Humeral head translation decreases with muscle loading

Angela E. Kedgley, MESc,[a,b] Geoffrey A. Mackenzie, BESc,[a,b] Louis M. Ferreira, BSc, BESc,[a,b] Darren S. Drosdowech, MD, FRCSC,[a,c] Graham J. W. King, MD, MSc, FRCSC,[a,c,e] Kenneth J. Faber, MD, MHPE, FRCSC,[a,c] and James A. Johnson, PEng, PhD,[a,b,c,d,e] *London, Ontario, Canada*

This study was conducted to determine the effect of in vitro passive and active loading on humeral head translation during glenohumeral abduction. A shoulder simulator produced unconstrained active abduction of the humerus in 8 specimens. Loading of the supraspinatus, subscapularis, infraspinatus/teres minor, and anterior, middle, and posterior deltoid muscles was simulated by use of 4 different sets of loading ratios. Significantly greater translations of the humeral head occurred both in 3 dimensions ($P < .001$) and in the sagittal plane ($P < .005$) during passive motion when compared with active motion from 30° to 70° of abduction. In the sagittal plane, passive abduction experienced a resultant translation of 3.8 ± 1.0 mm whereas the active loading ratios averaged 2.3 ± 1.0 mm. There were no significant differences in the translations that were produced by the 4 sets of muscle-loading ratios used to achieve active motions. This study emphasizes the importance of the musculature in maintaining normal ball-and-socket kinematics of the shoulder. (J Shoulder Elbow Surg 2008;17:132-138.)

Because the glenohumeral joint is largely unconstrained by the bony anatomy, the dynamic constraints (ie, the musculotendinous units) play an even greater role in maintaining joint stability than in other joints. These units may contribute to stability in a variety of different ways. Passive tension from the bulk effect of the muscle is one factor[25] that is disputed, with some authors stating that passive tension from the muscles plays a greater role than that of other soft tissues[3] and others regarding passive tension from the muscles as having little effect.[23] The relative compression of the articular surfaces as a result of muscle contraction may also create stability.[18,19,21,22] This may occur regardless of whether the muscle forces across the glenohumeral joint are balanced, as the balancing may be provided by the ligaments.[22] Tightening of passive ligamentous constraints provides significant restraint at the extremes of joint motion.[1,6,19,20,33] The barrier effect of the contracted muscle[22,31] and redirection of the joint reaction force to the center of the articular surface by a balancing of antagonist muscle forces[19] may also contribute to glenohumeral joint stability. Despite the aforementioned roles, the joint reaction force may not be directed into the articular surface, and hence, dislocation may occur.

The role of the deltoid muscle in stabilization is unclear. Some authors suggest that the deltoid does not provide any significant stabilization of the humeral head inferiorly,[23] whereas others suggest that it does provide restraint inferiorly.[5,24] Anteriorly, the deltoid increases in prominence as a stabilizer in abduction and external rotation, as the glenohumeral joint becomes increasingly unbalanced.[15]

Notwithstanding the fact that the glenohumeral joint is considered to be a ball-and-socket joint,[7,9,21,26,28,32] the motion of the humeral head on the glenoid surface may be more correctly modeled as a combination of rotations and translations.[6,7,14,26,35] Several authors have examined translation in the glenohumeral joint during active abduction, both in vivo and in vitro. Poppen and Walker[26] conducted an in vivo study that used radiographic data to determine the excursion of the humeral head during glenohumeral abduction. Karduna et al[12] performed a static in vitro analysis on cadaveric specimens, with the arm positioned both passively and actively, placing the humerus at 0°, 30°, and 60° of glenohumeral abduction and varying degrees of internal and external rotation.

