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## **A biomechanical confirmation of the relationship between critical shoulder angle (CSA) and articular joint loading.**

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1 **A biomechanical confirmation of the relationship between critical shoulder angle (CSA)**  
2 **and articular joint loading**

3

4 **Abstract**

5 **Background:** The Critical Shoulder Angle (CSA) has been shown to be correlated with  
6 shoulder disease states. The biomechanical hypothesis to explain this correlation is that the  
7 CSA changes the shear and compressive forces on the shoulder. The objective of this study is  
8 to test this hypothesis by use of a validated computational shoulder model. Specifically, this  
9 study assesses the impact on glenohumeral biomechanics of modifying the CSA.

10 **Methods:** An inverse dynamics three-dimensional musculoskeletal model of the shoulder was  
11 used to quantify muscle forces and glenohumeral joint forces. The CSA was changed by  
12 altering the attachment point of the middle deltoid into a normal CSA (33°), a reduced CSA  
13 of 28°, and an increased CSA of 38°. Subject-specific kinematics of slow and fast speed  
14 abduction in the scapular plane, and slow and fast forward flexion measured by a 3D motion  
15 capture system were used to quantify joint reaction shear and compressive forces.

16 **Results:** Increasing the CSA results in increased superior-inferior forces (shearing forces;  
17 integrated over the range of motion;  $p < 0.05$ ). Reducing CSA results in increased latero-  
18 medial (compressive) forces for both the maximum and integrated sum of the forces over the  
19 whole motion;  $p < 0.01$ ).

20 **Discussion/Conclusion:** Changes in the CSA modify glenohumeral joint biomechanics with  
21 increasing CSA producing higher shear forces that would contribute to rotator cuff overuse,  
22 whereas reducing the CSA results in higher compressive forces which contribute to joint  
23 wear.

24 **Level of evidence:** Basic Science Study; Computer Modeling

25

26 **Keywords:** Critical shoulder angle (CSA), glenohumeral joint, joint wear, rotator cuff,  
27 computational shoulder model.

28

29 The shape of the scapula and especially of the acromion has historically been considered as a  
30 potential etiology for shoulder pathologies including those of the gleno-humeral joint (GHJ).

31 Codman<sup>10</sup>, Armstrong<sup>3</sup> and Neer<sup>26</sup> described the association of specific acromion shapes  
32 with degenerative rotator cuff tear (RCT), leading to the well-known impingement syndrome  
33 and the extrinsic mechanical conflict theory. Since then, a broader understanding of

34 degenerative RCT physiopathology<sup>35</sup> (ageing of the tendon, modification of local  
35 vascularization, genetic disposition) has challenged this theory, thus questioning the absolute  
36 role of the acromion shape in this process. Research over the past 15 years has continued to

37 focus on the shape of the acromion in the coronal plane, including proposed radiological  
38 parameters to describe the lateral extension of the acromion<sup>5,28</sup>. A recent and widely cited

39 study from Moor et al proposed the concept of the Critical Shoulder Angle (CSA)<sup>24</sup>, a  
40 measure which takes into account the tilt of the glenoid (inclination) and the lateral extension  
41 of the acromion. In a population of 279 patients, they found that the mean CSA was

42 significantly different between a disease-free shoulder group (33.1°), an RCT group (38.0°),  
43 and a primary osteoarthritis group (OA, 28.1°). Numerous clinical observational studies have  
44 confirmed these findings<sup>4,5,8,11-13,15,16,23,25,38,43</sup>.

45

46 The link between the clinical observations and the CSA are hypothesized to be biomechanical  
47 in nature<sup>2,27,22,14,40,42</sup>, by changing the magnitude and direction of the deltoid force. The

48 hypothesis of this study is that an increased CSA would result in higher shearing forces that  
49 would be associated with RCT and a decreased CSA would result in higher compressive  
50 forces, associated with primary osteoarthritis.

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The objective of this study is to test the hypothesis by assessing the impact on GHJ biomechanics of modifying the CSA by use of a validated computational shoulder model.

