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Conflict of interest: none.

Funding: This study was funded in part by Swiss Orthopaedics.

Institutional review board approval was not required for this basic science study.

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Abstract

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

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

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

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

Level of evidence: Basic Science Study; Computer Modeling

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

The shape of the scapula and especially of the acromion has historically been considered as a potential etiology for shoulder pathologies including those of the gleno-humeral joint (GHJ). Codman [10], Armstrong [3] and Neer [26] described the association of specific acromion shapes with degenerative rotator cuff tear (RCT), leading to the well-known impingement syndrome and the extrinsic mechanical conflict theory. Since then, a broader understanding of degenerative RCT physiopathology [35] (ageing of the tendon, modification of local vascularization, genetic disposition) has challenged this theory, thus questioning the absolute role of the acromion shape in this process. Research over the past 15 years has continued to focus on the shape of the acromion in the coronal plane, including proposed radiological parameters to describe the lateral extension of the acromion [5,28]. A recent and widely cited study from Moor et al proposed the concept of the Critical Shoulder Angle (CSA) [24], a measure which takes into account the tilt of the glenoid (inclination) and the lateral extension of the acromion. In a population of 279 patients, they found that the mean CSA was significantly different between a disease-free shoulder group (33.1°), an RCT group (38.0°), and a primary osteoarthritis group (OA, 28.1°). Numerous clinical observational studies have confirmed these findings [4,5,8,11–13,15,16,23,25,38,43].

The link between the clinical observations and the CSA are hypothesized to be biomechanical in nature [2,27,22,14,40,42], by changing the magnitude and direction of the deltoid force. The hypothesis of this study is that an increased CSA would result in higher shearing forces that would be associated with RCT and a decreased CSA would result in higher compressive forces, associated with primary osteoarthritis.

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 [9] 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 [7,44] for a driving task [29] and other tasks of daily living [19] and for muscle forces through comparison with electromyography [18]. The model is customized to each subject by scaling segment lengths and body segment parameters.[17] 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). [32]

Tests conditions

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

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

Outcomes measures

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

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

Statistical Analysis

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

Results

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

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

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

Discussion

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

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

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

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

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

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

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

Computational simulation models are frequently used to analyze human joint biomechanics and have been validated in studies that quantified articular loading in activities of daily living, and the biomechanical consequences of a pathology, surgery, and arthroplasty [9,20,31,39]. Concerning shoulder modelling, the UKNSM (formerly Newcastle Shoulder Model) [9] is one the oldest validated inverse dynamics-based models [33]. Body segment parameters are based on a large group of young living patients [21]. There are limitations with such models and so, although, the model has been validated by comparison to instrumented joint replacements and electromyography, such computer simulations remain “models” and are, therefore, a surrogate of direct in-vivo biomechanical measurements.

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

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

Conclusion

Through a validated computational shoulder model, combined with in vivo motion analysis experiments, this study demonstrates that changes in the CSA modify GHJ biomechanics. Increasing the CSA results in higher shear forces, requiring increased rotator cuff use to neutralize the shear which is potentially damaging in the long term. Decreasing the CSA results in a higher joint compressive force which leads to increased joint wear.

These findings support previous clinical observational and biomechanical studies that alterations in CSA may have a role in common shoulder pathologies such RCT or OA. Consequently, surgical restoration to a "normal" CSA is recommended when treating patients with such pathologies, for example, lateral acromioplasty after rotator cuff repair or ensuring control of glenoid inclination when conducting arthroplasty surgery.

References

1. Apreleva M, Parsons IM, Warner JJ, Fu FH, Woo SL. Experimental investigation of reaction forces at the glenohumeral joint during active abduction. J. Shoulder Elbow Surg. 2000 Oct;9(5):409–417.

2. Apreleva M, Parsons IM, Warner JJP, Fu FH, Woo SL-Y. Experimental investigation of reaction forces at the glenohumeral joint during active abduction. J. Shoulder Elbow Surg. 2000 Sep 1;9(5):409–417.

