Effects of Upper Trunk Rotation on Shoulder Joint Torque Among Baseball Pitchers of Various Levels

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High rotational torques during baseball pitching are believed to be linked to most overuse injuries at the shoulder. This study investigated the effects of trunk rotation on shoulder rotational torques during pitching. A total of 38 pitchers from the professional, college, high school, and youth ranks were recruited for motion analysis. Professional pitchers demonstrated the least amount of rotational torque (p = .001) among skeletally mature players, while exhibiting the ability to rotate their trunks significantly later in the pitching cycle, as compared to other groups (p = .01). It was concluded that the timing of their rotation was optimized as to allow the throwing shoulder to move with decreased joint loading by conserving the momentum generated by the trunk. These results suggest that a specific pattern in throwing can be utilized to increase the efficiency of the pitch, which would allow a player to improve performance with decreased risk of overuse injury.

Overuse injuries to the shoulder and elbow joints remain the most common medical problem in all levels of baseball (McFarland and Wasik, 1998; Conte et al., 2001). Although a number of factors influence injury etiology, joint injuries resulting from repetitive microtrauma have been presumably linked to excessive joint loads commonly associated with overarm throwing (Albright et al., 1978). The shoulder joint is particularly vulnerable to these loads and consequently draws significant clinical attention in throwers. The high prevalence of pain

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and other signs and symptoms of shoulder injuries in baseball pitchers can be attributed in part to the physiological effects of these joint loads brought about from excessive rotational torques associated with throwing (Fleisig et al., 1995; Lyman et al., 2001; Werner et al., 2001; Hutchinson and Ireland, 2003). Although pitching involves a certain amount of rotational torque to produce effective throws, it is believed that high torques generate loads that exceed the elastic tensile strengths of the surrounding joint soft tissues (Radin et al., 1978; Fleisig et al., 1995). Thus, determining the specific biomechanical patterns that place the shoulder at higher risk could lead to coaching or training methods designed to correct inefficient pitching mechanics that produce high rotational torques.

Efficient throwing mechanics is predicated on a pitcher's ability to perform a sequence of movements in body segments, which progresses from the legs, pelvis, and trunk to the smaller, distal arm segments. The momentum generated by the larger segments is transferred to the adjacent distal segments by appropriately timing the movement of the pelvis and trunk in a manner that ideally follows the summation of speed principle in which a segment initiates its movement when the adjacent proximal segment reaches its maximum angular velocity (Putnam, 1993). The trunk segment, in particular, has been shown to contribute considerably to the total body angular momentum, as demonstrated in such activities as the high jump, tennis serve, and kicking (Dapena, 1978; Putnam, 1991, 1993; Bahamonde, 2000). Improper timing in sequential body motion can result in a loss of angular momentum transferred from the larger segments to the

throwing arm. As a result, pitchers tend to generate more internal torque to compensate for this dissipation (Fleisig et al., 1996; Fleisig et al., 2000). This principle is at times described by coaches as "throwing with too much arm," in which the energy from trunk rotation is transferred to the upper arm too early and dissipated instead of being applied to the hand and ball. This process may lead to an increase in potentially harmful torques generated at the shoulder joint (Fleisig et al., 1996). However, there currently is a paucity of quantitative data that substantiates this relationship for baseball pitching. Therefore, the purpose of this study was to compare the biomechanical patterns of trunk rotation and shoulder joint torque during baseball pitching between professional, collegiate, high school, and youth players.

METHODS

Subject Preparation

A total of 38 baseball pitchers participated in this study after providing written informed consents approved by the hospital's institutional review board. The throwing motions of 6 professional (age = 24 ± 2, height = 190.2 ± 9.6 cm, mass = 100.6 ± 12.3 kg), 11 collegiate (age = 19 ± 2, height = 192.6 ± 6 cm, mass = 94.8 ± 7.6 kg), 12 high school (age = 16 ± 2, height = 175.5 ± 30.6 cm, mass = 82.7 ± 15.4 kg), and 9 youth pitchers (age = 12 ± 1, height = 155.2 ± 8 cm, mass = 50.9 ± 12.5 kg) were data-captured and statistically analyzed using a between-group comparative design. All pitchers were considered relatively healthy with no significant bodily injury at the time of testing.

