Impact of Scapular Notching on the Glenoid Fixation in Reverse Total Shoulder Arthroplasty-An in-Vitro and Finite Element Study

Min Zhang PhD[1,2], Sarah Junaid PhD[2, 3], Thomas Gregory MD[2, 4], Ulrich Hansen PhD[2], Cheng-Kung Cheng PhD[1]

[1] Beijing Advanced Innovation Centre for Biomedical Engineering, Beihang University, Beijing, China, 102402. School of Biological Science and Medical Engineering, Beihang University, Beijing, China, 100083;

[2]Mechanical Engineering Department, Imperial College London, UK.

[3] Engineering and Applied Sciences, Aston University, Birmingham, UK.

[4]Department of Orthopaedic Surgery, Avicenne Teaching Hospital, APHP, University ParisXIII, Bobigny, France

Min Zhang: zhangminsky123@msn.com

Sarah Junaid: s.junaid@aston.ac.uk

Thomas Gregory: tms.gregory@gmail.com

Ulrich Hansen: u.hansen@imperial.ac.uk

Cheng-Kung Cheng: ckcheng2009@gmail.com

Corresponding author: Name: Min Zhang zhangminsky123@msn.com

Cheng-Kung Cheng ckcheng2009@gmail.com

Mailing address: No. 37, Xueyuan Road, Haidian District, Beijing, China 100083

Word count for abstract: 248; Word count for the main text: 4746

Declarations of interest: none

Abstract

Background: The high incidence of scapular notching in reverse total shoulder arthroplasty (RTSA) has spurred several methods to minimize the bone loss. However, up to 93 % of RTSA with accompanying scapular notching have been reported to maintain good implant stability for over 10 years. The purpose of this study is to investigate the correlation between scapular notching and glenoid fixation in RTSA.

Methods: An in-vitro setup was used to measure the notch-induced variations of the strain on the scapular surface and the micromotion at the bone-prosthesis interface during arm abductions of 30°, 60° and 90°. Finite element analysis (FEA) was used to study the bone and screw stresses as well as the bone-prosthesis micromotion in cases of a grade 4 notch during complicated arm motions.

Results: The notch resulted in an apparent increase of inferior screw stress in the root of the screw cap and the notch-screw conjunction. However, the maximal stress (172 MPa) along the screw after notch is still much less than the fatigue strength of the titanium screw (600 MPa) under cyclic loading. The bone-prosthesis micromotion results did not present significant notch-induced variations.

Conclusions: Scapular notching will not lead to significant effects on the initial stability of glenoid component in RTSA. This finding may explain the long-term longevity of RTSA in cases of severe scapular notching. The relationship between scapular notching and weak regions along the inferior screw may explain why fractures of the inferior screw are sometimes reported in patients with RTSA clinically.

Introduction

Scapular notching is a result of mechanical impingement between the humeral cup and the scapular neck, which often leads to implant wear and the generation of polyethylene (PE) debris. The PE particles can trigger localised osteolytic reactions and further enlarge the bone notch. Scapular notching is a frequently reported complication of Grammont reverse total shoulder arthroplasty (RTSA), occurring in 44 % to 93 % of patients [1-4]. Notching can appear within the first few postoperative months of a patient undergoing RTSA and may continue to progress over time [1, 4, 5]. This condition is also sometimes accompanied by screw fracture and implant loosening [6, 7]. Thus the presence of scapular notching has long been a clinical concern [6-8] and numerous publications have reported on efforts to minimize bone-prosthesis impingement and scapular notching [9-11]. However, a recent review of longevity studies for RTSA reported that the postoperative survivorship of RTSA is 70 % at 15 years, or when viewing prosthesis failure alone as the reason for revision the survivorship rate reaches 85 % at 15 years [1]. Moreover, at a follow-up of 10 or more years, 93 % of patients with RTSA had scapular notching, 48 % of whom being grade III or IV [1]. It is not yet clear whether scapular notching is associated with implant survivorship, particularly whether a severe notch promotes aseptic glenoid loosening, which has been reported in 12 % of Grammont RTSAs [8].

