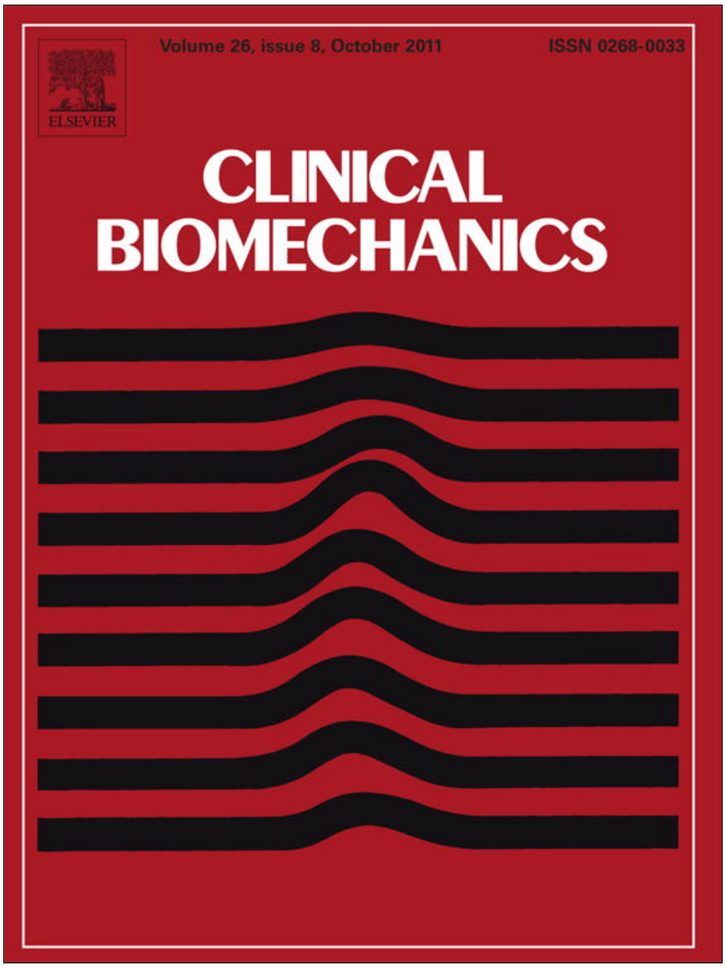
Volume 26, issue 8, October 2011
ISSN 0268-0033
ELSEVIER
CLINICAL
BIOMECHANICS

Clinical Biomechanics 26 (2011) 804–810

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Postural dependence of passive tension in the supraspinatus following rotator cuff repair: A simulation analysis ☆

Katherine R. Saul [a,b,*], Solomon Hayon [c], Thomas L. Smith [d], Christopher J. Tuohy [d], Sandeep Mannava [d]

[a] Department of Biomedical Engineering, Wake Forest School of Medicine, Winston-Salem, NC 27157, USA
[b] Virginia Tech – Wake Forest University School of Biomedical Engineering and Sciences, Winston-Salem, NC 27157, USA
[c] Wake Forest University, Winston-Salem, NC 27157, USA
[d] Department of Orthopaedic Surgery, Wake Forest School of Medicine, Winston-Salem, NC 27157, USA

ARTICLE INFO

Article history:
Received 19 January 2011
Accepted 12 April 2011

Keywords:
Computation
Musculoskeletal
Strength
Shoulder
Rotator cuff

ABSTRACT

Background: Despite surgical advances, repair of rotator cuff tears is associated with 20–70% incidence of recurrent tearing. The tension required to repair the torn tendon influences surgical outcomes and may be dependent on the gap length from torn tendon that must be spanned by the repair. Detailed understanding of forces throughout the range of motion (ROM) may allow surgeons to make evidence-based recommendations for post-operative care.
Methods: We used a computational shoulder model to assess passive tension and total moment-generating capacity in supraspinatus for repairs of gaps up to 3 cm throughout the shoulder (ROM).
Findings: In 60° abduction, increased gap length from 0.5 cm to 3 cm caused increases in passive force from 3 N to 58 N, consistent with those seen during clinical repair. For reduced abduction, passive forces increased substantially. For a 0.5 cm gap, tension throughout the ROM (elevation, plane of elevation, and rotation) is within reasonable limits, but larger gaps are associated with tensions that markedly exceed reported pull-out strength of sutures and anchors. Peak moment for a large 3 cm gap length was 5.09 Nm, a 53% reduction in moment-generating capacity compared to uninjured supraspinatus.
Interpretation: We conclude that shoulder posture is an important determinant of passive forces during rotator cuff repair surgery. Choosing postures that reduce forces intraoperatively to permit repair of larger gaps may lead to failure postoperatively when the shoulder is mobilized. For larger defects, loss of strength in supraspinatus may be substantial following repair even if retear is prevented.

1. Introduction

Rotator cuff tears pose a significant clinical problem. It is estimated that 20% of adults may suffer from a rotator cuff tear, with prevalence increasing to near 50% in adults over age 70 (Yamamoto et al., 2010). Approximately 300,000 rotator cuff surgeries are performed annually in the United States with an estimated economic burden of $3 billion (Aurora et al., 2007). The injury can lead to decreased range of shoulder motion, joint instability, chondral injury, and decreased function (Aurora et al., 2007; Neri et al., 2009), all of which may ultimately contribute to a loss of upper limb function and independence in the affected population (Centers for Disease Control and Prevention, 2003; Rejeski et al., 2008).

Once torn, rotator cuff tendons do not heal spontaneously in humans (Yadav et al., 2009). Since rotator cuff tendons have poor primary healing potential, surgical repair is often undertaken (Hirose et al., 2004; Neri et al., 2009). Despite considerable surgical advances, these surgeries are still associated with a high rate (20–70%) of recurrent tearing (Aurora et al., 2007; Galatz et al., 2001; Goldberg et al., 2001). Retear outcomes are often associated with several risk factors such as large tear size (Galatz et al., 2001, 2004; Oh et al., 2009), high degree of muscle atrophy (Galatz et al., 2001; Oh et al., 2009), poor tendon quality (Davidson and Rivenburgh, 2000; Oh et al., 2009), and inappropriate postoperative rehabilitation protocols (Iannotti, 1994; Oh et al., 2009).

