to the decrease of superior shear force acting on the resected surface. However, the loss of inferior glenoid pole resulted in more weak cancellous bone

contacting with the implant, and thus caused the increase of bone-prosthesis interface micromotions there.

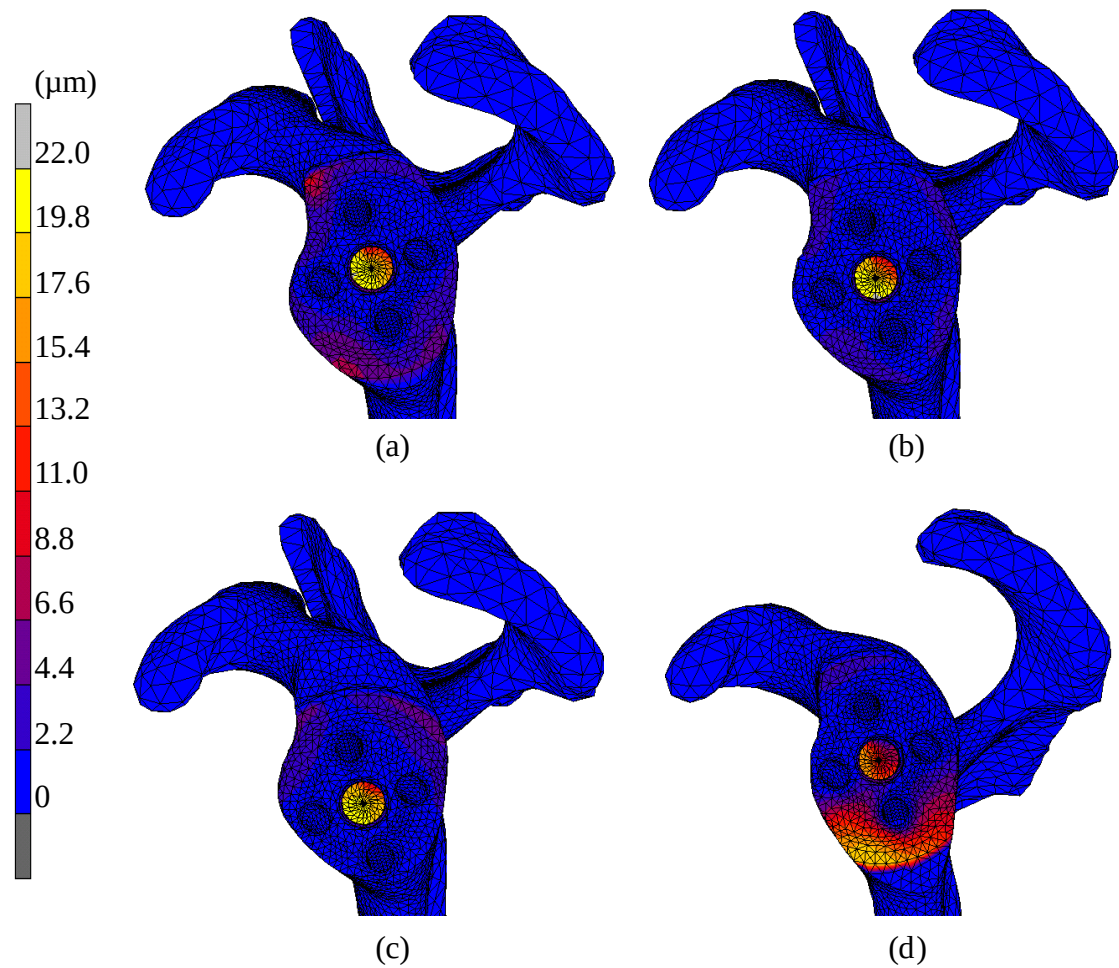


Figure 8-13 Interface micromotion distributions for the four implant positions in task 10.

In a study of standard 25 mm Encore reverse shoulder arthroplasty (Encore Medical), an inferior glenosphere tilt of approximately 15° yielded approximately $4.8 \mu\text{m}$ reduction of bone-prosthesis micromotions in shoulder abduction (Gutiérrez et al., 2007). Similar to the GH force during abduction process, task 10 also required very small anteroposterior shear force. The peak bone-prosthesis interface micromotion for this activity reduced $1.0 \mu\text{m}$ in the case of downward inclination of glenoid prosthesis, and showed a consistent trend to the finding of Gutiérrez et al (2007).

8.5 Conclusion

Inferior scapular notching is one of major concerns in the Grammont type of reverse shoulder arthroplasty (Rockwood, 2007). To minimize this problem, altered glenoid prosthesis positions (inferior positioning and tilting of the glenosphere) have been proposed in several articles (Nyffeler et al., 2005; Simovitch et al., 2007; Kontaxis, 2010). This study is aimed to investigate variations of glenoid prosthesis fixation (including the initial stability and long-term fixation) induced by the varied implant positions. Bone-prosthesis interface micromotions and postoperative adaptive bone resorption for the four varied implant position cases were evaluated. A conclusion is that inferior positioning of glenosphere may lead to inferior shift of GH force after the prosthesis implantation, and thus may result in early bone loss above the middle stem in comparison with those for the normal implant positions. Inclination of the glenoid prosthesis may lead to significant increase in the micromotions at the inferior bone-metaglène interface, and thus result in poor osseointegration and ultimately poor initial stability, even if it does reduce scapular notching.

CHAPTER 9

CONCLUSIONS AND FUTURE WORK

The purpose of this thesis is to investigate effects of scapular notching and strain-induced bone remodelling on the long-term failure of glenoid component in Delta CTA reverse shoulder arthroplasty (RSA). Meanwhile, this thesis has also predicted the long-term fixation of glenoid prosthesis in the case of new glenosphere positions (inferior positioning and downward tilting). As finite element (FE) method was employed as the core tool in this project, reliability of the FEMs was investigated with published data (Frich et al., 1997; Frich et al., 1998; Anglin et al., 1999) and in-vitro testing results before the main work. Summary and conclusions of this project are presented below.

9.1 Summary of the Results/Conclusions

9.1.1 Effects of Scapular Notching on the Fixation of Glenoid Prosthesis

Effects of scapular notching on the fixation of glenoid prosthesis were investigated from three aspects: (1) effects on the safety of inferior screw, (2) effects on the safety of the bone on the surface of inferior screw hole and (3) effects on the initial stability of glenoid component in Delta RSA.

9.1.1.1 Effects of Scapular Notching on the Safety of Inferior Screw

Inferior scapular notching was found to result in negative influences on the safety of inferior screw. It led to an increase of the maximum principal stresses of 23 MPa when standing up from an arm chair and an increase of 7.8 MPa when lifting to the head height. The predicted peak maximum principal stresses reached 172 MPa in the case of a grade 4 notch (Valenti, 2000). This value almost half of the fatigue stress of titanium material (415 MPa). Therefore, the screw may be safe even in the case of a big scapular notch. However, the damaged effects of inferior scapular notching on the screw safety may be aggravated in the sharp corner (i.e. thread of the inferior screw), where stress concentration may occur, and in the case of possible impingements between the screw and the humeral cup. Another result is that inferior scapular notch led to a significant increase of stresses in two places along the screw, one locating in the screw-to-metaglens junction and the other being in the intersection between the screw and the bone.

9.1.1.2 Effects of Scapular Notching on the Safety of the Bone on the Surface of Inferior Screw Hole

Due to the breakage of screws shown in some postoperative radiographs of Delta RSA, safety of the bone close to the screw was concerned. Bone stresses at the bone-screw interface in the case of an inferior notch classified as grade 4 in the Nerot system (Valenti, 2000) were examined. The results indicated that the inferior scapular notch led

to a maximum stress variation of 1.34 MPa for the case of standing up from an armchair. The predicted maximum principal stress of cancellous bone after the creation of a notch peaked at 3.32 MPa at the conjunction between the inferior screw hole and the surface of notch. This peak value is comparable with the reported fatigue failure stress (3.57 MPa) for 400 MPa epiphyseal cancellous bones (Guo et al., 1993), although the fatigue stress of scapula cancellous bone during arm motions is unknown. Thus, scapular notching may generate potential risk of bone fracture at the bone-screw interface, particularly in the region close to the notch.

