



# Serratus anterior weakness is a key determinant of arm-assisted standing difficulties

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## ABSTRACT

The ageing population has led to recent increases in musculoskeletal conditions, with muscle weakness a major contributor to functional decline. Understanding the early phases of muscle weakness will help devise treatments to extend musculoskeletal health. Little is understood of the effects of muscle weakness on everyday activities such as sit-to-stand, a determinant of mobility that, in the early stages of weakness, requires upper limb compensation. This experimental and computational modelling study investigated the effects of muscle weakness on upper-extremity muscle forces of 27 healthy adults when using arm rests. Weakness of 29 upper limb muscles was simulated by individually removing each from a musculoskeletal model. Serratus anterior weakness was highlighted as detrimental, with the model unable to fully solve the loadsharing redundancy in its absence, and forces at the elbow and glenohumeral joint and in other muscles were found to be profoundly increased. Its large number of fast-twitch muscle fibres, predisposed to atrophy with age, highlight the centrality of the serratus anterior as a key determinant of mobility in this critical task and a potential source of early immobility through its preferential loss of strength and thus point to the requirement for early clinical interventions to mitigate loss.

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## 1. Introduction

Recent large increases in the number of over 65 year olds has led to a rise in musculoskeletal (MSK<sup>1</sup>) conditions resulting in loss of mobility, increases in pain and social isolation [1], leading to a downward spiral of function [2]. This decline frequently starts with muscle weakness (MW) and loss of strength [3,4]; the study of MSK biomechanics to increase understanding of the aetiology of this weakness can facilitate targeted therapies. MSK modelling is a tool used to analyse the mechanics of movement, providing information on physiological parameters that are difficult to measure in vivo, such as joint contact and muscle forces.

Theoretical MW is a concept applied to MSK models to simulate the physiological diminishing of strength through alterations to the mathematical optimisation of muscle loadsharing; this has been widely explored in the literature [5–14] through, for example, arbitrary scalar reductions to the optimisation loadsharing constraints. These previous studies have mostly focused on the

effects of lower limb MW (Table 1), with the upper limbs poorly represented despite their importance in carrying out activities of daily living (ADLs).

Rising from a chair is a demanding task requiring high core and leg muscle strength [19–24], therefore difficulties in sit-to-stand (STS) can be attributed to lower limb MW. This leads to compensations such as increased trunk movement or use of arm rests to assist STS completion [25]. Previous investigations into STS were limited to functional muscle group power which cannot fully quantify the contributions of individual muscles. Synergistic and antagonistic muscle activity in movement make it difficult to define which muscles should be targeted for rehabilitation through physiotherapy or strength training. Further work is needed to focus the clinical application of improving STS performance, a motion noticeably missing from the literature in the context of upper limb MW.

One approach to identifying muscles vulnerable to age-related weakness is through muscle fibre-specific atrophy. Fibre type distribution changes with age, with a proportionally lower reduction in Type 1 muscle fibres (slow twitch, low fatigue) than the higher proportion reduction in the number and size of Type 2 fibres (fast twitch, high fatigue [26,27]); resulting in a drastic reduction in contractile properties for muscle classed as the latter. A summary of fibre type distributions for upper limb muscles is shown in

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<sup>1</sup> MSK – musculoskeletal modelling, MW – muscle weakness, STS – sit-to-stand, SA – Serratus Anterior, JRF – joint reaction forces, GH – glenohumeral

**Table 1**  
Studies simulating muscle weakness. LL = lower limb, UL = upper limb.

Study	Muscle group	Software/MSK model used	Assessment of the effects of MW	Outcomes
Afschrift et al. [9]	LL	Generic MSK models in OpenSim [15]	Tolerance of measured kinematics of 3 activities to MW: Quantified as the presence of a difference between required joint torques imposed by task, and moment generation by a muscle model, termed the capability gap (CG). Muscle force upper bounds reduced for all muscles	Reducing the maximum muscle force by 30% led to a CG during stair ascent/descent, and 40% for gait. Sit-to-stand most tolerant, with up to 70/80% weakness before a CG was found
Carmichael & Liu [8]	UL	Holzbaur et al. [16]	Matrix of impairments (of various severities) based on stroke patient data, applied to strength estimation model. Assessed difference in CG, as above, when drinking from a bottle	Not all impairment severities hindered user capability, the CG between subject strength and task requirement varied throughout motion. MW is not always inhibiting
van der Krogt et al. [10]	LL	Generic MSK models in OpenSim [15]	Assessment of the tolerance levels of gait to MW. All muscles weakened, then as groups, in 20% steps. Tolerance quantified by changes in joint angles, use of reserve muscles or an unsuccessful muscle force loadsharing optimisation	Up to 40% MW tolerated. Compensations could cause further overuse issues through increased antagonistic action
Bull et al. [6]	LL	UKNSM [17]	Inhibiting muscle activation during various ADLs	Tasks theoretically unachievable in weakened state
van Drongelen et al. [13]	UL	Delft Shoulder and Elbow Model (DSEM) [18]	Investigating the relationship between MW and shoulder joint loading during weight lifting. MW of muscles related to a lesion level at known innervation points on spine. Loadsharing optimisation success in FD model also assessed	When over 80% of muscles had a reduced maximum force, no solutions were found. At 45%, simulations were possible. Minor effects seen on GH contact force. Shows how kinematics can be altered at various weakness tolerance levels in order to complete task
Komura et al. [14]	LL	Delp et al. [15]	Full deactivation of individual muscles to observe effects on other muscles and joint contact forces during gait	Vital muscle paralysis did not stop motion, made possible by synergy re-arrangement. Rehabilitation should train alternative muscles. Some effects were seen on joint contact forces despite a lack of affiliation to deactivated muscles

supplementary data - Table S1. This table also includes muscle name abbreviations used in this study.

