EVALUATION OF THE EFFECT OF SCAPULAR MOTION ON MUSCULOSKELETAL MODELING OF THE SHOULDER AND UPPER EXTREMITY

by

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A dissertation submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Doctor of Philosophy in Biomechanics and Movement Science

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ABSTRACT

Musculoskeletal modeling possesses the ability to provide information on physiological parameters that cannot be directly measured. However, the validity of the results must be assessed to ensure that the model is sufficiently robust to recreate the mechanics occurring in vivo. The most widely used model of the upper extremity, MoBL ARMS, possesses limitations in representing natural scapular kinematics. Motion capture offers reliable quantification of humerothoracic (HT) motion, but accurate measurements of scapulothoracic (ST) and glenohumeral (GH) contributions to dynamic HT motion are difficult to obtain. In an effort to circumvent this issue, the MoBL ARMS model prescribes ST kinematics that relate to HT motion via regression equations, creating a relationship referred to as scapular rhythm. It is unknown how well this model replicates natural scapular kinematics for motions that generally follow the rhythm, such as shoulder abduction, and those that do not, such as forward reach. Furthermore, it is unknown how scapular kinematic differences affect the torque demands on the GH joint and ultimately influence model-predicted muscle activations. The purpose of this study was to evaluate the validity of the MoBL ARMS model, and specifically, the validity of using prescribed scapular kinematics to replicate natural, unconstrained scapular motion. The expected outcomes include a clear understanding of 1) how well the scapular rhythm model can recreate natural ST motion during shoulder abduction and forward reach, 2) how kinematic differences affect GH torque demands, and 3) how kinematic differences influence model-predicted muscle activations. The orientations of the trunk, scapula, and upper

extremity segments of five healthy subjects were measured with motion capture during two motions—shoulder abduction and forward reach. Simultaneously, fine-wire and surface electromyography (EMG) was collected for 17 muscles of the shoulder. Simulations were run on the published and publicly available MoBL ARMS model that prescribes ST kinematics, and a modified version of the model that allows for natural, unprescribed scapular kinematics. Differences between the two versions of the model indicated how well prescribed kinematics followed natural scapular motion and how ST kinematic errors influenced GH joint torques. Finally, model-predicted muscle activations were compared with EMG. Results demonstrated that the scapular rhythm of the MoBL ARMS model is not capable of replicating ST kinematics to a clinical meaningful degree (within 10°). Despite the substantial kinematic errors, only modest errors in GH torque were observed. Model-predicted muscle activations displayed moderate agreement with EMG for on/off timing; however, correlations between the model's activations and EMG were poor. Agreement between the model and EMG was greater for prime movers than for stabilizing and inactive muscles. These results show that the model may be able to recreate the activations of muscles which drive a motion, but is not sensitive enough to reliably predict the activities of those with subtler functions. Agreement between the model-predicted muscle activations and EMG did not appear to be related to ST kinematic errors. Considering the lack of a clear link between ST kinematic error magnitude and improved activation agreement, it can be inferred that other factors play a substantial role in determining model results. These findings can aid researchers in choosing which future applications are suitable for investigation with the MoBL ARMS model and assist in interpretation of model results.

Chapter 1

INTRODUCTION

1.1 Background

Upper extremity function is important to activities of daily living and quality of life. Biomechanical analyses can provide valuable insight into how the upper extremity functions in both healthy and pathological populations. Measurement of motion of the upper extremity segments can provide detailed information about joint kinematics and net moments present at each joint. In addition, muscle activation patterns can also be detected, but information on individual muscle forces cannot be measured. Musculoskeletal modeling possesses the ability to estimate physiological parameters that cannot be directly measured[1,2]. By approximating bones as rigid bodies and muscles as force actuators, musculoskeletal models can computationally estimate muscle forces and activations necessary to recreate dynamic movements measured by motion capture. Additionally, by virtue of being a computer-based model, this type of analysis can be used to simulate various clinical interventions to predict outcomes before the intervention ever takes place[1]. Modeling has the potential to provide clinicians with a wealth of detailed patient-specific information that could be used to optimize treatment decisions and improve clinical outcomes.

To date, musculoskeletal models have primarily been created for the lower extremity to examine tasks such as gait, while models of the upper extremity are far less common. One factor contributing to this disparity is that the structural complexity of the shoulder joint makes it inherently difficult to accurately model. Large ranges of

motion in all three GH rotational DOFs present complications in calculating and expressing joint orientations using Euler angles. The abundance of small, stabilizing musculature surrounding the shoulder can be problematic to represent across wide ranges of motion. Moreover, maintaining physiological fidelity of the wrapping paths and moment arms of these muscles across the joint for the wide variety of possible upper extremity motions is extremely challenging.

Perhaps the most difficult aspect of modeling the shoulder lies in proper implementation of scapular kinematics. Motion capture using retroreflective markers placed on superficial anatomical landmarks is a non-invasive, clinically applicable procedure that can accurately quantify 3D kinematics for many body segments[3]. This technique is capable of measuring overall HT motion, but is unable to discern the GH and ST contributions due to independent movement of the scapula beneath the skin[4]. Several methods have been developed to estimate ST kinematics including tracking jigs[5], magnetic tracking sensors[6], ultrasound-based methods[4] or marker clusters fixed to the acromion process of the scapula[7]. One of the methods for estimating scapular motion established a set of regression equations that coupled all three ST rotational DOFs to measured HT elevation[8]. These regression equations were not subject-specific and were developed based on group tendencies of healthy young adults. Additionally, the positions used to develop the equations captured varying degrees of ST upward/downward rotation, but did not encompass a wide range of motion (ROM) for either ST internal/external rotation or ST anterior/posterior tilt.

Existing models of the shoulder and upper extremity[9–13] implement various ways to handle the issue of scapular motion, however the mostly widely available and utilized of these, the MoBL ARMS model[9,13], employs the scapular rhythm developed

by de Groot and Brand (2001)[8]. Since this rhythm prescribes scapular motion, ST muscle forces required to achieve the kinematics cannot be measured. Despite these limitations, previous research has used the MoBL ARMS model to provide valuable information on GH and distal upper extremity mechanics. The MoBL ARMS model has been used estimate the ability of different nerve transfer surgeries to restore force generating capacity at the shoulder[14] and contribution to functional motions of the upper extremity[15]. Additionally, the model has been applied to better understand the capability of two proposed mechanisms that may contribute to shoulder deformities in patients with brachial plexus birth palsy[16]. This study also specifically identified which muscles produced force vectors known to cause osseous deformity at the GH joint in the brachial plexus birth palsy population[16]. Building on this research, the MoBL ARMS model was employed to investigate the effect of specific scenarios of muscle imbalance and impaired muscle growth associated with C5-C6 nerve injuries in the brachial plexus birth palsy population on shoulder motion and GH joint forces[17].

Although these studies clearly demonstrate the immense utility of musculoskeletal modeling at the shoulder and upper extremity, they all assume normal scapular kinematics by utilizing a prescribed scapular rhythm. It is unknown how well a scapular rhythm based on group tendencies applies to every healthy individual or individual patients. Pathologies such as brachial plexus birth palsy have been associated with abnormal scapular kinematics and greater ST contributions to overall HT motion[18]. Consequently, a prescribed scapular rhythm based on healthy adults will likely not adequately represent the scapular motion of these patients. Lastly, the scapular rhythm employed in this model focused on ST upward/downward rotation and did not incorporate a wide ROM for either ST internal/external rotation or ST

anterior/posterior tilt. Motions such as shoulder abduction primarily involve ST upward rotation with little ST internal/external rotation, thus following the rhythm. However, motions such as forward reach possess a greater amount of ST internal rotation which the rhythm will likely have difficulty approximating. It is unknown how well this model tracks natural scapular kinematics for motions that generally follow the prescribed scapular rhythm, such as shoulder abduction, and those that do not, such as forward reach. Furthermore, it is unknown how differences in scapular kinematics affect the requisite torque demands (i.e. joint torques required for the model to maintain its posture at each time point throughout a motion – *see Glossary for full definition*) on the GH joint. Finally, it is unknown how kinematic and kinetic differences influence simulation results for model-predicted muscle activation patterns.

This study aims to address these unknowns by evaluating the effect of prescribed scapular kinematics on model results. Two versions of the MoBL ARMS model will be tested – the first prescribes scapular motion to a rhythm and the second removes this constraint to allow for natural, unconstrained scapular kinematics. Differences between the two versions of the model will serve to demonstrate how prescribed scapular kinematics influence model results. The expected outcomes of this study include a clear understanding of 1) how well the scapular rhythm model can track natural ST motion during two different motions, 2) how scapular kinematic differences affect GH torque demands on the model, and 3) how scapular kinematic differences influence model-predicted muscle activations.

1.2 Specific Aims & Hypotheses

1.2.1 Aim 1 – Scapulothoracic Kinematics

To assess the ability of a 3-dimensionsal (3D) musculoskeletal model that employs a prescribed scapular rhythm to track scapular kinematics during a motion that follows the scapular rhythm – shoulder abduction – and a motion that does not - forward reach

Hypothesis 1.1: Clinically meaningful differences (>10□) in ST angle will exist during both motions between a) a model that constrains the scapula to move according to a rhythm and b) the same model modified to allow for natural, unprescribed scapular motion. These differences are expected to be >10□ for over 50% of each motion trial.

Hypothesis 1.2: Waveform symmetry and correlation analyses will demonstrate strong shape and trend agreement (Symmetry index and Pearson r>0.7) between the ST angle curves of the two models for all three degrees of freedom (DOF) during both motions.

Hypothesis 1.3: The magnitude of the maximum and mean differences in ST angle between models for each DOF will be less than 50% of FS model's ST excursion on that DOF during each trial

Hypothesis 1.4: For all the kinematics parameters in Hypotheses 1.1, 1.2, and 1.3 the differences between the two models will be greater during forward reach than shoulder abduction

1.2.2 Aim 2 – Glenohumeral Kinetics

To utilize 3D musculoskeletal modeling to evaluate the effect of scapular kinematics on the GH torques required during shoulder abduction and forward reach

Hypothesis 2.1: There will be no significant differences in mean and maximum values of GH torques for any DOF between the two models during both motions. Additionally, for each of the 3 DOFs, the GH torques between models will be in functional agreement with each other (i.e. both models require torques of the same function about a DOF - see Glossary for full definition)

Hypothesis 2.2: Waveform symmetry and correlation analyses will demonstrate a strong shape and trend agreement (Symmetry index and Pearson r>0.7) between GH torques of the two models for all three DOFs during both motions

Hypothesis 2.3: For all kinetic parameters in Hypotheses 2.1 and 2.2, the differences between the two models will be greater during forward reach than shoulder abduction

1.2.3 Aim 3 – Muscle Activations

To compare muscle activation patterns predicted by a 3D musculoskeletal model that employs a scapular rhythm with recorded electromyography (EMG) during shoulder abduction and forward reach

Hypothesis 3.1: The on/off timing of the muscle activations predicted by the model will be in agreement with the recorded EMG for at least 50% of the trial duration for both motions trials

Hypothesis 3.2: Waveform symmetry and correlation analyses will demonstrate moderate or greater shape and trend agreement (Symmetry index and Pearson r>0.4) between model-predicted muscle activations and EMG during both motions

Hypothesis 3.3: For all muscle activation parameters in Hypotheses 3.1 and 3.2, the differences between the two models will be greater during forward reach than shoulder abduction

1.3 Innovation

This study will assess the validity of applying a musculoskeletal model that employs a prescribed scapular rhythm to upper extremity motions with natural, unconstrained scapular motion. The knowledge gained from this work will describe the magnitude and characteristics of ST kinematic errors during motions with that generally follow a scapular rhythm, such as shoulder abduction, and those that do not, such as forward reach. Moreover, this study will provide an understanding of the degree to which kinematic errors impact the accuracy of model-predicted muscle activation patterns which will aid researchers in contextualizing model results and in choosing which future applications are suitable for investigation with a model utilizing a scapular rhythm.

Chapter 2

METHODS

2.1 Subjects

Five healthy adult subjects were recruited for this study. All subjects were between the ages of 18 and 65 and had no history of shoulder injury in the investigated limb. Prior to data collection, informed consent was obtained from all subjects. This study has been approved by the University of Delaware Institutional Review Board (Appendix B).

2.2 Data Collection

Motion capture and EMG was simultaneously recorded. Fourteen fine-wire electrodes were inserted into 12 muscles of the shoulder and upper extremity by a trained physician (Table 1). All fine-wire electrodes were placed under aseptic conditions with a sterilized hypodermic needle into muscle locations specified by Leis and Schenk et al.[19]. The fine-wire electrodes were bipolar, nylon-coated, stainless steel wires with 2mm of exposed sensors approximately 0.3-0.5mm apart. A muscle stimulation unit and visual signal inspection were used to confirm accurate placement of intramuscular leads. Fine-wire EMG data was recorded at 1920Hz with a Konigsberg system (Konigsberg Instruments Inc., Monrovia, CA). An additional six surface electrodes were placed on five superficial muscles (Table 1). Leads were positioned over the center of each muscle belly after shaving and cleaning the skin with isopropyl alcohol; a ground lead was affixed over the sternum. The surface

electrodes used were 99.9% pure silver, 5x1mm bars with 10mm between leads. Surface EMG data was recorded simultaneously at 1920Hz with a wireless Delsys Trigno system (Delsys Inc., Natick, MA).

Table 1 – Muscles recorded with fine-wire and surface EMG.

Fine-wire EMG Muscles		**Surface EMG Muscles**
Upper Trapezius	Supraspinatus	Anterior Deltoid
Middle Trapezius	Infraspinatus	Posterior Deltoid
Lower Trapezius	Teres Minor	Pectoralis Major
Rhomboid Major	Subscapularis	Latissimus Dorsi
Rhomboid Minor	Teres Major	Biceps Brachii
Levator Scapulae	Pectoralis Minor	Triceps Brachii
Serratus Anterior	Coracobrachialis	

Dynamic maximal voluntary contractions (dMVC) were collected for each lead[20,21]. The dMVC trial for each muscle began with the subject's arm in a neutral position. Investigators instructed the subjects to move their arm to perform the motion associated with the muscle of interest. Prior to recording each dMVC trial, the subjects were allowed to practice the dMVC motion to ensure that the desired muscle activation was being achieved and to allow for visual inspection and modification of the signals to confirm appropriate amplification without clipping. The investigator applied manual resistance to the subject's arm to elicit maximal effort throughout the entire motion. Manual resistance was preferred for this study as it allowed the investigator to adjust the level of resistance so that a full ROM could be completed. For each recorded dMVC trial, the subject was instructed to "perform the motion as quickly and forcefully as possible". Following placement of all EMG leads and dMVC testing, 6mm retroreflective markers were positioned over the anatomical landmarks

listed in Table 2. An acromion marker cluster (AMC)[7] was affixed to the acromion process of the scapula with the central marker of the triad positioned directly over the landmark. The AMC was used to estimate scapular kinematics during both motion trials[7]. Figure 1 displays the experimental setup.

Table 2 – Marker locations for motions capture.

T1 Spinous Process	Medial Epicondyle of Humerus
T8 Spinous Process	Lateral Epicondyle of Humerus
Manubrium	Radial Styloid Process
Acromion Process	Ulnar Styloid Process
Trigonum Spinae	Dorsum of 3rd Metacarpal Head
Inferior Angle of Scapula	

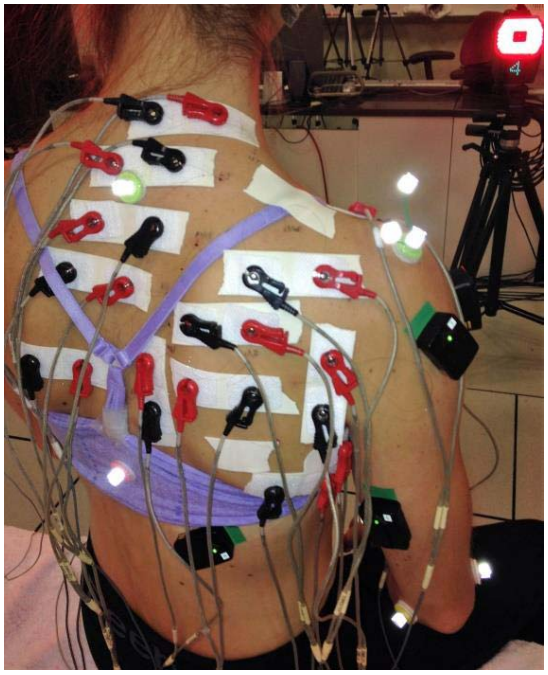

Figure 1 – Experimental setup with EMG electrodes and motion capture.

A 12 camera motion capture system (Motion Analysis Corp., Santa Rosa, CA) collecting at 60Hz recorded the 3D marker positions during a static calibration trial and both motion trials – shoulder abduction and forward reach. For the static trials, subjects placed their arms in a neutral position while a trained physician palpated the trigonum spinae and inferior angle of the scapula and placed two additional 6mm, 2D retroreflective markers on these landmarks; these markers were removed for dynamic trials. For shoulder abduction trials, the subjects began in a neutral position, moved to maximum abduction in the frontal plane, then returned to neutral along the same path. For forward reach trials, subjects started in a neutral position, moved to maximum forward reach while staying mainly in the sagittal plane, then returned to neutral along the same path. A metronome was used to pace the speed of each motion to minimize the effect of muscle contraction acceleration changes on EMG signals.

2.3 Data Processing

All EMG signals had the direct current bias removed and were band-pass filtered from 10-760Hz using a zero phase-shift, 4^{th} order Butterworth filter; an additional 60Hz band-stop filter was applied to mitigate electrical noise. Additionally, signals were full-wave rectified and enveloped by a low-pass, zero phase-shift, 4^{th} order Butterworth filter with a 4Hz cutoff frequency. The peak value of the signal within each muscle's dMVC trial was used for normalization of the respective signals during the abduction and forward reach motion trials. Finally, all dynamic EMG signals were re-sampled to 1,000 data points spanning the duration of each motion trial for final analysis. EMG processing was performed using custom software in LabVIEW 12 (National Instruments, Austin, TX).

In the static calibration trial, the relationship between the location of the trigonum spinae and inferior angle markers relative to the orientation of the AMC was ascertained. The trigonum spinae and inferior angle marker positions were virtually recreated using Cortex 2.5 (Motion Analysis Corp., Santa Rosa, CA) based on the orientation of the AMC throughout both motion trials.

All simulations in this study were performed on the MoBL ARMS dynamic musculoskeletal model of the upper extremity[13] (Figure 2) using the OpenSim 3.1 software package[1]. The MoBL ARMS model contained a thorax, clavicle, scapula, humerus, radius, ulna, as well as bones of the wrist and hand. All segments lengths were based on a 50th percentile male[22,23]. The joints of the model permitted six DOF of the thorax, two rotational DOF at the sternoclavicular joint, three rotational DOF at the acromioclavicular joint, three rotational DOF at the GH joint, one rotational DOF at the elbow, forearm pronation/supination, as well as wrist flexion/extension and deviation. The three DOF of GH motion was represented using the following convention: 1) plane of elevation, 2) elevation, and 3) internal/external rotation. Despite possessing two rotational DOFs at the sternoclavicular joint and three rotational DOFs at the acromioclavicular joint, motion of the shoulder girdle was completely determined by a set of regression equations known as a scapular rhythm[8]. This scapular rhythm constrained the scapula and clavicle to move in a prescribed pattern based on one independent DOF – GH elevation.

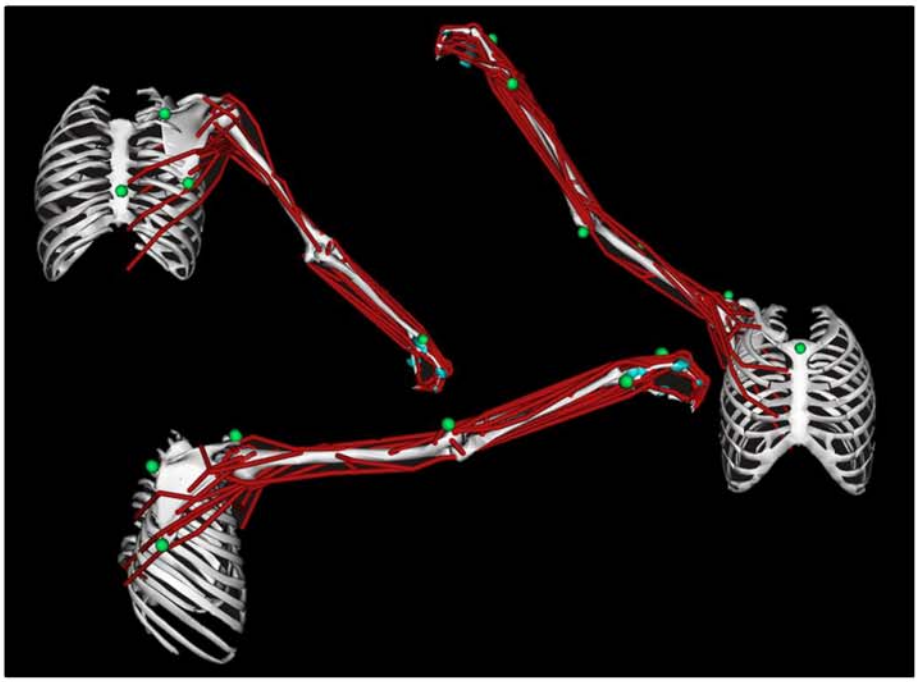

Figure 2 – The MoBL ARMS musculoskeletal model of the shoulder and upper extremity. Shown positioned in (A) early shoulder abduction, (B) full shoulder abduction, and (C) full forward reach.

