

The Long Head of the Biceps Tendon Has Minimal Effect on In Vivo Glenohumeral Kinematics

A Biplane Fluoroscopy Study

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Background: The in vivo stabilizing role of the long head of the biceps tendon (LHB) is poorly understood. While cadaveric studies report that the loaded LHB constrains translations in all directions, clinical data suggest that there is no clinically demonstrable alteration in glenohumeral position after LHB tenodesis or tenotomy. The purpose of this study was to investigate potential alterations in glenohumeral kinematics after LHB tenodesis during 3 dynamic in vivo motions using a biplane fluoroscopy system.

Hypothesis: Our hypothesis was that there would be no difference in glenohumeral translations greater than 1.0 mm between shoulders after biceps tenodesis and healthy contralateral shoulders.

Study Design: Controlled laboratory study.

Methods: Five patients who underwent unilateral, open subpectoral tenodesis performed abduction, a simulated late cocking phase of a throw, and simulated lifting with both their tenodesed shoulder and their contralateral healthy shoulder inside a biplane fluoroscopy system. Dynamic 3-dimensional glenohumeral positions and electromyography activity of the biceps brachii muscle were determined and compared.

Results: Significant glenohumeral translations occurred in both shoulders for abduction (3.4 mm inferiorly; $P < .01$) and simulated late cocking (2.6 mm anteriorly; $P < .01$). The mean difference for each motion in glenohumeral position between the tenodesed and the contralateral healthy shoulders was always less than 1.0 mm. The tenodesed shoulders were more anterior (centered) during abduction (0.7 mm; $P < .01$) and for the eccentric phase of the simulated late cocking motion (0.9 mm; $P < .02$). No significant differences were found during the simulated lifting motion and in the superior-inferior direction.

Conclusion: The effect of biceps tenodesis on glenohumeral position during the motions studied in vivo was minimal compared with physiological translations and interpatient variability.

Clinical Relevance: Our findings demonstrated that LHB tenodesis does not dramatically alter glenohumeral position during dynamic motions, suggesting the risk for clinically significant alterations in glenohumeral kinematics after tenodesis is low in otherwise intact shoulders.

Keywords: long head of the biceps tendon; tenodesis; glenohumeral kinematics; biplane fluoroscopy; impingement

The in vivo stabilizing role of the long head of the biceps tendon (LHB) is poorly understood. However, its role as a pain generator in the shoulder was described in 1948 by Hitchcock and Bechtol,¹⁰ who were also among the first to advocate biceps tenodesis. Tenodesis or tenotomy of the LHB has been reported to be an effective pain-relieving treatment option for recalcitrant biceps tendinitis and tendinopathy with low complication rates.^{24,27,31,33,39} However, the LHB has been reported in several in vitro

studies to have the mechanical ability to stabilize the glenohumeral joint.^{14,22,28,29,32,43} Therefore, tenodesis or tenotomy could theoretically alter glenohumeral joint kinematics, creating a subsequent concern for potential pathological changes in patients after these procedures. These concerns have sparked significant debate about the best indications for biceps tenodesis or tenotomy and for which patient populations the procedure is suitable. Moreover, the importance of investigating the stabilizing role of the LHB in vivo is emphasized by these debates.

Evidence of the ability of the LHB to constrain glenohumeral motion is based primarily on cadaveric studies, which reported that the LHB has stabilizing effects on the glenohumeral joint anteriorly, posteriorly, superiorly,

and inferiorly, suggesting concavity compression as its primary mechanism for providing glenohumeral stability.^{14,28,29,32,43} Others have suggested a pivotal role of the LHB in preventing superior humeral head migration.²² In vitro biomechanical investigations have also reported that the LHB significantly constrains rotational range of motion.^{32,43} Overall, cadaveric models have suggested that without the stabilizing presence of the LHB, increased glenohumeral translations of between 5 and 25 mm can occur in the superior-inferior and anterior-posterior planes.^{14,22,28,32,43}

To date, however, translations of this magnitude have not been demonstrated in vivo. In the shoulders of patients with isolated ruptures of the LHB, when compared with their contralateral healthy shoulders, average superior position differences of the humeral head of 2.2 mm (maximum difference of 6 mm) at 45°, 90°, and 120° of abduction were measured by Warner and McMahon⁴⁰ using serial, static radiographs. The authors theorized this amount of superior translation could contribute to an iatrogenic impingement syndrome. However, in the clinical literature, the stabilizing role of the LHB has not been as evident.^{4,39} For instance, the distance between the acromion and the humeral head in static radiographs was not altered after tenotomy.³⁹ Similarly, Boileau et al⁴ have further questioned the stabilizing role of the LHB, showing no clinical evidence of instability or proximal humeral migration in patients who were treated with tenodesis compared with those treated by superior labral anterior posterior (SLAP) repair. Moreover, significantly more patients were able to return to their previous level of sports participation when compared with SLAP repair in this study.

Clearly, surgeons still face a confusing and contradictory body of evidence surrounding the biomechanical function of the LHB. While cadaveric studies show a stabilizing effect of the LHB, clinical data suggest that there is no clinically demonstrable increased humeral head motion after LHB tenodesis or tenotomy. Therefore, the purpose of this study was to investigate the effects of the LHB on glenohumeral kinematics in vivo using highly sensitive and accurate 3-dimensional imaging. To do this, we used a biplane fluoroscopy system to compare shoulders in patients who had undergone an isolated LHB tenodesis with their healthy contralateral shoulder during abduction, a simulated late cocking phase of a throw, and a simulated lifting task (loaded forward flexion). Our hypothesis was that there would be no differences in glenohumeral translations greater than 1.0 mm between the shoulders.