Most muscles contain an equal balance of Type 1 and 2 fibres. However, a small selection, for example serratus anterior (SA) and pectoralis, contain over 60% Type 2. This may suggest an increased susceptibility to weakening with age due to the associated greater loss of Type 2 fibres, and further investigation into the specific weakness of these muscles could explain ADL difficulties. The function of SA during STS has not previously been investigated. Considering its role in stabilising the scapula during elevation [28], weakness of this muscle may prove detrimental to the success of an MSK model in producing a solution to the loadsharing optimisation.

The aim of this study was to investigate the theoretical effects of localised MW on STS performance in the upper limbs of a young population. It is hypothesised that weakness of individual muscles of the upper limb can be identified as key determinants of the ability to perform the STS motion. This could then potentially be used to investigate initial points of early compensations which could lead to age-related dysfunction.

## 2. Materials and methods

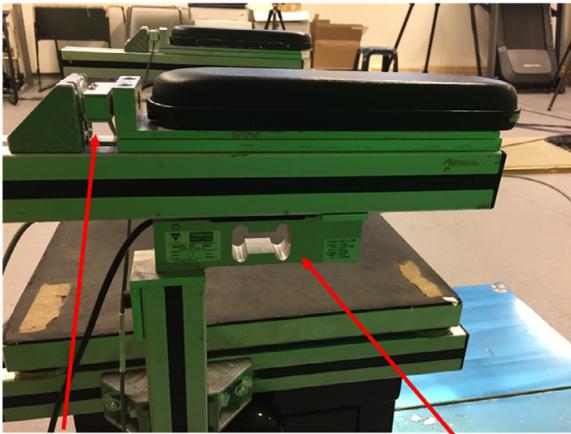
Power analysis (alpha of 0.05 and power of 80%) using previous work investigating differences in rising strategy in the young and old [29] found that a minimum group size of 25 was required.

27 healthy male subjects (aged  $28 \pm 5$  years) participated in the study. Ethical approval was received through the Joint Research Compliance Office and Imperial College Research Ethics Committee.

**Table 2**  
Anatomical locations of retro-reflective markers – right hand only tracked.

Segment	Anatomical location
Thorax	7th cervical vertebra
	Sternum jugular notch
	Sternum manubrium
	Sternum xiphoid process
Clavicle	Acromioclavicular joint
	Sternoclavicular joint
Scapula	Scapula tracker [30] affixed to the scapula spine and acromion
	Cluster of three markers placed on the upper arm
Humerus	Lateral epicondyle
	Medial epicondyle
	Cluster of three markers placed on the forearm
Forearm	Ulnar styloid
	Radial styloid
	Cluster of three markers placed on the back of the hand
Hand	Second metacarpal
	Fifth metacarpal

Retro-reflective markers were placed on the hands, arms, shoulders and torso of each participant (Table 2) using hypoallergenic tape and motion was captured using a 10-camera optical motion capture system (Vicon Motion Systems, Oxford, UK). A validated scapula locator was used to establish a local coordinate frame for each participant and trackers were used to measure kinematics dynamically [30]. External force data was obtained through load cells placed in the arm rests of an instrumented chair [31] with a set seat height of 46 cm (Fig. 1). A single point load cell was used to measure vertical hand forces (Vishay, model 1042-0050-G506R)



Tension/  
Compression  
load cell

Single point load cell

**Figure 1.** Instrumented arm rest. A positive force is defined as being in the superior (single point load cell) and anterior (tension/compression load cell) directions.

and a tension/compression load cell was used for the measurement of anterior-posterior hand forces (Omega, model LCM703-25). Each participant was asked to stand from a seated position 3 times, ensuring use of the arm rests.

STS phases were adopted from the literature [23,31–33]. The start was defined as trunk movement initiation and, having left the seat, the point at which hip flexion angular velocity first reached  $0^\circ/\text{sec}$  signified the end. Trials were normalised from 0 – 100% of a STS movement.

The United Kingdom National Shoulder model (UKNSM - [17]) was used to quantify muscle and joint reaction forces (JRF). The UKNSM includes four joints, three ligaments, 87 muscle elements (grouped into 29 functional parts, forming 21 muscles) and all upper limb bones (except the hand), including the thorax. The lower part of the trunk (the abdomen) is not included in this model. Bone geometry for anatomical scaling was taken from the Visible Human Dataset [34] and muscle morphology (muscle origins and insertions and physiological cross sectional areas - PCSA) from published cadaveric work [35–37]. Body segment parameters were calculated using De Leva [38]. Scapula movement was constrained through a kinematic optimisation, which defined the scapula thoracic gliding plane based on functional movements and anatomical landmarks [39,40]. Reflective marker positions were used to scale the MSK model to each participant using the aforementioned anatomical datasets. The formation of the equations of motion for each body segment informs an inverse dynamics algorithm to calculate intersegmental forces and moments. These subsequently inform a loadsharing optimisation to determine individual muscle forces through the minimisation of a cost function (sum of muscle stresses squared [17]) and within set muscle force boundaries. Muscles were modelled as linear actuators; active and passive contributions are not required for inverse dynamics-based optimisation.

