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An integrated model of active glenohumeral stability

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ABSTRACT

We present the first model of the glenohumeral joint implementing active muscle driven humeral positioning and stabilization without a priori constraints on glenohumeral kinematics. Previously established methods were used to predetermine the path, activation timing and resultant force contribution of 27 individual muscle segments at any given joint position. Artificial boundary conditions were applied in a three-dimensional finite element model of the joint and progressively released until the humeral head was completely free to rotate and translate within the fixed glenoid according to the compressive component of the predetermined resultant force. The shear component was then added such that no boundary conditions other than muscular force were applied. The framework was exploited to simulate elevation as a composite of instantaneous positions and theoretically demonstrate that joint stability can be achieved exclusively through muscular activity. Predicted muscle moment arms, muscle activation timing, humeral head translations, joint contact forces and stability ratio were comparable with existing experimental and in vivo data. This framework could be valuable for subject specific modeling and may be used to address clinical hypotheses related to shoulder joint stability that cannot be pursued using simplified modeling approaches.

1. Introduction

The glenohumeral joint is the most frequently dislocated major joint (Cave et al., 1974; Hovelius, 1982), with up to 8% patients experiencing complications even 25 years after the first dislocation (Hovelius et al., 2008). Joint instability, defined as the inability to maintain the humeral head centered in the glenoid fossa (Matsen and Fu, 1993), also represents a serious complication after conventional (Franta et al., 2007; Wirth and Rockwood, 1996) and reverse shoulder replacement (Sirveaux et al., 2004; Wall et al., 2007). Adequate treatments to restore joint stability in the presence of glenoid defects (Beran et al., 2010), atypical labrum anatomy (Kanatli et al., 2010) or with unconstrained prostheses (Gonzalez et al., 2011) remain limited.

The role of the passive stabilizing structures in the shoulder has been studied extensively (Debski et al., 1999a, 199b; Motzkin et al., 1998; O'Connell et al., 1990; Terry et al., 1991), but their relative importance in stability remains uncertain (Veeger and van der Helm, 2007). Glenohumeral stability is primarily ensured by coordinated rotator cuff action to provide concavity compression (Lippitt and Matsen, 1993). However, the study of active muscular stabilization remains neglected (Apreleva et al., 1998; McMahon and Lee, 2002; Schiffern et al., 2002) because the number of involved muscles and their infinite potential interactions makes such studies challenging, the rotator cuff muscles simultaneously move and stabilize the humerus (McMahon and Lee, 2002) and muscles can stabilize or destabilize the joint depending on joint position (Labriola et al., 2005; McMahon and Lee, 2002; Werner et al., 2007).

Muscular stabilization has been studied in cadaver tests (Halder et al., 2001a; Lee et al., 2000; McMahon and Lee, 2002) restricted to a subset of muscles, or describing muscle force vector components (Ackland and Pandy, 2009; Konrad et al., 2007; Labriola et al., 2005) that cannot account for muscle timing and interaction for movement and stabilization (Veeger and van der Helm, 2007).

Six degrees of freedom (DOF) numerical models of the glenohumeral joint are clearly required (Favre et al., 2009b; Hill et al., 2008; Veeger and van der Helm, 2007) to allow realistic simulations of the timing and extent of muscle stabilization. First departures from the ball-and-socket assumption constrained the intersection of the resultant force to remain within the glenoid boundaries (Favre et al., 2005; van der Helm, 1994; Yanagawa et al., 2008). The absence of humeral translations in these models has precluded analysis of tissue deformations, contact area and

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contact pressure (Buchler et al., 2002). Attempts to allow limited humeral translations (restricted by a non-physiological spring) have also been made, but during axial rotation only (Buchler et al., 2002). The joint was centered actively in another model (Terrier et al., 2007, 2008), but was limited to 2D, to the deltoid and rotator cuff and to abduction. The large variability in daily glenohumeral movements and loads call for models that have not been tailored to meet special cases of loading and motions.

The purpose of this study was to develop the first three-dimensional finite element (FE) model simulating the six DOF in the glenohumeral joint, and implementing active muscle driven humeral positioning and stabilization without *a priori* defined kinematic constraints. We then tested whether glenohumeral joint stability could be achieved during elevation through muscular action only (without the ligamentous capsule), as this is a matter of considerable controversy (Matsen, 2002).

