Original Articles

Computational analysis of subscapularis tears and pectoralis major transfers on muscular activity

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ABSTRACT

Background: Pectoralis major is the most common muscle transfer procedure to restore joint function after subscapularis tears. Limited information is available on how the neuromuscular system adjusts to the new configuration, which could explain the mixed outcomes of the procedure. The purpose of this study is to assess how muscles activation patterns change after pectoralis major transfers and report their biomechanical implications.

Methods: We compare how muscle activation change with subscapularis tears and after its treatment by pectoralis major transfers of the clavicular, sternal, or both these segments, during three activities of daily living and a computational musculoskeletal model of the shoulder.

Findings: Our results indicate that subscapularis tears require a compensatory activation of the supraspinatus and is accompanied by a reduced co-contraction of the infraspinatus, both of which can be partially recovered after transfer. Furthermore, although the pectoralis major acts asynchronously to the subscapularis before the transfer, its activation pattern changes significantly after the transfer.

Interpretation: The capability of a transferred muscle segment to activate similarly to the intact subscapularis is found to be dependent on the given motion. Differences in the activation patterns between intact subscapularis and the segments of pectoralis major may explain the difficulty in adapting psycho-motor patterns during the rehabilitation period. Thereby, rehabilitation programs could benefit from targeted training on specific motion and biofeedback programs. Finally, the condition of the anterior deltoid should be considered to improve joint function.

1. Introduction

A torn rotator cuff muscle may severely impair the mobility of the shoulder, resulting in restricted range-of-motion, pain, difficulties to perform common daily life activities and reduced quality of life (Gumina and Candela, 2017; Minagawa et al., 2013). Irreparable tears can be treated with a muscle transfer, where the insertion site of a functioning muscle is detached and transferred onto the insertion area of the torn muscle. The aim of such procedure is to restore the function of the torn muscle, while maintaining as much of the original function carried out by the transferred muscle as possible (Axe, 2016; Clark and Elhassan, 2018; Elhassan et al., 2010).

The subscapularis muscle is the largest muscle of the rotator cuff, also the only one with a relatively anterior position (Keating et al., 1993). Subscapularis tears have been increasingly reported, owing to imaging improvements and better awareness (Gerber et al., 1996; Gerber and Krushell, 1991; Lee et al., 2018b), with muscle transfer being a common treatment to restore the subscapularis function. Since the initial study by Wirth et Rockwood (Wirth and Rockwood, 1997), the latissimus dorsi, teres major (Elhassan et al., 2014), and pectoralis minor (Paladini et al., 2013; Wirth and Rockwood, 1997) have all been considered for muscle transfer for subscapularis tears. The most common choice, however, is the pectoralis major (PMA), with the options being a transfer of the entire muscle (Jost et al., 2003), the sternal part only (Elhassan et al., 2008; Valenti et al., 2011; Valenti et al., 2015), or the clavicular part only (Gavrilidis et al., 2010; Resch et al., 2000; Valenti et al., 2011; Valenti et al., 2015). Although PMA transfer has been shown to be less favorable in patients suffering from massive...
rotator cuff tears (Jost et al., 2003; Nelson et al., 2014; Shin et al., 2016), positive outcomes of PMA transfer include reduced pain (Gavriilidis et al., 2010; Moroder et al., 2017; Resch et al., 2000; Shin et al., 2016), improved activities of daily living (ADL) (Gavriilidis et al., 2010; Jost et al., 2003), higher Constant score (Elhassan et al., 2008; Ernstbrunner et al., 2019; Jost et al., 2003; Resch et al., 2000; Shin et al., 2016), and a better joint stability (Resch et al., 2000; Wirth and Rockwood, 1997). The choice of which muscle to transfer is typically based on anatomical considerations, with the literature consisting mainly of retrospective clinical studies assessing the efficacy of different transfers.

Previous biomechanical studies of muscle transfers have mainly focused on the transfer of the latissimus dorsi or the teres major for supero-posterior rotator cuff tears (Favre et al., 2008; Magermans et al., 2004a; Magermans et al., 2004b), while PMA transfer studies were limited to in-vitro kinematics or in-silico static analysis of simple motions (Jastifer et al., 2012; Konrad et al., 2007). However, in order to provide guidance on targeted rehabilitation, one needs to evaluate the complex active muscle interplay of the entire shoulder girdle for activities representative of daily life activities. Also, the transferred muscle is typically assumed to activate in-phase with the intact SSC (Gavriilidis et al., 2010; Shin et al., 2016), but if the activation pattern of the transferred muscle changes meaningfully compared to its preoperative state, the neurological adaptation necessary to cope with the new situation after a transfer may be more challenging for some patients.