Wuelker et al[35] used an in vitro loading device, similar to the one used in this study, to produce glenohumeral abduction. Loading ratios based on the cross-sectional areas of the muscles were used to distribute the simulated muscle forces. McMahon et al[21] also conducted a joint simulator study, actively positioning specimens at discrete abduction angles and using 4 sets of muscle force ratios. They found that humeral

From the [a]Bioengineering Research Laboratory, Hand and Upper Limb Centre, St Joseph's Health Care London, and Departments of [b]Biomedical Engineering, [c]Surgery, [d]Mechanical Engineering, and [e]Medical Biophysics, University of Western Ontario.

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Reprint requests: James A. Johnson, PEng, PhD, Hand and Upper Limb Centre, St Joseph's Health Care London, 268 Grosvenor St, London Ontario, Canada, N6A 4L6 (E-mail: *jajohnso@uwo.ca*).

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head translations were not significantly different with different sets of muscle-loading ratios and that the SD of the humeral head translation increased with increased abduction angle.

The physiologically correct method of simulating the forces applied by the muscles surrounding any joint has yet to be determined. In previous motion simulations of the glenohumeral joint, several approaches have been taken. These include applying loads equally to every muscle,[2] proportional to the physiologic cross-sectional areas (pCSAs) of the muscles,[5,10,29,36] proportional to some combination of electromyographic (EMG) data and cross-sectional areas of the muscles,[8] or proportional to some combination of EMG, cross-sectional area, lever arms of muscles, and clinical knowledge.[20,30]

Given the highly unconstrained nature of the glenohumeral joint, it is possible that varying the ratios between the simulated muscle loads may affect the translation of the humeral head during motion. Therefore, humeral head translation during glenohumeral abduction was measured for passive motion, as well as active motion generated via 4 different sets of loading ratios.

MATERIALS AND METHODS

Testing procedure

Eight fresh-frozen, cadaveric upper forequarters (mean age, 58 ± 11 years [range, 47-71 years]; 4 male and 4 female; 3 left and 5 right) were prepared for mounting in a glenohumeral joint simulator, as shown in Figure 1.[13] The specimens were inspected for any visible pathologies. Two specimens had a 1-cm tear in the supraspinatus. A third specimen had a defect in the joint capsule. No advanced osteoarthritis was found. In preparation for testing, the rotator cuff muscles were dissected from their respective fossae, and the scapula was cleaned of soft tissues. The joint capsule was not disrupted. The soft tissues covering the deltoid muscle were left as intact as possible. The distal humerus and forearm were removed, and a rod was cemented into the humeral canal and weighted to compensate for the removed arm mass. Each specimen was mounted to the abduction simulator by potting the scapula in a Delrin case with dental cement (Instant Tray Mix; Lang Dental Manufacturing, Wheeling, IL). This was attached to a base to which 6 sets of pulleys were also affixed. Six muscle groups were simulated: the supraspinatus, subscapularis, infraspinatus/teres minor, and anterior, middle, and posterior deltoids. Each muscle group was connected to a computer-controlled pneumatic actuator to produce motion via a stainless steel cable that passed through a set of pulleys that dictated the line of action of the muscle. The 3 cables that represented the deltoid were attached to the deltoid tuberosity by a cortical screw. The 3 rotator cuff tendons were sutured to plastic connectors, to which the cables were attached.

Four sets of loading ratios were applied to achieve active scapular plane abduction. These were based on (1) equal loads to all cables (constant-constant), (2) mean pCSAs of