The tension required to repair the torn rotator cuff tendon to the humeral head may influence surgical outcomes, and may be influenced by tear gap length. This tear gap is caused by a loss of tendon tissue due to the tear itself or damaged tissue removed during surgery. It can take an increased amount of tension to repair the tendon back to the humeral head secondary to muscle atrophy and retraction (Galatz et al., 2004; Gimbel et al., 2004a; Oh et al., 2009), which cause larger tear gaps. The passive forces in the repaired tendon have been characterized during surgical repair for the supraspinatus, a frequently injured rotator cuff muscle (Davidson and Rivenburgh, 2000). High intraoperative tensions have been associated with

☆ Investigation performed in the Department of Biomedical Engineering, Wake Forest School of Medicine, Medical Center Boulevard, Winston-Salem, NC 27157, USA.

* Corresponding author at: Department of Biomedical Engineering, Wake Forest School of Medicine, Medical Center Boulevard, Winston-Salem, CA 27157, USA.
E-mail address: ksaul@wakehealth.edu (K.R. Saul).

0268-0033/$ – see front matter
doi:10.1016/j.clinbiomech.2011.04.005

decreased post-operative active force-generating capability (Hersche and Gerber, 1998), lower scores on clinical outcome measures of shoulder function (e.g. Constant scale) (Davidson and Rivenburgh, 2000), and decreased pain relief (Davidson and Rivenburgh, 2000). In addition, high forces lead to increased force on the sutures themselves. However, the forces measured intraoperatively represent tensions observed only in the repair posture. The force that a muscle generates is highly dependent on its length, and therefore on the posture of the joints that it crosses (Gordon et al., 1966; Zajac, 1989). Consequently, the forces in the repaired muscle during post-operative immobilization and rehabilitation may differ from those measured intraoperatively, and it is difficult to obtain these forces *in vivo*. Previous researchers have demonstrated that passive tension in the supraspinatus depends on abduction posture, and have emphasized the importance of postoperative positioning (Hersche and Gerber, 1998). A more detailed understanding of the forces produced at the shoulder throughout the full range of motion for a variety of gap lengths may allow surgeons to make evidence-based recommendations for post-operative immobilization and rehabilitation, while optimally protecting the repair site.

Computational musculoskeletal modeling is an approach that permits characterization of the function of muscles for a wide range of postures and movements, through the integration of anatomical data into mathematical representations of joint motions and muscle force-generating properties. Musculoskeletal simulation has been previously used to evaluate surgical repair strategies in the upper (Murray et al., 2002; Saul et al., 2003) and lower (Fox et al., 2009) limbs. The goal of this study was to characterize the dependence of passive force and active moment-generating capacity on shoulder posture and tear size in supraspinatus following surgical repair of gap lengths up to 3 cm using a computational model.

2. Methods

We used a previously developed 15 degree of freedom kinematic model of the upper limb (Holzbaur et al., 2005) to perform simulations of surgical rotator cuff repair of the supraspinatus muscle–tendon unit. This model represents the anthropometry (Gordon et al., 1989; Winter, 1990) and muscle force-generating capacity of a 50th percentile young adult male. For this study, we used the shoulder portion of the model, which includes representations of the three degrees of freedom of the glenohumeral joint and constrained movement of the clavicle and scapula. The kinematic convention recommended by the International Society of Biomechanics (Wu et al., 2005) is used to describe the three degrees of freedom of the shoulder, which include axial rotation around the long axis of the humerus, thoracohumeral elevation (angle between the torso and humerus), and elevation plane (0° is abduction; 90° is forward flexion) (Fig. 1). The motion of the clavicle and the scapula is prescribed to vary with elevation of the humerus to produce a normal movement of the shoulder girdle (de Groot and Brand, 2001).

The muscles crossing the glenohumeral joint, including the muscles of the rotator cuff, are represented with 18 Hill-type muscle–tendon actuators. These actuators incorporate anatomically-based parameters (Langenderfer et al., 2004; Murray et al., 2000) derived from muscle cross-sectional area, muscle length, tendon length, and pennation angle that define the force-generating capacity of each muscle (Zajac, 1989). This type of muscle model allows us to calculate both the force that a muscle–tendon unit generates when the muscle is fully excited and the passive force generated from stretch in the actuator at a given length. In addition, the forces and moments produced by any combination of muscles and activations can also be calculated, though in this study we will focus on the behavior of the supraspinatus in isolation. The paths of the actuators are defined to match anatomical description using points attached to the underlying bones and wrapping surfaces meant to represent underlying bone and muscle surfaces. Muscles that have wide attachments (e.g. deltoid) or multiple tendons (e.g. biceps brachii) are represented by multiple paths. These paths have been validated using experimentally measured moment arms that mathematically define the path of a muscle (Hughes et al., 1998; Liu et al., 1997; Otis et al., 1994). We can also use the model to calculate the moment-generating capacity of a muscle at a given joint; the moment that a muscle produces is a common measure of strength at a joint (Engin and Kaleps, 1980; Garner and Pandy, 2001; Otis et al., 1990; Winters and Kleweno, 1993). Moment is defined as the product of muscle force and the moment arm of a muscle. We can calculate the moment produced at the shoulder throughout the range of motion.