Two versions of the MoBL ARMS dynamic musculoskeletal model of the upper extremity[13] were investigated in this study. The first version was the scapular rhythm (SR) model which is the default MoBL ARMS model, thus it constrained scapular kinematics to depend on the rhythm described above. The second version of the model had a free scapula (FS) and allowed for unprescribed ST kinematics. In order for the MoBL ARMS model to permit unprescribed scapular motion the coordinate coupler constraints on the sternoclavicular and acromioclavicular joints were removed. The result was two models – one that prescribed scapular kinematics, SR, and one that allowed for unprescribed or free scapular kinematics, FS. Both the SR and FS versions of the MoBL ARMS model possessed fifty Schutte-type muscle-

tendon actuators which represented 32 muscles of the upper extremity[24]. Each actuator was scaled by optimal fiber length, peak isometric force, pennation angle, and tendon slack length[25]. All values for optimal fiber length and pennation angle were derived from anatomical studies as outlined in Holzbaur et al. (2005)[9]. Tendon slack lengths were calculated to align with operating fiber lengths for muscle fibers[26] or were based on active and passive moment measurements from human subjects[9]. Peak isometric force for each muscle-tendon actuator were based on muscle volume and isometric strength data from healthy adults[27,28]. All the paths of each muscle-tendon actuator in the model produced muscle moment arms that were similar to published cadaveric data[29–36].

Model segments were scaled based on the marker positions when each subject was in a resting neutral position. Model segments were also scaled based on subject mass. The Inverse Kinematics tool within OpenSim was used to calculate model joint kinematics from the motion capture marker data. All shoulder abduction inverse kinematics trials began with the arm as close to the neutral starting position without placing the model's shoulder joint in a gimbal lock configuration. All shoulder abduction inverse kinematics trials ended at maximum HT abduction. All forward reach inverse kinematics trials began with the arm as close to the neutral starting position without placing the model's shoulder joint in a gimbal lock configuration and ended at maximum forward reach position. For both motions, the FS model used the acromion process, trigonum spinae, and inferior angle markers to calculate ST joint angles, whereas the SR model let the model's scapular rhythm coupling constraints determine ST joint angles based on humeral elevation. In order to determine anatomical joint orientations (i.e. not model joint orientations), OpenSim's Body

Kinematics Analysis tool was ran on the inverse kinematics results. This analysis output XYZ Euler angles which described the 3D orientation of each body segment relative to the ground coordinate system. These XYZ Euler angles were used to construct rotation matrices for the thorax, scapula and humerus segments which were then used to calculate the anatomical HT, GH, and ST joint angles.

Using the inverse kinematics results, the Inverse Dynamics tool within OpenSim[1] was used to compute the requisite GH joint torques during both motions. For more intuitive interpretation, the GH torques were transformed and expressed as such: GH abduction/adduction torque reported about the orientation of the thorax's anterior/posterior axis, GH flexion/extension torques reported about the orientation of the thorax's medial/lateral axis, and GH internal/external rotation torques reported about the long axis of the humerus.

Finally, the Computed Muscle Control (CMC) algorithm[37] within OpenSim[1] was used to estimate muscle activation patterns during the inverse kinematics results for both motions. Minimization of the sum of the squared muscle activations was used as a CMC cost function. For this analysis, a third version of the MoBL ARMS model was used. This model version was packaged with the official MoBL ARMS tutorial had undergone several simplifications to make it suitable for CMC. These simplifications included: reduced degrees of freedom at the thorax, locked degrees of freedom for pronation/supination, wrist flexion/extension, and wrist deviation. Due to complications scaling wrap objects and muscle paths, the model was not able to be scaled to each subject's individual segment size, thus default model size was used for all subjects. However, the model was scaled to each subject's mass. Additionally, due to the lack of ST musculature, the ST joint was constrained to move according to the

same scapular rhythm in the SR model. Lastly, the musculature distal to the elbow, which is not of interest for this study, was removed in order to allow for faster simulation times. The model-estimated muscle activation results from CMC were band-pass filtered from 10-760Hz using a zero phase-shift, 4^{th} order Butterworth filter and then enveloped with a 4^{th} order Butterworth filter with a cutoff of 4Hz.The muscle activations were then re-sampled to 1,000 data points spanning the duration of both motion trials for analysis.

2.4 Data Analyses

Aim 1: Model-estimated ST and GH kinematics were compared to directly measured kinematics to evaluate how well each model replicated natural scapular and humeral motion. To evaluate the magnitude of the differences in ST kinematics between models, ST angle differences >10° were considered clinically meaningful. This criterion is based on the opinions of several physicians with expertise in the shoulder and upper extremity. Both the maximum and mean ST angle difference between the SR and FS models were assessed for each motion trial. Additionally, the percentage of the trial with >10° ST angle difference was calculated and evaluated. For the 3 individual ST DOFs, the magnitude of the difference between models was investigated. For each DOF, differences >50% of FS model's ST excursion on that DOF during each trial were considered significant.

Waveform symmetry analysis[38] and Pearson's correlation analysis assessed the trend and shape of ST kinematics between the SR and FS models. Both of these analyses were used to compare the ST angle components resolved about the X, Y, and Z axes where X=upward/downward rotation, Y=internal/external rotation, Z=anterior/posterior tilt.

Waveform symmetry analysis is a technique that utilizes eigenvectors to compare the two waveforms. This analysis accounts for four measures of symmetry and normalcy: trend phase, trend symmetry/normalcy, range amplitude ratio, and range offset. Waveform symmetry ratio was the value used to assess similarities and differences between two curves. Waveform symmetry ratio can range from -1 to 1, with a value of 1 indicating identical curves, a value of 0 signifying no trend similarities between curves, and a value of -1 representing a completely opposite trend between curves.

Pearson's r-values between the SR and FS curves for each ST DOF were calculated and interpreted according to the recommendations of Dancey and Reidy[39]. By this categorization, an r-value of between 0.9 to 1 represents a very strong correlation, between 0.7 to 0.89 is a strong correlation, between 0.4 to 0.69 signifies moderate correlation, and <0.39 are considered weak correlations. Finally, all of the results for the parameters described above were compared between the two motions. All kinematic analyses were within-subject; however inter-subject variability was investigated. Both individual and group kinematic parameters were investigated.

Aim 2: Model-estimated torques were compared to directly calculated GH torques to evaluate how well each model replicated the mechanics that occurred in vivo. The magnitudes of the difference in requisite GH torques between models were evaluated qualitatively, due to lack of consensus for a clinically meaningful value. Mean and maximum differences were compared between the SR and FS models for GH torques about each of the three axes; GH torque differences were expressed as a percentage of the maximum value of the FS model within each trial. In addition to investigating mean and maximum GH torque differences between models, function

was also considered. If both the SR and FS models required a torque of the same function (e.g. GH abduction torque needed for both models) for a given GH DOF they were considered to be in agreement. However, if the function of the GH torques between models were not the same (e.g. SR model requires GH abduction torque while FS model requires GH adduction torque), they were considered to be in disagreement. Similar to Aim 1, Waveform symmetry ratio and Pearson's r-values were used to quantify the trend and shape of GH torque patterns about each axis between the SR and FS models. Lastly, all of the results for the parameters described above were compared between the two motions. All kinetic analyses were within-subject; however inter-subject variability was investigated. Both individual and group kinetic parameters were investigated.

Aim 3: Model-predicted muscle on/off activation timings were compared with EMG for each of the 12 muscles common to both the model and the EMG. Timing agreement between the model and EMG was defined as the duration of each motion trial when model-predicted activations and EMG were either both on or both off (*see Glossary*). Complete agreement throughout the trial yields a value of 100% whereas no agreement yields a value of 0%. This methodology has been used effectively in previous modeling research[40]. For the comparison of on/off activation timing muscles were classified and grouped based on their functional role during each motion. Muscles responsible for the causing the motion were prime movers, non-prime mover muscles which were on for >20% of the trial were classified as stabilizers, and muscles which were on for <20% of the trial were deemed inactive. Waveform symmetry ratio and Pearson's r-value were used to quantify the trend and shape of individual muscle activations between the model-predicted activations and EMG

normalized to dMVC. Muscles exhibiting minimal EMG activity (active less than 20% of the movement) did not undergo waveform symmetry and correlation analyses. Lastly activation timing agreement and trend analysis results were compared between motions. All muscle activation analyses were within-subject; however inter-subject variability was investigated. Both individual and group muscle activation parameters were investigated.

Chapter 3

RESULTS

3.1 Model Scaling

Model scaling errors were slightly larger than the OpenSim Recommended Best Practices for scaling[41] (maximum marker error<2cm; RMSE<1cm) for some subjects' models. The scaling errors for the SR models were marginally greater than those of the FS models for all subjects. On average the SR models had 0.49cm larger scaling RMSEs and 0.67cm larger maximum marker error than the respective FS model. The markers with the largest errors were almost exclusively located on the hand and wrist for all subjects.

3.2 Aim 1: Scapulothoracic Kinematics

3.2.1 Abduction

Inverse Kinematic Errors: Inverse kinematic errors for both the SR and FS models were nearly within the OpenSim Recommended Best Practices[41] (RMSE<2cm; maximum marker error within 2-4cm) for all subjects. It should be noted that the OpenSim Recommended Best Practices were developed based on lower extremity data, thus it is unknown how well results from the upper extremity should adhere to these guidelines. On average the SR models had 0.83cm larger Inverse Kinematics RMSEs and 0.79cm larger maximum marker error than the respective FS

model. The markers with the largest errors were almost exclusively located on the hand and wrist for all subjects.

Comparison with Direct Measures: A comparison between model-estimated and directly measured ST angles revealed no clinically meaningful differences between the FS model and direct measures during shoulder abduction. Additionally, the FS model followed the shape trends of natural ST kinematics. The SR model possessed significant differences from the direct measures, especially at upper levels of humeral elevation. The linear prediction of the SR model for ST kinematics did not capture the nuances of natural ST kinematics. The FS and SR models of all subjects displayed similar behavior when compared to directly measured ST angles. Data from a representative subject is shown in Figures 3, 4, and 5.

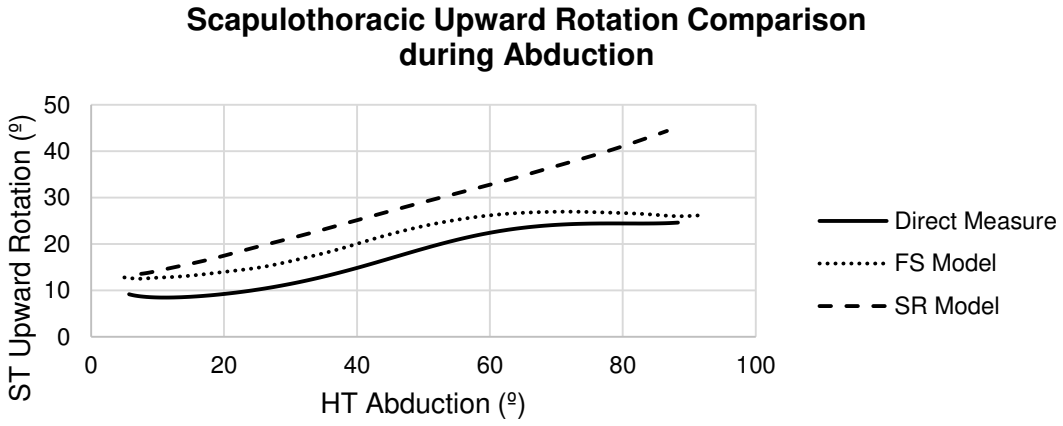

Figure 3 – Comparison of model-estimated scapulothoracic upward rotation with directly measured kinematics during shoulder abduction for a representative subject. All subjects' models displayed similar agreement with direct measures.

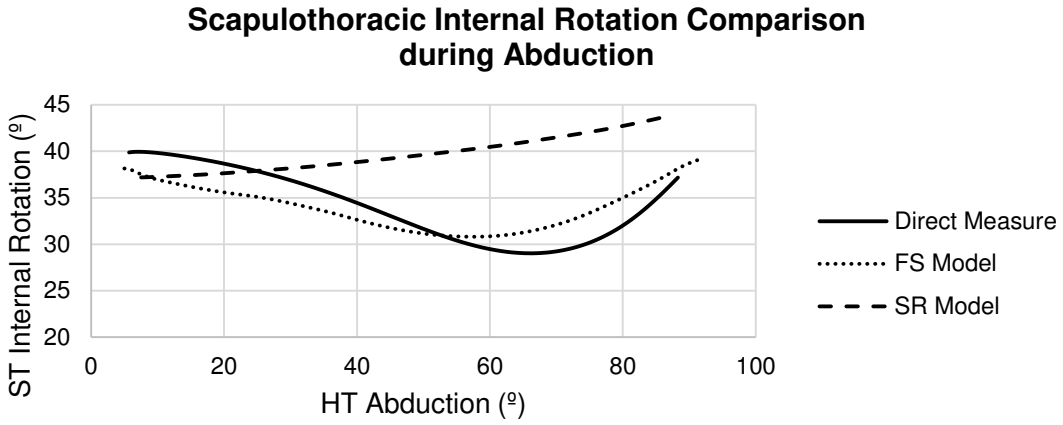

Figure 4 – Comparison of model-estimated scapulothoracic internal rotation with directly measured kinematics during shoulder abduction for a representative subject. All subjects' models displayed similar agreement with direct measures.

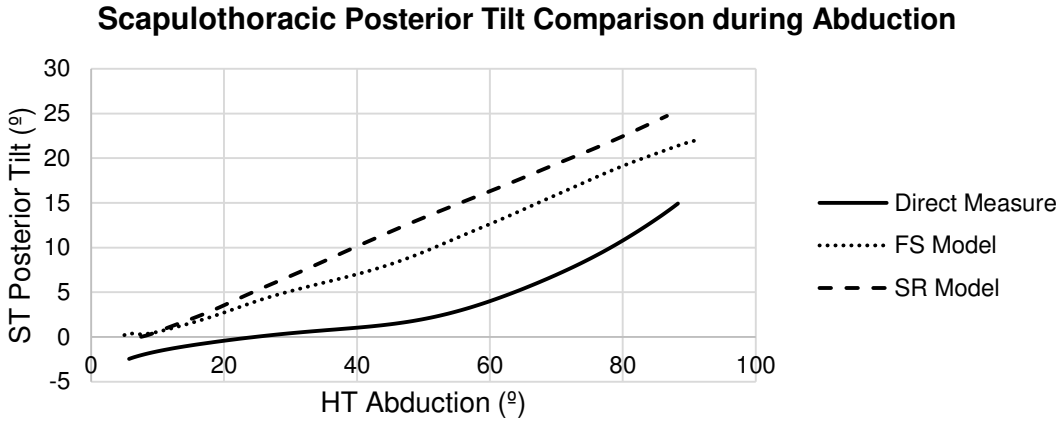

Figure 5 – Comparison of model-estimated scapulothoracic posterior tilt with directly measured kinematics during shoulder abduction for a representative subject. All subjects' models displayed similar agreement with direct measures.

Model-estimated GH kinematics were also compared with direct measures and revealed similar results as the ST kinematics comparison, however the differences between both models and the direct measures were more pronounced on the GH internal rotation and flexion DOFs. The more substantial differences on GH internal rotation and flexion DOFs may affect the ability of the models to accurately estimate GH joint torques. The FS and SR models of all subjects displayed similar behavior when compared to directly measured ST angles. Data from a representative subject is shown in Figures 6, 7, and 8.

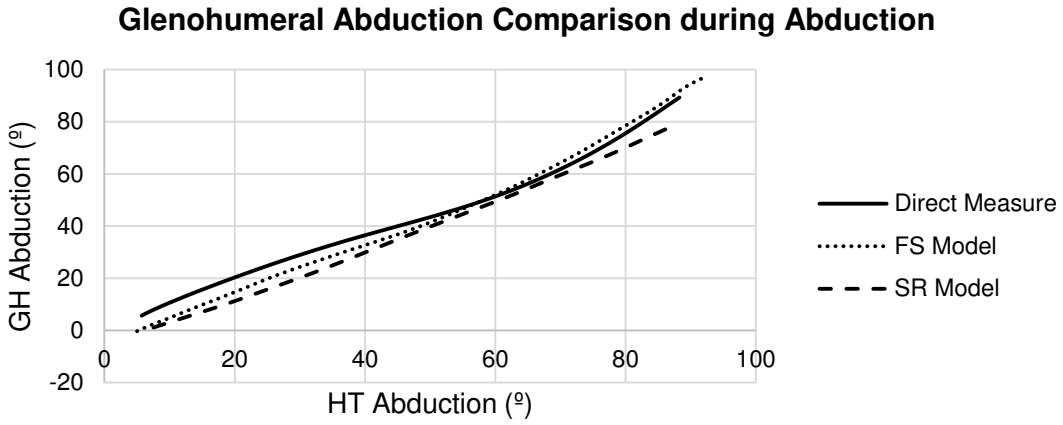

Figure 6 – Comparison of model-estimated glenohumeral abduction with directly measured kinematics during shoulder abduction for a representative subject. All subjects' models displayed similar agreement with direct measures.

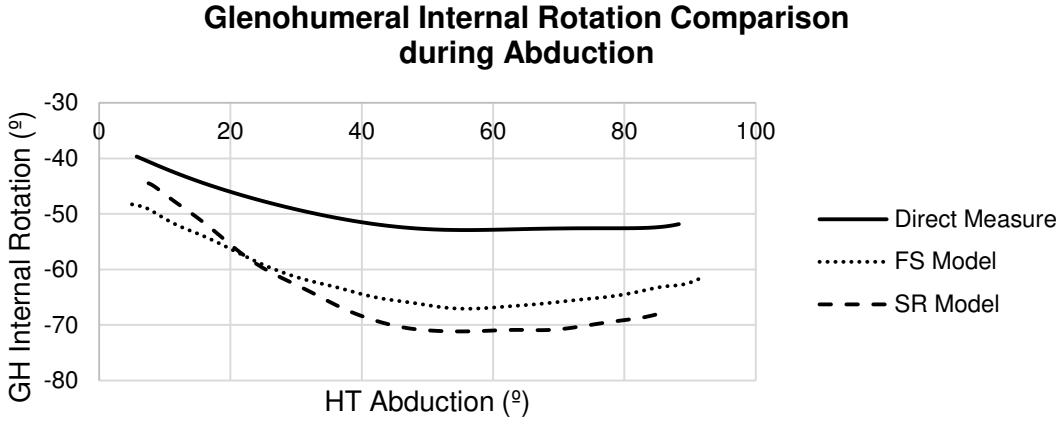

Figure 7 – Comparison of model-estimated glenohumeral internal rotation with directly measured kinematics during shoulder abduction for a representative subject. All subjects' models displayed similar agreement with direct measures.

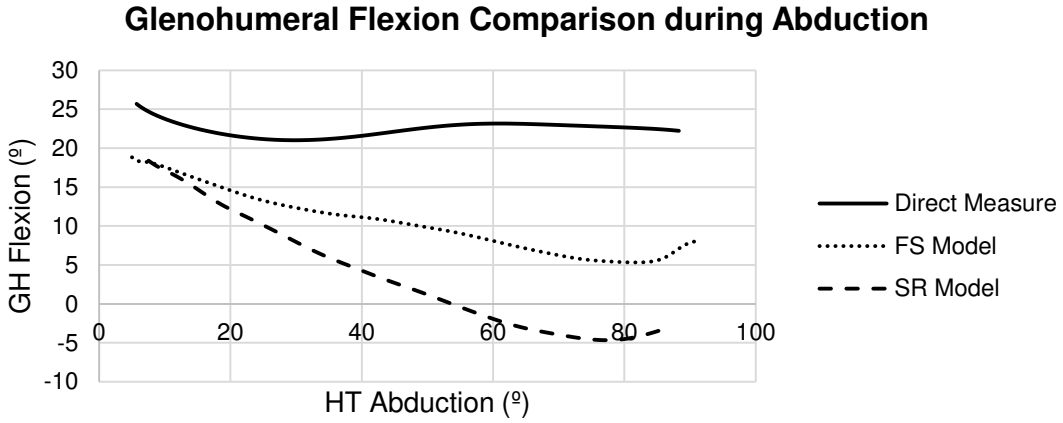

Figure 8 – Comparison of model-estimated glenohumeral flexion with directly measured kinematics during shoulder abduction for a representative subject. All subjects' models displayed similar agreement with direct measures.

ST Angle Difference: Clinically meaningful differences (>10°) in 3D ST resultant angle between the SR and FS models were observed for all five subjects during shoulder abduction (Figure 9). The differences in 3D ST resultant angle generally increased with greater levels of HT abduction for all subjects. The maximum differences in 3D ST resultant angle ranged from 11.11° to 34.69° between subjects, while mean differences fell between 8.80° and 18.48°. The percentage of the abduction trial with differences above 10° varied from 46% to 100% between subjects. Averaged across all subjects, the maximum and mean differences in 3D ST resultant angles were 24.61° and 14.25°, respectively. Across all subjects, the average percentage of the abduction trial with differences >10° was 74.80%. A full summary of 3D ST resultant angle differences can be found in Table 3.

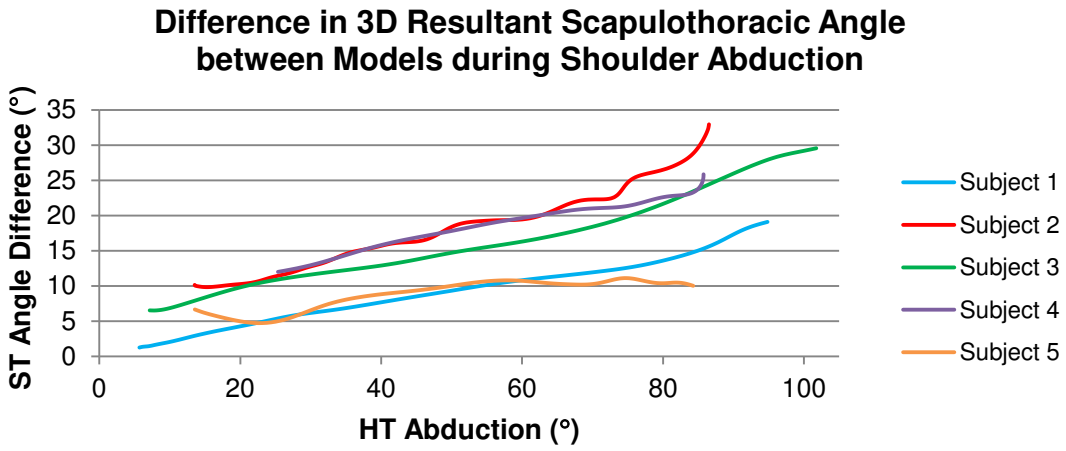

Figure 9 – Differences in 3D resultant scapulothoracic angle between the scapular rhythm (SR) and free scapula (FS) models during shoulder abduction. Differences >10° are clinically meaningful.

