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Finite element modelling of the glenohumeral capsule can help assess the tested region during a clinical exam

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The objective of this research was to examine the efficacy of evaluating the region of the glenohumeral capsule being tested by clinical exams for shoulder instability using finite element (FE) models of the glenohumeral joint. Specifically, the regions of high capsule strain produced by glenohumeral joint positions commonly used during a clinical exam were identified. Kinematics that simulated a simple translation test with an anterior load at three external rotation angles were applied to a validated, subject-specific FE model of the glenohumeral joint at 60° of abduction. Maximum principal strains on the glenoid side of the inferior glenohumeral ligament (IGHL) were significantly higher than the maximum principal strains on the humeral side, for all three regions of the IGHL at 30° and 60° of external rotation. These regions of localised strain indicate that these joint positions might be used to test the glenoid side of the IGHL during this clinical exam, but are not useful for assessing the humeral side of the IGHL. The use of FE models will facilitate the search for additional joint positions that isolate high strains to other IGHL regions, including the humeral side of the IGHL.

Keywords: inferior glenohumeral ligament; soft tissue mechanics; strain; finite element

Introduction

Approximately 5.6 million people will dislocate their glenohumeral joint (Hovelius 1982; Nelson and Arciero 2000) during their lifetime and 80% of these dislocations will occur in the anterior direction (Cave et al. 1974). Common injuries resulting from anterior dislocation are detachment of the inferior glenohumeral ligament (IGHL) from the anterior glenoid and labrum (Bankart 1923, 1938) and humeral avulsion of the glenohumeral ligaments (Warner and Beim 1997; Bokor et al. 1999; Schippinger et al. 2001; Bui-Mansfield et al. 2002; Chhabra et al. 2004; Richards and Burkhart 2004; Sailer and Imhof 2004). Physical diagnostic exams are the most crucial step for diagnosis of the location of injury to the capsule (Matsen 1991; Pollock and Bigliani 1993; Mallon and Speer 1995; Brenneke et al. 2000), but the exams are relatively imprecise and the glenohumeral joint positions used for these exams are not standardised between physicians. Treatments for these injuries depend on the region of the capsule that is injured (Gerber and Ganz 1984), but misdiagnosis of the injured region has been blamed for over 38% of recurring injuries (Hawkins and Hawkins 1985; Cooper and Brems 1992; Lusardi et al. 1993).

During these exams, clinicians apply forces to the humerus to translate the humeral head with respect to the glenoid. These forces also produce strains in the glenohumeral capsule (Malicky et al. 2002; Moore et al. 2008b), which is the primary passive stabiliser of the glenohumeral joint. The magnitudes of the resulting translations are then graded (Rockwood et al. 1998). Assessments are based on the application of a manual maximum force so that a firm end point is reached, restricting further translation (Harryman et al. 1992; Lippitt and Matsen 1993; Rockwood et al. 1998). The orientation of the glenohumeral joint has been shown to influence both the magnitude of the translations (Moore et al. 2004) and the clinician's diagnostic reproducibility (Levy et al. 1999; Tzannes et al. 2004). Finally, as the external rotation angle is increased patients often feel a sense of apprehension and/or discomfort caused by the exam (Gerber and Ganz 1984; Silliman and Hawkins 1993; Lo et al. 2004). Currently, the patient's indication of apprehension and/or pain is useful for the physician to help diagnose the injury, but this method is purely subjective, depending as much on the patient's pain threshold as on the extent of the injury.

Poor clinical outcomes, inconsistent clinical exams and complex glenohumeral capsule anatomy have motivated researchers to investigate the function of the specific regions of the glenohumeral capsule by evaluating their strain distributions (Turkel et al. 1981; Bigliani et al. 1992; Brenneke et al. 2000; Malicky et al. 2001; Moore 2008b).

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Subject-specific finite element (FE) modelling of the glenohumeral joint is a useful tool for predicting capsule strains (Debski et al. 2005; Ellis et al. 2007; Moore et al. 2008a) and this method is able to predict experimentally measured strains (Ellis et al. 2006; Moore et al. 2008a).

