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Analyzing glenohumeral torque-rotation response *in vivo*

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Abstract

Background—Because the human shoulder has many degrees of freedom that allow redundant means of producing the same net humerothoracic motion, there are many impediments to objective, repeatable assessment of shoulder function *in vivo*. Devices designed to date have suffered from poor reliability. In this study we introduce a new device and methods to evaluate human shoulder kinematics and evaluate its reproducibility from subject to subject and from day to day.

Methods—This was a controlled laboratory study. Using electromagnetic motion sensors to record the position and orientation of the thorax, scapula, and humerus, we quantified the kinematic response of twenty four normal shoulders in response to known internal-external torque application. A four-parameter logistic function was selected to characterize the strident features of the torque-rotation relationship.

Findings—Our analysis in conjunction with the measurement technique described herein, allowed the passive glenohumeral internal-external range of motion to be differentiated from other motion components and was determined to within 9.6% of full scale over three repeated trials. Range of motion was the most reliable biomechanical outcome, more so than computed indices of glenohumeral flexibility and hysteresis. The exact profile of the torque-rotation response, and therefore the repeatability of the calculated outcomes, was unique from shoulder to shoulder.

Interpretation—The development of the capacity for precise, non-invasive measurement of shoulder biomechanics over time is a requisite step towards optimizing treatment of shoulder injury and disease. Our current methods are superior to previous attempts at trying to non-invasively evaluate the biomechanics of the glenohumeral joint.

Keywords

shoulder; kinematics; biomechanics

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INTRODUCTION

Multiple components contribute to global shoulder motion, making reliable, objective assessments of joint function *in vivo* a challenge. This is especially true when evaluating patients who lie in the part of the spectrum between normal function and strikingly obvious impairment. Unfortunately, this same population may garner the greatest benefit from early and/or novel therapeutic interventions. A healthy shoulder exhibits greater mobility than any other human joint. Moreover, because the shoulder helps to control the position of the hand in space, restriction of shoulder girdle motion can lead to serious functional impairment. In furthering our understanding of shoulder impairments and their treatment, there is a limit to the information that can be derived from cadaveric studies and current mathematical models, since dynamic soft tissue behavior governs shoulder stability and mobility to a large extent (An 2002). Thus, *in vivo* examination of shoulder biomechanics is particularly useful for establishing a basis for this understanding.

The current study focuses primarily on the internal-external (IE) rotation component of glenohumeral motion, which is related to chronic instability (Gill et al., 1997) and often incapacitated by traumatic injury and arthritis (Jolles et al., 2004). Loss of internal and external rotation is not well tolerated by patients, as it interferes with many activities of daily living (Gill et al., 1997; Rundquist and Ludewig 2004). Active external rotation loss is also an important differentiator of rotator cuff tear patients (Silver et al., 2006). However, clinical assessments of internal and external rotation suffer poor intra- and inter-observer reproducibility (Bovens et al., 1990;). This may, in part, be due to the inherent subjectivity in conducting such assessments under unknown loading conditions (Novotny et al., 2000). Most clinical indices are also categorical in nature, which renders them less precise than a continuous measure. Before understanding the role of shoulder internal-external rotation in instability and in surgical and rehabilitation outcomes, a reliable, non-invasive technique is needed to accurately assess rotational range of motion (ROM) and flexibility *in vivo*. The rationale for the current investigation was to develop an improved method to measure rotational response to controlled torque application. With such a technique, it was hypothesized that in normal subjects all the parameters used to describe the glenohumeral biomechanics would be relatively invariant over three testing periods.

METHODS

Twelve asymptomatic subjects (six males and six females) with no history of shoulder surgery or pathology were recruited for study. All subjects gave their informed consent to the protocol, which had been approved by our Institutional Review Board. Both shoulders were tested. Clinical assessment of function was documented using the American Shoulder and Elbow Surgeons (ASES) scale (Richards et al., 1994), which contains ratings or measurements for range of motion, pain, strength, and instability. After receiving a satisfactory clinical assessment, each shoulder of each subject was tested during three separate sessions. The first two sessions occurred successively on the same day; the final session was conducted one week afterwards at the same time of day.