Table 3 – Mean and maximum 3D resultant scapulothoracic angle differences between the scapular rhythm (SR) and free scapula (FS) models and percentage of the trial with differences >10° for all subjects during shoulder abduction.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Max Diff. (°)	19.10	34.69	29.56	28.61	11.11	24.61
Mean Diff. (°)	9.29	18.10	16.56	18.48	8.80	14.25
% of Trial >10°	46.00	95.00	85.00	100.00	48.00	74.80

ST Upward Rotation: The maximum ST upward rotation difference between the FS and SR models ranged from 8.43° to 19.35° between subjects. To aid in interpretation of the results, kinematic differences were also expressed as a percentage of the FS model's ST upward rotation excursion throughout the trial; this will be referred to as reference ROM. The maximum differences in ST upward rotation were between 32.55% and 120.77%. The mean differences in ST upward rotation between models for all subjects ranged from 3.53° to 12.30° or 17.53% and 65.18% of each subject's reference ROM. For maximum ST upward rotation difference, 3 of 5 subjects had differences that were greater than 50% of their reference ROM. For mean ST upward rotation difference, 1 of 5 subjects had a difference that was greater than 50% of their reference ROM. Averaged across all subjects, the maximum and mean differences in ST upward rotation were 14.62° (77.94% ROM) and 6.34° (32.95% ROM), respectively. A full summary of ST upward rotation differences between models is presented in Table 4.

Table 4 – Mean and maximum scapulothoracic upward rotation angle differences between the scapular rhythm (SR) and free scapula (FS) models for all subjects during shoulder abduction.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Reference ROM (°)	14.78	20.13	18.87	25.90	21.16	20.17
Max Diff. (°)	17.85	19.35	18.25	8.43	9.21	14.62
Max Diff. (% of ROM)	120.77	96.13	96.71	32.55	43.53	77.94
Mean Diff. (°)	6.06	3.53	12.30	6.08	3.71	6.34
Mean Diff. (% of ROM)	41.00	17.54	65.18	23.47	17.53	32.95

Comparison of the ST upward rotation trends between the FS and SR models during abduction indicated strong to very strong trend similarities (Table 5). A full summary of the trend analysis for all three ST DOFs during abduction is presented in Table 5. Data for the subjects with the strongest and weakest trends for ST upward rotation are displayed in Figure 10.

Table 5 – Kinematic trend analysis for scapulothoracic upward rotation, internal rotation, and anterior tilt during shoulder abduction.

	ST Upward Rotation		ST Internal Rotation		ST Anterior Tilt	
	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R
Subject 1	0.964	0.936	-0.315	0.076	0.969	0.996
Subject 2	0.942	0.873	0.618	0.452	0.983	0.715
Subject 3	0.958	0.989	-0.933	-0.954	0.972	0.969
Subject 4	0.930	0.984	-0.865	-0.848	0.963	0.989
Subject 5	0.972	0.993	0.827	0.778	0.955	0.980

Negative	Weak	Moderate	Strong	Very Strong
<0	0 - 0.39	0.4 – 0.69	0.7 – 0.89	0.9 - 1

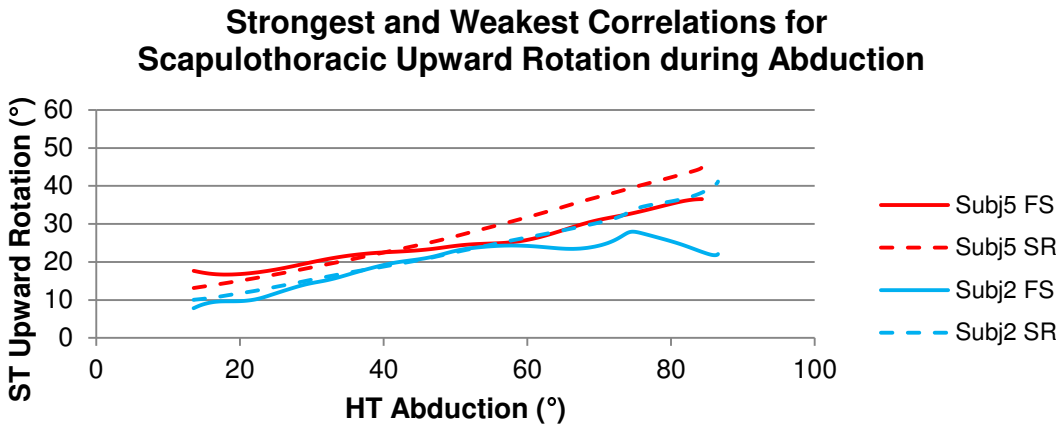

Figure 10 – Subjects with the strongest (r = 0.993) and weakest (r = 0.873) correlations for scapulothoracic upward rotation between their scapular rhythm (SR) and free scapula (FS) models during shoulder abduction.

ST Internal Rotation: The maximum ST internal rotation difference between the FS and SR models ranged from 9.64° to 27.02° between subjects. To aid in interpretation of the results, kinematic differences were also expressed a percentage of the FS model's ST internal rotation excursion throughout the trial; this will be referred to as reference ROM. The maximum differences in ST internal rotation were between 66.94% and 352.28%. The mean differences in ST internal rotation between models for all subjects were between 6.12° and 17.15° or 48.26% and 223.60% of each subject's reference ROM. For maximum ST internal rotation difference, all subjects had differences that were greater than 50% of their reference ROM. For mean ST internal rotation difference, 4 of 5 subjects had a difference that was greater than 50% of their reference ROM. Averaged across all subjects, the maximum and mean differences in ST internal rotation were 16.82° (189.81% ROM) and 10.93° (122.06%

ROM), respectively. A full summary of ST internal rotation differences between models is presented in Table 6.

Table 6 – Mean and maximum scapulothoracic internal rotation angle differences between the scapular rhythm (SR) and free scapula (FS) models for all subjects during shoulder abduction.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Reference ROM (°)	8.48	9.45	8.79	7.67	14.40	9.76
Max Diff. (°)	9.83	17.86	19.77	27.02	9.64	16.82
Max Diff. (% of ROM)	115.92	188.99	224.91	352.28	66.94	189.81
Mean Diff. (°)	6.12	14.99	9.46	17.15	6.95	10.93
Mean Diff. (% of ROM)	72.17	158.62	107.62	223.60	48.26	122.06

Comparison of the ST internal rotation trends between the FS and SR models during abduction indicated a range of correlation strengths (Table 5). Three subjects demonstrated weak or negative correlations values for both symmetry ratio and Pearson's r-value for ST internal rotation. The remaining two subject possessed moderate and strong symmetry ratios and Pearson's correlations. Data for the subjects with the strongest and weakest trends for ST internal rotation are displayed in Figure 11.

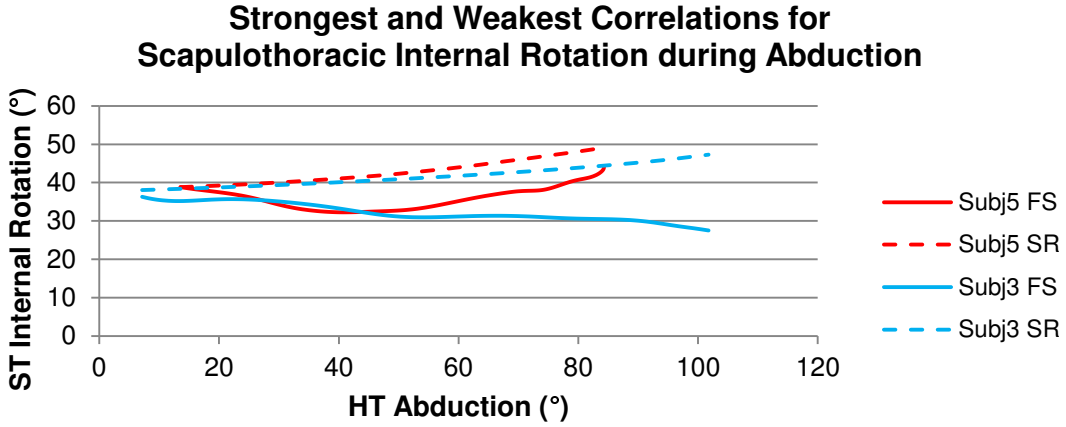

Figure 11 – Subjects with the strongest (r = 0.778) and weakest (r = -0.954) correlations for scapulothoracic internal rotation between their scapular rhythm (SR) and free scapula (FS) models during shoulder abduction.

ST Anterior Tilt: The maximum ST anterior tilt difference between the FS and SR models ranged from 3.24° to 24.45° between subjects. To aid in interpretation of the results, kinematic differences were also expressed a percentage of the FS model's ST anterior tilt excursion throughout the trial; this will be referred to as reference ROM. The maximum differences in ST anterior tilt were 14.56% to 467.50%. The mean differences in ST anterior tilt between models for all subjects were between 2.19° and 7.52° or 8.13% and 143.79% of each subject's reference ROM. For maximum ST anterior tilt difference, 3 of 5 subjects had differences that were greater than 50% of their reference ROM. For mean ST anterior tilt difference, 1 of 5 subjects had a difference that was greater than 50% of their reference ROM. Averaged across all subjects, the maximum and mean differences in ST anterior tilt were 10.89° (138.14% ROM) and 3.82° (45.29% ROM), respectively. A full summary of ST anterior tilt differences between models is presented in Table 7.

Table 7 – Mean and maximum scapulothoracic anterior tilt angle differences between the scapular rhythm (SR) and free scapula (FS) models for all subjects during shoulder abduction.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Reference ROM (°)	22.25	5.23	11.72	11.51	23.12	14.77
Max Diff. (°)	3.24	24.45	12.97	8.78	5.01	10.89
Max Diff. (% of ROM)	14.56	467.50	110.67	76.28	21.67	138.14
Mean Diff. (°)	2.19	7.52	4.74	2.79	1.88	3.82
Mean Diff. (% of ROM)	9.84	143.79	40.44	24.24	8.13	45.29

Comparison of the ST anterior tilt trends between the FS and SR models during abduction indicated strong to very strong trend similarities (Table 5). Data for the subjects with the strongest and weakest trends for ST anterior tilt are displayed in Figure 12.

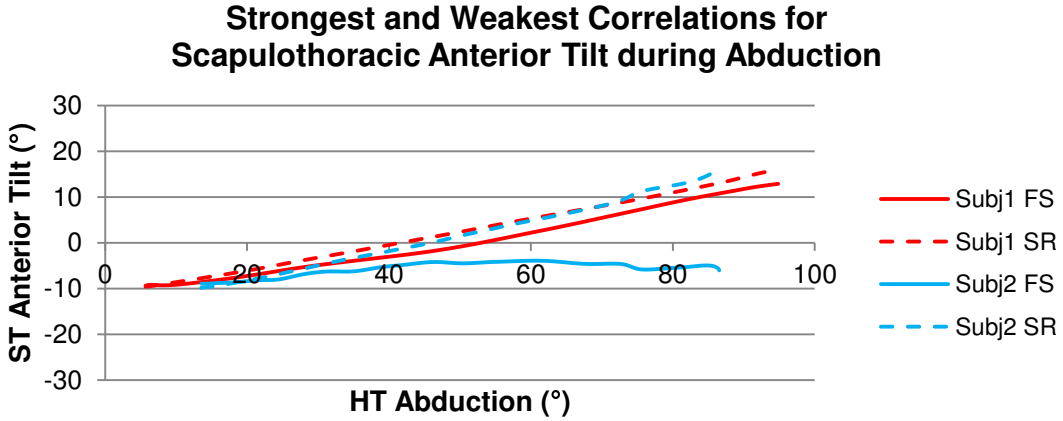

Figure 12 – Subjects with the strongest (r = 0.996) and weakest (r = 0.715) correlations for scapulothoracic anterior tilt between their scapular rhythm (SR) and free scapula (FS) models during shoulder abduction.

3.2.2 Forward Reach

Inverse Kinematic Errors: Inverse kinematic errors for both the SR and FS models were nearly within the OpenSim Recommended Best Practices[41] (RMSE<2cm; maximum marker error within 2-4cm) for all subjects. On average the SR models had 0.34cm larger Inverse Kinematics RMSEs and 0.02cm larger maximum marker error than the respective FS model. The markers with the largest errors were almost exclusively located on the hand and wrist for all subjects.

Comparison with Direct Measures: A comparison between model-estimated and directly measured ST angles revealed no clinically meaningful differences between the FS model and direct measures during forward reach. Additionally, the FS model followed the shape trends of natural ST kinematics. The SR model possessed significant differences from the direct measures, especially at upper levels of humeral elevation. The linear prediction of the SR model for ST kinematics did not capture the nuances of natural ST kinematics. The FS and SR models of all subjects displayed similar behavior when compared to directly measured ST angles. Data from a representative subject is shown in Figures 13, 14, and 15.

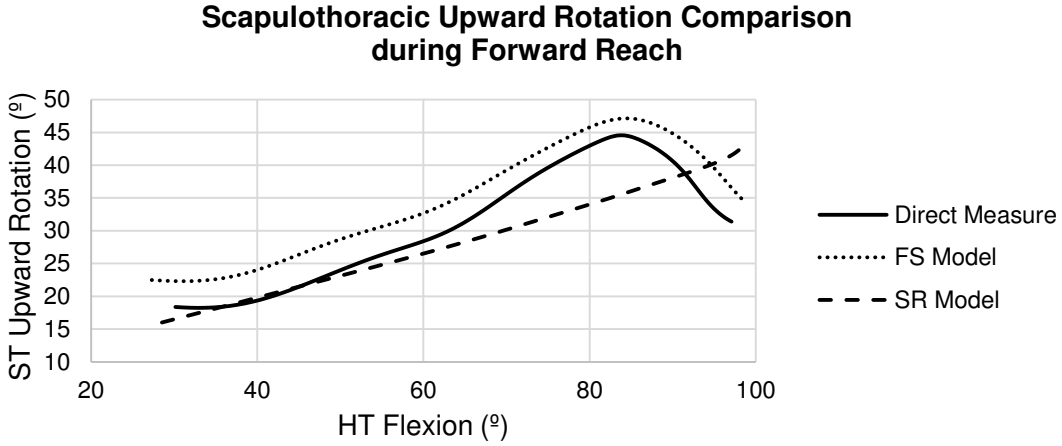

Figure 13 – Comparison of model-estimated scapulothoracic upward rotation with directly measured kinematics during forward reach for a representative subject. All subjects' models displayed similar agreement with direct measures.

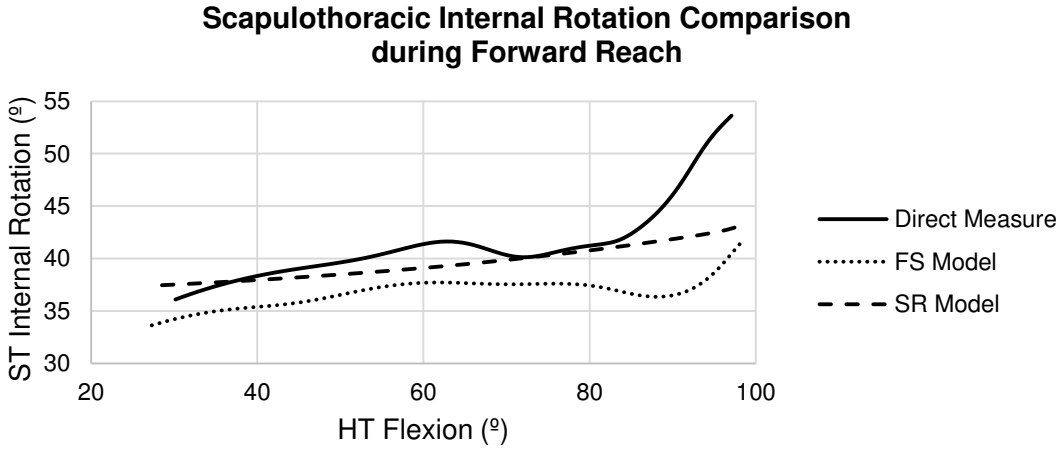

Figure 14 – Comparison of model-estimated scapulothoracic internal rotation with directly measured kinematics during forward reach for a representative subject. All subjects' models displayed similar agreement with direct measures.

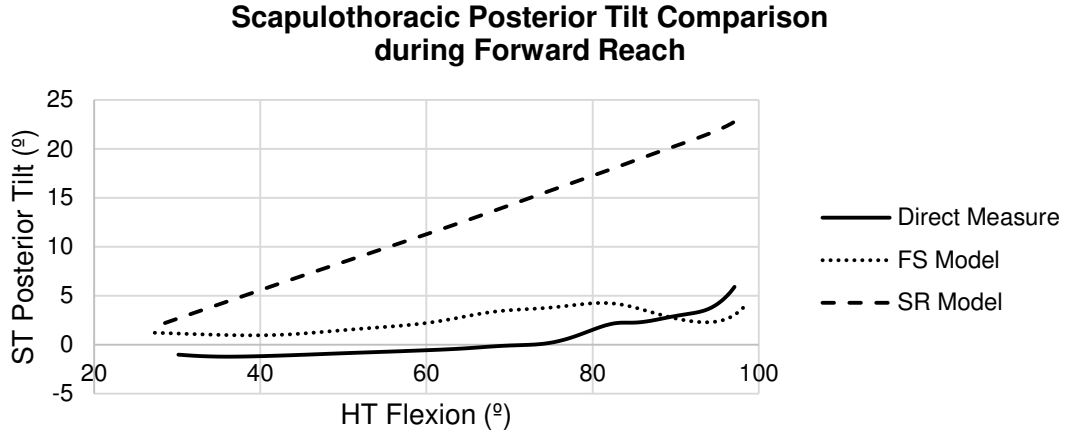

Figure 15 – Comparison of model-estimated scapulothoracic posterior tilt with directly measured kinematics during forward reach for a representative subject. All subjects' models displayed similar agreement with direct measures.

Model-estimated GH kinematics were also compared with direct measures. Both models effectively replicated the directly measured GH motion. The FS and SR models of all subjects displayed similar behavior when compared to directly measured ST angles. Data from a representative subject is shown in Figures 16, 17, and 18.

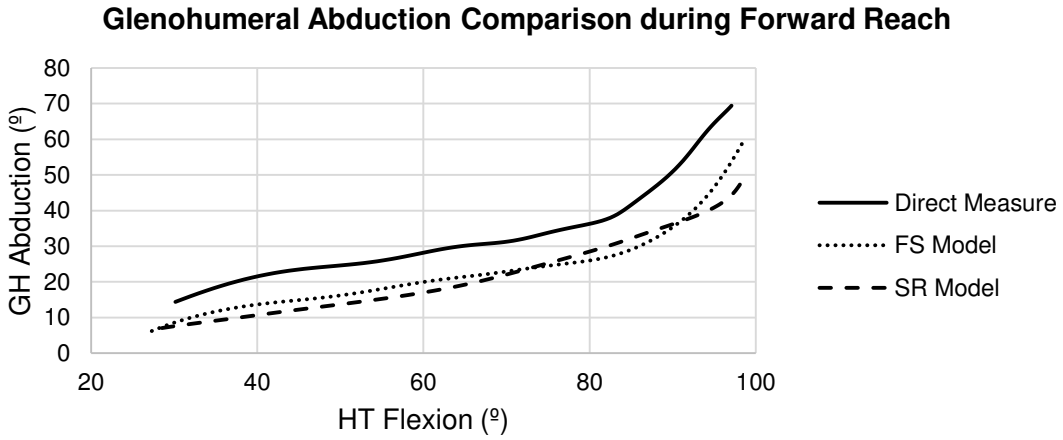

Figure 16 – Comparison of model-estimated glenohumeral abduction with directly measured kinematics during forward reach for a representative subject. All subjects' models displayed similar agreement with direct measures.

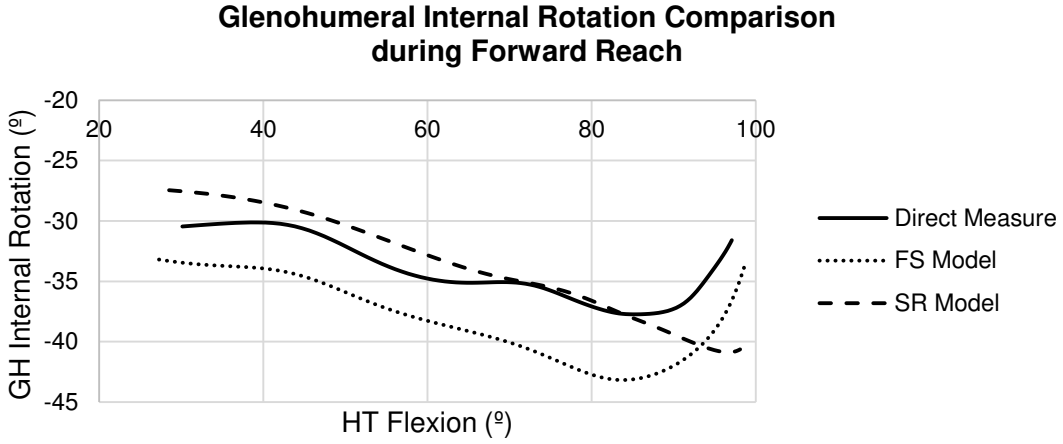

Figure 17 – Comparison of model-estimated glenohumeral internal rotation with directly measured kinematics during forward reach for a representative subject. All subjects' models displayed similar agreement with direct measures.