Numerical models can simulate muscle activity during motion. Effects of various clinical procedures can then be tested in a reproducible fashion on the same anatomy, e.g. allowing for a direct comparison of different muscle transfers.

In this work, we compare the activation patterns of shoulder muscles during three ADL (eating, washing, and combing) for the following scenarios: intact shoulder, a pre-surgical condition with an isolated full tear of the subscapularis, and 3 different PMA transfers (clavicular, sternal, and both) using a multi-body, muscle-driven, finite element model of the shoulder (Péan and Goksel, 2020). Our hypothesis is that PMA transfers lead to meaningful differences in muscle activation patterns with respect to the preoperative setting. The purpose of this study is to predict the changes in muscles activation patterns induced by various pectoralis major transfers, and provide a better understanding of its biomechanical effects during three daily life activities.

2. Methods

In this section, our previously published model is first briefly described. Then, we introduce the morphing of this model to an intact subject for which motion capture data was available. The EMG and JRF obtained for this intact model are validated next. Finally, the model is used to compute muscle activations during several ADL, which allows us to study the active compensation mechanism of the pectoralis major after several types of transfers with respect to both the intact anatomy and to a pathological case with torn subscapularis.

2.1. Musculoskeletal model of the shoulder

The biomechanical model and the specific simulation parameters used herein were described in detail in previous works (Péan et al., 2021; Péan and Goksel, 2020). Some of the main features are summarized herein, for sake of completeness. Our multi-body model involves four rigid structures (thorax, clavicle, scapula, and humerus) and 19 muscles segments representing 13 muscles listed in Table 1, developed to study the active compensation mechanism of the pectoralis major after several types of transfers with respect to both the intact anatomy and to a pathological case with torn subscapularis.

<table>
<thead>
<tr>
<th>Name</th>
<th>Abbr.</th>
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<tbody>
<tr>
<td>Deltoid Anterior</td>
<td>DAN</td>
</tr>
<tr>
<td>Deltoid Middle</td>
<td>DMI</td>
</tr>
<tr>
<td>Deltoid Posterior</td>
<td>DPO</td>
</tr>
<tr>
<td>Infraspinatus</td>
<td>ISP</td>
</tr>
<tr>
<td>Latisimus Dorsi</td>
<td>LD</td>
</tr>
<tr>
<td>Levator Scapulae</td>
<td>LS</td>
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<tr>
<td>Pectoralis Major Abdominal</td>
<td>PMAA</td>
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<tr>
<td>Pectoralis Major Clavicular</td>
<td>PMAC</td>
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<tr>
<td>Pectoralis Major Sternal</td>
<td>PMAS</td>
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<tr>
<td>Pectoralis Minor</td>
<td>PMI</td>
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<tr>
<td>Rhomboid</td>
<td>RM</td>
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<tr>
<td>Serratus Anterior</td>
<td>SAN</td>
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<tr>
<td>Subscapularis</td>
<td>SSC</td>
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<tr>
<td>Supraspinatus</td>
<td>SSP</td>
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<td>Teres Major</td>
<td>TMA</td>
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<td>Teres Minor</td>
<td>TMI</td>
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<tr>
<td>Trapezius Inferior</td>
<td>TRI</td>
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<tr>
<td>Trapezius Middle</td>
<td>TRM</td>
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<tr>
<td>Trapezius Superior</td>
<td>TRS</td>
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Table 1 List of muscle segments in the model with their abbreviations.