MATERIALS AND METHODS

Patients

Five patients (3 men and 2 women; age, 41 ± 14 years; height, 1.77 ± 0.09 m; weight, 87 ± 23 kg) participated in this study and signed an informed consent form approved by the institutional review board of the Vail Valley Medical Center. All patients matched very strict inclusion criteria: (1) unilateral isolated subpectoral tenodesis of the LHB with concomitant arthroscopic subacromial decompression for the treatment of chronic recalcitrant biceps tenosynovitis that did not respond to nonoperative treatment, (2) no further shoulder injury such as rotator cuff tear or instability, (3) a minimum of 4 months past surgery and cleared for unrestricted motion and activities, (4) age greater than 18 years, (5) no previous injury or surgery to the contralateral shoulder, and (6) full subjective and objective motion and function restored in the operative arm and normal motion and function in the control arm. Upon enrollment, an orthopaedic surgeon took a detailed medical history and performed a full clinical shoulder examination bilaterally to assess shoulder function and to examine for signs of shoulder pathological abnormalities.

Exercise Testing

Each patient performed 3 motions with each shoulder independently while seated inside a biplane fluoroscopy system: abduction, a simulated late cocking phase of a throw, and a simulated lifting task, which are described in more detail below. The motions in this study were selected to address different questions, which are further addressed in detail in the discussion. Each shoulder was imaged separately with the glenohumeral joint of interest centered within the biplane fluoroscopy system. The patients practiced each motion several times before it was recorded using the biplane fluoroscopy system.

For abduction, the patients were seated with their backs straight and their arm relaxed at their side (Figure 1A). From this position, the patients started a full range of motion abduction of 2 seconds' duration with the elbow fully extended while holding their thumb up and in the plane of the motion to control arm rotation (Figure 1B).

For the simulated late cocking phase of a throw, each patient completed a loaded, eccentric external rotation from the abducted, externally rotated position followed

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One or more of the authors has declared the following potential conflict of interest or source of funding: Dr Elser held a visiting research scholar position funded by Arthrex Inc. Dr Millett received royalties from and is a paid consultant for Arthrex Inc. This study was funded by the Gumbo Foundation and by the Steadman Philippon Research Institute, a 501(c)(3) nonprofit institution supported financially by private donations and corporate support from the following entities: Smith & Nephew Endoscopy, Arthrex Inc, Siemans Medical Solutions USA, OrthoRehab, Össur Americas, Opedix, Conmed Linvatec, and SBI.

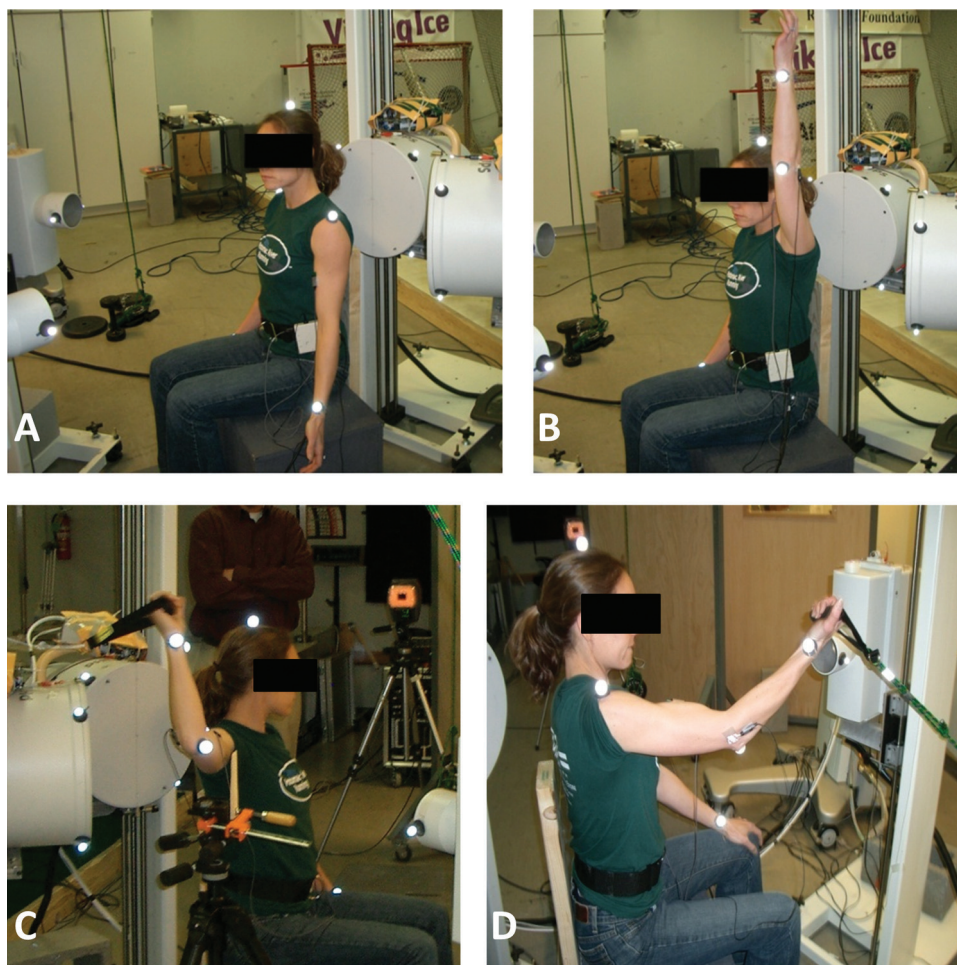


Figure 1. Patient seated in the biplane fluoroscopy system performing abduction (starting position [A] and end position [B]), with shoulder positioned in the abducted, externally rotated position waiting for a verbal cue (C), and performing the simulated lifting task by forward flexing while pulling on a weighted pulley system (D).

by concentric internal rotation. Specifically, the patients were seated with their backs straight and their arm raised 90° to the side, their arm externally rotated 90°, and their elbow flexed 90° (palm forward). From this position, the patients held a rope-pulley system with a 2.25-kg weight attached (Figure 1C). Upon verbal cue, the weight was dropped. The patients were instructed to allow the weight to pull their arm as far into external rotation as they were comfortable with (eccentric loading) and to then quickly internally rotate (concentric loading) until approximately 15° of internal rotation relative to the starting position. The entire motion lasted approximately 1.5 seconds.

