Medical Engineering & Physics xxx (2014) xxx–xxx

Contents lists available at ScienceDirect

Medical Engineering & Physics

journal homepage: www.elsevier.com/locate/medengphy

The effect of different humeral prosthesis fin designs on shoulder stability: A computational model

Chia-Ming Chang[a,1], Wen-Lin Yeh[b,1], Wen-Chuan Chen[c], Colin J. McClean[a], Yi-Long Chen[d], Yu-Shu Lai[c,*], Cheng-Kung Cheng[a,c,**]

[a] Institute of Biomedical Engineering, National Yang-Ming University, Taipei, Taiwan
[b] Department of Surgery, Chang Gung memorial Hospital, Taoyuan, Taiwan
[c] Orthopaedic Devices Research Center, National Yang-Ming University, Taipei, Taiwan
[d] Division of Neurosurgery, Taipei City Hospital, Taipei, Taiwan

ARTICLE INFO

Article history:
Received 5 November 2013
Received in revised form 14 March 2014
Accepted 14 March 2014

Keywords:
Total shoulder arthroplasty
Fin of humeral prosthesis
Computational model
Joint stability

ABSTRACT

Humeral prostheses commonly use a fin structure as an attachment point for the supraspinatus muscle in total shoulder arthroplasty (TSA), but these fins may cause injury to the muscle during implantation, inadvertently influencing stability. In order to prevent supraspinatus injury, the effect of different humeral prostheses on shoulder joint stability needs to be investigated. A commercially available prosthesis and two modified humeral prostheses that substituted the fin structure for 2 (2H) or 3 holes (3H) were evaluated using computational models. Glenohumeral abduction was simulated and the superioinferior/anterioposterior stability of the shoulder joint after TSA was calculated. The results revealed that the 2H design had better superioinferior stability than the other prostheses, but was still less stable than the intact shoulder. There were no obvious differences in anterioposterior stability, but the motion patterns were clearly distinguishable from the intact shoulder model. In conclusion, the 2H design showed better superioinferior stability than the 3H design and the commercial product during glenohumeral joint abduction; the three prostheses show similar results in anterioposterior stability. However, the stability of each tested prosthesis was not comparable to the intact shoulder. Therefore, as a compromise, the 2H design should be considered for TSA because of its superior stability.

1. Introduction

A typical total shoulder arthroplasty (TSA) is performed by replacing the articular surface of the glenohumeral joint with a metallic humeral prosthesis and a glenoid component made of ultra-high molecular weight polyethylene (UHMWPE). TSA is a common treatment method for shoulder rheumatoid arthritis, severe osteoarthritis, complex fracture of the proximal humerus, and carcinoma [1–3]. Previous studies have reported on the effectiveness of TSA, including the increased passive and active range of motion (ROM), improved function of the shoulder joint, and pain relief [1,2]. For non-union humeral fracture, TSA can offer pain relief, but the restoration of ROM depends on the quality of intact bone tissue [3]. In addition, the condition of the supraspinatus is a major factor in the success of this procedure. Patients with an intact supraspinatus have a better prognosis and can return to normal daily life sooner. TSA with an intact supraspinatus can reach 120° in shoulder flexion at one week post-surgery, but for an injured supraspinatus, such movements cannot be performed until 8–12 weeks [4].

TSA failure typically occurs with component loosening or instability. Aldinger et al. reported an 8.6% failure rate at 2 years follow-up [5]. Most failures occur from implant problems, non-union, and abnormal biomechanical performance [6]. A larger head radius of the humeral prosthesis would elevate the rotation center against the glenoid component. This offset can lead to more severe wear in the glenoid component [6–9]. Bohsali et al. demonstrated that over-sized humeral heads will result in anterior instability and implant failure [6].

* Corresponding author at: Orthopaedic Devices Research Center, National Yang-Ming University, No. 155, Sec. 2, Linong St., Shih-Pai, Taipei 11221, Taiwan.
** Corresponding author at: Orthopedic Biomechanics Laboratory, Department of Biomedical Engineering, National Yang-Ming University, No. 155, Sec. 2, Linong St., Shih-Pai, Taipei 11221, Taiwan. Tel.: +886 2 2826 7020; fax: +886 2 2822 8557.
E-mail addresses: uesu@ortho.ym.edu.tw (Y.-S. Lai), ckcheng@ym.edu.tw (C.-K. Cheng).
[1] Authors with equal contribution.

http://dx.doi.org/10.1016/j.medengphy.2014.03.005
1350-4533/

Fig. 1. Shoulder II humeral prosthesis (Zimmer, Warsaw, Indiana).

The position of the implant is also a key factor influencing failure of TSA surgery. Previous studies showed that malpositioning between the humeral prosthesis and glenoid component will cause implant breakage [5,9]. Another factor is the condition of the interface between the bone and implant. Loosening at the bone–glenoid interface often leads to dislocation of the shoulder joint when the glenoid is no longer secured within the scapula and cannot stabilize the humeral head (named rocking-horse phenomenon) [10]. Bohsali et al. reported that instability of the shoulder joint after TSA may be due to rotator cuff injury, anterior deltoid dysfunction or anterior capsule tearing [6].