9.1.1.3 Effects of Scapular Notching on the Initial Stability of Glenoid Prosthesis

Bone-prosthesis interface micromotions after the creation of a grade 4 inferior scapular notch were examined. The maximum micromotion increased only 2.7 μm in the case of erosive inferior scapular neck, peaking at 59.6 μm . This peak value is comparable with the required threshold (50 μm) for postoperative bony ingrowth, thus indicated a possible good initial stability of the glenoid prosthesis in the case of inferior scapular notching.

9.1.2 Effects of Strain-induced Bone Remodelling on the Fixation of Glenoid Prosthesis

Adaptive process of bone remodelling after the implantation of Delta CTA RSA was simulated in this study. Apparent bone density was predicted to reduce significantly (at a level of approximately 63%) in the region close to the resected surface and the middle peg after 8-year follow-up applications of Delta RSA. In addition, the force after the prosthesis implantation transmitted along the middle peg from the resected surface to the tip, and from the posterior to the anterior glenoid. Due to the reduction of glenohumeral joint force for the Delta RSA, bone at the end of stem also resorbed, although the degrees in this region was lower than those close to the back of metaglene.

9.1.3 Correlation between Scapular Notching and Strain-induced Bone Remodelling

Strain-induced bone resorption progressed from the back side of metaglene and the middle peg to the peripheral regions. The process of this type of bone loss may reduce the bone densities and accelerate the generation of inferior scapular notching, as weak bone is more likely to break under the impingement of the humeral cup against the scapular neck. Inferior scapular notching initiated from the inferior and lateral rim of the glenoid and developed to the medial and superior region. The process of this type of bone loss could lead to increase of bone stresses/strains and thus slow down the progress of bone resorption or even lead to bone deposition.

9.1.4 Factors for Postoperative Bone Loss after the RSA Implantation

Factors leading to the postoperative loss of supporting bone for the glenoid prosthesis were analysed. The erosion of glenoid bone could be divided into superior and inferior loss. Superior bone erosion could be mainly caused by the strain-induced bone resorption and osteolysis, which is induced by wear particles of the polyethylene humeral cup. Bone erosion in the inferior region is very complicated. It could include two situations. In the case of no scapular notches being presented, only bone resorption is the predominant factor for the postoperative bone loss. Additionally, bone in this situation is the weakest under the middle peg and dense at the inferior rim of the scapular neck. In the case of a scapular notch being presented, bone shows high densities under the peg and low densities in the inferior periphery. Factors for the bone loss in this situation are complicated. Bone loss in the corner of the metaglene and the middle stem may be predominantly caused by strain-induced bone resorption and polyethylene particle-induced osteolysis. Bone loss in the inferior rim of scapular neck may be predominantly subject to the impingements between the humeral cup and the scapular neck and osteolysis. Bone loss close to the stem tip could be predominantly caused by the adaptive bone remodelling and osteolysis, as this region is far away from origins of both bone adaptation and scapular notching.

9.1.5 Predictions of Glenoid Prosthesis Fixations in the Case of Various Glenoid Prosthesis Positions

Long-term glenoid prosthesis fixation in the case of new surgical techniques (inferior positioning and downward titling of the glenosphere) was predicted and compared with the fixation in the case of conventional surgical technique. In this study, the prosthesis fixation was assessed by two factors: strain-induced bone resorption and initial stability.

9.1.5.1 Bone Remodelling Predictions for Various Glensphere Positions

Adaptive process of bone remodelling for various glenosphere positions were predicted. The results showed that bone resorption developed in the similar trends irrespective of glenosphere positions. They all started in the regions close to the resected surface and the middle peg and extended to the peripheral regions. Bone resorption for the normal position or the inferior inclination of glenoid prosthesis almost evenly distributed above and under the middle stem. In the case of inferior positioning, the amount of bone resorption was greater above than under the middle stem. Due to the significant inferior shift of GH force in the case of inferior positioning of the glenosphere, postoperative bone resorption occurred earlier than those in cases of other glenoid prosthesis positions.

9.1.5.2 Micromotion Predictions for Various Glensphere Positions

Micromotion was used as a parameter to investigate the variations of glenoid prosthesis fixation induced by the various implant positions with relative to the scapula. It is found that downward tilting of the glenoid prosthesis led to high micromotion, peaking at 137.4 μm when standing up from an armchair and thus may result in poor osseointegration and ultimately poor initial stability, in terms of the required threshold of bone ingrowth (50 μm) proposed by Pilliar (1986). The peak micromotions without glenosphere tilting were all within or around the required threshold of osseointegration, thus predicted a good postoperative environment for bone ingrowth.

9.2 Future Work

FE modelling used in this study is capable to investigate and analyse effects of scapular notching and strain-induced bone remodelling on the long-term fixation of glenoid component in Delta CTA RSA. The FE model also provided reasonable predictions and analyses for the possible effects of inferior positioning and downward titling of the glenosphere on the long-term prosthesis fixation. However, in the discussion of this study, it is clear that there were some limitations in the FE modelling and the in-vitro testing. In addition, this study concluded that the proposed inferior positioning of glenoid prosthesis may result in early bone resorption. In addition, the downward tilting may lead to poor environment for the bony ingrowth and ultimately result in poor initial stability. The proposed improvements of these studies and recommendations for the future prosthesis designs are presented as follow:

9.2.1 Inclusion of Muscle Loading

In this study, muscle forces in the shoulder joint are not considered in the FE models. It is unclear whether the muscle loading may affect the predicted results presented in this study. In a study of the total hip prosthesis, Bitsakos (2005) reported that the amount of postoperative bone loss with muscle forces was different from that without muscle forces. Muscles in the shoulder contribute greatly to the forces acting upon the joint during physiological daily activities. Thereby, the inclusion of muscle loading in the FE model could provide realistic loading conditions, and predict realistic bone remodelling responses after the prosthesis implantation.

9.2.2 Validation of the Bone Remodelling Response

In this study, the dead zone in the strain-adaptive bone remodelling algorithm was set as 1.5. This value has only been validated with the clinical DEXA scan results in the hip replacement studies (Kerner et al., 1999; Bitsakos, 2005), but has never been correlated to the bone scans associated with shoulder joint replacements. It is unclear whether this threshold level is anatomic site dependent. Thereby, it would be valuable to validate our FE bone adaptation predictions with DEXA scan, which is the “gold standard” for bone density quantification in clinical practice.

In this study, the radiographic bone loss of a patient who was treated with Delta RSA and had shoulder dysfunction at 8-year follow-up was used as a reference to investigate bone density reductions at long-term follow up, and to predict the possible correlation between bone resorption and loss of supporting bone for the Delta glenoid prosthesis. It would be better to create a time constant which could relate the simulated time to the physical time, based on in-vivo and in-vitro studies for shoulder joint.

9.2.3 Micro FE Modelling of the Cementless Bone-prosthesis Interface Micromotion

Micromotion was used as a mechanical parameter to predict the possibility of occurrence of osseointegration and subsequently to assess the initial stability of glenoid prosthesis. The interface micromotion value is at the same scale to the pore size of the trabeculae and the pore size of the prosthesis attaching on the bone. It is unclear whether macro-scale pores at the bone-prosthesis interface may influence the micromotions in reality. More future work could be carried out on the correlation between the osseointegration process and the possible intimate interaction of the trabeculae against the pores on the prosthesis surface.

9.2.4 Recommendations for the Future Prosthesis Designs

One of the contributions in this project is that it predicted the long-term glenoid fixation in the case of new surgical techniques (inferior positioning or tilting of the glenoid prosthesis). Due to the inferior location of glenoid prosthesis, postoperative glenohumeral joint contact force after the prosthesis implantation of Delta RSA may shift inferiorly relative to that preoperatively, thus may lead to early bone resorption. Due to the downward tilting of glenosphere, more weak bone contact with the prosthesis and generated high micromotions. Thus this prosthesis position may lead to poor environment for bony ingrowth and subsequently result in poor initial stability. Therefore new prosthesis designs or surgical techniques which could synchronously minimize scapular notching, strain-induced bone resorption and could improve the osseointegration is needed.