3. Armstrong JR. Excision of the acromion in treatment of the supraspinatus syndrome; report of 95 excisions. J. Bone Joint Surg. Br. 1949 Aug;31B(3):436–442.

4. Balke M, Liem D, Greshake O, Hoeher J, Bouillon B, Banerjee M. Differences in acromial morphology of shoulders in patients with degenerative and traumatic supraspinatus tendon tears. Knee Surg. Sports Traumatol. Arthrosc. Off. J. ESSKA. 2016 Jul;24(7):2200–2205. doi:10.1007/s00167-014-3499-y

5. Banas MP, Miller RJ, Totterman S. Relationship between the lateral acromion angle and rotator cuff disease. J. Shoulder Elbow Surg. 1995 Dec;4(6):454–461.

6. Beeler S, Hasler A, Götschi T, Meyer DC, Gerber C. Critical shoulder angle: Acromial coverage is more relevant than glenoid inclination. J. Orthop. Res. 2019 Jan 1;37(1):205–210. doi:10.1002/jor.24053

7. Bergmann G, Graichen F, Bender A, Kääb M, Rohlmann A, Westerhoff P. In vivo glenohumeral contact forces--measurements in the first patient 7 months postoperatively. J. Biomech. 2007;40(10):2139–2149. doi:10.1016/j.jbiomech.2006.10.037

8. Blonna D, Giani A, Bellato E, Mattei L, Caló M, Rossi R, et al. Predominance of the critical shoulder angle in the pathogenesis of degenerative diseases of the shoulder. J. Shoulder Elbow Surg. 2016 Aug;25(8):1328–1336. doi:10.1016/j.jse.2015.11.059

9. Charlton IW, Johnson GR. A model for the prediction of the forces at the glenohumeral joint. Proc. Inst. Mech. Eng. [H]. 2006 Nov;220(8):801–812. doi:10.1243/09544119JEIM147

10. Codman EA. The shoulder; rupture of the supraspinatus tendon and other lesions in or about the subacromial bursa. Boston, Mass: T. Todd Company, printers; 1934.

11. Daggett M, Werner B, Collin P, Gauci M-O, Chaoui J, Walch G. Correlation between glenoid inclination and critical shoulder angle: a radiographic and computed tomography

study. J. Shoulder Elbow Surg. 2015 Dec;24(12):1948–1953. doi:10.1016/j.jse.2015.07.013

12. Garcia GH, Liu JN, Degen RM, Johnson CC, Wong AC, Dines DM, et al. Higher critical shoulder angle increases the risk of retear after rotator cuff repair. J. Shoulder Elbow Surg. 2017;26(2):241–245. doi:10.1016/j.jse.2016.07.009

13. Gerber C, Catanzaro S, Betz M, Ernstbrunner L. Arthroscopic Correction of the Critical Shoulder Angle Through Lateral Acromioplasty: A Safe Adjunct to Rotator Cuff Repair. Arthrosc. J. Arthrosc. Relat. Surg. Off. Publ. Arthrosc. Assoc. N. Am. Int. Arthrosc. Assoc. 2018 Mar;34(3):771–780. doi:10.1016/j.arthro.2017.08.255

14. Gerber C, Snedeker JG, Baumgartner D, Viehöfer AF. Supraspinatus tendon load during abduction is dependent on the size of the critical shoulder angle: A biomechanical analysis. J. Orthop. Res. Off. Publ. Orthop. Res. Soc. 2014 Jul;32(7):952–957. doi:10.1002/jor.22621

15. Gomide LC, Carmo TC do, Bergo GHM, Oliveira GA, Macedo IS. Relationship between the critical shoulder angle and the development of rotator cuff lesions: a retrospective epidemiological study. Rev. Bras. Ortop. 2017 Jul;52(4):423–427. doi:10.1016/j.rboe.2017.06.002

16. Heuberer PR, Plachel F, Willinger L, Moroder P, Laky B, Pauzenberger L, et al. Critical shoulder angle combined with age predict five shoulder pathologies: a retrospective analysis of 1000 cases. BMC Musculoskelet. Disord. 2017 Jun 15;18(1):259. doi:10.1186/s12891-017-1559-4

17. Johnson GR, Spalding D, Nowitzke A, Bogduk N. Modelling the muscles of the scapula morphometric and coordinate data and functional implications. J. Biomech. 1996 Aug 1;29(8):1039–1051.