During each test session, the measurement system was configured and applied, data were collected, and then the sensors were removed and the subject taken out of the device. Following a brief warm up, electromagnetic position and orientation sensors (MiniBIRD 800, Ascension Technology Corp., Milton, VT) were secured to the skin via medical adhesive tape and benzoin prep at the level of the first thoracic vertebra (T1), the posterolateral aspect of the acromion, and the distal humerus. Subjects were seated in the “beach chair position” (reclined backwards 10°), and their back and shoulders were supported with a vacuum pack (Olympic Medical, Seattle, WA) and a strap which crossed

their torso diagonally (Figure 1). The arm was abducted such that the subject could reach full internal rotation without contacting his or her body or the seat (60° of abduction in the scapular plane). Digitization of specific anatomical locations (thorax: jugular notch, C7/T1, T8, T12, xyphoid process; humerus: medial and lateral epicondyles; scapula: posterolateral acromion, inferior angle, medial border of scapular spine) with the subject in neutral position (arm at side; thumb forward) referenced the electromagnetic sensors to the underlying bony anatomy. The proximal humerus was located by the center of rotation method (Veeger 2000). The arm was secured in a custom-molded thermoplastic brace (Orthoplast, Depuy Casting, Raynham, MA) with the elbow bent to 90°. The brace interfaced with a six degree-of-freedom load cell (Mini45, ATI, Apex, NC) that was mounted rigidly to a rotation platform with a hinged gear joint (Gray and Prior Machine Co., Bloomfield, CT) to accommodate naturally occurring changes in abduction-adduction throughout the internal-external ROM (IE ROM)). A moment was applied to induce maximum internal rotation of the humerus, and then reversed to induce maximum external rotation. An additional electromagnetic sensor recorded the position and orientation of the load cell throughout testing. Load cell and motion sensor data were recorded at 100 Hz, and smoothed using a fourth-order, zero-phase-shift Butterworth filter with a 20 Hz cutoff frequency (Winter 1990). During analysis, force and torque components were transformed from the load cell center into the local (moving) coordinate axes of the humerus (see Equation 1). Four repetitions of internal-external rotation were completed during each session. The Euler angle sequence describing glenohumeral joint motions followed the International Shoulder Group convention (Wu et al., 2005). Based on the orientation of the forearm (assuming the long axis of the forearm was perpendicular to the transepicondylar axis), and on body segment parameter estimates (Dempster and Gaughran 1967), the moment contributed by the mass of the forearm and hand was differentiated from the applied coupled torque. The direct linear transformation to obtain glenohumeral IE rotation torque was:

$$[\mathbf{S}]^T \times [\mathbf{B}] \times [\mathbf{H}] \times \mathbf{M}_{LoadCell} - [\mathbf{S}] \times \mathbf{r} \times \mathbf{F}_{Global} + [\mathbf{S}] \times \mathbf{l} \times 0.016 \times mg \quad (1)$$

where S = shoulder (humeral) orientation matrix, B = load cell sensor orientation matrix, H = transfer function between load cell and load cell sensor, M = moments registered at the load cell, F = load cell forces transformed to the global coordinate system, m = body mass, g = gravity vector, l = forearm vector, and r = vector from load cell to humeral origin.

The torque-rotation curve can be conceptualized as having two distinct phases, each covering the full IE ROM. One portion represents induced movement in the direction of increasing internal rotation from the release of external rotation torque and return to neutral, and through to the maximum application of the internal rotation torque (“internal loading”); the other section traverses the sequence from release of internal rotation torque, back through neutral, to application of a maximum external rotation torque (“external loading”). From the 100-Hz sampling, data from each phase were binned into 0.2-Nm increments for curve fitting. Each phase of the sigmoidal rotation-versus-torque curve was then fit with a four-parameter logistic, using PROC NLIN (an iterative nonlinear regression method) in SAS (Version 9.1.3, SAS Institute, Inc., Cary, NC). The parameterization employed was:

$$Orientation = A + \frac{(B - A)}{1 + e^{(torque - C)/D}} \quad (2)$$

where A represents an upper asymptote, B is a lower asymptote, C is the midpoint, and D is the slope at C. Simultaneous least-squares fitting of the parameters was accomplished via

the Newton-Gauss algorithm (Whittaker and Robinson 1967). To verify the parameterization (*i.e.*, no substantial difference in convergence incidence or solution obtained), an alternate algorithm (Marquardt-Levenberg) was invoked.