2. Methods

A 3D deformable model of the glenohumeral joint was combined with previous methods of muscle wrapping (Favre et al., 2010) and muscle force estimation (Favre et al., 2005) to simulate active joint balance and stabilization (Supplementary Fig. 1). Each step can be automatically performed for any instantaneous, static joint position. First, muscle paths are computed for a position of interest (Supplementary Fig. 1A). Second, this anatomical information is used to estimate the muscle forces required to balance a given external moment and stabilize the joint (Supplementary Fig. 1B). The resultant force is finally applied in a 3D FE model for simulation of joint contact and humeral head translations (Supplementary Fig. 1C).

For the FE model creation, a fresh frozen human scapula free of visible deformities was CT scanned (400 μm isotropic resolution, Philips Brilliance 40, Philips, Best, NL). The bony contour was outlined using a global threshold (ImageJ, Bethesda, MD) and 3D reconstruction was performed in Matlab (MathWorks, Natick, MD). The full joint being unavailable, the humerus geometry was imported from the Bel repository (Van Sint Jan et al., 2004). The humeral head diameter (48 mm) matched the glenoid height (33 mm) according to the anthropometric relationships (McPherson et al., 1997).

The bones were meshed with quadratic tetrahedral elements (Viceconti et al., 1998; Cifuentes and Kalbag, 1992; Ramos and Simoes, 2006) in ANSYS (Workbench 12.0, Ansys Inc., Canonsburg, PA) and imported into Marc Mentat 2008r1 (MSC.Software, Santa Ana, CA). In this study focusing on active stabilization, the capsule was not modeled. The cartilage and labrum geometries were modeled following the published anatomical data. The humeral head cartilage surface was defined as a sphere (Boileau and Walch, 1997; Soslowsky et al., 1992) with 1 mm thickness (Fox et al., 2008). The humerus was positioned with respect to the scapula by aligning their respective coordinate systems (Wu et al., 2005). The humerus was translated medio-laterally so as to create a 1.3 mm clearance in the glenoid cavity center, corresponding to the glenoid cartilage thickness in this location (Yeh et al., 1998). By congruency of both cartilage surfaces (Soslowsky et al., 1992), the lateral border of the glenoid cartilage was made to coincide with the humeral cartilage surface. The labrum thickness was designated according to the published data (Howell and Galinat, 1989). Isotropic linear-elastic bone material properties (18 GPa Young's modulus and 0.3 Poisson's ratio) (Currey et al., 2001) and hyperelastic neo-Hookean cartilage and labrum material properties were chosen with the strain-energy density function (Buchler et al., 2002) defined as

$$W = C_{10}(I_1 - 3)$$

with

$$C_{10} = E/4(1+\nu)$$

with E representing the elastic modulus, ν is the Poisson's ratio and I_1 is the first invariants of the Cauchy–Green tensor.

The values of $C_{10}=1.79$ and $C_{10}=12.5$ were attributed to the cartilage ($E=10$ MPa and $\nu=0.4$) (Buchler et al., 2002) and labrum, respectively ($E=70$ MPa, $\nu=0.4$) (Smith et al., 2009). Cartilage contacts were frictionless (Terrier et al., 2008).

The nodes on the scapula lying 2 cm medial from the most medial point on the glenoid surface were fixed. The humerus was initially fixed in translation, but rotations were always permitted. First, the humerus was displaced toward the glenoid until an initial touching contact was automatically detected by the FE software. Second, the compressive joint reaction force component previously calculated in the muscle force estimation was applied at the humeral head center (located according to Meskers et al., 1998) to keep the two bones touching while progressively removing the medio-lateral translational constraint on the humerus (using the "gradual release" Marc Mentat function). Third, the remaining humeral inferior–superior and antero-posterior translational constraints were gradually released, leaving the head fully free to translate, rotate and center within the glenoid by the compressive force. Finally, the shear joint reaction force component calculated in the muscle force estimation was applied and the model was equilibrated to assume a self-aligned, "physiological" configuration.