For the simulated lifting task, each patient performed forward flexion over a 2-second period with the elbow flexed at approximately 30° and the forearm supinated while pulling on a weighted pulley system that was attached to the floor approximately 2 m in front of them (Figure 1D). The initial load was 30 N and peaked at approximately 100 N during the motion.

Data Collection

Data collection consisted of 3 parts for each shoulder: (1) a bilateral computed tomography (CT) scan, (2) motion recording using a biplane fluoroscopy system, and (3) surface electromyography (EMG) recordings of biceps brachii muscle activity throughout the captured motion.

Each patient underwent a high-resolution CT scan of both shoulders with an Aquilion 64 CT scanner (Toshiba America Medical Systems, Tustin, California). The scans included both clavicles and went distally to the inferior angles of the scapulae axially and included both humeri mediolaterally. A sequence of axial images was obtained using clinical shoulder scan technique factors and sharp bone CT reconstruction (voxel size of approximately $0.95 \times 0.95 \times 0.5 \text{ mm}^3$).

The custom biplane fluoroscopy system was constructed from 2 BV Pulsera C-arms (Philips Medical Systems, Best, the Netherlands), of which the control systems were

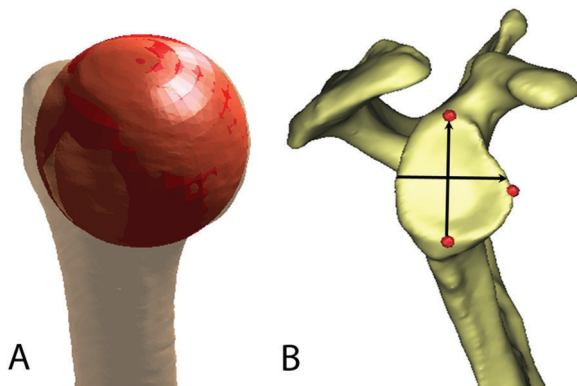


Figure 2. (A) The humeral head center was determined by fitting a sphere to the articular surface of the humeral head. (B) The glenoid coordinate system was based on the most superior, inferior, and anterior points on the computed tomography-based models.

accurately synchronized. Two coupled, high-speed, high-resolution (1024×1024) digital cameras (Phantom V5.1, Vision Research, Wayne, New Jersey) were interfaced with the image intensifiers of the fluoroscopy systems using a custom interface. The radiographic generators and image intensifiers were mounted on a custom gantry to allow freedom of movement inside the system. The C-arms were operated in continuous fluoroscopy mode, and the exposures were captured at 100 frames per second.

The system was calibrated by imaging a square grid for image distortion correction⁷ followed by a small calibration cube to determine the x-ray focus positions and relative position and orientation of the 2 fluoroscopes (Kaptein et al¹⁵). The biplane fluoroscopy system was validated using standard validation techniques.^{3,13,16} Motion data on 4 cadaveric shoulders were collected in the same manner as in this in vivo study. Bias and precision² relative to bead tracking were calculated from 30 frames² during abduction for each specimen, and average bias and precision were 0.2 ± 0.5 mm, 0.3 ± 0.3 mm, and 0.3 ± 0.4 mm for anterior-posterior, superior-inferior, and distraction-compression translations, respectively. These values were similar to those previously reported for the knee on our system (0.2 ± 0.3 mm, -0.1 ± 0.1 mm, -0.05 ± 0.1 mm in translations and $0.1^\circ \pm 0.1^\circ$, $0.3^\circ \pm 0.2^\circ$, $0.1^\circ \pm 0.3^\circ$ in rotations, respectively),³⁶ and they were consistent with previously reported studies using similar biplane fluoroscopy technology.^{1,3,23}

The surface EMG signals for each biceps brachii muscle were collected at 1200 Hz (Bagnoli-8, DelSys Inc, Boston, Massachusetts) and synchronized with the biplane fluoroscopy system using the data collection module of a motion analysis system (Motion Analysis Corporation, Santa Rosa, California). The EMG electrodes were placed over the center of each muscle belly and in line with the muscle fibers. Prior to completing the motions in the biplane fluoroscopy system, 3 isometric maximum voluntary contractions (MVC) were recorded with the arm at the side and the elbow flexed at 90° to provide the EMG signals during the motions with a maximum effort reference.

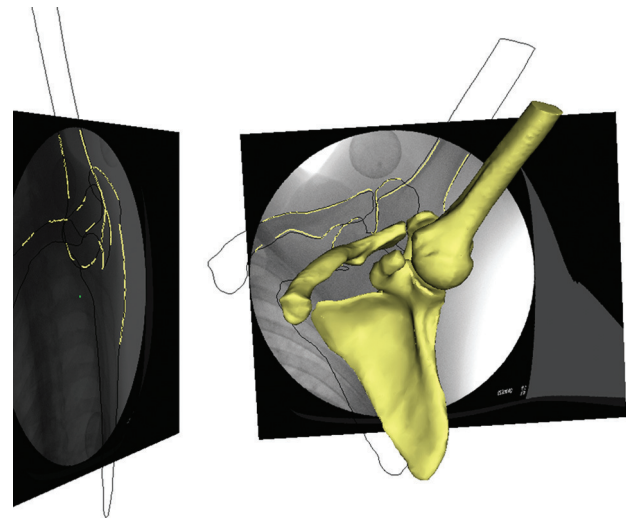


Figure 3. Matched bone geometries for an abduction frame. The algorithm matches the detected bone contours (pale lines) with the projected bone contours (black lines).