The logistic parameters (A,B,C,D in Figure 2) do not directly correspond to clinical parameters. However, the scale factor ('D' in equation 2) could be conceptualized as a metric of midrange flexibility (or compliance) of the glenohumeral joint. Two other outcomes (IE ROM and neutral zone width) were derived from the logistic parameters and represent range-of-motion and hysteresis of the glenohumeral joint, respectively. IE ROM was defined as the amount of rotation about the humeral axis relative to the glenoid that occurred between the upper and lower asymptotes that bound the limits of internal and external rotation (*i.e.*, A-B). Neutral zone width, a hysteresis measure defined as the absolute torque difference between the midpoint of the external loading rotation curve (C_{external}) and the midpoint of the corresponding internal loading rotation curve (C_{internal}), was also examined.

Variance components were of interest. Therefore, a nested, random-effects multivariate, analysis of variance (mANOVA) was used to explore the variability of the four logistic parameters. The individual, trial, and cycle from which the curve was taken were treated as random factors. Fixed factors were gender and side (left or right).

Although a single four-parameter logistic which fit the average of four repetitions was used universally for data analysis, this was compared to analysis of just the final cycle to determine whether or not it typified shoulder motion (*i.e.*, whether or not there was a preconditioning effect). It was likewise compared to averaging parameters that were fit to individual cycles.

RESULTS

The study cohort had a mean age of 24.7 (standard deviation (SD): 4.8 years, range: 19-36), and a Body Mass Index of 25.2 (SD: 4.4 kg/m², range: 19.2-34.7). All subjects were right-handed. Two of the twelve had generalized joint laxity (Beighton et al., 1973). Every shoulder tested scored 100 on the ASES scale, indicating no pain and full function, and had normal range of motion and strength (Table 1).

Results derived from the four logistic equation parameters are summarized in Table 2. Across all trials of all shoulders the mean IE ROM was 226.0° (SD: 34.4°). Variation in the defined IE ROM per shoulder over three trials on average was reproducible to within 9.6% (SD: 4.9%) of the total ROM measurement of 23.1° (SD: 11.2°), and this ranged from 2.1% to 22.8 %. The midrange flexibility (scale factor D) was reproducible to within 20.6% (SD: 11.3%) of the total midrange flexibility measurement (range: 0.5 to 3.8 °/Nm). Likewise, variation in neutral zone width was greater than IE ROM reproducibility, with trial-to-trial values corresponding to within 28.0% (SD: 18.7%) of the total neutral zone measurement.

In terms of fixed factors, IE ROM for all trials of male shoulders, with internal and external rotation values averaged, was 224.9° (SD: 33.0°), compared to 226.9° (SD: 23.2°) for females (p=0.87). Variability (*i.e.*, standard deviation) across three measurement trials for male shoulders was comparable to females (p=0.35). Neutral zone width did not differ to a statistically significant amount between males and females (p=0.18), nor did variability in neutral zone width (p=0.71). Midrange flexibility, however, was higher in males (p<0.01), as well as significantly more variable (p<0.01). Dominant (right) shoulders exhibited 222.5° (SD: 31.6°) of IE ROM, which was slightly less than non-dominant (left) shoulders (Average; 229.3° SD: 24.6°; p=0.09). Variability across three measurement trials was also similar for left and right shoulders (p=0.40). Neutral zone width was comparable from side-

to-side in its magnitude ($p=0.37$) and variability ($p=0.41$). Midrange flexibility results between sides were similar too ($p=0.15$ and $p=0.08$ for magnitude and variability respectively).

Relative to the other variance components, within trial (or cycle-to-cycle) variability of the logistic parameters was negligible. As expected, between trial variability was larger than within trial variability. Ignoring other factors besides cycle and trial, the standard deviation of the trial-to-trial variability was over 2.5 times that of the cycle-to-cycle variability within trial. The IE ROM parameter across four cycles was on average 14.1° (or 6.2%) for any single trial of a single shoulder. This average is skewed by outliers, as the standard deviation of the IE ROM across four trials ranged from 0.8 to 92.2° . This is a reflection of the fact that the exact shape of the curve, and therefore the repeatability of the results, was shoulder-joint dependent.