Using strain to identify positions in which connective soft tissues stabilise a diarthrodial joint has been used extensively for the anterior cruciate ligament (ACL) of the knee (Henning et al. 1985; Renstrom et al. 1986; Woo et al. 1987; Butler 1989; Howe et al. 1990). These studies led to the development of clinical exams (Katz and Fingeroth 1986) to diagnose knee instability and injury to the ACL. It is generally accepted that an anterior load should be applied to the tibia while the knee is at 30° of flexion to test for ACL injury. In this position, the injured knee will usually have increased translation compared to the contralateral, uninjured knee. Although, clinical exams for the ACL are standardised and commonly used, extensive research was needed for their development. The starting point was to improve the understanding of knee positions that strained the ACL. Due to the complexity of the strains in the glenohumeral capsule during joint motion (Malicky et al. 2001; Malicky et al. 2002; Moore et al. 2008b), a method to correlate glenohumeral joint positions and the capsule strains produced by these positions is needed. As with the ACL, identifying the positions in which the glenohumeral capsule is strained and where those strains occur in the capsule is the first step to developing clinical exams.

The objective of this research is to conduct a feasibility study to examine the efficacy of developing clinical exams using FE models of the glenohumeral joint by locating the regions of highly strained capsule tissue at multiple joint positions. The long-term research goal is to find joint positions for clinical exams that isolate the region of the capsule being tested, so that accurate diagnoses can be made, without subjective patient input and causing discomfort to the patient. Based on a qualitative analysis of previously reported experimental strains in the glenohumeral capsule (Moore et al. 2008b), it was hypothesised that the simple translation test with an anterior load would strain the glenoid side of the IGHL in the FE model more than the humeral side at 30° and 60° of external rotation. Data addressing this hypothesis could suggest that these joint positions would be ideal for diagnosing injuries to this frequently injured region of the tissue.

Methods

The details of the construction of the subject-specific FE model of the glenohumeral joint used in this study and its validation with experimental strain data were reported previously (Moore et al. 2008a); a brief description follows. The geometry of the humerus, scapula and capsule was obtained from a computed tomography (CT) data set of a shoulder cadaver specimen (male, 45 years old, left) while the humerus and scapula were in the reference position and the capsule was in its reference configuration. For the FE model, bones and humeral cartilage were represented with rigid bodies and the glenohumeral capsule was modelled with shell elements. Based on material testing of the capsular tissue from this shoulder, an isotropic hypoelastic constitutive model with regionally varying subject-specific elastic modulii and Poisson's ratios of 0.495 were used to represent the three regions of the IGHL (anterior band (AB-IGHL), axillary pouch, and posterior band (PB-IGHL)) and the anterior-superior and posterior capsule regions. Model validation was performed by comparing predicted strains from the FE model to experimentally measured strains during kinematics that simulated the simple translation test. Eight of the 11 sampling regions were found to be within two times the repeatability of the experimental strain measurements ($\pm 7\%$).

For this study, the kinematics applied to the FE model simulated the simple translation test with an anterior load performed at three external rotation angles that are commonly used by physicians to examine anterior stability. To experimentally simulate the clinical exam, a cadaveric shoulder specimen (same as that used for the validated FE model) was mounted in a robotic/universal force-moment sensor testing system that has been extensively used previously (Debski et al. 2005; Ellis et al. 2007; Moore et al. 2008a, 2008b). The humerus was secured within a thick-walled aluminium cylinder and fixed in a custom clamp mounted to the base of the system. The scapula was rigidly attached to the end-effector of the manipulator through another specially designed clamp and the universal force-moment sensor. The coordinate system of the robotic/universal force-moment sensor testing system was then defined as the anatomic coordinate system of the glenohumeral joint as previously described (Debski et al. 1999; Burkart and Debski 2002).