We simulated the repair of the supraspinatus for tendon defects up to 3.0 cm in length. To do this, we shortened the parameter representing tendon length in the model — tendon slack length — by the length of the defect. The path of the supraspinatus was unchanged. This is consistent with clinical goals to restore the intact path of the muscle by pulling the muscle–tendon unit back to the insertion footprint on the humeral head. The reduction in tendon slack length represents the loss of tendon tissue from the tear or removal intraoperatively when the defect is trimmed before repair. We performed simulations for the uninjured condition (tendon length = 3.95 cm), and for gap lengths from 0.5 cm to 3.0 cm in 0.5 cm increments. We calculated the passive muscle–tendon force and total (active plus passive) moment generated by the supraspinatus, and evaluated the change in force and moment with posture and gap length. Total moment calculations assumed full activation of the muscle. We compared passive forces to experimentally measured pull-out strength of suture and anchor techniques typically used in these repairs (191–287 N) (Demirhan et al., 2003; Ma et al., 2006) and

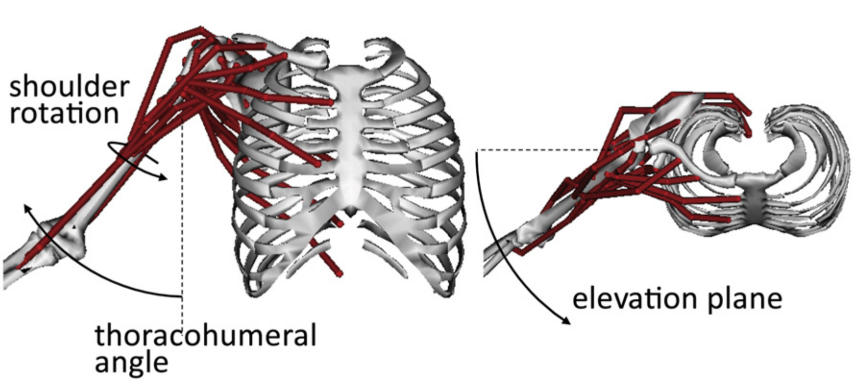

Fig. 1. Degrees of freedom defined in the musculoskeletal model. Axial shoulder rotation, thoracohumeral angle, and the plane of elevation are implemented using the ISB recommendations (Wu et al., 2005). Neutral postures are indicated with the dashed lines. For elevation plane, 0° is the abduction plane, and 90° is forward flexion.

identified postures for which passive forces remained under this threshold. We calculated the moment-generating capacity of the supraspinatus to elevate the humerus in the abduction plane with the shoulder in neutral rotation (an estimate of abduction strength, one of the primary functions of the supraspinatus).

3. Results

Passive force in the supraspinatus increases with gap length when the shoulder is elevated in the abduction plane (Fig. 2). For a 0.5 cm gap, tension throughout the range of motion is within reasonable limits, with all postures resulting in force under 28.5 N, well under reported pull-out force (Demirhan et al., 2003; Ma et al., 2006). For reference, a pull out force of 215 N, as measured by Demirhan et al. (2003) for a repair using 2 corkscrews combined with a single Mason–Allen transosseous suture, is noted in the figures. The highest passive forces are seen when the shoulder is in a neutral elevation posture. In a typical shoulder positioning for rotator cuff repair (60° abduction), an increase in gap length from 0.5 cm to 3 cm caused increases in supraspinatus passive force from 3.4 N to 61.5 N. However, when abduction is reduced to 0°, passive forces increased from 28.5 N to 518.1 N. For gap lengths of 2.5 and 3.0 cm, there were postures (less than 5° and 22° elevation, respectively) for which passive force exceeded the representative pull-out strength value of 215 N (Demirhan et al., 2003), and a larger range of postures exceeding the lower reported strengths, as low as 191 N (Ma et al., 2006).

Changing the elevation plane and the axial rotation of the arm also altered the passive force produced by the supraspinatus (Figs. 3 and 4). For a 3.0 cm gap length, the range of elevation planes that do not exceed the pull-out force increased with increasing thoracohumeral angle. Postures closer to the abduction plane rather than forward flexion plane were associated with lower passive forces, with a minimum force in the 20° elevation plane. Similarly, the range of axial rotation postures that do not exceed the pull-out force increase with increasing thoracohumeral angle, with lower forces associated with external rotation.

We observed decreases in the peak abduction moment-generating capacity of the supraspinatus with increasing gap length (Fig. 5). Peak moment for the largest gap length of 3 cm was 5.09 Nm, a 53% reduction in moment-generating capacity compared to an uninjured muscle–tendon unit. Differences among gap length conditions were most pronounced when the arm was more adducted.

4. Discussion

Our data indicate that larger tear size is associated with increased passive forces through the range of motion and decreased active moment-generating capacity of the supraspinatus. In addition, we conclude that shoulder posture is an important determinant of passive forces during and following rotator cuff repair surgery. In a typical shoulder positioning for rotator cuff repair (60° abduction), we calculated that an increase in gap length from 0.5 cm to 3 cm caused increases in supraspinatus passive force from 3 N to 58 N. These passive forces are in line with those seen intraoperatively during clinical repair of the rotator cuff (0–45 N, mean 11 N) (Davidson and Rivenburgh, 2000), providing validation for our simulations. However, smaller thoracohumeral angles were associated with large increases in passive force, and a range of postures was identified for which direct failure of the suture and anchoring techniques was predicted.