The difficulty in obtaining muscle force boundaries was addressed by creating an unbounded optimisation, a method previously used by Prinold [40] to identify pivotal muscles in extreme activities, and in this study is used to model the infinite capacity required to achieve the task. This meant removing any physiological maximum force for the remaining muscles. To separately test the effect of the unbounded optimisation on individual muscles, the percentage activation (muscle force normalised to maximum muscle force) was compared between full strength and weakened

states to observe how MW may breach these maximum values, calculated as PCSA multiplied by a maximum muscle stress of  $100 \text{ Ncm}^{-2}$  [17].

MW was simulated by separately removing each of the 29 muscles from the UKNSM, leaving only 28 functional muscle lines of action for each repetition of the analysis. Previous studies (see Table 1) made use of specific levels of weakness, as opposed to the scenario of complete removal used in the current study. There is little evidence in the literature to justify using subject-specific levels of weakness between that of fully functional and weakened. The measurement of these levels for individual muscles is impractical, and fell outside the scope of the current work.

STS performance was evaluated based on the number of successful loadsharing solutions produced by the mathematical optimisation and the effect this had on the activation of the remaining muscles and the changes in glenohumeral and elbow JRF.

Failed optimisations were calculated per trial. In the context of this study, a ‘failed’ optimisation referred to the mathematical optimisation being unable to find a solution to the muscle force load-sharing problem. Data from failed outputs were excluded. For each participant, the number of time frames with failed optimisations were found for each muscle disabled for each of the three trials. These time frames were converted into a percentage of the trial length (to the nearest integer), and if a failure occurred at that percentage for at least two of the trials, it was included in the subject’s failure count. The number of time frames with failures was then summed for each participant.

The Shapiro-Wilk criterion was used to test for normality and statistical significance was assessed, either with paired-sample *t* tests or related-sample sign tests with an alpha level of 0.05, for the following: difference in peak muscle force (and summation of all peak muscle forces), JRF, and remaining muscle forces between full strength and MW scenarios.

### 3. Results

All data were non-normally distributed, hence Median data is presented. The prevalence of failed optimisations is shown in Fig. 2. SA proved the most debilitating, with the UKNSM unable to produce solutions during portions of the motion for 18 participants. All other participants not included in the figure had 100% optimisation success rates.

The effects of removing a muscle from the optimisation on peak JRF and remaining muscle forces are shown in Table 3. Each muscle removed resulted in significant increases and/or decreases in peak force in at least one other muscle ( $p < 0.0005$  to  $p = 0.043$ ). 14 resulted in significant changes in glenohumeral JRF ( $p < 0.0005$  to  $p = 0.021$ ) and 8 for the elbow ( $p < 0.0005$  to  $p = 0.021$ ). Removal of the SA muscle resulted in the most extreme changes: 18 out of the 28 remaining muscles had significantly increased peak muscle forces ( $p < 0.0005$  to  $p = 0.021$ ), with 4 showing decreased values ( $p < 0.0005$  to  $p = 0.007$ ). This also resulted in large significant increases in both the glenohumeral and elbow JRF (Table 3;  $p < 0.0005$ ).

These tables and following figures include data for the SA muscle which are considered non-physiological owing to the unbounded loadsharing optimisation. These have been highlighted in red in tables and placed on separate axes for graphs, and have been included to emphasise the effects of SA’s absence. The data should not be considered as factual, but instead used to interpret scenarios where subjects would most likely have to alter their kinematics to realistically achieve the task.

Fig. 3 presents the median sum of all remaining peak forces for each muscle removed from the UKNSM. Despite the presence of significant differences between full-strength and all other

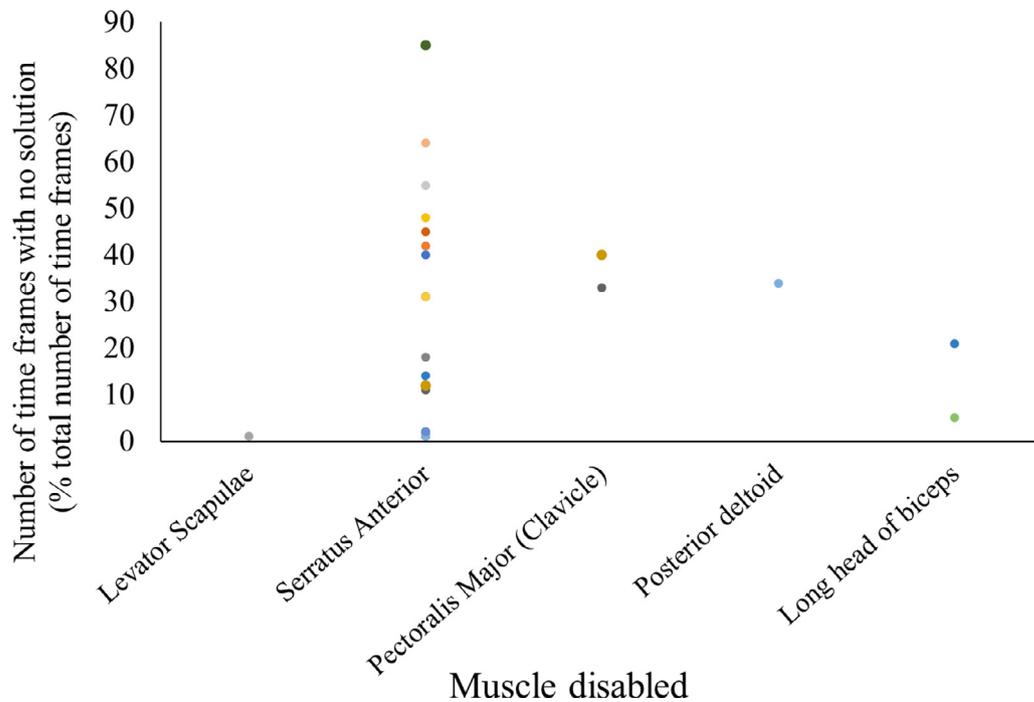


Figure 2. Mean percentage of failed time frames for all participants whose data included failed optimisations. Each mark represents a different participant.