**Material and Methods**

The United Kingdom National Shoulder Model (UKNSM)

An inverse dynamics three-dimensional musculoskeletal model of the upper limb<sup>9</sup> was used to quantify muscle and glenohumeral joint forces. This model quantifies forces in 87 muscle elements, three ligaments and joint reaction forces of the sternoclavicular, acromioclavicular, scapulothoracic, glenohumeral and elbow joint (Figure 1). The analysis begins by solving the determinate inverse dynamics intersegmental moments based on measured kinematics. Then an optimization algorithm is used to minimize the sum of muscle stresses squared to solve the muscle load-sharing redundancy. The model has been validated for glenohumeral joint force measures through comparison to instrumented anatomical shoulder replacement measurements<sup>7,44</sup> for a driving task<sup>29</sup> and other tasks of daily living<sup>19</sup> and for muscle forces through comparison with electromyography<sup>18</sup>. The model is customized to each subject by scaling segment lengths and body segment parameters.<sup>17</sup> In brief, clavicle and scapula segments were homogeneously scaled based on relative segment lengths between model and subject. An ellipsoid represented the scapulo thoracic gliding-plane; this was non-homogeneously scaled using an optimization procedure to minimize the difference between digitized anatomical landmarks and the final ellipsoid. A partially closed chain method is used to optimize scapula and clavicle kinematics, in which the least squares difference to the measured scapular and clavicle kinematics is minimized and the scapula medial border is constrained to not penetrate the thorax wall (represented by an elipsoid).<sup>32</sup>

76 Tests conditions

77 The scapula within the UKNSM was imported into computer aided engineering software to  
78 allow the effect of medializing and lateralizing the acromion to be modelled, thus changing  
79 the CSA and therefore the three-dimensional position of the attachment point of the middle  
80 deltoid. This was implemented in the software by sketching new points and lines parallel to  
81 the plane of measurement of the CSA and coincident with the original middle deltoid  
82 attachment point. The changes were made to create 3 different CSAs: a normal CSA ( $33^\circ$ ), a  
83 reduced CSA of  $28^\circ$ , and an increased CSA of  $38^\circ$  that reflect the changes previously reported  
84 as clinically significant <sup>24</sup>.

85

86 To test the effect of different subject-specific kinematics, a previously obtained dataset of  
87 kinematics was used <sup>20</sup>. These kinematics were measured using a nine camera optical motion  
88 tracking system with a set of twenty-one retro-reflective markers used to track the thorax,  
89 clavicle, humerus, and forearm segments <sup>36,37,45</sup>. The kinematics of six healthy male subjects  
90 were used (age 25 years  $\pm$  2 years) who each performed four motions with a 2 kg hand load to  
91 provide a resistance to motion without fatiguing the subjects. These motions were: slow and  
92 fast speed abduction (in the scapular plane), and slow and fast forward flexion <sup>34</sup>. Each subject  
93 performed three trials per motion. These datasets were passed through the UKNSM after  
94 appropriate subject scaling <sup>32</sup>.

95 Outcomes measures

96 All output measures were normalized to humero-thoracic elevation and all forces were  
97 normalized to the subject's body weight and integrated over the range of motion from  $30^\circ$  to  
98  $120^\circ$  of humero-thoracic elevation. The following parameters were analyzed to test the effects  
99 of CSA changes on GHJ biomechanics:

100 - the magnitude (maximum and integrated over the whole motion) of the GHJ shearing forces  
101 (SF), representing the superior-inferior (SI) forces and the antero-posterior (AP) forces acting  
102 in the sagittal plane that are exerted on the joint during motion.

103 - the magnitude (maximum and integrated over the whole motion) of the GHJ compressive  
104 forces (CF), representing the forces acting in the frontal plane (latero-medial) that are exerted  
105 on the joint during motion.

106

### 107 Statistical Analysis

108 Repeated measures two-way ANOVA (RM 2-Way ANOVA), with Bonferroni post-hoc  
109 correction was conducted on the results using SPSS software (2014; IBM Corp., Armonk,  
110 NY, USA). The assumptions of the RM 2-Way ANOVA method were tested for all measures.  
111 The independent variables are CSA (normal, increased, reduced) and Motion (fast and slow,  
112 forward flexion and scapular plane abduction); the dependent variables are the joint forces  
113 (integrated AP shear, max AP shear, integrated SI shear, max SI shear, integrated CF, max  
114 CF). Mauchly's test indicated that the assumption of sphericity had been violated for the  
115 main effects of CSA and Motion for several measures, therefore degrees of freedom were  
116 corrected using Greenhouse-Geisser estimates of sphericity ( $\epsilon$ ).