Several other researchers have also noted the importance of considering the posture of the shoulder and its effect on tension in the rotator cuff. Certainly, it is well known that the force-generating capacity of muscles in the rotator cuff is dependent on the posture of the arm, as summarized by Cole et al. (2007). Others have explored this issue with regard to passive tension in surgical repair. Andres et al. (2010) demonstrated in an ovine model that increased abduction is associated with reduced tension and glenoid contact pressures. Reilly et al. (2004) also identified a dependence of passive tension in the supraspinatus on abduction angle, using a combination of intraoperative measurements and cadaveric measurements. In that study, an increase of 30° in abduction posture from neutral led to an average decrease in passive tension of 34 N, and an imposition of this tension in a cadaveric repair led to gap development. This may indicate progression toward failure of the repair site. In our study, we predict a reduction in passive tension for the same postures for a gap length of 1 cm to be 37.1 N. Hersche and Gerber (1998) also reported increases in tension of 45 N with reduced abduction of 60°. Our simulations predict a reduction of 44.7 N for 60° for a 1 cm gap length. These studies provide additional support for our findings and simulations, which agree well with these previous studies, while we

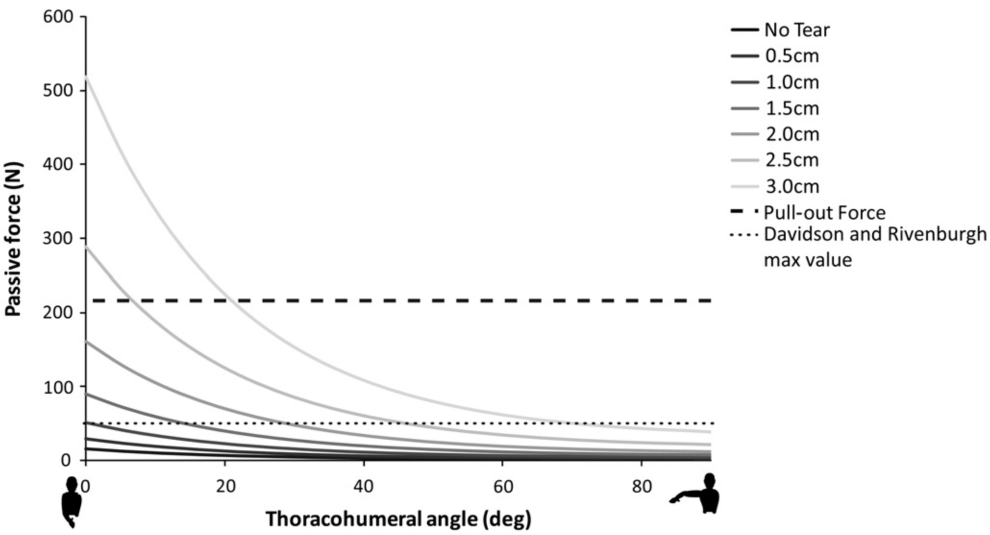

Fig. 2. Supraspinatus passive force versus abduction. Gap lengths from 0 cm (no tear) to 3.0 cm are indicated with the solid curves. A representative pull-out strength for sutures and anchoring of 215 N (Demirhan et al., 2003) is indicated with the dashed black line. Larger gap lengths and smaller thoracohumeral angles are associated with increased passive forces. Passive forces at 60° abduction for all gap lengths are consistent with observed intraoperative passive forces (0–45 N) (Davidson and Rivenburgh, 2000); the maximum reported value is indicated (dotted black line).

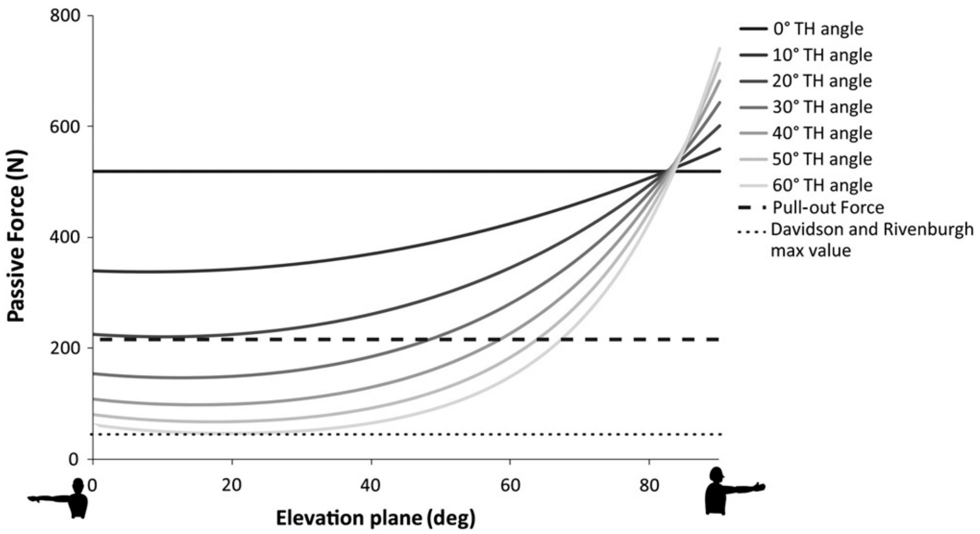

Fig. 3. Supraspinatus passive force versus elevation plane for a 3 cm gap length. Lower thoracohumeral angles (darker solid curves) and more forward flexed postures are associated with higher passive forces that can exceed a representative pull-out strength for sutures and anchoring of 215 N (Demirhan et al., 2003) (dashed black line). Passive force at 60° abduction (0° elevation plane) for this gap length is consistent with the maximum observed intraoperative passive forces (45 N) (Davidson and Rivenburgh, 2000) (dotted black line).

expanded the analysis to consider the effects of posture on tension beyond abduction alone.

One important clinical consideration is the appropriate range of postures that should be permitted immediately following surgical repair of a ruptured tendon, either during the use of a sling for post-operative immobilization or during rehabilitation. For 2.5 and 3 cm gap lengths, tension markedly exceeded pull-out strength of suture and anchor techniques typically used in these repairs in some postures. While other gap length conditions did not experience forces larger than the pull-out strength, there was still a marked increase in passive forces at lower thoracohumeral angles. High forces that do not exceed the pull-out strength threshold may still interfere with healing. Mechanotransduction is essential for tendon-to-bone healing. Tensions that are too high or too low have been shown in the pre-clinical setting to affect the ultimate biomechanical properties of the tendon-to-bone interface (Galatz et al., 2004; Gimbel et al., 2004a, 2004b). Post-operative rehabilitation provides the patient with early motion and possibly an earlier return to recreational or work related activities (Klintberg et al., 2009). The surgeon must balance early motion with the healing and repair site biomechanical integrity in order to attain the best surgical and clinical outcome (Koo and Burkhart, 2010). Our data address the biomechanical implications of common movements and postures, indicating that external rotation, abducted elevation planes, and larger thoracohumeral angle postures were associated with decreased passive forces. In particular, thoracohumeral angle had the largest effect on the passive force. These postures would therefore be best suited for sling positioning and for early rehabilitation to reduce the likelihood of retear during healing.