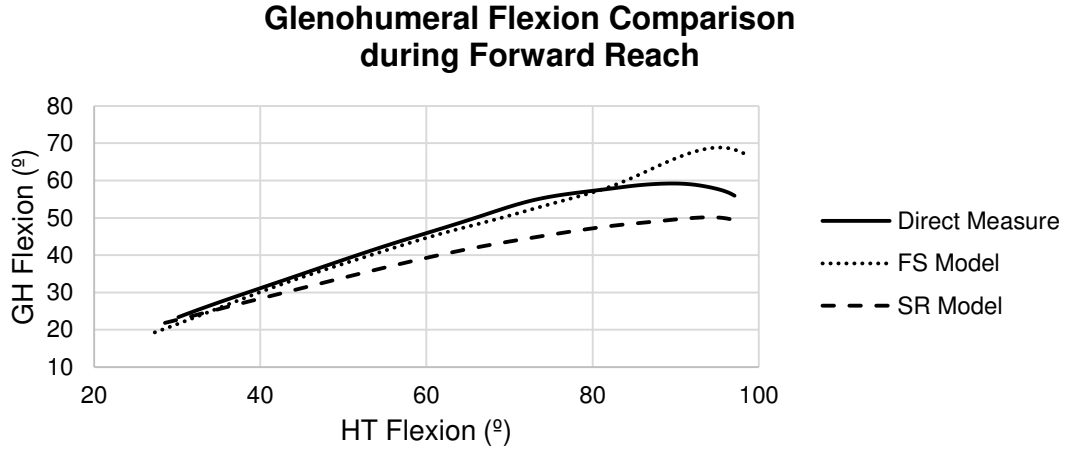

Figure 18 – Comparison of model-estimated glenohumeral flexion with directly measured kinematics during forward reach for a representative subject. All subjects' models displayed similar agreement with direct measures.

ST Angle Difference: Clinically meaningful differences (>10°) in 3D ST resultant angle between the SR and FS models were observed for all five subjects during forward reach (Figure 19). The differences in 3D ST resultant angle generally increased with greater levels of HT flexion for all subjects. The maximum differences in 3D ST resultant angle ranged from 15.98° to 28.02° between subjects, while mean differences fell between 12.02° and 17.42°. The percentage of the forward reach trial with >10° differences ranged from 45% to 100% between subjects. Averaged across all subjects, the maximum and mean differences in 3D ST resultant angles were 23.37° and 13.72°, respectively. Across all subjects, the average percentage of the forward reach trial with differences >10° was 73.80%. A full summary of 3D ST resultant angle differences can be found in Table 8.

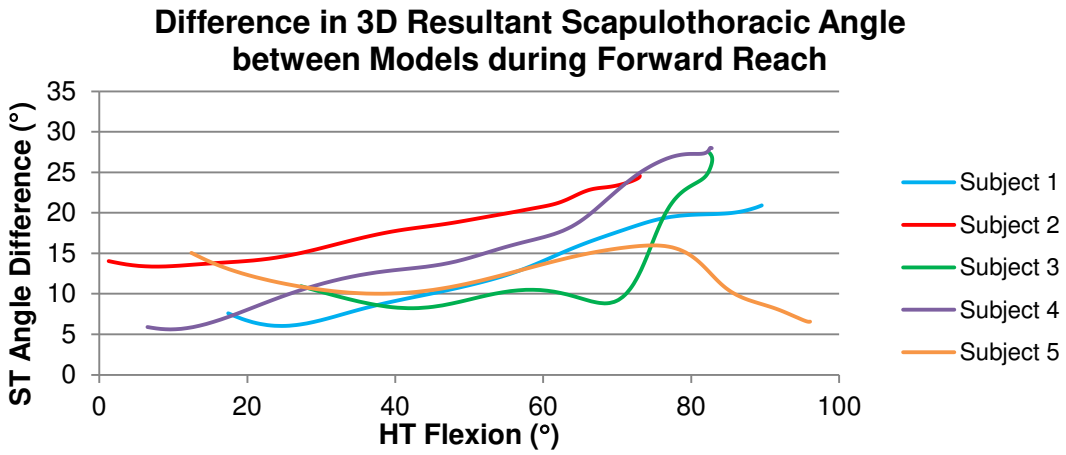

Figure 19 – Differences in 3D resultant scapulothoracic angle between the scapular rhythm (SR) and free scapula (FS) models during forward reach. Differences >10° are clinically meaningful.

Table 8 – Mean and maximum 3D resultant scapulothoracic angle differences between the scapular rhythm (SR) and free scapula (FS) models and percentage of the trial with differences >10° for all subjects during forward reach.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Max Diff. (°)	20.92	24.58	27.37	28.02	15.98	23.37
Mean Diff. (°)	12.85	17.42	11.54	14.75	12.02	13.72
% of Trial >10°	62.00	100.00	45.00	75.00	87.00	73.8

ST Upward Rotation: The maximum ST upward rotation difference between the FS and SR models during forward reach ranged from 6.91° to 18.65° between subjects. To aid in interpretation of the results, kinematic differences were also expressed a percentage of the FS model's ST upward rotation excursion throughout the trial; this will be referred to as reference ROM. The maximum differences in ST upward rotation were 24.06% to 138.56%. The mean differences in ST upward rotation between models for all subjects were between 4.10° and 6.74° or 13.61% and

50.07% of each subject's reference ROM. For maximum ST upward rotation difference, 2 of 5 subjects had differences that were greater than 50% of their reference ROM. For mean ST upward rotation difference, 1 of 5 subjects had a difference that was greater than 50% of their reference ROM. Averaged across all subjects, the maximum and mean differences in ST upward rotation were 11.57° (63.66% ROM) and 5.76° (29.79% ROM), respectively. A full summary of ST upward rotation differences between models is presented in Table 9.

Table 9 – Mean and maximum scapulothoracic upward rotation angle differences between the scapular rhythm (SR) and free scapula (FS) models for all subjects during forward reach.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Reference ROM (°)	24.52	30.13	13.46	25.62	16.56	22.06
Max Diff. (°)	11.47	7.25	18.65	6.91	13.57	11.57
Max Diff. (% of ROM)	46.78	24.06	138.56	26.97	81.94	63.66
Mean Diff. (°)	6.59	4.10	6.74	4.84	6.54	5.76
Mean Diff. (% of ROM)	26.88	13.61	50.07	18.89	39.49	29.79

Comparison of the ST upward rotation trends between the FS and SR models during forward reach indicated strong to very strong trend similarities for four of five subjects (Table 10). One subject exhibited negative correlations for both the symmetry ratio and Pearson's r-value. A full summary of the trend analysis for all three ST DOFs during abduction is presented in Table 10. Data for the subjects with the strongest and weakest trends for ST upward rotation are displayed in Figure 20.

Table 10 – Kinematic trend analysis for scapulothoracic upward rotation, internal rotation, and anterior tilt during forward reach.

	ST Upward Rotation		ST Internal Rotation		ST Anterior Tilt	
	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R
Subject 1	0.907	0.826	0.859	0.884	0.982	0.795
Subject 2	0.965	0.997	0.929	0.962	0.991	0.711
Subject 3	-0.753	-0.612	0.944	0.974	-0.91	-0.731
Subject 4	0.942	0.982	-0.968	-0.387	-0.955	-0.437
Subject 5	0.967	0.977	0.916	0.902	0.893	0.885

Negative	Weak	Moderate	Strong	Very Strong
<0	0 - 0.39	0.4 – 0.69	0.7 – 0.89	0.9 - 1

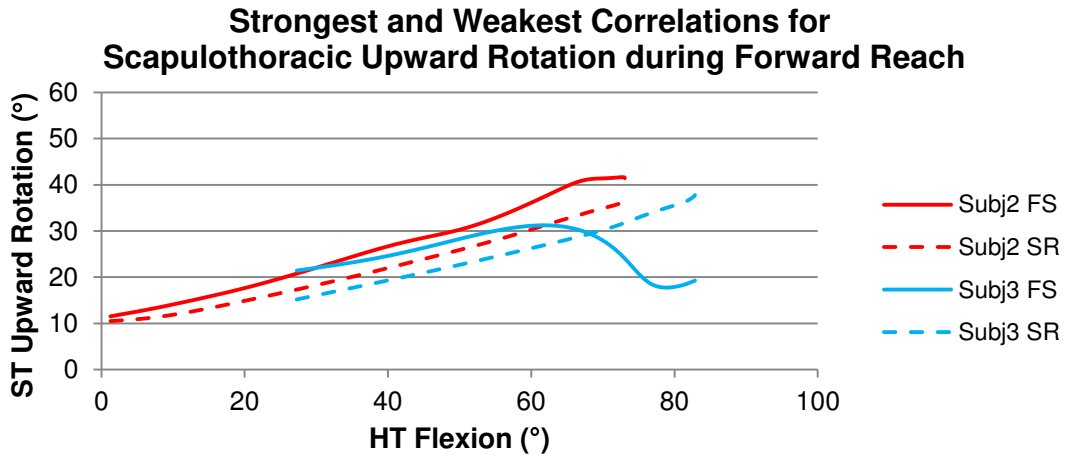

Figure 20 – Subjects with the strongest (r = 0.997) and weakest (r = -0.612) correlations for scapulothoracic upward rotation between their scapular rhythm (SR) and free scapula (FS) models during forward reach.

ST Internal Rotation: The maximum ST internal rotation difference between the FS and SR models ranged from 4.70° to 14.93° between subjects. To aid in interpretation of the results, kinematic differences were also expressed a percentage of the FS model's ST internal rotation excursion throughout the trial; this will be referred

to as reference ROM. The maximum differences in ST internal rotation were 50.98% to 698.06%. The mean differences in ST internal rotation between models for all subjects were between 2.30° and 13.73° or 24.95% and 463.59% of each subject's reference ROM. For maximum ST internal rotation difference, all subjects had differences greater than 50% of their reference ROM. For mean ST internal rotation difference, 3 of 5 subjects had a difference greater than 50% of their reference ROM. Averaged across all subjects, the maximum and mean differences in ST internal rotation were 10.20° (206.97% ROM) and 7.51° (143.89% ROM), respectively. A full summary of ST internal rotation differences between models is presented in Table 11.

Table 11 – Mean and maximum scapulothoracic internal rotation angle differences between the scapular rhythm (SR) and free scapula (FS) models for all subjects during forward reach.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Reference ROM (°)	9.22	11.14	10.56	2.06	11.88	8.97
Max Diff. (°)	4.70	14.93	8.16	14.38	8.85	10.20
Max Diff. (% of ROM)	50.98	134.02	77.27	698.06	74.49	206.97
Mean Diff. (°)	2.30	13.73	6.49	9.55	5.49	7.51
Mean Diff. (% of ROM)	24.95	123.25	61.46	463.59	46.21	143.89

Comparison of the ST internal rotation trends between the FS and SR models during forward reach indicated strong to very strong trend similarities for four of five subjects (Table 10). One subject exhibited negative correlations for both the symmetry ratio and Pearson's r-value. Data for the subjects with the strongest and weakest trends for ST internal rotation are displayed in Figure 21.

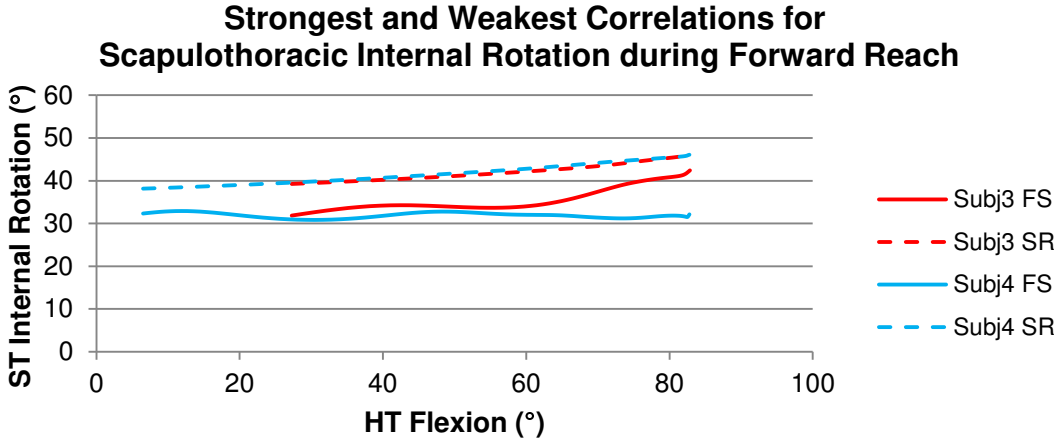

Figure 21 – Subjects with the strongest (r = 0.974) and weakest (r = -0.387) correlations for scapulothoracic internal rotation between their scapular rhythm (SR) and free scapula (FS) models during forward reach.

ST Anterior Tilt: The maximum ST anterior tilt difference between the FS and SR models ranged from 12.25° to 24.12° between subjects. To aid in interpretation of the results, kinematic differences were also expressed a percentage of the FS model's ST anterior tilt excursion throughout the trial; this will be referred to as reference ROM. The maximum differences in ST anterior tilt were 72.19% to 526.60%. The mean differences in ST anterior tilt between models for all subjects were between 5.36° and 10.30° or 41.25% and 273.94% of each subject's reference ROM. For maximum ST anterior tilt difference, all subjects had differences that were greater than 50% of their reference ROM. For mean ST anterior tilt difference, 4 of 5 subjects had a difference that was greater than 50% of their reference ROM. Averaged across all subjects, the maximum and mean differences in ST anterior tilt were 19.52° (362.14% ROM) and 8.06° (149.92% ROM), respectively. A full summary of ST anterior tilt differences between models is presented in Table 12.

Table 12 – Mean and maximum scapulothoracic anterior tilt angle differences between the scapular rhythm (SR) and free scapula (FS) models for all subjects during forward reach.

Parameter	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Avg.
Reference ROM (°)	3.76	4.60	6.88	5.33	16.97	7.51
Max Diff. (°)	19.80	21.86	19.55	24.12	12.25	19.52
Max Diff. (% of ROM)	526.60	475.22	284.16	452.53	72.19	362.14
Mean Diff. (°)	10.30	8.46	5.36	9.20	7.00	8.06
Mean Diff. (% of ROM)	273.94	183.91	77.91	172.61	41.25	149.92

Comparison of the ST anterior tilt trends between the FS and SR models during forward reach revealed a range of trend similarities (Table 10). Three of five subjects demonstrated strong to very strong trends; however, the remaining two subjects displayed negative correlations. Data for the subjects with the strongest and weakest trends for ST anterior tilt are displayed in Figure 22.

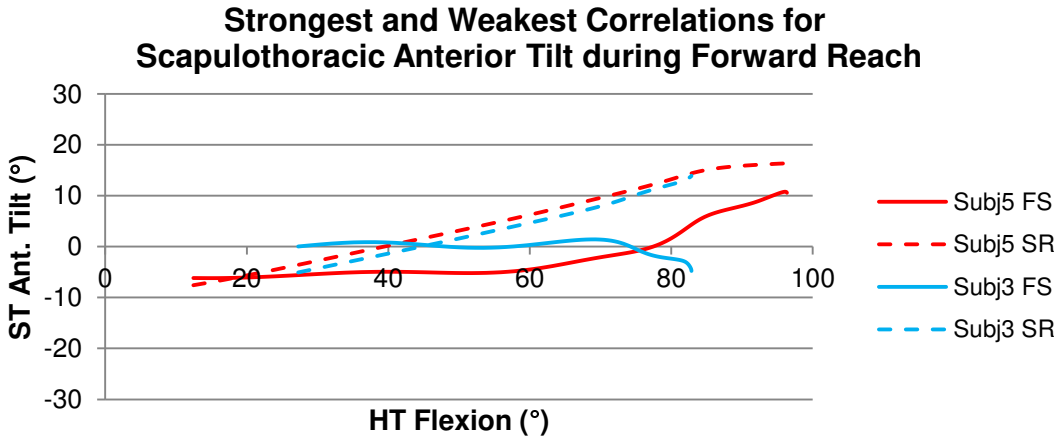

Figure 22 – Subjects with the strongest (r = 0.885) and weakest (r = -0.731) correlations for scapulothoracic anterior tilt between their scapular rhythm (SR) and free scapula (FS) models during forward reach.

3.3 Aim 2: Glenohumeral Kinetics

3.3.1 Abduction

Comparison with Direct Measures: A comparison between model-estimated and directly calculated GH torques revealed that both the FS and SR models did not accurately recreate the mechanics occurring in vivo during shoulder abduction. Both the magnitude and the pattern of the model torques displayed appreciable disagreement. All subjects model-estimated GH torques were in similarly poor agreement with direct measures, consequently data from only one representative subject is shown in Figures 23, 24, and 25.

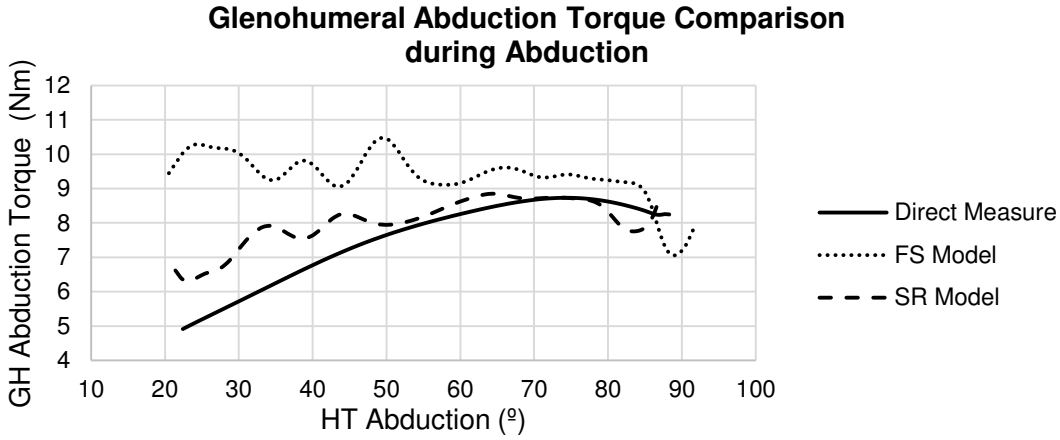

Figure 23 – Comparison of model-estimated glenohumeral abduction torques with directly measured torques during shoulder abduction for a representative subject. All subjects displayed different, but comparatively poor agreement.

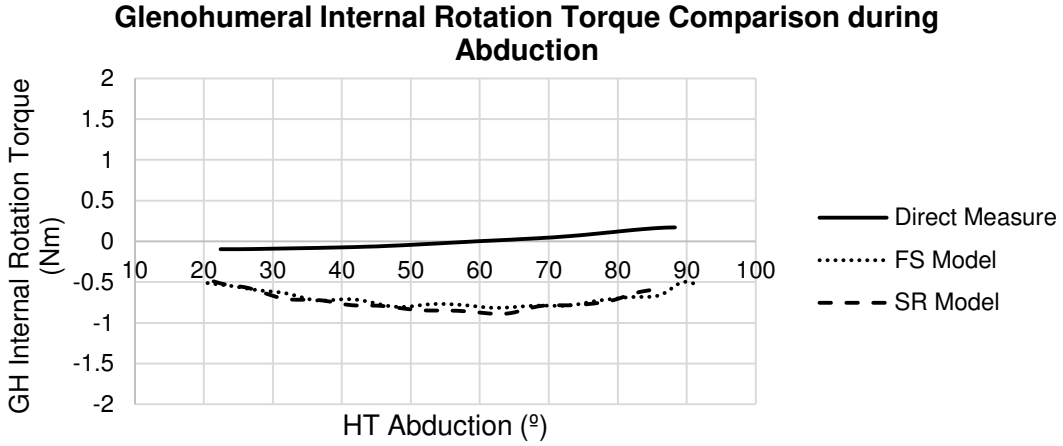

Figure 24 – Comparison of model-estimated glenohumeral internal rotation torques with directly measured torques during shoulder abduction for a representative subject. All subjects displayed different, but comparatively poor agreement.

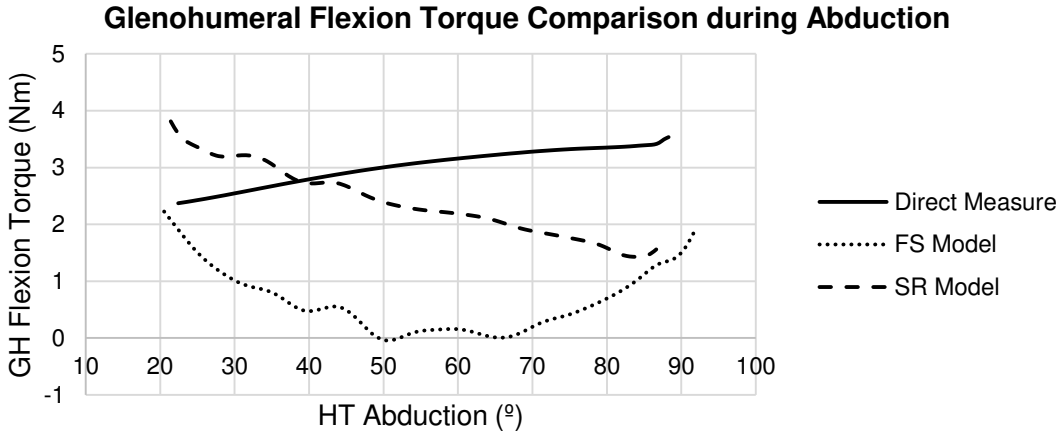

Figure 25 – Comparison of model-estimated glenohumeral flexion torques with directly measured torques during shoulder abduction for a representative subject. All subjects displayed different, but comparatively poor agreement.

