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Errors Associated With Utilizing Prescribed Scapular Kinematics to Estimate Unconstrained, Natural Upper Extremity Motion in Musculoskeletal Modeling

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Abstract

Musculoskeletal modeling is capable of estimating physiological parameters that cannot be directly measured, however, the validity of the results must be assessed. Several models utilize a scapular rhythm to prescribe kinematics, yet it is unknown how well they replicate natural scapular motion. This study evaluated kinematic errors associated with a model that employs a scapular rhythm using 2 shoulder movements: abduction and forward reach. Two versions of the model were tested: the original MoBL ARMS model that utilizes a scapular rhythm, and a modified MoBL ARMS model that permits unconstrained scapular motion. Model estimates were compared against scapulothoracic kinematics directly measured from motion capture. Three-dimensional

scapulothoracic resultant angle errors associated with the rhythm model were greater than 10° for abduction (mean: 16.4° , max: 22.4°) and forward reach (mean: 11.1° , max: 16.5°). Errors generally increased with humerothoracic elevation with all subjects reporting greater than 10° differences at elevations greater than 45° . Errors associated with the unconstrained model were less than 10° . Consequently, use of the original MoBL ARMS model is cautioned for applications requiring precise scapulothoracic kinematics. These findings can help determine which research questions are suitable for investigation with these models and assist in contextualizing model results.

Keywords

shoulder; scapulothoracic; biomechanics; scapula; model

Musculoskeletal modeling is capable of estimating physiological parameters that cannot be directly measured,^{1,2} however, the validity of the results must be assessed. A substantial challenge of modeling the shoulder lies in proper implementation of scapular kinematics.^{3,4} Scapular motion is subject-specific and involves a unique combination of rotational and translational motion. Consequently, it is difficult to parameterize individualized scapular motion while retaining physiological moment arms of the muscles spanning the glenohumeral joint. One nonindividualized approach commonly employed in modeling utilizes regression equations that couple scapulothoracic motion to humerothoracic elevation, thereby creating a relationship known as scapular rhythm.^{5,6} Scapular rhythm represents a generic approach to the estimation of scapulothoracic orientation that is rarely used for precise measurement of an individual. The rhythm's regression equations were developed from a limited number of positions and based on healthy group tendencies.⁵ Consequently, the rhythm is not subject-specific, will not represent abnormal motion, and may not be suitable for certain motions. Despite these limitations, several widely-used and freely-available models utilize a scapular rhythm^{4,7-10} to prescribe kinematics, yet it is unknown how well these models replicate natural scapular motion. Errors associated with the scapular rhythm have been previously quantified but only for a single plane of humeral elevation⁵ and never within a modeling environment.

The purpose of this study was to assess the ability of a model, MoBL ARMS,^{4,7} which employs a prescribed scapular rhythm to replicate natural scapular kinematics during 2 shoulder movements: abduction and forward reach. Two versions of the model were tested: the original MoBL ARMS model that utilizes a scapular rhythm, and a modified MoBL ARMS model that permits unconstrained scapular motion. Model estimates were compared against scapular kinematics directly measured from motion capture using an acromion marker cluster.¹¹ A systematic review of methods to measure scapular motion reports average acromion marker cluster errors of 5° to 7° on each axis of motion during abduction and forward flexion.¹² This amount of error is approximately equivalent to 10° of 3D resultant scapulothoracic helical angle. We hypothesized that differences in 3D resultant scapulothoracic angle between the scapular rhythm model and direct measures would be greater than 10° for the majority of both motions. Additionally, we hypothesized that the unconstrained model would have 3D resultant scapulothoracic errors less than 10° . We also

hypothesized that strong positive correlations would exist between model-predicted and directly measured kinematics for both models. Finally, since the positions used to develop the scapular rhythm did not emphasize forward reaching,⁵ we hypothesized that errors associated with the rhythm model would be greater during forward reach than abduction.

Methods

Five healthy adult subjects (ages 26–35, 2 male, 3 female) participated. Informed consent was obtained in accordance with the institution's human subjects review board.