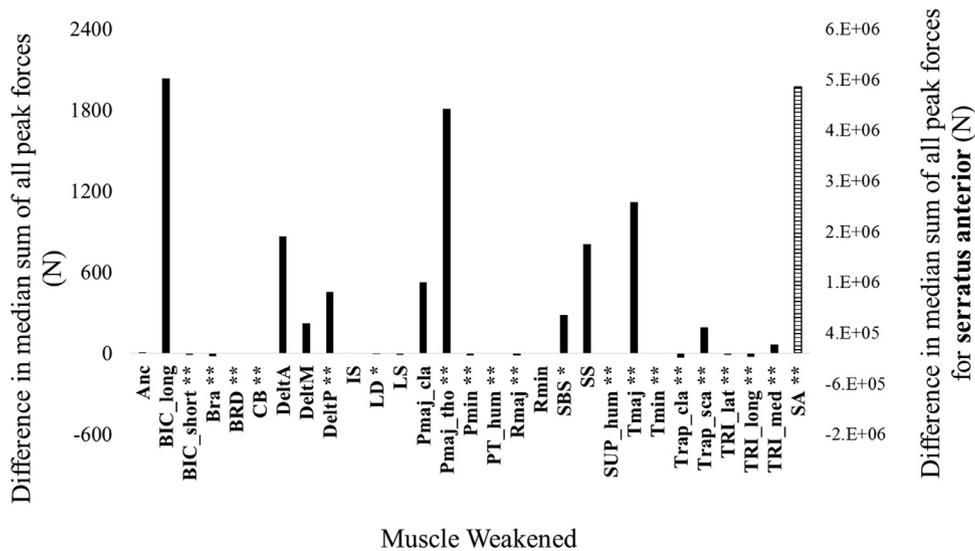


Figure 3. Difference in median sum of all peak muscle forces between full strength and weakened muscle states. Serratus anterior (dashed) has been placed on a secondary axis due to the large forces associated with its removal. \* ( $p < 0.05$ ) and \*\* ( $p < 0.0005$ ) indicate statistically significant differences to the full-strength state.

‘weakened’ states, only removal of 7 of the 29 muscles modelled in the UKNSM caused a significant increase in total peak muscle force when compared to the full-strength model. 13 of the 29 muscles caused a significant decrease in total peak muscle force, however, the absolute differences were small (ranging from 0.02% to 1.02% relative to the full strength) compared to the increases in total peak muscle force (1.96% to 54.52%, N/A for SA).

The effect of essentially removing a maximum muscle force for all muscles resulted in few that would theoretically exceed a maximum level (Supplementary data and Table 4). BIC\_long required extreme activation under removal of many muscles, and, removing SA resulted in the majority of other muscles requiring in excess of 100% activation for a successful optimisation solution.

#### 4. Discussion

Localised MW does not exclusively affect immediate joints and muscles with similar lines of action. It has previously been shown that antagonistic action of compensatory muscles will also increase [10]. One example, from this study, is how removal of the pectoralis major (Pmaj) affected elbow JRF. The prime elbow movers are the biceps and triceps. Both these muscles were found to significantly increase in peak force under Pmaj removal, whose’ role in arm adduction is synergised by TRI\_long (long head of the triceps), which saw an increase of approximately 200% during pectoralis weakness. This action, plus the increased antagonistic action of both the short (590% relative to full strength) and long (114%) heads of the biceps brachii, subsequently increased elbow JRF.

**Table 3**

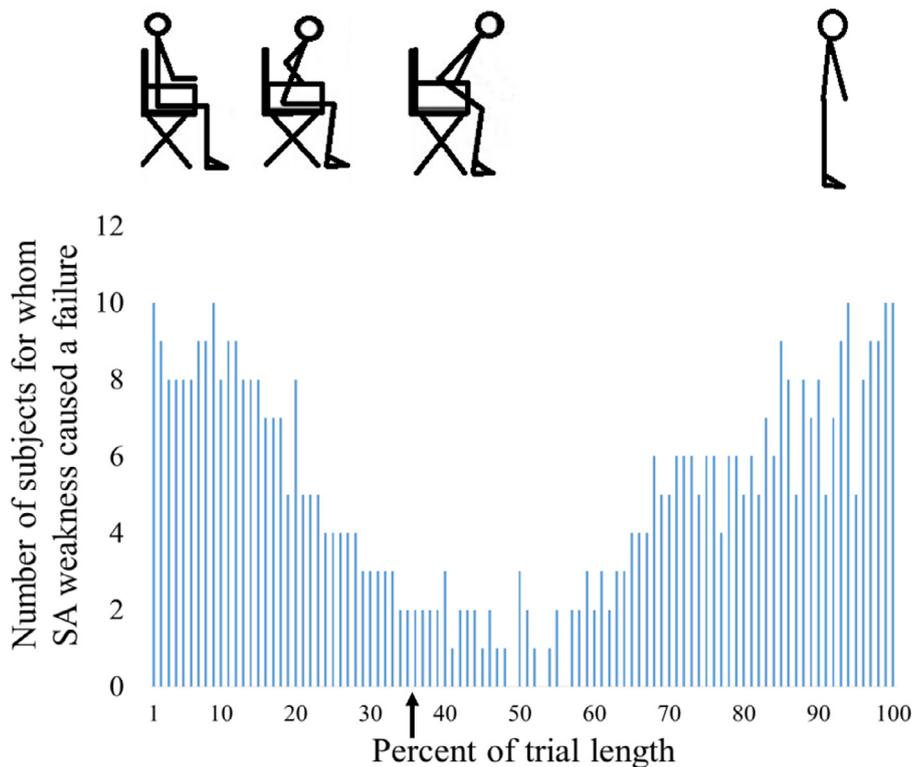
Peak forces for each muscle modelled in the UKNSM under removal of listed muscles. Bold ( $p < 0.05$ ) or underlined ( $p < 0.0005$ ) values show that removing the muscle in this row caused a statistically significant difference in peak muscle force for the muscle in this column. 100% indicates no change.