Abduction angle was allowed to fluctuate during the course of applied IE rotation. Increased abduction appeared to facilitate greater IE ROM. For each shoulder, the mean abduction angle was regressed with the IE ROM. The coefficient of determination (R^2) averaged 0.65 (SD: 0.36). While the strength of this relationship varied from $R^2 < 0.001$ to $R^2 > 0.999$, in 64% of shoulders, the coefficient of determination was greater than 0.75.

DISCUSSION

In a previous study in our laboratory, an attempt was made to evaluate torque-rotation response of the glenohumeral joint by immobilizing the shoulder girdle and measuring rotation of a motion tracker attached to the upper arm (Novotny et al., 2000). This approach produced IE ROM data over three trials that were, on average, reproducible to within 24.7% (SD: 19.6%) of the total IE ROM measurement. The high variability seen in that study motivated this improved approach. The lower value of 9.6% found in the current study underscores the importance of monitoring the position and orientation of the entire shoulder complex when measuring internal-external rotation biomechanics of the humerus relative to the scapula, rather than our prior approach (Novotny et al., 2000) that attempted to restrict scapulohumeral movement or assumed that motion at this interface is negligible. Improved repeatability also may arise from accounting for the (changing) orientation of the humerus relative to the load cell throughout the maneuver.

Furthermore, empirical torque cut-offs universally instituted in the previous study (internal and external torques of 4 Nm to bound ROM and 1 Nm to bound a neutral zone (Novotny et al., 2000)) may not always be appropriate. Applying these definitions to the present study, higher variability in IE ROM ensued, as some shoulders were not at or near their rotational limit by 4 Nm of applied torque. Using a fixed (2-Nm) neutral zone width likewise leads to disparities in what torque-rotation phenomena fall within the neutral zone. Mean neutral zone width in the current investigation was 1.40 Nm, and this width varied across shoulders by over an order of magnitude (0.3-3.4 Nm). Even if a more optimal torque cutoff could be chosen, data suggest limits be set separately for internal and external rotation, since there was evidence of asymmetry between rotation directions.

Clinical shoulder ROM measures are imprecise, and this may limit their prognostic value. External rotation of the shoulder from neutral, for example, has only fair inter-tester reliability and varies 11.7° when measured and re-measured by the same examiner with a hand-held goniometer (Bovens et al., 1990). Internal rotation, typically measured by spinal level, is marginally reproducible and suffers poor correlation to actual glenohumeral rotation measured in degrees (Ginn et al., 2006; Mallon et al., 1996; Wakabayashi et al., 2006). Moreover, flexibility cannot be measured or inferred with a hand-held goniometer. Before

applying a measurement technique to compare cohorts, it is important to develop a full understanding of its capabilities and limitations. Both the maximum attainable precision, and the precision necessary to discriminate long-term clinical successes from clinical failures, is uncertain. Hallström *et al.* (2006) demonstrated that the standard deviation associated with measurements of internal-external rotation performed twice (actively) by six patients was 4.2° , when measured with an accurate radiographic technique. This may represent a best-case scenario for shoulder kinematics.

There are many different schemes in the literature for measuring kinematic variability (mean ensemble curves, values derived from curve parameterization, etc.). Essentially, what is evident from inspection of all sets of curves from all shoulders in the present study is that shape of the torque-rotation curve is shoulder dependent. This implies that there may not be a single mathematical model which is ideal for all data. Mathematically, shape can be thought of as the property of a closed curve that remains after removing dilations, translations, and rotations (Wentz 2000; Zhao and Stough 2005). Properties of the area distribution (Zernike or Legendre moments, for example) may be more discriminating than logistic fits. However, the significance of statistical differences in one or more of the area moments is not readily apparent. The four-parameter logistic represents a reasonable model for the rotation-versus-torque behavior, but small sample size makes it very difficult to choose any one model unequivocally between several competing model fits (Ratkowsky 1983).

The four-parameter logistic is based on a dependent-independent variable relationship that is bounded by two asymptotes. Internal-external ROM was higher than expected. Fundamentally, the logistic paradigm may tend to overestimate ROM; it can be argued that extrapolation is not applicable because there are (abrupt) anatomic limits on maximum attainable ROM. The IE ROM measured using the internal loading and IE ROM measured using external loading were closely related ($R^2 = 0.75$), with a slope that was very close to unity (Figure 3), and not significantly different ($p=0.76$). Midrange flexibility (scale factor 'D') from internal loading was also correlated to external loading ($R^2 = 0.64$, slope = 0.97), and not significantly different ($p=0.64$).