Based on personal talks with a shoulder surgeon (Dr. Denny T. T. Lie from Singapore General Hospital), scapular notching is not frequently observed in Singapore. This finding is different from the published reports, in which postoperative incidence (50% - 96%) of scapular notching was very high for Delta RSA. It is noticed that most current clinical results with regard to scapular notching were obtained from European or American hospitals. It could be the generally different sizes of scapulae between Asian and European or American patients that lead to various incidence of scapular notching. For the comparatively small Asian scapulae, glenosphere in Delta RSA may mostly overlap the inferior rim of glenoid thus reduce impingements of humeral cup against the scapular neck, although the glenosphere position is not changed. However, the limit glenosphere sizes (36 mm and 42 mm in diameter) may be too small to overlap the inferior rim of glenoid for most Europeans, thus scapular notching are frequently observed. Therefore, in the author's opinion, the future work could be carried out on the study of bigger than normal glenosphere sizes. The big glenosphere may not only minimize scapular notching but also avoid the problems of early bone resorption and less osseointegration induced by the inferior positioning and downward tilting of glenosphere.

REFERENCES

- Abdul-Kadir MR, Hansen U, Klabunde R, Lucas D, Amis A. 2008. Finite element modelling of primary hip stem stability: the effect of interference fit. *Journal of Biomechanics* 41:587-594.
- Abu-Rajab RB, Watson WS, Walker B, Roberts J, Gallacher SJ, Meek RM. 2006. Peri-prosthetic bone mineral density after total knee arthroplasty. Cemented versus cementless fixation. *The Journal of Bone and Joint Surgery [Br]* 88:606-613.
- Ahir SP, Walker PS, Squire-Taylor CJ, Blunn GW, Bayley JIL. 2004. Analysis of glenoid fixation for a reversed anatomy fixed-fulcrum shoulder replacement. *Journal of Biomechanics* 37:1699-1708.
- An YH, Draughn RA. 1999. *Mechanical Testing of Bone and the Bone-Implant Interface*. USA: CRC Press.
- Ando M, Imura S, Omori H, Okumura Y, Bo A, Baba H. 1999. Nonlinear three-dimensional finite element analysis of newly designed cementless total hip stems. *Artificial Organs* 23:339-346.
- Anglin C, Tolhurst P, Wyss UP, Pichora DR. 1999. Glenoid cancellous bone strength and modulus. *Journal of Biomechanics* 32:1091-1097.
- Anthony AS, Brian DB. 2006. *Sports Medicine*: Lippincott Williams & Wilkins
- Bailie DS, Llinas PJ, Ellenbecker TS. 2008. Cementless humeral resurfacing arthroplasty in active patients less than fifty-five years of age. *The Journal of Bone and Joint Surgery [Am]* 90:110-117.
- Basmajian JV, Bazant FJ. 1959. Factors preventing downward dislocation of the adducted shoulder joint. An electromyographic and morphological study. *The Journal of Bone and Joint Surgery [Am]* 41-A:1182-1186.
- Bayraktar HH, Keaveny, TM. 2004. Mechanisms of uniformity of yield strains for trabecular bone. *Journal of Biomechanics* 37:1671-1678.
- Beddow FH, Elloy MA. 1982. Clinical experience with the Liverpool shoulder replacement. In: Bayley I, Kessel L, eds. *Shoulder Surgery*. New York: Springer-Verlag:164-167.

- Bertin KC, Freeman MA, Samuelson KM, Ratcliffe SS, Todd RC. 1985. Stemmed revision arthroplasty for aseptic loosening of total knee replacement. *The Journal of Bone and Joint Surgery [Br]* 67:242-248.
- Biegler FB, Reuben JD, Harrigan TP, Hou FJ, Akin JE. 1995. Effect of porous coating and loading conditions on total hip femoral stem stability. *Journal of Arthroplasty* 10:839-847.
- Bitsakos C. 2005. Computer Simulation of Periprosthetic Bone Remodelling after Total Hip Arthroplasty. In. London, UK: Imperial College London.
- Bobyn JD, Pilliar RM, Cameron HU, Weatherly GC. 1980. The optimum pore size for the fixation of porous-surfaced metal implants by the ingrowth of bone. *Clinical Orthopaedics and Related Research*:263-270.
- Bobyn JD, Pilliar RM, Cameron HU, Weatherly GC. 1981. Osteogenic phenomena across endosteal bone-implant spaces with porous surfaced intramedullary implants. *Acta Orthopaedica Scandinavica* 52:145-153.
- Bodey WN, Yeoman PM. 1983. Prosthetic arthroplasty of the shoulder. *Acta Orthopaedica Scandinavica* 54:900-903.
- Boileau P, Trojani C, Walch G, Krishnan SG, Romeo A, Sinnerton R. 2001. Shoulder arthroplasty for the treatment of the sequelae of fractures of the proximal humerus. *Journal of Shoulder and Elbow Surgery* 10:299-308.
- Boileau P, Watkinson D, Hatzidakis AM, Hovorka I. 2006. Neer Award 2005: The Grammont reverse shoulder prosthesis: results in cuff tear arthritis, fracture sequelae, and revision arthroplasty. *Journal of Shoulder and Elbow Surgery* 15:527-540.
- Boileau P, Watkinson DJ, Hatzidakis AM, Balg F. 2005. Grammont reverse prosthesis: design, rationale, and biomechanics. *Journal of Shoulder and Elbow Surgery* 14:147S-161S.
- Branemark PI. 1983. Osseointegration and its experimental background. *Journal of Prosthetic Dentistry* 50:399-410.
- Brekelmans WA, Poort HW, Slooff TJ. 1972. A new method to analyse the mechanical behaviour of skeletal parts. *Acta Orthopaedica Scandinavica* 43:301-317.
- Broström LA, Wallensten R, Olsson E, Anderson D. 1992. The Kessel Prosthesis in total shoulder arthroplasty. A five-year experience. *Clinical orthopedics* 277:155-160.

- Brown IW, Ring PA. 1985. Osteolytic changes in the upper femoral shaft following porous-coated hip replacement. *The Journal of Bone and Joint Surgery [Br]* 67:218-221.
- Burke CS, Roberts CS, Nyland JA, Radmacher PG, Acland RD, Voor MJ. 2006. Scapular thickness--implications for fracture fixation. *Journal of Shoulder and Elbow Surgery* 15:645-648.
- Cameron HU, Pilliar RM, Macnab I. 1976. The rate of bone ingrowth into porous metal. *Journal of Biomedical Materials Research* 10:295-302.
- Cann CE, Genant HK. 1980. Precise measurement of vertebral mineral content using computed tomography. *Journal of Computer Assisted Tomography* 4:493-500.
- Carlsson AS, Gentz CF, Linder L. 1983. Localized bone resorption in the femur in mechanical failure of cemented total hip arthroplasties. *Acta Orthopaedica Scandinavica* 54:396-402.
- Carter DR. 1982. The relationship between in vivo strains and cortical bone remodeling. *Critical Reviews in Biomedical Engineering* 8:1-28.
- Carter DR. 1984. Mechanical loading histories and cortical bone remodeling. *Calcified Tissue International* 36:S19-S24.
- Carter DR, Harris WH, Vasu R, Caler WE. 1981. The mechanical and biological response of cortical bone to in vivo strain histories. In: Cowin SC. ed. *Mechanical Properties of Bone*. New York: American society of mechanical engineers publication
- Carter DR, Hayes WC. 1977. The compressive behaviour of bone as a two-phase porous structure. *The Journal of Bone and Joint Surgery [Am]* 59-A:954-962.
- Chamay A, Tschantz P. 1972. Mechanical influences in bone remodeling. Experimental research on Wolff's law. *Journal of Biomechanics* 5:173-180.
- Chong DYR. 2009. Biomechanical Analysis of Fixation and Bone Remodelling of Total Knee Replacement. In: *Mechanical Engineering*. London: Imperial College London.
- Churchill RS, Brems JJ, Kotschi H. 2001. Glenoid size, inclination, and version: an anatomic study. *Journal of Shoulder and Elbow Surgery* 10:327-332.
- Ciarelli MJ, Goldstein SA, Kuhn JL, Cody DD, Brown MB. 1991. Evaluation of orthogonal mechanical properties and density of human trabecular bone from the major metaphyseal regions with materials testing and computed tomography. *Journal of Orthopaedic Research* 9:674-682.