Previous studies on the fixation strength of the glenoid baseplate in RTSA included in-vitro testing and finite element analysis (FEA). In-vitro testing can closely replicate the conditions in the body, but the range of arm motions is restricted and this method can only provide limited information on what is happening within the joint. Roche et al. [11] used an in-vitro setup to evaluate initial implant fixation through bone-prosthesis micromotion after scapular notching. However, only arm abduction was simulated. Finite element analysis can simulate any joint movement through a range of complicated activities and is beneficial for assessing stresses and forces that cannot be easily measured using other means. This study is aimed to use an in-vitro setup and FEA to quantitatively assess the correlation between scapular notching and glenoid fixation in RTSA. The fixation was assessed according to initial implant stability and screw stability. In-vitro testing was used to investigate the notch-induced variations of bone strain and bone-prosthesis micromotion under 30°, 60° and 90° of humeral abductions respectively. For more complex shoulder movements (lifting an object to head height and standing up from an armchair), FEA was used to further study the effect of scapular notching on the glenoid fixation with regards to screw safety, screw stability and initial implant stability with the parameters of screw stress, bone stress on the surface of the screw hole and bone-prosthesis micromotion.

Given the high incidence of scapular notching but low revision rates for RTSA, it was hypothesized that a Grade 4 scapular notch would have little effect on the stability of RTSA during the simulated daily activities.

Materials and methods

1. In-vitro Testing

Three cadaveric scapulae (provided by Science Care, USA) (Table 1) without any history of shoulder disease or surgery were used for the in-vitro evaluation. The method for preparing the cadaveric scapulae for testing is described in a previous publication by the authors [12]. The cadaveric shoulders were first taken out from a -20 °C freezer and thawed at room temperature the night before the in-vitro testing began. Then, the scapula was separated from each shoulder and soft tissues on the surface of each scapula were removed. For the purpose of setting the coordinate system with respect to the glenoid bone, the labrum on the glenoid was carefully removed. Bone strains on the scapular surface and bone-prosthesis micromotions in both the unnotched and notched conditions were measured with the aim of evaluating the effect of scapular notching on implant stability. Methods preparing and measuring these two parameters in the in-vitro testing are described below.

<u>Preparation for measuring bone strains on the scapular surface</u>

On each of the scapulae, eight uniaxial strain gauges (FLA-2-11, Tokyo Sokki Kenkyujo Co., Ltd.) were attached at approximately 10 mm and 25 mm beneath the glenoid articular surface and around the glenoid at each level (Figure 1). These two levels were chosen with the purpose of investigating strain close to, and at a small distance from the glenoid. Strain gauges on the anterior, posterior and superior surfaces of the scapula were roughly perpendicular to the glenoid articular surface. The strain gauges located on the inferior surface were orientated parallel to the lateral border. The procedure of fixing a strain gauge on the bone surface conformed to the method introduced by Miles and Tanner [13] and is detailed below. The location where a strain gauge would be attached was firstly specified and marked with a black permanent marker. Then, the periosteum on the target location for the strain gauge was cleared and the bone surface was abraded with a piece of 400 grit silicon-carbide paper. As suggested by Wright and Hayes [14], the targeted bone surface was prepared with CSM-2 degreaser, a thin layer of M-Bond catalyst, a thin layer of M-Bond 200 adhesive, and one drop of M-Bond 200 adhesive in this order (Vishay Measurements Group U.K. Ltd). Finally, one strain gauge was attached and pressed with a finger for approximately one minute on the target surface. All the strain gauges were connected to a calibrated model P3 strain recorder (accuracy 1 $\mu\varepsilon$) (Vishay Measurements Group U.K. Ltd) for strain measurements.

Setup of bone-prosthesis interface micromotion test

Referring to Figure 2 (A), each scapula was secured in a container filled with polymethylmethacrylate (PMMA) bone cement (Stryker Simplex®) at the approximately one third of bone from the medial side. The coordinate system (Figure 2 (A)) was defined in accordance with the system proposed by Terrier et al. [15], with the middle point of the glenoid fossa being the origin (O) of the coordinate system. The X-axis was orientated from posterior to anterior, the Y-axis was orientated from inferior to superior, and the Z-axis was defined as being perpendicular to the glenoid articular surface. An experienced orthopaedic shoulder surgeon implanted each shoulder joint with a Delta CTA RTSA (Depuy Synthes Company, Warsaw, USA) using the procedure detailed in the 2005 version of the Delta implant surgical guide (Depuy Synthes Company, Warsaw, USA). The relative movement (micromotion) at the bone-prosthesis interface was measured using a Linear Variable Differential Transformer (LVDT) (DP/2/S, Solartron Metrology, UK) (Resolution 0.01 μm) (Figure 2 (B)). Each LVDT was firmly fixed on the metal glenoid component in the RTSA with an external rod (Figure 2 (B)). Movement of the probe on the LVDT corresponded to the relative movement between the metal glenoid component and the position where the probe on the LVDT touches the bone. The probes were initially positioned as close as possible to the bone-prosthesis interface. Four calibrated LVDTs were fixed to the superior, inferior, anterior and posterior of the metal glenoid implant.

