

ARE SMALL FEMALES MORE VULNERABLE TO LOWER NECK INJURIES WHEN SEATED SUFFICIENTLY AWAY FROM THE STEERING WHEEL IN A FRONTAL CRASH?

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Paper Number 17-0147

ABSTRACT

The size and shape of occupants dictates their biomechanical response in any frontal crash. Seatbelt is the primary occupant restraint equipment that reduces risk of serious injuries. The biomechanical response of the occupant is further improved in presence of airbag and advance technologies such as load limiter and pretensioner. Sufficient research and tests are available to assess the biomechanical response of a small female driver sitting close to the steering wheel in a frontal crash. However, no or very less data is available for the response of a small female driver sitting sufficiently away from the steering wheel in a frontal crash scenario. This study is more focused on the lower neck biomechanics of a small driver sitting sufficiently away from the steering wheel.

A validated MADYMO model of a standard regulatory frontal crash test is used to test the hypothesis using the factorial design. The factors selected are the steering-chest distance, D-ring height adjustment and the pulse type. The study shows that small females are vulnerable to high lower neck flexion moments with increase in the steering-chest distance (SCD).

INTRODUCTION

The seatbelt in a vehicle is the primary restraining device for any crash mode. Especially, in a frontal crash, the seatbelt dissipates the kinetic energy of the occupant in a controlled manner to increase their level of crash protection. Furthermore, the seatbelt performance is enhanced by advanced technologies such as pre-tensioner and load limiter in the presence of supplemental airbag systems. The optimum restraint performance in a frontal crash depends on all the restraining safety systems working together including the proper pelvic restraint from the vehicle seat. This overall optimum performance ensures the seatbelt force transmission on desired stronger skeletal locations preventing any seatbelt induced injuries due to submarining [1,2,3,4,5, and 6]. The head and neck biomechanics in a frontal crash depends on the overall restraint performance in the absence of any head impact with the non-deploying vehicle interior. The inertia of the head produces neck flexion moment and neck shear load. The lower cervical spine is subjected to higher flexion moment compared to upper level. However, the shear load remains almost constant along the entire length. Figure 1 shows the schematics of flexion moment magnitude along the cervical spine induced by head inertia in a frontal crash for a properly belted occupant.

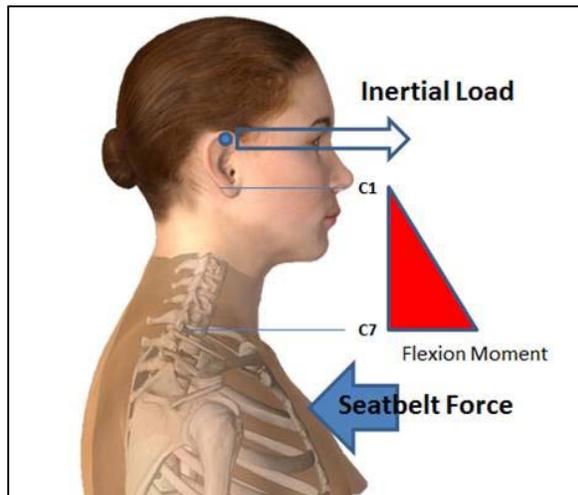


Figure 1. Flexion moment magnitude along the cervical spine induced by the head inertial load.

The motivation for this study stems from a real life accident where a properly belted small statured female sustained severe lower neck injury. She sustained C5-C6 Bilateral Facet Dislocation (BFD)

causing her spinal cord injury. She chose to sit sufficiently away from the steering wheel (SCD \approx 16 inches). Her vehicle featured a conventional single loop seatbelt system equipped with advanced technologies such as belt pre-tensioner and load limiter. The D-ring and the anchor point of the belt are mounted on the vehicle body.

Bilateral facet dislocation (BFD) in the cervical spine includes hyperflexion, flexion-distraction, and flexion-compression as injury mechanisms [7,8]. The flexion moment is a significant component of this injury mechanism. Punjabi et al.[9] successfully produced BFD in the laboratory by applying inertial load on the superior vertebrae when the inferior vertebrae were fixed. In the frontal crash, the flexion moment acting on the lower cervical spine is dictated by the seatbelt performance in dissipating the occupant thorax energy.