18. Klemt C, Nolte D, Ding Z, Rane L, Quest RA, Finnegan ME, et al. Anthropometric Scaling of Anatomical Datasets for Subject-Specific Musculoskeletal Modelling of the Shoulder. Ann. Biomed. Eng. 2019 Apr;47(4):924–936. doi:10.1007/s10439-019-02207-2

19. Klemt C, Prinold JA, Morgans S, Smith SHL, Nolte D, Reilly P, et al. Analysis of shoulder compressive and shear forces during functional activities of daily life. Clin. Biomech. Bristol Avon. 2018;54:34–41. doi:10.1016/j.clinbiomech.2018.03.006

20. Klemt C, Prinold JA, Morgans S, Smith SHL, Nolte D, Reilly P, et al. Analysis of shoulder compressive and shear forces during functional activities of daily life. Clin. Biomech. Bristol Avon. 2018 May;54:34–41. doi:10.1016/j.clinbiomech.2018.03.006

21. de Leva P. Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. J. Biomech. 1996 Sep;29(9):1223–1230.

22. Ludewig PM, Phadke V, Braman JP, Hassett DR, Cieminski CJ, LaPrade RF. Motion of the shoulder complex during multiplanar humeral elevation. J. Bone Joint Surg. Am. 2009 Feb;91(2):378–389. doi:10.2106/JBJS.G.01483

23. Mantell MT, Nelson R, Lowe JT, Endrizzi DP, Jawa A. Critical shoulder angle is associated with full-thickness rotator cuff tears in patients with glenohumeral osteoarthritis. J. Shoulder Elbow Surg. 2017 Dec;26(12):e376–e381. doi:10.1016/j.jse.2017.05.020

24. Moor BK, Bouaicha S, Rothenfluh DA, Sukthankar A, Gerber C. Is there an association between the individual anatomy of the scapula and the development of rotator cuff tears or osteoarthritis of the glenohumeral joint?: A radiological study of the critical shoulder angle. Bone Jt. J. 2013 Jul;95-B(7):935–941. doi:10.1302/0301-620X.95B7.31028

25. Moor BK, Wieser K, Slankamenac K, Gerber C, Bouaicha S. Relationship of individual scapular anatomy and degenerative rotator cuff tears. J. Shoulder Elbow Surg. 2014 Apr;23(4):536–541. doi:10.1016/j.jse.2013.11.008

26. Neer CS. Anterior acromioplasty for the chronic impingement syndrome in the shoulder: a preliminary report. J. Bone Joint Surg. Am. 1972 Jan;54(1):41–50.

27. Nikooyan AA, Veeger HEJ, Westerhoff P, Graichen F, Bergmann G, van der Helm FCT. Validation of the Delft Shoulder and Elbow Model using in-vivo glenohumeral joint contact forces. J. Biomech. 2010 Nov 16;43(15):3007–3014. doi:10.1016/j.jbiomech.2010.06.015

28. Nyffeler RW, Werner CML, Sukthankar A, Schmid MR, Gerber C. Association of a large lateral extension of the acromion with rotator cuff tears. J. Bone Joint Surg. Am. 2006 Apr;88(4):800–805. doi:10.2106/JBJS.D.03042

29. Pandis P, Prinold JAI, Bull AMJ. Shoulder muscle forces during driving: Sudden steering can load the rotator cuff beyond its repair limit. Clin. Biomech. Bristol Avon. 2015 Oct;30(8):839–846. doi:10.1016/j.clinbiomech.2015.06.004

30. Parsons IM, Apreleva M, Fu FH, Woo SLY. The effect of rotator cuff tears on reaction forces at the glenohumeral joint. J. Orthop. Res. Off. Publ. Orthop. Res. Soc. 2002 May;20(3):439–446. doi:10.1016/S0736-0266(01)00137-1

31. Praagman M, Stokdijk M, Veeger HE, Visser B. Predicting mechanical load of the glenohumeral joint, using net joint moments. Clin. Biomech. Bristol Avon. 2000 Jun;15(5):315–321.