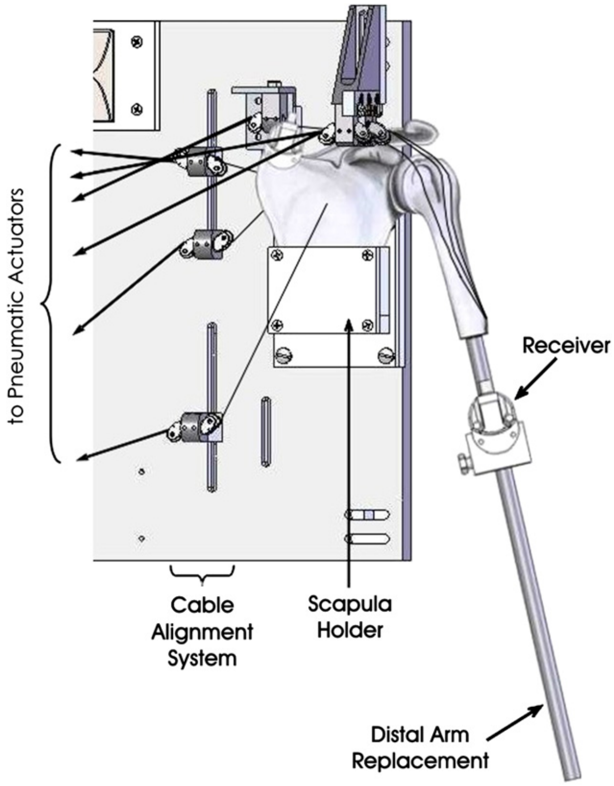

Figure 1 Schematic of testing apparatus. The scapula was potted in a holder, and the distal portion of the humerus was removed and replaced with a weight. Forces were applied to simulate the anterior, middle, and posterior deltoid muscles and the rotator cuff muscles—the supraspinatus, subscapularis, and infraspinatus and teres minor in combination—by computer-controlled pneumatic actuators. The 3 cables that simulated the deltoid were attached to the deltoid tuberosity. The 3 rotator cuff muscle cables were sutured to the corresponding tendons. These cables then passed through an alignment system that provided the appropriate lines of action for the muscles. The shoulder simulator was used in conjunction with an electromagnetic tracking system. One receiver was attached to the scapula and one to the humerus.

the muscles (pCSA), (3) constant values of the product of EMG data and pCSAs (constant EMG), and (4) variable ratios of the EMG and pCSA data, which changed as a function of abduction angle (variable EMG).[13] Before the initiation of active motion, small tone loads were applied to each of the rotator cuff muscles to center the humeral head in the glenoid fossa. The tone loads were chosen individually for each specimen to be the lowest load that could keep the glenohumeral joint reduced without abducting the arm. The same load was applied to each rotator cuff muscle, and this varied between 20 and 40 N. Abduction was performed in a smooth and continuous quasistatic manner, with the full range of motion taking place over 15 to 20 seconds. The motion of the humerus was not constrained, in either the plane of abduction or the angle of rotation. Despite this, the plane of elevation remained within 10° of the scapular plane throughout abduction. During passive motion, the investigator ensured that the joint maintained a reduced position. Approximately 90° of abduction was achieved during both passive and active scapular-plane abduction, as measured between the scapula and the humerus.

