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Influence of Reverse Shoulder Implant Positioning on Patient-Specific Muscle Forces: A Simulation Study

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Abstract

A reverse total shoulder arthroplasty (RTSA) is a common treatment used to stabilize the shoulder and improve range of motion in patients with torn rotator cuff muscles. Shoulder stability relies on the shoulder muscles. With rotator cuff tears, the RTSA enables the deltoid muscle to become the primary stabilizer of the shoulder joint. To improve stability, RTSAs increase the deltoid muscle moment arm and decrease the required torque about the shoulder joint for movement. Currently, there is not standardized, objective method for a surgeon to position an implant on a specific patient. This study is part of a larger, ongoing project to optimize the deltoid muscle force for a population of RTSA patients and create a tool to determine the ideal placement of the implant based on the optimal deltoid force. The current study was a step towards the overall goal by investigating patient-specific muscle model parameters and establishing a framework for understanding deltoid muscle function in RTSA patients. This goal was achieved through a parameter sensitivity study and optimization studies.

Acknowledgements

Thank you to Rehoboth Innovations LLC. and Dr. David Walker for your support of me working on this research project. Dr. Kinney, I will always be grateful for the mentorship and guidance that you have given me over the past three years. Thank you to Elijah Kuska in the Computational Biomechanics Lab for your help and mentorship and paving the way for me. And finally a special thank you to my mom for helping and encouraging me over the last 21 years. To my dad, even though you are no longer with us I know you are still smiling down on me and encouraging me every step of the way.

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1. Introduction

1.1 TSA vs. RTSA

Total shoulder arthroplasty (TSA) is a shoulder joint replacement surgery recommended for patients with severe shoulder injuries including rotator cuff muscle tears and/or severe arthritis [1]. In a TSA surgery, the surgeon replaces the patient's shoulder (glenohumeral) joint with an artificial shoulder joint. Anatomically, the glenohumeral joint is a ball and socket joint that typically allows for six degrees of freedom. In a normal shoulder, the humeral head is the ball component of the joint and the glenoid fossa, a concave section of the scapula, is the socket. The purposes of the TSA procedure are to relieve pain, increase the patient's range of motion (ROM), and improve the patient's ability to complete upper extremity functional tasks [2]. Since its approval in the United States in 2004, surgeons have performed a modification of the TSA called a reverse total shoulder arthroplasty (RTSA) [3]. In a RTSA, the orientation of the ball-and-socket components of the glenohumeral joint are reversed; a glenosphere (ball) is implanted on the glenoid cavity of the scapula and a stem is drilled into the humerus with a concave articulation (socket) on the head of the humerus [4]. The ball and socket in the normal shoulder joint allows for large ROM of the arm. However, because the socket is shallow rather than deep it has limited bony stability. The stability of the glenohumeral joint is derived from strong rotator cuff muscles. When someone with a normal shoulder joint reaches upwards the humeral head translates upwards in the socket. Individuals with weak shoulder muscles, common in this patient population, may experience excessive upward translation and, overtime, this could lead to impingement of the common tendon of the rotator cuff muscles and to rotator cuff muscle tears [1]. The RTSA implant is designed to reduce stress on the shoulder joint and its musculature by increasing stability of the joint by enabling the deltoid muscle to become the primary stabilizer and eliminating upward translation of the head of the humerus [5].

1.2 Deltoid Muscle Function, Force, and Moment Arm

Muscles in the body move limbs by contracting and producing a torque about a joint, which is equivalent to the cross product of the moment arm and the muscle force vectors. The torque produced by the muscle is given by Equation (1).

$$\vec{\tau} = \vec{r} \times \vec{F} \tag{1}$$

In Equation (1), $\vec{r}$ represents the muscle moment arm and $\vec{F}$ is the force produced by the muscle. A RTSA increases the stability of the shoulder by increasing the moment arm of the deltoid. The increased leverage enables the deltoid to compensate for torn rotator cuff muscles without exerting an extremely high amount of force to lift the arm [6]. This is important because the high compensatory deltoid muscle forces could deleteriously contribute to notching of the scapula [7]. The amount of force a muscle produces is also related to its length via the force-length curve (Figure 1). Optimal isometric force (fMo) for a muscle is produced at the muscle optimal fiber length (lMo). The ideal range of the lMo values is between 0.5-1 where the muscle will produce only active force; from 1*lMo to 1.5*lMo the muscle produces passive force which can overstretch the muscle and lead to injury [8].

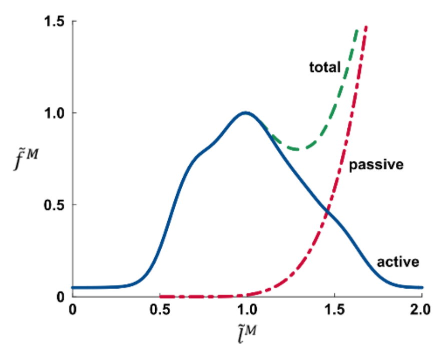

Figure 1. Force-length curve

Anatomically, the deltoid is divided into three subregions: anterior, lateral, and posterior deltoid. These three components have different origins and insertion points and control different shoulder movements. In a normal shoulder, the anterior and lateral deltoid regions have positive abduction moment arms, meaning that they contribute to an abduction moment in the shoulder, while the posterior deltoid has a negative abduction moment arm, indicating that it contributes to and adduction moment in the shoulder [9,10]. In a previous cadaveric RTSA study conducted by Ackland et al., researchers concluded that all three subregions of the deltoid, including the posterior deltoid, had positive abduction moment arms; meaning all three deltoid subregions contributed to an abduction moment in the shoulder. However, while these studies are beneficial there are many limitations to cadaveric studies. A recent study by Walker et al, used in vivo motion capture and fluoroscopy data to create scaled OpenSim models for 14 RTSA subjects and calculated the moment arms for all three deltoid subregions. This study determined that the subregions of the deltoid, post RTSA, function the same as in a normal shoulder [11].

