Internal and external rotation of the shoulder: Effects of plane, end-range determination, and scapular motion

Sean P. McCully, MS,[a] Naveen Kumar, BS,[b] Mark D. Lazarus, MD,[c] and Andrew R. Karduna, PhD,[a] *Eugene, OR, and Philadelphia, PA*

The purpose of this study was to determine whether plane, end-range determination, or scapular motion affects shoulder range-of-motion measurements. In 16 healthy subjects, instrumentation with a magnetic tracking device was used to measure shoulder internal and external range of motion. The arm was supported while it was rotated either actively or passively with a measured torque. There was a significant main effect of plane for internal rotation ($P < .001$) but not for external rotation ($P = .584$). Passive humerothoracic motion was significantly greater than active humerothoracic motion for internal rotation ($P < .006$) and external rotation ($P < .01$). Active and passive humerothoracic motion was significantly greater than active and passive glenohumeral motion in 6 of the 7 active conditions and all 7 passive conditions ($P < .002$). Our results suggest that significant amounts of scapulothoracic motion may impact measurements of isolated glenohumeral joint motion.
(J Shoulder Elbow Surg 2005;14:602-610.)

The shoulder is one of the most mobile joints in the human body and moves in a complex 3-dimensional pattern. This motion is accomplished through coordinated interactions between 3 diarthrodial articulations: the glenohumeral, acromioclavicular, and sternoclavicular joints, of which the former has the largest range of motion. However, this mobility comes at a price, as this joint is the most frequently dislocated in the body. Whereas active muscle contraction and glenoid geometry are primarily responsible for stability in the mid ranges of motion, the ligaments and capsular structures are mainly responsible for stability at the end ranges of motion.[36,46] A failure of any of these stabilizers can negatively affect shoulder kinematics and may result in decreased glenohumeral joint function.[10]

From a biomechanical perspective, the glenohumeral joint is typically described as having the following 3 degrees of rotational freedom: plane of motion, elevation, and internal and external rotation.[2] Although many of the traditional studies of shoulder motion have primarily focused on shoulder elevation,[20,43] there has been considerable interest of late in measuring internal and external rotation along the long axis of the humerus.[9,45] Study of this motion is important for two main reasons. First, the available range of internal and external rotation impacts shoulder function, from simple activities of daily living, such as hair combing, to more complex tasks required by athletes and occupational workers. Depending on the level of force applied throughout the shoulder joint, osseous and soft-tissue adaptations can result from repetitive shoulder motions. For example, bodybuilders have a decreased internal range of motion,[5] whereas professional baseball pitchers have an increased external range of motion coupled with a decreased internal range of motion.[7,15,19,22] Second, measurements of internal and external rotation can be used as indicators of capsular tightness. Cadaveric studies using either selective cutting protocols[11,24,34,39,50] or strain measurements,[14,42,49,52] as well as numerical models,[13,40] have been used to assess the extent to which various portions of the capsule limit rotation. Clinically, measurements of capsular tightness are used in the assessment of patients with impingement syndrome.[51,54]

The American Academy of Orthopaedic Surgeons' current recommendations for clinical measurement of shoulder rotation is by goniometer for external rotation with the arm at the side and for internal and external rotation with the arm at 90° of humeral abduction.[25] In addition, internal rotation with the arm at the side is assessed by having the patient reach behind his or her back and noting what vertebral level can be reached with the thumb. These measurements are cost-effective and easy to perform and have fair to good intrarater and interrater reliability.[28] However, from a biomechanical perspec-

From the [a]Department of Human Physiology, University of Oregon, Eugene, and [b]Drexel University School of Medicine and [c]Department of Orthopaedic Surgery, Thomas Jefferson University, Philadelphia.

Reprint requests: Andrew R. Karduna, PhD, University of Oregon, Department of Human Physiology, Eugene, OR 97403 (E-mail: *karduna@uoregon.edu*).

1058-2746/2005/$30.00
doi:10.1016/j.jse.2005.05.003

tive, there are 3 key limitations. First, as with all goniometric assessments, end range is determined by clinical feel, as opposed to an objective assessment of torque. In a laboratory setting, investigators can measure the manual torque applied to an in vivo joint during range-of-motion measurements—this approach has been used for the ankle,[47] cervical spine,[38] and shoulder.[41] The second concern is that, although goniometers may be designed and used to assess glenohumeral motion, they are really measuring both glenohumeral and scapulothoracic motion. Previous studies have shown that motion of the scapula can have a significant effect on both goniometric[4,9] and vertebral level[37] measurements. Isolating glenohumeral motion typically requires a fixation technique to prevent unwanted scapular motion. This approach is difficult to perform and may introduce unwanted artifact into the measurement. The third issue is that the effect of the plane of motion has not been well documented, as most studies have focused on a single plane of motion.