Joint is implicitly incorporated via non-interpenetration constraints based on mesh collision detection at each timestep. The muscle parts were modeled as membrane elements, automatically meshed with B-spline surfaces (Péan and Goksel, 2020), which wrap on the bones using non-interpenetration constraints. Material model is modeled as a combination of an embedding matrix using a co-rotated Hooke’s law and a fiber component with passive and active parts modeled according to Blemker et al. (Blemker et al., 2005). The active part of the material represents muscle activation by a scalar parameter, normalized between 0 and 1, which is set by minimizing the differences in position and velocity between body markers from motion capture and the corresponding landmarks on the bone model. This is achieved by solving a Quadratic Problem at each timestep during the motion using the linearized model dynamics, which yields the aforementioned muscle activation values to input at the next timestep. A stability constraint enforcing the JRF to remain within the glenoid (Ambrosio et al., 2011; Blana et al., 2008; Bolsterlee et al., 2013; Dickerson et al., 2007; Favre et al., 2005; Sagl et al., 2019) was not necessary in the current study as JRFs pointed within the glenoid at all times.

2.2. Subject-specific modelling

Eating, washing, and combing trajectories of intact subjects from a publicly available dataset (Bolsterlee et al., 2014) were simulated. This dataset contains motion capture of several upper limb landmarks: 4 on the torso; 4 on the scapula; 3 on the clavicle; and 2 on the humerus (with exact locations defined by Wu et al. (Wu et al., 2005)). These landmarks were also annotated on our computational model. All motions for all 9 marker locations were transformed to a fixed thorax coordinate frame at all time instances by computing a rigid transformation of the thorax JRF to its initial position at t = 0. We follow the procedure described in (Péan et al., 2021) to register our model to the subject’s markers and anthropomorphic measurements.

2.3. Electromyography (EMG)

We used the raw EMG measurements from the above-mentioned dataset (Bolsterlee et al., 2014), which includes the following muscle parts: Deltoid (anterior, medial, posterior), Infraspinatus, Latisimus Dorsi, Pectoralis Major (clavicular, sternal), Trapezius (inferior, transverse, superior). Following the work of Staudenmann, et al. (Staudenmann et al., 2007), the raw EMG data was first filtered by a high-pass 1st order Butterworth filter at 250 Hz, then rectified, and filtered by a low-pass 1st order Butterworth filter at 2 Hz. The similarity between the model activation prediction and the processed EMG was evaluated using
Spearman’s $\rho$, yielding a correlation coefficient $x_{corr}$ that describes the monotonic relationship between the variables while obviating any parametric model assumption between them.

2.4. Comparison to in-vivo joint reaction forces

Simulated joint reaction force (JRF) magnitudes were compared to in-vivo measurements obtained from instrumented prostheses (Bergmann et al., 2007; Westerhoff et al., 2009). Comparison was performed for the combing motion, the only common activity between both datasets. In our results, four measurements (herein called S1, S2, S3, and S4) are included from the public Orthoload database.

2.5. Simulation of pectoralis major transfer

PMA is commonly divided into three regions, the clavicular part (PMAC), the sternal part (PMAS), and the abdominal part (PMAA), which were represented in our model by three independent muscle segments that can be activated separately. In line with previous work (Shin et al., 2016), transfer of PMAC and PMAS were studied. These were computationally performed by moving their respective insertion sites to the SSC insertion site, see Fig. 1.

Intact computational scenario with physiologically functioning SSC was taken as the baseline. An isolated, irreparable tear of the subscapularis was then simulated by removing the muscle from the simulation, therefore removing both its passive and active force-generating capacity; referred to as case “−SSC” with minus representing removal. Transfers of the clavicular part (+PMAC), sternal part (+PMAS), or both (+PMASC) were then performed on the model. For all conditions, the simulated activations for the intact case positively correlate with an average distance error for the humerus landmarks (EM, EL) of 1.2 cm for eating, 0.7 cm for washing, and 3.4 cm for combing (Supplementary Table 1). Landmark positions with a simulated SSC tear lie on average within 1.2 cm of the intact positions, and improves to below 0.9 cm after different transfers (Supplementary Table 2). The simulated JRF shows trends and magnitudes similar to the in-vivo data (Supplementary Table 2).

2.6. Analysis of the tear and the muscle transfers

Muscle transfer procedures aim not only to replace the function of the torn muscle by the transferred muscle, but also to keep the remaining muscles functioning as normal as possible, e.g. preventing them from overloading to avoid further tears. Specifically, the loss of the physiological contribution of the transferred muscle should not lead to secondary functional impairments. Accordingly, first we analyzed the impact of the tear and the following transfers on neighbouring muscles (SSP, ISP, and DAN), by comparing their activation levels to the intact case. Co-contraction of ISP with SSC is evaluated forimesteps where SSC activation is above an arbitrary threshold of 1% in the intact case. Similarly, changes in the activation signals of PMAC and PMAS were compared with their intact state.