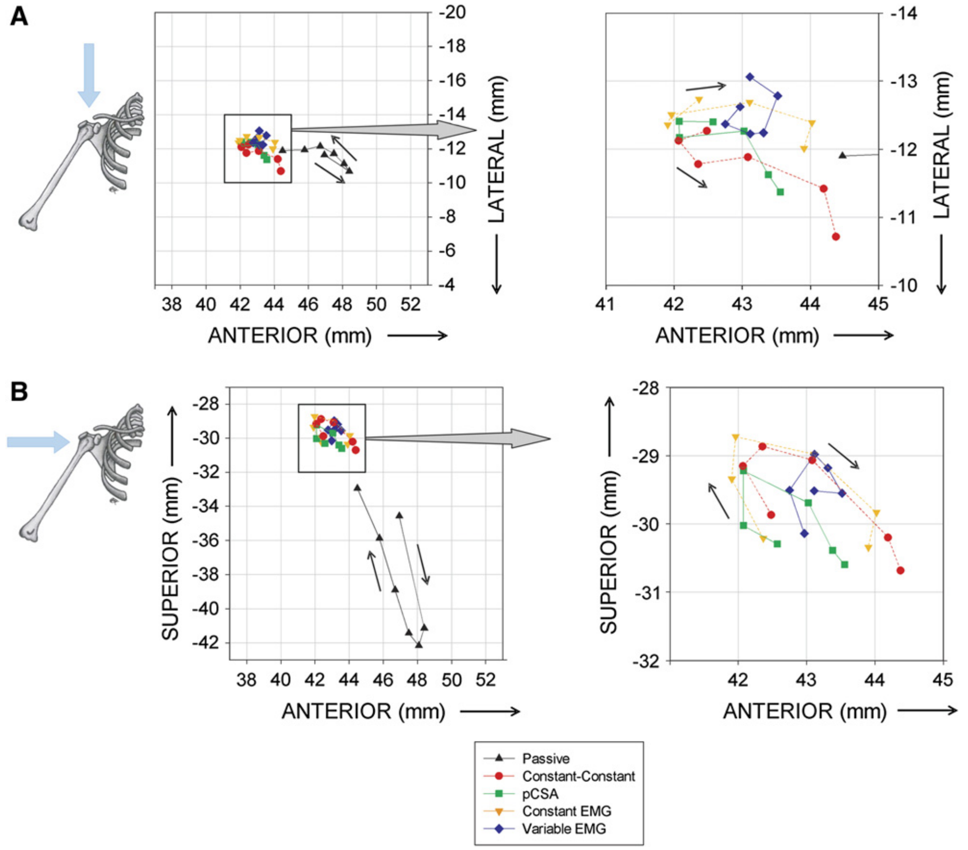

Figure 2 Humeral head path during abduction: transverse plane (**A**) and sagittal plane (**B**). Motion pathways of the center of the humeral head are shown for 1 specimen in the scapular coordinate system for 1 trial of passive motion (*black triangles*) and 4 methods of active motion: constant-constant (*red circles*), pCSA (*green squares*), constant EMG (*orange inverted triangles*), and variable EMG (*blue diamonds*). Points are plotted every 10° beginning at 20°. The *arrows* on the plots indicate the direction of increasing angle of abduction. The plots on the *left* show all motions, whereas the enlarged plots on the *right* show the active motions only.

An electromagnetic tracking system (Flock of Birds; Ascension Technologies, Burlington, VT) was used to quantify the joint motion and to provide abduction angle feedback for the variable EMG loading ratio. To determine the position of the humeral head relative to the scapula as a function of abduction, various bony landmarks were digitized on the humerus and scapula. These points were used to create coordinate systems on the scapula and humerus.[34] The origin of the coordinate system on the humerus was the center of the humeral head, created by applying a sphere-fit algorithm to the points obtained from a trace digitization of the articular surface, after motion testing. The mean error in the determination of the humeral head center, based on digitization of a 30-mm-diameter patch, is 0.46 mm. The repeatability is approximately 1.0 mm. The origin of the scapular coordinate system was the acromial angle, the most dorsolateral point on the scapula.

Data analysis

Humeral head translation in each direction, superior-inferior, anterior-posterior, and medial-lateral, was calculated as the change in position of the humeral head center in the scapular coordinate system from a given abduction angle compared with its reference position at 30° abduction. In this way, all 8 specimens could be compared with the same reference position. For each specimen, motions were conducted 5 times, and the results from these 5 trials were averaged.

Statistical analysis was performed by use of 1- and 2-way repeated-measures analyses of variance. This was followed by multiple comparisons via the Student-Newman-Keuls technique with significance defined at $P < .05$.

RESULTS

The effect of the various loading methods on the translation of the humeral head is demonstrated in Figure 2. This shows the path of the humeral head center in the transverse and sagittal planes of the scapula, as it moved during passive and active arm abduction for 1 specimen. The plane essentially parallel to the glenoid surface is the sagittal plane of the scapula. More translation occurred during passive motion