32. Prinold JAI, Bull AMJ. Scaling and kinematics optimisation of the scapula and thorax in upper limb musculoskeletal models. J. Biomech. 2014 Aug 22;47(11):2813–2819. doi:10.1016/j.jbiomech.2014.05.015

33. Prinold JAI, Masjedi M, Johnson GR, Bull AMJ. Musculoskeletal shoulder models: a technical review and proposals for research foci. Proc. Inst. Mech. Eng. [H]. 2013 Oct;227(10):1041–1057. doi:10.1177/0954411913492303

34. Prinold JAI, Villette CC, Bull AMJ. The influence of extreme speeds on scapula kinematics and the importance of controlling the plane of elevation. Clin. Biomech. Bristol Avon. 2013 Dec;28(9–10):973–980. doi:10.1016/j.clinbiomech.2013.10.008

35. Rees JL. The pathogenesis and surgical treatment of tears of the rotator cuff. J. Bone Joint Surg. Br. 2008 Jul;90(7):827–832. doi:10.1302/0301-620X.90B7.19874

36. Shaheen AF, Alexander CM, Bull AMJ. Effects of attachment position and shoulder orientation during calibration on the accuracy of the acromial tracker. J. Biomech. 2011 Apr 29;44(7):1410–1413. doi:10.1016/j.jbiomech.2011.01.013

37. Shaheen AF, Alexander CM, Bull AMJ. Tracking the scapula using the scapula locator with and without feedback from pressure-sensors: A comparative study. J. Biomech. 2011 May 17;44(8):1633–1636. doi:10.1016/j.jbiomech.2011.02.139

38. Spiegl UJ, Horan MP, Smith SW, Ho CP, Millett PJ. The critical shoulder angle is associated with rotator cuff tears and shoulder osteoarthritis and is better assessed with radiographs over MRI. Knee Surg. Sports Traumatol. Arthrosc. Off. J. ESSKA. 2016 Jul;24(7):2244–2251. doi:10.1007/s00167-015-3587-7

39. Van Drongelen S, Van der Woude LH, Janssen TW, Angenot EL, Chadwick EK, Veeger DH. Mechanical load on the upper extremity during wheelchair activities. Arch. Phys. Med. Rehabil. 2005 Jun;86(6):1214–1220. doi:10.1016/j.apmr.2004.09.023

40. Viehöfer AF, Gerber C, Favre P, Bachmann E, Snedeker JG. A larger critical shoulder angle requires more rotator cuff activity to preserve joint stability. J. Orthop. Res. Off. Publ. Orthop. Res. Soc. 2016;34(6):961–968. doi:10.1002/jor.23104

41. Viehöfer AF, Gerber C, Favre P, Bachmann E, Snedeker JG. A larger critical shoulder angle requires more rotator cuff activity to preserve joint stability. J. Orthop. Res. 2016;34(6):961–968.

42. Viehöfer AF, Snedeker JG, Baumgartner D, Gerber C. Glenohumeral joint reaction forces increase with critical shoulder angles representative of osteoarthritis-A biomechanical analysis. J. Orthop. Res. Off. Publ. Orthop. Res. Soc. 2016;34(6):1047–1052. doi:10.1002/jor.23122

43. Watanabe A, Ono Q, Nishigami T, Hirooka T, Machida H. Differences in Risk Factors for Rotator Cuff Tears between Elderly Patients and Young Patients. Acta Med. Okayama. 2018 Feb;72(1):67–72. doi:10.18926/AMO/55665

44. Westerhoff P, Graichen F, Bender A, Halder A, Beier A, Rohlmann A, et al. In vivo measurement of shoulder joint loads during activities of daily living. J. Biomech. 2009 Aug 25;42(12):1840–1849. doi:10.1016/j.jbiomech.2009.05.035

45. Wu G, van der Helm FCT, Veeger HEJD, Makhsous M, Van Roy P, Anglin C, et al. 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. 2005 May;38(5):981–992. https://doi.org/10.1016/j.jbiomech.2004.05.042