- Codsi MJ, Iannotti JP. 2008. The effect of screw position on the initial fixation of a reverse total shoulder prosthesis in a glenoid with a cavitory bone defect. *Journal of Shoulder and Elbow Surgery* 17:479-486.
- Cofield RH. 1984. Total shoulder arthroplasty with the Neer prosthesis. *The Journal of Bone and Joint Surgery [Am]* 66:899-906.
- Constant CR, Murley AH. 1987. A clinical method of functional assessment of the shoulder. *Clinical Orthopaedics and Related Research* 214:160-164.
- Cowin SC. 1984. Mechanical modeling of the stress adaptation process in bone. *Calcified Tissue International* 36 Suppl 1:S98-103.
- Cowin SC. 1993. Bone stress adaptation models. *Journal of Biomechanical Engineering* 115:528-533.
- Cowin SC. 2001. *Bone Mechanics Handbook*. Boca Raton, Florida, USA: CRC Press
- Cowin SC, Hegedus DH. 1976. Bone remodeling. I: Theory of adaptive elasticity. *Journal of Elasticity* 6:313-326.
- Cowin SC, Van Buskirk WC. 1979. Surface bone remodeling induced by a medullary pin. *Journal of Biomechanics* 12:269-276.
- Currey JD. 1986. Power law models for the mechanical properties of cancellous bone. *Engineering in Medicine* 15:153-154.
- Davis AM, Damani M, White LM, Wunder JS, Griffin AM, Bell RS. 2005. Periprosthetic bone remodeling around a prosthesis for distal femoral tumors: longitudinal follow-up. *The Journal of Arthroplasty* 20:219-224.
- De Wilde L, Mombert M, Van Petegem P, Verdonk R. 2001. Revision of shoulder replacement with a reversed shoulder prosthesis (Delta III): report of five cases. *Acta Orthopaedica Belgica* 67:348-353.
- De Wilde LF, Audenaert EA, Berghs BM. 2004. Shoulder prostheses treating cuff tear arthroplasty: a comparative biomechanical study. *Journal of Orthopaedic Research* 22:1222-1230.
- Biomechanics in Dentistry http://www.feppd.org/ICB-Dent/campus/biomechanics_in_dentistry/ldv_data/mech/basic_titanium.htm#top
- Edwards TB, Kadakia NR, Boulahia A, Kempf JF, Boileau P, Nemoz C, Walch G. 2003. A comparison of hemiarthroplasty and total shoulder arthroplasty in the treatment of primary glenohumeral osteoarthritis: results of a multicenter study. *Journal of Shoulder and Elbow Surgery* 12:207-213.

- Fenlin JM, Jr. 1975. Total glenohumeral joint replacement. *Orthopaedic Clinics of North America* 6:565-583.
- Fevang BT, Lie SA, Havelin LI, Skredderstuen A, Furnes O. 2009. Risk factors for revision after shoulder arthroplasty: 1,825 shoulder arthroplasties from the Norwegian Arthroplasty Register. *Acta Orthopaedica* 80:83-91.
- Frankle M, Levy JC, Pupello D, Siegal S, Saleem A, Mighell M, Vasey M. 2006. The reverse shoulder prosthesis for glenohumeral arthritis associated with severe rotator cuff deficiency. a minimum two-year follow-up study of sixty patients surgical technique. *The Journal of Bone and Joint Surgery [Am]* 88 Suppl 1 Pt 2 p178-190.
- Frich LH, Jensen NC, Odgaard A, Pedersen CM, Søjbjerg JO, Dalstra M. 1997. Bone strength and material properties of the glenoid. *Journal of Shoulder and Elbow Surgery* 6:97-104.
- Frich LH, Odgaard A, Dalstra M. 1998. Glenoid bone architecture. *Journal of Shoulder and Elbow Surgery* 7:356-361.
- Friedman RJ. 1994. *Arthroplasty of the Shoulder*. New York: Thieme medical publishes, Inc. New York. p1-26
- Friedman RJ, Hawthorne KB, Genez BM. 1992. The use of computerized tomography in the measurement of glenoid version. *The Journal of Bone and Joint Surgery [Am]* 74:1032-1037.
- Frost HM. 1964. *The laws of bone structure*. Charles CT, Springfield, Illinois, USA.
- Fyhrie DP, Carter DR. 1986. A unifying principle relating stress to trabecular bone morphology. *Journal of Orthopaedic Research* 4:304-317.
- Galante J, Rostoker W, Ray RD. 1970. Physical properties of trabecular bone. *Calcified Tissue International* 5:236-246.
- Gibson LJ. 1985. The mechanical behaviour of cancellous bone. *Journal of Biomechanics* 18:317-328.
- Grammont PM, Baulot E. 1993. Delta shoulder prosthesis for rotator cuff rupture. *Orthopedics* 16:65-68.
- Grammont PM, Trouilloud P, Laffay JP, Deries X. 1987. Etude et realisation d'une nouvelle protheses d'epaule. *Rhumatologie* 39:17-22.
- Grassi FA, Murena L, Valli F, Alberio R. 2009. Six-year experience with the Delta III reverse shoulder prosthesis. *Journal of Orthopaedic Surgery (Hong Kong)* 17:151-156.

- Guery J, Favard L, Sirveaux F, Oudet D, Mole D, Walch G. 2006. Reverse total shoulder arthroplasty. Survivorship analysis of eighty replacements followed for five to ten years. *The Journal of Bone and Joint Surgery [Am]* 88:1742-1747.
- Gupta S. 2002. Stress Analysis of the Scapula: Design Considerations of Glenoid Prostheses. In: *Mechanical engineering and marine technology*. Delft: Delft University of Technology.
- Gupta S, van der Helm FCT, van Keulen F, Kaptein BL. 2004. Development and experimental validation of a three-dimensional finite element model of the human scapula. *Proceedings of the Institution of Mechanical Engineers. Part H, Journal of engineering in medicine* 218:127-142.
- Gupta S, van der Helm FCT. 2004. Load transfer across the scapula during humeral abduction. *Journal of Biomechanics* 37:1001-1009.
- Gutiérrez S, Greiwe RM, Frankle MA, Siegal S, Lee III WE. 2007. Biomechanical comparison of component position and hardware failure in the reverse shoulder prosthesis. *Journal of Shoulder and Elbow Surgery* 16:S9-S12.
- Gutiérrez S, Levy JC, Frankle MA, Cuff D, Keller TS, Pupello DR, Lee III WE. 2008. Evaluation of abduction range of motion and avoidance of inferior scapular impingement in a reverse shoulder model. *Journal of Shoulder and Elbow Surgery* 17:608-615.
- Haddad RJ, Cook SD, Thomas KA. 1987. Biological fixation of porous-coated implants. *The Journal of Bone and Joint Surgery [Am]* 69:1459-1466.
- Harman M, Frankle M, Vasey M, Banks S. 2005. Initial glenoid component fixation in “reverse” total shoulder arthroplasty-A biomechanical evaluation. *Journal of Shoulder and Elbow Surgery* 14:162S-167S.
- Harris WH, Jasty M. 1985. Bone ingrowth into porous coated canine acetabular replacements: the effect of pore size, apposition, and dislocation. *Hip*:214-234.
- Hart RT. 2001. Bone Modeling and Remodeling: Theories and Computation. In: Cowin SC. ed. *Bone Mechanics Handbook*. Boca Raton, Florida, USA: CRC Press.
- Hart RT, Davy DT, Heiple KG. 1984. Mathematical modeling and numerical solutions for functionally dependent bone remodeling. *Calcified Tissue International* 36 Suppl 1:S104-109.
- Hatzidakis AM, Norris TR, Boileau P. 2005. Reverse shoulder arthroplasty; indication, technique and results. *Techniques in Shoulder and Elbow Surgery* 6:135-154.