For each subject, a set of 2.5-cm spherical reflective markers was placed on the skin overlying 34 anatomical landmarks secured with adhesive tape. An upper body marker set (Figure 1) was combined with the Helen Hayes lower body marker set (Kadaba et al., 1990) to create a full body marker set used to bilaterally define the hip, knee, ankle, shoulder, elbow, and wrist joints as well as the upper and lower limb segments.

Setup and Protocol

Motion capture was conducted using eight visible-red cameras interfaced with the Real-Time Motion
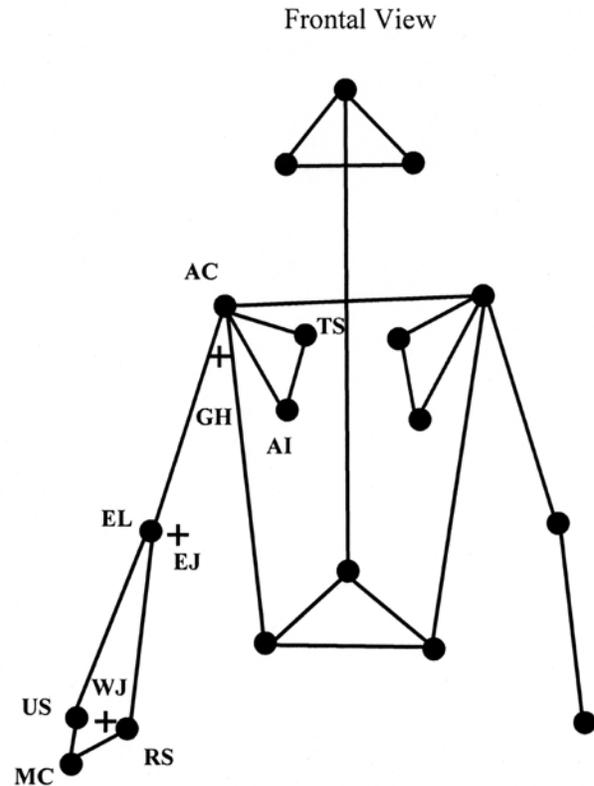

Frontal View

Figure 1 — Marker locations used with the upper body kinematic model. Real markers (•) were placed on the skin overlying the acromion process (AC), spine of the scapula (TS), inferior angle (AI), lateral epicondyle (EL), ulnar styloid (US), radial styloid (RS), and the base of the fifth metacarpal (MC). Estimated joint centers were defined using virtual markers (+) for the glenohumeral (GH), elbow (EJ), and wrist (WJ) joints.

Capture System (Motion Analysis Corp., Santa Rosa, CA). Strobe control was set at 200 Hz, and camera rate was fixed at 120 Hz for full pixel resolution. Seed and wand calibration procedures were employed on each day of data collection. The average 3-D residual error for the motion capture system was 1.2 ± 0.6 mm, which is the degree of accuracy in which the system can reconstruct the location of each marker in the capture volume. Marker tracks were processed using marker identification techniques and digital signal processing that incorporated a Butterworth filter at a cut-off frequency of 18 Hz. The marker-based optical system was housed in a 130-m^2 motion analysis laboratory with cameras specifically positioned to allow for a 5 m (L) × 2 m (W) × 4 m (H) calibrated volume of space, ideal for capturing throwing motion off a 2.7 m (L) × 2.5 m (W) × 0.3 m (H) indoor mound (ProMounds, Inc., Winthrop, MA).

After performing a preparation routine for stretching and warm-up throwing, each pitcher threw up to 15 fastball pitches off the indoor mound to a simulated strike zone area at a regulation distance of 18.4 m (16 m for youth pitchers) from the pitching rubber. Pitchers or their coaches were asked to rate each pitch on a scale from 1 to 5, with 1 being the *worst* and 5 being the *best* using a subjective criteria of ball location and body posture at release. The highest rated pitch that hit the strike zone with reliable marker data was ultimately selected for analysis. Since most throwers deliver a pitch with consistent mechanics (Pappas et al., 1985; Fleisig et al., 1999), only one pitch per subject was analyzed, in agreement with a number of previous investigations that employed similar single trial analysis (Feltner & Dapena, 1986; Werner et al., 1993).