Measurement in the unnotched bone condition

All the scapulae with the strain gauges and RTSA were firstly used for the measurement in the unnotched bone condition. The test setup is shown in Figure 2 (B). The bone container holding the unnotched scapula was secured on the platform of an Instron machine (Instron Ltd, UK). The superoinferior direction of the scapula was aligned with the matching humeral cup (Depuy) and the pneumatic cylinder. The humeral cup was fixed to the actuator in the Instron machine (Instron Ltd, UK) and supplied the vertical force. The pneumatic cylinder was fixed with the platform of the Instron machine (Instron Ltd, UK) and applied the horizontal force. Maximum glenohumeral force values in the arm motions of 30°, 60° and 90° abductions were obtained from the study of Terrier and associates [15] (Supplementary) and executed by the pneumatic cylinder and the actuator in the Instron machine. The strain value measured by each strain gauge around the glenoid under each abduction angle was recorded. The output from each LVDT was also recorded. In order to reduce the effect of the viscoelastic properties of bone on the results, a five-minute restoration period was allowed for each scapula before the start of the next loading case [16]. Due to possible impingement between the rod for securing the inferior LVDT on the implant and the humeral cup at 30° and 60° abductions, inferior micromotions under these two conditions were not recorded. The test was repeated three times for each abduction angle and the average value was used to represent the strain and micromotion for that angle.

Measurement in the notched bone condition

After all testing of unnotched samples was complete, a Nerot-Sirveaux grade 4 inferior artificial notch (Figure 3 (A)) was hand made in each scapula with the most medial border of the notch being roughly 10 mm below the inferior rim of the glenoid component [17]. The positioning of the notch was consistent with those reported in clinical literature [6, 18]. The strain gauges used for the testing on the unnotched scapulae were used and remained in place during the notching procedure. Prior to the testing in the notched condition, the positions of the strain gauges were verified to be the same as in the unnotched condition testing. Gauges that were broken or damaged were replaced with new ones at the same positions. The notched scapulae were then moved back into the Instron machine. The same operation method of position of the bone container on the Instron platform as used in the previous testing was used. The same loading conditions (arm abduction to 30°, 60° and 90°) for the unnotched bone were applied. Strains and micromotions around the glenoid were recorded and compared to the pre-notched results. A student's t-test was used to investigate the effects of a severe notch on bone strain and micromotion. A p-value of less than 0.05 was considered statistically significant.

2. Analysis of the Effect of Scapular Notching on Implant Stability in Daily Activities with Finite Element Modelling

Before further analysis of the effect of scapular notching on the implant stability in complex daily activities with finite element modelling (FEM), the notch-induced changes in bone strain and bone-prosthesis micromotion in the arm abductions of 30°, 60° and 90° predicted from the FEM were validated with the results from the previous experiments. The believable FEM which had been validated with the in-vitro testing results would be used for the further study in daily activities.

Validation of the finite element modelling

The method of building the FEM of a scapula with a Delta CTA prosthesis was described in our previous work [19], and consists of the following steps. CT images (Table 1) of all three

Commented [JS1]: How many cycles were tested for each loading case? What was the loading frequency? 1 Hz?

Commented [JS2]: So a total of 9 tests for each scapula?

Commented [JS3]: This must be a paired student t-test? Since you compared the same sample before and after the test.

Commented [JS4]: So you already validated from a previous study? Did you publish that data? If so, would be important to reference here.

Commented [JS5]: I think you can delete this sentence as you are mentioning the same thing as the previous 2 sentences.