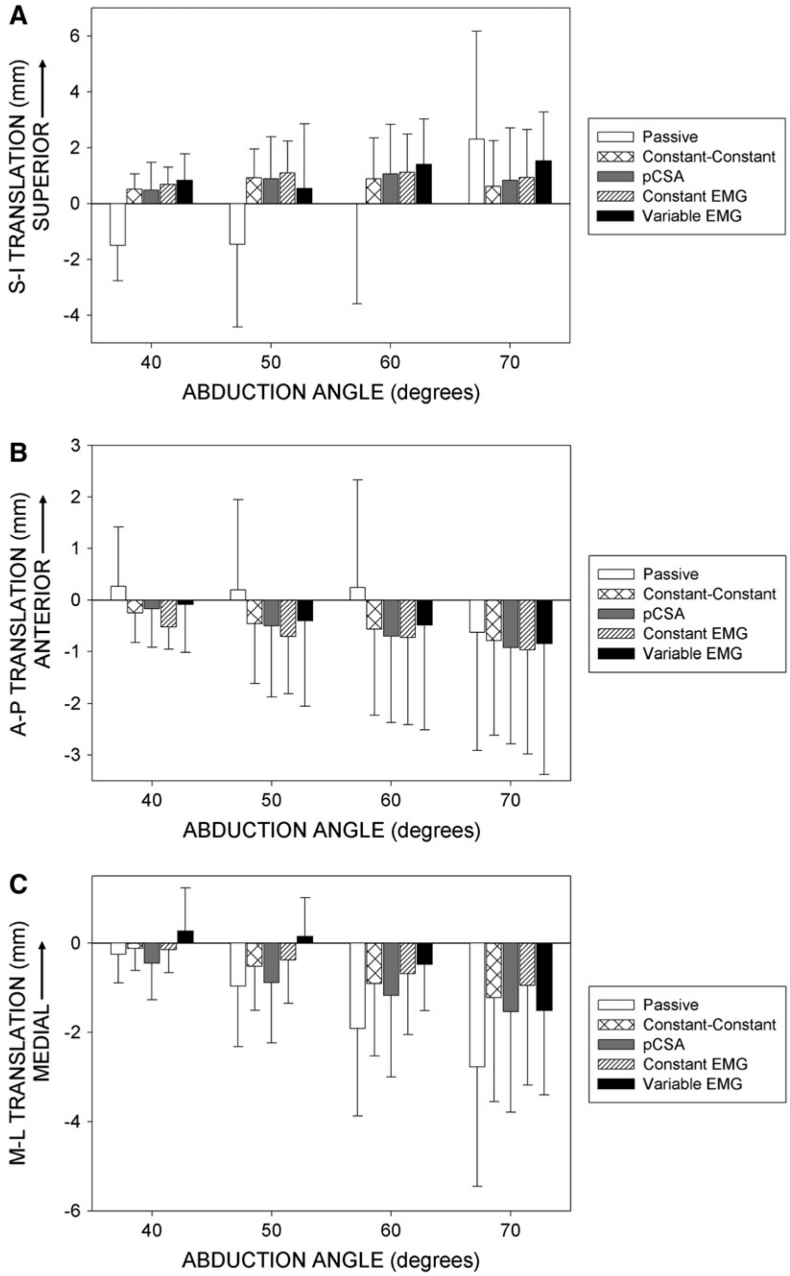

Figure 3 Humeral head translation along each axis: superior-inferior (*S-I*) (**A**), anterior-posterior (*A-P*) (**B**), and medial-lateral (*M-L*) (**C**). Resultant translations (mean ± 1 SD) for all specimens in comparison to the position of the humeral head at 30° of abduction are presented for 40°, 50°, 60°, and 70° of abduction. *Open bars*, Passive motion; *bars with large crosses*, constant-constant; *gray bars*, pCSA; *bars with small stripes*, constant EMG; *black bars*, variable EMG.

than active ($P < .005$), with no significant differences between the active loading methods ($P > .6$). Between 30° and 60° of abduction, passive abduction produced a resultant translation in the sagittal plane of 3.8 ± 1.0 mm whereas the active loading ratios averaged 2.3 ± 1.0 mm. In the coronal plane of the scapula, passive abduction yielded a resultant translation of 4.1 ± 1.4 mm whereas active abduction averaged 2.0 ± 1.5 mm.