The aim of this study was to identify the significant factors exacerbating the lower neck flexion moments of a small female in a frontal crash scenario when she chose to sit sufficiently away from the steering wheel. Females are at higher risk compared to belted male counterpart during vehicle collisions [10]. This study tests the hypothesis as below

“ Farther seat position increases the risk of lower neck BFD for a small female occupant in a vehicle with a conventional belt system.”

METHODOLOGY

Sensitivity analysis using factorial design is employed to identify the significant factors and their interactions dictating the lower neck flexion values of a small stature female in a frontal crash.

Online NHTSA crash test database was searched [11] to identify frontal barrier test with a 5th percentile ATD (Anthropomorphic Test Device) in which the ATD has positioned sufficiently away from the steering wheel (more than 8 inches steering-to-chest distance). No test is available with the 5th percentile ATD in a situation where the Steering-to-Chest distance (SCD) was greater than 8 inches. Crash test NHTSA 4419 was selected in which the 5th ATD was positioned close to the steering with approx. 8 inches of SCD. The vehicle in the test featured a conventional single loop seatbelt system equipped with advanced technologies such as pre-tensioner and load limiter.

Computational Model Overview

MADYMO [12] model is employed to quantify desired response of lower neck flexion value to be used for the Factorial design. The ATD and the vehicle interior were modeled and positioned according to the data produced in the test report. Figure 2 shows the initial model setup. Red spheres in the model denote pre-test measurements of various locations. The SCD in the model is set to 200mm (7.87 inches) as per the test measurement. The computational model features seatbelt pretensioner and load limiter.

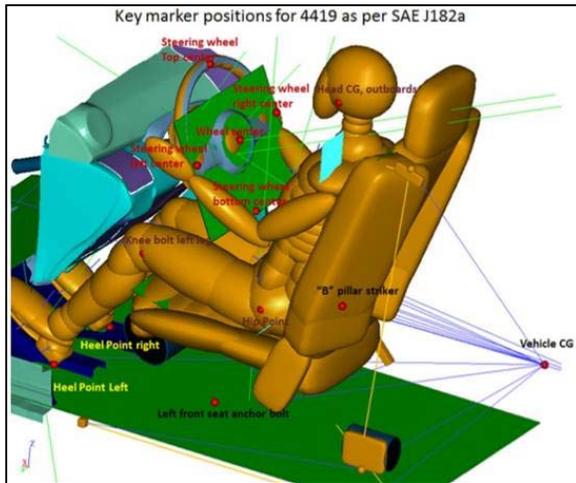


Figure 2. Model set up using 5th ATD.

Model Validation

The biomechanical response of the ATD and the seatbelt loads are compared to check for the computational model validity. Figure 3 shows the kinematic comparison of the test and the model. Figure 4 shows the shoulder belt comparison between the test and the model. Figure 5 demonstrates the comparison of resultant head acceleration between the test and the model. Figure 6 shows the comparison of chest deflection. Figure 7 displays the comparison between the thorax acceleration. Figure 8 indicates the correlation between the model and the test for pelvic acceleration. The comparison of the model and the test 4419 plots shows a good correlation. Table 1 demonstrates the comparison of upper neck Nij neck injury measure and Table 2 indicates the comparison of HIC (Head Injury Criteria) between the test and the model.