Data Analysis

Data analysis consisted of 4 steps: (1) 3-dimensional bone geometry reconstruction of the humerus and scapula from CT data, (2) coordinate system assignment and geometry transformation, (3) bone position and orientation determination in the biplane fluoroscopy data, (4) and post-processing to extract the 3-dimensional glenohumeral translations.^{5,35,36}

The bones were identified in the CT images, and the 3-dimensional geometry of the humerus and scapula were reconstructed using commercial software (Mimics, Materialise, Plymouth, Michigan). Custom-written software (Matlab, The Mathworks, Natick, Massachusetts) assigned anatomic coordinate systems to the bone geometries⁴¹ and transformed the bones to positions suitable for position and orientation tracking. The humeral head center was determined by automatically fitting a sphere to the articular portion of the humeral head (Figure 2A). A glenoid coordinate system was created based on the most superior, inferior, and anterior points on the glenoid rim. The glenoid center was calculated to be midway between the superior and inferior glenoid rim points (Figure 2B).

Determination of the bone position and orientation from the biplane fluoroscopy data was performed using validated commercial software (Model-Based RSA, Medis Specials, Leiden, the Netherlands).^{12,13,16,17} In this software, the contours of the humerus and scapula were automatically extracted from the biplane fluoroscopy images using Canny edge detection and manually assigned to each bone. Subsequently, a fully automatic 6 degrees-of-freedom optimization algorithm¹⁸ determined the position and orientation, which optimally matched the detected contours with the projected contours from the imported bone geometries (Figure 3).¹⁷ From the bone position and orientation for each frame, the position of the

humeral head center was determined (in mm) relative to the glenoid coordinate system. Glenohumeral position was quantified in the superior-inferior and anterior-posterior directions for the tenodesed and healthy shoulders during each motion.

For shoulder abduction, the glenohumeral positions were determined for specific arm elevation angles from 20° to 150° of arm elevation in 10° increments. For the simulated late cocking motion, the glenohumeral positions were determined for 3 arm positions: initial position (90° abducted, 90° externally rotated position), maximum external rotation, and final internal rotation. To investigate the eccentric and concentric phases of the motion in more detail, the eccentric phase (from initial to maximum external rotation position) as well as the concentric phase (from maximum external rotation to final internal rotation position) were resampled in 20% increments from 0% to 100% to normalize for patient variations in timing of the motion. Lastly, for simulated lifting, the glenohumeral positions were determined for specific arm elevation angles from 40° of arm elevation to 130° in 10° increments.

The raw EMG data were processed with a 50-millisecond root mean square (RMS) moving window (1-millisecond increments). The EMG reference values were calculated using the average of the peak RMS values of 3 MVCs. Muscle activation was quantified by calculating average and peak values expressed as a percentage of the MVC (%MVC) over the full range of motion for the abduction and simulated lifting tasks and for the eccentric external rotation and concentric internal rotation phases of the simulated late cocking motion.³⁴

Statistical Analysis

For all motions, statistical analysis was performed in both the superior-inferior and anterior-posterior directions. The statistical significance level was set to .05.

For abduction and the simulated lifting task, the statistical analysis of the glenohumeral positions consisted of a 2-way repeated-measures ANOVA with the independent measures of shoulder (tenodesed vs healthy) and arm elevation angle (abduction: 20°-150° in 10° increments; simulated lifting: 40°-130° in 10° increments). Peak and averaged EMG amplitudes over the entire motion range were compared between the 2 shoulders using a paired *t* test.

For the simulated late cocking motion, a 2-way repeated-measures ANOVA with independent factors of shoulder (tenodesed vs healthy) and arm position (initial, maximum external rotation, and final internal rotation) was performed to detect differences in glenohumeral positions. For each phase (eccentric external rotation and concentric internal rotation), a 2-way repeated-measures ANOVA with independent factors of shoulder (tenodesed vs healthy) and percentage phase (0%, 20%, 40%, 60%, 80%, and 100%) was performed to detect differences in glenohumeral translations during each phase. The mean \pm 1 standard deviation and the confidence interval of the translations that occurred between the 3 arm positions were calculated by subtracting the glenohumeral position at the start of the motion from the glenohumeral position at maximum

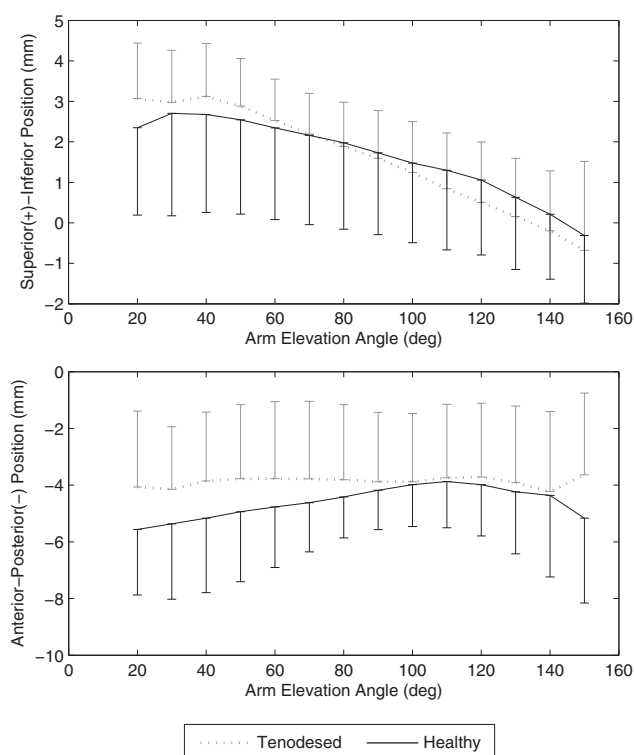


Figure 4. Glenohumeral superior-inferior (top) and anterior-posterior (bottom) positions (mean \pm 1 standard deviation) as a function of arm elevation angle for both the tenodesed (dashed) and healthy (solid) shoulders during abduction.