scapulae were used to reconstruct the geometry of the bone in Avizo 5 (Mercury Systems, Andover, USA). Each reconstructed scapula model was implanted with a Delta CTA RTSA with guidance from an experienced orthopaedic shoulder surgeon and following the surgical technique for this type of prosthesis (2005 revision, Depuy Synthes Company, Warsaw, USA). The glenoid component and screw positions of each scapula in the FEM was consistent with those of the same bone in the previous cadaveric testing. Each FEM of an implanted scapula was used to create two models: with and without a scapular notch (Figure 3 (B)). For each notched model, a Nerot-Sirveaux grade 4 notch [6] was simulated to be consistent with the notch created in the same cadaveric scapula. All the notched and unnotched FEMs were imported into the software MSC Marc (MSC Software Corporation, Santa Ana, USA) for finite element (FE) pre-processing and modelling. Methods of FEM in MSC Marc for the notched and unnotched bone were the same. Each model of the bone with a Delta CTA RTSA was composed of isotropic and linear elastic tetrahedral elements. The material properties of each element in the FE model of the scapula were determined by the CT values and the density-modulus relationship proposed by Carter and Hayes [20]. The FEM of the three cadaveric scapulae in the intact condition were validated against results from in-vitro cadaveric testing in our previous work [12]. The Young's modulus of the cobalt-chrome baseplate and the glenohumeral sphere was set as 210 GPa [21], and that of the titanium screws for securing the glenoid component were set as 110 GPa [21]. The Poisson's ratio for all the elements was 0.3. The bone-prosthesis interface was unbonded with a friction coefficient of 0.4 [21], which has been shown to be consistent with in-vitro conditions [22]. The screws were assumed to provide firm fixation, and thus to be rigidly bonded with the bone. The FE models used the same coordinate system, arm abduction angles and boundary conditions as the in-vitro testing. Similarly, the strain in the FE models was recorded at the same points where the strain gauges were located in the in-vitro test and in the same direction as the gauge orientation. The relative movement between the glenoid baseplate and the position of the LVDT probe on the bone was also calculated. Convergence testing for each analyzed scapula showed that a mesh size of 1.5 mm in the region of the glenoid and 3.0 mm in the remaining bone was able to produce reliable strains and micromotions [12]. The mean notch-induced strain change in the position of each strain gauge for the three scapula models was calculated. In addition, the bone-prosthesis micromotion in each direction of the glenoid from the three subjects was also averaged. Because of unavoidable differences between the in-vitro and FE models in accordance to notch shape, implant position, and screw location, a comparison was made between the in-vitro and FE models to assess the effect of scapular

Commented [JS6]: What about the UHMWPE numeral cup lining?

Commented [JS7]: What was the % convergence?

notching on strain and bone-prosthesis micromotion. This comparison was used to assess the accuracy of the FE model.

Effect of scapular notching on implant stability in daily activities

After validating the FE models as described above, the models were used to simulate two complicated shoulder movements: 1. lifting a block to head height, and 2. standing up from an armchair. These two activities have been reported to produce the greatest glenohumeral contact forces and anteroposterior shear forces out of 13 daily shoulder activities in patients with RTSA [23]. The force values for these two activities presented by Kontaxis et al. [23] were used (Supplement). Principal stresses along the screws and on the surface of the screw holes as well as bone-prosthesis micromotions were evaluated. A student's t-test was used to assess the effect of scapular notching on the stability of the glenoid implant. A p-value of less than 0.05 was considered significant.

Results

In-vitro testing

The strains recorded from each strain gauge from the three cadaveric scapulae were averaged and are presented in Figure 4 (A) and (B). The results indicated that the presence of a notch did not lead to significant effects on the bone strains around the scapula (p=0.86). While the magnitude of the strain changes varied depending on gauge location and activity being performed. The loading-dependent characteristic of notch-induced bone strain presents a necessity for more realistic and complicated loading simulations.

Mean bone-prosthesis interface micromotions in each LVDT position from the three subjects were presented in Figure 4 (C). It is shown that the notch did not significantly impact the bone-prosthesis relative movements around the glenoid component (p=0.84).

Validation of finite element modelling with the experimental measurements

Notch-induced strain variations from the FE models of the three subjects were averaged in each strain gauge position and illustrated with the in-vitro results in Figure 5. The FE results for the notch-induced strain variations around the glenoid displayed a consistent trend with those from the in-vitro testing. Both the FE and experimental data presented an apparent notch-induced reduction in strain variations from the position close to the glenoid to that far away around the glenoid. The maximal difference between the FE notch-induced variation around the glenoid and that obtained from the experimental results was 14 με and occurred in the lateral posterior glenoid surface.

The comparison between the FE and experimental micromotion variations indicated that the FE model of scapulae can predict the same levels of micromotions to the in-vitro testing. The maximum FE-experimental difference in the notch-induced micromotion variations around the glenoid was 0.5 μm.

Commented [JS8]: This seems very small, is it definitely 0.5 microns?