The moment-generating capacity of the repaired supraspinatus is altered substantially for larger tear sizes. Even without representing changes in muscle force-generating capacity, or atrophy, with age, injury, or gender, we still see a reduction in the peak total moment generated by the muscle due solely to the stretch in muscle from a shortened tendon. A combination of lengthening of the muscle

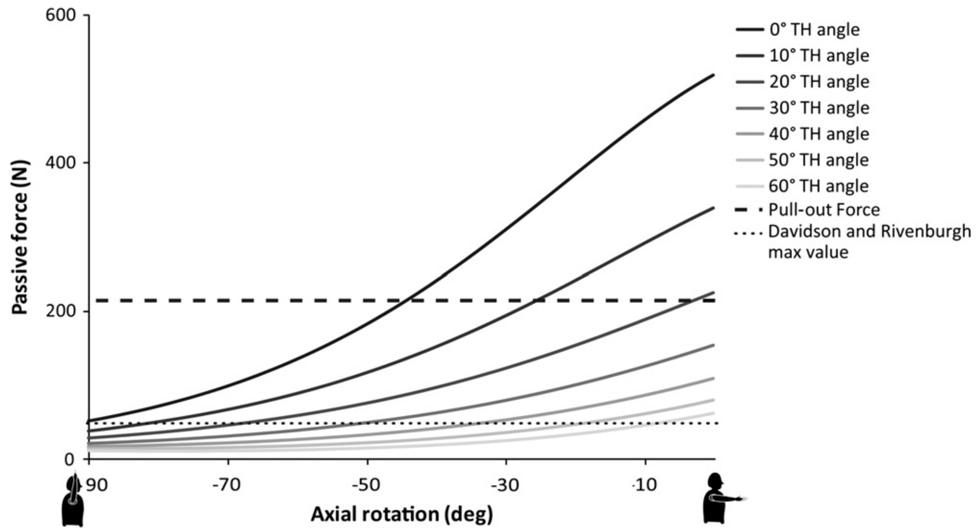

Fig. 4. Supraspinatus passive force versus axial rotation for a 3 cm gap length. Lower thoracohumeral angles (darker solid curves) and more internally rotated postures are associated with higher passive forces that can exceed a representative pull-out strength for sutures and anchoring of 215 N (Demirhan et al., 2003) (dashed black line). Passive force at 60° abduction and 0° rotation for this gap length is consistent with the maximum observed intraoperative passive force (45 N) (Davidson and Rivenburgh, 2000) (dotted black line).

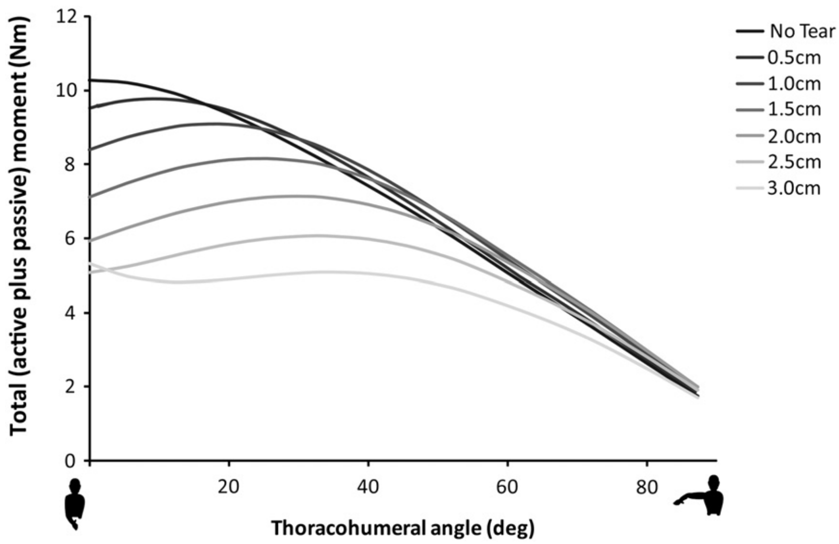

Fig. 5. Total (active plus passive) isometric moment-generating capacity for supraspinatus versus abduction. Gap lengths from 0 cm (no tear) to 3.0 cm are indicated with the solid curves. Larger gap lengths are associated with reduced peak isometric moment-generating capacity, with reductions up to 53% for a 3 cm gap. A combination of lengthening of the muscle causing a reduction in muscle active force and an alteration of the interaction between the force and moment arm profiles of the supraspinatus results in an overall reduction in total moment.

causing a reduction in muscle active force and an alteration of the interaction between the force and moment arm profiles of the supraspinatus results in an overall reduction in total moment. While we can identify postures which will protect the repair immediately following surgery from high passive forces, function is altered relative to the uninjured muscle even if retear is avoided. This suggests that with current repair techniques and single stage surgery, repair of large tear sizes significantly reduces the supraspinatus' function. This is consistent with other reports in the literature. For example, Davidson and Rivenburgh (2000) recommended against repairs requiring more than 36 N of tension due to reduced isokinetic strength and increased pain observed following surgical repair with this high tension. Unfortunately, direct testing of the muscle–tendon unit is difficult in the intraoperative surgical setting; thus, computational modeling is advantageous in that it allows for the assessment of force-generating capacity.