Proper placement of the RTSA implant is complex due to the three translation and three rotation variables, i.e., six degrees of freedom, for both the glenoid and the humerus. There are functional consequences to the deltoid muscle if the RTSA implant is placed such that the moment arm of the deltoid is too short or too long [12]. Previous research suggests that a more medial and inferior placement of the center of rotation (COR) for the RTSA implant will cause the moment arm of the deltoid muscle to be longer and as a result the muscle will produce less force to generate the required torque for shoulder activities [4]. However, even with medial and inferior placement of the implant, many RTSA patients do not regain full ROM and some experience scapular notching leading to bone loss [7]. It is not clear why these negative outcomes occur. One possible reason may be implant positioning. Determination of the optimal placement of RTSA implants is complex given the multiple degrees of freedom of the glenohumeral joint. In addition, implants come in various sizes and shapes to fit the patient's body and there is no standardized or objective assessment surgeons use to fit an implant to a specific patient [4,7,12].

1.3 Simulation and Optimization

There is a need to provide surgeons with an objective method to consistently position RTSA implants in their patients. During surgery, it is possible for a surgeon to test for scapular notching and ROM; however, it is difficult to measure muscle function and forces. Simulation and optimization methods have been used to understand the relationships between the moment arm and functionality of a muscle, to predict muscle forces, and to analyze the impact a surgery may have on a muscle moment arm [13].

Computationally predicted joint contact forces can also be applied to a model to determine the optimal joint implant positioning. A study conducted by Serrancoli et al. used patient-specific muscle parameters to directly validated knee joint contact forces. By doing so Serrancoli et al. also indirectly validated muscle forces because muscle forces are the primary contributors to joint contact forces[14].

The effect of patient-specific muscle parameters on modeling realistic muscle function in the RTSA population is unknown. Calibration of patient-specific muscle parameters via optimization is feasible, but can be time consuming. Due to the fast workplace environment, surgeons cannot afford to wait a long time for optimizations to converge. In order to decrease convergence time and apply these tools clinically, muscle parameter optimizations must be provided a realistic initial guess that is representative of the patient's muscle function. To our knowledge, previous studies have not established guidelines for adjusting muscle parameter values from literature, especially across a group of subjects. Reduction of passive force produced by muscles may be a mechanism for adjusting parameter values and obtaining a reasonable initial guess for future parameter calibration.

This thesis is a first step in a larger, ongoing research study with the goal to optimize the deltoid muscle force for a population of RTSA patients and create a tool that predicts the ideal patient-specific placement of the implant based on the optimal deltoid muscle force. This tool would give orthopedic surgeons the ability to objectively determine the optimal implant placement prior to surgery, decreasing surgery time and increasing the likelihood of success for the procedure. The purpose of this thesis was to investigate patient-specific muscle model parameters and establish a framework to understand deltoid muscle function in RTSA patients. The goal was achieved through two studies. First, a parameter sensitivity study, in which the parameters were adjusted to reduce passive force and the resulting muscle function was analyzed. Second, an optimization study that used a framework to further calibrate the muscle parameters for each subject and then predict the deltoid muscle force for RTSA subjects performing a dynamic arm abduction.

2. Methods

2.1 Experimental Data

Electromyographic (EMG) muscle activity, motion capture, and fluoroscopic data of a group of RTSA patients were previously collected and provided by Rehoboth Innovations, LLC [15,16]. A combination of the fluoroscopic images and the motion capture data were used to define the motion in a musculoskeletal shoulder model for each subject. The models were modified from Holzbaur et al [17] to include the RTSA implant and were scaled to the subject specific dimensions. Each model included two degrees of freedom at the shoulder (abduction/adduction and flexion/extension) and 15 major shoulder muscles (anterior, lateral, and posterior deltoid; four rotator cuff muscles; thoracic, lumbar, and iliac latissimus dorsi; clavicular, sternal, and costal pectoralis major; upper trapezius; and teres major). OpenSim [18], an open-source software, was used to perform biomechanical modeling and to calculate musculotendon lengths, muscle moment arms, and joint moments for the three deltoid muscles that are the focus of this study. These values were calculated for the three subregions of the deltoid for each subject as the subject performed isometric muscle contractions at arm elevation angles of 0, 45, and 90 degrees and dynamic abduction trials.

2.2 Parameter Sensitivity Study

Data from eight RTSA subjects with two different implants were used in a parameter sensitivity study (Table 1). The optimal muscle fiber length (lMo) and tendon slack length (lTs) values of the three subregions of the deltoid were modified from the literature values [19] using scaling factors. Scaling factors were chosen manually with the goal of determining common factors for each deltoid muscle across all of the subjects so that the muscle tendon length (lMtilda) values were within the active force range, 0.5-1.0*lMo, on the force length curve. In addition to the manually chosen scaling factors for the lMo and lTs parameters, the fMo literature values were multiplied by a scaling factor of 2.5 as per common practice [20]. A custom MATLAB code was used to generate plots of the lMtilda values, predicted muscle activation as compared to the normalized experimental EMG data from isometric trials, and force contribution from each of the three deltoid muscles before and after the literature parameter values were modified.