Effect of scapular notching on implant stability in daily activities

Distributions of the maximum principal stress along the inferior screw from the three subjects before and after notching were predicted with FE analysis. It showed the same trend of stress distribution along the inferior screw for the three subjects. One subject's stress distribution when standing up from an armchair are illustrated in Figure 6 (A). It exhibited that high stresses appeared in the root of screw cap. The scapular notch resulted in an increase in the maximum principal stress for all the three subjects. The averaged maximum principal stresses from the three subjects in cases of a Nerot-Sirveaux grade 4 inferior notch reached 72.5 MPa (SD 4.8 MPa) while lifting a block to the head height and 172.0 MPa (SD 6.2 MPa) while standing up from an armchair. When averaging the notch-induced change of the maximum principal stresses on each cross section along the inferior screw at 2 mm intervals, all three subjects' results presented consistent trends in the two shoulder activities. One subject's results were illustrated in Figure 6 (B). Both simulated arm activities led to apparent notch-induced increase of the maximum principal stress in the root of the screw cap and the conjunction between the notch and the inferior screw. The results also indicated that large glenohumeral contact force resulted from the activity of standing up from an armchair led to the most apparent increase of stress after scapular notching.

Distribution of the maximum principal stress on the surface of the inferior screw hole before and after notching was used to assess the possibility of notch-induced bone fracture. The results from all three subjects showed the same stress distribution. High stresses appeared close to the screw tip as shown in Figure 7, which is one subject's stress distributions before and after scapular notching when standing up from an armchair. In addition, it was found that the bone stress on the surface of the inferior screw hole after scapular notching increased, with the mean maximum principal value for the three subjects in the two simulated shoulder joint activities being 3.3 MPa (SD 0.9).

Micromotion distributions at the bone-prosthesis interface for the three subjects before and after scapular notching were calculated. Figure 8 presents distributions of one subject's bone-prosthesis micromotion when rising from an armchair. The results indicated that there were not significant variations in the bone-prosthesis micromotion ($p=0.87$). The mean peak notch-induced increase of bone-prosthesis micromotion for the three subjects was 2.7 μm (SD 0.6) when standing up from an armchair and 1.2 μm (SD 0.1) when lifting a block to head height.

Discussion

Both in-vitro testing and FE analysis methods were utilized to investigate the effect of inferior scapular notching on the glenoid fixation in RTSA. The most important finding is that (1) notch-induced stress variation was loading and location dependent. (2) An inferior scapular notch led to apparent increase in the root of the screw cap as well as the screw-notch interface. (3) The bone stress on the surface of the inferior screw hole increased after scapular notching. (4) A severe inferior scapular notch resulted in few variations in the micromotion at the bone-prosthesis interface during daily arm activities.

Strains on the surface of three cadaveric scapulae before and after scapular notching under 30°, 60° and 90° arm abductions were measured using in-vitro testing. The results showed that notch-induced strain variation was loading and location dependent. The region close to the notch was generally impacted by the notch more than the region far away from the notch. It is possibly because no bone supports the inferior screw in the region of bone loss, and thus bone close to the notch suffered more stresses.

The FEM for predicting the strains and micromotions in the bone condition of an inferior scapular notch were validated with the completed in-vitro testing. The maximum FE-experimental deviation of the notch-induced strain variations was 14 $\mu\varepsilon$, and that of the bone-prosthesis micromotion changes was 0.5 µm. The differences between the FE predictions and the experimental results could have been induced by the unavoidable inconsistent notch geometries and positions of the glenoid prosthesis in the FEM to those in the in-vitro testing. The slight changes of the location of the glenoid component in RTSA and the notch surface created by hand may have led to variations of force transmitted from the glenoid prosthesis to the bone. The contact condition at the interface between the non-locked screws (the anterior and posterior screws) and the bone is possibly another explanation for the FE-experimental variations. In the FE model, the non-locked screws were assumed firmly secured. The real condition may not have been the same as the assumption in the FE modeling, and may have led to different experimental results. However, the FEM of the three scapulae when they were in the intact condition had been validated against the results from the in-vitro cadaveric testing in our previous work [12]. Moreover, the notch-induced strain variations predicted from the FEM displayed a consistent trend to those measured from the in-vitro testing in the same loading and fixation conditions. Thus, the FEM was able to predict believable strain variations induced by the inferior scapular notch. The maximum notch-induced change of bone-prosthesis micromotion (0.5 μm) was much lower than the threshold for bone integration (50 μm) [24], thus the FEM was able to predict the effect of the inferior scapular notching on the bone ingrowth after RTSA implantation.