Markers were placed on T1 and T8 spinous processes, manubrium, and medial and lateral humeral epicondyles. An acromion marker cluster¹¹ was affixed to the acromion process of the scapula with the central marker of the triad positioned directly over the landmark. The acromion marker cluster was used to estimate scapular kinematics during both motion trials.¹¹ Subjects placed their arms in a static neutral position and 2 additional markers were positioned on the trigonum spinae and inferior angle of the scapula; these markers were removed for dynamic trials. For abduction and forward reach, subjects began in a neutral position then moved to approximately 90° humerothoracic elevation within the frontal and sagittal planes, respectively. Three-dimensional marker positions were recorded with a 12-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA) collecting at 60 Hz.

In the static neutral trial, the relationship (ie, transformation matrix [R]) was established between the orientation of the scapula, as determined by palpation, and the orientation of the acromion marker cluster. This relationship was then utilized to estimate scapular orientation during the dynamic trials based on the orientation of the acromion marker cluster; this technique has been described in previous literature.^{11,13,14} Scapulothoracic joint angles were calculated using a YXZ Euler sequence in accordance with International Society of Biomechanics (ISB) recommendations.¹⁵ Humerothoracic joint angles were calculated using XZY and ZXY Euler sequences for abduction and forward reach, respectively, as only the first rotation was of interest for this study.

All simulations were performed on the MoBL ARMS dynamic musculoskeletal model of the upper extremity⁷ using OpenSim 3.3.¹ The MoBL ARMS model's unscaled segment lengths and properties is based on the anthropometry of a 50th percentile male.¹⁶ The model includes 15 degrees of freedom that define the kinematics of the shoulder, elbow, forearm, wrist, and hand.⁴ All model joint kinematic conventions follow recommendations by the ISB.¹⁵ Two versions of the model were investigated. The first was the unaltered model that employs the scapular rhythm as described above. The constraints enforcing the scapular rhythm were re-moved to create a second modified version of the model that permitted unprescribed, free scapular motion; dynamic scapular orientation for this version of the model was estimated by the acromion marker cluster.

Model segments were scaled based on marker positions from the static neutral position. OpenSim's Inverse Kinematics tool was used to calculate model joint kinematics. Simulations for both motions began with the arm close to the neutral position without

placing the shoulder in gimbal lock, and ended at approximately 90° humerothoracic elevation. The free scapula model used the acromion marker cluster to determine dynamic scapular orientation, whereas the rhythm model prescribed scapular orientation as a function of humeral elevation.

Anatomical joint orientations (ie, not model joint orientations) were calculated with OpenSim's Body Kinematics Analysis tool throughout the inverse kinematics results. XYZ Euler angles describing the orientation of each segment were used to construct rotation matrices for each segment throughout each trial. Using the location of the model markers and the rotation matrices for each segment, segment coordinate systems for the thorax, scapula, and humerus were constructed in the same method described above for motion capture. Model-predicted scapulothoracic joint angles were then calculated using the same method described above.¹⁵

Model-estimated scapulothoracic kinematics were compared to kinematics directly measured from motion capture. Three-dimensional resultant scapulothoracic helical angle¹⁷ errors greater than 10° were considered meaningful as this threshold is approximately equivalent to measurement errors associated with the acromion marker cluster.¹² Finally, the percentage of the trial with greater than 10° differences was calculated and evaluated.

Differences between each model and direct measures were investigated on each scapulothoracic axis where X = upward/ downward rotation, Y = internal/external rotation, Z = anterior/ posterior tilt. Relative differences on each axis were examined by expressing the differences as a percent of the excursion on that axis throughout each motion. Pearson's correlation analysis assessed trends between each model and direct measures on each scapulothoracic axis. Pearson's *r*-values were interpreted as strong ($r \geq .7$), moderate ($.7 > r \geq .4$), and weak ($r < .4$).¹⁸ Finally, all results were compared between motions.

Results

Model scaling and inverse kinematic errors were within the Open-Sim Recommended Best Practices for both models.¹⁹

For abduction, the scapular rhythm model produced >10° mean and maximum scapulothoracic resultant angle differences for all subjects, compared to only 1 subject's free model (Figure 1A). Errors for the rhythm model were substantially larger than the free model (Figure 1A and Table 1). Meaningful differences (>10°) were maintained for a majority of each subject's abduction trial with the rhythm model, compared to only a minimal portion of the trial with the free model (Table 1). Errors of the rhythm model increased with humerothoracic elevation for all subjects, while the free model errors were more consistent across all levels of elevation (Figure 1A). Both models reported strong correlations with directly measured scapulothoracic upward rotation (Figure 2A and Table 2). The free model had strong correlations with directly measured scapulothoracic internal rotation, while rhythm model correlations were negative for all subjects. Correlations for the free model on scapulothoracic posterior tilt during abduction were strong for 4 subjects and negative for 1 subject. The rhythm model trends for scapulothoracic posterior tilt ranged

from negative to strong. On all scapulothoracic axes, kinematic errors were substantially larger for the rhythm model (Figure 2A–C and Table 2).