Given that the standard clinical assessment of this motion involves a measurement of passive humerothoracic rotation, the purpose of this study was to determine differences that might be a result of the 3 factors discussed above—that is, the effects of (1) testing plane, (2) end-range determination (active vs passive), and (3) scapular motion.

MATERIALS AND METHODS

Study design and subjects

Sixteen subjects participated in this study (age range, 20-32 years). All indicated that they had no history of cervical or shoulder pain or pathology, and all were recruited from a diverse university population. Subjects were excluded from the study if they had (1) less than 135° of active humeral elevation in the scapular plane, (2) prior shoulder surgery, (3) shoulder injury in the past 6 months, or (4) presence of shoulder pain preventing the correct execution of tests, including an inability to achieve relaxation during testing. The subjects comprised 8 women and 8 men, with a mean age of 23 ± 3 years, a mean height of 171 ± 10 cm, and a mean mass of 68 ± 10 kg. The dominant shoulder of each subject was tested. Approval for this study was obtained from the Institutional Review Board of the University of Oregon, Eugene, OR. All subjects were informed of the nature and details of the study and gave written and verbal consent before their participation.

Electromyographic measurements

A Myopac Jr (Run Technologies, Mission Viejo, CA) unit with 4 dual-lead channels was used for collecting and processing electromyography (EMG) recordings from superficial shoulder musculature. The chief purpose of these recordings was to ensure that there was minimal muscle activity during passive positioning of the arm. EMG activity was recorded from the pectoralis major (sternal head), trapezius (upper fibers), middle deltoid, and infraspinatus. Initial identification of muscle locations was determined based on the recommendations of Cram et al,[16] with subject motion and manual palpation as the final determinants for electrode placement. Surface EMG was recorded by use of a bipolar lead with 2 pediatric electrodes (Blue Sensor Electrodes, Ambu, Olstykke, Denmark) located 3.4 cm apart (center-to-center distance) and placed parallel to the primary muscle fiber alignment. A single-lead grounding electrode was placed on the dominant-side clavicle for signal noise reduction. The system has a common mode rejection ratio of at least 90 dB and a high-pass (10 Hz) and low-pass (1000 Hz) filter. Data were collected at a sampling rate of 1200 Hz. For the passive motions, the data were run through a root mean square algorithm with a window of 50 milliseconds.

To normalize EMG activity levels during arm motion, maximum voluntary contractions (MVCs) of the muscles were obtained during a 5-second contraction, with the amplitude of the contraction being determined as the root mean square value over the middle 2 seconds of the contraction. The following test positions were used: pectoralis major, each humerus in 90° of flexion with the palms of each hand pressed together[33,35]; middle deltoid, 90° of shoulder abduction with the elbow at 90° of flexion and the forearm parallel to the floor[1]; upper fibers of the trapezius, humerus at the side and a shrug of the shoulders[16]; and infraspinatus, 90° of humeral flexion in the scapular plane with the elbow fixed at 90° with 30° of internal rotation.[32] The larger of the two amplitudes from the middle deltoid and shrugging test was used for the upper trapezius fiber MVC value.

Shoulder kinematic measurements

A Polhemus 3Space Fastrak device (Polhemus, Colchester, VT) was used for collecting 3-dimensional in vivo kinematics of the shoulder complex. The room was surveyed for metallic objects that may have disrupted the electromagnetic field. We found minimal effects of metal in the room, and the torque cell had a minimal effect on the humeral receiver. Subjects were asked to sit with their thoracic spine, scapula, and humerus exposed for receiver placement. Women were asked to wear a sports bra, which allowed access to the entire scapular region. Subjects sat with their feet flat on the floor at a comfortable width apart, back against the chair, and eyes fixed forward. This position was maintained throughout marker placement and digitization procedures.