With the validated FEM of implanted scapulae, two complicated physical daily shoulder activities were simulated. The predicted notch-induced stress changes along the inferior screw depicted that a notch led to apparent increase of screw stress in the root of the screw cap and the screw-notch interface. The two regions of big notch-induced stress variation predicted from the FEM are a line with the positions of screw fractures reported from the clinical practices [6,25]. The agreement between FE prediction and the clinical observation presented that the FEM of an implanted scapula with a scapular notch could predict believable results when the effects of the severe notch on the inferior screw were analyzed. In this study, the predicted maximal principal stress of the inferior screw in the bone condition of a Nerot-Sirveaux grade 4 notch was 172 MPa and occurred when standing up from an armchair, which resulted in the largest glenohumeral joint contact force in the 13 daily arm activities reported by Kontaxis [23]. This value was much lower than the fatigue strength of the inferior screw material (titanium, 600 MPa) in daily life [26]. It documented that the inferior screw in a scapula implanted with a RTSA was comparatively safe even in the bone condition of a severe inferior scapular notch. The incidence of breakage of the inferior screw accompanied with the scapular notching in clinical practice was 2% reported from Sirveaux and associates [6] and 1% in the Grassi and co-workers' study [25]. The screw fracture was possibly caused by the movement of the humeral component into the notch and the impact to the inferior screw [27]. It may also be induced by the stress concentration in the inferior screw thread, reducing the screw fatigue life. Some incorrect surgical techniques, such as overtensioning of deltoid muscle observed in clinical practice [8] , could be another factor leading to screw fracture in the case of scapular notching. The results of this study documented that the notch-induced stress variation was loading-dependent. Overtensioning of deltoid muscle may increase the glenohumeral contact force and induce higher stresses than our predictions. Generally, the inferior screw is comparatively safe even in the presence of a severe inferior notch. However, if the inferior screw breaks, the root of the screw cap and the bone-notch interface are the regions of highly potential risk.

The maximal principal stresses on the surface of the inferior screw hole after scapular notching were analyzed. The peak stress in the cancellous bone on the surface of the inferior screw hole reached 3.3 MPa (SD 0.9). This value was lower than the regional ultimate strength (13 MPa - 110 MPa) [28-30] and failure strength (9 MPa - 15 MPa) [28] of cancellous bone, but on the same

Commented [JS9]: Did you include normal joint compression in your experiments and FE simulation?

Commented [JS10R9]: If not, perhaps this might be another explanation as to why some screws fail level as the fatigue failure strength (3.57 MPa) for the epiphyseal cancellous bone with Young's modulus of 400 MPa after 1 million cycles [31]. The finding suggests that scapular notching may increase the risk of bone fracture close to the inferior screw hole and may explain the possible screw loosening in the presence of scapular notching, which were reported to cover 40% of glenoid loosening [6].

Micromotions at the bone-prosthesis interface were analysed to assess the effects of a severe inferior scapular notch on the initial stability of glenoid prosthesis in RTSA. The results showed that few variations in the notch-induced bone-prosthesis micromotions were observed after scapular notching, with a peak increase of approximately 2.7 μm (SD 0.6) when rising from an armchair and 1.2 μm (SD 0.1) when lifting a block to head height. The maximum predicted bone-prosthesis micromotion of the implanted scapula accompanied by a severe scapular notch was 59.8 μm, which is on the same level as the threshold for bone growth (50 μm) [24] and predicted a generally effective bone-prosthesis environment for the bone osteointegration. This finding was in line with the report from Nyffeler et al. [18], in which an eight-month follow-up retrieved Delta Ш RTSA in the scapula accompanied by a Grade 3 inferior notch was generally well supported by the bone biological attachments.

There are several limitations. Firstly, the unavoidable inconsistence in the notch geometries, the positions of the glenoid prosthesis and screw fixations, between the experiment and the FEM, limit the precision of statistical comparison. In our previous work, the FEM of the three cadaveric scapulae in the intact condition were validated against results from in-vitro cadaveric testing [12]. Moreover, the differences between the FE predicted notch-induced variations of inferior screw stress and those from experiments were much smaller than the fatigue strength of the titanium screw material. The FE-experimental variations of bone-prosthesis micromotions were also much lower than the threshold for bone ingrowth. Therefore, the FEM of a scapula accompanied by an inferior notch can produce a consistent result to the reality. Secondly, only severe inferior notch (Nerot-Sirveaux grade 4) was used in this study, although scapular notches are also observed in the anterior and posterior scapulae [17]. Because an inferior notch is one of the most significant with regards to bone loss, as well as screw fractures that were reported in the bone being associated with the inferior scapula notch in clinic [6, 25], a severe inferior scapular notch is appropriate in assessing the implant fixation. Thirdly, the assessment of bone fracture was limited by the use of the fatigue failure value from the bovine cancellous bone with Young's modulus of 400 MPa [31]. A proper fatigue failure limitation from scapular trabecular bone in daily life would improve the accuracy of our assessment. Finally, the use of

LVDTs precluded the ability to measure the relative bone-prosthesis movement in the inferior scapula. Future iterations of this test paradigm may use slightly different motion capture techniques (i.e. Laser extensometer) to capture the displacements in all the regions around the glenoid (anterior, posterior, inferior, superior).