Lower body segments of the pelvis, upper leg, lower leg, and feet were defined using a widely accepted method based on the Helen Hayes marker set (Kadaba et al., 1990). In addition, local coordinate systems were defined for the trunk, upper arm, forearm, and hand segments (Figure 2) and we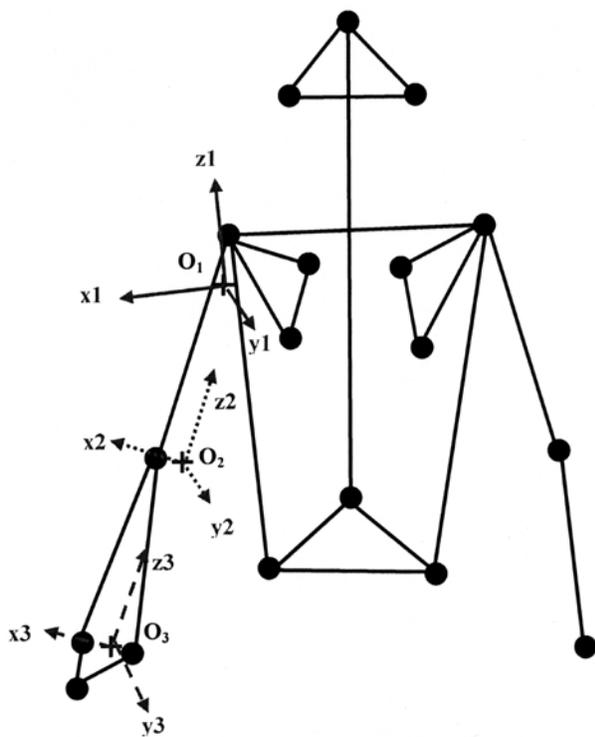re used to calculate 3-D rotations at the shoulder and elbow joints. The specific upper extremity kinematic model including joint center definitions, coordinate systems, joint coordinate system, and angular conventions are beyond the scope of this discussion and are described in detail in the Appendix. Pelvic kinematic data was calculated relative to the fixed coordinate system of the laboratory, allowing for trunk rotations to be calculated relative to the pelvis using the thoracic coordinate system. The clinical convention of forward flexion, lateral bending, and rotation were computed as a sequence of rotations in the sagittal, frontal, and transverse planes, respectively, using joint coordinate system calculations that are analogous to the typical Euler angle sequence used for trunk kinematics (Kadaba et al., 1990; Crawford et al., 1996). The transverse plane rotation of the trunk relative to the pelvis provided what some baseball coaches refer to as "hip-shoulder separation" and was considered neutral when this degree of separation was zero.

Kinetics

Joint torques of the throwing shoulder were calculated using the inverse dynamics technique described by Feltner and Dapena (1989). For the purpose of this study, only internal-external rotational torque, which is the joint moment about the long axis of the humerus on the throwing side, was analyzed for comparison. Inertial properties of the throwing arm segments used for this analysis were estimated using anthropometric ratios previously measured from adult male cadavers (Clauser et al., 1969). The mass of a baseball 23 cm in circumference was 0.14·kg. Rotational torques were expressed in absolute units (newton meters) and in terms normalized by body weight (BW) and height (H) (BW·H).

Data Analysis

Specific points in trunk rotation and shoulder kinetics were extracted as a function of the normalized pitching cycle, defined from front-foot contact (0%) to ball release (100%). Trunk rotation peak magnitudes and onset times were recorded in addition to the peak internal rotation torque at the throwing side shoulder. Peak trunk rotation was measured at the point in the pitching cycle when the difference in transverse plane rotation between the pelvis and torso reached its maximum magnitude. Data were compared across all competitive groups

Figure 2 — Segmental coordinate systems for the upper body kinematic body using estimated joint centers GH, EJ, and WJ as local origins O_1, O_2, and O_3, respectively.

(independent variable) using a four-level completely randomized ANOVA with a post hoc Bonferroni test whenever significant ($\alpha = .05$) differences were found in the aforementioned dependent variables.

RESULTS

No significant kinematic differences in trunk movement were found across all four groups of pitchers except for the starting time of trunk rotation, defined as the onset time of peak trunk rotation (Table 1). Professional pitchers rotated their torsos toward home plate much later in the pitch cycle ($34 \pm 5\%$, $p = .01$) compared with lower level pitchers, who rotated at an average below 14% into the pitching cycle. Eleven pitchers from the high school and youth levels rotated their trunk before front-foot contact (0% PC). Despite these temporal differences, the magnitude of peak trunk rotation ($52 \pm 8°$, $p = .48$) did not differ significantly across all levels (Table 1).