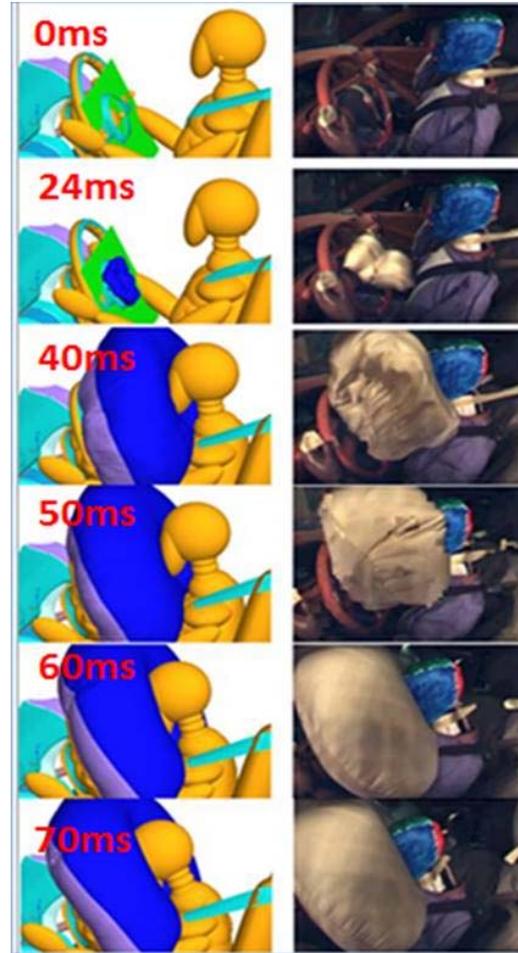


Figure 3. Head and neck kinematics comparison.

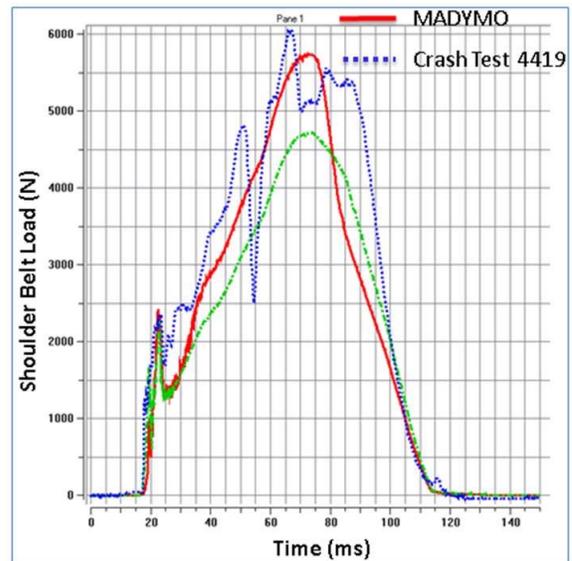


Figure 4. Shoulder belt load comparison.

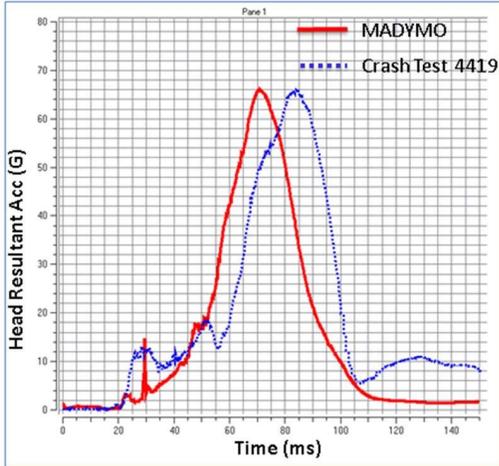


Figure 5. Head resultant acceleration comparison.

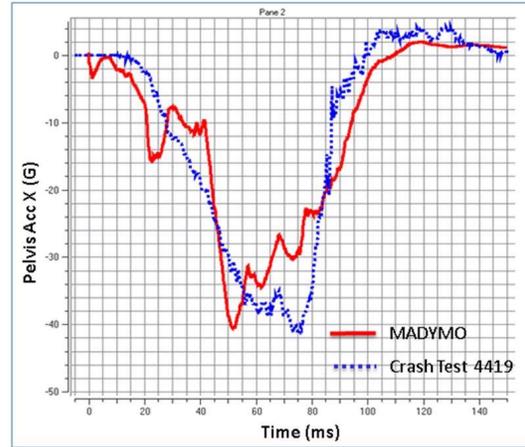


Figure 8. Pelvic acceleration comparison.

