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THE EFFECT OF LIGAMENT STIFFNESS ON SHOULDER CARTILAGE PRESSURE AND KINEMATICS

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INTRODUCTION

Rotator cuff tears are considered one of the primary causes of shoulder pain and dysfunction in adults. More than 50% of patients over the age of 60 and 80% of patients over the age of 80 experience a cuff tear [1]. Unfortunately, the success rate of rotator cuff repair is variable with many resulting in a re-tear. Revision surgeries can be as high as 30% for isolated supraspinatus tendon tears [2].

Rotator cuff tears not only decrease the mechanical properties of the cuff muscles, but also have secondary effects on the other tissues within the joint – specifically, ligaments and cartilage. In the presence of rotator cuff tears, the mechanical properties of the coracoacromial ligament have been shown to decrease, the area of glenohumeral (GH) cartilage degeneration increases, and glenoid cartilage thickness and modulus of elasticity decreases [3-6]. Thus, the overall shoulder joint stability and function would likely be affected by the change in these properties. However, the degree to which these changes affect function is not well understood.

There is a wide variation in the measured mechanical properties of shoulder joint ligaments [7-8]. It is essential to know how the variation of ligament stiffness affects the kinematics and cartilage pressure in the GH joint in order to understand why there is a high incidence of re-tears. To investigate this, we developed a three-dimensional finite element (FE) model of the shoulder joint.

METHODS

Computed tomography (CT) data files of a male shoulder (voxel size: 0.34 x 0.34 x 0.49 mm) were used to reconstruct the geometries of the humerus and scapula (Figure 1A) using image processing software (Amira v5.6, Visualization Sciences Group, Germany). Bones surfaces were created and exported into Geomagic Studio (v2014, Geomagic Inc., USA) to smooth, refine, and reduce noise in the solid computer-aided design models (Figure 1B).

Construction of the FE model and all simulations were performed using Abaqus (v6.13, Daussalt Systems, France). Shell elements were used for the humerus and scapula (elastic modulus, E = 17 GPa, poisson's ratio, v = 0.30) [9]. A layer of elements were offset from the proximal and distal ends of the humerus and scapula to model the articular cartilage. Humeral cartilage was defined as 0.5 mm thick and scapular cartilage as 1 mm thick and modeled using a neo-Hookean hyperelastic material model (shear modulus, G = 6.8 MPa, v = 0.45) [10]. Frictionless contact was defined between the humerus and scapula. Within the glenohumeral joint are four primary ligaments that provide stability: coracohumeral (CHL), superior glenohumeral (SGHL), middle glenohumeral (MGHL), and inferior glenohumeral (IGHL). These ligaments were modeled as axial springs (Figure 1C). Stiffness values from the literature for the CHL, SGHL, and IGHL were used to model the ligaments [7-8]. No stiffness values were available for the MGHL; therefore, the average of the SGHL and IGHL were used. A local coordinate system was defined corresponding to the medial-lateral (ML), superior-inferior (SI), and anterior-posterior (AP) directions.

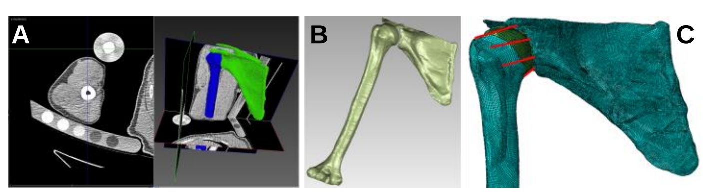

Figure 1: A) Geometry created in Amira, B) exported from Geomagic, C) FE model of shoulder joint with ligaments (red)

To simulate abduction, the humerus was rotated about the local AP axis. Rotations about the ML and SI axes were zero. Translational components of humerus were left unconstrained. During first 30° of abduction, the scapula was fixed and then followed a 2:1 relation for upward rotation with respect to the humerus [11].

Simulations were run for three different stiffness values (Table 1). Low stiffness was defined as the average stiffness minus one standard deviation, and high stiffness as the average stiffness plus one standard deviation. The kinematics of the humerus, peak and mean cartilage contact pressures were calculated. The relationship between contact pressures and abduction was analyzed using linear regressions. F-tests were used to test for significant differences in contact pressure as a result of changing the stiffness values (α = 0.05).