Kinetic analysis revealed differences in rotational torque at the throwing shoulder. Onset time for peak shoulder internal rotation torque occurred when the throwing shoulder reached maximum external rotation in the late cocking phase. The absolute magnitudes of internal rotation torque differed significantly among all four pitching levels (Figure 3), with youth pitchers exhibiting the least amount of internal rotation torque (33 ± 3 N·m) as

Table 1 Trunk Rotation (Mean ± *SD*) Magnitudes and Onset Times for All Pitching Levels

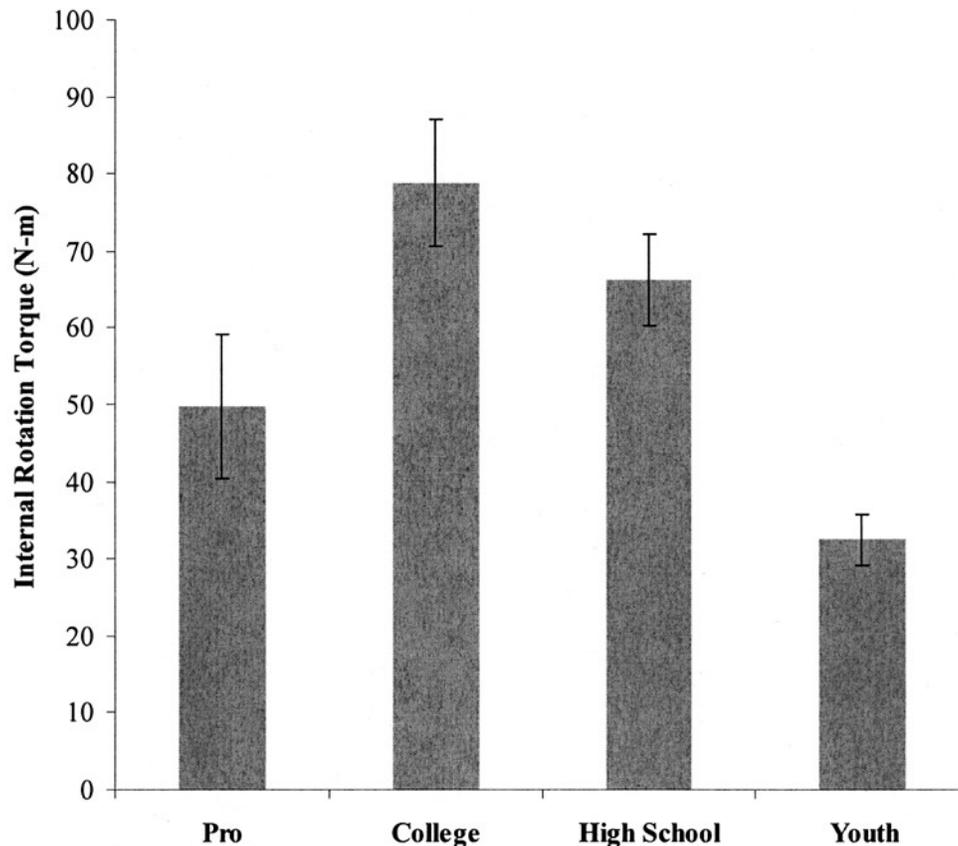| Competitive level	Rotation (deg)*	Onset time (% PC)†
Youth	50.1 ± 6.9	5.0 ± 0.7
High school	54.9 ± 9.1	6.4 ± 1.3
College	51.5 ± 8.2	14.2 ± 1.5
Professional	49.9 ± 5.2	34.3 ± 5

*$p = .48$, †$p = .01$.

Figure 3 — Peak shoulder internal rotation torque (N·m) across all four groups. Professional players exhibited the least rotation torque (50 ± 9 N·m, p = .04) among skeletally mature pitchers.

compared with high school (66 ± 6 N·m, $p < .001$) and collegiate (78 ± 9 N·m, $p < .001$) pitchers. Professional level pitchers threw with less internal rotation torque (50 ± 9 N·m) than collegiate pitchers (78 ± 9 N·m, $p = .04$). When analyzing rotation torques normalized by body weight and height (Figure 4), professional pitchers exhibited significantly less normalized peak torque (25 ± 3% BW·H) than pitchers from the college (43 ± 5% BW·H, $p = .01$), high school (49 ± 5% BW·H, $p = .003$), and youth (40 ± 3% BW·H, $p = .02$) levels. No significant differences in normalized torque were found among all three amateur groups.