- Hedtmann A, Werner A. 2007. Shoulder arthroplasty in rheumatoid arthritis. *Orthopade* 36:1050-1061.
- Hefzy MS, Singh SP. 1997. Comparison between two techniques for modeling interface conditions in a porous coated hip endoprosthesis. *Medical Engineering and Physics* 19:50-62.
- Helgason B, Taddei F, Palsson H, Schileo E, Cristofolini L, Viceconti M, Brynjolfsson S. 2008. A modified method for assigning material properties to FE models of bones. *Medical Engineering and Physics* 30:444-453.
- Hopkins AR, Hansen UN, Amis AA. 2005. Finite element models of total shoulder replacement: application of boundary conditions. *Computer Methods in Biomechanics and Biomedical Engineering* 8:39-44.
- Hopkins AR, Hansen UN, Amis AA, Taylor M, Gronau N, Anglin C. 2006. Finite element modelling of glenohumeral kinematics following total shoulder arthroplasty. *Journal of Biomechanics* 39:2476-2483.
- Hopkins AR, Hansen UN, Bull AMJ, Emery R, Amis A. 2008. Fixation of the reversed shoulder prosthesis. *Journal of Shoulder and Elbow Surgery* 17:974-980.
- Huiskes R, van Rietbergen B. 2005. In *Basic Orthopaedics Biomechanics and Mechano-biology*, 3rd edn, ed.: Philadelphia, PA: Lippincott, Williams & Wilkins.
- Huiskes R, Weinans H, Grootenboer HJ, Dalstra M, Fudala B, Slooff TJ. 1987. Adaptive bone-remodeling theory applied to prosthetic-design analysis. *Journal of Biomechanics* 20:1135-1150.
- Humphrey CS, Kelly II JD, Norris TR. 2008. Optimizing glenosphere position and fixation in reverse shoulder arthroplasty, Part Two: The three-column concept. *Journal of Shoulder and Elbow Surgery* 17:595-601.
- Iannotti JP, Gabriel JP, Schneck SL, Evans BG, Misra S. 1992. The normal glenohumeral relationships. An anatomical study of one hundred and forty shoulders. *The Journal of Bone and Joint Surgery [Am]* 74:491-500.
- Inman VT, Saunders JBdM, Abbott LC. 1944. Observation on the function of the shoulder joint. *The Journal of Bone and Joint Surgery [Am]* 26:1-30.
- Jasty M, Bragdon CR, Zalenski E, O'Connor D, Page A, Harris WH. 1997. Enhanced stability of uncemented canine femoral components by bone ingrowth into the porous coatings. *The Journal of Arthroplasty* 12:106-113.

- John M, Pap G, Angst F, Flury MP, Lieske S, Schwyzer HK, Simmen BR. 2010. Short-term results after reversed shoulder arthroplasty (Delta III) in patients with rheumatoid arthritis and irreparable rotator cuff tear. *International Orthopaedics* 34:71-77.
- Jonsson E, Brattstrom M, Lidgren L. 1988. Evaluation of the rheumatoid shoulder function after hemiarthroplasty and arthrodesis. *Scandinavian Journal of Rheumatology* 17:17-26.
- Jonsson E, Egund N, Kelly I, Rydholm U, Lidgren L. 1986. Cup arthroplasty of the rheumatoid shoulder. *Acta Orthopaedica Scandinavica* 57:542-546.
- Joshi D. 2008. Total shoulder replacement-the third most replaced joint in America. [http://www.pearldiverinc.com/pdi/articles.jsp?q=/var/www/html/pearldiver/market/ext/html/Total-Shoulder-Replacement-The-3rd-Most-Replaced-Joint-in-America.html&t=Total Shoulder Replacement - The 3rd Most Replaced Joint in America!&c=ext](http://www.pearldiverinc.com/pdi/articles.jsp?q=/var/www/html/pearldiver/market/ext/html/Total-Shoulder-Replacement-The-3rd-Most-Replaced-Joint-in-America.html&t=Total%20Shoulder%20Replacement%20-%20The%203rd%20Most%20Replaced%20Joint%20in%20America!&c=ext).
- Joshi D. 2009. Reverse shoulder replacement soaring popularity. [http://www.pearldiverinc.com/pdi/articles.jsp?q=/var/www/html/pearldiver/market/ext/html/Reverse-Shoulder-Replacement-Soaring-Popularity-07-20-09.html&t=Reverse Shoulder Replacement Soaring Popularity&c=ext](http://www.pearldiverinc.com/pdi/articles.jsp?q=/var/www/html/pearldiver/market/ext/html/Reverse-Shoulder-Replacement-Soaring-Popularity-07-20-09.html&t=Reverse%20Shoulder%20Replacement%20Soaring%20Popularity&c=ext).
- Juvenspan M, Nourissat G, Doursounian L. 2005. The use of a reverse shoulder system for the treatment of massive irreparable rotator cuff tears. http://www.maitrise-orthop.com/viewPage_us.do?id=797.
- Kalouche I, Abdelmoumen S, Crepin J, Mitton D, Guillot G, Gagey O. 2008. Mechanical properties of glenoid cancellous bone. In: 16th Congress of European Society of Biomechanics. p S324.
- Kalouche I, Crepin J, Abdelmoumen S, Mitton D, Guillot G, Gagey O. 2010. Mechanical properties of glenoid cancellous bone. *Clinical Biomechanics* 25:292-298.
- Kalouche I, Sevivas N, Wahegaonker A, Sauzieres P, Katz D, Valenti P. 2009. Reverse shoulder arthroplasty: does reduced medialisation improve radiological and clinical results? *Acta Orthopaedica Belgica* 75:158-166.
- Katz D, O'Toole G, Cogswell L, Sauzieres P, Valenti PH. 2007. A history of the reverse shoulder prosthesis. *International Journal of Shoulder Surgery* 1:108-113.

- Keaveny TM, Hayes WC. 1993. A 20-Year Perspective on the Mechanical-Properties of Trabecular Bone. *Journal of Biomechanical Engineering-Transactions of the Asme* 115:534-542.
- Kelley MT. 2006. Nonsurgical management of osteoarthritis of the knee. *Journal of the American Academy of Physician Assistants* 19:26-32.
- Kelly II JD, Humphrey CS, Norris TR. 2008. Optimizing glenosphere position and fixation in reverse shoulder arthroplasty, Part One: The twelve-mm rule. *Journal of Shoulder and Elbow Surgery* 17:589-594.
- Kerner J, Huiskes R, van Lenthe GH, Weinans H, van Rietbergen B, Engh CA, Amis AA. 1999. Correlation between pre-operative periprosthetic bone density and post-operative bone loss in THA can be explained by strain-adaptive remodelling. *Journal of Biomechanics* 32:695-703.
- Kessel L, Bayley JIL. 1982. The Kessel total shoulder replacement. In: Bayley JIL, Kessel L, eds. *Shoulder Surgery*. New York: Springer-Verlag. p160-164.
- Klawitter JJ, Bagwell JG, Weinstein AM, Sauer BW. 1976. An evaluation of bone growth into porous high density polyethylene. *Journal of Biomedical Materials Research* 10:311-323.
- Kölbel R, Friedebold G. 1973. Shoulder joint replacement. *Arch Orthop Unfallchir* 76:31-39.
- Kontaxis A. 2010. Biomechanical Analysis of Reverse Anatomy Shoulder Prosthesis Design In: Department of Mechanical and System Engineering. Newcastle: Newcastle University, UK.
- Kontaxis A, Johnson, G. R. 2009. The biomechanics of reverse anatomy shoulder replacement - A modelling study. *Clinical Biomechanics* 24:254-260.
- Kopperdahl DL, Keaveny TM. 1998. Yield strain behavior of trabecular bone. *Journal of Biomechanics* 31:601-608.
- Kuiper JH, Huiskes R. 1996. Friction and stem stiffness affect dynamic interface motion in total hip replacement. *Journal of Orthopaedic Research* 14:36-43.
- Lama F, Bhatiaa DN, Mostofib SB, van Rooyena K, de Beera JF. 2007. Biomechanical considerations of the normal and rotator cuff deficient shoulders and the reverse shoulder prosthesis. *Current Orthopaedics* 21:40-46.
- Lanyon LE. 1984. Functional strain as a determinant for bone remodeling. *Calcified Tissue International* 36 Suppl 1:S56-61.