Table 1. Subject demographic information for parameter sensitivity study

Subject	Gender	Age at testing	Implant type	Height [in]	Weight [lbs]
1	M	63	Encore	68	150
2	F	73	Exactech	62	194
3	F	73	Exactech	67	160
4	M	66	Encore	66	144
5	F	76	Encore	62	174
6	F	82	Encore	65	155
7	F	76	Exactech	61	200
8	M	75	Encore	68	186

2.3 Optimization

Data from the models from four of the eight RTSA subjects from the parameter sensitivity study and a custom MATLAB optimization framework created by Rehoboth Innovations LLC. were used to further calibrate the muscle model parameters generated from the parameter sensitivity study and predict the anterior, lateral, and posterior deltoid muscle forces during abduction. A calibration phase and dynamic prediction phase were performed. The calibration phase used a two-level optimization (outer and inner level) [14] to calibrate the muscle model parameters to match the experimental data while the subjects performed isometric contractions of their shoulder muscles at 0°, 45°, and 90° of arm elevation. The experimental data matched was the flexion-extension and abduction-adduction moments of the shoulder. The optimization used a nonlinear least squares outer level optimization and a quadratic programming inner level optimization to reduce the time to convergence. The cost function included terms to minimize muscle activation, reserve actuator contributions, passive muscle forces, and to ensure that the lMtilda values were within the desired range (0.5-1.0*lMo) of the force-length curve. The weights of each of the cost function terms were adjusted until the following conditions were satisfied: the activations were mostly realistic, the force contributions were less than 2.5*fMo, the reserve actuator contributions were minimal, and the lMtilda values were all between 0.5-1*lMo. The outer-level optimized the muscle model parameters and the inner-level optimization used the parameters from the outer-level optimization to optimize the muscle activations needed to match the experimental isometric joint moment data. This occurred in a cyclic pattern until the optimization converged.

Table 2. Subject demographic information for optimization study

Subject	Gender	Age at testing	Implant type	Height [in]	Weight [lbs]
1	F	76	Exactech	61	200
2	F	73	Exactech	62	194
3	M	75	Encore	68	186
4	F	82	Encore	65	155

Following the parameter calibration, the dynamic abduction motion data were used to predict the muscle forces needed to match the joint moments during a dynamic motion. Muscle activations, force contributions, and lMtilda values for all three subregions of the deltoid were predicted using the subject-specific muscle lMo and lTs parameter values from the calibration phase. Matching of the joint moment data was evaluated by comparing the experimentally measured joint moment, the predicted joint moment contributions from the muscles only, the reserve actuators only, and the sum of the contributions from the muscles and the reserve actuators. Predictions were made under two conditions: 1) matching the predicted joint moment contributions and the experimental joint moments about both the flexion-extension and abduction-adduction axes 2) matching the predicted joint moment contributions and the experimental joint moment about only the abduction-adduction axis. RMSE values were calculated for dynamic simulations to compare the muscle contributions to the experimental joint moment for dynamic abduction motion for the four subjects for the two matching conditions. Finally, the results from both conditions were analyzed to determine if the models and optimization were capable of accurately predicting and optimizing the deltoid muscle behavior for all four subjects.

3. Results

3.1 Parameter Sensitivity Study

Common scaling factors were found for the lateral and posterior deltoid muscles, but not for the anterior deltoid (Table 3). The anterior deltoid scaling factor for six of the eight subjects was 1.5, but was 1.79 and 2.22 for the other two subjects. The scaling factors for the lateral and posterior deltoids were 1.13 and 1.25 respectively.

Table 3. Scaling factors for each subject and deltoid muscle component

	Deltoid Component		
Subject	**Anterior**	**Lateral**	**Posterior**
1	1.79	1.13	1.25
2	1.5	1.13	1.25
3	1.5	1.13	1.25
4	1.5	1.13	1.25
5	1.5	1.13	1.25
6	1.5	1.13	1.25
7	2.2	1.13	1.25
8	1.5	1.13	1.25

The ranges for the average lMtilda values across all eight subjects for each of the three deltoid components prior to adjusting the literature parameters were mostly out of the desired range of 0.5-1. Some of the subjects had lMtilda values for one or more deltoid muscle components within the desired range prior to adjustment, but others, as depicted by the ranges, produced large passive forces. There was an inconsistency among the subjects as to which deltoid component required the most parameter adjustment. Table 4 shows the average lMtilda values before and after parameter adjustment for all three deltoid components.

Table 4. Average lMtilda values for deltoid components across all eight subjects before and after parameter adjustment

	Average lMtilda	
Deltoid	**Literature Parameters**	**Adjusted Parameters**
Anterior	1.4-3.05	0.7-0.91
Lateral	0.85-1.05	0.625-0.82
Posterior	0.84-1.08	0.675-0.84

3.2 Optimization

Following the parameter sensitivity study, the lMo and lTs values were further calibrated via the two-level isometric calibration optimization. For all subjects, additional adjustment of the lMo and lTs values were needed in comparison to the original literature parameter values and to the parameter sensitivity study values (Tables 5 and 6). Following the isometric calibration optimization, the lMo and lTs values varied across all four subjects and three subregions of the deltoid. Overall, following the isometric calibration, the lMtilda values were all within the desired range between 0.5-1*lMo and there was general agreement between the experimentally measured EMG data for the three deltoid muscles at 0, 45, and 90 degrees of isometric shoulder abduction. The predicted contribution of the muscles to the flexion-extension and abduction-adduction-joint moment matched the experimental joint moments with small root-mean-square error (RMSE) of 2.0 or less for three of the four subjects. Subject 1 had a larger reserve actuator contribution to the moment at 90 degrees abduction and therefore, the RMSE value was 4.5 for the flexion-extension moment and 5.8 for the abduction-adduction moment.

Table 5. Optimal muscle fiber length values from the literature, following the parameter sensitivity study, and following the optimization study. Literature lMo values were the same for all four subjects across each of the three subregions of the deltoid and were the original input parameters into the parameter sensitivity study. After the parameter sensitivity study the lMo values were the same across all four subjects for the lateral and posterior deltoid muscles, but not for the anterior deltoid due to the different scaling factor used for subject 1. The lMo values varied across all four subjects after the isometric calibration optimization.