	Anc	BIC_long	BIC_short	Bra	BRD	CB	DeltA	DeltM	DeltP	IS	LD	LS	Pmaj_clav	Pmaj_thor	Pmin	PT_hum
Median full strength (N)	43	227	12	40	10	1	175	256	132	17	23	50	46	443	24	1
Muscle removed	Median muscle force post muscle removal (% of full-strength force values)															
Anc	100	99	100	100	91	<b>101</b>	<b>101</b>	<b>100</b>	100	<b>100</b>	<b>100</b>	<b>105</b>	109	100	<b>100</b>	<b>108</b>
BIC_long	<b>98</b>	<b>914</b>	<b>155</b>	<b>148</b>	27	<b>111</b>	<b>252</b>	118	<b>40</b>	<b>168</b>	<b>100</b>	121	<b>163</b>	<b>186</b>	<b>115</b>	
BIC_short	100	<b>100</b>	119	<b>100</b>	<b>191</b>	100	100	100	<b>107</b>	102	100	100	100	100	100	<b>100</b>
Bra	100	100	<b>179</b>	<b>198</b>	100	100	<b>100</b>	<b>101</b>	<b>115</b>	<b>103</b>	100	100	100	100	<b>102</b>	<b>123</b>
BRD	100	100	<b>122</b>	<b>120</b>	100	100	100	<b>100</b>	<b>105</b>	100	100	100	100	100	<b>100</b>	<b>100</b>
CB	100	100	<b>111</b>	100	100	100	100	100	100	100	100	100	100	100	100	100
DeltA	<b>164</b>	<b>142</b>	98	<b>99</b>	<b>100</b>	73		<b>168</b>	100	91	93	<b>171</b>	<b>175</b>	<b>121</b>	100	<b>177</b>
DeltM	<b>129</b>	106	<b>49</b>	87	64	82	<b>110</b>		<b>206</b>	112	<b>161</b>	<b>145</b>	<b>122</b>	98	<b>53</b>	<b>154</b>
DeltP	100	100	100	<b>140</b>	<b>118</b>	100	100	<b>173</b>		<b>386</b>	<b>124</b>	99	105	100	<b>141</b>	100
IS	<b>100</b>	100	<b>103</b>	<b>102</b>	<b>100</b>	<b>100</b>	100	100	100		100	100	100	100	100	100
LD	<b>100</b>	100	<b>133</b>	<b>118</b>	<b>102</b>	118	100	105	100	97		98	100	<b>100</b>	<b>103</b>	100
LS	<b>101</b>	100	99	100	100	91	<b>100</b>	<b>103</b>	100	100	99	<b>115</b>	100	<b>100</b>	<b>108</b>	100
Pmaj_clav	124	126	<b>172</b>	114	101	<b>145</b>	67	156	103	<b>107</b>	91	169		<b>114</b>	117	154
Pmaj_thor	<b>161</b>	<b>228</b>	<b>619</b>	<b>97</b>	77	<b>4336</b>	<b>149</b>	<b>94</b>	99	150	<b>795</b>	<b>176</b>	<b>366</b>		117	<b>146</b>
Pmin	100	100	<b>100</b>	<b>115</b>	102	<b>100</b>	100	<b>100</b>	<b>103</b>	100	<b>102</b>	<b>102</b>	98	100		100
PT_hum	<b>100</b>	100	100	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>	<b>101</b>	100	<b>100</b>	<b>100</b>	100	100	100
Rmaj	<b>101</b>	102	<b>101</b>	106	100	<b>127</b>	100	100	<b>101</b>	100	120	<b>109</b>	<b>103</b>	<b>100</b>	<b>107</b>	<b>108</b>
Rmin	<b>101</b>	100	100	100	100	100	<b>100</b>	101	100	100	103	<b>108</b>	<b>104</b>	<b>100</b>	98	<b>100</b>
SA	<b>5 × 10<sup>5</sup></b>	<b>4 × 10<sup>5</sup></b>	<b>8 × 10<sup>5</sup></b>	<b>5 × 10<sup>5</sup></b>	<b>1 × 10<sup>6</sup></b>	<b>2 × 10<sup>7</sup></b>	191	<b>4 × 10<sup>5</sup></b>	<b>2 × 10<sup>5</sup></b>	<b>1 × 10<sup>6</sup></b>	<b>3</b>	128	514	<b>122</b>	<b>3 × 10<sup>3</sup></b>	<b>7 × 10<sup>6</sup></b>
SBS	100	100	<b>123</b>	100	98	<b>100</b>	102	<b>100</b>	<b>199</b>	100	100	100	100	100	100	100
SS	<b>138</b>	<b>118</b>	<b>61</b>	<b>98</b>	<b>84</b>	36	<b>108</b>	104	103	<b>187</b>	84	<b>115</b>	185	102	99	123
SUP_hum	<b>100</b>	100	<b>100</b>	<b>100</b>	<b>100</b>	100	<b>100</b>	<b>100</b>	100	<b>100</b>	100	<b>100</b>	100	100	<b>100</b>	<b>100</b>
Tmaj	<b>157</b>	<b>134</b>	86	106	93	73	106	93	<b>203</b>	88	<b>841</b>	<b>187</b>	181	<b>86</b>	<b>62</b>	<b>146</b>
Tmin	100	100	<b>100</b>	100	100	100	100	100	100	101	<b>100</b>	<b>100</b>	100	100	100	100
Trap_cla	<b>103</b>	100	100	<b>107</b>	<b>97</b>	100	<b>100</b>	100	<b>101</b>	<b>105</b>	<b>79</b>	102	89	<b>100</b>	<b>88</b>	<b>100</b>
Trap_sca	<b>114</b>	<b>117</b>	<b>159</b>	125	113	<b>209</b>	102	104	104	106	84	100	154	94	95	<b>123</b>
TRI_lat	<b>109</b>	92	<b>87</b>	100	100	64	<b>100</b>	101	100	100	<b>114</b>	<b>98</b>	102	101	<b>100</b>	<b>115</b>
TRI_long	<b>121</b>	<b>86</b>	96	100	100	<b>164</b>	99	100	<b>101</b>	100	<b>84</b>	<b>115</b>	<b>123</b>	<b>99</b>	100	108