Table 1: Ligament stiffness, K (N/mm) [7-8]

	CHL	SGHL	MGHL	IGHL
Low	30.8	15.9	15.6	15.4
Avg	36.7	17.4	21.4	25.4
High	42.6	18.9	27.1	35.4

RESULTS

With increasing abduction angle, both peak and mean scapula cartilage pressure increased regardless of stiffness value (R^2 range = 0.71 - 0.86) (Figure 2A, B). Similar trends were found for the humeral cartilage. Decreasing stiffness resulted in lower peak scapula contact pressure (p=0.047), but had no effect on the mean contact pressure. However, increasing stiffness did not change either peak or mean scapula cartilage pressure. Humerus cartilage pressure was also not changed by increasing ligament stiffness, but decreasing stiffness decreased both peak and mean humerus cartilage pressures (p=0.014 for peak, p<0.001 for mean).

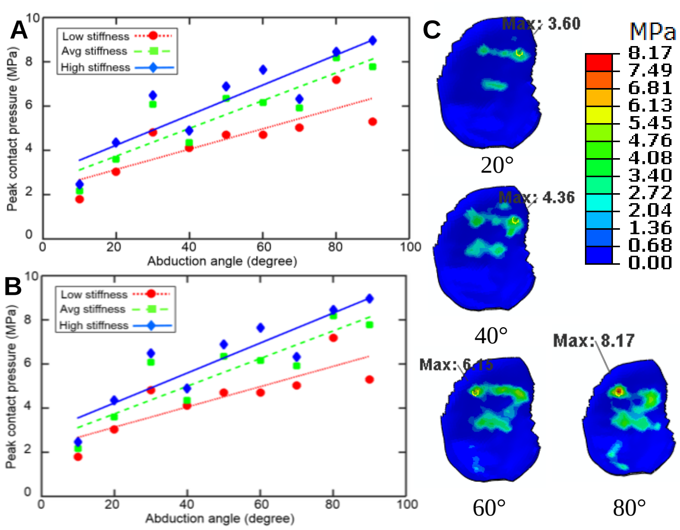

Figure 2: Scapula cartilage pressure during abduction: (A) peak, B) mean and C) pressure map for average stiffness at 20°, 40°, 60°, and 80° abduction

Decreasing ligament stiffness caused the humerus to move more posteriorly (p=0.02) whereas increasing stiffness resulted in no change (Figure 3). Superior-inferior translations were sensitive to both increasing (p=0.012) and decreasing stiffness (p=0.0044). However, changing ligaments stiffness did not change medial-lateral translations.

DISCUSSION

The purpose of this study was to investigate the influence of ligament stiffness on humerus kinematics and cartilage pressure. Our results indicate that cartilage pressure was only sensitive to decreases in ligament stiffness. Surprisingly, increasing stiffness contributed to changes in superior-inferior translations but did not result in changes in cartilage pressure.

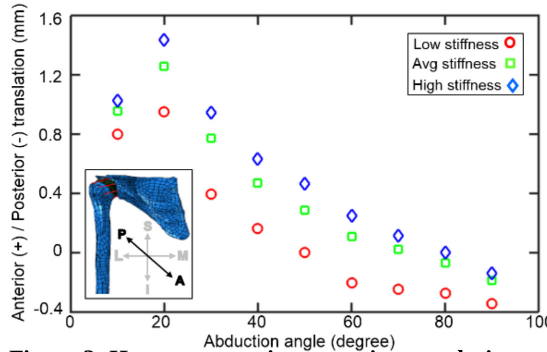

Figure 3: Humerus anterior-posterior translations during abduction

Since, there was no variation in medial-lateral translation, anterior-posterior translation may be the driving kinematic degree of freedom of cartilage contact pressure during abduction.

Our results show that humerus translation and cartilage pressure are sensitive to ligament stiffness. Others have shown that the mechanical properties of shoulder ligaments decrease due to rotator cuff tears [3-4]. If the injured rotator cuff tendons alone are treated during a repair, without regard for restoring ligament stiffness, the overall shoulder joint function may not be the same as the normal healthy shoulder. Moreover, the assessment of kinematics and muscle function as a metric of joint function (e.g. ability to lift the arm) may mask changes in cartilage pressure resulting in secondary consequences such as osteoarthritis.

Though our FE model has not been experimentally validated, our results are consistent with previous studies in the prediction of glenohumeral contact pressure. For example, others have reported mean cartilage pressure of 1.09±0.35MPa at 60° abduction [12], and our model results ranged from 1.02MPa - 1.44MPa. We report peak scapula cartilage pressure at 90° abduction ranging from 5.29MPa - 8.96MPa depending on ligament stiffness, and others have measured peak contact pressure of 4.2MPa for single arm weight and 9.2MPa for double arm weight at the same angle [13]. The anterior-posterior translation was also similar to previous findings [14-15].

In conclusion, our results indicate that ligament stiffness results in changes in contact pressure. Small kinematic changes resulted in significant changes in contact pressure suggesting that care should be taken when interpreting patient function. Altered ligaments stiffness may affect cartilage contact pressure even after rotator cuff repair.

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