Optimal fiber length (lMo)				
Muscle	**Subject**	**Literature (cm)**	**Parameter sensitivity (cm)**	**Calibrated (cm)**
Anterior Deltoid	1	9.8	21.8	35.1
	2		14.7	15.2
	3		14.7	3.1
	4		14.7	12.6
Lateral Deltoid	1	10.8	12.2	23.5
	2		12.2	13.7
	3		12.2	22.4
	4		12.2	11.4
Posterior Deltoid	1	13.7	17.1	13.9
	2		17.1	15.0
	3		17.1	15.4
	4		17.1	13.1

Table 6. Tendon slack length values from the literature, following the parameter sensitivity study, and following the optimization study. Literature lTs values were the same for all four subjects across each of the three subregions of the deltoid and were the original input parameters into the parameter sensitivity study. After the parameter sensitivity study the lTs values were the same across all four subjects for the lateral and posterior deltoid muscles, but not for the anterior deltoid due to the different scaling factor used for subject 1. The lTs values varied across all four subjects after the isometric calibration optimization.

Tendon slack length (lTs)				
Muscle	**Subject**	**Literature (cm)**	**Parameter sensitivity (cm)**	**Calibrated (cm)**
Anterior Deltoid	1	9.7	21.5	39.7
	2		14.6	8.8
	3		14.6	15.7
	4		14.6	10.2
Lateral Deltoid	1	11.0	12.4	4.0
	2		12.4	7.5
	3		12.4	3.1
	4		12.4	9.2
Posterior Deltoid	1	3.8	4.8	3.8
	2		4.8	4.5
	3		4.8	3.3
	4		4.8	4.1

When matching both the flexion-extension and abduction-adduction moments, the RMSE values for the flexion-extension moment were higher for all four subjects than the abduction-adduction moments (Table 7). Subject four had the smallest RMSE values for this condition with 0.10 for the flexion-extension moment and 0.07 for the abduction-adduction moment. The predicted muscle activations for all four subjects when matching both the flexion-extension and abduction-adduction moments showed all three subregions of the deltoid activated at various time points during the duration of the dynamic abduction motion.

Table 7. Root-mean-square error for each of the four subjects when matching both the flexion-extension and abduction-adduction joint moments. A smaller RMSE value means the predicted muscle contribution and experimental data of the muscle contribution to the moment is matched.

Root-mean-square error (RMSE)		
Subject	**Flexion-Extension Moment**	**Abduction-Adduction Moment**
1	4.48	3.40
2	6.10	1.74
3	4.87	3.34
4	0.10	0.07

For all four subjects, when only the abduction-adduction moment was matched, the RMSE values for the abduction-adduction moment decreased (Table 8) in comparison to the RMSE values shown in Table 7. Subject 3 had the lowest RMSE value of 0.02 and subject 2 had the highest, 0.04. When matching the abduction-adduction moment only, the predicted muscle activations for the anterior and posterior deltoid were minimal during the duration of the motion, while the lateral deltoid activation was larger than expected.

Table 8. Root-mean-square error for each of the four subjects when matching only the abduction-adduction joint moments. A smaller RMSE value means the predicted muscle contribution and experimental data of the muscle contribution to the moment is matched.

Root-mean-square error (RMSE)	
Subject	**Abduction-Adduction Moment**
1	0.03
2	0.04
3	0.02
4	0.04

4. Discussion

This study is part of a larger, ongoing project aiming to optimize the deltoid muscle force for a population of RTSA patients and generate a tool that orthopedic surgeons could use to determine the optimal patient-specific placement of the implant. With this tool, a surgeon would have the ability to determine the implant placement prior to surgery, making the procedure more efficient and improving surgical outcomes. The current study was a step towards that overall goal by investigating patient-specific parameters and establishing the framework for understanding deltoid muscle function in RTSA patients. This goal was achieved through the parameter sensitivity and optimization studies.

4.1 Parameter Sensitivity Study

The purpose of the parameter sensitivity study was to determine if reduction of passive force was a mechanism to calibrate muscle model parameters for the RTSA patient population. The goal was to find a method for adjusting parameters that was standardized for as many subjects as possible. We aimed to find common scaling factors for the lMo and lTs parameters to adjust the parameters away from the literature values for all RTSA subjects. Then the isometric calibration optimization used the inputs from the parameters sensitivity study to further calibrate the lMo and lTs values for each subject. Common scaling factors across all subjects were found for the lateral and posterior deltoid, but not for the anterior deltoid. This is not surprising as the range for the lMtilda values of the anterior deltoid using the literature parameter values was very wide. Larger scaling factors for the anterior deltoid were required for two of the eight subjects. It is likely that these two subjects were acting in the extreme passive end of the force length curve. Therefore, these subjects required larger scaling factors to cause the muscle to function on the active force region of the force-length curve.

The parameter sensitivity study was successful in improving the muscle activation and force predictions to align with expected physiological behavior. Using the literature parameter values, many subjects displayed a trend in which the muscle activation was minimal, but the predicted muscle forces in the magnitude of 10^4 N, which is very high (see Figure 2 for a representative subject). As the muscles were not activated, the excessive muscle forces were primarily due to passive force production. High passive force production is not consistent with expected physiological behavior of muscles as previous studies have shown that muscles tend to operate on the active region of the force-length curve [21,22]. Following the parameter adjustment, the muscles operated primarily in the active region of the force-length curve and the muscle activation and force contributions became more realistic for all three deltoid muscles for all eight subjects.