In order to replace the active function of SSC, the transferred muscle should be activated when SSC was itself active in the intact case (Omid and Lee, 2013; Shin et al., 2016). This is evaluated as the percentage of time when PMAC or PMAS is active at the same time as SSC was active in the intact case. In this analysis, SSC, PMAC and PMAS were considered active when activation signals were above an arbitrary threshold of 1%. In addition, spearman correlation coefficients were computed to evaluate how closely the transferred PMAC or PMAS act in-phase with the intact SSC. Spearman correlation coefficients were also computed to assess how in-phase the PMAC activation is after its transfer, compared to its intact condition. Finally, stability of the glenohumeral joint is reported for all cases as the deviation of the JRF from a pure compressive direction, defined as the direction normal to the glenoid plane.

3. Results

3.1. Model validation

The three ADL could be successfully simulated by the intact model, with an average distance error for the humerus landmarks (EM, EL) of 1.2 cm for eating, 0.7 cm for washing, and 3.4 cm for combing (Supplementary Table 1). Landmark positions with a simulated SSC tear lie on average within 1.2 cm of the intact positions, and improves to below 0.9 cm after different transfers (Supplementary Table 2). The simulated JRF shows trends and magnitudes similar to the in-vivo data (Supplementary Fig. 1). Simulated activations for the intact case positively correlate ($x_{corr} > 0.5$) with EMG for at least one motion for all reported parts except PMAS (Supplemental Table 3).

3.2. Muscle activity

Activation of the SSP with the torn SSC was consistently higher than in the intact case, with a mean difference of 11.2, 7.2, and 2.3%, respectively, for the eat, wash, and comb motions. Muscle transfers generally led to a reduction of the activation of the SSP compared to the torn SSC case (Table 2) but not to the level of the intact baseline, as clearly visible, for example, for the eating motion (Fig. 2).

Activation of the ISP in the torn SSC case was lower than in the intact case (Fig. 3). The mean difference of activation during co-contraction periods are: $-14.9$, $-8.1$, and $-4.2%$, respectively, for the eating, washing, and combing motion. Muscle transfers generally led to an increase of the activation of the ISP compared to the torn SSC case (Table 2) but not to the level of the intact baseline, as clearly visible, for example, for the eating motion (Fig. 2).

The activation of the anterior part of the deltoid increased after PMAC transfer (Fig. 4). The transfer of the clavicular part resulted in an average increase in DAN activation of 5, 10.3, and 13.5%, while the transfer along with the sternal part increased on average by 6, 20.6, and 16.2% for the eat, wash, and comb motion, respectively.

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Supplementary Table 1: Landmark positions with a simulated SSC tear lie on average within 1.2 cm of the intact positions, and improves to below 0.9 cm after different transfers (Supplementary Table 2). The simulated JRF shows trends and magnitudes similar to the in-vivo data (Supplementary Fig. 1). Simulated activations for the intact case positively correlate ($x_{corr} > 0.5$) with EMG for at least one motion for all reported parts except PMAS (Supplemental Table 3).

Supplementary Table 2: Simulated JRF magnitudes were compared to in-vivo measurements obtained from instrumented prostheses (Bergmann et al., 2007; Westerhoff et al., 2009). Comparison was performed for the combing motion, the only common activity between both datasets. In our results, four measurements (herein called S1, S2, S3, and S4) are included from the public Orthoload database.

Supplementary Table 3: Muscle transfers generally led to a reduction of the activation of the SSP compared to the torn SSC case (Table 2) but not to the level of the intact baseline, as clearly visible, for example, for the eating motion (Fig. 2).

Fig. 1. Intact PMA (a) simulated transfer of (b) clavicular part, (c) sternal part, and (d) sternoclavicular part. Pectoralis major clavicular, sternal, abdominal regions are shown in yellow, green, blue, respectively. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)
After their respective transfer, PMAC and PMAS activation coincided more with intact SSC activation for the eating and combing motions (Table 3). For the washing motion, PMAC was already highly active when the intact SSC was active, and it improved slightly after transfer. However, the opposite was seen with PMAS. The activation of PMAC and PMAS in the intact case are negatively correlated with the intact SSC activation for all motions (Fig. 5). Although no clear trend was observed for the eating and combing motions ($\chi_{corr} \leq 0.19$), during washing motion PMAC pattern after transfer correlates positively with the intact SSC ($\chi_{corr} \geq 0.64$), see Table 4. PMAS, instead, positively correlates after its transfer with the intact SSC for the eating motion ($\chi_{corr} \geq 0.21$).