The initial joint orientation in the testing system was 60° of glenohumeral abduction, 0° of horizontal abduction and 0° of external rotation. The horizontal abduction angle was held constant throughout the entire experimental protocol. Force control was then used to apply a 22 N compressive load (medially directed) to the humerus while the forces in the two orthogonal directions were minimised (~ 0 N). This centred the humeral head within the glenoid cavity and determined the joint position at 60° of glenohumeral abduction and 0° of external rotation. At this joint position, a 25 N anterior load was applied to the humerus, while maintaining the 22 N compressive force, and the resulting kinematics were recorded by recording the locations of registration blocks attached to the humerus and scapula with an external digitiser (Microscribe 3DX, Immersion Corporation, San Jose, CA, USA). Preliminary testing indicated that, with the skin and musculature removed, an anterior load of 25 N, while maintaining a compressive load of 22 N, would translate the humeral head to the edge of the glenoid without resulting in dislocation or subluxation.

To simulate the clinical exam at 30° and 60° external rotation, an increasing moment with a maximum of 3 Nm was applied to the humerus about its longitudinal axis while maintaining the 22 N joint compressive force until the joint positions corresponding to 30° and ~60° external rotation, respectively, were reached. At these joint positions, the 25 N anterior load was applied to the humerus, while maintaining the 22 N compressive force, and the resulting kinematics were recorded.

The kinematics at each external rotation angle, recorded by the external digitiser during testing, were input into the FE code as previously described (Debski et al. 2005; Ellis et al. 2007; Moore et al. 2008a). The coordinates of the registration blocks in both the CT and kinematic datasets allowed for correlation of the two datasets. Motion was described during the FE simulations using incremental translations and rotations (Maker 1995) from the reference position to the positions during the simple translation test, based on the experimental measurements of the locations of the registration blocks on the humerus and scapula in each position. The non-linear FE code NIKE3D was used for all analyses (Maker 1995). LSPOST (Livermore Software Technology Corporation, Livermore, CA, USA) was used to visualise and process the predicted strains.

To compare strains within regions of the IGHL, the AB-IGHL, axillary pouch and PB-IGHL were divided midway between their glenoid and humeral insertion sites, yielding a total of six IGHL sub-regions for analysis (Figure 1). The nodal maximum principal strains in each sub-region were averaged for each external rotation angle. An unbalanced GLM two-way ANOVA procedure in SPSS (SPSS, Inc., Chicago, IL, USA) was used to compare regional strains at each rotation angle. Post-hoc comparisons were performed using the Tukey test. Statistical comparisons were only considered to be significant if the differences were statistically significant ($p < 0.05$) and the average strain difference was greater than the repeatability of the experimental strain measurements (±3.5%).

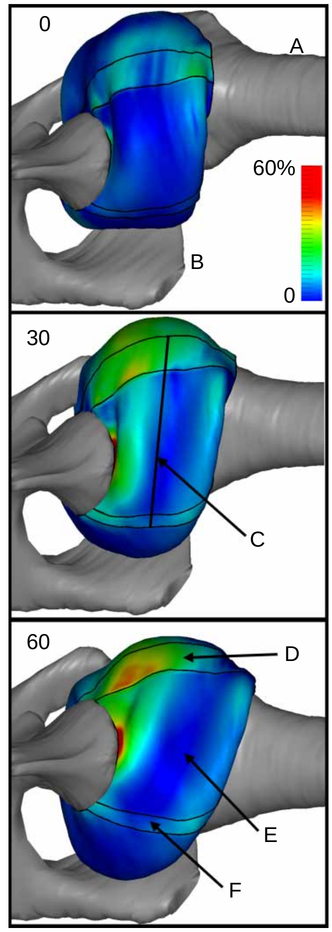

Figure 1. Inferior view (left shoulder) of fringe plots of IGHL first principal strains at 60° of abduction and full anterior translation at 0°, 30° and 60° of external rotation. A, Humerus; B, glenoid; C, IGHL mid-line; D, AB-IGHL; E, axillary pouch and F, PB-IGHL. The glenoid side of the IGHL is consistently loaded more than the humerus side at 30° and 60° of external rotation.