TRI_med	<b>120</b>	92	<b>88</b>	<b>95</b>	<b>76</b>	91	<b>102</b>	<b>106</b>	<b>99</b>	99	<b>113</b>	102	126	102	<b>126</b>	<b>131</b>
Median full strength (N)	Rmaj	Rmin	SA	SBS	SS	SUP_hum	Tmaj	Tmin	Trap_clav	Trap_scap	TRI_lat	TRI_long	TRI_med	GH	JRF	Elbow JRF
	48	22	383	22	166	4	209	1	52	224	172	108	234	886	729	
Muscle removed	Median muscle force post muscle removal (% of full-strength force values)															
Anc	100	<b>102</b>	<b>104</b>	100	<b>101</b>	<b>106</b>	<b>102</b>	100	<b>100</b>	<b>102</b>	<b>101</b>	<b>109</b>	<b>104</b>	<b>102</b>	100	
BIC_long	<b>220</b>	161	<b>191</b>	<b>35</b>	<b>409</b>	100	<b>231</b>	<b>0</b>	<b>85</b>	<b>207</b>	81	<b>16</b>	93	<b>208</b>	<b>58</b>	
BIC_shor	100	<b>100</b>	100	<b>107</b>	100	100	100	<b>88</b>	100	100	100	100	100	100	100	100
Bra	100	101	100	<b>103</b>	<b>100</b>	<b>100</b>	100	<b>125</b>	100	100	100	100	100	100	100	100
BRD	100	100	100	<b>101</b>	100	100	100	100	100	100	100	100	100	100	100	100
CB	100	100	100	<b>100</b>	100	100	100	100	100	100	100	100	100	<b>100</b>	100	100
DeltA	<b>108</b>	<b>160</b>	<b>194</b>	79	111	<b>186</b>	<b>109</b>	63	<b>203</b>	<b>109</b>	<b>130</b>	107	<b>164</b>	116	<b>130</b>	
DeltM	101	<b>151</b>	<b>141</b>	<b>77</b>	<b>108</b>	<b>142</b>	<b>70</b>	<b>0</b>	<b>106</b>	<b>138</b>	106	<b>87</b>	<b>133</b>	<b>91</b>	107	
DeltP	73	100	100	85	100	100	<b>160</b>	<b>2950</b>	110	100	100	100	99	<b>105</b>	94	
IS	100	100	<b>100</b>	<b>173</b>	100	100	100	<b>163</b>	100	100	100	100	100	100	100	100
LD	<b>102</b>	100	100	105	100	100	<b>102</b>	100	100	100	100	<b>100</b>	100	100	100	103
LS	<b>108</b>	<b>121</b>	<b>103</b>	99	<b>100</b>	<b>108</b>	<b>101</b>	88	<b>107</b>	101	<b>95</b>	<b>111</b>	99	<b>103</b>	100	
Pmaj_clav	105	157	177	117	122	<b>125</b>	100	50	91	120	125	95	128	114	119	
Pmaj_thor	80	169	<b>203</b>	<b>237</b>	<b>188</b>	122	93	<b>0</b>	<b>73</b>	93	<b>150</b>	<b>240</b>	<b>151</b>	<b>148</b>	<b>190</b>	
Pmin	103	100	<b>100</b>	100	100	100	100	100	<b>98</b>	98	105	100	100	99	100	
PT_hum	<b>100</b>	<b>100</b>	<b>100</b>	100	<b>100</b>	<b>100</b>	100	100	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>	<b>100</b>
Rmaj		<b>121</b>	<b>100</b>	100	100	<b>100</b>	<b>100</b>	100	<b>100</b>	<b>111</b>	108	103	99	100	100	
Rmin	111		100	100	<b>100</b>	<b>103</b>	101	100	<b>103</b>	<b>105</b>	<b>100</b>	<b>103</b>	100	<b>101</b>	100	
SA	<b>37</b>	<b>79</b>		<b>6 × 10<sup>5</sup></b>	<b>8 × 10<sup>4</sup></b>	<b>2 × 10<sup>7</sup></b>	<b>7 × 10<sup>3</sup></b>	<b>6 × 10<sup>7</sup></b>	<b>2 × 10<sup>4</sup></b>	<b>149</b>	<b>300</b>	<b>6</b>	<b>3 × 10<sup>5</sup></b>	<b>10</b>	<b>3 × 10<sup>5</sup></b>	
SBS	<b>81</b>	100	103		100	100	102	<b>213</b>	105	100	100	91	100	100	100	
SS	105	<b>167</b>	<b>190</b>	<b>239</b>		139	94	75	95	119	<b>126</b>	<b>58</b>	<b>126</b>	92	<b>103</b>	
SUP_hum	<b>100</b>	<b>100</b>	<b>100</b>	100	<b>100</b>		100	100	100	<b>100</b>	<b>100</b>	100	<b>100</b>	<b>100</b>	<b>100</b>	100
Tmaj	78	<b>170</b>	<b>198</b>	96	104	<b>136</b>		<b>175</b>	<b>153</b>	<b>131</b>	86	<b>185</b>	100	110	<b>110</b>	
Tmin	100	100	100	101	100	100	100		100	100	100	100	100	100	100	
Trap_cla	<b>102</b>	<b>104</b>	100	101	101	<b>100</b>	<b>101</b>	88	<b>116</b>	<b>98</b>	<b>102</b>	99	<b>101</b>	100	100	
Trap_sca	<b>256</b>	<b>224</b>	104	108	<b>113</b>	<b>131</b>	101	38	<b>218</b>		97	<b>117</b>	<b>112</b>	<b>108</b>	<b>103</b>	
TRI_lat	<b>100</b>	100	<b>92</b>	99	<b>101</b>	<b>106</b>	102	100	<b>100</b>	99		<b>122</b>	<b>163</b>	101	<b>96</b>	
TRI_long	92	<b>102</b>	<b>124</b>	100	<b>96</b>	<b>100</b>	112	<b>100</b>	<b>109</b>	103	<b>110</b>		<b>124</b>	<b>93</b>	<b>81</b>	
TRI_med	103	105	101	<b>99</b>	<b>106</b>	<b>122</b>	102	88	100	<b>104</b>	<b>213</b>	<b>140</b>		<b>107</b>	95	