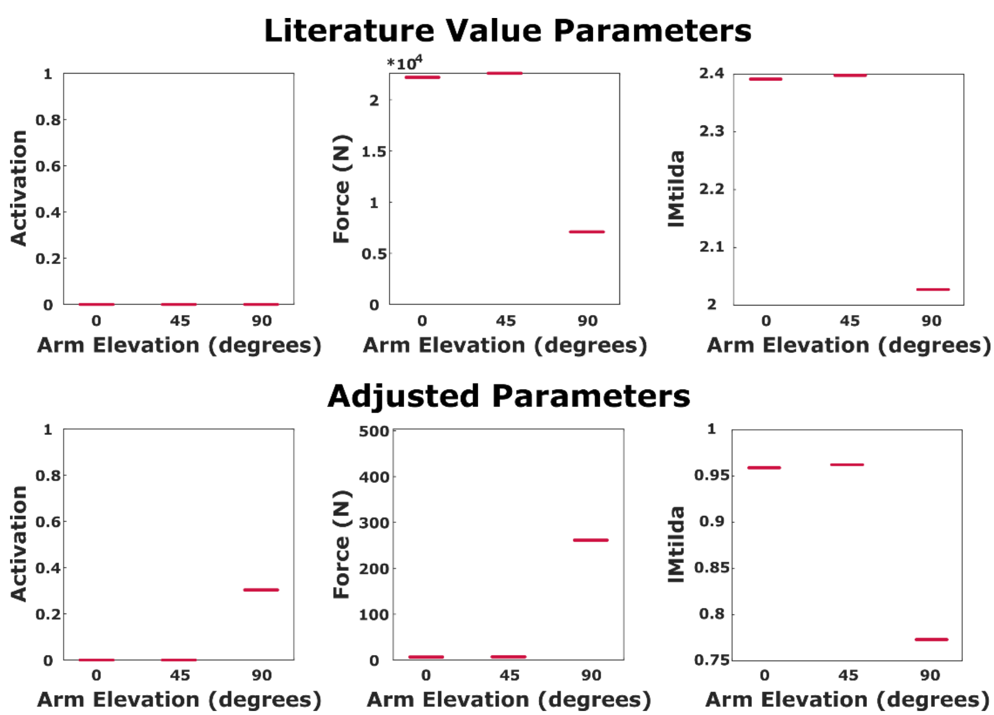

Figure 2. Anterior Deltoid activation, force, and lMtilda for literature and adjusted muscle parameters for a representative subject

With the literature parameter values, the anterior deltoid for a representative subject was not active, 0% activation, at all arm elevation angles but produced up to 20,000 N of primarily passive force (Figure 2). It is not physiologically realistic for a muscle to be inactive and to produce such a large amount of force. In addition, the lMtilda values for this subject reached as high as 2.4*lMo at 45˚ of arm elevation. This lMtilda value is in the extreme passive-force region of the force-length curve (Figure 1) and is not typically represented on the curve, as physiologically, muscles do not produce this much passive force. By adjusting the parameters to reduce passive force production and shift the lMtilda value for each of the three arm elevations into the desired range of 0.5-1*lMo, the anterior deltoid muscle activation and force became more physiologically realistic. When the muscle was inactive it did not produce force, and when the muscle was around 30% active it produced approximately 250 N of force, a reasonable magnitude for the anterior deltoid muscle. The resulting muscle activation after parameter adjustment mimicked the trend observed in the normalized experimental EMG data from the isometric trials. Similar results were observed in all eight subjects in the parameter sensitivity study.

By scaling the lMo and lTs parameters to cause the subregions of the deltoid to operate in the active force region of the force-length curve, the muscle activations and force predictions were improved to align with expected physiological behavior for all eight subjects. However, while the parameter sensitivity study should provide the calibration optimization with a more realistic initial guess to speed up the time to convergence, additional adjustments to the lMo and lTs parameters were necessary. The purpose of these adjustments were to further calibrate the lMo and lTs parameters to the individual subjects, but the results of the parameter sensitivity study should provide the calibration optimization with a more realistic initial guess to speed up the time to convergence. This will facilitate the application of simulation methods in a clinical setting.

4.2 Optimization

The goal of the optimization study was to predict deltoid muscle forces for four RTSA subjects performing a dynamic arm abduction motion. To achieve this goal, an isometric calibration optimization was used to further calibrate the muscle model parameters for all three deltoid muscles and all four subjects individually. Then, the dynamic abduction simulations were performed using the calibrated parameters to determine the deltoid muscle behavior for all four RTSA subjects.

The calibrated lMo and lTs values were adjusted so that the contribution of the reserve actuators was minimized for the isometric trials, the deltoid muscle activation reasonably agreed with the experimental EMG data, and the deltoid muscle force contributions were less than the maximum isometric strength literature values [19]. For three of the four subjects, lMo and lTs parameters were found such that the reserve actuator contributions at all three arm elevations was minimized. Figure 3 shows the flexion-extension and abduction-adduction joint moments for Subject 4. This subject had the lowest RMSE values after the isometric calibration of all of the subjects. This is represented by this plot indicating that the reserve actuator contribution to the joint moment (green line) is approximately zero at all three arm elevation angles and the prediction of muscle contributions (red dashed line) matched closely with the experimental ID moment data (black line). Subject 3 displayed similar reduction of reserve contribution to Subject 4. Subject 1 followed this trend except for at 90˚ arm elevation and Subject 2 was unable to turn off the reserve actuator contribution to the flexion-extension joint moment at 45˚ arm elevation.

The predicted muscle activations did not perfectly match up with the normalized, experimental surface EMG data recorded for the three deltoids. Although this is a limitation of our results, EMG data is very noisy and surface EMG has the tendency to pick up signals from different muscles which can generate interaction of signals [23]. Because the optimization in this study focuses on the three subregions of the deltoid only and not the muscle activity of nearby shoulder muscles such as the upper trapezius this is most likely the cause of the error. Despite this limitation, for all four subjects the lateral deltoid had the highest predicted force contribution at each of the three isometric arm elevations. This is consistent with what has been reported in the literature for reverse total shoulder subjects[6,11,12]. Therefore, we have confidence that the calibration optimization was able to predict realistic muscle forces in the RTSA subjects.