With a torn SSC, PMAC and PMAS both activate similarly to their respective intact condition ($\chi_{corr} \geq 0.67$), see Table 5. However, PMAC and PMAS activation patterns mostly changed significantly after their transfer, compared to their intact activations.

On average, joint stability was worse when the SSC was torn compared to the intact condition, with an increased deviation of 17, 6, and 1$^\circ$ in average for the eating, washing, and combing motion, respectively. After PMA transfers, stability was generally slightly improved compared to the torn SSC case, with a decreased deviation in the best case of 6, 1 and 3$^\circ$ (Supplemental Table 4).

### 4. Discussion

In this study, we report validations with two essential model outputs for the intact anatomy. The computed muscle activation generally matched positively with the EMG of the reference subject for a large set of muscles, showing that our model can predict realistic activation motion PMAC pattern after transfer correlates positively with the intact SSC ($\chi_{corr} \geq 0.64$), see Table 4. PMAS, instead, positively correlates after its transfer with the intact SSC for the eating motion ($\chi_{corr} \geq 0.21$).

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During the combing motion, the SSC is mostly inactive but the observed effects of SSC tear on ISP co-contraction may be the result of passive forces by SSC tissue.

Third, a torn SSC induces significant changes in activation and loading of the SSP. The higher SSP load may explain why SSC tears, when not treated, often propagate in time superiorly leading to SSP tears (Lee et al., 2018a; Lenart and Ticker, 2017). Our simulated transfers led to a reduction of the SSP activation compared to the torn SSC case, explained by the transferred PMA taking over the SSC function. Therefore, an early diagnosis of such pathology and its repair with a muscle transfer might limit SSP overuse and hence the risk of tear propagation from an isolated tear into a massive rotator cuff pathology. This is especially important with such transfers having been shown to be less successful in patients suffering from massive tears (Jost et al., 2003; Nelson et al., 2014; Shin et al., 2016).

Furthermore, the transferred muscle is typically assumed to activate in-phase with the intact SSC (Gavriilidis et al., 2010; Shin et al., 2016). This assumption often leads to omitting any biofeedback exercises during rehabilitation (Omid and Lee, 2013). Our simulations showed that PMAC and SSC do not act in-phase during complex movements, but rather complementary to one another. The activation pattern of the transferred muscle changes meaningfully compared to its initial state. Therefore, the neurological adaptation necessary to cope with the new situation after a transfer may be challenging for some patients, as they would need to re-learn a completely different neuromuscular control pattern. Note that we focused in this work on the capability of transferred PMA to actively take over the role of the subscapularis. The absence of changes in the activation patterns does not exclude that the desired function can be achieved by the passive effect of the muscle in its new location.

The activation patterns generated by the simulations are mechanically optimal with respect to following a target, while being minimal with respect to overall activation. However, despite positive correspondences with EMG in the intact case, the simulated activation might not be necessarily neurologically possible in the general case as some biological structures and constraints are omitted in the model, e.g., co-contraction for enforcing joint stability.

In this work, we considered PMA transfers on a single subject, for a limited number of ADL. Subject variability may be a potential contributor in the observations and further studies are required to generalize the results of this work. While the three considered activities are common ADL covering relatively different ranges, they still do not cover the full range of shoulder motions. Simulating supplementary movements might provide additional insights.

Due to the absence of ground truth for the optimal rest stretch ratio in our muscles models, we assumed that the rest position for all muscles was at the standard anatomical position, whereas in reality, many muscles are not at their intrinsic optimal rest length in the standard anatomical position (Mount et al., 2003). This might, for example, have restricted the range of motion, especially over head, due to some muscles not stretching as much as physiologically possible. Alternatively, some muscles may also have been unable to contract as much as physiologically possible. We indeed observed this for latissimus dorsi and teres major for the combing motion, which required an increased value of optimal rest stretch ratio compared to default value. Our model does not simulate inter-muscles collisions, which prevents the evaluation of PMA transfers underneath or over the conjoint tendon, which would allow a stability vs. risk tradeoff (Nelson et al., 2014; Thompson et al., 2020).