117

### 118 Results

119 Statistically significant differences were found for the integrated SI shear force, the integrated  
120 CF and the maximum CF (Table 1). These statistical results are presented in Figure 2. In  
121 summary, an increased CSA angle resulted in 111% higher integrated SI shear forces  
122 ( $p < 0.05$ ) when compared to normal. For the compressive forces, a reduced CSA angle  
123 resulted in a significant increase of the CF (integrated: 2.8% increase,  $p < 0.01$ ; maximum:  
124 2.8% increase,  $p < 0.05$ ) compared to normal CSA and also when compared to increased CSA

125 (integrated: 5.5% increase,  $p < 0.01$ ; maximum 5.3% increase,  $p < 0.05$ ). There were no main  
126 effects found for Motion, nor an interaction effect of Motion and CSA.

127 The mean maximum SI forces were 13% BW, 12% BW and 21% BW for reduced, normal  
128 and increased CSA, respectively. The maximum CF were 132% BW, 136% BW and 129%  
129 BW for reduced, normal and increased CSA, respectively.

130

### 131 **Discussion**

132 This is the first study to assess the effects of changes in CSA on GHJ biomechanics through a  
133 computational shoulder model in order to test the mechanically-based hypothesis. The results  
134 confirm the initial hypothesis based on literature<sup>41</sup>, that is to say increasing the CSA results in  
135 increased superior-inferior forces (shearing forces), whereas decreasing the CSA results in  
136 increased latero-medial (compressive force) forces (Figure 3).

137

138 Our results are also consistent with those of two cadaver studies previously published about  
139 the subject. Even if the model and analysis method are different, Vielhofer and Gerber found  
140 that with an increased CSA, the ratio of glenohumeral joint shear to joint compression forces  
141 is increased (peak difference of 23% at 50° of thoraco-humeral abduction compared to a  
142 normal CSA)<sup>40</sup>, requiring substantially increased compensatory supraspinatus loads  
143 (increased by 13-33% between 33° and 37° of elevation compared to a normal CSA)<sup>14</sup>.

144

145 These cadaver studies combined with the study presented here provide simple mechanical  
146 explanations for the results:

- 147 • an increased CSA would result in a lateralized proximal deltoid insertion.  
148 Consequently, the vector force of the deltoid during abduction would be more vertical,  
149 requiring a greater horizontal force from the cuff to stabilize the joint (by



150 counterbalancing the shearing forces). Ultimately, this increase in muscle use could  
151 potentially, in combination with other factors, lead to a degenerative RCT in the long-  
152 term; and

- 153 • a reduced CSA would result in a medialized proximal deltoid insertion. Consequently,  
154 the vector force of the deltoid during abduction would be more oblique (resultant  
155 internal translation), which in addition to the vector force of the cuff (mostly  
156 horizontal), could increase the load (compressive forces) on the GHJ. Finally, this  
157 could lead to OA in the long-term.

158

159 It is known that small elevated mechanical loading is associated with the instigation and  
160 progression of osteoarthritis<sup>1,30</sup>, suggesting that, although small in percentage terms, the  
161 statistical differences in key mechanical variables found here may also be clinical significant.

162

163 Computational simulation models are frequently used to analyze human joint biomechanics  
164 and have been validated in studies that quantified articular loading in activities of daily living,  
165 and the biomechanical consequences of a pathology, surgery, and arthroplasty<sup>9,20,31,39</sup>.

166 Concerning shoulder modelling, the UKNSM (formerly Newcastle Shoulder Model)<sup>9</sup> is one  
167 the oldest validated inverse dynamics-based models<sup>33</sup>. Body segment parameters are based on  
168 a large group of young living patients<sup>21</sup>. There are limitations with such models and so,  
169 although, the model has been validated by comparison to instrumented joint replacements and  
170 electromyography, such computer simulations remain “models” and are, therefore, a surrogate  
171 of direct in-vivo biomechanical measurements.