The Fastrak is a magnetic tracking device that consists of a global positioning transmitter, 3 receivers (thoracic, scapular, and humerus), and a digitizing probe, which is hardwired to the system electronics unit. As defined by the manufacturer, when the transmitter is within 76 cm of the sensors, the system is accurate to 0.8 mm. The transmitter was firmly attached to the seat back of the chair at a distance of 50 cm from the thoracic receiver and was leveled. Because the receivers were always within this range of the transmitter, we are confident that we can trust the accuracy measurements provided by the manufacturer (Polhemus). In addition, a bone pin study validating this surface-mounted scapular-tracking technique has been previously described.[30] The first receiver was placed on the spinous process of the third thoracic vertebra by use of spirit gum adhesive and Micropore surgical tape (3M Health-

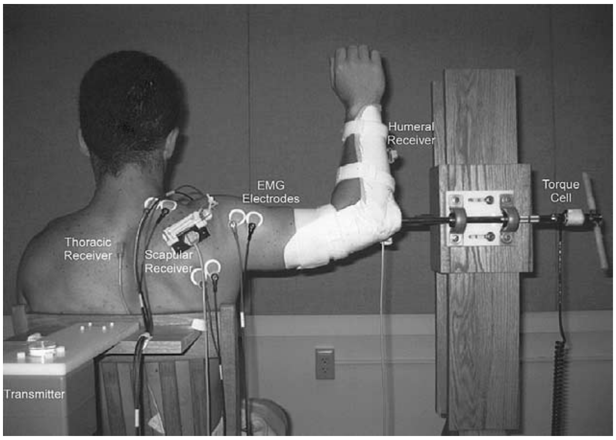

Figure 1 Image depicts experimental setup with transmitter and receiver locations, EMG electrode placement, and supporting wood stand with attached torque cell jig.

Care, St. Paul, MN) to fix the receiver to the skin. The second receiver was mounted on the distal forearm portion of a custom-made Polyform splint (Sammons Preston Roylan, Bolingbrook, IL) positioned on the dominant elbow with elastic straps. The final receiver was positioned over the scapula after it was mounted on a custom-made and previously validated scapular-tracking device machined from plastic[30] (Figure 1). The base of the scapular tracker is plastic and has a hinge joint that conforms to the spine of the scapula. From this base, an adjustable arm extends and contacts the acromion. The base and the arm contacting the acromion were attached to the skin with adhesive-backed Velcro strips placed above and below the spine of the scapula and on the flat part of the acromion, just proximal to the origin of the deltoid muscle. The scapular tracker remained well fixed to the scapula during motion from these Velcro attachments.[30]

A series of standardized embedded axes were established. These axis systems were derived from a series of anatomic landmarks proposed by the shoulder subcommittee of the International Society of Biomechanics committee for standardization and terminology.[53] The landmarks on each bony segment are digitized in the following order: seventh cervical vertebra, eighth thoracic vertebra, sternal notch, xiphoid process, acromioclavicular joint line, root of the scapular spine, inferior angle, medial epicondyle, lateral epicondyle, and center of humeral head. All landmarks were on the surface of each subject and could be digitized directly, with the exception of the humeral head. The center of the humeral head was determined as the point that moved the least with regard to the scapula when the humerus was moved throughout short arcs of mid-range glenohumeral motion and was calculated by use of a least squares algorithm.[27]

Standard matrix transformation methods were applied to determine the rotational matrix of the humerus with respect to the thorax or scapula.[53] Humeral rotations were represented by use of a standard Euler angle sequence, where the first rotation defined the plane of elevation, the second rotation defined the amount of elevation, and the last rotation represented the amount of internal and external rotation.[2]

Shoulder kinetic measurements

Investigator-applied internal and external rotational torque of the shoulder was measured with a torque cell (model No. SWS-100; Transducer Technologies, Temecula, CA). The torque cell was connected to a square drive extension and mounted to a fixed wooden stand, which supported the weight of the arm (Figure 1). The square drive extension (directed in line with the long axis of the humerus) was connected to the Polyform elbow-immobilizer splint. Pilot work showed highly variable torque tolerances for each subject in any given plane. Therefore, a torque threshold (4 Nm) was adopted and defined the maximum end range of motion for both internal and external rotation.[41] Because the weight of the forearm was not negligible, the torque due to gravity was also estimated as a function of the rotation angle. Similar to the gravity correction used for knee motion on an isokinetic machine,[23] this required a measurement of the torque due to gravity with the forearm supported horizontally.

Polyform immobilizer splint

A custom-made Polyform immobilizing splint was used to measure internal and external humeral rotation accurately for all data collection. The immobilization splint was positioned on the humerus and forearm, fixing the elbow at 90° of flexion. The immobilizing splint acts as a rigid structure, allowing secure placement of the humeral receiver, and connects to the square drive extension on the wooden stand (Figure 1). With the humeral plane fixed, the immobilizing splint's design maximizes uniaxial measurements within that plane. Clinically, internal rotation with the arm at the side is measured by the subject reaching behind his or her back and touching the highest vertebral level with the thumb. Because of the nature of our immobilization splint (elbow fixed), subjects were unable to perform this task.