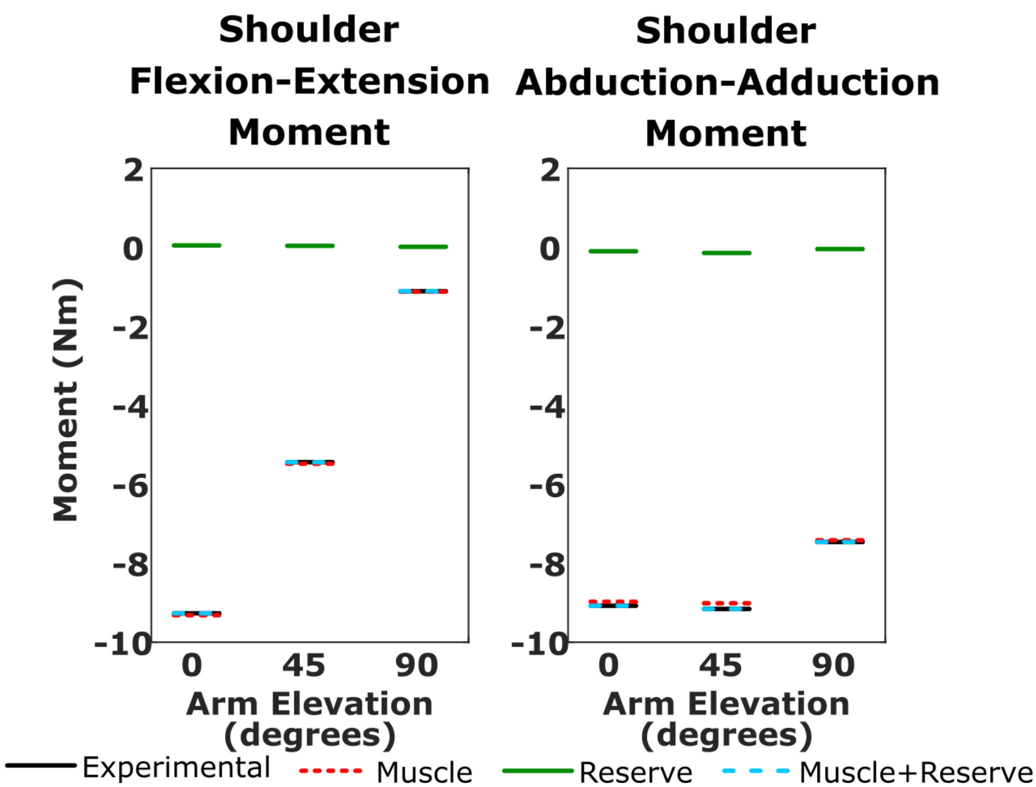

Figure 3. Flexion-extension and abduction-adduction joint moments for the isometric abduction motion at 0, 45, and 90˚ for Subject 4. The goal of the isometric calibration optimization was to determine values for the lMo and lTs muscle parameters to minimize the reserve actuator contribution (green) and match the predicted muscle contribution (red dashed) with the experimental joint moments (black).

With the dynamic simulations, the RMSE values for all four subjects when matching only the abduction-adduction joint moment were all very small (≤0.04). This indicates that the shoulder models, optimization framework, and muscle parameters were sufficient to match the predicted muscle contribution to the experimental joint moment data when matching only the abduction-adduction moment (Figure 4). However, the activations for the anterior and posterior deltoid muscles were minimal (Figure 5), which is known not to be physiologically accurate[16].

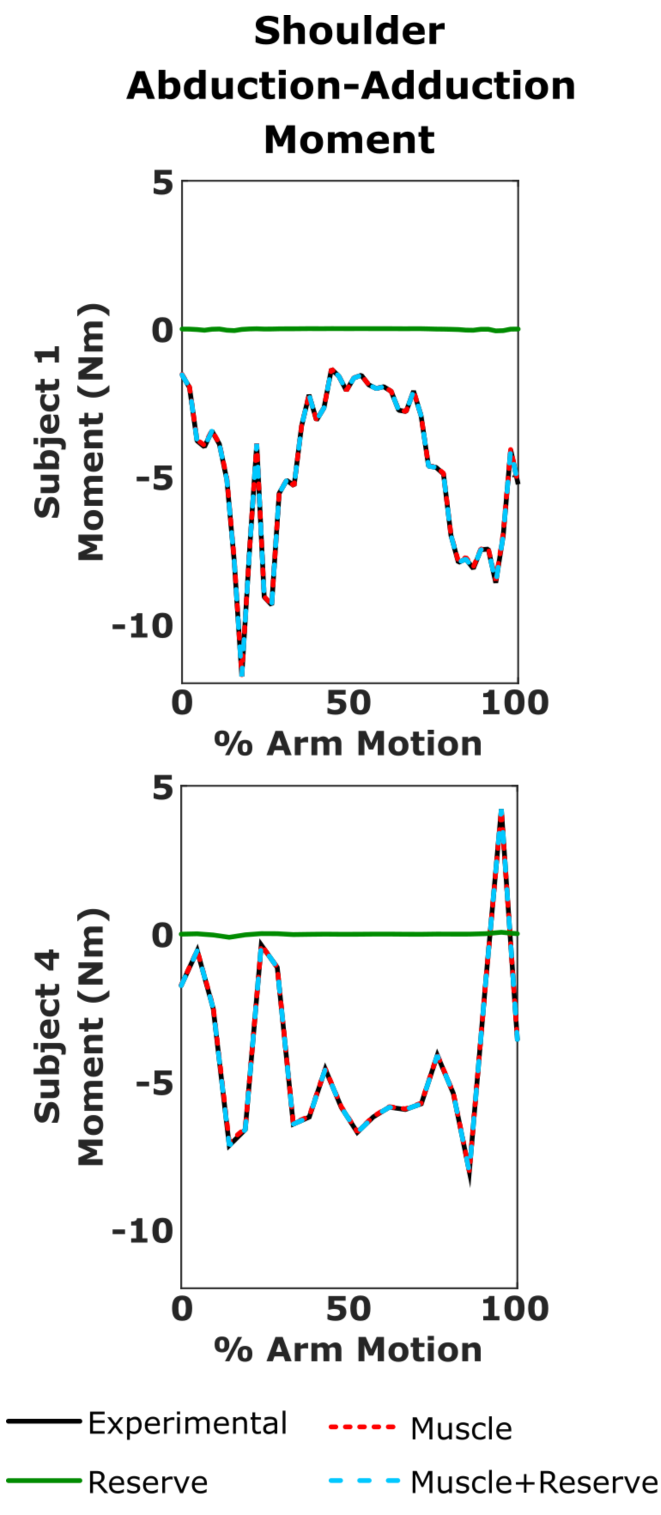

Figure 4. Abduction-adduction joint moments for Subject 1 (top) and Subject 4 (bottom) after the dynamic simulation only matching the abduction-adduction moment. The reserve contribution (green) was close to zero and the muscle contribution (red) matched with the experimental joint moment (black). The x-axis represents normalized arm motion for each subject's range of arm abduction.