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

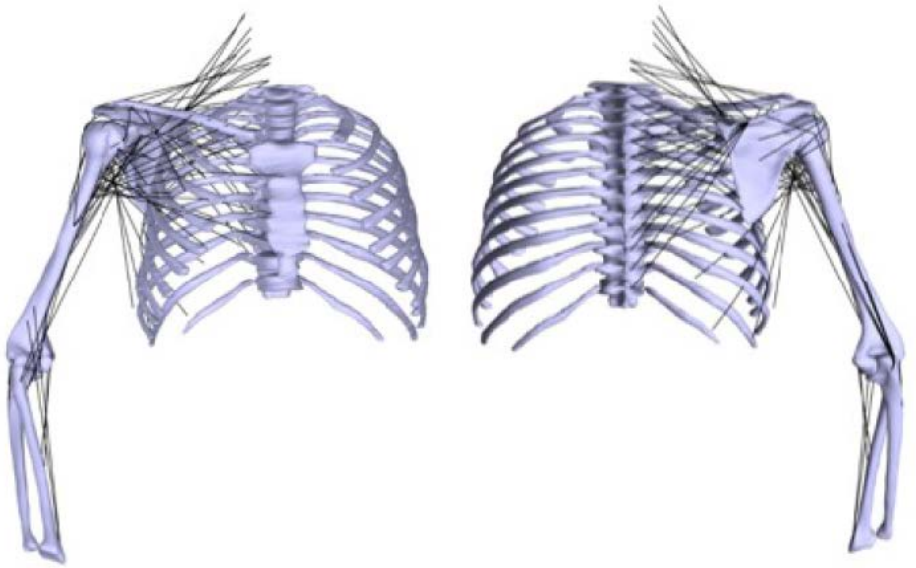
Figure 2. Statistically significant changes in joint forces due to changed CSA. These conditions that showed a main effect in the ANOVA analysis. The error bars indicate standard deviation.

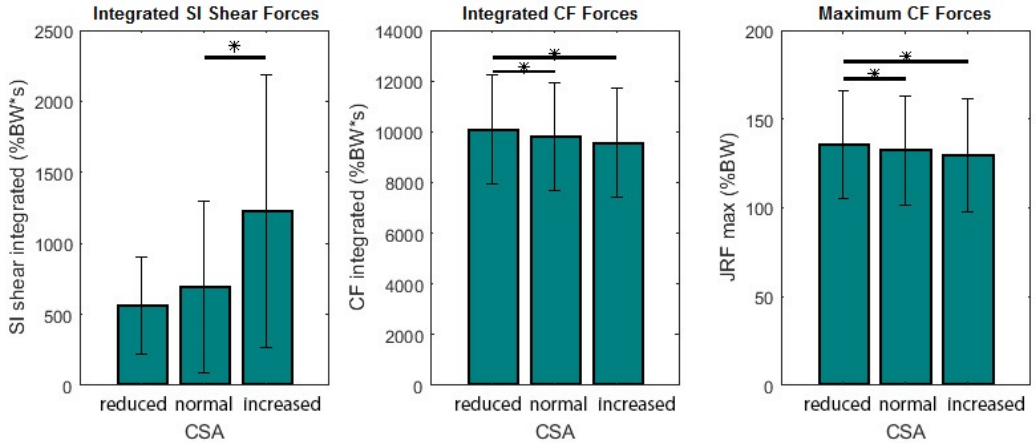
Figure 3. Vector plot of the maximum joint reaction forces in the medio-lateral and superior-inferior direction for the three CSA angles. The integrated shear force in the superior-inferior direction (SI) was significantly larger for increased CSA compared to normal CSA. The reduced CSA showed significantly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