- Lanyon LE, Goodship AE, Pye CJ, MacFie JH. 1982. Mechanically adaptive bone remodelling. *Journal of Biomechanics* 15:141-154.
- Lehtinen JT, Tingart MJ, Apreleva M, Warner JJP. 2004. Total, trabecular, and cortical bone mineral density in different regions of the glenoid. *Journal of Shoulder and Elbow Surgery* 13:344-348.
- Lévigne C, Boileau P, Favard L, Garaud P, Mole D, Sirveaux F, Walch G. 2008. Scapular notching in reverse shoulder arthroplasty. *Journal of Shoulder and Elbow Surgery* 17:925-935.
- Lévigne C, Garret J, Boileau P, Alami G, Favard L, Walch G. 2010 November. Scapular notching in reverse shoulder arthroplasty: Is it important to avoid it and how? *Clinical Orthopaedics and Related Research*. DOI 10.1007/s11999-010-1695-8
- Levitz CL, Lotke PA, Karp JS. 1995 December. Long-term changes in bone mineral density following total knee replacement. *Clinical Orthopaedics and Related Research* 68-72.
- Levy J, Blum S. 2009. Inferior scapular notching following encore reverse shoulder prosthesis. *Orthopedics* 32:775.
- Levy JC, Virani N, Pupello D, Frankle M. 2007. Use of the reverse shoulder prosthesis for the treatment of failed hemiarthroplasty in patients with glenohumeral arthritis and rotator cuff deficiency. *The Journal of Bone and Joint Surgery [Br]* 89:189-195.
- Levy O, Copeland SA. 2004. Cementless surface replacement arthroplasty (Copeland CSRA) for osteoarthritis of the shoulder. *Journal of Shoulder and Elbow Surgery* 13:266-271.
- Li MG, Nilsson KG. 2000. The effect of the preoperative bone quality on the fixation of the tibial component in total knee arthroplasty. *The Journal of Arthroplasty* 15:744-753.
- Lim D, Seliktar R, Sun W, Holman J, Nunes L. 2002. Mechanical property distribution and finite element modeling of the shoulder based on quantitative computerized tomography. *Proceedings of the Ieee 28th Annual Northeast Bioengineering Conference*:247-248.
- Linde F, Pongsoipetch B, Frich LH, Hvid I. 1990. Three-axial strain controlled testing applied to bone specimens from the proximal tibial epiphysis. *Journal of Biomechanics* 23:1167-1172.

- Lo IK, Litchfield RB, Griffin S, Faber K, Patterson SD, Kirkley A. 2005. Quality-of-life outcome following hemiarthroplasty or total shoulder arthroplasty in patients with osteoarthritis. A prospective, randomized trial. *The Journal of Bone and Joint Surgery [Am]* 87:2178-2185.
- Lonner JH, Klotz M, Levitz C, Lotke PA. 2001. Changes in bone density after cemented total knee arthroplasty: influence of stem design. *The Journal of Arthroplasty* 16:107-111.
- Lord G, Bancel P. 1983. The madreporic cementless total hip arthroplasty. New experimental data and a seven-year clinical follow-up study. *Clinical Orthopaedics and Related Research* 176: 67-76.
- Lotz JC, Gerhart TN, Hayes WC. 1990. Mechanical properties of trabecular bone from the proximal femur: a quantitative CT study. *Journal of Computer Assisted Tomography* 14:107-114.
- Lotz JC, Gerhart TN, Hayes WC. 1991. Mechanical properties of metaphyseal bone in the proximal femur. *Journal of Biomechanics* 24:317-329.
- Lucas DB. 1973. Biomechanics of the shoulder joint. *Archives of Surgery* 107:425-432.
- Lugli T. 1978. Artificial shoulder joint by Pean (1893): the facts of an exceptional intervention and the prosthetic method. *Clinical Orthopaedics and Related Research*:215-218.
- Muscle MagFitness.com. Osteoarthritis Affects Over 21 Million Americans. <http://www.musclemagfitness.com/disease-and-conditions/bone-disease/-osteoarthritis-affects-over-21-million-americans.html>.
- Mansat P, Barea C, Hobatho MC, Darmana R, Mansat M. 1998. Anatomic variation of the mechanical properties of the glenoid. *Journal of Shoulder and Elbow Surgery* 7:109-115.
- Martin RB. 1984. Porosity and specific surface of bone. *Critical Reviews in Biomedical Engineering* 10:179-222.
- Martini FH. 2001. *Fundamentals of Anatomy and Physiology*, Seventh edition: Prentice-Hall, Inc.
- Maurel N, Diop A, Grimberg J. 2005. A 3D finite element model of an implanted scapula: importance of a multiparametric validation using experimental data. *Journal of Biomechanics* 28:1865-1872.

- Maurel N, Diop A, Grimberg J, Elise S. 2002. In vitro biomechanical analysis of glenoids before and after implantation of prosthetic components. *Journal of Biomechanics* 35:1071-1080.
- McFarland EG, Sanquarjit, P, Tasaki A, Keyurapan E, Fishman EK, Fayad LM. 2006. the reverse shoulder prosthesis: a review of imaging features and complications. *Skeletal Radiology* 35:488-496.
- Michel MC, Guo XD, Gibson LJ, McMahon TA, Hayes WC. 1993. Compressive fatigue behavior of bovine trabecular bone. *Journal of Biomechanics* 26:453-463.
- Miles AW, Tanner KE. 1992. *Strain Measurement in Biomechanics*. London: Chapman & Hall.
- Mimar R, Limb D, Hall RM. 2008. Evaluation of the mechanical and architectural properties of glenoid bone. *Journal of Shoulder and Elbow Surgery* 17:336-341.
- Morgan EF, Bayraktar HH, Keaveny TM. 2003. Trabecular bone modulus-density relationships depend on anatomic site. *Journal of Biomechanics* 36:897-904.
- Morgan EF, Bayraktar HH, Yeh OC, Majumdar S, Burghardt A, Keaveny TM. 2004. Contribution of inter-site variations in architecture to trabecular bone apparent yield strains. *Journal of Biomechanics* 37:1413-1420.
- Morgan EF, Keaveny TM. 2001. Dependence of yield strain of human trabecular bone on anatomic site. *Journal of Biomechanics* 34:569-577.
- Morgan EF, Yeh OC, Chang WC, Keaveny TM. 2001. Nonlinear behavior of trabecular bone at small strains. *Journal of Biomechanical Engineering* 123:1-9.
- Mulhall KJ, Ghomrawi HM, Engh GA, Clark CR, Lotke P, Saleh KJ. 2006. Radiographic prediction of intraoperative bone loss in knee arthroplasty revision. *Clinical Orthopaedics and Related Research* 446:51-58.
- Naveed MA, Kitson J, Bunker TD. 2011. The Delta III reverse shoulder replacement for cuff tear arthropathy: a single-centre study of 50 consecutive procedures. *The Journal of Bone and Joint Surgery [Br]* 93:57-61.
- Neer CS, 2nd. 1974. Replacement arthroplasty for glenohumeral osteoarthritis. *The Journal of Bone and Joint Surgery [Am]* 56:1-13.
- Neer CS, 2nd, Craig EV, Fukuda H. 1983. Cuff-tear arthropathy. *The Journal of Bone and Joint Surgery [Am]* 65:1232-1244.
- Neer CS, 2nd, Watson KC, Stanton FJ. 1982. Recent experience in total shoulder replacement. *The Journal of Bone and Joint Surgery [Am]* 64:319-337.

- Neer CS, Brown TH, Jr., McLaughlin HL. 1953. Fracture of the neck of the humerus with dislocation of the head fragment. *The American Journal of Surgery* 85:252-258.
- Neer CS, Watson, KC, Stanton FJ. 1982. Recent experience in total shoulder replacement. *The Journal of Bone and Joint Surgery [Am]* 64-A:319-337.
- Nyffeler RW, Werner CM, Simmen BR, Gerber C. 2004. Analysis of a retrieved Delta III total shoulder prosthesis. *The Journal of Bone and Joint Surgery [Br]* 86:1187-1191.
- Nyffeler RW, Werner CML, Gerber C. 2005. Biomechanical relevance of glenoid component positioning in the reverse Delta III total shoulder prosthesis. *Journal of Shoulder and Elbow Surgery* 14:524-528.
- Orr TE, Carter DR, Schurman DJ. 1988. Stress analyses of glenoid component designs. *Clinical Orthopaedics and Related Research*:217-224.
- Peng L, Bai J, Zeng X, Zhou Y. 2006. Comparison of isotropic and orthotropic material property assignments on femoral finite element models under two loading conditions. *Medical Engineering and Physics* 28:227-233.
- Petersen MM, Jensen NC, Gehrchen PM, Nielsen PK, Nielsen PT. 1996. Ten relation between trabecular bone strength and bone mineral density assessed by dual photon and dual energy X-ray absorptiometry in the proximal tibia. *Calcified Tissue International* 59:311-314.
- Petersen MM, Nielsen PT, Lauritzen JB, Lund B. 1995. Changes in bone mineral density of the proximal tibia after uncemented total knee arthroplasty. A 3-year follow-up of 25 knees. *Acta Orthopaedica Scandinavica* 66:513-516.
- Pilliar RM, Lee JM, Maniopoulos C. 1986. Observations on the effect of movement on bone ingrowth into porous-surfaced implants. *Clinical Orthopaedics and Related Research* 208:108-113.
- Pressel T, Lengsfeld M, Leppek R, Schmitt J. 2000. Bone remodelling in humeral arthroplasty: follow-up using CT imaging and finite element modeling--an in vivo case study. *Archives of Orthopaedic and Trauma Surgery* 120:333-335.
- Pugh JW, Rose RM, Radin EL. 1973. A structural model for the mechanical behavior of trabecular bone. *Journal of Biomechanics* 6:657-670.
- Radnay CS, Setter KJ, Chambers L, Levine WN, Bigliani LU, Ahmad CS. 2007. Total shoulder replacement compared with humeral head replacement for the