GH Abduction Torques: The FS and SR models were in functional agreement (*see Glossary*) for all subjects throughout each of their respective abduction trials. The mean GH abduction torque difference between the models fell between 6.14% and 12.09% of the maximum FS torque within the trial for each subject. The maximum GH abduction torque difference between models ranged from 13.53% to 28.55% of the maximum FS torque within the trial for each subject. The raw magnitudes of the GH abduction torques were small (<3.0 Nm, *see Glossary for definition of 'small')* and similar between the FS and SR models. The difference in maximum torque between the FS and SR models for all five subjects ranged from 0.15 to 1.31 Nm. A comparison of GH abduction torques between the FS and SR models during abduction are displayed for a representative subject in Figure 26. Trend analysis revealed strong to very strong correlations between the GH abduction torques of the FS and SR models during abduction (Table 13).

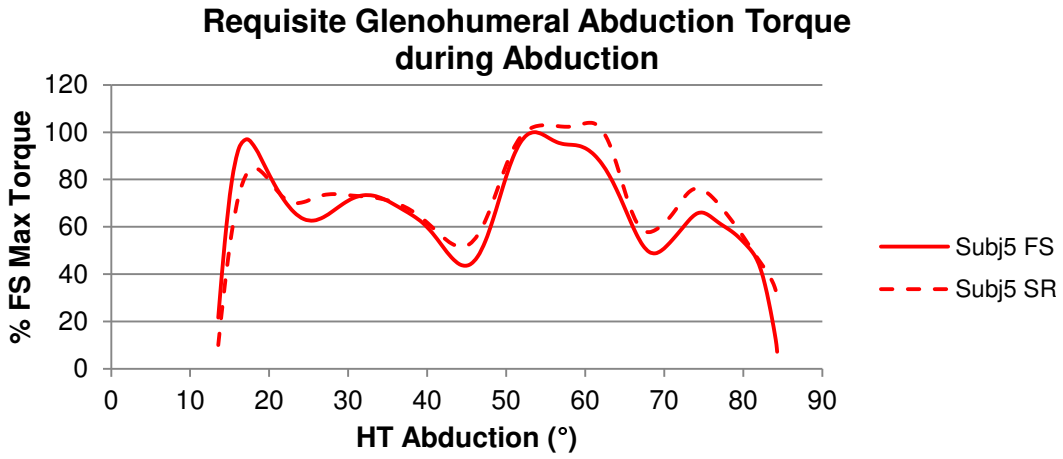

Figure 26 – Requisite glenohumeral abduction torques for the scapular rhythm (SR) and free scapula (FS) models during shoulder abduction. Data for one representative subject displayed and expressed as a percentage of the maximum FS torques within the trial.

Table 13 – Kinetic trend analysis for requisite glenohumeral abduction, internal rotation, and flexion torques during abduction.

	GH Abduction		GH Internal Rotation		GH Flexion	
	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R
Subject 1	0.892	0.864	0.976	0.992	0.895	0.770
Subject 2	0.966	0.944	0.943	0.900	0.641	0.523
Subject 3	0.884	0.922	0.939	0.886	-0.522	-0.058
Subject 4	0.718	0.719	0.980	0.950	0.510	0.212
Subject 5	0.974	0.951	0.968	0.992	-0.286	0.056

Negative	Weak	Moderate	Strong	Very Strong
<0	0 - 0.39	0.4 – 0.69	0.7 – 0.89	0.9 - 1

GH Internal Rotation Torque: The FS and SR models were in functional agreement for all subjects throughout each of their respective abduction trials. The mean GH internal rotation torque difference between the FS and SR models fell between 1.74% and 12.03% of the maximum FS torque within the trial for each subject. The maximum GH internal rotation torque difference between models ranged from 7.77% to 27.41% of the maximum FS torque within the trial for each subject. The raw magnitudes of the GH internal rotation torques were very small (<3.0 Nm, *see Glossary for definition of 'small')* and similar between the FS and SR models. The difference in maximum torque between the FS and SR models for all five subjects ranged from 0.009 to 0.119 Nm. A comparison of GH internal rotation torques between the FS and SR models during abduction are displayed for a representative subject in Figure 27. Trend analysis revealed strong to very strong correlations between the GH internal rotation torques of the FS and SR models during abduction (Table 13).

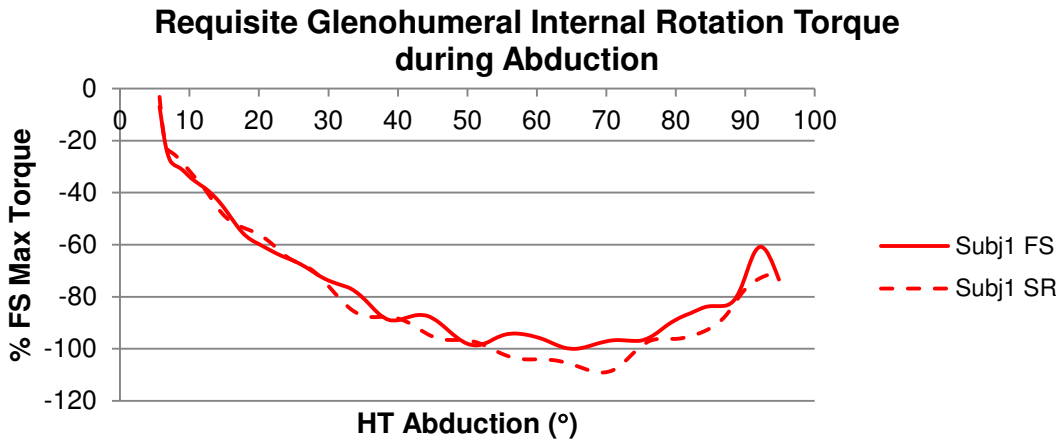

Figure 27 – Requisite glenohumeral internal rotation torques for the scapular rhythm (SR) and free scapula (FS) models during shoulder abduction. Data for one representative subject displayed and expressed as a percentage of the maximum FS torque within the trial.

GH Flexion Torques: The FS and SR models were in functional agreement for 3 of 5 subjects throughout each of their respective abduction trials. One subject had a brief functional disagreement between the FS and SR models at the beginning of the trial. Another subject's models were in functional disagreement for the majority of the abduction trial with the FS model requiring primarily GH extension torques while the SR model required flexion torques. The mean GH flexion torque difference between the FS and SR models ranged from 24.75% to 77.98% of the maximum FS torque within the trial for each subject. The maximum GH flexion torque difference between models fell between 46.68% and 215.75% of the maximum FS torque within the trial for each subject. The raw magnitudes of the GH flexion torques for the FS and SR models varied between subjects with maximum torque difference between the FS and SR models ranging from 0.05 to 3.81 Nm. A comparison of GH flexion torques between the FS and SR models during abduction are displayed for a representative

subject in Figure 28. The correlations between the GH flexion torques of the FS and SR models during abduction varied greatly between subjects (Table 13). For symmetry ratio, one subject had strong correlations, while two had moderate correlations and two had negative correlations. The Pearson's r-values were as follows: one strong, one moderate, two weak, and one negative correlation.

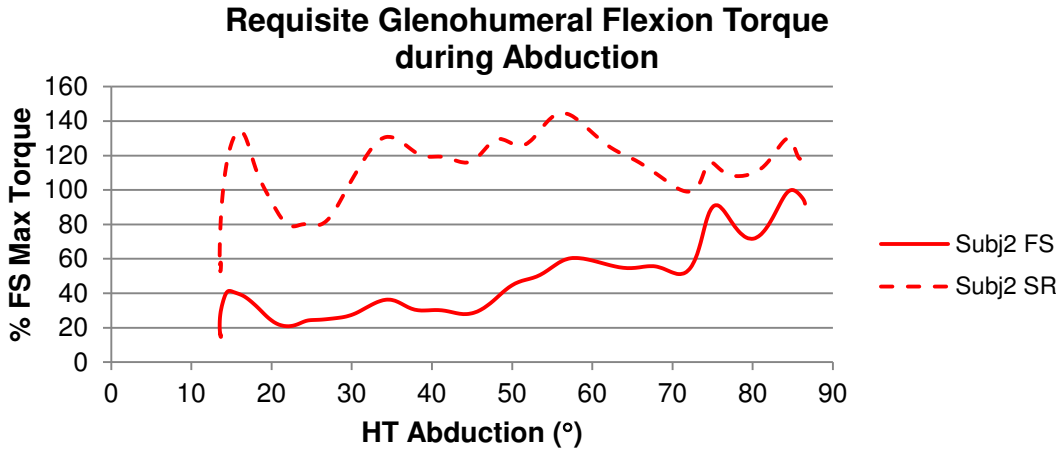

Figure 28 – Requisite glenohumeral flexion torques for the scapular rhythm (SR) and free scapula (FS) models during shoulder abduction. Data for one representative subject displayed and expressed as a percentage of the maximum FS torque within the trial.

3.3.2 Forward Reach

Comparison with Direct Measures: A comparison between model-estimated and directly calculated GH torques revealed that both the FS and SR models did not accurately recreate the mechanics occurring in vivo during forward reach. Both the magnitude and the pattern of the model torques displayed appreciable disagreement. All subjects model-estimated GH torques were in similarly poor agreement with direct

measures, consequently data from only one representative subject is shown in Figures 29, 30, and 31.

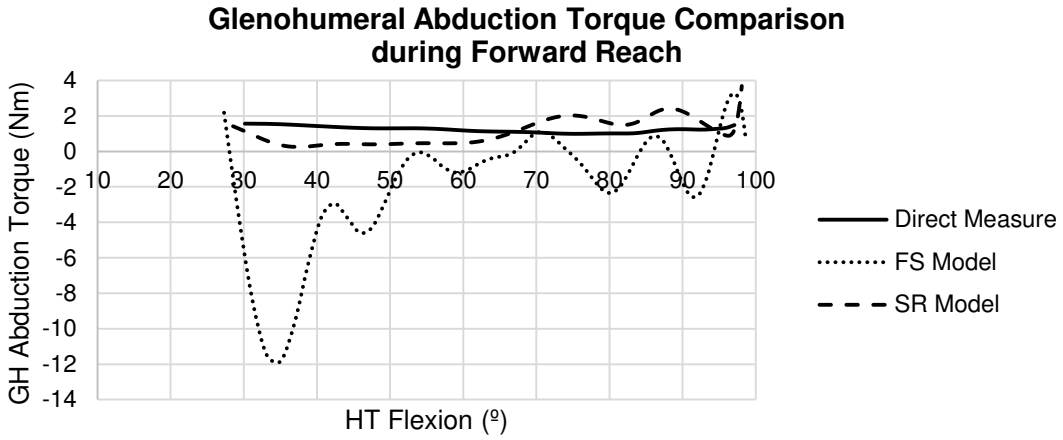

Figure 29 – Comparison of model-estimated glenohumeral abduction torques with directly measured torques during forward reach for a representative subject. All subjects displayed different, but comparatively poor agreement.

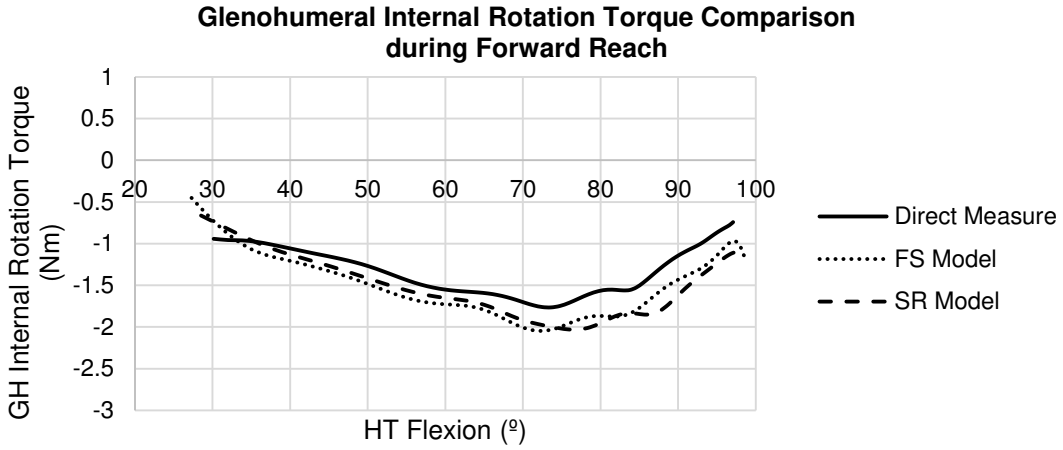

Figure 30 – Comparison of model-estimated glenohumeral internal rotation torques with directly measured torques during forward reach for a representative subject. All subjects displayed different, but comparatively poor agreement.

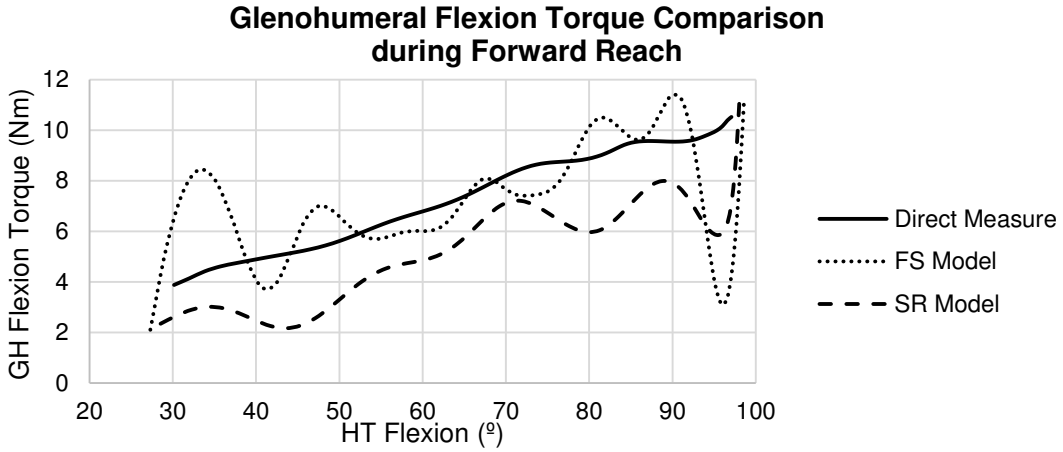

Figure 31 – Comparison of model-estimated glenohumeral flexion torques with directly measured torques during forward reach for a representative subject. All subjects displayed different, but comparatively poor agreement.

GH Abduction Torques: The FS and SR models were in functional agreement for all subjects throughout each of their respective forward reach trials. The mean GH abduction torque difference between the FS and SR models ranged from 5.46% to 16.63% of the maximum FS torque within the trial for each subject. The maximum GH abduction torque difference between models fell between 14.07% and 47.16% of the maximum FS torque within the trial for each subject. The raw magnitudes of the GH abduction torques for the FS and SR models varied between subject with the difference in maximum torque between the FS and SR models ranging from 0.64 to 3.13 Nm. A comparison of GH abduction torques between the FS and SR models during forward reach are displayed for a representative subject in Figure 32. Trend analysis revealed moderate to very strong correlations between the GH abduction torques of the FS and SR models during forward reach (Table 14).

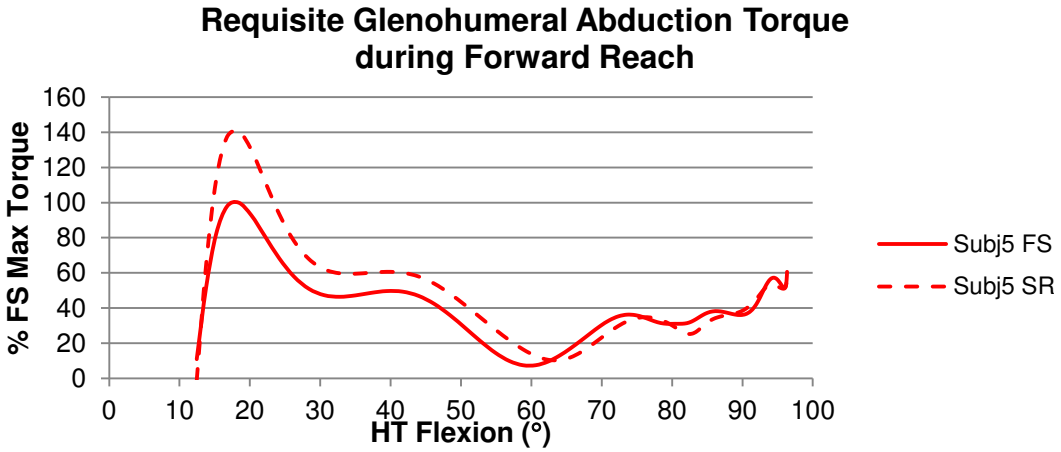

Figure 32 – Requisite glenohumeral abduction torques for the scapular rhythm (SR) and free scapula (FS) models during forward reach. Data for one representative subject displayed and expressed as a percentage of the maximum FS torque within the trial.

Table 14 – Kinetic trend analysis for requisite glenohumeral abduction, internal rotation, and flexion torques during forward reach.

	GH Abduction		GH Internal Rotation		GH Flexion	
	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R	Symmetry Ratio	Pearson's R
Subject 1	0.72	0.719	0.985	0.985	0.867	0.941
Subject 2	0.775	0.646	0.977	0.967	0.932	0.947
Subject 3	0.966	0.994	0.984	0.977	0.961	0.984
Subject 4	0.938	0.951	0.949	0.87	0.877	0.887
Subject 5	0.928	0.946	0.988	0.998	0.914	0.945

Negative	Weak	Moderate	Strong	Very Strong
<0	0 - 0.39	0.4 – 0.69	0.7 – 0.89	0.9 - 1

GH Internal Rotation Torques: The FS and SR models were in functional agreement for all subjects throughout each of their respective forward reach trials. The mean GH internal rotation torque difference between the FS and SR models ranged from 2.80% to 16.68% of the maximum FS torque within the trial for each subject. The maximum GH internal rotation torque difference between models fell between 6.35% and 34.16% of the maximum FS torque within the trial for each subject. The raw magnitudes of the GH internal rotation torques were very small (<3.0 Nm, *see Glossary for definition of 'small')* and similar between the FS and SR models. The difference in maximum torque between the FS and SR models for all five subjects ranged from 0.02 to 0.17 Nm. A comparison of GH internal rotation torques between the FS and SR models during forward reach are displayed for a representative subject in Figure 33. Trend analysis revealed strong to very strong correlations between the GH internal rotation torques of the FS and SR models during abduction (Table 14).

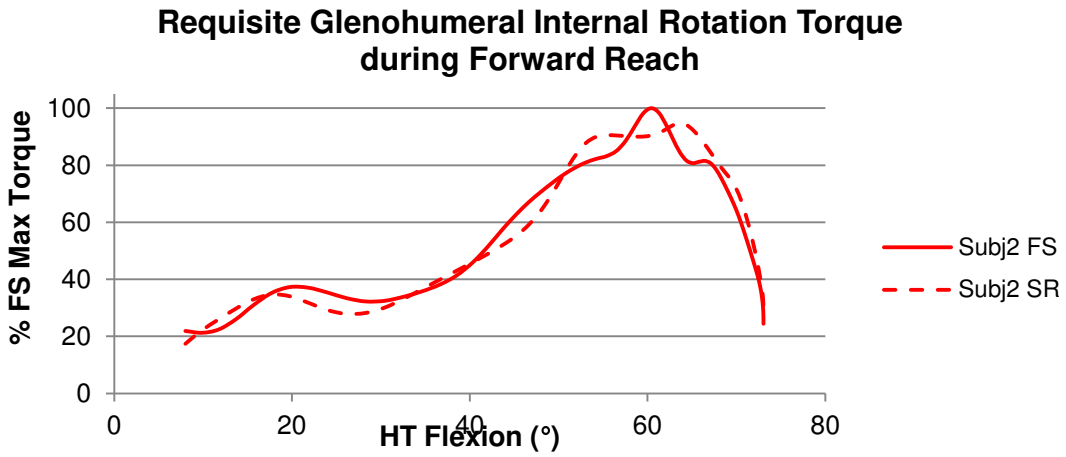

Figure 33 – Requisite glenohumeral internal rotation torques for the scapular rhythm (SR) and free scapula (FS) models during forward reach. Data for one representative subject displayed and expressed as a percentage of the maximum FS torque within the trial.

GH Flexion Torques: The FS and SR models were in functional agreement for all subjects throughout each of their respective forward reach trials. The mean GH flexion torque difference between the FS and SR models ranged from 5.59% to 9.92% of the maximum FS torque within the trial for each subject. The maximum GH flexion torque difference between models fell between 16.73% and 44.25% of the maximum FS torque within the trial for each subject. The raw magnitudes of the GH abduction torques for the FS and SR models varied between subjects with the difference in maximum torque between the FS and SR models ranging from 0.20 to 2.02 Nm. A comparison of GH flexion torques between the FS and SR models during forward reach are displayed for a representative subject in Figure 34. Trend analysis revealed moderate to very strong correlations between the GH flexion torques of the FS and SR models during forward reach (Table 14).

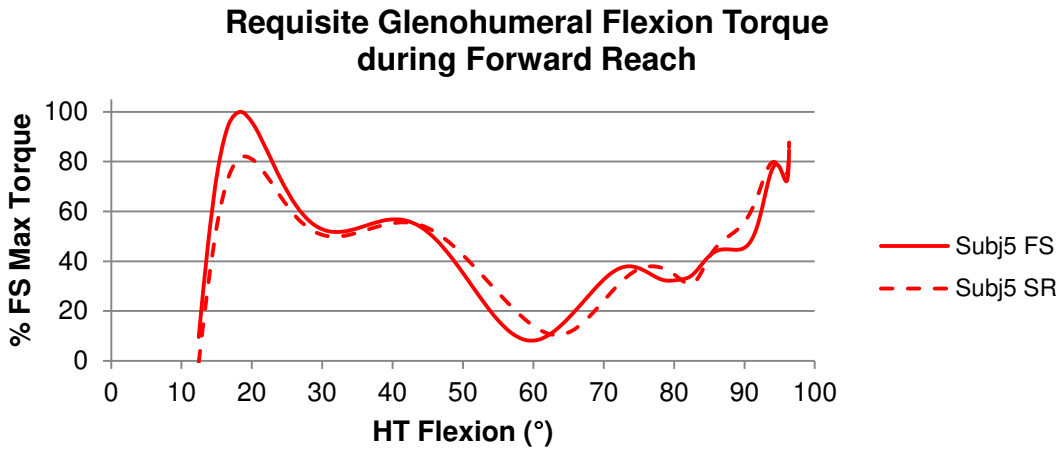

Figure 34 – Requisite glenohumeral flexion torques for the scapular rhythm (SR) and free scapula (FS) models during forward reach. Data for one representative subject displayed and expressed as a percentage of the maximum FS torque within the trial.