external rotation (eccentric phase) and by subtracting the glenohumeral position at maximum external rotation from the glenohumeral position at final internal rotation (concentric phase). The amount of external rotation from the start of the motion to the maximum external rotation position was recorded and compared between the 2 shoulders using a repeated-measures *t* test. Lastly, a 2-way repeated-measures ANOVA with independent factors of shoulder (tenodesed vs healthy) and phase (eccentric vs concentric) was performed for average and peak EMG values. Post hoc Bonferroni-corrected *t* tests were performed when significant effects were found in the ANOVA.

RESULTS

The isolated biceps tenodesis was performed on the left shoulder of 3 patients and on the right shoulder of 2 patients. All patients were right-hand dominant. All patients had range of motion asymmetries between shoulders of less than 10° in all planes. Rotator cuff motor strength testing was grade 5/5 bilaterally in all patients by clinical examination. No patient had greater than grade 1 laxity in anterior, posterior, and inferior translations by the load and shift test in either shoulder. In the postoperative shoulders, no residual anterior shoulder pain was present, and no biceps deformities were noted.

TABLE 1

Mean \pm Standard Error for the Peak and Average Electromyography Amplitude (%MVC) for the Healthy and Tenodesed Shoulders During Abduction, Both Phases of the Simulated Late Cocking Motion, and the Simulated Lifting Task^a

Motion	Electromyography Amplitude (%MVC)			
	Peak		Average	
	Healthy	Tenodesed	Healthy	Tenodesed
Abduction	20.9 \pm 16.7	14.1 \pm 12.2	9.5 \pm 6.8	7.9 \pm 7.5
Late cocking (ecc)	7.6 \pm 7.4	9.6 \pm 7.2	5.4 \pm 5.1	4.5 \pm 3.8
Late cocking (con)	14.5 \pm 11.6	19.1 \pm 8.9	7.4 \pm 5.7	9.2 \pm 5.4
Lifting	125.9 \pm 40.4	114.5 \pm 42.4	45.6 \pm 14.0	61.2 \pm 16.5

^aMVC, maximum voluntary contraction; ecc, eccentric phase; con, concentric phase.

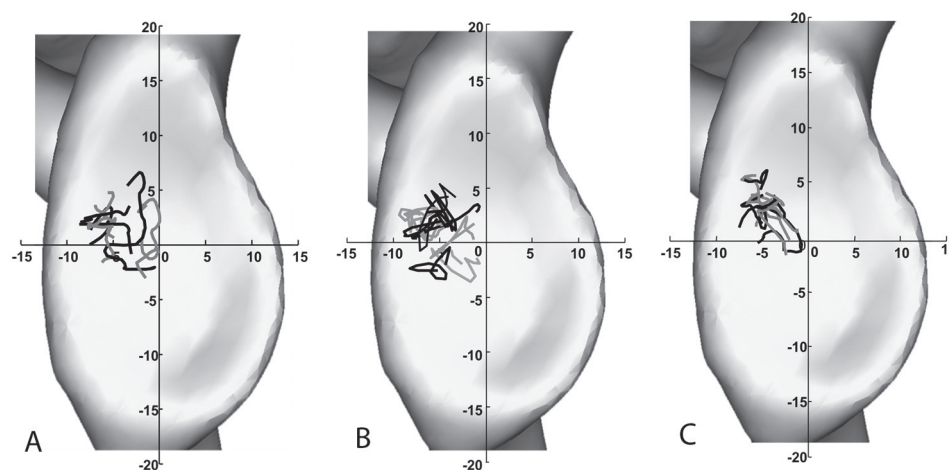


Figure 5. Path of glenohumeral motion graphed for all patients, with gray lines representing the tenodesed shoulders and black lines representing the healthy contralateral shoulders: (A) abduction, (B) simulated late cocking phase of a throw, and (C) simulated lifting task. The glenoid image is scaled to represent the approximate average glenoid size and is provided for visual reference only.

Abduction

Figure 4 demonstrates the glenohumeral positions for the tenodesed and healthy shoulders versus the discrete arm elevation angles. On average for both shoulders, the humeral head center was positioned approximately 1 to 2 mm superiorly and 4 to 5 mm posteriorly of the midpoint between the most superior and most inferior points of the glenoid rim. In the superior-inferior direction, the positions of the tenodesed shoulders were on average 0.1 mm inferior of the healthy shoulder positions (not significant), while the average translation for both shoulders as a function of arm elevation was 3.4 mm ($P < .01$). No interaction effects were found. Thus, the mean position difference in the tenodesed shoulder was 2.9% of the physiological translations that occurred in both shoulders. Anterior-posterior positions demonstrated a significant mean difference of 0.7 mm between shoulders ($P < .01$), with the tenodesed shoulders being more anterior than the healthy shoulders (Figure 4). No significant differences because of

arm elevation angle or interaction effects were found. The maximum difference between shoulders at any arm elevation angle occurred at 20° of arm elevation and was 0.7 mm for superior-inferior position and 1.5 mm for anterior-posterior position. Figure 5A shows the individual paths of glenohumeral motion (superior-inferior position graphed vs anterior-posterior position) of the tenodesed and healthy shoulders and visualizes the small differences between the shoulders.