- treatment of primary glenohumeral osteoarthritis: a systematic review. *Journal of Shoulder and Elbow Surgery* 16:396-402.
- Rancourt D, Shirazi-Adl A, Drouin G, Paiement G. 1990. Friction properties of the interface between porous-surfaced metals and tibial cancellous bone. *Journal of Biomedical Materials Research* 24:1503-1519.
- Reilly DT, Burstein AH. 1975. The elastic and ultimate properties of compact bone tissue. *Journal of Biomechanics* 8:393-405.
- Rho JY, Hobatho MC, Ashman RB. 1995. Relations of mechanical properties to density and CT numbers in human bone. *Medical Engineering and Physics* 17:347-355.
- Rice JC, Cowin SC, Bowman JA. 1988. On the dependence of the elasticity and strength of cancellous bone on apparent density. *Journal of Biomechanics* 21:155-168.
- Rittmeister M, Kerschbaumer F. 2001. Grammont reverse total shoulder arthroplasty in patients with rheumatoid arthritis and nonreconstructible rotator cuff lesions. *Journal of Shoulder and Elbow Surgery* 10:17-22.
- Roberts CC, Ekelund AL, Renfree KJ, Liu PT, Chew FS. 2007. Radiologic assessment of reverse shoulder arthroplasty. *Radiographics* 27:223-235.
- Roberts SN, Foley AP, Swallow HM, Wallace WA, Coughlan DP. 1991. The geometry of the humeral head and the design of prostheses. *The Journal of Bone and Joint Surgery [Br]* 73:647-650.
- Robertson DM, Pierre L, Chahal R. 1976. Preliminary observations of bone ingrowth into porous materials. *Journal of Biomedical Materials Research* 10:335-344.
- Roche C, Flurin PH, Wright T, Crosby LA, Mauldin M, Zuckerman JD. 2009. An evaluation of the relationships between reverse shoulder design parameters and range of motion, impingement, and stability. *Journal of Shoulder and Elbow Surgery* 18:734-741.
- Rockwood CA Jr. 2007. The reverse total shoulder prosthesis. The new kid on the block. *The Journal of Bone and Joint Surgery [Am]* 89:233-235.
- Rothwell AG, Taylor J, Wright M, Devane P, Tobin H, Gaarkeuken P, Henwood A, Miles K, Hobbs T, Frampton C. 2007. The New Zealand joint registry nine year report January 1999 to December 2007. In. New Zealand orthopaedic association
- Saha AK. 1950. Mechanism of shoulder movements and a plea for the recognition of "zero position" of glenohumeral joint. *Indian Journal of Surgery* 12:153-165.

- Saha AK. 1971. Dynamic stability of the glenohumeral joint. *Acta Orthopaedica Scandinavica* 42:491-505.
- Saha AK. 1983. The classic. Mechanism of shoulder movements and a plea for the recognition of "zero position" of glenohumeral joint. *Clinical Orthopaedics and Related Research* 3-10.
- Sanchez-Sotelo J, Cofield RH, Rowland CM. 2001. Shoulder hemiarthroplasty for glenohumeral arthritis associated with severe rotator cuff deficiency. *The Journal of Bone and Joint Surgery [Am]* 83-A:1814-1822.
- Seitz P, Rueggsegger P, Gschwend N, Dubs L. 1987. Changes in local bone density after knee arthroplasty. The use of quantitative computed tomography. *The Journal of Bone and Joint Surgery [Br]* 69:407-411.
- Sharma GB, Debski RE, McMahon PJ, Robertson DD. 2009. Adaptive glenoid bone remodeling simulation. *Journal of Biomechanics* 42:1460-1468.
- Sharma GB, Debski RE, McMahon PJ, Robertson DD. 2010. Effect of glenoid prosthesis design on glenoid bone remodeling: adaptive finite element based simulation. *Journal of Biomechanics* 43:1653-1659.
- shouldersurgeon.com. Shoulder Replacement Surgery (arthroplasty). http://www.shouldersurgeon.com/shoulder_replacement_surgery/.
- Simovitch RW, Zumstein MA, Lohri E, Helmy N, Gerber C. 2007. Predictors of scapular notching in patients managed with the Delta III reverse total shoulder replacement. *The Journal of Bone and Joint Surgery [Am]* 89:588-600.
- Sirveaux F. 1997. La prothèse de Grammont dans le traitement des arthropathies de l'épaule à coiffe détruite. A propos d'une série multicentrique de 42 cas. *Faculté de médecine de Nancy, these de l'université de Nancy* 1:245.
- Sirveaux F, Favard L, Oudet D, Huquet D, Walch G, Mole D. 2004. Grammont inverted total shoulder arthroplasty in the treatment of glenohumeral osteoarthritis with massive rupture of the cuff. Results of a multicentre study of 80 shoulders. *The Journal of Bone and Joint Surgery [Br]* 86:388-395.
- Soballe K, Hansen ES, Brockstedt-Rasmussen H, Pedersen CM, Bunger C. 1992. Bone graft incorporation around titanium-alloy- and hydroxyapatite-coated implants in dogs. *Clinical Orthopaedics and Related Research* 282-293.
- Soininvaara TA, Miettinen HJ, Jurvelin JS, Alhava EM, Kroger HP. 2004. Bone mineral density in the proximal femur and contralateral knee after total knee arthroplasty. *Journal of Clinical Densitometry* 7:424-431.

- Stölken JS, Kinney, JH. 2003. On the importance of geometric nonlinearity in finite-element simulations of trabecular bone failure. *Bone* 33:494-504.
- Stone KD, Grabowski JJ, Cofield RH, Morrey BF, An KN. 1999. Stress analyses of glenoid components in total shoulder arthroplasty. *Journal of Shoulder and Elbow Surgery* 8:151-158.
- Sumner DR, Galante JO. 1990. Bone ingrowth. In: Envars CM, ed. *Surgery of the Musculoskeletal System*. New York: Churchill Livingstone p151-176.
- American Academy of Orthopaedic Surgeons. 2011. Rotator cuff tear
<http://orthoinfo.aaos.org/topic.cfm?topic=a00064>.
- Susan BH. 2007. *Basic biomechanics*. United States: McGraw-Hill.
- Swanson AB, de Groot Swanson G, Maupin BK, Wei JN, Khalil MA. 1986. Bipolar implant shoulder arthroplasty. *Orthopedics* 9:343-351.
- Swanson AB, de Groot Swanson G, Sattel AB, Cendo RD, Hynes D, Jar-Ning W. 1989. Bipolar implant shoulder arthroplasty. Long-term results. *Clinical Orthopaedics and Related Research* 227-247.
- Terrier A, Reist A, Merlini F, Farron A. 2008. Simulated joint and muscle forces in reversed and anatomic shoulder prostheses. *The Journal of Bone and Joint Surgery [Br]* 90:751-756.
- Tonino AJ, Davidson CL, Kloppner PJ, Linclau LA. 1976. Protection from stress in bone and its effects. Experiments with stainless steel and plastic plates in dogs. *The Journal of Bone and Joint Surgery [Br]* 58:107-113.
- Torchia ME, Cofield RH, Settergren CR. 1997. Total shoulder arthroplasty with the Neer prosthesis: Long-term results. *Journal of Shoulder and Elbow Surgery* 6:495-505.
- Torrens C, Corrales M, Gonzalez G, Solano A, Caceres E. 2009. Morphology of the scapula relative to the reverse shoulder prosthesis. *Journal of Orthopaedic Surgery (Hong Kong)* 17:146-150.
- Treharne RW. 1981. Review of Wolff's Law and its proposed means of operation. *Orthopaedic Review* 10:35-47.
- Valenti PH, Boutens D, Nerot C. 2001. Delta three reversed prosthesis for osteoarthritis with massive rotator cuff tear: long term results. In: Walch G, Boileau, P., Mole, D., editor.: *2000 Protheses d' epaule. recul de 2 a 10 ans*. Montpellier, France: Sauramps Medical.