DISCUSSION

It is generally accepted that baseball pitching produces high stresses at the shoulder that could lead to joint injury (Albright et al., 1978; Fleisig et al., 1995; McFarland & Wasik, 1998). The rotational torques at the throwing shoulder measured in this study were lower in magnitude than those measured previously, but their clinical relevance still warrants significant attention, because rotational torques are most likely the underlying cause of both soft tissue and bony injuries (Feltner & Dapena, 1986; Fleisig et al., 1995; Fleisig et al., 1999; Sabick et al., 2004). It has been suggested that such joint loads are affected by the manner in which the proximal body segments move during the delivery (Putnam, 1993; Fleisig et al., 1996); however, this relationship has yet to be quantitatively analyzed.

Baseball pitching, like other overhand sports activities, requires the employment of momentum sequentially transferred from the larger body segments to the smaller distal segments, all of which collectively contribute to the overall force and velocity output of the throw (Putnam, 1991, 1993; Bahamonde, 2000). This transfer of momentum would ultimately require less contribution from the smaller distal segments in generating relatively high throwing velocities. Since the torso represents more
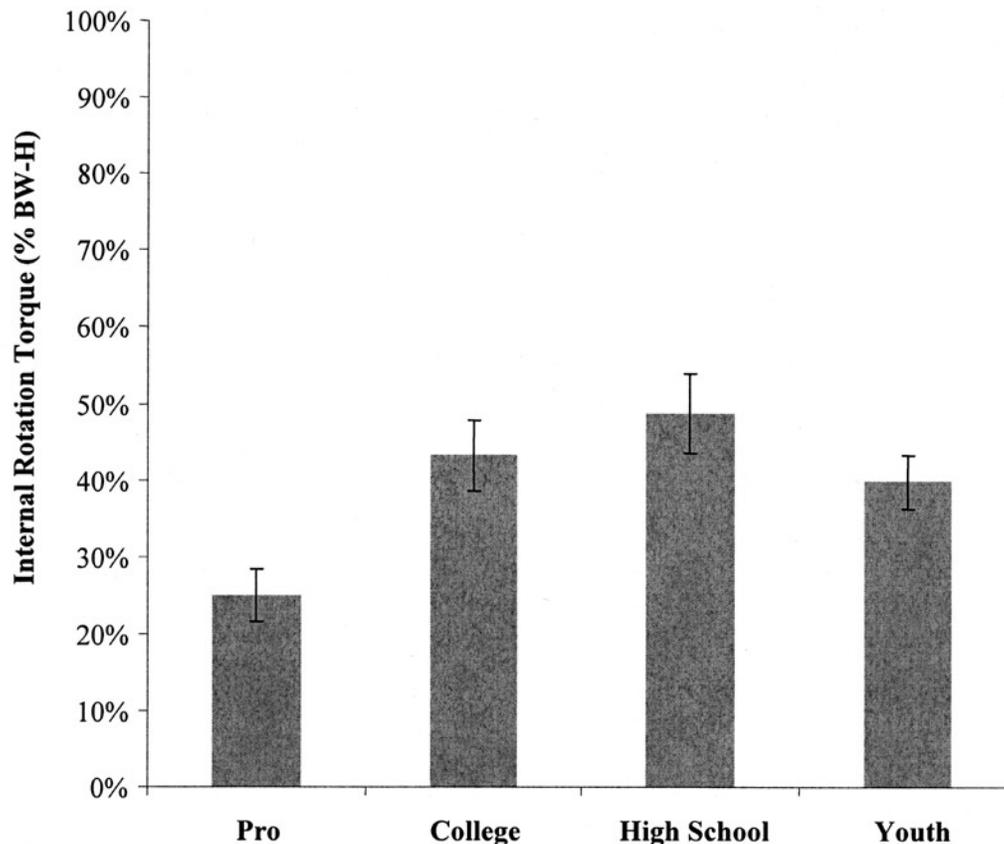

Figure 4 — Peak shoulder internal rotation torque normalized to body weight and height (%BW·H) across all four groups. Professional players exhibited the least rotation torque (25 ± 3% BW·H, p = .001) among skeletally mature pitchers.