Conclusion

This study is aimed to investigate effects of scapular notching on the fixation of glenoid component in Grammont RTSA. Both the in-vitro testing and FEM results presented few notch-induced variations of bone-prosthesis micromotions. The stress values along the inferior titanium screw in the implanted scapula accompanied by an inferior notch were lower than the screw fatigue strength (600 MPa) and documented that the inferior screw was comparatively safe even in the presence of a severe inferior notch on the scapular neck. These findings may explain the long-term longevity of RTSA in the case of severe scapular notching. However, the relationship between the inferior scapular notch, the weak regions along the inferior screw (the root of the screw cap and the screw-notch conjunction) and the slightly notch-induced increase of the bone stresses on the surface of the inferior screw hole, is possibly an explanation for the positions of the inferior screw fracture and the screw loosening accompanied by scapular notching.

Acknowledgement

We wish to thank Dr Kontaxis from Newcastle University for the provision of force data for Delta reverse shoulder arthroplasty in physiological daily activities.

Reference

1. Ernstbrunner L, Andronic O, Grubhofer F, Camenzind RS, Wieser K and Gerber C. Long-term results of reverse total shoulder arthroplasty for rotator cuff dysfunction: a systematic review of longitudinal outcomes. J Shoulder Elbow Surg 2019; 28(4): 774-781. doi:10.1016/j.jse.2018.10.005
2. Favard L, Levigne C, Nerot C, Gerber C, De Wilde L and Mole D. Reverse prostheses in arthropathies with cuff tear: are survivorship and function maintained over time? Clin Orthop Relat Res 2011; 469(9): 2469-2475. doi:10.1007/s11999-011-1833-y
3. Gerber C, Canonica S, Catanzaro S and Ernstbrunner L. Longitudinal observational study of reverse total shoulder arthroplasty for irreparable rotator cuff dysfunction: results after 15 years. J Shoulder Elbow Surg 2018; 27(5): 831-838. doi:10.1016/j.jse.2017.10.037
4. Gruber S, Schoch C and Geyer M. The reverse shoulder arthroplasty Delta Xtend: mid-term results. Der Orthopade 2017; 46(3): 222-226. doi:10.1007/s00132-016-3355-5
5. Lévigne C, Garret J, Boileau P, Alami G, Favard L and Walch G. Scapular notching in reverse shoulder arthroplasty. J Shoulder Elbow Surg 2008; 17(6): 925-935. doi:10.1016/j.jse.2008.02.010
6. Sirveaux F, Favard L, Oudet D, Huquet D, Walch G and Mole D. Gammont inverted total shoulder arthroplasty in the treatment of glenohumeral osteoarthritis with massive rupture of the cuff: Results of a multicentre study of 80 shoulders. Bone Joint J 2004; 86(3): 388-395.
7. Farshad M and Gerber C. Reverse total shoulder arthroplasty--from the most to the least common complication. International Orthopaedics 2010; 34: 1075-1082. doi:10.1007/s00264-010-1125-2
8. Boileau P. Grammont reverse prosthesis: design, rationale, and biomechanics. J Shoulder Elbow Surg 2005; 14: 147-161.