Testing protocol

Before data collection, the Beighton hypermobility test was used to measure the degree of general joint laxity.[6] The glenohumeral joint was then preconditioned by placing it in 90° of elevation in the coronal plane while the investigator passively rotated the shoulder. Subjects verbally confirmed when a good stretch was felt in the shoulder capsule, and the investigator held the position for 10 seconds. Three internal and three external stretches were used to complete shoulder preconditioning. Although preconditioning is not used clinically, it was chosen for this study because it ensures that every subject started the test with the same tissue loading history. Without preconditioning, multiple plane measurements might have introduced order effects during data collection, with a larger range of motion observed for those measurements made at the end of the protocol.

Subjects were asked to perform both active and passive shoulder rotation in 4 conditions: 90° of humeral elevation in the coronal, scapular, and sagittal planes and with the arm at the side. Scapular plane orientation was defined as

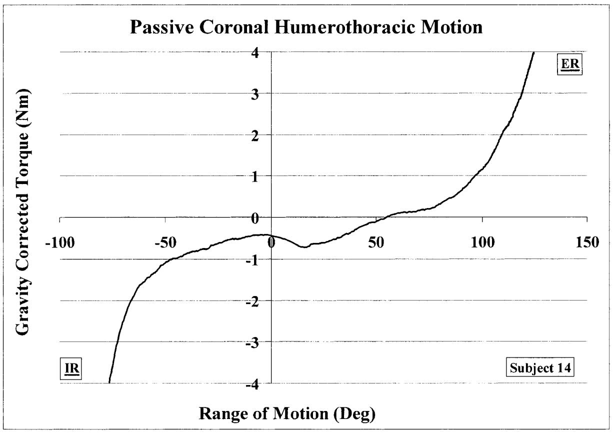

Figure 2 Representative plot (subject 14) of passive humerothoracic angle versus gravity-corrected torque for internal rotation (*IR*) and external rotation (*ER*) in coronal plane.

approximately 30° to 35° anterior to the coronal plane. To ensure proper placement for a given humeral position, real-time confirmation of the position of the humerus was checked through onscreen visual feedback from the magnetic tracking device. Because there were 4 humeral planes, 2 muscle activities (active and passive), and 2 rotations (internal and external), order effects were accounted for with the use of a balanced Latin square design.[44]

For active motion, both maximal internal rotation and external rotation were assessed for each humeral position. Because of physical contact between the forearm and abdomen, only external rotation was assessed with the arm at the side. Two trials were collected, one in which internal rotation preceded external rotation and one in which the reverse was true. The subject was placed in the desired humeral plane and was seated in a chair facing forward, not able to view the computer monitor. The investigator was responsible for ensuring that the subject maintained proper arm position and motivation during each trial. Once the subject could perform the active motion properly, data collection began. On the basis of our digitization protocol, the zero point (0°) of neutral rotation was defined as the point when the subject's forearm was essentially parallel to the floor.

The protocol for assessing passive motions was identical to that for active motion, except arm motion was controlled by the investigator. Tester-assisted torque was applied to the torque cell coupled to the Polyform splint, resulting in shoulder rotation. Subjects relaxed as much as possible in order to manipulate the shoulder with minimal active muscular resistance. Maximum passive external and internal rotation angles were recorded, as well as the required amount of gravity-corrected torque to obtain these angles (Figure 2). To prevent the investigator from applying too much torque, an audible sound would chime when 4 Nm was met or exceeded, at which time the investigator would reduce the torque and begin rotating in the opposite direction. If subjects began to feel discomfort or instability in their shoulder during any part of the testing procedure, they could tell the investigator to cease their rotation. Therefore, either the subject would allow the computer-preset threshold of 4 Nm gravity-corrected torque to define the end range of motion or the subject's discomfort defined the end range of motion for each plane. EMG from each passive trial was analyzed after each trial to determine percent muscle activation levels. Trials were only accepted if the EMG remained below a threshold set at 15% of MVC. Pilot data provided a range of passive EMG percentages, and many subjects were unable to relax below a 15% MVC threshold. We acknowledge the accepted value (15% MVC) as an arbitrary number, but it was shown to be a reasonable and reliable threshold value for all subjects.