Previous authors have reported a range of pull-out strengths for common arthroscopic suturing and anchoring techniques, from 191 to 287 N (Demirhan et al., 2003; Ma et al., 2006) for a variety of common techniques including two-simple, massive cuff, Mason–Allen, double-row fixation, and corkscrew anchors (Arthrex, Munich, Germany). Ma et al. (2004) also reported 233 N and 246 N pull-out strengths for massive cuff and modified Mason–Allen approaches, respectively. However, this study also found substantially lower strengths of 72 N and 77 N for simple and horizontal suturing techniques, respectively. While we use the higher range of values, and in particular 215 N as reported by Demirhan et al. (2003) as a representative value for our analyses, it is important to note that lower forces can lead to failure for some types of suturing approaches. Our results suggest that for more substantial tears with larger gap lengths, suturing and anchoring techniques with higher pull-out strengths are indicated.

Our simulations do not incorporate representations of changes or adaptations in the muscle that may occur with injury or following surgery. Rotator cuff tears are known to be associated with atrophy of the injured muscle (Galatz et al., 2001; Oh et al., 2009), fatty infiltration, muscle retraction, and loss of tendon mechanical quality (Davidson and Rivenburgh, 2000; Oh et al., 2009). Changes in these properties affect the generation of force and moment by the affected muscle. In particular, atrophy and fatty infiltration would reduce the active force and moment-generating capacity of a muscle by reducing the volume of muscle tissue that can generate force. Muscle retraction due to an overall shortening of the muscle fibers and loss of sarcomeres in series would be likely to increase the passive force generated by the muscle throughout the range of motion when the tear is repaired, due to the force-length property of muscle (Gordon et al., 1966; Zajac, 1989). Specifically, since the retracted fibers would need to stretch to longer lengths to span the same distance, passive force would increase even more than simulated here. A loss of tendon mechanical integrity would likely reduce the suture pull out strength. Therefore, it is likely that the clinical implications predicted by the simulations described here are a "best-case" scenario when considered with respect to muscle behavior after injury-induced muscle changes. Additional muscle changes as described here would contribute to even more overall weakening of the supraspinatus accompanied by higher passive forces and lower pull-out strengths. Therefore, gap size and post-operative shoulder posture may be considered clinically to limit negative post-surgical consequences.

The model used in these simulations represented average muscle properties for a 50th percentile young adult male. Rotator cuff impairment is frequently associated with an older patient population (Yamamoto et al., 2010), and it is known that muscle atrophy and loss of strength are frequently associated with aging (Akagi et al., 2009; Klein et al., 2001). Models that are made to better represent specific individuals by representing the relative muscle size and particular muscle injury for individual patients may be valuable for individualized complex analyses of movement or surgical outcomes for which relative strength of multiple muscles may have a larger influence. However, these types of models are difficult and time-consuming to develop, requiring detailed imaging of the upper limb muscles and computationally expensive analysis of these images and would be unlikely to provide an improved understanding of the postural dependence of the biomechanics of a single muscle in isolation, as in this study. The model used here with average muscle properties is well-suited for answering the questions posed in this study, because the postural dependences of passive muscle–tendon behavior is determined primarily by the geometric configuration of the muscle and the relative length of the tendon and muscle in the supraspinatus specifically. These properties are unlikely to change substantially for the population of interest, except with regard to changes in tendon length due to surgery. While variations in muscle volume or cross-sectional area are likely to be seen among individuals, and do significantly influence muscle force and moment (Holzbaur et al., 2007), this would only influence the magnitudes of the results seen, and not the effects of gap length or posture. Therefore, the insights gained from this study with regard to the postures that reduce passive forces in the repaired tendon, and the large passive forces associated with large gap lengths can be used clinically to guide decisions made when planning a treatment strategy. For individual patients, their particular clinical condition should always be considered first before applying biomechanical principles to practice, as there are always other clinical considerations, such as overall patient health or concomitant neuromuscular injuries or disorders, that may influence surgical decision-making. Models that better represent an aging population may provide additional insight into the possible clinical outcomes, including potential compensatory strategies used when repair of the rotator cuff is not possible. The model could be altered using population averages for muscle volumes in older men and women, which may provide a basis for incorporating salient characteristics of an older adult population without the high cost of developing individual patient models.

Another limitation of the model is that individual muscles act in isolation from one another; there are no connections among adjacent muscles. It is known that forces are transmitted to adjacent muscles (as summarized by Huijing, 2009), and that the muscles of the rotator cuff may be particularly connected in both the muscle and tendon regions. These interactions are important to consider, and may be important for questions such as predicting locations for subsequent tearing following a primary tear or whether there are alterations in the paths of adjacent muscles following an isolated tear due to retraction of a muscle. Finite element modeling of the rotator cuff muscles is one approach that may ultimately address these issues; however, to date finite element modeling of muscle is computationally expensive (Blemker and Delp, 2005) and has not yet incorporated interconnections among adjacent muscles. To the authors' knowledge, existing finite element models of the shoulder (e.g. Adams et al., 2007; Suarez et al., 2009) do not incorporate three-dimensional representations of active or passive muscle structures. We assume that the normal path of the supraspinatus is restored by the surgery; by considering only the forces in the muscle postoperatively, we attempt to limit the effect of neglecting the intramuscular interactions. Further, because the passive and active moments produced at the shoulder compare well with experimentally measured values for the uninjured shoulder (Engin and Kaleps, 1980; Holzbaur et al., 2005; Winters and Kleweno, 1993), and because the passive forces we predict match well with intraoperative forces for supraspinatus (Davidson and Rivenburgh, 2000), we feel confident in the capability of this biomechanical model to answer the questions posed in this study.

We conclude that more abducted postures, while reducing forces intraoperatively to permit repair of larger gaps, may ultimately lead to increased failure postoperatively when the shoulder is adducted. Furthermore, for larger tendon defects, the loss of strength in the supraspinatus may be substantial following repair even if retear is prevented. This study elucidates the biomechanical consequences of large tendon gap length repairs in the supraspinatus, and provides insights as to why this is associated with poorer outcomes clinically. Methods for reducing the gap lengths that must be bridged, such as tendon grafts or graft substitutes, may be a promising route for limiting the negative biomechanical consequences of necessary surgical repair.