172

173 Apart from technical considerations of the computational model, this study has some other  
174 limitations. Tests were only performed in simplified motions of pure scapular abduction or

175 flexion, whereas most daily activities have ranges of motion that combine abduction and  
176 flexion with obligate rotations<sup>20</sup>. Moreover, the model is based on a range of subjects with  
177 normal anatomy, and this study varied the CSA in isolation, whereas there might be other  
178 changes present with a change in CSA angle, such as glenoid version or muscle stiffness that  
179 may influence the model outputs. Additionally, a change in anatomy might also result in a  
180 further change in kinematics, although this was mitigated in this study by using a set of data  
181 from six healthy subjects, rather than just from one subject. Finally, we could have studied  
182 the different components of the CSA separately but even if the lateral acromial roof extension  
183 has a greater influence in pathogenesis of degenerative RCT and concentric OA than acromial  
184 height or glenoid inclination, the CSA remain the best factor to predict these pathologies<sup>6</sup>.

185

## 186 **Conclusion**

187 Through a validated computational shoulder model, combined with in vivo motion analysis  
188 experiments, this study demonstrates that changes in the CSA modify GHJ biomechanics.

189 Increasing the CSA results in higher shear forces, requiring increased rotator cuff use to  
190 neutralize the shear which is potentially damaging in the long term. Decreasing the CSA  
191 results in a higher joint compressive force which leads to increased joint wear.

192 These findings support previous clinical observational and biomechanical studies that  
193 alterations in CSA may have a role in common shoulder pathologies such RCT or OA.

194 Consequently, surgical restoration to a “normal” CSA is recommended when treating patients  
195 with such pathologies, for example, lateral acromioplasty after rotator cuff repair or ensuring  
196 control of glenoid inclination when conducting arthroplasty surgery.

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340

341 **Figure 1. Illustration of the UKSNM.**

342

343 **Figure 2.** Statistically significant changes in joint forces due to changed CSA. These  
344 conditions that showed a main effect in the ANOVA analysis. The error bars indicate standard  
345 deviation.

346

347 **Figure 3.** Vector plot of the maximum joint reaction forces in the medio-lateral and superior-  
348 inferior direction for the three CSA angles. The integrated shear force in the superior-inferior  
349 direction (SI) was significantly larger for increased CSA compared to normal CSA. The  
350 reduced CSA showed significantly higher compressive forces (CF) compared to normal CSA  
351 and increased CSA.

352

353 **Table 1.** Repeated measures ANOVA. Shown are the main effects for the separate force  
354 directions (AP, SI, CF), and the pairwise comparisons for the force directions that  
355 significantly differed between CSA conditions. Nor = Normal CSA (33°), Red = Reduced  
356 CSA (28°), Inc = Increased CSA (38°).

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358



**Table 1.** Repeated measures ANOVA. Shown are the main effects for the separate force directions (AP, SI, CF), and the pairwise comparisons for the force directions that significantly differed between CSA conditions. Nor = Normal CSA (33°), Red = Reduced CSA (28°), Inc = Increased CSA (38°).

| Variable            | Mauchly's sphericity                        | Main effect      |                          | Pairwise CSA |           |          |
|---------------------|---|------------------|--------------------------|--------------|-----------|----------|
|                     |   |                  |                          | Nor-Red      | Nor- Inc  | Red- Inc |
| integrated AP shear | $\chi^2(2)=9.5$ ,<br>p<.01, $\epsilon=.52$  | p = 0.598        |                          |              |           |          |
| max AP shear        | $\chi^2(2)=11.6$ ,<br>p<.01, $\epsilon=.51$ | p = 0.595        |                          |              |           |          |
| integrated SI shear |   | <b>p = 0.026</b> | <b>F(2,10) = 5.34</b>    |              | p = 0.044 |          |
| max SI shear        | $\chi^2(2)=[]$ ,<br>p<.05, $\epsilon=.55$   | p = 0.068        |                          |              |           |          |
| integrated CF       | $\chi^2(2)=6.23$ ,<br>p<.05, $\epsilon=.56$ | <b>p = 0.003</b> | <b>F(1.1, 5.6)=24.55</b> | p = 0.001    |           | p = 0.01 |
| max CF              | $\chi^2(2)=6.84$ ,<br>p<.05, $\epsilon=.55$ | <b>p = 0.009</b> | <b>F (2,10) =14.97</b>   | p = 0.008    |           | p = 0.03 |