Data reduction and analysis

For all calculations, data were averaged over the 2 trials collected (internal rotation first and external rotation first). The following dependent variables were calculated: passive humeral motion with respect to the thorax (P_{HT}), passive humeral motion with respect to the scapula (P_{GH}), active humeral motion with respect to the thorax (A_{HT}), and active humeral motion with respect to the scapula (A_{GH}). For all analyses, internal rotation data were analyzed separately from external rotation data. To evaluate intrasession reliability, intraclass correlation coefficients [ICC (3, 1)] were run on P_{HT}, A_{HT}, P_{GH}, and A_{GH}.

To determine the effect of plane on passive motion, a 1-way repeated-measures analysis of variance (ANOVA) was run on P_{HT}. To determine the effect of muscle activity, a 2-way repeated-measures ANOVA was run on humerothoracic motion, having two within-subject factors: plane of motion (coronal, scapular, sagittal) and muscle activity (P_{HT} vs A_{HT}). To determine the effect of scapulothoracic motion, a 2-way repeated-measures ANOVA was run on the passive data, having two within-subject factors: plane of motion (coronal, scapular, sagittal) and type of motion (P_{HT} vs P_{GH}). This ANOVA was repeated for the active data (A_{HT} vs A_{GH}). When appropriate, follow-up 1-way ANOVAs and paired *t* tests were performed. An additional paired *t* test was used for the position of the arm at the side. Because there were a total of 4 ANOVAs, the α level was set at .0125 (.05/4).

RESULTS

Some subjects did not reach the 4-Nm threshold during data collection. In order of planes, the percentages of subjects not reaching the 4-Nm threshold were as follows: external rotation with the arm at the side, 12% (2/16); external rotation in the coronal plane, 12% (2/16); internal rotation in the coronal plane, 12% (2/16); external rotation in the scapular plane, 12% (2/16); internal rotation in the scapular plane, 37% (6/16); external rotation in the sagittal plane, 25% (4/16); and internal rotation in the sagittal plane, 44% (7/16).

General joint laxity was measured via the Beighton hypermobility scoring system before data collection with a range of 1 (no joint laxity) to 8 (multiple joint laxity).[6] This hypermobility test served as a descriptive tool, which demonstrated that none of our subjects

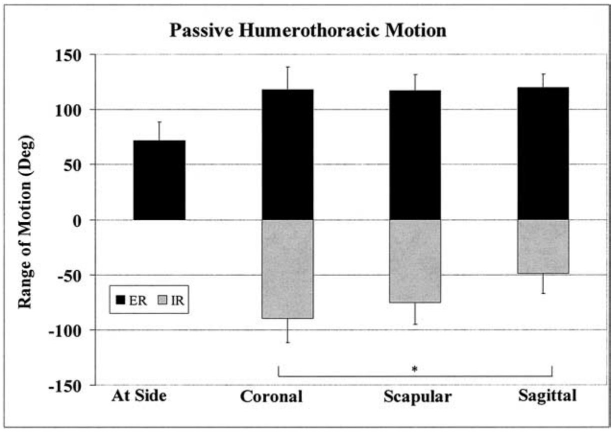

Figure 3 Mean (± SD) passive humerothoracic motion across all 7 conditions tested. A repeated-measures ANOVA showed a significant main effect of plane for internal rotation (*IR*) but not for external rotation (*ER*). *Asterisk*, $P < .05$.

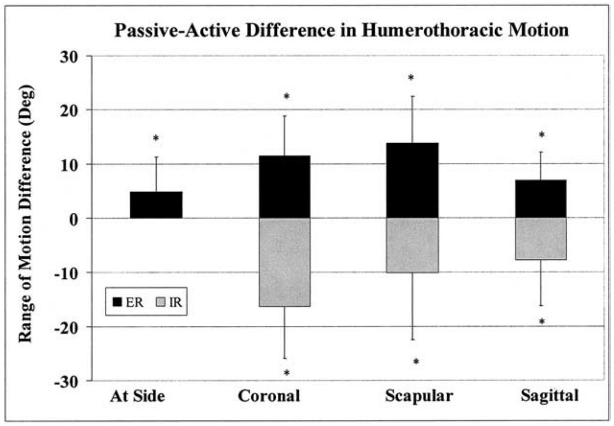

Figure 4 Mean (±SD) differences between passive and active humerothoracic motion. *ER*, External rotation; *IR*, internal rotation. *Asterisk*, $P < .05$.

Table I Humerothoracic motion by plane, rotation, and muscle activity

	Arm at side (°)	Coronal (°)	Scapular (°)	Sagittal (°)
Active				
ER	67 ± 17	107 ± 19	103 ± 14	113 ± 11
IR	N/A	−73 ± 18	−65 ± 11	−41 ± 14
Passive				
ER	72 ± 17	118 ± 21	117 ± 15	120 ± 12
IR	N/A	−90 ± 22	−75 ± 20	−49 ± 19

Data are presented as mean ± SD.
ER, External rotation; *IR,* internal rotation; *N/A,* not applicable.

had abnormal generalized joint laxity. The mean laxity score for the present study was 2.3. Women were shown to have higher mean scores (3.1) than men (1.4).