than half of the total body mass and consequently generates a significant amount of angular momentum, appropriate timing of trunk rotation during the throwing motion is necessary to maximize this momentum transfer (Joris et al., 1985). It has been presumed that even slight changes in timing could result in reduced output and potentially harmful joint loads at the throwing shoulder (Fleisig et al., 1996). However, little quantified data exists in the current literature to support this presumption. The current study demonstrated that pitchers who rotated their torsos later in their delivery exhibited less shoulder internal rotation torque. Although direct relationships in kinetic data produced with inverse dynamics are speculative at best, these findings suggest that pitchers may generate the necessary momentum to throw a ball at acceptable velocities without overloading the shoulder joint by sequentially utilizing the larger body segments (Joris et al., 1985; Fleisig et al., 1996). It appeared that the sequence of body segmental motion was compromised when the trunk rotated too early, because part of the rotational energy was lost and made up for by the upper extremity, but further research on angular momentum is needed to substantiate this pattern.

The idea of the torso providing the highest segmental contribution to overhand movements has been studied in other sports. Bahamonde (2000) reported the trunk generated the most amount of angular momentum in the sagittal plane in the performance of a tennis serve. Although the pitching motion, unlike the tennis serve, largely consists of movements in the transverse plane, the contribution of the trunk to the total body momentum is significant in the primary plane of movement. In kicking, the linear motion of the hip is primarily facilitated by rotation of the trunk about its longitudinal axis (Putnam, 1993). This transfer of momentum in the transverse plane is similar to the mechanism observed during pitching as the distal segments of the arm rotate as a result of trunk movement. Thus, in order to conserve angular momentum, the rotational angular velocity about the long axis of the humerus is nearly five times that of the trunk (Putnam, 1993; Fleisig et al., 1996). The probability of inducing resultant loads, higher than the elastic strengths of the shoulder structures constraining the joint, increases as upper extremity muscular contributions augment the joint angular velocity in an attempt to compensate for an early onset of trunk rotation, as seen in the less experienced players

captured in this study. The same mechanism may be employed in fatigued players in their attempt to maintain proper mechanics over the course of a game as demonstrated in the changes in joint position and demand in shoulder muscles during the late inning pitches (Murray et al., 2001; Mullaney et al., 2005). It is unclear, however, whether these changes are direct results of fatigue itself or compensatory mechanisms to minimize injury.

The sequence of proximal to distal segmental motion can be used to explain the magnitude and onset of rotational torques at the shoulder joint during a pitch. Following pelvic rotation, the trunk rotates toward home plate. As the throwing shoulder abducts, this trunk rotation would cause the glenohumeral joint to externally rotate to a position outside the normal active range of motion of the joint (Sabick et al., 2004). This position is often referred to as a "lag" because the motion of the throwing arm increasingly lags behind trunk rotation when it initiates earlier in the pitching cycle (Putnam, 1993). In this study, the shoulder internal rotation torque, which counters this movement, peaked at maximum external rotation. Hence, early trunk rotation most likely caused a shoulder lag that led to an increase in internal rotation torque observed in the less experienced throwers. Based on our findings, it would seem that potentially detrimental rotational torques could be minimized if trunk rotation was delayed so as to allow the shoulder joint to "catch up" with segmental body motion. The exact point or range within the pitching cycle for optimum trunk rotation has yet to be determined and should be addressed in future investigations.

Technical differences even between similar motion analysis systems may have accounted for the lower internal rotation torques found in this study as compared to those reported previously. In addition, markers placed on the skin of the throwing arm could have accounted for errors in the kinematic calculations despite efforts to minimize them, such as placing markers on bony prominences with the least amount of skin motion. Gordon and Dapena (2006) reported that upper arm kinematic measurements were unreliable owing to movement artifacts just prior to ball impact during a tennis serve. However, because rotational torques measured in this study were extracted at or near maximum external rotation of the throwing arm, the values estimated were reported with an acceptable degree of error.

In summary, this study represents the first known attempt to quantitatively compare the shoulder joint torques of baseball pitchers of various levels as a function of trunk rotation. The findings indicated that pitchers who exhibited trunk rotation later in the pitching cycle also demonstrated less rotational torques but its direct cause-effect relationship has yet to be determined. Thus, a possible mechanism for minimizing throwing-related overuse injuries has been presented, but further research is needed to clearly define how trunk rotation can be modified to reduce such pathomechanics. Furthermore, estimations in segmental linear and angular momentum as well as the effects of fatigue during pitching should be addressed in subsequent analyses.