The biceps EMG activation levels were medium to low for this exercise, with peak activation levels of 21% MVC and average activation levels below 10% MVC (Table 1). There was no significant difference in EMG activation between the tenodesed and healthy contralateral shoulders.

Simulated Late Cocking

Figure 6 demonstrates the glenohumeral positions for the tenodesed and healthy shoulders at the initial position

(abducted and externally rotated position), the maximum external rotation, and final internal rotation arm positions. Throughout the motion for both shoulders, the humeral head was positioned approximately 1 to 2 mm superior and 5 to 7 mm posterior of the midpoint between the most superior and inferior points on the glenoid rim (Figure 5B). In the anterior-posterior direction during the eccentric phase, there was a significant difference between shoulders ($P < .02$), with the humeral head positioned on average 0.9 mm more anterior (more centered) in the tenodesed shoulder compared with the healthy shoulder. A significant anterior translation of 2.6 mm ($P < .01$) was also observed in both shoulders between the maximum external rotation and final internal rotation position. No other significant differences between shoulders, arm positions, or interaction effects were found. The largest glenohumeral position difference between shoulders was 1.2 mm in the superior-inferior direction in the final internal rotation arm position, with the healthy shoulders being superior to the tenodesed shoulders (not significant). Translations in the tenodesed shoulders were always smaller than the translations in the healthy shoulders in all planes (Table 2). There was no significant difference in the amount of external glenohumeral rotation between the shoulders from the initial position until the maximum external rotation position ($47.8^\circ \pm 16.5^\circ$ vs $51.9^\circ \pm 8.2^\circ$ for the tenodesed and healthy shoulders, respectively).

The biceps EMG activation levels were considered low for this exercise, with the highest peak activation level of 19.1% MVC (Table 1) and average activation levels always below 10% MVC. The EMG activation was significantly higher for the concentric internal rotation phase versus the eccentric external rotation phase (peak: $P < .01$; average: $P < .01$), but there was no significant difference in activation between the tenodesed and healthy contralateral shoulders and no significant interaction effect.

Simulated Lifting

Figure 7 demonstrates the glenohumeral positions for the tenodesed and healthy shoulders versus the discrete arm elevation angles. Relative to the origin, the humeral head center was positioned approximately 2 to 3 mm superiorly and 4 to 6 mm posteriorly (Figure 5C). No significant differences were found in either direction. In the superior-inferior direction, the tenodesed side was positioned on average 0.5 mm more superior (not significant) than the healthy side, with a maximum position difference between shoulders of 1.0 mm at 70° of arm elevation. In the anterior-posterior direction, the tenodesed side was positioned on average 0.1 mm anterior (not significant) of the healthy side, with a maximum position difference between shoulders of 1.2 mm at 130° of arm elevation.

Biceps EMG activation was high with peak activation at approximately 120% MVC, even higher than during the controlled isometric MVC trials, and average biceps activation up to 61% MVC (Table 1). There was no significant difference in EMG activation between the tenodesed and the healthy shoulders.

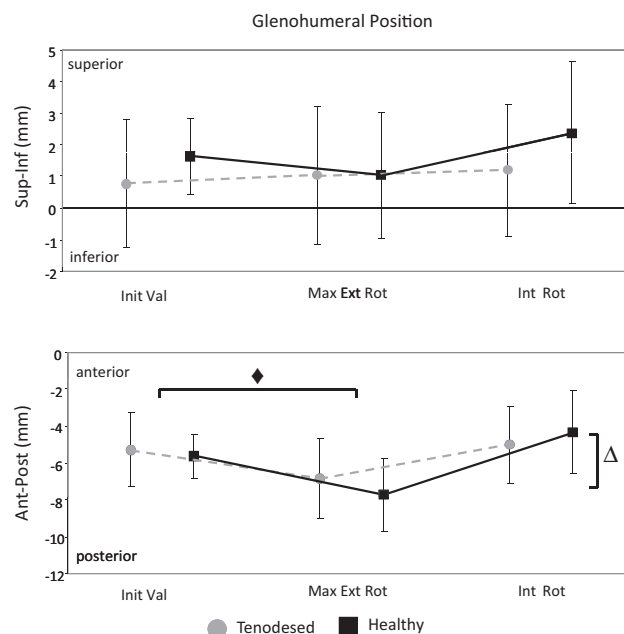


Figure 6. Mean and standard deviation for glenohumeral position in the superior-inferior (top) and anterior-posterior (bottom) directions for the tenodesed (circle) and healthy (square) shoulders for the 3 arm positions of initial position, maximum external rotation, and final internal rotation (♦ and Δ indicate significant differences).

DISCUSSION

This study presented a biplane fluoroscopy analysis of glenohumeral positions during 3 dynamic motions in patients with isolated biceps tenodesis compared with their unaffected clinically normal contralateral shoulder. We found that the difference in glenohumeral motion between the biceps tenodesis and contralateral healthy shoulders averaged across a motion was always less than 1.0 mm. While statistically significant differences were found between the tenodesed and healthy shoulders, with the tenodesed shoulder more anterior (centered) during abduction (0.7 mm) and the eccentric phase of the simulated late cocking motion (0.9 mm), we believe these differences are unlikely to be of clinical significance. No significant differences were found in the superior-inferior direction and during the simulated lifting motion. The maximum difference in glenohumeral position between the shoulders in any direction, at any time point, or for any arm position never exceeded 1.5 mm. Both shoulders had significant glenohumeral translations occurring during abduction (3.4 mm inferiorly) and simulated late cocking (2.6 mm anteriorly), demonstrating that the between-shoulder differences were smaller than the physiological translations that occur naturally in the shoulder. Thus, our results suggest that the risk for clinically significant alterations in glenohumeral joint kinematics because of biceps tenodesis or tenotomy is low in otherwise intact shoulders.