All intraclass correlation coefficient values exceeded 0.85, indicating good to excellent reliability. The results of the 1-way ANOVAs for P_{HT} showed a significant main effect of plane for internal rotation ($P < .001$) but not for external rotation ($P = .584$) (Figure 3). Results for both internal and external rotation comparing P_{HT} and A_{HT} are presented in Table I. These results demonstrate a significant effect of muscle activity ($P < .001$). Paired *t* tests for internal rotation demonstrated a significantly larger range of motion passively compared with actively for the coronal ($P < .001$), scapular ($P = .005$), and sagittal ($P = .002$) planes. Similarly, for external rotation, a significantly larger range of motion for passive motion as compared with active was demonstrated for the coronal ($P < .001$), scapular ($P < .001$), and sagittal ($P < .001$) planes, as well as with the arm at the side ($P = .009$) (Figure 4).

The results for both internal and external rotation comparing A_{HT} and for A_{GH} are presented as difference scores and demonstrate a significant main effect for motion type ($P < .001$). Paired *t* tests for internal rotation demonstrated a significantly larger range of humerothoracic motion for the coronal ($P < .001$) and sagittal ($P < .001$) planes, but not for the scapular ($P = .08$) plane. For external rotation, significantly larger humerothoracic motions were demonstrated for the coronal ($P < .001$), scapular ($P < .001$), and sagittal ($P < .001$) planes, as well as with the arm at the side ($P < .001$) (Figure 5, *A*).

The results for both internal and external rotation comparing P_{HT} and for P_{GH} are presented as difference scores and demonstrate a significant main effect for motion type ($P < .001$). Paired *t* tests for internal rotation demonstrated a significantly larger range of humerothoracic motion for the coronal ($P < .001$), sagittal ($P < .001$), and scapular ($P = .001$) planes. For external rotation, significantly larger humerothoracic motions were demonstrated for the coronal ($P < .001$), scapular ($P < .001$), and sagittal ($P < .001$) planes, as well as with the arm at the side ($P < .001$) (Figure 5, *B*).

DISCUSSION

On the basis of the work of Novotny et al,[41] a technique was developed to allow for a quantitative assessment of passive glenohumeral internal and external rotation with a measurable torque. Key additions provided by the present study allowed for measurements of both active and passive motion across different humeral planes and simultaneous measurement of glenohumeral and humerothoracic motion. Our findings establish a comprehensive description of rotational motion of the humerus in healthy individu-

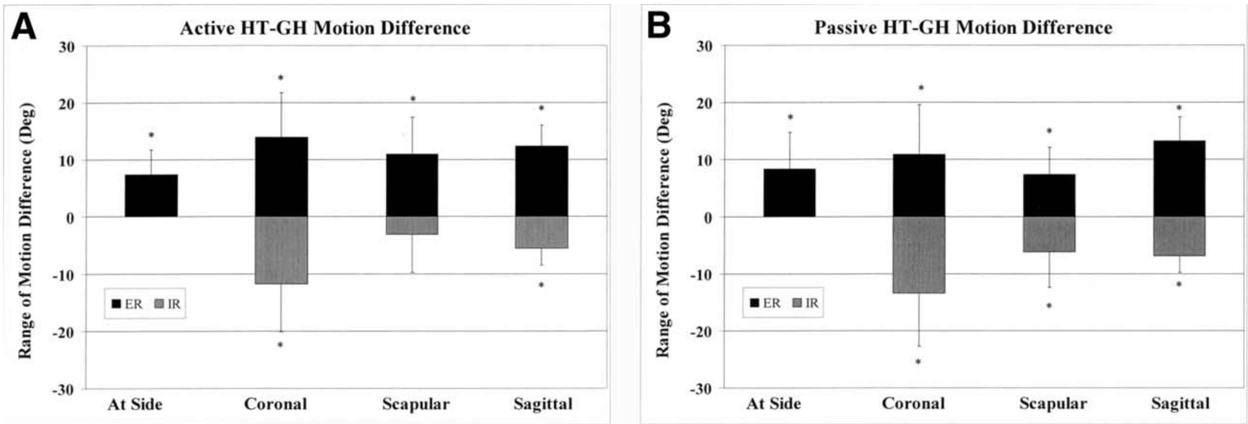

Figure 5 A, Mean (± SD) differences between active humerothoracic (*HT*) and active glenohumeral (*GH*) motion. *Asterisk*, $P < .05$. **B**, Mean (± SD) differences between passive HT and passive GH motion. *Asterisk*, $P < .05$. *ER*, External rotation; *IR*, internal rotation.

als and allow for a better understanding of joint function.