The labrum and cartilage implementation was controlled by comparing stability ratios (defined as the shear force required to dislocate the joint with a 50 N compressive load) in eight directions against experimental measurements with dissected capsule (Halder et al., 2001b; Lazarus et al., 1996; Lippitt and Matsen, 1993). The glenoid was oriented in a plane perpendicular to the compressive force to simulate the experiments. Dislocation was defined as the point where the humeral head center displacement increased abruptly (Supplementary Fig.2).

The deformable model was combined with two established methods to apply physiologically relevant glenohumeral joint resultant forces. The muscle paths for 27 muscle segments were computed using the current bone geometries (Supplementary Fig. 1A) within an automated wrapping paradigm (Favre et al., 2010). For muscles that do not attach on the scapula (latissimus dorsi and pectoralis major), the origin sites were measured using a previously published physical model (Favre et al., 2005, 2009a). The resulting muscle moment arms and lines of action were input to an algorithm for distributing muscle forces according to their relative mechanical advantage for counteracting external moment components (Supplementary Fig. 1B) (Favre et al., 2005, 2009a). If joint stability was less than a critical threshold, supplementary rotator cuff activity was incrementally applied until the joint was stabilized. Originally the algorithm considered the joint to be stable if the resultant force fell within the glenoid boundaries (Favre et al., 2005). Here the introduced deformability of the glenoid boundaries was accounted by considering the joint stable if the stability ratio was below the values computed in the FE model as described above.

Scapular plane glenohumeral elevation from 0° to 80° was simulated in increments of 10°, and outputs were compared against the experimental evidence. Muscle moment arms were compared with published experiments (Favre et al., 2005; Hughes et al., 1998; Kuechle et al., 1997; Liu et al., 1997; Nyffeler et al., 2004; Otis et al., 1994; Poppen and Walker, 1978). These anatomical data were used to estimate the muscle forces required to balance the arm weight (35 N) applied at the arm center of mass in each glenohumeral position (Poppen and Walker, 1976). The external force vector was rotated with a 2:1 proportion through the range of elevation to account for the scapulohumeral rhythm (Poppen and Walker, 1976). The on–off patterns of muscle activity were compared with the available electromyographic (EMG) data (Habermeyer et al., 1987; Inman et al., 1944; Yanagawa et al., 2008, Kronberg et al., 1990). The joint reaction forces were compared with previously published simulations (Oizumi et al., 2006; Poppen and Walker, 1978; Terrier et al., 2008; van der Helm, 1994; Yanagawa et al., 2008) and in vivo telemetric data (Nikooyan et al., 2010). To assess the

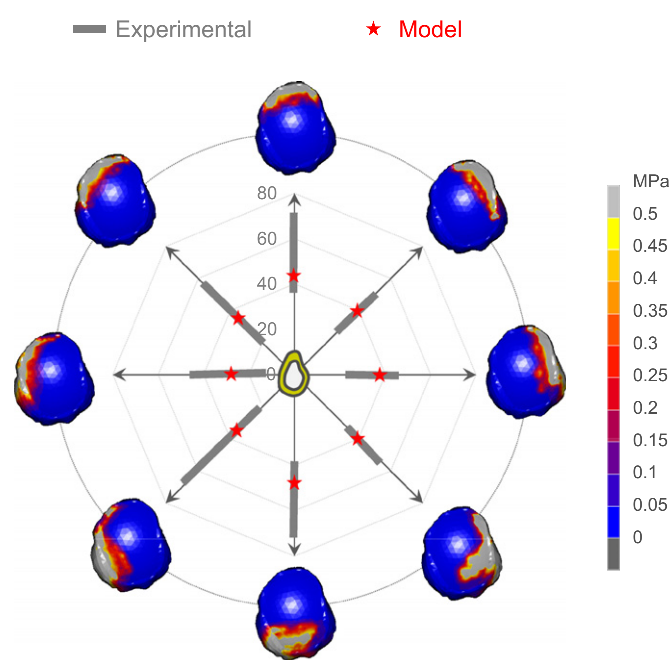

Fig. 1. Stability ratio comparison of simulated (red stars) and reported experimental (gray areas) values. The corresponding von Mises stress distributions at the time just prior to dislocation show that the humerus moved in the direction of the applied force. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)