9. Chou J, Malak SF, Anderson IA, Astley T and Poon PC. Biomechanical evaluation of different designs of glenospheres in the SMR reverse total shoulder prosthesis: Range of motion and risk of scapular notching. J Shoulder Elbow Surg 2009; 18(3): 354-359. doi:10.1016/j.jse.2009.01.015
10. Edwards TB, Trappey GJ, Riley C, Ơ Connor DP, Elkousy HA and Gartsman GM. Inferior tilt of the glenoid component does not decrease scapular notching in reverse shoulder arthroplasty: results of a prospective randomized study. J Shoulder Elbow Surg 2012. doi:10.1016/j.jse.2011.08.057
11. Roche CP, Stroud NJ, Martin BL, Steiler CA, Flurin P-H, Wright TW, et al. The impact of scapular notching on reverse shoulder glenoid fixation. J Shoulder Elbow Surg 2013; 22(7): 963-970. doi:10.1016/j.jse.2012.10.035
12. Zhang M. Effects of Scapular Notching and Bone Remodelling on Long-Term Fixation of the Glenoid Component in Reverse Shoulder Arthroplasty. 2012. PhD, Thesis. Imperial College, London, UK.
13. Dabestani M. In *vitro* strain measurement in bone. In: Miles AW and Tanner KE, editor. Strain Measurement in Biomechanics. London: Chapman & Hall; 1992. p. 58-69. (ISBN No. 978-94-011-2330-3).
14. Wright TM, Hayes, W. C. Strain gage application on compact bone. J Biomech 1979; 12(6): 471-475.
15. Terrier A, Reist, A., Merlini, F., Farron, A. Simulated joint and muscle forces in reversed and anatomic shoulder prostheses. Bone Joint J 2008; (90-B): 751-756. doi:10.1302/0301-620X.90B6
16. Chong D. Biomechanical Analysis of Fixation and Bone Remodelling of Total Knee Replacement. 2009. PhD, Thesis. Imperial College, London, UK.

17. Simovitch RW, Zumstein MA, Lohri E, Helmy N and Gerber C. Predictors of scapular notching in patients managed with the Delta III reverse total shoulder replacement. J Bone Joint Surg Am 2007; 89(3): 588-600. doi:10.2106/JBJS.F.00226
18. Nyffeler RW, Werner CM, Simmen BR and Gerber C. Analysis of a retrieved Delta III total shoulder prosthesis. Bone Joint J 2004; 86: 1187-1191.
19. Zhang M, Junaid S, Gregory T, Hansen U and Cheng C-K. Effect of baseplate positioning on fixation of reverse total shoulder arthroplasty. Clin Biomech 2019; 62: 15-22. doi:10.1016/j.clinbiomech.2018.12.021
20. Carter DR and Hayes WC. The compressive behaviour of bone as a two-phase porous structure. The Journal of Bone & Joint Surgery American Volume 1977; 59: 954-962.
21. Hopkins AR, Hansen, U. N., Bull, A. M. J., Emery, R., Amis, A. Fixation of the reversed shoulder prosthesis. J Shoulder Elbow Surg 2008; 17(6): 974-980. doi:10.1016/j.jse.2008.04.012
22. Harman M, Frankle, M., Vasey, M. and Banks, S. Initial glenoid component fixation in “reverse” total shoulder arthroplasty-A biomechanical evaluation. J Shoulder Elbow Surg 2005; 14: 162S-167S. doi:10.1016/j.jse.2004.09.030
23. Kontaxis A. Biomechanical analysis of reverse anatomy shoulder prosthesis. 2010. PhD, Thesis. Newcastle University, Newcastle UK.
24. Pilliar RM, Lee, J. M., Maniatopoulos, C. Observations on the effect of movement on bone ingrowth into porous-surfaced implants. Clin Orthop Relat Res 1986; 208: 108-113.
25. Grassi FA, Murena L, Valli F and Alberio R. Six-year experience with the Delta III reverse shoulder prosthesis. J Orthop Surg (Hong Kong) 2009; 17(2): 151-156. doi:10.1177/230949900901700205
26. Wright T, Robert F and Closkey J. Properties of biomaterials used in joint replacements. In: Shanbhag A, Rubash H and Jacobs J, editor. Joint Replacement and Bone Resorption

Pathology, Biomaterials, and Clinical Practice. New York: Taylor & Francis Group; 2006. p. 123-145. (ISBN No. 10：0-8247-2954-4).

27. Roberts CC, Ekelund AL, Renfree KJ, Liu PT and Chew FS. Radiologic assessment of reverse shoulder arthroplasty. Radiographics 2007; 27(1): 223-235. doi:10.1148/rg.271065076
28. Anglin C, Tolhurst P, Wyss UP and Pichora DR. Glenoid cancellous bone strength and modulus. J Biomech 1999; 32(10): 1091-1097.
29. Frich LH, Jensen NC, Odgaard A, Pedersen CM, Søjbjerg JO and Dalstra M. Bone strength and material properties of the glenoid. J Shoulder Elbow Surg 1997; 6(2): 97-104.
30. Mimar R, Limb D and Hall RM. Evaluation of the mechanical and architectural properties of glenoid bone. J Shoulder Elbow Surg 2008; 17(2): 336-341. doi:10.1016/j.jse.2007.07.024
31. Michel MC, Guo X-DE, Gibson LJ, McMahon TA and Hayes WC. Compressive fatigue behavior of bovine trabecular bone. J Biomech 1993; 26(4-5): 453-463.