TABLE 2

Mean \pm Standard Deviation (Confidence Interval) of the Superior-Inferior (SI) and Anterior-Posterior (AP) Translations During the Eccentric Phase From the Initial Position to Maximum External Rotation and During the Concentric Phase From Maximum External Rotation to Final Internal Rotation

	Eccentric Phase		Concentric Phase	
	Tenodesed	Healthy	Tenodesed	Healthy
SI	0.28 ± 1.65 (-1.77 to 2.33)	-0.58 ± 0.98 (-1.81 to 0.64)	0.15 ± 1.15 (-1.27 to 1.57)	1.33 ± 1.11 (-0.05 to 2.71)
AP	-1.55 ± 1.46 (-3.37 to 0.26)	-2.09 ± 2.35 (-5.01 to 0.83)	1.83 ± 1.70 (-0.28 to 3.95)	3.41 ± 2.64 (1.29 to 5.52)

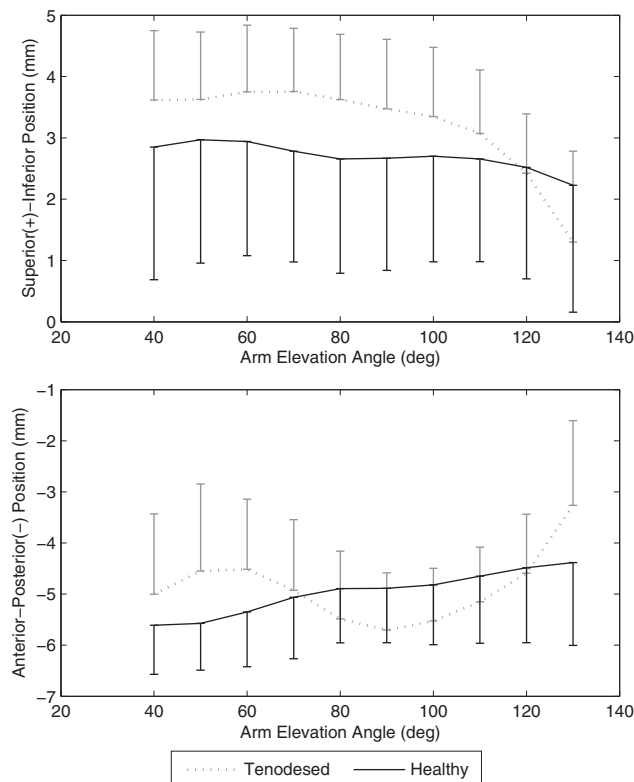


Figure 7. Glenohumeral superior-inferior (top) and anterior-posterior (bottom) positions (mean \pm 1 standard deviation) as a function of arm elevation angle for both the tenodesed (dashed) and healthy (solid) shoulders during simulated lifting (forward flexion).

The results of this study agree with those of previous clinical studies,^{4,39} which have not reported any demonstrable effects of biceps tenodesis or tenotomy on glenohumeral position. The differences between shoulders presented here are smaller than those reported in a previous static radiography study⁴⁰ and much smaller than those demonstrated in previous in vitro experiments.^{14,22,28,29,33,43} Therefore, while the in vitro experiments clearly show that the LHB has the mechanical capability to constrain glenohumeral motion, its overall contribution to glenohumeral stability in vivo in otherwise healthy shoulders, as evidenced by glenohumeral translations, was minimal. Our results suggest that the other

stabilizing structures of the shoulder, such as the surrounding musculature, are producing significantly higher forces at the glenohumeral joint in vivo, such that the contribution of the biceps provides a minimal secondary stabilizing role when these other stabilizing structures are intact.

It is entirely unknown at this time whether average position differences of less than 1.0 mm have clinically and biologically meaningful effects on the glenohumeral joint, particularly in the long term. After biceps tenodesis or tenotomy, patients do not complain of perceived instability, and our patients had no detectable instability or laxity on clinical examination. In addition, the physiological translations measured in this study were several times larger than the differences between shoulders that were found. Moreover, the between-patient variability at any time during the motions (ie, standard deviation) was larger than the between-shoulder differences. These all suggest that the potential for long-term deleterious effects of biceps tenodesis is likely to be small in otherwise healthy and stable shoulders. However, long-term follow-up of LHB tenodesis patients is needed to definitively address this issue.

The motions in this study were selected to address several distinct questions. Abduction was selected because many have proposed that the LHB has a humeral head-depressing effect, and as such, this motion provides a direct comparison with the only previous in vivo study by Warner and McMahon,⁴⁰ who reported average superior glenohumeral migration in shoulders with a biceps rupture of 2.2 mm with extremes of 6 mm during abduction. Our data did not find shoulder position differences this large. It is possible this was entirely because of the application of the newer and more accurate technology available today. Projection radiographs are prone to positioning and projection errors.³⁷ Moreover, during abduction, there is scapular protraction and posterior tilting, which may not have been compensated for. In biplane fluoroscopy, bone position and orientation, and therefore glenohumeral translations, were accurately determined in 3 dimensions, eliminating these potential sources of error. Furthermore, for all of the patients in our study, magnetic resonance imaging (MRI) scans were obtained, and arthroscopic procedures were performed, which confirmed that the rotator cuffs were intact. Alternatively, it is unknown whether glenohumeral positions are different between serial static postures or during dynamic motions, which should be investigated in the future.