For forward reach, each subject's rhythm model achieved and maintained meaningful ($>10^\circ$) scapulothoracic resultant angle errors (Table 1 and Figure 1B); errors for the free model were less than 10° . Both models had strong correlations with directly measured scapulothoracic upward rotation and internal rotation during forward reach for all subjects (Figure 2D, E and Table 2). The free model's correlations with direct measures for scapulothoracic posterior tilt ranged from moderate to strong; the rhythm model reported negative correlations for all subjects. Scapulothoracic kinematic errors were consistently larger for the rhythm model (Figure 2D–F and Table 2).

Discussion

Due to difficulties implementing unconstrained scapular kinematics, many upper extremity musculoskeletal models utilize a scapular rhythm to prescribe scapular motion. The results of this study show that the scapular rhythm employed by the MoBL ARMs model produced meaningful scapular kinematic errors ($>10^\circ$) for the majority of the abduction and forward reach. Scapulothoracic resultant angle differences were slightly larger during abduction than forward reach, suggesting that the model's scapular rhythm may not perform worse for motions that incorporate more scapulothoracic internal rotation and posterior tilt. Errors associated with the free model were within the measurement error of the acromion marker cluster (10°), thus demonstrating that faithful representation of scapular kinematics is possible within a modeling environment.

The rhythm model's errors generally increased with humerothoracic elevation during abduction with meaningful differences being present in all subjects at elevations greater than 44.8° . During forward reach, errors greater than 10° were present in all subjects by 39.9° of humerothoracic elevation. These findings indicate that use of models with a scapular rhythm should be limited to low levels of humerothoracic elevations for applications requiring precise scapulothoracic kinematics. However, utilization of these type of models even at low levels of elevation is still cautioned as some subjects achieved meaningful errors ($>10^\circ$) at as little as 28.3° and 18.2° of elevation during abduction and forward reach, respectively.

Strong correlations found for upward rotation were in agreement with de Groot and Brand's conclusion that the scapular rhythm's upward rotation was more strongly correlated with humeral elevation than other axes.⁵ Trends for internal rotation between the rhythm model and direct measures were strong during forward reach, but negative during abduction. Motion of the scapula is influenced by the contour of the thoracic cage, yet during abduction the scapula is free to lift off the thorax making internal rotation more independent. Previously published data also found that internal rotation is less predictable as a function of humeral elevation than the other scapulothoracic axes.⁵ Posterior tilt correlations for the rhythm model were generally weak during abduction and negative during forward reach. The rather minimal amount of motion about the posterior tilt axis maybe responsible for these weak correlations.

The RMSE values between the rhythm model and direct measures for upward rotation and posterior tilt were similar to those previously reported for the rhythm⁵; however the values for internal rotation were slightly larger than those previously reported. This may be due to the fact that the acromion marker cluster reference standard is less accurate at estimating scapulothoracic internal rotation than other axes.^{12,20}

The primary assumption in this investigation is that scapulothoracic kinematics estimated by the acromion marker cluster represent true scapular motion. In lieu of a clinically applicable gold standard, the acromion marker cluster is the recommended method to estimate scapular motion¹² and has been validated.^{11,21–25} Single calibration acromion marker cluster errors range from 3.5° to 9.2°,¹² and thus must be considered when interpreting the results of this study. The greatest acromion marker cluster errors occur at humerothoracic elevations above 90°–120°^{11,13,22,25} and during motions of humeral rotation.²⁶ To mitigate errors associated with acromion marker cluster, only single plane humerothoracic elevation motions were analyzed and predominantly constrained to elevations below 90°.

This study is one of the first^{27,28} to provide an analysis of scapular motion in musculoskeletal modeling and the first to quantify scapular kinematic errors associated with models utilizing a scapular rhythm. Results demonstrated that a model employing a scapular rhythm does not replicate scapular motion within a meaningful threshold of 10°. Errors associated with the free version of the model that permitted unconstrained, natural scapular motion were well within this 10° tolerance level demonstrating that accurate representation of scapular motion is achievable within a modeling environment. It remains unknown how the kinematic errors associated with the scapular rhythm affect model predicted muscle activation and force patterns. These findings can aid researchers in choosing which research questions are suitable for investigation with the models utilizing prescribed scapular kinematics and assist in contextualizing model results.