A comprehensive validation of the model would require data from patients before and after the transfer procedure. However, due to the unavailability of such information, the model was validated for the intact anatomy against EMG of the motion-captured source subject and available JRF data from subjects with an instrumented prosthesis. In the absence of motion data for the various pathological conditions considered in the current study, our model followed the same trajectories for patterns. Also, the simulated joint reaction force conformed to in-vivo measurements of subjects with instrumented anatomical prosthesis.

In addition, the model was able to reproduce several published clinical observations: First, the transfer of PMAC leads to a substantial increase in DAN activation, see Fig. 4, where the activation of the intact DAN (in blue) is consistently lower than when PMAC is transferred (yellow and purple curves). Over time, the increased usage of DAN may lead to muscle hypertrophy. This model observation is in agreement with the clinical long-term post-operative observation of DAN excessively bulging after PMA transfers (Resch et al., 2000). In addition, current surgical indications include a functioning deltoid as a prerequisite for the PMA transfer (Nelson et al., 2014).

Second, co-contraction of the ISP, an antagonist of SSC, is reduced for all motions in the torn SSC case, when compared to the intact case. Such a reduction in co-contraction may negatively affect joint stability (Gasbarro et al., 2017; Péan et al., 2021), by reducing the compressive JRF component and therefore increasing the shear-to-compressive force ratio. We observed the PMA transfer to partially restore co-contraction for eating and combing motions, which may improve joint stability.

### Table 4

Spearman correlation coefficients between the activation signal of SSC in the intact case and the activation signals of PMAC and PMAS in intact case and after both transfer. Empty cells indicate no correlation ($\rho=\text{NaN}$), as PMAS shows no activation in the intact case.

<table>
<thead>
<tr>
<th></th>
<th>PMAC</th>
<th>PMAS</th>
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<tbody>
<tr>
<td></td>
<td>intact +PMAC +PMASC</td>
<td>intact +PMAS +PMASC</td>
</tr>
<tr>
<td>EAT</td>
<td>-0.55</td>
<td>-0.38</td>
</tr>
<tr>
<td>WASH</td>
<td>-0.64</td>
<td>0.64</td>
</tr>
<tr>
<td>COMB</td>
<td>-0.70</td>
<td>-0.19</td>
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</table>

### Table 5

Spearman correlation coefficients between activation signals in the intact case, and with torn SSC and after both transfer. Empty cells indicate no correlation ($\rho=\text{NaN}$), as PMAS shows no activation in the intact case.

<table>
<thead>
<tr>
<th></th>
<th>PMAC</th>
<th>PMAS</th>
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<tbody>
<tr>
<td></td>
<td>-SSC</td>
<td>+PMAC</td>
</tr>
<tr>
<td>EAT</td>
<td>0.96</td>
<td>0.62</td>
</tr>
<tr>
<td>WASH</td>
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<td>0.06</td>
</tr>
<tr>
<td>COMB</td>
<td>0.92</td>
<td>0.19</td>
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</table>

A comprehensive validation of the model would require data from patients before and after the transfer procedure. However, due to the unavailability of such information, the model was validated for the intact anatomy against EMG of the motion-captured source subject and available JRF data from subjects with an instrumented prosthesis. In the absence of motion data for the various pathological conditions considered in the current study, our model followed the same trajectories for
all conditions, although the kinematics to perform the same ADL might be altered after a transfer.

5. Conclusions

The simulated muscle activation patterns differ greatly after transfers compared to preoperative state, which may explain the inability of some patients to adapt after a PMA transfer. In particular, rehabilitation exercises after transfer of the clavicular part of the pectoralis major might benefit from focusing on washing and combing motions, where activation patterns were antagonistic to the preoperative state. A transfer of the clavicular part of the pectoralis major alone, or together with the sternal part, may not be successful when the patient suffers from a weak anterior deltoid. In general, the simulated muscle transfers considered in this study were shown to reduce the activation on the supraspinatus, decreasing the risks of tear propagation to the supraspinatus. The simulated muscle transfers also restored antagonistic contraction of the infraspinatus, potentially improving joint stability.

Declaration of Competing Interest

Authors declare no conflict of interest.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2021.105541.

References