The four-parameter logistic model converged at plausible values for 95.1% of the data. Approximately 43% of the non-convergent data were from a single subject who had been identified as having generalized ligamentous laxity. Again, this is consistent with shape of the torque-rotation curve being shoulder-dependent. Lax joints may accommodate extra translation of the humerus on the glenoid, which in turn may allow the joint to keep seeking an optimal position for continued rotation. Such a phenomenon could produce a torque-rotation response without marked asymptotes. On the other hand, the other subject with generalized joint laxity did not display anomalous torque-rotation data. Furthermore, Jansson *et al.* (2004) found that generalized ligamentous laxity was not necessarily related to rotational shoulder joint laxity.

McCully *et al.* (2005) concluded that “significant amounts of scapulothoracic motion may impact measurements of isolated glenohumeral joint motion.” Although rotation about the long axis of the humerus relative to the glenoid was measured explicitly, some of the internal-external rotation measured was likely facilitated by scapulothoracic sliding (Bagg and Forrest 1988; Ebaugh *et al.*, 2005; Youdas *et al.*, 1994), which did occur. The propensity for bony impingement between the glenoid and the humerus, as well as the degree of ligamentous constraint, clearly depends upon scapulothoracic position. While the relationship between glenohumeral and scapulothoracic rotation has been widely studied during active humeral elevation (Inman *et al.*, 1944), few studies can be found in the literature looking at this relationship during glenohumeral IE rotation. Finley *et al.* (2003)

studied the effect of sitting posture on scapular kinematics as measured with a cutaneous electromagnetic tracking system. In addition to posture, fatigue, resistance level, passive positioning, and muscle strength all affect scapular kinematics (Tsai et al., 2003). In the present repeatability study, rotations of the clavicle were not explicitly tracked, as this would be problematic with a skin-based system, but a method (an image-based technique, for example) allowing sternoclavicular and acromioclavicular monitoring would enhance our understanding of how the different components of the shoulder girdle work in concert.

From the collected ASES data, external rotation measured at 0° abduction was not correlated with external rotation measured at 90° abduction for either active ($R^2=0.14$, $p=0.07$) or passive ($R^2=0.003$, $p=0.79$) motion. Likewise, Rundquist *et al.* (2004) found that idiopathic internal shoulder rotation deficits were more evident in abduction than with the arm at the side. Not only does the degree of internal-external rotation vary with shoulder abduction, but the reliability of external rotation also depends on abduction, with external rotation gauged (via goniometry) at 90° abduction producing better reliability than 0° abduction (Riddle et al., 1987). This has potential implications for our reproducibility results because during our tests, abduction position was allowed to vary from test-to-test and individual-to-individual, provided that there was adequate clearance during internal rotation. Clearly, ligamentous constraint changes with abduction angle (Ernlund and Warner 2000; Ferrari 1990; Harryman et al., 1990; O'Brien et al., 1995). In the majority of shoulders evaluated presently, there was a strong relationship between glenohumeral IE ROM and abduction. Other researchers (Hallström and Kärrholm 2006) have reported coupling between external humeral rotation and abduction of the arm during active motions. Because the IE rotation in our protocol was large relative to shoulder flexion-extension and ab/adduction, the latter two rotations may be subject to kinematic crosstalk, depending on the coordinate system orientation.

Several potential sources of variability exist in the present study. The electromagnetic motion tracking system has finite limits; each sensor is accurate to within 1.8 mm and 0.5°, but working in conjugate, actual accuracy is highly dependent on the proximity of the sensors to each other, to the transmitter, to ferromagnetic and paramagnetic objects, and to electromagnetic fields that are generated by all electrical equipment. The use of interfering objects and fields was minimized in the study. Inevitably, reproducibility of sensor placement and precise location of anatomic landmarks may differ between trials. With skin-based electromagnetic motion tracking devices, the assumption is made that motion measured at the skin is an acceptable surrogate for the actual rigid body motions of underlying bone. Skin- and bone-based shoulder rotations measured with electromagnetic sensors corresponded closely in one cadaveric study (LaScalza et al., 2002), and scapular movements exhibit similar patterns (albeit not necessarily similar numerical values) between skin-based and bone-based tracking modalities *in vivo* (Karduna et al., 2001). This has not been established for the humerus.