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Figure 1 —.

3D scapulothoracic resultant angle differences between each model, free scapula (FS, solid lines) and scapular rhythm (SR, dashed lines), and direct measures for all subjects during humerothoracic (HT) abduction (A) and forward reach (B). Differences $>10^\circ$ are considered meaningful.

Figure 2 —.
Comparisons between model-predicted and directly measured scapulothoracic kinematics on each axis for 1 representative subject during humerothoracic (HT) abduction (A, B, and C) and forward reach (D, E, and F).

Table 1

3D Scapulothoracic Resultant Angle Differences Between Each Model, Free Scapula (FS) and Scapular Rhythm (SR), and Direct Measures During Abduction and Forward Reach

Parameter	Abduction		Forward Reach	
	FS Model	SR Model	FS Model	SR Model
Max. angle difference (°)	8.0 (2.3)	22.4 (5.2) [*]	4.8 (1.3)	16.5 (3.2) [*]
Mean angle difference (°)	5.0 (1.1)	16.4 (4.1) [*]	2.4 (0.4)	11.1(0.4) [*]
% of trial with >10° differences	1.6 (3.6)	86.0 (15.7) [‡]	0.0 (0.0)	71.2(19.1) [‡]

Note. Data presented as group mean (SD).

^{*} Meaningful difference (> 10°).

[‡] Meaningful difference (> 10°) for > 50% of the motion trial.

Table 2

Scapulothoracic Angle Differences and Trend Comparisons Between Each Model, Free Scapula (FS) and Scapular Rhythm (SR), and Direct Measures During Abduction and Forward Reach

Parameter	Scapulothoracic Upward Rotation			Scapulothoracic Internal Rotation			Scapulothoracic Posterior Tilt		
	FS Model	SR Model		FS Model	SR Model		FS Model	SR Model	
Abduction									
Mean reference ROM (°)	16.4 (1.9)			7.9 (1.9)			3.7 (1.8)		
RMSE (°)	4.4 (0.9)	8.8 (3.5)		1.2 (0.4)	14.0 (3.2)		2.5 (1.2)	3.0 (1.2)	
RMSE (%ROM)	27.2 (4.9)	52.9 (20.0)		16.3 (4.1)	192.5 (88.5)		96.9 (83.9)	111.1 (80.0)	
Abs. max. error (°)	6.6 (1.7)	12.4 (4.2)		2.3 (0.5)	18.8 (4.1)		4.4 (2.1)	5.2 (1.6)	
Abs. max. error (%ROM)	39.9 (8.0)	75.8 (26.7)		30.6 (10.5)	251.5 (85.7)		169.0 (144.5)	185.9 (121.7)	
Pearson's <i>r</i> -value	0.99 (0.01)	0.99 (0.01)		0.90 (0.08)	-0.69 (0.37)		0.59 (0.79)	0.12 (0.90)	
Forward reach									
Mean reference ROM (°)	18.2 (2.9)			14.3 (3.5)			5.5 (3.2)		
RMSE (°)	1.8 (0.6)	6.7 (3.1)		1.6 (0.7)	5.4 (3.5)		0.8 (0.3)	5.7 (3.1)	
RMSE (%ROM)	9.6 (2.9)	39.2 (21.6)		12.7 (8.6)	36.7 (23.8)		19.3 (13.0)	117.7 (77.4)	
Abs. max. error (°)	3.2 (1.3)	11.8 (5.0)		3.3 (1.6)	8.3 (3.5)		1.6 (0.7)	9.9 (5.6)	
Abs. max. error (%ROM)	17.5 (7.1)	69.0 (35.7)		26.2 (20.2)	56.7 (20.6)		41.3 (31.8)	195.5 (104.2)	
Pearson's <i>r</i> -value	.98 (.05)	.95 (.05)		.98 (.02)	.98 (.02)		.86 (.17)	-.72 (.26)	

Note. Data presented as group mean (SD). Differences are reported in degrees and as a percentage of the total motion about each axis from direct measures. ROM = range of motion; RMSE = root mean square error. Mean reference ROM displays mean amount of motion on each axis as determined by direct measures