Terrier et al. evaluated the supraspinatus during abduction by finite element analysis and showed that without the supraspinatus superior displacement would increase 1.6-fold and contact force by 8% [11]. Suarez et al. revealed that an injured supraspinatus may lead to increased micromotion of the glenoid component, and an injured infraspinatus may lead to component dislocation [12]. This increased micromotion may hinder integration between the bone and implant, which could easily lead to dislocation.

Humeral prostheses commonly use a fin structure as an attachment point for the supraspinatus muscle in TSA. However, it has been reported that these fins may lead to tearing of the supraspinatus attachment [13]. The fin is typically erected on the stem of the prosthesis, and has 1–3 holes for muscle attachment (Fig. 1); the attachment of an intact supraspinatus is 23–25 mm in length and 16 mm in width [14,15]. However, it is inevitable that there will be differences in muscle orientation and position between an intact shoulder and a reconstructed one, as current designs cannot adequately replicate normal shoulder anatomy.

Impairment of the supraspinatus may lead to shoulder instability and poor performance after TSA. In addition, the fin, which is designed for reconstructing the supraspinatus, may in fact lead to muscle injury and cannot sufficiently mimic normal muscle attachment. Based on these problems, the purpose of this study was to investigate shoulder stability after implantation of a commercially available prosthesis and two newly designed humeral prostheses that forewent the fin structure in favor of 2 or 3 holes in the humeral head.

2. Method

CT images of a 74-year-old female volunteer, without musculoskeletal pathology or obvious degenerative arthritis, were used for reconstructing the bone geometry (humerus, scapula, and clavicle) and muscles (anterior/middle/posterior deltoid, supraspinatus, subscapularis, infraspinatus, and teres minor). Muscle attachments were confirmed by a senior orthopedic surgeon.

Comparisons of shoulder stability were made between the intact shoulder, TSA with commercial prosthesis, and TSA with 2 new prosthesis designs; new designs were based on the commercial product (P), Shoulder II Humeral Prosthesis (Zimmer, Warsaw, Indiana) (Fig. 1). In these 2 new designs, the fin was replaced with 2–3 holes at the insertion of the supraspinatus on the humeral head (Fig. 2). In the 2 hole design (2H), holes were added at the anterior region and posterior region for attachment. For the 3 hole prosthesis (3H), a third hole was added between the holes in the 2H design.

Kinematic analyses were performed by MSC.ADAMS R3 (MSC Software, Santa Ana, CA). Sagittal, frontal, and translation planes followed the local coordinate system put forward by the International Society of Biomechanics [16]. Abduction of the glenohumeral joint was simulated in the scapular plane, and the angle of abduction was defined as the angle between the central line of the humerus and the line through the acromion and parallel with the rib cage [17]. Boundary conditions fixed the scapula and clavicle but left the humerus unconstrained. For the intact shoulder, the articular surface was set as contact with impact function and the stiffness was defined according to which stiffness could best replicate that of intact shoulders referenced from previous studies. In addition, several contact stiffnesses were tested for simulating the contact behavior of an intact articular surface in the glenohumeral joint. Contact between the humeral prosthesis and glenoid component was set as frictional with a frictional coefficient of 0.04 [18]. Based on anthropometric data, the weight of the whole upper limb was set as 37.5 N [19]. According to previous studies, an additional weight of 10 kg was added in hand [20,21].

The muscles were simulated by 3–6 vectors in each muscle from anterior to posterior [22], and the location of these vectors was based on CT image. The magnitude of muscle forces depended on the abduction angle [11]. The ratios between muscle force vectors were estimated from the product of physiological cross-sectional area and the amplitude of the electromyography signal, as shown in Table 1 [22]. In order to ensure the supraspinatus passed around the humeral head, a pulley-spring beads system was used (Fig. 3) [23]. Shoulder stability was defined as the displacement between the center of the humeral head and center of the glenoid component, and was recorded in superioinferior and anterioposterior directions during abduction.

3. Results

A resultant contact stiffness of 1828.94 N/mm proved to be most similar to stiffness results published in previous studies [24], with a displacement difference of +0.04 mm at 45° abduction and −0.42 mm at 60° abduction (Fig. 4).

Regarding stability in the superioinferior direction (Fig. 5), all prostheses showed motion patterns similar to the intact shoulder. The intact shoulder and three prostheses move superiorly with increasing abduction and inferior after 45°. However, there were noticeable differences in maximum superior displacement, which occurred at an abduction angle of around 45° in all models. The intact shoulder has the least displacement at 3.28 mm. The commercial ‘finned’ prosthesis and the 3H had the greatest superior displacement at 7.14 mm.

Please cite this article in press as: Chang C-M, et al. The effect of different humeral prosthesis fin designs on shoulder stability: A computational model. Med Eng Phys (2014), http://dx.doi.org/10.1016/j.medengphy.2014.03.005