Acknowledgments

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APPENDIX

Kinematic Model Definitions

To measure the dynamic range of motion of the shoulder joint during pitching, a sequential kinematic model based on the joint coordinate system (JCS) was employed. Methods previously used to measure shoulder motion are primarily based on projection angles (Feltner & Dapena, 1986; Dillman et al., 1993), which are susceptible to errors as a result of out-of-plane motion (Soutas-Little et al., 1987). Other methods used in clinical applications include variations of a standardized protocol based on Euler angles (Schmidt et al., 1999; van Der Helm, 2002; Rab et al., 2002). However, in an effort to follow a set of standardized descriptions of joint kinematics that are rigorous, nonconflicting, and relevant across various disciplines, we incorporated a variation of the method proposed by the Standardization and Terminology Committee (STC) of the International Society of Biomechanics (ISB), which defined a set of coordinate systems for various joints of the lower body (Wu et al., 2002) as well as for the upper body (Wu et al., 2005) based on the JCS (Cole et al., 1993; Wu and Cavanagh, 1995).

The shoulder, elbow, and wrist joint centers were defined relative to the reflective markers using anthropometric offsets provided by the regression analysis of Veeger et al. (1997) and were identified as virtual markers in the EVa Real Time software (Motion Analysis Corporation, Santa Rosa, CA) (Appendix Table). The glenohumeral (GH) joint was estimated relative to three markers on the shoulder girdle. Despite the fact that scapula motion is compromised by surface motion, the shoulder girdle triad has been shown to be more "rigid" than trunk markers (i.e., L AC, R AC, xyphoid) within elevation range of motion less than 120° (Karduna et al., 2001). The wrist joint center was located midway between the radial (RS) and ulnar styloid (US) markers. Elbow joint was referenced to the plane defined by the GH joint, EL (lateral epicondyle), and the wrist joint center, with the assumptions that elbow motion was primarily uniplanar (flexion–extension) and forearm pronation–supination occurred about the long axis perpendicular to the wrist joint axis. The flexion–extension axis of the elbow was approximated as long as the three points were not collinear as typically observed in a large range of elbow motion (Wang et al., 1998). This elbow joint estimation was adopted over a convention defining the joint axis through an intermediate coordinate

Appendix Table   Joint Center Definitions (Right Side) Utilized in the Upper Body Model With Offsets Determined by Regression Analysis (Veeger et al., 1997)

Joint center	Real markers (Figure 1)	*x*	*y*	*z*
Shoulder (GH)	AC, TS, AI	35%	7%	−6.7%
Wrist (WJ)	US, RS	0%	50%	0%
Elbow (EJ)	GH, EL, WJ	100%	0%	15%

Note. *y*- and *z*-axis offsets have opposite signs for the left shoulder and elbow joint centers. AC: acromion process, TS: scapula spine, AI: inferior angle, US: ulnar styloid, RS: radial styloid, and GH: glenohumeral joint.

system because marker placement on the medial epicondyle, or on any point on the upper arm, was inappropriate owing to high surface motion artifacts during pitching. As wand markers are typically used with such marker sets, the coordinate data would have been considered unusable as a result of irreparable noise contamination.

Specific segmental axes, joint coordinate systems, and angular conventions used for the kinematic measurements of the study were defined for the upper extremity. The trunk was represented as two rigid segments, representing the left and right shoulder girdles, contrary to one trunk segment described in most common conventions adopted for shoulder joint rotations (Veeger et al., 1997; Rab et al., 2002). Glenohumeral rotation was estimated relative to the ipsilateral "half" of the trunk. This segmental coordinate system was defined using markers on the inferior angle (AI), spine (TS), and acromion process (AC) of the scapula (Figure 2):

O_1	Origin (GH)
$\mathbf{x}_1 = (GH - TS)/\lvert GH - TS \rvert$	Line directed laterally from TS to GH
$\mathbf{y}_1 = \mathbf{x}_1 \times \mathbf{p}_1$	Line directed anteriorly perpendicular to the plane formed by $\mathbf{x}_1$ and $\mathbf{p}_1$
$\mathbf{z}_1 = \mathbf{x}_1 \times \mathbf{y}_1$	Line directed in the superior direction perpendicular to the plane formed by $\mathbf{x}_1$ and $\mathbf{y}_1$

where $\mathbf{p}_1 = (AI - TS)/\lvert AI - TS \rvert$.

