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Modeling a rotator cuff tear: Individualized shoulder muscle forces influence glenohumeral joint contact force predictions

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ABSTRACT

Background: Rotator cuff tears in older individuals may result in decreased muscle forces and changes to force distribution across the glenohumeral joint. Reduced muscle forces may impact functional task performance, altering glenohumeral joint contact forces, potentially contributing to instability or joint damage risk. Our objective was to evaluate the influence of rotator cuff muscle force distribution on glenohumeral joint contact force during functional pull and axilla wash tasks using individualized computational models.
Methods: Fourteen older individuals (age 63.4 yrs. (SD 1.8)) were studied; 7 with rotator cuff tear, 7 matched controls. Muscle volume measurements were used to scale a nominal upper limb model's muscle forces to develop individualized models and perform dynamic simulations of movement tracking participant-derived kinematics. Peak resultant glenohumeral joint contact force, and direction and magnitude of force components were compared between groups using ANCOVA.
Findings: Results show individualized muscle force distributions for rotator cuff tear participants had reduced peak resultant joint contact force for pull and axilla wash ($P \le 0.0456$), with smaller compressive components of peak resultant force for pull ($P = 0.0248$). Peak forces for pull were within the glenoid. For axilla wash, peak joint contact was directed near/outside the glenoid rim for three participants; predictions required individualized muscle forces since nominal muscle forces did not affect joint force location.
Interpretation: Older adults with rotator cuff tear had smaller peak resultant and compressive forces, possibly indicating increased instability or secondary joint damage risk. Outcomes suggest predicted joint contact force following rotator cuff tear is sensitive to including individualized muscle forces.

1. Introduction

Rotator cuff tears (RCT) are a prevalent musculoskeletal injury in older individuals (Yamamoto et al., 2010). Rotator cuff muscles provide shoulder stability by situating the humeral head in the glenoid fossa (Ackland and Pandy, 2009) through combined action of concave-compression and anterior-posterior (transverse) and superior-inferior force couples (Lippitt et al., 1993; Lippitt and Matsen, 1993). Glenohumeral

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joint contact force (JCF) is a quantitative measure of joint stability, with reduced compression and forces directed outside the glenoid rim indicative of instability (Marchi et al., 2014; Steenbrink et al., 2009; Van Drongelen et al., 2013). The JCF will be dynamically driven by both limb kinematics during a task and the forces generated by muscles. Kinematics of task performance are known to be altered following RCT in older adults (Vidt et al., 2016a). While the proportion of individual muscle volumes at the shoulder is preserved across healthy adult age groups (Holzbaur et al., 2007a; Saul et al., 2015a; Vidt et al., 2012), RCT can alter the volume of injured muscles (Vidt et al., 2016b), dramatically changing the muscle force distribution across the joint. However, the extent to which muscle volume, and thus force-generating capacity, is altered may vary across individuals. If RCT causes a force imbalance at the glenohumeral joint, it may result in loss of dynamic joint control (Magarey and Jones, 2003), induce dyskinesis, or precipitate abnormal joint loading scenarios that lead to deleterious wear of articular cartilage on the glenoid and humeral head (Hsu et al., 2003). Understanding the specific effects of altered muscle force and altered kinematics after RCT on the associated glenohumeral JCF can provide insight into secondary consequences of RCT injury.

Since JCFs cannot be measured in vivo without invasive procedures (e.g. instrumented joint replacement), computational modeling is a useful approach to explore biomechanical consequences of injured muscle-tendon units and altered movement, and elucidate factors contributing to risk for subsequent glenohumeral joint damage. Several detailed upper limb models are currently available (Dickerson et al., 2007; Holzbaur et al., 2005; Nikooyan et al., 2011; Saul et al., 2015b; van der Helm, 1994), but these models represent a single specimen or average force-generating capabilities of young adults. Inclusion of age- and injury-related muscle changes is essential for more accurate predictions of the force profile of older individuals. In previous studies, models incorporating subject-derived strength characteristics have shown improved predictive qualities (Mogk et al., 2011; Nikooyan et al., 2010) for individual patients or specific populations. Specifically, the individual pattern of injury across individuals and the combined influence of aging may alter rotator cuff muscle force distribution, and thus predicted JCF, at the glenohumeral joint.

Our objective was to evaluate the effect of individualized muscle force and kinematics following rotator cuff injury on glenohumeral JCF predictions. To do this, we developed individualized computational models by scaling model muscle force-generating characteristics to correspond with subject-derived measurements of rotator cuff muscle volume. Dynamic simulations of movement were performed with individualized computational models and subject-derived kinematics of two upper limb tasks to examine the influence of muscle force distribution across the glenohumeral joint on predicted glenohumeral JCF. We hypothesized that altered muscle forces for RCT patients would result in a JCF profile that included reduced compressive forces and JCF directed closer to the glenoid fossa boundary.