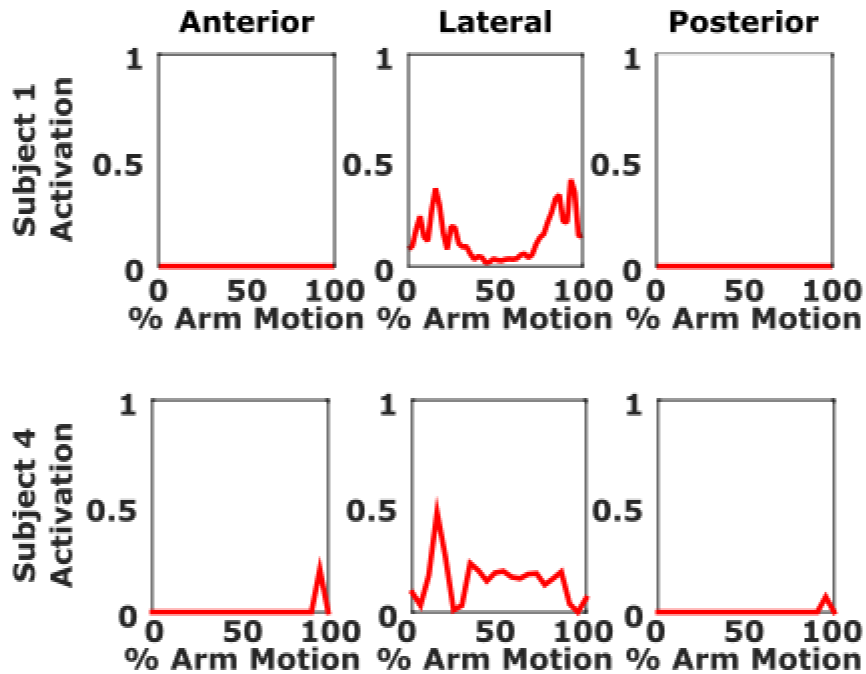

Figure 5. Deltoid muscle activation for Subject 1 (top) and Subject 4 (bottom) from the dynamic simulations when matching the abduction-adduction moment only. Activation is a unitless quantity in which 0 represents 0% activation when the muscle is not contracting and 1 represents 100% activation when the muscle is fully active and contracting. During abduction, it makes sense that the lateral deltoid would be the most active for these subjects, but the anterior and posterior deltoids should also be contributing to the motion to be consistent with previous results [16]. The x-axis represents normalized arm motion for each subject's range of arm abduction.

Agreement between the predicted muscle contributions and the experimental joint moments was not as strong when matching both the flexion-extension and abduction-adduction moments as indicated by the higher RMSE values for this condition (Table 7). However, subjects had varying levels of agreement. Subject 4 had very close agreement between the simulation predictions and the experimental data while the reserve actuators were opposing the muscle contribution for the flexion-extension joint moment for the three other subjects, including Subject 1 (Figure 6). The further illustrates the need for patient-specific modelling, optimization framework, and muscle parameters. The lack of agreement observed is most likely due to the simplicity of the model. The subregions of the deltoid are not the primary shoulder flexors/extensors and the results of the dynamic simulations showed that the deltoid muscles alone were not sufficient to match the predicted and experimental flexion-extension joint moment. Despite the lack of agreement in the joint moment data, the muscle activations predicted in this condition were more realistic with the anterior and posterior deltoids activating for longer periods throughout the dynamic arm abduction simulation (Figure 7).

It is likely that the ideal solution lies somewhere in between the results from matching only the abduction-adduction joint moment and the results from matching both the flexion-extension and abduction-adduction joint moments. The subregions of the deltoid contribute to both flexion-extension and abduction-adduction and therefore it would be beneficial to generate a model that is robust enough to match both moments. The results of this study indicate that additional shoulder muscles may be necessary to add to the model and optimization to more accurately represent muscle function in RTSA subjects.