The humeral head translation in each direction, superior-inferior, anterior-posterior, and medial-lateral, is shown in Figure 3. The majority of the translation tended to occur in the superior-inferior direction for all methods of abduction. Although the investigator attempted to

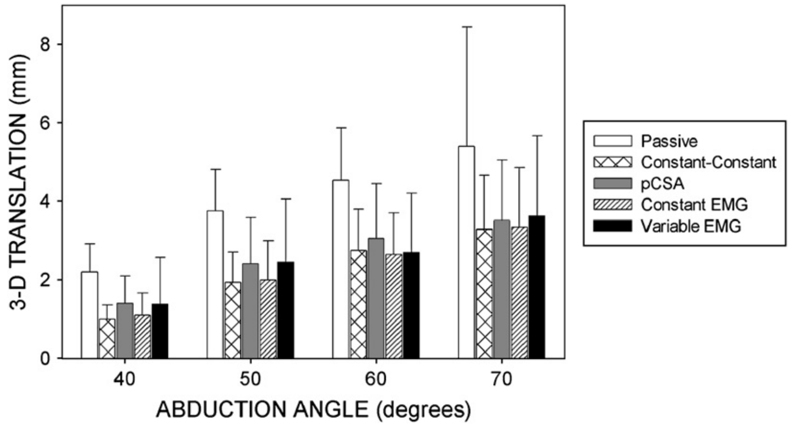

Figure 4 Three-dimensional (*3D*) humeral head translation during abduction. Resultant translations (mean ± 1 SD) for all specimens in comparison to the position of the humeral head at 30° of abduction are presented for 40°, 50°, 60°, and 70° of abduction. *Open bars*, Passive motion; *bars with large crosses*, constant-constant; *gray bars*, pCSA; *bars with small stripes*, constant EMG; *black bars*, variable EMG.

maintain a centralized position of the humeral head during passive motion testing, the head center was located significantly inferior ($P < .001$) to its position during active motion. There was no difference in head position among the 4 different active motion protocols ($P > .6$). The superior-inferior position of the humeral head did not vary ($P > .4$) as a function of abduction angle between 30° and 80° for active loading techniques; however, it did vary ($P < .001$) for passive motion.

Similarly, there was a difference in the anterior-posterior position between passive and active motion ($P < .001$) but no difference in head position among the 4 different active motion protocols ($P > .9$). The humeral head was situated more anteriorly in the passive cases ($P < .001$). In the medial-lateral direction, a difference was measured between the passive and active loading ratios ($P < .01$) at angles of abduction of less than 50°, with the humeral head being more lateral in the passive cases. No difference was seen at higher angles.

Figure 4 shows the 3-dimensional resultant translation of the humeral head position, referenced from 30° of abduction. Greater translation occurred in passive motion compared with active motion between 30° and 60° of abduction ($P < .001$). No difference in humeral head translation was found among the 4 methods used to simulate active motions ($P > .4$).

The intraspecimen repeatability was found to be higher for active motions than for passive motions ($P < .04$). Repeatability for passive motions averaged 1.7 mm in 3 dimensions, whereas it averaged 0.4 mm for constant-constant, 0.8 mm for pCSA, 0.4 mm for constant EMG, and 1.0 mm for variable EMG.

DISCUSSION

This study showed that translation of the humeral head decreased with active simulation of abduction and that the humeral head was positioned more superiorly and posteriorly with simulation of active motion. However, the ratios between the forces applied to the muscles to simulate active motion did not affect the position. This is possibly a result of the direct influence of the compressive forces generated by simulated muscle loading, which effectively compresses the joint and improves stability. Thus, active loading has an important stabilizing influence on the humeral head; this is consistent with findings at the elbow.[4,11] Unfortunately, passive motions were performed by a single investigator, and as a result, inter-investigator reliability trials for passive motion are not available.