2. Methods

The Wake Forest Health Sciences Institutional Review Board approved this study; all participants provided written informed consent. Fourteen older individuals (age 63.4 yrs. (SD 1.8)) participated (Table 1), including 7 participants (4 M/3F) with a supraspinatus tendon tear (RCT group) and 7 age-, (within 2 years) sex-matched controls. Rotator cuff tear participants who presented to our institution's orthopaedic clinic with symptoms of shoulder pain and were diagnosed with at least a high grade partial-thickness (> 50% tendon thickness) degenerative, MRI-confirmed supraspinatus tendon tear were recruited; 5 participants had full-thickness tear, 2 had partial-thickness supraspinatus tear (Table 1). Asymptomatic control participants were recruited from the community, did not have history of shoulder pain or injury, and were further screened using a modified Jobe's test for asymptomatic RCT (sensitivity: 81%; specificity: 89%) (Gillooly et al., 2010). The Jobe's test was performed with the participant's arms elevated 90° in the scapular plane with neutral arm rotation while a small, downward force (~2 kg) was manually applied. Exclusionary criteria included a test eliciting pain or weakness. In accordance with prior studies of participants with RCT (Vidt et al., 2016a; Vidt et al., 2016b), the injured arm was studied for RCT participants, and the dominant arm was investigated for asymptomatic controls (Table 1).

2.1. Functional task kinematics

Participants completed 2 functional tasks based on everyday activities while seated (chair height = 0.53 m). Loaded functional pull and axilla wash (Fig. 1A) were chosen for assessment because they represented both planar and multiplane tasks, and statistically significant differences in self-selected kinematics were identified between RCT and control groups for these tasks in prior work (Vidt et al., 2016a). Using previously described methods (Li et al., 2016), seven Hawk motion capture cameras (Motion Analysis Corporation, Santa Rosa, CA, USA) were used to track positions of twelve 1 cm-diameter retroreflective markers placed on anatomical locations on the upper limb and torso (Li et al., 2016) as participants performed each task. Participants were instructed on task start and finish positions, but could freely choose their joint postures and speed of movement. Three trials of each task were recorded, with 60 s of rest between trials and 2 min rest between tasks. Participants were instructed to stop and notify study staff if they felt any pain or discomfort during performance of any task. The second trial of each task for each participant was used for analysis. Marker data were post-processed using Cortex software (Cortex, Motion Analysis Corporation, Santa Rosa, CA, USA) and smoothed with the program's internal 6 Hz Butterworth filter. Joint kinematics were calculated using a nominal upper limb model (Saul et al., 2015b) in OpenSim (v.3.1) (Delp et al., 2007) using methods described below.

2.2. Model development

To calculate joint kinematics for functional pull and axilla wash tasks, the nominal dynamic upper limb model (Saul et al., 2015b) was scaled to each subject's anthropometry using OpenSim's scaling tool and marker positions recorded from motion capture with a static trial. Joint kinematics for each functional task were calculated using inverse kinematics and the scaled model. Briefly, inverse kinematics calculates joint angles of each model degree of freedom using a least squared algorithm to minimize distance between marker locations recorded using motion capture and positions of virtual markers (cf. Fig. 1B, pink spheres) in the model (Delp et al., 2007). Joint angle trajectories were filtered off-line with a zero-phase digital filter with a custom Matlab program (The Mathworks, Natick, MA, USA).

To calculate glenohumeral JCF with subject-derived joint kinematics, individualized computational models used for dynamic simulations of movement were developed using the nominal dynamic upper limb model as a foundation. This model includes joint descriptions of Saul et al. (Saul et al., 2015b) and maintains kinematic descriptions originally described by Holzbaur et al. (Holzbaur et al., 2005) Individualized models in this work maintained the bony geometry and kinematic descriptions from the nominal model, including representing scapulo-humeral articulation as a ball-and-socket joint, scapulo-humeral rhythm according to regression equations reported by de Groot and Brand (De Groot and Brand, 2001), and axis descriptions of thoracohumeral motion (elevation plane, elevation, axial rotation) according to International Society of Biomechanics recommendations (Wu et al., 2005). Range of motion of shoulder generalized coordinates and associated muscle paths were augmented to permit the full range of observed thoracohumeral motion for recorded tasks and maintain proper interaction of muscle actuators with their associated wrapping surfaces. Elevation plane range of motion was expanded to allow −95° to 130° of rotation and humeral axial rotation range of motion was