3.4 Aim 3: Muscle Activations

3.4.1 Abduction

Muscle activation percent on/off timing agreements between CMC and EMG for all subjects are listed in Table 15. Averaged across all subjects, 5 of 12 muscles reported >50% timing agreement: anterior deltoid, posterior deltoid, teres major, infraspinatus, and supraspinatus. Three of five subjects achieved >50% timing agreement averaged across all muscles with 64.6% and 48.0% being the highest and lowest across muscle on/off timing agreements.

Table 15 – Percentage agreement of muscle activation on/off timing between EMG and CMC during shoulder abduction.

Functional Role	Muscle	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Mean
Prime Mover	Anterior Deltoid	92.7	77.4	97.9	100.0	100.0	93.6
	Posterior Deltoid	93.9	41.4	100.0	91.2	98.9	85.1
	Supraspinatus	100.0	99.7	100.0	100.0	100.0	99.9
Stabilizer	Infraspinatus	81.8	90.4	11.8	95.9	100.0	76.0
	Subscapularis	43.8	27.5	97.0	3.4	41.7	42.7
	Teres Minor	22.8	70.8	45.2	6.7	61.0	41.3
	Triceps Brachii	40.4	52.8	12.0	0.0	97.5	40.5
	Coracobrachialis	23.9	36.9	5.4	0.0	8.4	14.9
Inactive	Latissimus Dorsi	25.3	19.6	32.3	91.5	68.9	47.5
	Teres Major	74.1	57.8	50.6	78.2	79.1	68.0
	Biceps Brachii	21.8	4.1	16.8	33.4	0.0	15.2
	Pectoralis Major	16.4	0.6	7.3	38.3	19.8	16.5
	Across Muscle Mean	53.1	48.3	48.0	53.2	64.6	

Figure 35 displays the percent agreement between CMC and EMG for on/off activation timing. Muscles are grouped by functional role—prime movers are responsible for generating the motion, stabilizers are on for >20% of the trial, and inactive muscles are on for <20% of the trial. Prime movers were on for the majority of the trial and showed very high on/off timing agreement between CMC and EMG for all subjects. Muscles contributing a stabilizing role demonstrated lower levels of agreement than prime movers and had a greater variability in agreement values between subjects. Inactive muscles displayed the lower timing agreement than prime movers and stabilizers. Additionally, the inactive muscles displayed high variability between subjects.

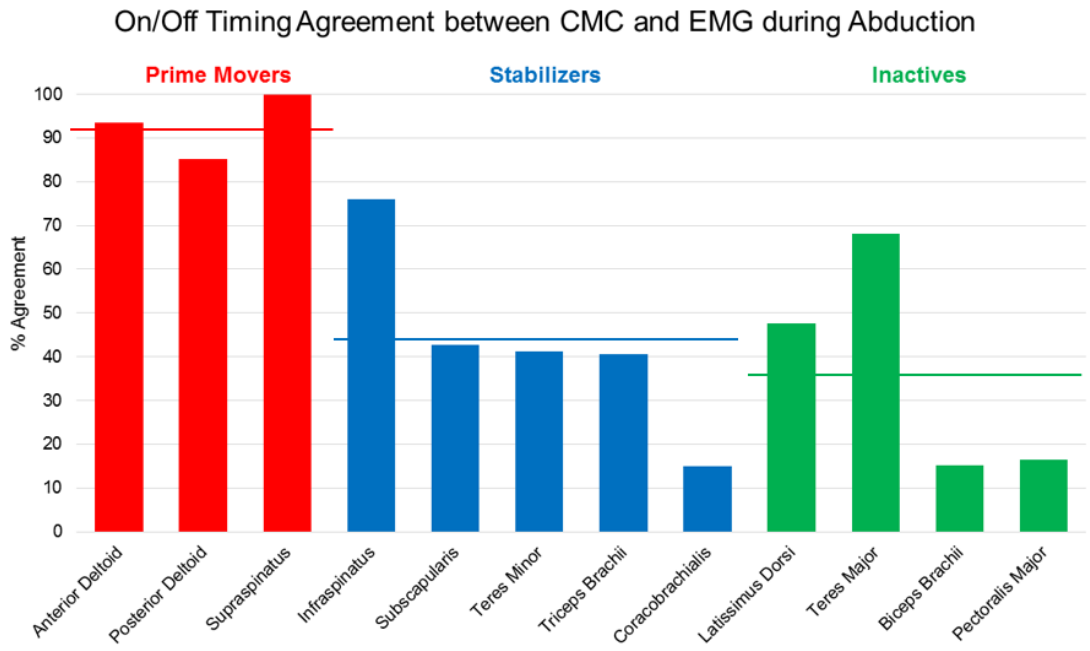

Figure 35 – Muscle activation on/off timing comparison between EMG and CMC during shoulder abduction. Agreement from all subjects averaged for each muscle. Horizontal line indicates average agreement for each muscle group.

All data for EMG vs. CMC correlation analyses for abduction is presented in Table 16. Muscles which were on for <20% of the abduction trial according to EMG were excluded from trend analysis. Trends between EMG and CMC varied greatly with the majority of muscles having weak or negative correlations. Prime mover muscles did not have notably improved correlations compared to the stabilizers. Correlation values for each muscle were inconsistent between subjects. Activations from CMC generally had far greater fluctuations throughout the trial when compared to EMG. Both EMG and CMC deltoid muscle activation patterns during abduction for a representative subject are presented in Figure 36. Figure 37 displays EMG and CMC

activation patterns of the rotator cuff muscles during abduction for a representative subject.

Table 16 – Pearson's R-values between EMG and CMC activation signals during shoulder abduction. Muscles that were active <20% of the trial were excluded from analysis.

Muscle	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Mean
Pectoralis Major	N/A	N/A	N/A	N/A	N/A	N/A
Anterior Deltoid	0.270	-0.198	0.634	0.485	0.117	0.262
Posterior Deltoid	0.774	-0.446	0.733	0.627	0.046	0.347
Latissimus Dorsi	N/A	N/A	N/A	N/A	N/A	N/A
Biceps Brachii	N/A	N/A	N/A	N/A	N/A	N/A
Triceps Brachii	0.081	-0.099	N/A	0.099	N/A	0.000
Subscapularis	0.416	-0.219	0.217	0.657	0.313	0.277
Teres Minor	-0.344	-0.094	N/A	0.254	0.732	0.137
Teres Major	N/A	N/A	-0.625	0.698	N/A	0.037
Infraspinatus	0.276	0.144	-0.616	0.607	0.709	0.434
Supraspinatus	0.747	0.197	0.493	0.244	0.633	0.463
Coracobrachialis	-0.390	-0.506	-0.402	0.254	0.100	-0.189
Across Muscle Mean	0.229	-0.153	0.175	0.436	0.379	

Negative	Weak	Moderate	Strong	Very Strong
<0	0 - 0.39	0.4 – 0.69	0.7 – 0.89	0.9 - 1

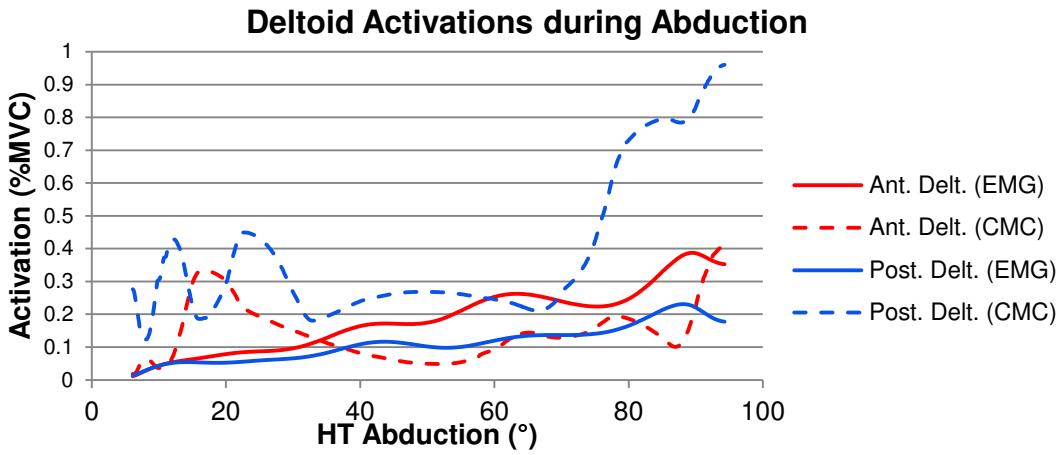

Figure 36 – Muscle activation patterns between EMG and CMC for deltoid during shoulder abduction. Data from one subject displayed. Subject 1: anterior deltoid (r=0.270) and posterior deltoid (r=0.774).

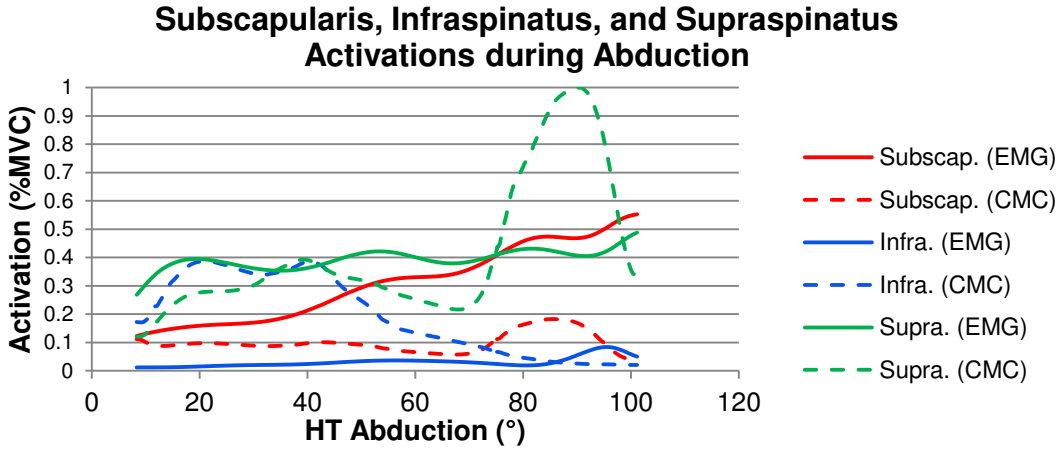

Figure 37 – Muscle activation patterns between EMG and CMC for rotator cuff muscles during shoulder abduction. Data from one subject displayed. Subject 3: subscapularis (r=0.217), infraspinatus (r=-0.616), and supraspinatus (r=0.493).

3.4.2 Forward Reach

Muscle activation percent on/off timing agreements between CMC and EMG for all subjects are listed in Table 17. Averaged across all subjects, 6 of 12 muscles reported >50% timing agreement: anterior deltoid, posterior deltoid, subscapularis, teres minor, infraspinatus, and supraspinatus. Four of five subjects achieved >50% timing agreement averaged across all muscles with 57.2% and 49.9% being the highest and lowest across muscle on/off timing agreements.

Table 17 – Percent agreement of muscle activation on/off timing between EMG and CMC during forward reach.

Functional Role	Muscle	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Mean
Prime Mover	Anterior Deltoid	100.0	97.4	96.0	100.0	100.0	98.7
	Supraspinatus	97.6	100.0	81.2	92.0	100.0	94.2
Stabilizer	Posterior Deltoid	44.7	24.1	72.7	58.2	54.0	50.7
	Infraspinatus	100.0	100.0	100.0	100.0	100.0	100.0
	Subscapularis	86.0	33.0	80.0	76.3	100.0	75.1
	Teres Minor	64.1	97.4	49.3	64.9	92.5	73.6
	Coracobrachialis	0.0	22.4	0.0	8.7	0.0	6.2
Inactive	Triceps Brachii	80.4	23.0	70.2	21.7	39.3	46.9
	Latissimus Dorsi	31.0	3.3	45.4	22.1	50.1	30.4
	Teres Major	71.9	50.6	0.0	8.5	42.4	34.7
	Biceps Brachii	0.0	33.4	59.8	26.8	0.0	24.0
	Pectoralis Major	10.4	13.9	30.2	34.0	0.0	17.7
	Across Muscle Mean	57.2	49.9	57.1	51.1	56.5	

Figure 38 displays the percent agreement between CMC and EMG for on/off activation timing. Muscles are grouped by functional role—prime movers are responsible for generating the motion, stabilizers are on for >20% of the trial, and inactive muscles are on for <20% of the trial. Prime movers were on for the majority of the trial and showed very high on/off timing agreement between CMC and EMG for all subjects. Muscles contributing a stabilizing role demonstrated lower levels of

agreement than prime movers and had a greater variability in agreement values between subjects. Inactive muscles displayed the lower timing agreement than prime movers and stabilizers. Additionally, the inactive muscles displayed high variability between subjects.

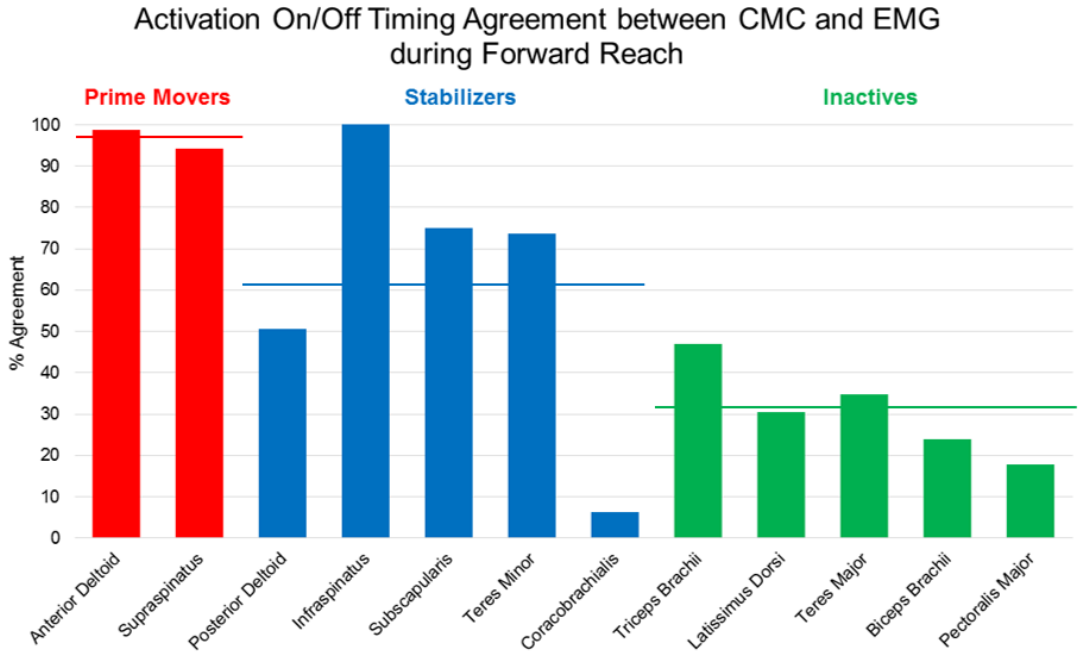

Figure 38 – Muscle activation on/off timing comparison between EMG and CMC during forward reach. Agreement from all subjects averaged for each muscle. Horizontal line indicates average agreement for each muscle group.

All data for EMG vs. CMC correlation analyses for forward reach is presented in Table 18. Muscles which were on for <20% of the abduction trial according to EMG were excluded from trend analysis. Correlations between EMG and CMC were primarily negative. Prime mover muscles did not have improved correlations

compared to the stabilizer muscles. Activations from CMC generally had far greater fluctuations throughout the trial when compared to EMG. Both EMG and CMC deltoid muscle activation patterns during forward reach for a representative subject are presented in Figure 39. Figure 40 displays EMG and CMC activation patterns of the rotator cuff muscles during forward reach for a representative subject.

Table 18 – Pearson's R-values between EMG and CMC activation signals during forward reach. Muscles that were active <20% of the trial were excluded from analysis.

Muscle	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Group Mean
Pectoralis Major	N/A	N/A	N/A	N/A	N/A	N/A
Anterior Deltoid	-0.171	0.203	0.551	-0.566	0.078	0.019
Posterior Deltoid	-0.021	-0.444	-0.095	-0.606	-0.780	-0.389
Latissimus Dorsi	N/A	N/A	N/A	N/A	N/A	N/A
Biceps Brachii	N/A	-0.565	0.108	N/A	N/A	-0.228
Triceps Brachii	N/A	-0.538	0.402	N/A	N/A	-0.068
Subscapularis	-0.325	-0.492	-0.090	-0.099	-0.151	-0.231
Teres Minor	-0.422	-0.512	N/A	-0.041	0.554	-0.105
Teres Major	N/A	N/A	0.131	-0.848	-0.402	-0.373
Infraspinatus	-0.712	-0.615	0.401	0.238	0.457	-0.046
Supraspinatus	-0.078	-0.423	0.830	0.031	-0.385	-0.005
Coracobrachialis	0.580	-0.368	-0.575	0.710	0.305	0.130
Across Muscle Mean	-0.164	-0.417	0.185	-0.148	-0.040	

Negative	Weak	Moderate	Strong	Very Strong
<0	0 - 0.39	0.4 – 0.69	0.7 – 0.89	0.9 - 1

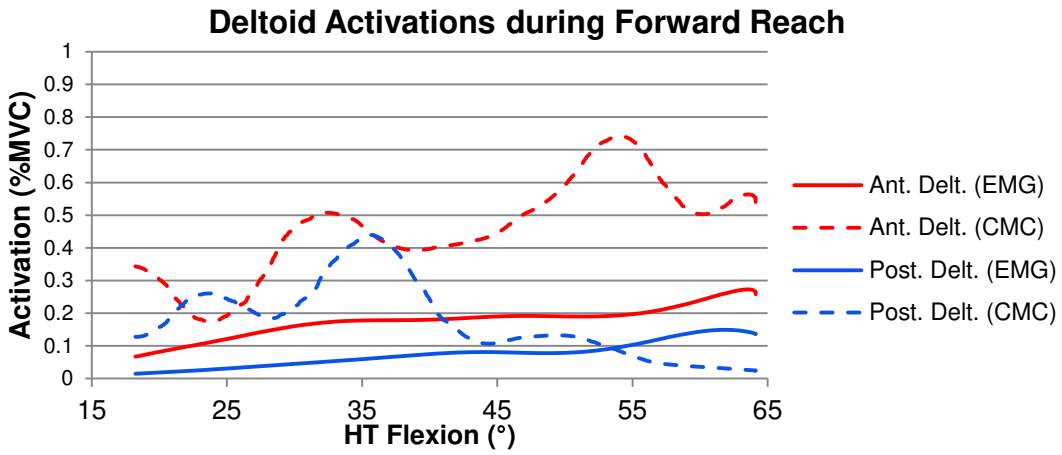

Figure 39 – Muscle activation patterns between EMG and CMC for deltoid during forward reach. Data from one subject displayed. Subject 5: anterior deltoid (r=0.078) and posterior deltoid (r=-0.780).

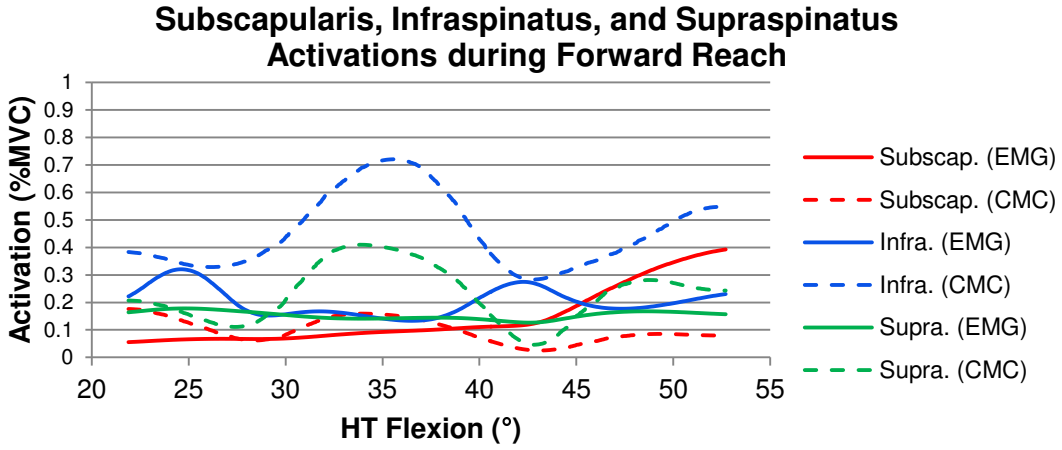

Figure 40 – Muscle activation patterns between EMG and CMC for rotator cuff muscles during forward reach. Data from one subject displayed. Subject 1: subscapularis (r=-0.325), infraspinatus (r=-0.712), and supraspinatus (r=-0.078).