In brief, the major findings of the present study are as follows. In the 4 humeral planes (7 conditions tested), passive humerothoracic motion was significantly greater than active humerothoracic motion in every condition. Measurements of passive humerothoracic internal rotation had a significant humeral plane effect, whereas passive external humerothoracic rotation measurements showed no effect of plane. A significant effect of scapulothoracic motion (defined as the difference between humerothoracic and glenohumeral motion) was shown in 6 of the 7 active conditions and all 7 passive conditions tested (Figure 5, A and B).

The range of motion in external rotation with the arm at the side in the present study (72°) is within the range reported by other groups studying this motion (54°[19] to 81°[7]). Similarly, total humerothoracic motion in the coronal plane in the present study (208°) agrees well with previous reports in the literature (190°[17] to 224°[12]). Presumably, because of the recommendation by the American Academy of Orthopaedic Surgeons,[25] the vast majority of studies in the literature only assessed rotational range of motion in these two positions. Total scapular plane motion in the present study (192°) was larger, approximately 31° higher, than was reported in the one study we could find that evaluated humeral rotation at 90° of elevation in the scapular plane.[35] Other studies have evaluated rotational motion in the scapular plane but at only 45° of elevation.[17,41] Interestingly, we were unable to find any studies reporting internal or external humeral rotation with the arm in the sagittal plane.

Differences between active and passive range of motion emphasize the importance of assessing range of motion in a clinical setting consistently. This is important, considering that reports are inconsistent with regard to how end range is determined. Whereas some use active positioning alone[28] or with the effects of gravity,[21] others use passive positioning determined by capsular end feel,[4,45] 2 lb[29] or 4 lb[5] of force, scapular liftoff,[54] or pain.[3]

The results of the present study indicate that in all planes of motion tested, in vivo passive humerothoracic motion was significantly greater than active humerothoracic motion, which is consistent with previous in vivo[26] and cadaveric[31] reports. The most dramatic effect was observed for internal rotation in the coronal plane, with mean passive motion exceeding mean active motion in excess of 15° (Figure 4). The limited range of motion observed actively could be a result of either a mechanical or neurologic limitation of torque production. Mechanical limitations include nonfavorable lines of action and insufficient muscle filament overlapping causing a reduction in contractile force. Neurologic limitations may result from negative feedback mechanisms, where the central nervous system attempts to prevent unwanted motion by inhibiting muscle contraction.[18,48]

With regard to the effect of plane on range of motion, our original hypothesis was that as the arm was moved into a more anterior plane, there would be a decrease in internal range of motion and an increase in external range of motion. This hypothesis was based on the assumption that end range was mainly a result of tightening of the inferior glenohumeral ligament.[49] Although this hypothesis was supported for internal rotation, it was not supported for external rotation (Figure 3). To help understand these results, we compared the glenohumeral range of motion data in the present study with similar data collected in cadavers.[31] There is a clear agreement with the internal rotation trend (Figure 6). However, for external rotation, whereas the present study showed

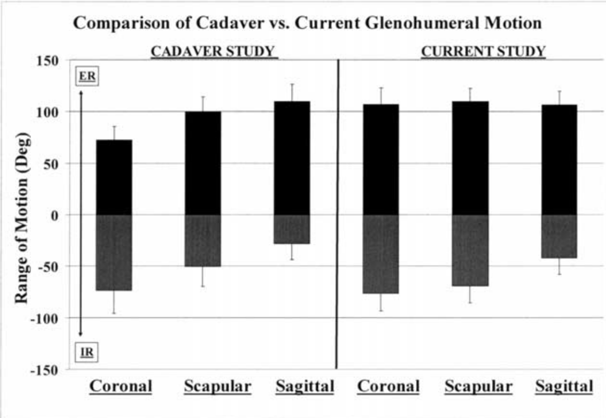

Figure 6 Comparison between passive glenohumeral motion from cadavers[31] and passive glenohumeral motion in this study. All cadaveric specimens showed the expected trend of decreasing internal rotation (*IR*) and increasing external rotation (*ER*) as the humerus was moved to a more anterior plane. The present study showed a similar trend for internal rotation but not for external rotation.

no effect of plane, the cadaveric study showed the expected pattern of an increasing external range of motion as the plane became more anterior (Figure 6). In fact, for both internal and external rotation, every cadaveric specimen demonstrated the pattern of sagittal greater than scapular greater than coronal for external rotation and the reverse for internal rotation. In the cadaveric study, the joints were dissected down to the capsule. However, the present study was performed in vivo, and many of the secondary passive tissues (muscles, skin, fascia, and intracapsular pressure) and potential active constraint (muscle coactivation) could have increased the net resistance during rotational motion.