In any study involving repeated trials, the learning effect (in this case, the learned behavior is how to relax) may improve performance, while demotivation/apprehension may deteriorate performance. Both of these factors are likely involved in the task of allowing an examiner to rotate a study subject's shoulder to its limits. Nonetheless, this passive motion approach was chosen over active motion because it is not known whether or not voluntary joint motion should be perfectly repeatable from cycle-to-cycle or day-to-day (Mitra et al., 1997). The rate of moment application in our testing was regulated by the examiners' discretion, and this may also be a source of variability; with active shoulder ROM, a correlation is reported between peak ROM and movement velocity (Rundquist and Ludewig 2004). We compared analysis performed on the average of all 4 cycles to analysis performed on the 4th cycle only and found no difference in all outcomes, and this led us to conclude that there was not a preconditioning effect.

Flexibility is the inverse of stiffness, and is here defined as the range of angular displacement when the passive/active structures are stretched over a prescribed torque range. Because endpoint stiffness decreases with increased trunk and shoulder mobility (Perreault et al., 2000), higher endpoint flexibility is expected in this study compared to those employing more constraint. We found that flexibility changes as a function of applied torque throughout the IE ROM. Midrange flexibility (D) was more variable across trials than IE ROM. Musculoskeletal flexibility is influenced by a number of factors, including viscoelastic stress relaxation response, stretch tolerance, and electromyographic activity (Magnusson et al., 1997). Variance in the midrange flexibility also suggests varying ability to relax during passive manipulation of the shoulder and/or differing degrees of muscle bulk about the shoulder joint. Studies under anesthesia with a neuromuscular blocking agent may elucidate whether or not electromyographic activity plays a key role in midrange measurement repeatability.

Reproducibility of IE ROM did not appear to depend on sex or limb dominance. Previous studies of the shoulder (Barnes et al., 2001) and of other joints have shown that females average greater ROM than males. Our study demonstrated trivial IE ROM (2°) differences between males and females, which did not reach statistical significance. Sex differences in muscle guarding have been found in the knee (Wojtys et al., 2003); perhaps a related phenomenon in the shoulder is responsible for the observed midrange flexibility difference that we observed between males and females.

CONCLUSION

Precise, non-invasive measurement of shoulder biomechanics over time is a requisite step towards optimizing surgical and rehabilitative treatment of shoulder injury and disease. Our current methods of measuring IE ROM and neutral zone width (hysteresis) of the glenohumeral joint are superior to other previous attempts at trying to non-invasively evaluate the biomechanics of this joint *in vivo*. This approach also includes the capability to describe midrange flexibility of the glenohumeral joint; however, the reproducibility this measurement was inferior to that of IE ROM. The ability of the current methodology to produce consistent torque-rotation data across trials varies. A larger dataset would be necessary to attempt covariate-based prediction of exactly which shoulders can be measured reliably over time. Due to the large residual error that could not be directly attributed to known independent variables, before using this device to make clinical comparisons, repeated shoulder measurement trials should be collected at the initial visit, and any shoulders exhibiting large variation in ROM over trials should be excluded from further study. Finally, the precision of a measurement is not a reflection of its accuracy. A requisite next step for future studies will involve comparison of results obtained using this method to those obtained simultaneously using uniplanar radiostereometric analysis *in vivo* to track three-dimensional bony movements.

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Figure 1.

Subjects were seated in the “beach chair position” (reclined backwards 10°), and their back and shoulders were supported with a vacuum pack (Olympic Medical, Seattle, WA) and a strap which crossed their torso diagonally. Electromagnetic position and orientation sensors (MiniBIRD 8000, Ascension Technology Corp., Milton, VT, USA) were secured to the subjects' skin via medical adhesive tape and Bensoin prep. The sensors were positioned at the level of the first thoracic vertebra (T1), the posterolateral aspect of the acromion, and the distal humerus. The arm was abducted (55-60 degrees) and secured in a custom-molded thermoplastic brace with the elbow bent to 90°. The brace interfaced with a six degree-of-freedom load cell that was mounted rigidly to a rotation platform with a hinged gear joint to accommodate naturally occurring changes in abduction-adduction throughout the internal-external range-of-motion (IE ROM)). A moment was applied to induce maximum internal rotation of the humerus, and then reversed to induce maximum external rotation.