3.5 Summary of Results

Aim 1: To assess the ability of a 3-dimensionsal (3D) musculoskeletal model that employs a prescribed scapular rhythm to track scapular kinematics during a motion that follows the scapular rhythm – shoulder abduction – and a motion that does not - forward reach

Hypothesis 1.1: Clinically meaningful differences (>10°) in ST angle will exist during both motions between a) a model that constrains the scapula to move according to a rhythm and b) the same model modified to allow for natural, unprescribed scapular motion. These differences are expected to be >10° for over 50% of each motion trial. **Outcome:** Clinically meaningful differences (>10°) in 3D ST resultant angle were found during both abduction and forward reach trials. The first portion of **Hypothesis 1.1** would be accepted for all subjects. The group averages for difference in resultant ST angle was >10° for >50% of both motion trials. Thus, the second portion of **Hypothesis 1.1** would be accepted based on a group average for abduction and forward reach.

Hypothesis 1.2: Waveform symmetry and correlation analyses will demonstrate strong shape and trend agreement (Symmetry index and Pearson r>0.7) between the ST angle curves of the two models for all three degrees of freedom (DOF) during both motions. **Outcome:** During abduction trend similarities between models were strong for ST upward rotation and ST anterior tilt, but weak for ST internal rotation. Accordingly, **Hypothesis 1.2** would be accepted for ST upward rotation and ST anterior tilt, but rejected for ST internal rotation during abduction. During forward reach trend similarities between models were generally strong for ST upward rotation and ST internal rotation, but weak for ST anterior tilt. Thus, **Hypothesis 1.2** would be

accepted for ST upward rotation and ST internal rotation, but rejected for ST anterior tilt during forward reach.

Hypothesis 1.3: The magnitude of the maximum and mean differences in ST angle between models for each DOF will be less than 50% of FS model's ST excursion on that DOF during each trial. **Outcome:** For all 3 ST DOFs, differences greater than 50% of the FS model's ST reference ROM on that DOF were present for the majority of subjects during both motions. Consequently, **Hypothesis 1.3** was rejected for abduction and forward reach.

Hypothesis 1.4: For all the kinematics parameters in Hypotheses 1.1, 1.2, and 1.3 the differences between the two models will be greater during forward reach than shoulder abduction. **Outcome:** Mean and maximum differences in ST resultant angle between models were extremely similar between motions. Differences and trends between models were also similar, thus **Hypothesis 1.4** would be rejected.

Aim 2: To utilize 3D musculoskeletal modeling to evaluate the effect of scapular kinematics on the GH torques required during shoulder abduction and forward reach

Hypothesis 2.1: There will be no significant differences in mean and maximum values of GH torques for any DOF between the two models during both motions. Additionally, for each of the 3 DOFs, the GH torques between models will be in functional agreement with each other. **Outcome:** Maximum and mean differences in GH torques between models were relatively small (*see Glossary for definition of small)* for the GH abduction and internal rotation DOFs, but were appreciable for the GH flexion DOF during abduction. Maximum and mean differences in GH torques between models were relatively small for all 3 GH DOFs during forward reach.

Consequently, the first portion of **Hypothesis 2.1** would be rejected for abduction, but accepted for forward reach. Models were in functional agreement for the GH abduction and internal rotation DOFs throughout abduction, however functional disagreement was present for GH flexion DOF. Both models were in functional agreement for all 3 GH DOFs during forward reach. Accordingly, the second portion of **Hypothesis 2.1** would be rejected for abduction but accepted for forward reach.

Hypothesis 2.2: Waveform symmetry and correlation analyses will demonstrate a strong shape and trend agreement (Symmetry index and Pearson $r>0.7$) between GH torques of the two models for all three DOFs during both motions. **Outcome:** Very strong correlations were found for GH torques about the abduction and internal rotation DOFs during abduction. However, a variety of negative, weak, and moderate correlations were found for the GH flexion DOF. **Hypothesis 2.2** would be accepted for the GH abduction and internal rotation DOFs, but rejected for the GH flexion DOF during abduction. Strong correlations were found for GH torques about all 3 DOFs during forward reach. **Hypothesis 2.2** would be accepted for all 3 GH DOFs during forward reach.

Hypothesis 2.3: For all kinetic parameters in Hypotheses 2.1 and 2.2, the differences between the two models will be greater during forward reach than shoulder abduction. **Outcome:** Differences for all kinetic parameters between the two models were not greater during forward reach, thus **Hypothesis 2.3** would be rejected.

Aim 3: To compare muscle activation patterns predicted by a 3D musculoskeletal model that employs a scapular rhythm with recorded electromyography (EMG) during shoulder abduction and forward reach.

Hypothesis 3.1: The on/off timing of the muscle activations predicted by the model will be in agreement with the recorded EMG for at least 50% of the trial duration for both motions trials. **Outcome:** Averaged across all subjects, 5 of 12 muscles achieved >50% activation on/off timing agreement during abduction compared with only 6 of 12 during forward reach. **Hypothesis 3.1** would be rejected for both motions.

Hypothesis 3.2: Waveform symmetry and correlation analyses will demonstrate moderate or greater shape and trend agreement (Symmetry index and Pearson r>0.4) between model-predicted muscle activations and EMG during both motions. **Outcome:** Only 2 of 9 eligible muscles had at least a moderate trend agreement between CMC and EMG during abduction. During forward reach 0 of 10 muscles analyzed had at least a moderate trend agreement between CMC and EMG. Based on these results **Hypothesis 3.2** would be rejected for both motions.

Hypothesis 3.3: For all muscle activation parameters in Hypotheses 3.1 and 3.2, the differences between the two models will be greater during forward reach than shoulder abduction. **Outcome:** Results demonstrated that model-predicted muscle activations were in similar agreement with EMG during abduction and forward reach, thus **Hypothesis 3.3** would be rejected.

Chapter 4

DISCUSSION & CONCLUSIONS

4.1 Aim 1: Scapulothoracic Kinematics

The comparison between model-estimated and directly measured ST kinematics revealed that there were no clinically meaningful differences between the FS model and direct measures; however, significant differences were present between the SR model's ST kinematics and direct measures. This comparison demonstrates that the assumption that the FS model's estimates represent natural ST kinematics is valid. The scapular rhythm employed by the MoBL ARMs model was unable to track natural, unconstrained scapular kinematics to a clinically meaningful degree for the majority of the abduction and forward reach trials. The SR model's ST kinematic errors generally increased with HT elevation and meaningful differences were present in all subjects at HT elevations greater than 54° during both motions. These findings indicate that use of the MoBL ARMS model for applications requiring precise ST kinematics should be limited to motions with less than 54° of HT elevation. Moreover, use of the MoBL ARMS model involving less than 54° of HT elevation should still be cautioned as some subjects achieved clinically meaningful ST kinematic errors at as little as 13° and 3° of HT elevation during abduction and forward reach, respectively.

Scapulothoracic kinematic errors were nearly identical between abduction and forward reach. The mean and maximum 3D ST resultant angle differences between abduction and forward reach were each within 2° of one another. Additionally, the duration of each motion trial with differences >10° were nearly identical (74.8% of

abduction vs. 73.8% of forward reach). These findings provide evidence that the scapular rhythm employed by the MoBL ARMS model does not track scapular kinematics substantially worse for motions that do not follow the scapular rhythm than those that do.

Errors in 3D ST resultant angle are not distributed in the same way across the 3 ST DOFs for abduction versus forward reach. The largest differences between models were found for ST internal rotation during abduction and ST anterior tilt during forward reach, respectively. While ST upward rotation differences between models were substantial, the relative differences were smaller than the other 2 DOFs due to the larger excursion about the upward rotation axis. Kinematic trend analysis revealed very strong relationships between models for ST upward rotation during both motions. This result is expected as the scapular rhythm regression equations developed by de Groot and Brand (2001) reported ST upward rotation as possessing the strongest significant positive relationships with humeral elevation[8]. These findings also can be explained from a functional standpoint as upward rotation of the scapula actively contributes to raising the upper limb to the desired end point during HT elevation tasks. Trends for ST internal rotation were considerably weaker during abduction with 3 of 5 subjects reporting negative correlations. The data used to establish the scapular rhythm regression equations demonstrated that internal rotation was less predictable as a function of humeral elevation than the other 2 ST DOFs[8]. The ST internal rotation relationship between models was much stronger during forward reach. For ST anterior tilt, strong trends were found between models for abduction, while less consistent trends were present during forward reach. These findings for ST internal rotation and anterior tilt may be a result of the scapula being constrained to follow the contour of

the thoracic cage during HT motion. Throughout shoulder abduction, ST anterior tilt is more rigorously constrained while ST internal rotation is more independent. However, during forward reach the motion of the scapula around the lateral surface of the thoracic cage mandates ST internal rotation to be more dependent while ST anterior tilt is less constrained.

Subjects exhibited similar trends and magnitudes of ST kinematic errors throughout each motion, but there was appreciable inter-subject variability. The kinematic errors tended to be smaller for subjects with segment sizes closer to those of the 50% male[22,23] used in the creation of the default MoBL ARMS model[9,13]. However, no clear relationship existed between errors reported by the Scaling and Inverse Kinematics tools within OpenSim and deviation from the default model size. The three female subjects possessed larger ST kinematic errors than the two male subjects. It appeared that the more substantial kinematic errors in the female subjects could be attributed to greater errors on the ST internal rotation DOF than their male counterparts; errors on the ST upward rotation and anterior tilt DOFs were similar between genders. While total excursion and patterns were comparable, the female subjects maintained consistently less ST internal rotation throughout each motion than males. The scapular rhythm regression equations were developed on a predominantly male group of subjects[8] which may explain why females had larger ST kinematic errors in this study. These findings provide evidence that the model may be less able to accurately track scapular motion for individuals whose ST kinematics deviate further from those of the young, healthy, predominantly male subject pool used to develop the regression equations for the scapular rhythm. Accordingly, the use of the

model on patient populations known to exhibit abnormal scapular kinematics should be cautioned if accurate ST kinematics are required.

4.2 Aim 2: Glenohumeral Kinetics

The comparison between model-estimated and directly measured GH torques revealed significant differences between direct measures and both models. Therefore, the discussion of comparison between the FS and SR models presented below should be interpreted with caution as neither model faithfully recreated the GH kinetics occurring in the body. Functional agreement for joint torques is a key criterion for evaluating how well the model is recreating the mechanics that occurred in vivo. If the scapular rhythm positions the model's GH joint in a configuration that requires torques of the opposite function than the subject actually experienced (e.g. abduction vs. adduction), the CMC algorithm will be inclined to overestimate muscle activity from antagonists and underestimate agonist activity. Throughout both abduction and forward reach the FS and SR models were in functional agreement for all 3 GH DOFs for nearly all subjects. Two subjects experienced periods of functional disagreement for GH flexion torques during shoulder abduction. The function about the GH flexion torque axis is very sensitive during shoulder abduction. Small aberrant movements of the arm out of the frontal plane can result in the model experiencing GH flexion/extension torques opposite of the function that occurred in vivo. This type of functional disagreement is an example of how errors in ST kinematics can manifest themselves as errors in GH kinetics and ultimately influence model-predicted muscle activations.

A second important factor to ensure that a musculoskeletal model is accurately representing the true mechanics of the body is the pattern of requisite joint torque. The

SR model demonstrated strong trend agreement with the FS model during both motions on nearly all 3 GH DOFs in this study, however neither model was in strong agreement with directly calculated GH torques. The only exceptions were for torques about the GH flexion DOF for some subjects during abduction. Trends in GH flexion torques were predominantly weak or poor because the minimal amount of motion about this axis during shoulder abduction makes torques highly sensitive to small, out of frontal plane changes in arms position.

There is a lack of consensus as to what classifies a given joint torque as small or large. Appropriate classification is dependent on the biomechanical scenario being investigated. This study assessed differences in GH joint torques between two models and ultimately would like to determine if these differences would significantly impact model-predicted muscle activations. Thus, in the context of this study a torque difference will be considered small if it is not expected to drastically affect the model-predicted results. To estimate the effect of joint torque differences on muscle activations, a calculation was performed (*see Glossary*) and estimated that torques less than 3 Nm should not result in >10% change in muscle activation. Therefore, torque differences of 3 Nm or less were defined as "small" for interpretation purposes in this study. Despite the substantial ST kinematic differences, the GH kinetics were comparable between models for both abduction and forward reach. Mean differences in GH torques between models were generally very small for both motions. The only exceptions were the two subjects with functional disagreement on the GH flexion DOF during abduction. Although the maximum differences in GH torques between models were more substantial for some DOFs when expressed as a percentage of the FS model's maximum torque, these differences translate to predominantly small errors

in raw torque (<3.5 Nm for all subjects and <2 Nm for majority of subjects). With the SR model in functional agreement with the FS joint torques, kinetic errors of this magnitude will impact model-predicted muscle activations, but should only lead to a modest overestimation/underestimation of agonist muscle activity without engendering excessive antagonist activity.

The differences in requisite GH torques between models were generally comparable between abduction and forward reach. In terms of functional agreement, mean and maximum differences, and trends, kinetic results were similar between motions for the GH abduction and internal rotation DOFs. However, differences in GH flexion torques between the FS and SR models were observed between motions. Functional agreement between models was better during forward reach than abduction. The maximum and mean differences averaged across subjects were larger during abduction than forward reach. The correlations between models were stronger for forward reach than abduction for GH flexion torques. Considering that the ST kinematic errors were nearly identical between motions, it is not completely unexpected that the GH kinetic errors would also be fairly similar between abduction and forward reach. This finding provides further evidence that the SR model is equally capable for analyzing shoulder motions that follow the scapular rhythm and those that generally do not.

4.3 Aim 3: Muscle Activations

The results for activation on/off timing agreement between CMC and EMG indicate that the model may be able to recreate the activations of muscles which drive a motion, but is not sensitive enough to reliably predict the activities of those with subtler and more nuanced functions. Prime movers for HT elevation tasks—deltoid

and supraspinatus—were on for the majority of both motions according to EMG and showed high on/off timing agreement with CMC. It is expected that both EMG and CMC would require significant activity from these muscles as they are agonists for both motions. Furthermore, previous modeling research has shown that the deltoid's contribution to net shoulder abduction moment vastly exceeds that of all other muscles[14]. Stabilizing muscles, such as the rotator cuff, were primarily on according to EMG, but demonstrated a wide range of agreements with CMC. The prevalence of small, stabilizing muscles across the GH joint provides a sort of functional redundancy in the model as many different muscles can provide the same function allowing for several possible combinations of muscle activities yielding a valid CMC solution. Muscles with little EMG activity displayed a variety of timing agreements across subjects. In cases of poor agreement, inactive muscles according to EMG (e.g. teres major), may have been erroneously utilized by CMC instead of a muscle with a similar function that was actually on and performing the desired action. This type of timing disagreement for inactive muscles may be due to the previously mentioned functional redundancy within the model.

In the current study, muscle activation on/off timing agreement between CMC and EMG averaged across all muscles ranged from 48.0% and 64.6%. Knarr et al. (2014) investigated timing agreement for the activity of lower extremity muscles during gait of individuals after stroke[40]. Results from that study found agreements between the model and EMG ranging from 50%-80% which were generally higher than those reported in this study. The prime movers and some stabilizer muscles in the current study displayed individual agreement values similar to those in the previous literature; however, muscles with little EMG activity tended to have inferior timing

agreement. In Knarr et al. (2014), only muscles possessing considerable EMG activity during gait were investigated. Therefore, the greater overall timing agreement found in that study may be due to the fact that muscles with little or no EMG activity were not examined. Other potential reasons for the current study reporting agreement values lower than previous literature may lie in fundamental differences between the upper and lower extremities. In the lower extremity, the pelvis provides a fairly rigid base segment for the lower limb to rotate about. By contrast, the scapula in the upper extremity is far from rigid, as it is only attached to the clavicle, and is free to float over the thoracic cage. Since the primary responsibility of the scapula is to support the humerus, the muscles crossing the shoulder not only drive upper extremity motion, but also must play an enormous role in stabilizing the GH joint, thus predicting their activations with a model may be more challenging than it is for muscles of the lower extremity.

Correlations between EMG and CMC activation patterns were poor. The prime movers did not exhibit stronger trends than the stabilizing musculature. These findings provide evidence that the model may not be sufficiently robust to accurately predict activations for the individual muscles of the shoulder. Computed muscle control results demonstrated far more fluctuation in muscle activation signal compared to EMG. Excessive oscillations in muscle activations patterns are likely due to errors in model-predicted GH joint torques compared with direct measures. Excessive oscillations in muscle activation patterns may also be a consequence of the functional redundancy within the upper extremity and the model. During CMC, as a muscle's activation level increases the optimizers penalizes its use more heavily. Should a muscle actuator's activation become too high, the CMC algorithm will look to utilize a

different muscle with a similar function to achieve the same desired joint torque. Therefore, the fluctuations in activations may be the CMC optimizer frequently switching between muscles with similar function which provide different, but equally valid solutions. The use of a different CMC optimization cost function may provide more stable muscle activation predictions as previous upper extremity modeling research has found that different cost functions may improve simulation results[42].

Computed muscle control generally overestimated the degree of muscle activation compared to EMG. One potential explanatory factor is the ease with which the model's muscle actuators can rotate the humerus. The relatively small amount of inertia about the long-axis of the humerus coupled with the lack of external forces on the distal end of the extremity enables relatively small muscle forces to cause large GH internal/external rotation joint accelerations. In order to counteract these joint accelerations, CMC may require more muscle activity from the GH internal/external rotators leading to the higher muscle activations than those measured by EMG. A second possible explanation could be the lack of subject-specific strength scaling. Subject-specific muscle parameters have been shown to alter model-predicted results[43]. For all subjects, the default muscle properties from MoBL ARMS model were used. However, most of subjects in this study were smaller than and likely had less muscle mass than the 50th percentile male used as default in the MoBL ARMs model. Therefore, the use of overly strong default muscle properties for the smaller subjects in this study would actually be more apt to result in an underestimation of muscle activations.

Results for muscle timing agreement and activation correlations between models were not significantly better for abduction than forward reach. On/off timing

agreement averaged across muscles was nearly identical (abduction: 53.4% vs. forward reach: 54.3%), the number of muscles with >50% agreement was comparable, and correlations between EMG and CMC were only slightly stronger during abduction than forward reach. These findings are not unexpected as the kinematic and kinetic differences between motions were similar. Muscle activation on/off timing agreement and correlations between CMC and EMG did not appear to be related to ST kinematic errors. Subjects with smaller ST kinematic errors did not have markedly improved muscle activation on/off timing or trend agreements between CMC and EMG. Results from CMC are dependent on a multitude of factors ranging from muscle properties (e.g. maximum isometric force, force-length, force-velocity, etc.), model geometry (e.g. moment arms), to optimization parameters (e.g. cost function). Considering the lack of a clear link between ST kinematic error magnitude and improved activation timing agreement, it can be inferred that errors in scapular motion do not completely dictate simulation outcomes and that other factors play a substantial role in determining model results.

4.4 Limitations & Future Work

This study has shown that the MoBL ARMS model possesses significant ST kinematic errors and demonstrated that the model is limited in its ability to predict muscle activations. Due to the many other parameters that influence model simulations, this study was unable to ascertain exactly how much of the errors in muscle activation could be directly attributed to ST kinematic errors. Future work investigating the isolated effect of some of these other factors could be performed to better understand how sensitive model-predicted muscle activations are to each. The use of subject-specific muscle parameters has been shown to improve model-predicted

muscle activity in the lower extremity[40,43], however the effects of these changes have not be examined for upper extremity modeling. Additionally, research into the choice of CMC optimization cost function would also be beneficial as it is unknown if the CMC cost functions developed for lower extremity modeling are ideal for upper extremity applications and previous upper extremity modeling literature has shown that choice of optimization cost function influences simulation results[42].

The primary assumption in this investigation is that the ST kinematics tracked by the AMC represented true scapular motion. Currently, a clinically applicable gold standard for measuring dynamic scapular orientation does not exist. In lieu of a reference standard the AMC is a recommended method to estimate scapular motion[44]. The accuracy of the AMC has been validated for single plane HT elevation and thus the motions chosen for evaluation in this study primarily involved planar HT elevation[7,45–49]. Additionally, the AMC has been shown to perform less accurately at HT elevations above approximately 90-120° due to substantial soft tissue motion over the acromion at higher elevations[5,7,46,49]. To mitigate the effect of this source of error the analyses in this study were predominantly constrained to HT elevations below 90°. A second limitation of this study was slight differences in trunk segment orientation between the SR and FS models. A major limitation of the analyses in this study was that the model-estimated torques were not in agreement with directly calculated GH torques. This lack of agreement can likely be attributed to the model's elaborate system of joints and non-physiological body segments to represent the shoulder complex. Should any of the model's joint axes become aligned during a motion, a gimbal lock scenario is encountered which can lead to errors in joint torques. Despite the fact that the model-estimated torques were in poor agreement with direct measures,

the comparisons between models are still valid as both models suffer from the same limitations. The errors in model-estimated GH torques will affect the CMC muscle activation predictions and thus interpretation of the results from Aim 3 should take this factor into account. Future work should be dedicated to improving the model's GH torque estimates. Due to the non-negligible amount of trunk motion throughout each trial, the trunk DOFs could not be held constant and still allow the SR model to represent physiological ST motion. In an effort to enforce identical trunk motion between the FS and SR models, higher tracking weights were assigned for the markers of the trunk segments during inverse kinematics optimization in OpenSim. The inverse kinematics results showed that the trunk orientation between models was similar (within ~5°), but not identical. These minor differences in trunk motion between models likely had no effect on the ST kinematic results, but positioned the models at marginally different orientations relative to gravity which would have a slight, but non-trivial, influence on the inverse dynamics results for requisite GH torques. Lastly, the on/off activation threshold was chosen as 5% of MVC for each muscle. This threshold was chosen with the assumption that activations above this value were volitional in healthy subjects and that activations below 5% could possibly be attributed to electrical noise or electrode motion.