- Valenti PH, Boutens, D., Nerot, C. 2001. Delta three reversed proshtesis for osteoarthritis with massive rotator cuff tear: long term results (> 5 years). In: Walch G, Boileau P, Mole D, editors. 2000. Shoulder Prostheses: two to ten year follow up. Montpellier: Sauramps Medical. P253-259.
- van der Helm FC. 1996. A standardized protocol for motion recordings of the shoulder. Proceedings of the First Conference of the International Shoulder Group (ISG). Delft, The Netherlands.
- van der Helm FCT. 1998. The 'reversed' glenohumeral endoprosthesis: the role of the rotator cuff muscles for stability and strength. *Journal of Biomechanics* 31 (Supplement 1):27.
- Vanhove B, Beugnies A. 2004. Grammont's reverse shoulder prosthesis for rotator cuff arthropathy. A retrospective study of 32 cases. *Acta orthopaedica Belgica* 80:219-225.
- Vasu R, Carter DR, Schurman DJ, Beaupre GS. 1986. Epiphyseal-based designs for tibial plateau components--I. Stress analysis in the frontal plane. *Journal of Biomechanics* 19:647-662.
- Viceconti M, Bellingeri L, Cristofolini L, Toni A. 1998. A comparative study on different methods of automatic mesh generation of human femurs. *Medical Engineering and Physics* 20: 1-10.
- Vince KG, Insall JN, Kelly MA. 1989. The total condylar prosthesis. 10- to 12-year results of a cemented knee replacement. *The Journal of Bone and Joint Surgery [Br]* 71:793-797.
- Walch G, Boileau P, Mole D. 2006. Reverse Shoulder Arthroplasty. Montpellier: Sauramps medical.
- Weaver JK, Chalmers J. 1966. Cancellous bone: its strength and changes with aging and an evaluation of some methods for measuring its mineral content. *The Journal of Bone and Joint Surgery [Am]* 48:289-298.
- Weinans H, Huiskes R, Grootenboer HJ. 1992. The behavior of adaptive bone-remodeling simulation models. *Journal of Biomechanics* 25:1425-1441.
- Werner CML, Steinmann PA, Gilbert M, Gerber C. 2005. Treatment of painful pseudoparesis due to irreparable rotator cuff dysfunction with the Delta III reverse-ball-and-socket total shoulder prosthesis. *The Journal of Bone and Joint Surgery [Am]* 87-A:1476-1486.

- Whiteside LA. 1993. Cementless revision total knee arthroplasty. *Clinical Orthopaedics and Related Research* 160-167.
- Wiater JM, Fabing MH. 2009. Shoulder arthroplasty: prosthetic options and indications. *Journal of the American Academy of Orthopaedic Surgeons* 17:415-425.
- Williams GR Jr, Rockwood CA Jr. 1996. Hemiarthroplasty in rotator cuff-deficient shoulders. *Journal of Shoulder and Elbow Surgery* 5:362-367.
- Wolff JD. 1892. *Das Gesetz der Transformation der Knochen*. Berlin: Hirschwald.
- Woo S. 1981. The relationships of changes in stress levels on long bone remodelling. In: Cowin SC, editor. *Mechanical properties of bone*. American Society of Mechanical Engineers Publication p107-129.
- Woodruff MJ, Cohen AP, Bradley JG. 2003. Arthroplasty of the shoulder in rheumatoid arthritis with rotator cuff dysfunction. *International Orthopaedics* 27:7-10.
- Wright TM, Hayes WC. 1979. Strain gage application on compact bone. *Journal of Biomechanics* 12:471-475.
- Zachariah SG, Sanders JE. 2000. Finite element estimates of interface stress in the trans-tibial prosthesis using gap elements are different from those using automated contact. *Journal of Biomechanics* 33:895-899.
- Zannoni C, Mantovani R, Viceconti M. 1998. Material properties assignment to finite element models of bone structures: a new method. *Medical Engineering and Physics* 20:735-740.

APPENDIX A

Summary of clinical results for reverse shoulder arthroplasty.

Authors	Implant	Pathology (Number)	Follow-up (Month)	Age (Years)	Score (/100)	Elevation (°)	Re-operation (Number)	Glenoid loosening (Number)
Sirveaux et al., 2004	Grammont	CTA: 80	44	72.8	CS: 23/66	73/138	3	Progressive radiolucency: 3 Failed glenoid prosthesis: 2
Naveed et al., 2011	Delta III	CTA: 44	39 (8-81)	81 (59-95)	ASES: 19/ (14-23/65)	55/105	4	2
John et al., 2009	Delta III	CTA: 17	24.3	67.3 (46.3-84.9)	CS: 19/59.5	-	0	0
Woodruff et al., 2003	-	CTA: 13	87	64	CS: /59	-	-	Radiolucent lines around all stems and metaglenes in 5 cases

Rittmeister and Kerschbaumer, 2001	-	RA: 8	54	60	CS: 17/63	-	-	Aseptic glenoid loosening: 2
Boileau et al., 2005	Delta	CTA: 21 Revision: 19 SF: 5	40	-	CS: 17/59	55/121	10	1
Valenti et al., 2001	-	CTA: 39	84	-	CS: 21/63	60/120	6	-
Werner et al., 2005	Delta III	CTA: 17 Revision: 41	97	68	CS: 29/64	42/100	19	3
Frankle et al., 2005	Encore	CTA: 60	33	71	ASES: 34.3/68.2	41/102	8	-
Guery et al., 2006	Delta	CTA: 51 RA:6 SF:2 Revision: 1	79.4	71	-	-	9	3

Fevang et al., 2009	Delta III	CTA: 63 RA: 118 SF:38 Others: 6	60-144	69	-	-	5 – and 10-year failure rates: 10% and 22%, respectively	13
Grassi et al., 2009	Delta III	23	42	-	CS: 22/56	65/133	2	2
De Wilde et al., 2001		Revision: 5	30	-	14/62	-	20%	-

ASES: American Shoulder and Elbow score

CTA: Cuff Tear Arthropathy

RA: Rheumatoid Arthritis

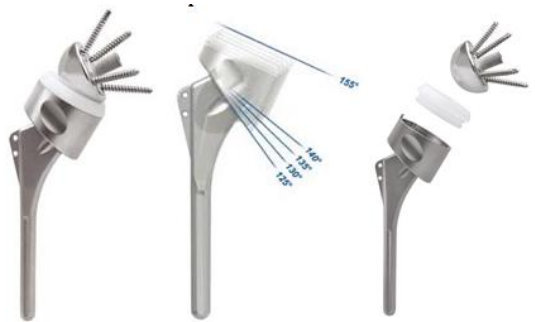
SF: Sequelae of Fracture

CS: Constant-Murley Score

APPENDIX B

Summary of various brands of reverse shoulder arthroplasties.

Company	Implant name	Picture
Depuy	Delta III	 The image shows the components of the Delta III reverse shoulder arthroplasty. It includes a humeral head with a central peg, a glenoid component with a central peg, a glenoid liner, and a humeral neck. The components are arranged in a vertical line, showing the assembly from the humeral head to the glenoid component.
Depuy	Delta CTA	 The image shows the components of the Delta CTA reverse shoulder arthroplasty. It includes a humeral head with a central peg, a glenoid component with a central peg, a glenoid liner, and a humeral neck. The components are arranged in a horizontal line, showing the assembly from the humeral head to the glenoid component.
Depuy	Delta Xtend	 The image shows the components of the Delta Xtend reverse shoulder arthroplasty. It includes a humeral head with a central peg, a glenoid component with a central peg, a glenoid liner, and a humeral neck. The components are arranged in a horizontal line, showing the assembly from the humeral head to the glenoid component.

Tornier	Aequalis®	
Tornier	Adapter	
Zimmer®	Zimmer Anatomical™ Reverse	

Zimmer®	Zimmer Trabecular Reverse	
Encore®	Encore	
Exactech	Equinox	

Stanmore	Bayley Walker	
Biomet	Verso	
Biomet	Tess modular	

FH Orthopaedics	ARROW	
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