Figure 2. Schematic of biomechanical parameters derived from the torque rotation curve: IE ROM (A-B), midrange flexibility (D), and neutral zone width ($|C_{\text{internal}} - C_{\text{external}}|$, where midpoint C is indicated with *).

Figure 3.

Close correspondence existed between IE ROM measured from the internal loading/external unloading curve and the external loading/internal unloading curve.

Table 1

Range of motion of study subjects, measured according to ASES guidelines.

	Active Average (SD)	Passive Average (SD)
Forward Flexion	161° (5°)	170° (6°)
External rotation (0° abduction)	75° (16°)	85° (14°)
External rotation (90° abduction)	103° (13°)	119° (13°)
Internal rotation (median spinal level)	T5	Between T4 and T5
Cross body abduction	20.6 cm (2.7 cm)	16.5 cm (2.8 cm)

Internal-External range of motion (IE ROM), midrange flexibility, and neutral zone width are given for all trials of all shoulders, along with demographic information. The tabulated I-E ROM and flexibility both represent averages of the data obtained in the internal loading direction and the data obtained in the external loading direction. (N/A indicates that there was no plausible convergence for the four-parameter logistic).

Table 2

AGE (years)	SEX	TRIAL	Left side			Right side		
			IE ROM (degrees)	Flexibility (degrees/Nm)	NZ width (Nm)	IE ROM (degrees)	Flexibility (degrees/Nm)	NZ width (Nm)
29	M	1	225.1	1.40	0.98	208.1	1.07	1.08
		2	201.6	2.07	1.14	183.6	1.05	1.47
		3	208.7	2.69	0.82	167.8	0.86	1.34
20	F	1	222.5	1.05	0.99	201.6	1.29	1.04
		2	243.2	1.08	1.10	224.7	1.32	1.30
		3	252.2	1.00	0.67	210.8	1.05	1.41
21	M	1	207.5	1.42	1.18	164.4	0.83	0.86
		2	174.1	1.00	0.73	156.2	1.24	1.41
		3	211.9	1.25	0.65	218.1	1.81	1.28
24	F	1	204.8	0.83	1.69	186.9	1.65	2.59
		2	249.0	0.82	N/A	207.1	1.90	2.04
		3	246.5	1.24	3.09	195.2	1.25	2.06
19	F	1	193.2	1.24	1.04	N/A	N/A	N/A
		2	161.6	0.99	N/A	193.6	1.32	2.03
		3	206.7	1.56	0.28	232.0	1.01	1.89
19	F	1	245.3	0.85	0.66	241.4	0.98	1.59
		2	276.0	1.11	0.38	218.8	0.99	0.72
		3	N/A	N/A	N/A	256.7	1.12	1.56
23	M	1	285.5	1.65	2.31	185.9	1.61	2.04
		2	285.8	1.34	1.13	260.7	1.77	1.66
		3	260.1	1.12	1.11	208.0	1.23	1.32
26	M	1	198.3	0.90	0.47	243.3	1.36	1.06
		2	251.2	1.95	1.99	268.3	1.31	1.18
		3	228.7	1.48	1.18	296.6	1.34	0.84

AGE (years)	SEX	TRIAL	Left side			Right side		
			IE ROM (degrees)	Flexibility (degrees/Nm)	NZ width (Nm)	IE ROM (degrees)	Flexibility (degrees/Nm)	NZ width (Nm)
24	M	1	263.4	2.00	1.59	199.9	1.05	2.35
		2	286.2	2.17	2.36	219.8	1.86	N/A
		3	200.0	1.26	1.35	181.2	1.49	2.21
36	M	1	222.8	3.54	2.53	259.9	2.70	3.17
		2	212.2	2.78	2.22	272.3	2.30	3.41
		3	238.3	3.42	1.63	253.4	2.53	3.40
27	F	1	220.1	0.92	1.07	248.5	0.94	1.37
		2	230.7	0.97	1.04	257.8	1.00	1.35
		3	228.0	0.83	0.62	302.5	1.08	1.53
28	F	1	243.9	1.19	0.71	213.5	0.70	0.38
		2	196.1	0.82	0.41	215.0	0.55	0.31
		3	208.6	0.82	0.54	254.4	0.80	0.61