A similar effect was seen by Komura et al. [14] during gait, who found that deactivating the gluteus maximus (crosses the hip) affected the forces at the knee and that the forces at the ankle increased when the rectus femoris (crosses the knee and hip) was removed. Komura also discovered that removal of the gluteus maximus affected the knee more than the hip. This can also be seen in

the STS data of this study where removal of BIC\_long increased the glenohumeral JRF more than the resulting decrease in elbow JRF.

Removal of SA caused significant, large increases in the peak activity of multiple muscles crossing both the shoulder and elbow. The prime role of SA is to stabilise the scapula during elevation, upward rotation and protraction. The synergists for this include a



**Figure 4.** Incidence of failed optimisations for a weakened serratus anterior over time. Top line of images represents key points in the STS motion. An arrow indicates the point of peak hand force.

**Table 4**

Muscles which, when removed, caused other muscles to produce greater than a theoretical maximum muscle force. The middle deltoid consistently produced force over 100% for full strength and all removed muscles and has been omitted from this table.

Removed muscle	Remaining muscles which produced forces exceeding the calculated maximum muscle force
BIC_long	SS, Tmaj
DeltA	BIC_long
Pmaj_thor	BIC_long
SA	Anc, BIC_long, BIC_short, BRA, BRD, CB, DeltP, IS, Pmin, PT_hum, SBS, SS, SUP_hum, Tmaj, Tmin, Trap_clav, TRI_long
Tmaj	BIC_long

great deal of the upper limb musculature, including the trapezius, rhomboids, pectoralis minor and levator scapulae. All, barring the rhomboids, saw increased muscle activity under SA removal and many possess secondary functions across the shoulder and elbow.

The Serratus Anterior was the only muscle to have any noticeable effect on optimisation success (Fig. 4), despite the lack of any constraints on muscle force for synergistic intervention. This would suggest that during parts of STS, SA is the only muscle capable of sufficiently stabilising the scapula. At the start, when participants have their hands placed on the arm rests, the scapula is in an upwardly-rotated and elevated position. Evidently no other muscles can provide the stability here. However, it stands to reason (considering the unrealistically-large forces seen in other muscles under SA weakness) that introducing more realistic muscle force upper boundaries for other muscles may increase failure rates around the point of peak external hand force. This was evident by the findings in the accompanying supplementary data, which showed that many muscles required greater than

100% activation to facilitate a successful optimisation in lieu of SA action.

Weakness of SA is a prime prerequisite of scapula winging [41], which cases a decline in functional ability during ADLs and sports [42]. A review by Cools et al. [43] found that weakness, not complete removal, of SA is commonly associated with other pathologies of the upper limb. This shows the clinical relevance of the work presented here; the results of removing SA could inform muscle strength training for synergistic groups. The effects on JRF is particularly relevant considering known associations between osteoarthritis (OA) and MW, discussed in detail by Segal & Glass [44]. Their review found multiple publications supporting the idea that weakness of the quadriceps is concurrent with OA incidence in lower limbs. A similar review by Loureiro et al. [45] on hip OA found a number of studies reporting increased MW in the arthritic limb of patients relative to the contralateral limb. No such publications associating upper limb OA and MW could be found, however, the findings for the lower body would likely apply to all joints considering the atrophy that occurs in older patients during a recovery period and extended muscle rest.