One explanation for these discrepancies is that in an in vivo setting, the capsule is the end restraint to internal but not external rotation. For example, in the sagittal plane, additional passive restraint during external rotation might come from the long head of the biceps brachii muscle. In addition, the discovery of Pacinian and Ruffini corpuscles around the capsule and within the musculotendinous fibers has led to new theories regarding how the glenohumeral joint may be stabilized at the end range of motion.[48] Blasier et al[8] found that shoulder proprioception was significantly more sensitive to externally versus internally rotated positions and found that this effect was more pronounced in persons with minimal joint laxity. Therefore, it may be that limitations in external rotation are indirectly influenced by capsular tension resulting in reciprocal efferent muscle activation. Coactivation of the subscapularis, teres major, and latissimus dorsi muscles would result in an increase in internal rotation torque, potentially protecting joint integrity during extreme externally rotated positions.[48] Because we did not measure the EMG activity of these muscles in the present study, we cannot confirm this hypothesis.

Past studies have predicted that glenohumeral motion provides the majority of motion as a result of the spherical head of the humerus articulating with the shallow surface of the glenoid on the scapula. The present study's ability to measure both active and passive humerothoracic and glenohumeral motion confirmed 2 key points. First, on average, the active glenohumeral articulation was responsible for approximately 89% (range, 84%-96%) of the total motion in the 7 humeral planes. The passive glenohumeral articulation, on average, was responsible for approximately 89% (range, 86%-94%) of the total motion in the 7 humeral planes. The largest percentage of active (90%-96%) and passive (93%-94%) glenohumeral motion arose from rotations in the scapular plane and lends support to the concept that this is the most suitable plane for isolating glenohumeral motion (Figure 5, *A* and *B*). Because the majority of motion is occurring at the glenohumeral joint, an accurate measurement of glenohumeral motion is crucial for clinical diagnosis. However, most clinical shoulder measurements are made with a goniometer, which records a combination of glenohumeral and scapulothoracic motion. The present study found that passive humerothoracic measurements overestimated glenohumeral measurements by as much as 24° in the coronal and sagittal planes and as little as 14° in the scapular plane (Figure 5, *B*). Therefore, if isolated glenohumeral motion is sought and it is not possible to measure scapular rotations, either the scapula must be stabilized[9] or the measurement stopped when the scapula translates or rotates away from the thorax.[4]

Several limitations of the present study must be addressed. The first is that angular velocities during passive trials were not held constant across subjects or planes. Reflexive feedback to the musculature may occur if the angular velocity is too fast, and tissue elongation (creep) may occur with a constantly applied force over time. However, subject relaxation (based on EMG thresholds) was maintained, and no reflexive protective activation was noted in the muscles that were monitored. The direct application of the scapular tracker to the skin may have introduced skin motion artifact in the measurement of glenohumeral motion. However, this method has been previously validated against bone pin measurements.[30] All shoulders were put through a preconditioning protocol before data collection. As mentioned previously, this was done to ensure a similar loading history for all subjects. However, the effect of preconditioning on such factors as subject motivation and proprioception was not assessed in the present study.

We acknowledge that 4 Nm may not have been enough torque to engage the capsular structures. However, Kuhn et al[34] found that 3.4 Nm was enough torque to engage the capsule in an active cadaveric model study. For all passive range-of-motion trials, there was either a predetermined threshold value of 4 Nm or a subject-defined threshold level. These threshold levels established the end range of motion during passive internal and external rotation and could be affected by sex, pain threshold, or overall laxity. Because differences exist in shoulder musculature between sexes, it may prove difficult to compare internal and external rotation between sexes based on the current predetermined torque thresholds.

In conclusion, an assessment of internal and external rotation is a standard part of a clinical examination of the shoulder. It is important for clinicians to understand that these measurements are dependent on the plane tested and how end range is determined. In addition, if the scapula is not stabilized, a significant amount of motion may be occurring at the scapulothoracic articulation.

We thank Jason M. Henkle for assistance with data collection and subject recruitment.

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