The upper arm reference frame was identified by defining the humeral long axis first followed by the anterior and lateral axes (Figurer 2):

O_2	Origin (EJ)
$\mathbf{z}_2 = (GH - EJ)/\lvert GH - EJ \rvert$	Long axis of the humerus directed proximally (EJ to GH)
$\mathbf{y}_2 = \mathbf{z}_2 \times \mathbf{p}_2$	Line directed anteriorly perpendicular to the plane formed by $\mathbf{z}_2$ and $\mathbf{p}_2$
$\mathbf{x}_2 = \mathbf{z}_2 \times \mathbf{y}_2$	Line directed laterally perpendicular to the plane formed by $\mathbf{z}_2$ and $\mathbf{y}_2$

where $\mathbf{p}_2 =$ elbow joint axis.

The segmental coordinate system for the forearm was similarly defined (Figure 2):

O_3	Origin (WJ)
$\mathbf{z}_3 = (EJ - WJ)/\lvert EJ - WJ \rvert$	Long axis of the forearm directed proximally (WJ to EJ)
$\mathbf{y}_3 = \mathbf{z}_3 \times \mathbf{p}_3$	Line directed anteriorly perpendicular to the plane formed by $\mathbf{z}_3$ and $\mathbf{p}_3$
$\mathbf{x}_3 = \mathbf{z}_3 \times \mathbf{y}_3$	Line directed laterally perpendicular to the plane formed by $\mathbf{z}_3$ and $\mathbf{y}_3$

where $\mathbf{p}_3 =$ wrist joint axis.

Because markers were no smaller than 2.5 cm (DIA), a reference frame for the hand segment was not defined. However, a single principal vector pointing from the wrist joint to the hand marker (MC) was defined to allow for wrist flexion–extension estimations.

A thoracic coordinate system was created in order to measure overall trunk kinematics relative to the pelvis. The origin of this coordinate system was defined as the midpoint of the two shoulder joints and approximated a fixed point on the midline of the thorax (Starr et al., 2000):

$O_4 = (GH_L + GH_R)/2$	Origin (midpoint between the two shoulder joints)
$\mathbf{x}_4 = (GH_R - O_4)/\lvert GH_R - O_4 \rvert$	Vector pointing from left to right shoulder
$\mathbf{z}_4 = (\mathbf{p}_4 - O_4)/\lvert \mathbf{p}_4 - O_4 \rvert$	Line directed superiorly from origin to AC midpoint
$\mathbf{y}_4 = \mathbf{x}_4 \times \mathbf{z}_4$	Line directed anteriorly perpendicular to the plane formed by $\mathbf{x}_4$ and $\mathbf{z}_4$

where $\mathbf{p}_4 = (AC_R + AC_L)/2$.

Joint Coordinate Systems

A joint coordinate system (JCS) was created at the shoulder using the hinge axis ($\mathbf{x}_1$) of the shoulder girdle segment as the first JCS axis ($\mathbf{e}_1$) and the long axis ($\mathbf{z}_2$) of the humeral segment as the third JCS axis ($\mathbf{e}_3$). The second, or *floating*, axis ($\mathbf{e}_2$) was defined as the vector perpendicular to the plane created by the two body-fixed axes. Similarly, the elbow JCS was defined using the humeral lateral axis ($\mathbf{x}_2$) and the forearm long axis ($\mathbf{z}_3$) as the two fixed axes.

The JCS convention allowed for rotations measured relative to one coordinate system per joint where the terminal position of a joint was dependent on the selection of the JCS fixed axes and not the rotation sequence itself. This convention is simply an alternative to expressing a specific sequence of Euler angle rotations (Chao, 1980).

Joint Angle Calculations

Shoulder rotations were measured using the axes defined by the aforementioned JCS:

$\mathbf{a}_1$ Horizontal Abduction–Adduction

$\mathbf{a}_2$ Elevation

$\mathbf{a}_3$ Internal–External Rotation

Because of the ambiguity of clinical descriptors of shoulder movement, elevation was used to describe the motions of flexion–extension and abduction–adduction as their terminal positions are noncommutative (Wu et al., 2005). Left shoulder rotations were calculated similarly using mirrored coordinate data. Internal–external rotation, which is rotation about the humerus' long axis, was differentiated in time for calculations of angular velocity for the throwing arm.

Data processing of elbow kinematics was limited to calculations of flexion–extension owing to the uniplanar assumption necessary for estimation of the joint axis using the specified marker set. Thus the flexion–extension angle was calculated as simply the angle between the long axes of the forearm and humerus:

$\mathbf{a}_4$ Flexion-Extension