There also appears to be a strong link between muscles classed as predominantly fast-twitch (higher percentage of Type 2 fibres) and increases in other muscles' activity when removed. Those muscles which were highlighted as being fast-twitch, with the exception of coracobrachialis, all incited large increases in other muscles' activity. The preferential degradation of this fibre with age could indicate that weakness of these muscles is physiologically relevant in determining STS difficulties in the elderly. Removal of these muscles significantly increased and/or decreased the peak GH and elbow JRF (Table 3), thus lending weight to a hypothesis that this induces discomfort in the upper limb during STS. Further work should investigate the relationship between a reduction in Type 2 muscle fibre and increased MW and joint pain.

The Serratus Anterior muscle is highlighted as having a large percentage of Type 2 muscle fibres in rats (other data are not available) and as a point of reference it was shown that rodents have higher proportions of Type 2 fibres in the vastus lateralis when compared to humans [46]. However, the similarities between ageing rat and human musculature [26] means this serves as a good estimation of SA composition. Evidently, a reduction in the size and number of Type 2 fibres with age would have a profound effect on SA. Considering the extreme effects of its theoretical removal, this offers one explanation into elderly STS difficulties [47]. Future work may look to set muscle force upper bounds and observe the effects on optimisation success, followed by working with scapula winging patients as an extreme case of SA weakness and tailoring rehabilitation to their specific needs.

Due to the lack of validated methods for providing accurate muscle force constraints, any results using such constraints would be subject to the accuracy of their calculation. However, for this study it meant there were no set strength limits, leading to muscle forces well above physiological relevance. Instead theoretical maximum muscle forces showed the extent to which individual muscles were overloaded when solving the loadsharing redundancy. Future work may develop methods by which individual muscle lines of action, or functional groups, can be given bespoke upper bounds for more relevant optimisation constraints. One possible route would be using dynamometry, although the number of movements required for a comprehensive strength profile may be impractical.

The aim of this study was to observe the effects of theoretical MW in the context of measured kinematics; it is therefore limited by the loadsharing optimisation of the MSK model used to calculate the original muscle and joint force data. The effect of changing the cost function on the ability to compensate for MW was not investigated, which may change the ability to produce successful solutions. Additionally, the kinematics were not altered in any way and therefore represents an extreme solution to the loadsharing problem. Kinematic changes may come as an alternative compensation, which could result in smaller muscle and JRF effects than those quantified here.

The study was limited to young subject data, as opposed to elderly kinematics which would be of interest considering the difficulties this age group have in standing from a seated position, even when using arm rest support. However, the overall aim of the study was to observe the effects of MW during 'ideal' movements, informing the potential long-term effects on STS ability. If elderly kinematics were observed, then compensations in participants may already be employed to adapt to existing weakness, giving instead the effects of further weakness. This study highlighted the effects on young STS performance; however, these findings may warrant further investigation into differences in elderly STS kinematics under varying levels of weakness/ability.

The current study investigated the effects of singular MW, presenting information on the localised effects of removing individual muscles on neighbouring muscles and joints. Future iterations should look at quantifying groups, or combinations of muscles which naturally weaken with age, giving a more informed method for observing the theoretical effects of general MW.

As the instrumented chair contained load cells for the measurement of hand loads in only two directions, medio-lateral (ML) components could not be measured. As the participants were restricted to movements predominantly in the sagittal plane, these ML forces are expected to be low though no published data are available to confirm this. Elderly groups may utilise support in the ML direction, particularly when an individual is seated in a chair with arm rests at a significant distance apart, for added stability and support. Future design iterations of the instrumented chair should include sensors for measurement of force in all directions.

For the current study, it is believed that the arm rests were sufficiently close together to minimise the requirement for ML support.

## 5. Conclusions

Kinematics of 27 healthy male subjects performing a STS activity were analysed using MSK modelling to simulate upper limb MW. Of the 29 muscles removed in this way, SA proved the most debilitating and vastly increased the activity of other compensatory muscles and upper body JRFs. Previous studies have identified the debilitating effects of SA fatigue and weakness; however, this is the first study to quantify its importance during STS, a key determinant of mobility and social inclusion. The high prevalence of Type 2 muscle fibres in SA suggests it could be vulnerable to age-related atrophy, thus hypothesising a potential cause behind STS difficulties in the elderly. Due to its role in maintaining scapula stability during elevated activities, motions requiring large shoulder movements become painful and almost impossible when SA is weakened. This may suggest routes for tailored rehabilitation, for instance targeted SA training as per published recommendations [48], with future work investigating ways to identify local MW. Other muscles with high proportions of fast-twitch muscle fibres were also shown to produce high levels of muscle force changes, further confirming the vulnerability of elderly functional ability to the natural degradation of Type 2 fibres.

The work in this study focused on the STS motion and has identified SA as a dominant muscle for this activity. The task was selected based on its importance for mobility. Such work could be used to advise clinical practice, whereby difficulties in activities could be linked to weakness of the aforementioned list of muscles.

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## Declaration of Competing Interest

The authors declare no conflict of interest with the content of this study.

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## Ethical Approval

Ethical approval was received through the Joint Research Compliance Office and Imperial College Research Ethics Committee.

## Supplementary materials

Supplementary material associated with this article can be found, in the online version, at doi:[10.1016/j.medengphy.2019.09.023](https://doi.org/10.1016/j.medengphy.2019.09.023).

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