Acknowledgments

Funding support for this work was provided by the WFU Graduate School of Arts and Sciences Summer Research Opportunities Program. We also thank the OpenSim development team at Stanford University for helpful conversations regarding model implementation. The study sponsors had no involvement in the study design; in the collection, analysis and interpretation of data; in the writing of the manuscript; nor in the decision to submit the manuscript for publication.

References

Adams, C.R., Baldwin, M.A., Laz, P.J., Rullkoetter, P.J., Langenderfer, J.E., 2007. Effects of rotator cuff tears on muscle moment arms: a computational study. J. Biomech. 40, 3373–3380.
Akagi, R., Takai, Y., Ohta, M., Kanehisa, H., Kawakami, Y., Fukunaga, T., 2009. Muscle volume compared to cross-sectional area is more appropriate for evaluating muscle strength in young and elderly individuals. Age Ageing 38, 564–569.
Andres, B.M., Lam, P.H., Murrell, G.A., 2010. Tension, abduction, and surgical technique affect footprint compression after rotator cuff repair in an ovine model. J. Shoulder Elbow Surg. 19, 1018–1027.
Aurora, A., Mccarron, J., Iannotti, J.P., Derwin, K., 2007. Commercially available extracellular matrix materials for rotator cuff repairs: state of the art and future trends. J. Shoulder Elbow Surg. 16, S171–S178.
Blemker, S.S., Delp, S.L., 2005. Three-dimensional representation of complex muscle architectures and geometries. Ann. Biomed. Eng. 33, 661–673.
Centers for Disease Control and Prevention, 2003. Trends in Aging — United States and Worldwide.
Cole, B.J., Elattrache, N.S., Anbari, A., 2007. Arthroscopic rotator cuff repairs: an anatomic and biomechanical rationale for different suture-anchor repair configurations. Arthroscopy 23, 662–669.
Davidson, P.A., Rivenburgh, D.W., 2000. Rotator cuff repair tension as a determinant of functional outcome. J. Shoulder Elbow Surg. 9, 502–506.
De Groot, J.H., Brand, R., 2001. A three-dimensional regression model of the shoulder rhythm. Clin. Biomech. 16, 735–743.
Demirhan, M., Atalar, A.C., Kilicoglu, O., 2003. Primary fixation strength of rotator cuff repair techniques: a comparative study. Arthroscopy 19, 572–576.
Engin, A.E., Kaleps, I., 1980. Active muscle torques about long-bone axes of major human joints. Aviat. Space Environ. Med. 51, 551–555.
Fox, M.D., Reinbolt, J.A., Ounpuu, S., Delp, S.L., 2009. Mechanisms of improved knee flexion after rectus femoris transfer surgery. J. Biomech. 42, 614–619.
Galatz, L.M., Griggs, S., Cameron, B.D., Iannotti, J.P., 2001. Prospective longitudinal analysis of postoperative shoulder function: a ten-year follow-up study of full-thickness rotator cuff tears. J. Bone Joint Surg. Am. 83-A, 1052–1056.
Galatz, L.M., Ball, C.M., Teefey, S.A., Middleton, W.D., Yamaguchi, K., 2004. The outcome and repair integrity of completely arthroscopically repaired large and massive rotator cuff tears. J. Bone Joint Surg. Am. 86-A, 219–224.
Garner, B.A., Pandy, M.G., 2001. Musculoskeletal model of the upper limb based on the visible human male dataset. Comput. Methods Biomech. Biomed. Engin. 4, 93–126.
Gimbel, J.A., Mehta, S., Van Kleunen, J.P., Williams, G.R., Soslowsky, L.J., 2004a. The tension required at repair to reappose the supraspinatus tendon to bone rapidly increases after injury. Clin. Orthop. Relat. Res. 258–265.
Gimbel, J.A., Van Kleunen, J.P., Mehta, S., Perry, S.M., Williams, G.R., Soslowsky, L.J., 2004b. Supraspinatus tendon organizational and mechanical properties in a chronic rotator cuff tear animal model. J. Biomech. 37, 739–749.
Goldberg, B.A., Nowinski, R.J., Matsen III, F.A., 2001. Outcome of nonoperative management of full-thickness rotator cuff tears. Clin. Orthop. Relat. Res. 99–107.
Gordon, A.M., Huxley, A.F., Julian, F.J., 1966. The variation in isometric tension with sarcomere length in vertebrate muscle fibres. J. Physiol. 184, 170–192.
Gordon, C.C., Churchill, T., Clauser, C.E., Bradtmiller, B., Mcconville, J.T., Tebbets, I., et al., 1989. 1988 Anthropometric Survey of U.S. Army Personnel: Methods and Summary Statistics. United States Army Natick Research, Development and Engineering Center, Natick, MA.
Hersche, O., Gerber, C., 1998. Passive tension in the supraspinatus musculotendinous unit after long-standing rupture of its tendon: a preliminary report. J. Shoulder Elbow Surg. 7, 393–396.
Hirose, K., Kondo, S., Choi, H.R., Mishima, S., Iwata, H., Ishiguro, N., 2004. Spontaneous healing process of a supraspinatus tendon tear in rabbits. Arch. Orthop. Trauma Surg. 124, 374–377.
Holzbaur, K.R., Murray, W.M., Delp, S.L., 2005. A model of the upper extremity for simulating musculoskeletal surgery and analyzing neuromuscular control. Ann. Biomed. Eng. 33, 829–840.
Holzbaur, K.R., Delp, S.L., Gold, G.E., Murray, W.M., 2007. Moment-generating capacity of upper limb muscles in healthy adults. J. Biomech. 40, 2442–2449.
Hughes, R.E., Niebur, G., Liu, J., An, K.N., 1998. Comparison of two methods for computing abduction moment arms of the rotator cuff. J. Biomech. 31, 157–160.
Huijing, P.A., 2009. Epimuscular myofascial force transmission: a historical review and implications for new research. International Society of Biomechanics Muybridge Award Lecture, Taipei, 2007. J. Biomech. 42, pp. 9–21.
Iannotti, J.P., 1994. Full-thickness rotator cuff tears: factors affecting surgical outcome. J. Am. Acad. Orthop. Surg. 2, 87–95.
Klein, C.S., Rice, C.L., Marsh, G.D., 2001. Normalized force, activation, and coactivation in the arm muscles of young and old men. J. Appl. Physiol. 91, 1341–1349.
Klintberg, I.H., Gunnarsson, A.C., Svantesson, U., Styf, J., Karlsson, J., 2009. Early loading in physiotherapy treatment after full-thickness rotator cuff repair: a prospective randomized pilot-study with a two-year follow-up. Clin. Rehabil. 23, 622–638.
Koo, S.S., Burkhart, S.S., 2010. Rehabilitation following arthroscopic rotator cuff repair. Clin. Sports Med. 29, 203–211, vii.