4.5 Conclusions

This study was the first to provide a detailed analysis of scapular motion in musculoskeletal modeling of the shoulder and upper extremity. Results demonstrated that the scapular rhythm employed by the model is not capable of accurately replicating scapular kinematics to a clinical meaningful degree for motions that generally follow the scapular rhythm and those that do not. Consequently, the MoBL

ARMS model is not recommended for applications requiring accurate ST kinematics. Despite the substantial errors in ST kinematics, only modest errors in GH torque were observed between models, however, neither model reliably recreated the GH torques occurring in vivo. The errors in model-estimated GH torques will affect the agreement between CMC results and EMG and thus should be taken into account when interpreting these results. Model-predicted muscle activations displayed moderate agreement with EMG for on/off timing, but correlations between CMC and EMG activations were poor. Agreement between the model and EMG was greater for prime movers than for stabilizing muscles and those with little EMG activity. These findings show that the model may be able to recreate the activations of muscles which drive a motion, but is not sensitive enough to reliably predict the activities of those with subtler and more nuanced functions. Muscle activation agreement between CMC and EMG did not appear to be related to ST kinematic errors. Considering the lack of a clear link between ST kinematic error magnitude and improved activation agreement, it can be inferred that other factors play a substantial role in determining model results. These findings can aid researchers in choosing which future applications are suitable for investigation with the MoBL ARMS model and assist in interpretation of model results.

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Appendix A

GLOSSARY

Requisite Torque:

The joint torque required for the model to maintain the posture in every time point throughout a motion trial.

"Small" Torque Difference:

Torque differences of 3 Nm or less were defined as "small" for this study.

There is a lack of consensus as to what classifies a given joint torque as small or large. Appropriate classification is dependent on the biomechanical scenario being investigated. This study assessed differences in GH joint torques between two models and ultimately would like to determine if these the differences would significantly impact model-predicted muscle activations. Thus, in the context of this study a torque difference will be considered small if it should not drastically affect model-predicted results. To estimate the effect of joint torque differences on muscle activations, a calculation (see below) estimated that torques less than 3 Nm should result in <10% changes in muscle activations. Therefore, torque differences of 3 Nm or less were defined as "small" for interpretation purposes in this study.

With the exception of 2 subjects with functional disagreement of torques between models, all subjects' maximum torque differences were less than 3.5 Nm with most maximum differences being <2.0 Nm. Using this maximum torque difference of 3.5 Nm for GH abduction torque during shoulder abduction as an example it is possible estimate the effect of torque difference on model-predicted activations in a

worst case scenario. An assumption is made that the prime movers for abduction—deltoid and supraspinatus—muscle actuators are the only ones utilized to generate the 3.5 Nm requisite abduction torque. On average the model reports each of these muscles as having a ~2.0 cm abduction moment arm throughout the motion and approximately 2800 N of combined maximum isometric force. This means that the largest maximum torque difference between models (3.5 Nm) would result only in a ~5% difference in activation for these muscles. This example neglects force-length and force-velocity relationships, but even if the muscles operating at sub-optimal lengths which permitted only 50% of their respective maximum isometric force, the difference in activation would be ~10%. This quick example makes several assumptions, but serves to show that the torque differences of magnitude observed in this study should not strongly influence model-predicted muscle activations.

Functional Agreement/Disagreement:

Functional agreement refers to instances throughout a motion trial where both the FS and SR models require torques of the same function (i.e. both models report positive values or both negative values) about a given DOF. Functional disagreement refers to instances where the models require torques of opposing functions (i.e. one model reports positive values while the other reports negative values and vice versa) about a given DOF.

For example, for the GH Abduction DOF if the requisite torque was positive for both models, the models would be in functional agreement as both require an abduction (i.e. not adduction) torque. Should one model report a positive and the other report a negative torque, this indicates that the models require torques of opposing

functions (i.e. adduction vs. abduction) and the models would be considered to be in function disagreement.

Activation On/Off Timing Agreement/Disagreement:

Functional agreement refers to periods of a motion trial where the on/off status of a muscle actuator are the same between CMC and EMG (i.e. both 'on' or both 'off'). Functional disagreement refers to periods of a motion trial where the on/off status of a muscle actuator are not the same between CMC and EMG (i.e. one 'on' and one 'off').

Appendix B

IRB APPROVAL DOCUMENTATION

RESEARCH OFFICE

210 Hullihen Hall
University of Delaware
Newark, Delaware 19716-1551
Ph: 302/831-2136
Fax: 302/831-2828

DATE: April 25, 2016

TO: Robert Tyler Richardson, MS
FROM: University of Delaware IRB

STUDY TITLE: [900968-1] Evaluation of the Effect of Scapular Motion on Musculoskeletal Modeling of the Shoulder and Upper Extremity

SUBMISSION TYPE: New Project

ACTION: DETERMINATION OF EXEMPT STATUS
DECISION DATE: April 25, 2016

REVIEW CATEGORY: Exemption category # *(4)*

Thank you for your submission of New Project materials for this research study. The University of Delaware IRB has determined this project is EXEMPT FROM IRB REVIEW according to federal regulations.

We will put a copy of this correspondence on file in our office. Please remember to notify us if you make any substantial changes to the project.

If you have any questions, please contact Nicole Farnese-McFarlane at (302) 831-1119 or nicolefm@udel.edu. Please include your study title and reference number in all correspondence with this office.

HUMAN SUBJECTS PROTOCOL
University of Delaware

Protocol Title: Evaluation of the Effect of Scapular Motion on Musculoskeletal Modeling of the Shoulder and Upper Extremity

Principal Investigator
Name: Robert Tyler Richardson
Department/Center: Kinesiology and Applied Physiology (KAAP)
Contact Phone Number: 484-886-9908
Email Address: trich@udel.edu

Advisor (if student PI):
Name: James G. Richards
Contact Phone Number: 302-831-0515
Email Address:jimr@udel.edu

Other Investigators:

Investigator Assurance:

By submitting this protocol, I acknowledge that this project will be conducted in strict accordance with the procedures described. I will not make any modifications to this protocol without prior approval by the IRB. Should any unanticipated problems involving risk to subjects occur during this project, including breaches of guaranteed confidentiality or departures from any procedures specified in approved study documents, I will report such events to the Chair, Institutional Review Board immediately.

1. **Is this project externally funded?** ☐ YES ☒ NO

If so, please list the funding source:

2. Research Site(s)

☒ University of Delaware

☐ Other (please list external study sites)

Is UD the study lead? ☒ YES☐ NO (If no, list the institution that is serving as the study lead)

3. **Project Staff**
Please list all personnel, including students, who will be working with human subjects on this protocol (insert additional rows as needed):

NAME	ROLE	HS TRAINING COMPLETE?
Robert Tyler Richardson	Principle Investigator	YES
James Richards	Advisor	YES

4. **Special Populations**
Does this project involve any of the following:

Research on Children?
No.

Research with Prisoners?
No.

If yes, complete the Prisoners in Research Form and upload to IRBNet as supporting documentation

Research with Pregnant Women?
No.

Research with any other vulnerable population (e.g. cognitively impaired, economically disadvantaged, etc.)? please describe
No.

5. **RESEARCH ABSTRACT** Please provide a brief description in LAY language (understandable to an 8th grade student) of the aims of this project.

The goal of this study is understand how well a computer model can represent natural shoulder motion and function. The motion of the trunk and arm as well as the activities of shoulder muscles will be measured during two arm motions. Simulations will be performed using a computer model of the body. This model will have simulated bones, joints, and muscles to represent those same parts of the human body. The model will try to reproduce the motion of each of the body segments and the activity from each of the shoulder muscles during each arm motion. The model results will be compared with the collected data. The results from this study will help give an understanding of how well a computer model can recreate the actual human body. The findings from this study will help researchers choose which future applications are suitable for investigation with this type of model.

6. **PROCEDURES** Describe all procedures involving human subjects for this protocol. Include copies of all surveys and research measures.

<u>The data collected in Dr. Garry Quinton's initial study includes:</u>
-a brief medical history focused solely on shoulder function and past/current treatments for

shoulder dysfunction
-a combination of surface and fine-wire electromyography
-motion capture data of the dynamic orientation of the trunk and upper extremity segments during 5 shoulder motions

The data from Dr. Garry Quinton's initial study (see above) will be used in this retrospective study. All subject data will be de-identified for this study.

Musculoskeletal modeling simulations will be performed using the previously collected data. Simulations will be performed on the MoBL ARMS dynamic musculoskeletal model of the upper extremity using the OpenSim 3.1 software package. Two versions of the MoBL ARMS model will be investigated. The first version will be the stock, published, and publicly available MoBL ARMS model which prescribes scapular kinematics to depend on a set of regression equations known as the scapular rhythm (SR). The second version of the model will be a modified version of the SR model which possessed a free scapula (FS) and allowed for unprescribed ST kinematics. The result was two models – one that prescribes scapular kinematics, SR, and one that allows for unprescribed or free scapular kinematics, FS.

Model segments will be scaled based on the marker positions when each subject is in a resting neutral position. Model segments will also be scaled based on subject mass. The Inverse Kinematics tool within OpenSim will be used to calculate model joint kinematics from the motion capture marker data. All shoulder abduction inverse kinematics trials will begin with the arm at neutral resting position and will end at maximum humerothoracic abduction. All forward reach inverse kinematics trials will begin with the arm at neutral resting position and ended at maximum forward reach position. For both motions, the FS model will use motion capture markers to calculate scapulothoracic joint angles, whereas the SR model will let the model's scapular rhythm coupling constraints determine scapulothoracic joint angles based on humeral elevation. In order to determine anatomical joint orientations (i.e. not model joint orientations), OpenSim's Body Kinematics Analysis tool will be used on the inverse kinematics results to calculate the anatomical humerothoracic, scapulothoracic, and glenohumeral joint angles.

Using the inverse kinematics results, the Inverse Dynamics tool within OpenSim will be used to compute the requisite glenohumeral joint torques during both motions.

Finally, the Computed Muscle Control algorithm within OpenSim will be used to estimate muscle activation patterns during the inverse kinematics results for both motions. These model-estimated muscle activations will be compared with measured electromyography data.

7. STUDY POPULATION AND RECRUITMENT

Describe who and how many subjects will be invited to participate. Include age, gender and other pertinent information.

This is a retrospective study that will use de-identified data from the dissertation of Dr. Robert Quinton. There were five adult subjects in this study – 2 males and 3 females – between the ages of 18-65.

Attach all recruitment fliers, letters, or other recruitment materials to be used. If verbal recruitment will be used, please attach a script.

N/A

Describe what exclusionary criteria, if any will be applied.

-Inability to understand and consent to participation in this study

-Inability to follow verbal and visual directions of the researcher to perform 5 shoulder motions
-Physically unable to sit upright in a chair for 2 hours with periodic breaks
-Have had botulinum toxin ("BOTOX") injection in any shoulder muscles in the past 12 months
-Have had a tendon release or tendon transfer surgery on the shoulder

Describe what (if any) conditions will result in PI termination of subject participation.
N/A

8. **RISKS AND BENEFITS**

List all potential physical, psychological, social, financial or legal risks to subjects (risks listed here should be included on the consent form).
None. Data has been de-identified.

In your opinion, are risks listed above minimal* or more than minimal? If more than minimal, please justify why risks are reasonable in relation to anticipated direct or future benefits.
No.

*(*Minimal risk means the probability and magnitude of harm or discomfort anticipated in the research are not greater than those ordinarily encountered in daily life or during the performance of routine physical or psychological examinations or tests)*

What steps will be taken to minimize risks?
Data has been de-identified.

Describe any potential direct benefits to participants.
None.

Describe any potential future benefits to this class of participants, others, or society.
None.

If there is a Data Monitoring Committee (DMC) in place for this project, please describe when and how often it meets.
No.

9. **COMPENSATION**

Will participants be compensated for participation?
No direct compensation will be paid as part of this retrospective study. At the time data was collected for Dr. Quinton's study subjects were compensated in the amount of $100.

If so, please include details.
See above.

10. **DATA**

Will subjects be anonymous to the researcher?
Yes.

If subjects are identifiable, will their identities be kept confidential? (If yes, please specify how)
N/A

How will data be stored and kept secure (specify data storage plans for both paper and electronic files. For guidance see http://www.udel.edu/research/preparing/datastorage.html)

All data has been de-identified. Data will be stored on a password protected computer that is either in direct possession of the principle investigator or in a locked cabinet in the principle investigator's office.

How long will data be stored?

Indefinitely

Will data be destroyed? ☐ YES ☒ NO (if yes, please specify how the data will be destroyed)

Will the data be shared with anyone outside of the research team? ☐ YES ☒ NO (if yes, please list the person(s), organization(s) and/or institution(s) and specify plans for secure data transfer)

How will data be analyzed and reported?

The open-source musculoskeletal modeling software program OpenSim will be used to analyze all data collected. Data will be reported in aggregate form only at professional conferences and in professional journals.

11. CONFIDENTIALITY

Will participants be audiotaped, photographed or videotaped during this study?

No.

How will subject identity be protected?

Data will be de-identified.

Is there a Certificate of Confidentiality in place for this project? (If so, please provide a copy).

No.

12. CONFLICT OF INTEREST

(For information on disclosure reporting see: http://www.udel.edu/research/preparing/conflict.html)

Do you have a current conflict of interest disclosure form on file through UD Web forms?

No.

Does this project involve a potential conflict of interest*?

No.

* As defined in the University of Delaware's Policies and Procedures, a potential conflict of interest (COI) occurs when there is a divergence between an individual's private interests and his or her professional obligations, such that an independent observer might reasonably question whether the individual's professional judgment, commitment, actions, or decisions could be influenced by considerations of personal gain, financial or otherwise.

If yes, please describe the nature of the interest:

N/A

13. CONSENT and ASSENT

____ Consent forms will be used and are attached for review (see Consent Template under Forms

and Templates in IRBNet)

____ Additionally, child assent forms will be used and are attached.

____ Waiver of Documentation of Consent (attach a consent script/information sheet with the signature block removed).

_X__ Waiver of Consent (Justify request for waiver)
Retrospective data for this study has been de-identified. The following statements were contained in the original study's consent form:
"The data collected for this study can also be used for future studies"
"…data we collect during this study may be released to other researchers to help with future studies"

14. **Other IRB Approval**
Has this protocol been submitted to any other IRBs?
No.

If so, please list along with protocol title, number, and expiration date.
N/A

15. **Supporting Documentation**
Please list all additional documents uploaded to IRBNet in support of this application.
None.

INFORMED CONSENT TO PARTICIPATE IN RESEARCH

Title of Project: (complete title of the project as it appears on the protocol form)

Principal Investigator(s): Robert Garry Quinton, D.O., James Richards, Ph.D., Freeman Miller, M.D.

You are being invited to participate in a research study. This consent form tells you about the study including its purpose, what you will be asked to do if you decide to take part, and the risks and benefits of being in the study. Please read the information below and ask us any questions you may have before you decide whether or not you want to participate.

WHAT IS THE PURPOSE OF THIS STUDY?

The purpose of this study is to learn more about how muscles work together to perform certain shoulder motions.

- We will look to see how muscles of the shoulder work in people who do not have any shoulder problems.
- We will look to see how muscles of the shoulder work in people with cerebral palsy and brachial plexus birth palsy.
- We will compare any differences and similarities between the ways the muscles work between the three groups of people.
- This research study will be used in a doctoral dissertation by the primary investigator.
- You will be one of approximately 15 participants in this study.

WHY ARE YOU BEING ASKED TO PARTICIPATE?

You are being asked to participate because you are an adult with no history of or current shoulder problems, or you are an adult with shoulder problems as a result of you diagnosis of cerebral palsy, or you are an adult with shoulder problems as a result of your diagnosis of brachial plexus birth palsy (also known as neonatal brachial plexus palsy).

You will be excluded from participating in this study if…

- You do not have the ability to understand and consent to your own participation in this study;
- You do not have the ability to follow verbal and visual directions of the researcher to perform 5 shoulder motions;
- You do not have the physical capability to sit upright in a chair for 2 hours with periodic breaks;
- You have had botulinum toxin ("BOTOX") injections in any of your shoulder muscles in the past 12 months;
- You have ever had a tendon release or tendon transfer surgery on your shoulder

Participant's Initials _____

WHAT WILL YOU BE ASKED TO DO?

As part of this study you will be asked to travel to the KAAP Biomechanics Laboratory at the University of Delaware where the study will take place. Your participation in this study will consist of one session lasting about 3 hours. During this 3-hour session you will undergo electromyography (EMG) testing of the muscles of the shoulder while you perform 5 different arm motions. EMG measures the electrical signal that comes from muscles. By measuring these signals we can learn what muscles are doing during a certain motion, when during that motion they are active, and how active.

EMG collection will be performed by a combination of two methods; fine-wire and surface. Fine-wire EMG collection consists of a very small disposable needle (smaller than the needle employed during a blood draw) that is placed in one muscle then quickly removed. Two wires that are smaller than the width of a human hair will remain with one end in the muscle and the other attached to the EMG machine. The electrical activity of the muscle is recorded through these wires by the EMG machine. At this point a very small electrical signal will be sent from the machine to your muscle which will make the muscle twitch but it will not be painful. This twitch is to help us confirm that the wires are in the correct muscle. Surface EMG collection consists of a small plastic rectangle that is applied to the skin using adhesive tape. This skin EMG sensor records the electrical activity of the muscle that is just below the skin surface. For this study a total of 15 fine-wire EMG sensors and 5 surface EMG sensors will be placed on each subject.

In order to record the arm motions that you will be performing we will use a motion capture system. This system uses reflective dots that we stick to your skin with adhesive tape. The system cameras can only see those reflective dots and cannot see you. The cameras will record the motion of the dots as your arm moves and a computer program matches that motion to the EMG signals from the sensors in your muscles.

Once all of the EMG sensors and the motion capture dots are in place, we will have you perform a maximum muscle activation test for each muscle we are testing (20). For these tests, all you have to do is place your arm in the instructed position, perform the instructed arm motion while resistance is applied to your arm by one of the researchers. Each of these 20 tests takes about 15 seconds. Once all of these tests have been completed we will ask you to perform 5 basic arm motions without any resistance. Each motion will be performed 3 times. Once all of these motions are performed, the EMG sensors and motion capture dots will be removed and your participation in this study will be complete.

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

Possible risks of participating in this research study include …

- There will be some discomfort when the fine-wire needle is inserted into each muscle but it will be quickly removed and the wires that remain in the muscle should not cause any discomfort. All wires will be placed by a trained and experienced physician to minimize discomfort.
- There is a very slight risk of infection from the insertion of the wires. However, all wires will be placed by a trained and experienced physician using sterile technique, this risk is minimal.

Participant's Initials _____

- There may be minimal bleeding from the insertion of the wires but this will easily be stopped with some pressure over the site. However, if you have a problem with bleeding or blood clotting due to a medical condition or a medication you are taking; please notify a researcher prior to participating.
- You may experience some bruising or aching where the wires were inserted the day after but this will resolve quickly on its own.
- The electrical signal sent to the muscle to confirm the wire location has a risk of interfering with any implanted electrical devices such as pacemakers or implanted defibrillators. If you have any of these devices implanted in you, please notify a researcher prior to participating.
- There is a slight risk of having a local allergic skin reaction to the adhesive. However, this reaction usually resolves shortly after the adhesive is removed and the site is washed with soap and water.

WHAT IF YOU ARE INJURED DURING YOUR PARTICIPATION IN THE STUDY?

If you are injured during research procedures, you will be offered first aid at no cost to you. If you need additional medical treatment, the cost of this treatment will be your responsibility or that of your third-party payer (for example, your health insurance). By signing this document, you are not waiving any rights that you may have if injury was the result of negligence of the university or its investigators.

WHAT ARE THE POTENTIAL BENEFITS?

There will be no direct benefits from participating in this study. However, indirectly any study participant or member of the general population who currently or in the future will participate in a physical therapy program for shoulder problems or may require surgery for a shoulder problem may potentially benefit from the information obtained from this study. The data collected for this study can also be used for future studies.

HOW WILL CONFIDENTIALITY BE MAINTAINED?

For this study, no personally identifiable information such as your name, date of birth, or social security number will be collected from you. This means that any data we collect on you will in no way be able to be linked to you personally. The information we collect on your medical history will not have your name or any other identifiable information kept with it. Your medical history and data will only be identified with a random subject number and will only be reported with that number. You will also not be photographed, audiotaped, or videotaped in any way. Since we are not collecting any identifiable information from you, the data we collect during this study may be released to other researchers to help with future studies.

WILL THERE BE ANY COSTS TO YOU FOR PARTICIPATING IN THIS RESEARCH?

There are no costs for participating in this study.

WILL YOU RECEIVE ANY COMPENSATION FOR PARTICIPATION?

To compensate you for your time and travel to participate in this study, you will be paid $100.

Participant's Initials _____

DO YOU HAVE TO TAKE PART IN THIS STUDY?

Taking part in this research study is entirely voluntary. You do not have to participate in this research. If you choose to take part, you have the right to stop at any time. If you decide not to participate or if you decide to stop taking part in the research at a later date, there will be no penalty or loss of benefits to which you are otherwise entitled. Your decision to stop participation, or not to participate, will not influence current or future relationships with the University of Delaware.

FINANCIAL INTEREST(S) OF THE RESEARCHERS:

The investigators conducting this study have no conflicting financial interests to disclose.

WHO SHOULD YOU CALL IF YOU HAVE QUESTIONS OR CONCERNS?

If you have any questions about this study, please contact the Principal Investigator, Dr. Robert Quinton at (302) 831-8212 or rquinton@udel.edu or Jim Richards at (302) 831-2054 or jimr@udel.edu.

If you have any questions or concerns about your rights as a research participant, you may contact the University of Delaware Institutional Review Board at hsrb-research@udel.edu or (302) 831-2137.

Your signature on this form means that: 1) you are at least 18 years old; 2) you have read and understand the information given in this form; 3) you have asked any questions you have about the research and those questions have been answered to your satisfaction; 4) you accept the terms in the form and volunteer to participate in the study. You will be given a copy of this form to keep.

Printed Name of Participant	Signature of Participant	Date
Person Obtaining Consent (PRINTED NAME)	Person Obtaining Consent (SIGNATURE)	Date

Participant's Initials _____