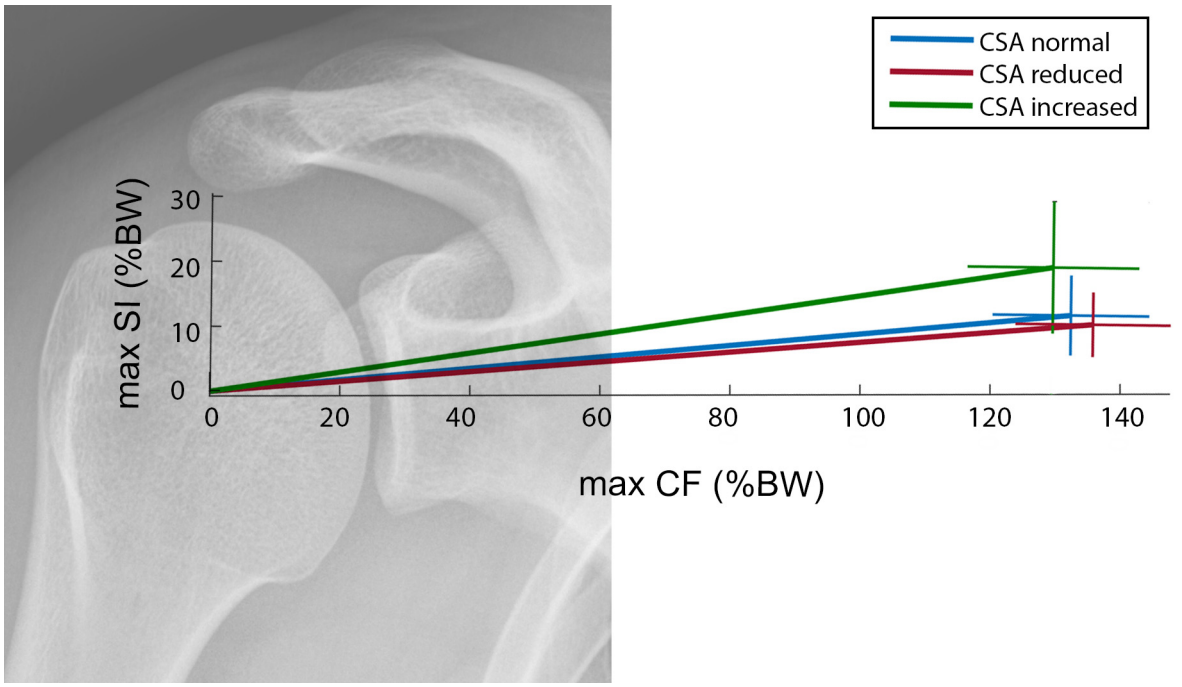
Table 1. Repeated measures ANOVA. Shown are the main effects for the separate force directions (AP, SI, CF), and the pairwise comparisons for the force directions that significantly differed between CSA conditions. Nor = Normal CSA (33°), Red = Reduced CSA (28°), Inc = Increased CSA (38°).

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	$\chi^2(2)=9.5$, $p<.01$, $\varepsilon=.52$	p = 0.598				
max AP shear	$\chi^2(2)=11.6$, $p<.01$, $\varepsilon=.51$	p = 0.595				
integrated SI shear		**p = 0.026**	**F(2,10) = 5.34**		p = 0.044	
max SI shear	$\chi^2(2)=[]$, $p<.05$, $\varepsilon=.55$	p = 0.068				
integrated CF	$\chi^2(2)=6.23$, $p<.05$, $\varepsilon=.56$	**p = 0.003**	**F(1.1, 5.6)=24.55**	p = 0.001		p = 0.01
max CF	$\chi^2(2)=6.84$, $p<.05$, $\varepsilon=.55$	**p = 0.009**	**F (2,10) =14.97**	p = 0.008		p = 0.03




Integrated SI Shear Forces
SI shear integrated (%BW*s)
2500
2000
1500
1000
500
0
reduced
normal
increased
CSA
Integrated CF Forces
CF integrated (%BW*s)
14000
12000
10000
8000
6000
4000
2000
0
reduced
normal
increased
CSA
Maximum CF Forces
JRF max (%BW)
200
150
100
50
0
reduced
normal
increased
CSA


CSA normal
CSA reduced
CSA increased
max SI (%BW)
30
20
10
0
0
20
40
60
80
100
120
140
max CF (%BW)