The mean humeral head translation in the coronal plane during active motion was 2.0 mm over 30° of abduction. This correlates well with the findings of Poppen and Walker,[26] who showed, using plane radiographs, that less than 1.5 mm of translation occurred on average in vivo in the scapular plane in normal subjects between each 30° arc of motion. Wuelker et al[35] reported greater superior and anterior translations than observed in our study. They found a mean of 5.7 mm of translation superiorly and 3.3 mm anteriorly between 30° and 90° of abduction. This exceeds the translations observed in this study, which showed less than 5 mm in 3 dimensions for this range of abduction. This may be partly a result of differences in the measurement of the humeral head center. In our study, a trace digitization of the humeral head surface was sphere-fit to determine the center. In contrast, Wuelker et al used an impression of the humeral head in plaster to provide a surface in which the humeral head was moved while tracking the position of humerus. This may have influenced the coordinate systems and, consequently, the values of the translation.

Our results differ from those of Karduna et al,[12] who reported no difference between humeral head translation under passive and simulated active abduction. They used a static joint simulator that utilized spring scales to apply loads to the simulated muscles. Whereas static joint simulators are beneficial for examining discrete joint positions, they cannot examine continuous kinematics. However, they did take into account the scapulothoracic motion, which may have had some effect on the humeral head translations.

Konrad et al[16] simulated active motion in a cadaveric shoulder testing device similar to that used in our study but with the addition of the pectoralis major and latissimus dorsi muscles. The rotator cuff muscles and the middle deltoid muscle were simulated using equal forces to each muscle group, and the pectoralis major and latissimus dorsi were allocated 10%, 20%, or 30% of the deltoid force. Although the addition of other muscle groups helps to improve the replication of the in vivo state, the application of non-varying ratios between muscle groups does not represent the most physiologically based method currently available by which to simulate muscle forces.

No difference was found in the translations in 1, 2, or 3 dimensions among the active motions created by the different loading ratios. This is agreement with the findings of McMahon et al[21] and seems to imply that any of the active ratios could be used in the laboratory to simulate glenohumeral abduction. It is recognized that although the loading ratios all result in decreased humeral head translation, they may result in differing glenohumeral joint reaction forces.

In the sagittal plane, the mean translation for passive motion was 3.8 mm, whereas the mean translation for active motion was 2.3 mm over the course of 30° of abduction. As other investigators have also found, this emphasizes the importance of the rotator cuff muscles in creating and maintaining the ball-and-socket kinematics of the shoulder.[17,27] Some believe this is the most important factor in maintaining the stability of the glenohumeral joint.[17,33] The fact that translations increased as the humerus approached 80° of abduction may have been a result of impingement of the greater tuberosity on the acromion. This may have prevented the arm from elevating farther and, thus, forced the humeral head to move inferiorly to allow abduction.

The weaknesses of this study include the fact that scapulothoracic motion was not simulated and the muscles were simulated by single cables. Given that most muscles have large attachment sites and are fan-shaped, the net-force line of action of some muscles may vary with abduction angle. Therefore, a single cable may serve only as an approximation to physiologic loading. In addition, the translations during active internal and external rotation, as well as those during forward flexion, also need to be examined. The strengths of this study include the use of a variable loading technique for the in vitro simulation of muscle loading. Although this technique did not produce significantly different results from those obtained by use of the constant loading ratios, it may lead to more physiologically based loading methods in the future.

The ability to replicate physiologic kinematics in a cadaveric model accurately is valuable in the pursuit of a greater understanding of the causes and remedies of many shoulder conditions. The study of the glenohumeral joint under controlled conditions may lead to improved in vitro modeling. This work emphasizes the importance of the musculature in creating and maintaining the ball-and-socket kinematics of the shoulder and helps validate the use of the glenohumeral joint simulator for in vitro testing. This study also suggests that the translational kinematics of the humeral head in 3 dimensions may be invariant with respect to the selection of muscle-loading ratios.

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