Results

While IGHL strains were relatively small and more evenly dispersed between the glenoid and humeral sides of the IGHL during FE simulations of the simple translation test at 0° of external rotation, more tissues were strained with much higher peak strains on the glenoid side of the IGHL when the simple translation test was simulated at 30° and 60° of external rotation (Figure 1). The end result of these loading conditions was highly strained tissue on the glenoid side of the IGHL, especially the AB-IGHL, with slack, essentially non-strained tissue on the humeral side of the IGHL, especially the PB-IGHL.

Quantitative comparisons between the glenoid and humeral sides of the IGHL support the qualitative evaluations. Maximum principal strains on the glenoid side of the IGHL were significantly higher than the maximum principal strains on the humeral side when the simple translation test was simulated at 30° and 60° of external rotation ($p < 0.05$ for all three regions at both rotation angles), but not at 0° of external rotation (Figure 2). Maximum principal strains on the glenoid side of the AB-IGHL were more than 1.5 times higher and nearly four times higher than on the humeral side at 30° and 60° of external rotation, respectively. The maximum principal

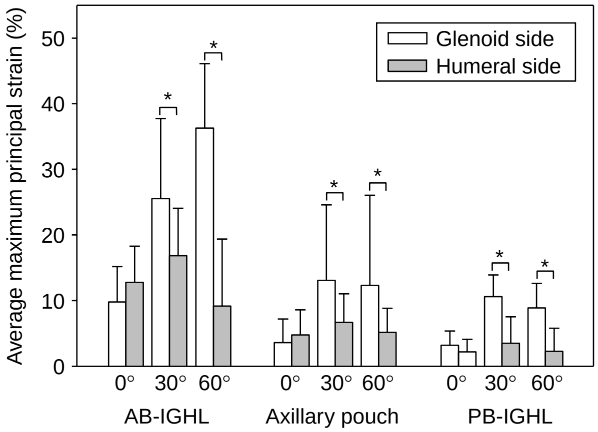

Figure 2. Maximum principal strains at 0°, 30° and 60° of external rotation on the glenoid and humeral sides of each IGHL region. Maximum principal strains were significantly higher on the glenoid side for each IGHL region at 30° and 60° of external rotation, but not at 0°. (*$p < 0.05$; mean ± SD).

strains in the axillary pouch on the glenoid side were more than two times higher than the humeral side at both 30° and 60° of external rotation. Maximum principal strains on the glenoid side of the PB-IGHL were over three times higher than on the humeral side at 30° and over four times higher at 60° of external rotation. There were no significant differences in the maximum principal strains between the humeral and glenoid sides of all three sub-regions of the IGHL when the simple translation test was conducted at 0° of external rotation. Finally, the strains in the AB-IGHL were significantly higher than the other regions on both the glenoid and humeral sides of the IGHL ($p < 0.05$ for all comparisons) with the glenoid side of the AB-IGHL consistently having the highest strains compared to all the other regions at every external rotation angle ($p < 0.05$ for all comparisons).

Discussion

In this study, the maximum principal strain throughout the IGHL was determined during a simple translation test at three external rotation angles using a validated subject-specific FE model of the glenohumeral joint. The strains on the glenoid side of each IGHL region were significantly higher than on the humeral side at 30° and 60° of external rotation, but not at 0°, which supported the hypothesis of this study. Further, the strain predictions from the subject-specific FE model compared well with experimental data from a sample of five shoulders subjected to the simple translation test that was reported previously (Moore et al. 2008b). Both studies indicate that the simple translation test with an anterior load performed at 30° and 60° of external rotation localises tissue strains to the glenoid side of the IGHL. This agreement further validates the results in the previous FE study (Moore et al. 2008a).