Langenderfer, J., Jerabek, S.A., Thangamani, V.B., Kuhn, J.E., Hughes, R.E., 2004. Musculoskeletal parameters of muscles crossing the shoulder and elbow and the effect of sarcomere length sample size on estimation of optimal muscle length. Clin. Biomech. 19, 664–670.

Liu, J., Hughes, R.E., Smutz, W.P., Niebur, G., Nan-an, K., 1997. Roles of deltoid and rotator cuff muscles in shoulder elevation. Clin. Biomech. 12, 32–38.

Ma, C.B., Macgillivray, J.D., Clabeaux, J., Lee, S., Otis, J.C., 2004. Biomechanical evaluation of arthroscopic rotator cuff stitches. J. Bone Joint Surg. Am. 86-A, 1211–1216.

Ma, C.B., Comerford, L., Wilson, J., Puttlitz, C.M., 2006. Biomechanical evaluation of arthroscopic rotator cuff repairs: double-row compared with single-row fixation. J. Bone Joint Surg. Am. 88, 403–410.

Murray, W.M., Buchanan, T.S., Delp, S.L., 2000. The isometric functional capacity of muscles that cross the elbow. J. Biomech. 33, 943–952.

Murray, W.M., Bryden, A.M., Kilgore, K.L., Keith, M.W., 2002. The influence of elbow position on the range of motion of the wrist following transfer of the brachioradialis to the extensor carpi radialis brevis tendon. J. Bone Joint Surg. Am. 84-A, 2203–2210.

Neri, B.R., Chan, K.W., Kwon, Y.W., 2009. Management of massive and irreparable rotator cuff tears. J. Shoulder Elbow Surg. 18, 808–818.

Oh, J.H., Kim, S.H., Ji, H.M., Jo, K.H., Bin, S.W., Gong, H.S., 2009. Prognostic factors affecting anatomic outcome of rotator cuff repair and correlation with functional outcome. Arthroscopy 25, 30–39.

Otis, J.C., Warren, R.F., Backus, S.I., Santner, T.J., Mabrey, J.D., 1990. Torque production in the shoulder of the normal young adult male. The interaction of function, dominance, joint angle, and angular velocity. Am. J. Sports Med. 18, 119–123.

Otis, J.C., Jiang, C.C., Wickiewicz, T.L., Peterson, M.G., Warren, R.F., Santner, T.J., 1994. Changes in the moment arms of the rotator cuff and deltoid muscles with abduction and rotation. J. Bone Joint Surg. Am. 76, 667–676.

Reilly, P., Bull, A.M., Amis, A.A., Wallace, A.L., Richards, A., Hill, A.M., et al., 2004. Passive tension and gap formation of rotator cuff repairs. J. Shoulder Elbow Surg. 13, 664–667.

Rejeski, W.J., Ip, E.H., Marsh, A.P., Miller, M.E., Farmer, D.F., 2008. Measuring disability in older adults: the International Classification System of Functioning, Disability and Health (ICF) framework. Geriatr. Gerontol. Int. 8, 48–54.

Saul, K.R., Murray, W.M., Hentz, V.R., Delp, S.L., 2003. Biomechanics of the Steindler flexorplasty surgery: a computer simulation study. J. Hand Surg. Am. 28, 979–986.

Suarez, D.R., Van Der Linden, J.C., Valstar, E.R., Broomans, P., Poort, G., Rozing, P.M., et al., 2009. Influence of the positioning of a cementless glenoid prosthesis on its interface micromotions. Proc. Inst. Mech. Eng. H 223, 795–804.

Winter, D., 1990. Biomechanics and Motor Control of Human Movement. Wiley, New York.

Winters, J.M., Kleweno, D.G., 1993. Effect of initial upper-limb alignment on muscle contributions to isometric strength curves. J. Biomech. 26, 143–153.

Wu, G., Van Der Helm, F.C., Veeger, H.E., Makhsous, M., Van Roy, P., Anglin, C., et al., 2005. ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—part II: shoulder, elbow, wrist and hand. J. Biomech. 38, 981–992.

Yadav, H., Nho, S., Romeo, A., Macgillivray, J.D., 2009. Rotator cuff tears: pathology and repair. Knee Surg. Sports Traumatol. Arthrosc. 17, 409–421.

Yamamoto, A., Takagishi, K., Osawa, T., Yanagawa, T., Nakajima, D., Shitara, H., et al., 2010. Prevalence and risk factors of a rotator cuff tear in the general population. J. Shoulder Elbow Surg. 19, 116–120.

Zajac, F.E., 1989. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. Crit. Rev. Biomed. Eng. 17, 359–411.