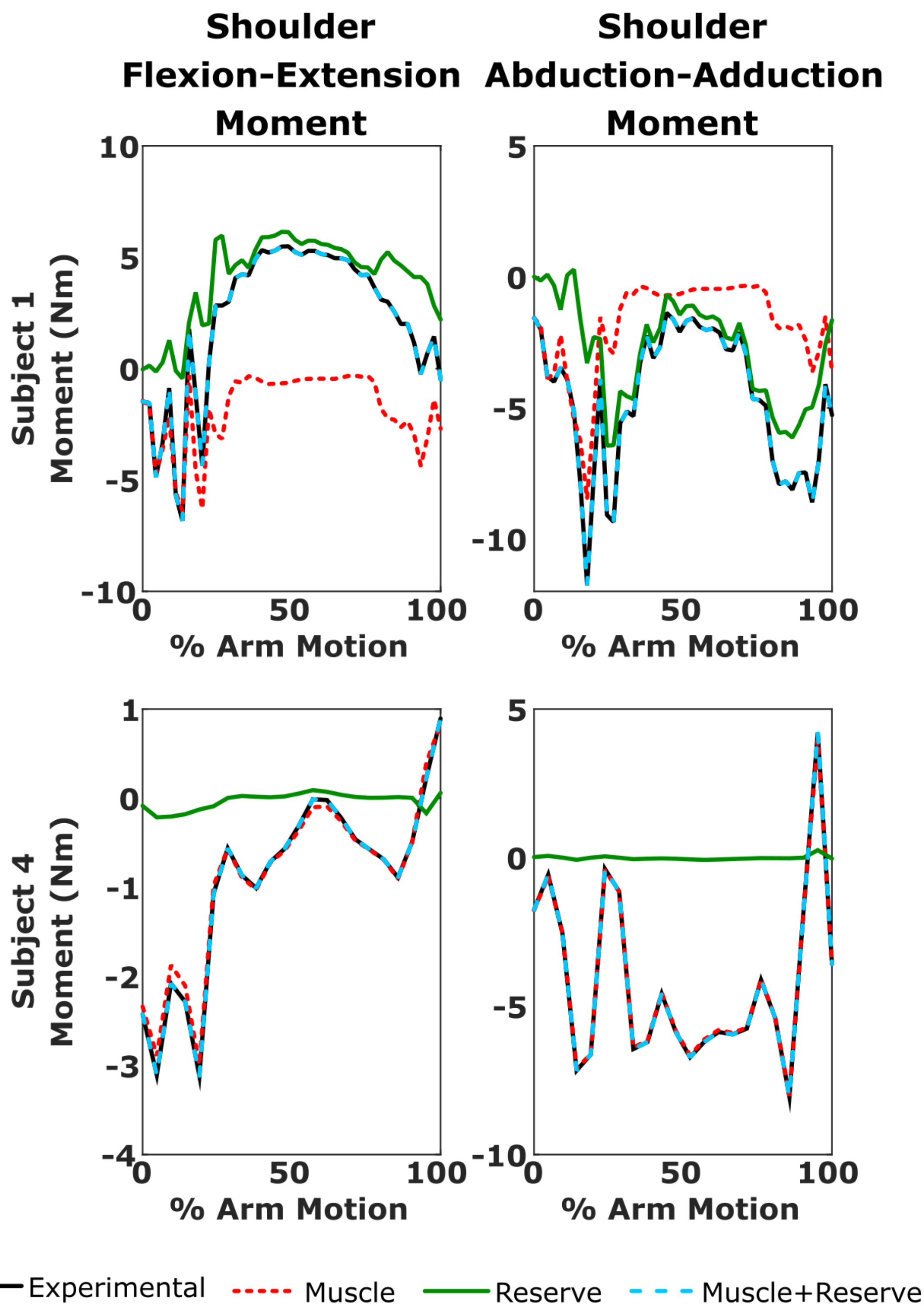

Figure 6. Flexion-extension and abduction-adduction joint moments for Subject 1 (top) and Subject 4 (bottom) after the dynamic simulation when matching the flexion-extension and abduction-adduction joint moments. The reserve contributions were not minimized as successfully as they were when only the abduction-adduction moment was matched. The x-axis represents normalized arm motion for each subject's range of arm abduction.

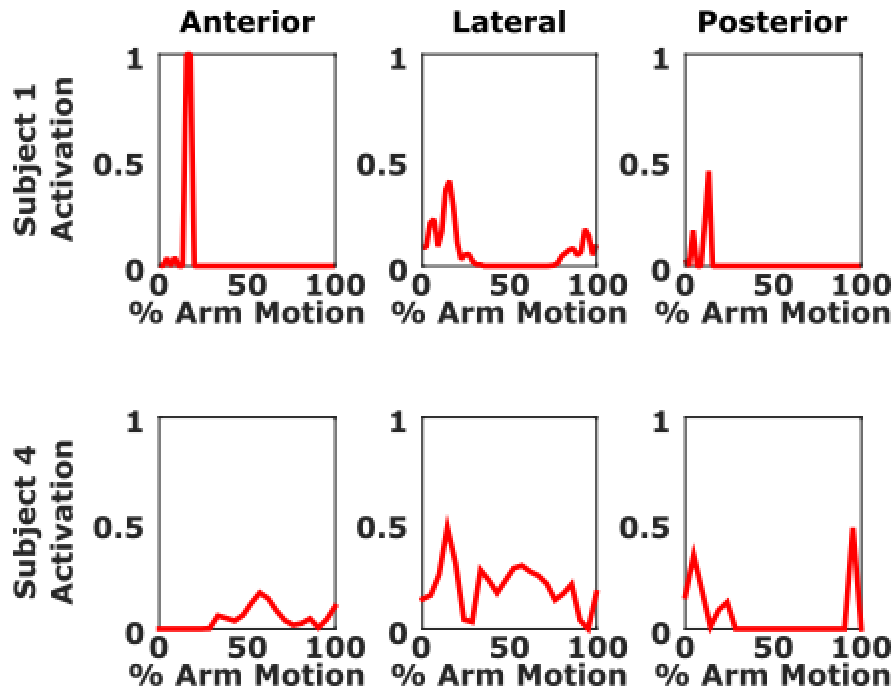

Figure 7. Deltoid muscle activation for Subject 1 (top) and Subject 4 (bottom) from the dynamic arm abduction simulations when matching both the flexion-extension and abduction-adduction joint moments. The x-axis represents normalized arm motion for each subject's range of arm abduction.

5. Conclusion

RTSA patients have varying height, weight, and muscle force capacity. Therefore, it is important that RTSA implants be placed in patient-specific locations to have optimal deltoid muscle function after surgery. To determine patient-specific implant placement, patient-specific models are needed. The purpose of this study was to develop a model and optimization framework to calibrate patient specific muscle model parameters for reverse total shoulder patients. The ultimate goal is to use the models and simulations to determine trends that may aid surgeons in identifying the exact location to place an RTSA implant to optimize the deltoid muscle forces thereby improving shoulder function.

Due to the fast workplace environment, surgeons do not have a great deal of time for optimizations to converge. Reduction of passive force appears to be a feasible process to adjust muscle model parameter values and improve patient-specific calibration of models. However, a more robust model including more shoulder muscles beyond the three deltoid components must be created to match both the flexion-extension and abduction-adduction moments. Future research should focus on collecting data of shoulder flexors and extensors and incorporating those muscles into the model and simulation to match the flexion-extension joint moment.

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