The simulated late cocking phase of a throw motion was selected based on the frequent clinical conundrum about whether biceps tenodesis should be performed on overhead-throwing athletes or overhead workers with SLAP tears. These specific populations are of particular interest to our practice, and others, because they present frequently. Some studies have even suggested better outcomes with biceps tenodesis as compared with SLAP repair.⁴ Unfortunately, a completely unrestricted throw was not feasible inside the biplane fluoroscopy system because of its 3-dimensional field of view of approximately the size of a basketball. In previous *in vitro* studies, specifically designed and reported to simulate a throw, the late cocking phase of a throw is simulated by positioning a cadaveric shoulder specimen in the abducted and externally rotated position and subsequently applying an external rotation torque.^{9,11,25,26} For our *in vivo* study, we opted to not solely apply an external rotation torque to the arm in the abducted and externally rotated position while controlling arm position but instead to apply the torque through a pulley system held by hand while allowing the arm to move freely. We considered this a step closer toward the goal of an unrestricted throw compared with a pure external rotation torque and believe this motion to be more representative of (although not identical to) the *in vivo* conditions of overhead throwing motion. Glenohumeral 3-dimensional rotations were compared between shoulders to ensure both shoulders moved in a similar way and were found to be not significantly different.

Our results demonstrate that there was a small mean difference (0.9 mm) in the anterior-posterior direction during this motion, while there was simultaneously an anterior translation of 2.6 mm in both shoulders. The EMG activity was low during this motion (peak of 19%), which is consistent with other EMG data on muscle activity during the late cocking phase of pitching (26%) and the football throw (20%).^{6,8,20} The glenohumeral position results could be misinterpreted as demonstrating that the biceps has a stabilizing effect because even at low activations, there is a measureable difference between the shoulders. However, the data actually demonstrated that the tenodesed shoulders always translated less than the normal shoulders, which is entirely contrary to what would be expected if the LHB had a stabilizing effect.

The simulated lifting task included in this study was designed to maximally activate the biceps throughout the forward flexion range of motion. This task is applicable to heavy and manual laborers and weight lifters who commonly have LHB abnormalities.³⁸⁻⁴⁰ The EMG data demonstrated that this goal was achieved by both the high peak activity and average biceps activation. However, no significant differences were found in this motion, and mean differences between shoulders were 0.5 mm or less, and maximum differences in any position or direction did not exceed 1.2 mm, suggesting that even with high levels of biceps activation, only small changes in glenohumeral motion were present. Thus, the data in this study demonstrate that glenohumeral position differences between shoulders were small with or without high biceps muscle activity.

In our study, the operative shoulders were compared with the patients' contralateral healthy shoulders. Alternatively,

we could have compared the same shoulder in patients before and after tenodesis. However, preoperative pain and dysfunction might have altered the motions of the shoulder, resulting in a biased comparison with the postoperative shoulder. Thus, our study design was as close as possible to the ideal study of comparing the same shoulder before and after tenodesis in patients with healthy preoperative shoulders, which we believe would be ethically unacceptable. In addition, our study design allowed for repeated-measures statistical analysis, increasing our ability to detect consistent small effects that occurred in the presence of larger between-patient variability due to differences in general glenohumeral positions.

Both a strength and limitation of the study was the accuracy of the biplane fluoroscopy system. On one hand, the accuracy of biplane fluoroscopy systems is the highest of *in vivo* techniques available to date with submillimeter accuracy in 3 dimensions.^{1,3,23} As demonstrated by the statistically significant submillimeter findings, this technique allows for the detection of very small differences between groups. On the other hand, if it is shown in the future that these small differences are clinically relevant, improvements in the technique are necessary to achieve even higher system accuracies. The accuracy of the system (0.7 mm) and the findings of this study clearly demonstrate that the stabilizing effect of the biceps is limited to approximately 1.0 mm or less in the otherwise healthy shoulder. This was consistently found among the 3 motions.

The data in this study do not address the possibility that the LHB is an important stabilizer when there are significant other injuries in the shoulder such as SLAP tears, glenohumeral instability, and/or rotator cuff tears. Itoi et al¹⁴ theorize in their *in vitro* biomechanical study that the role of the LHB and short head biceps becomes more important as shoulder stability decreases in the position of abduction and external rotation. In shoulders with rotator cuff tears, the force couple may also be imbalanced, which allows abnormal superior translation of the humeral head,^{19,30,42} and the role of the LHB in glenohumeral stability may be increased.²¹ Our findings warrant further well-designed trials of LHB tenodesis in patients with additional shoulder abnormalities such as SLAP tears using the same methodology.

CONCLUSION

This is the first biplane fluoroscopy study to examine the stabilizing role of the LHB under dynamic conditions *in vivo*. As hypothesized, the effect of biceps tenodesis on mean glenohumeral position *in vivo* was less than 1.0 mm and small when compared with physiological translations and interpatient variability. Our findings demonstrate that LHB tenodesis does not dramatically alter glenohumeral translation during dynamic motions in otherwise healthy shoulders. While long-term follow-up studies are needed to prove this point, our data suggest, based on current knowledge, a low probability of long-term deleterious effects following biceps tenodesis in an otherwise stable shoulder. These findings are useful to surgeons who are contemplating LHB tenodesis in individuals with LHB injury.

ACKNOWLEDGMENT

The authors thank Robert F. LaPrade, MD, PhD, for his invaluable input to the article. In addition, we thank Jacob P. Krong, Daniel S. Peterson, Elizabeth Hageman, and Andrea North for their support in the processing of the biplane fluoroscopy data and Dr Michael Decker for his assistance with the EMG processing.

The authors thank Arthrex Inc for funding the research fellow position at the Steadman Philippon Research Institute held by Dr Florian Elser at the time the research was performed. Furthermore, the authors thank Medis Specials BV for providing the tracking software for the biplane fluoroscopy system.

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