The similar strain magnitudes between 30° and 60° of external rotation for most sub-regions of the IGHL imply that clinicians are testing the glenoid side of the IGHL when this exam is performed anywhere from 30° to 60° of external rotation. The data also suggest that different regions of the IGHL, particularly the AB-IGHL, may be strained significantly more than other areas of the IGHL during the simple translation test, potentially allowing clinicians to further isolate the area being tested.

The glenoid side of the AB-IGHL also had the greatest amount of strain at all external rotation angles, supporting the general concept that the AB-IGHL is frequently injured at its glenoid insertion (Bankart 1923, 1938). In the context of the simple translation test with an anterior load, it appears that when this test is administered at 60° of external rotation, the area around the glenoid side of the AB-IGHL is isolated. It can also be inferred that the apprehension test, a clinical exam that determines the patient's apprehension to external rotation, is probably isolating the same area, with some bias to the glenoid side of the IGHL.

Both this study and the previous experimental study (Moore et al. 2008b) clearly show that the humeral side of the IGHL is not loaded when the simple translation test is administered at external rotation angles equal to or exceeding 30°. Further, when the results of these studies are taken together, it appears that the strains on the humeral side are fairly inconsistent between specimens when the simple translation test with an anterior load is performed at 0° of external rotation. For these reasons, the simple translation test with an anterior load may not be appropriate for testing the humeral side of the IGHL. Injuries such as humeral avulsion of the glenohumeral capsule (Warner and Beim 1997; Bokor et al. 1999; Schippinger et al. 2001; Bui-Mansfield et al. 2002; Chhabra et al. 2004; Richards and Burkhart 2004; Sailer and Imhof 2004) would probably be missed using this simple translation test.

This computational analysis had three primary weaknesses: sample size ($n = 1$); the exclusion of the labrum and the use of a hypoelastic constitutive equation in the FE model. Previous studies have shown that there are variations in the strain patterns between specimens (Moore et al. 2008a), but this study used only a single validated FE model based on a single specimen (Moore et al. 2008b). In the future, a population of FE models will be constructed and validated to predict the variation in the population on a subject-specific basis and assess the current trends predicted in the current study. Furthermore, the FE model used for this study does not include the labrum. The lack of a labrum will increase strains at the glenoid insertion (where the labrum would be), but will slightly decrease strains in the mid-substance tissue adjacent to the glenoid for every region of the IGHL (Drury et al. 2009). In this study, strains were averaged over large areas to compare the strain distribution between regions and joint position. Therefore, the IGHL sub-region (i.e. glenoid side of the AB-IGHL) that is tested by this clinical exam can be distinguished, but the location of the peak strain values might not be correct (i.e. mid tissue vs. labrum vs. labrum–capsule interface). Finally, a hypoelastic material model was used to represent the glenohumeral capsule material. Hypoelasticity is objective for finite deformations (i.e. large strains and rotations; Simo and Hughes 1998) and the limitations of using this material model for the IGHL have previously been discussed (Ellis et al. 2007). In the context of the current study, use of a hypoelastic material might change the magnitudes of the strains, but will have very little effect on their distribution. Even though a few weaknesses exist, our results are strongly supported by previous experimental data that examined five cadaveric shoulders (Moore et al. 2008a).

This study established a methodology to assess the capsule regions tested (strained) by clinical exams using a computational model. Once the region tested is identified, the next step, as was done for the ACL, is to look for side-to-side differences in translation between injured and uninjured shoulders in these joint positions using a repeatable methodology. Many of the discrepancies and complications with clinical exams for shoulder injuries (Levy et al. 1999; Tzannes et al. 2004) may be due to a lack of a systematic approach while developing the exams. In the future, diagnostic methods that are based on quantitative results from standardised clinical exams that do not cause discomfort to the patient or require their subjective responses would be advantageous. Our research group is currently in the process of constructing subject-specific FE models of the glenohumeral capsule that includes the labrum (Drury et al. 2009) and the proper constitutive model representing the capsule with a fibre reinforced, hyperelastic material model. We hope to determine effective joint positions for clinical exams using a population of subject-specific FE models of the glenohumeral joint.

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