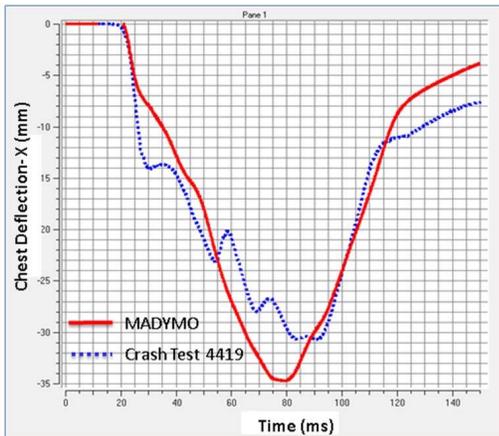


Figure 6. Chest deflection comparison.

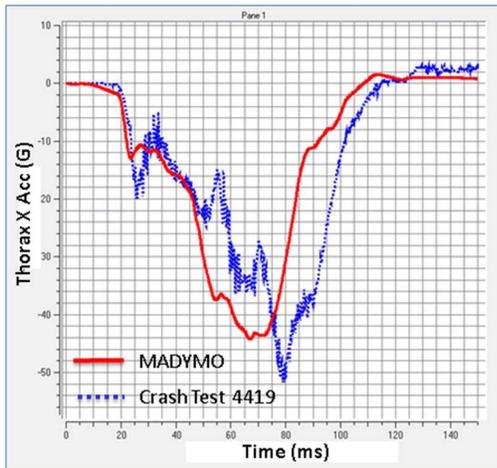


Figure 7. Thorax acceleration comparison.

Table 1.
Upper Neck Nij comparison

Driver Nij	N _{TE}	N _{CE}	N _{TF}	N _{CF}
NHTSA4419	0.64	0.01	0.21	0.01
MADYMO4419	0.19	0.01	0.27	0.06

Table 2.
HIC Comparison

Driver	HIC	HIC (36ms)	HIC (15ms)
NHTSA4419	686 T1: 60.2 ms T2: 92.8 ms	686 T1: 60.2 ms T2: 92.8 ms	451 T1: 71.1 ms T2: 86.1 ms
MADYMO4419	595 T1: 56.6 ms T2: 86.0 ms	595 T1: 56.6 ms T2: 86.0 ms	451 T1: 64.4 ms T2: 79.4ms

Sensitivity Analysis Using Factorial Design

The validated model is employed to conduct a factorial design at two levels. The factors such as SCD, D-ring height, and the pulse type are investigated for their joint effect on the lower neck flexion value. The NHTSA NCAP pulse is compared with the IIHS offset pulse. Table 3 shows the treatment combination and corresponding response as obtained using the computational model. The low and high levels of factors selected are SCD at 8 inches, and 12 inches, D-ring Height at 33 inches and 35 inches, and the pulse for NHTSA frontal and IIHS offset barrier. Figure 9 shows the half-normal probability

plot of factor effects shows the important SCD effect on the response compared to other factors. Figure 10 shows the Pareto chart showing the relative size of the effects clearly indicating the magnitude of SCD effect on the response [13].

Table 3.
Treatment combinations and response for the design.

SCD (Inches)	D-Ring Height Inches	Pulse Type	Peak Flexion (N.m)
8	33	NCAP	66
12	33	NCAP	169
8	35	NCAP	70
12	35	NCAP	186
8	33	IIHS	105
12	33	IIHS	187
8	35	IIHS	96
12	35	IIHS	160

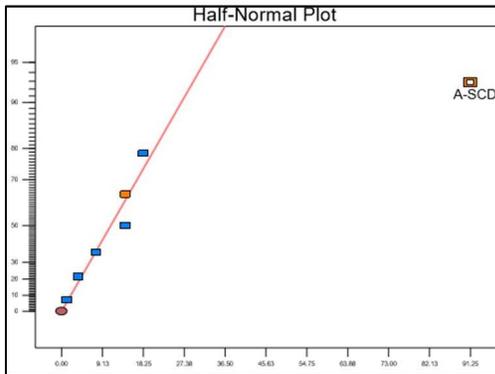


Figure 9. Half Normal plot with effect chosen(Appendix A).

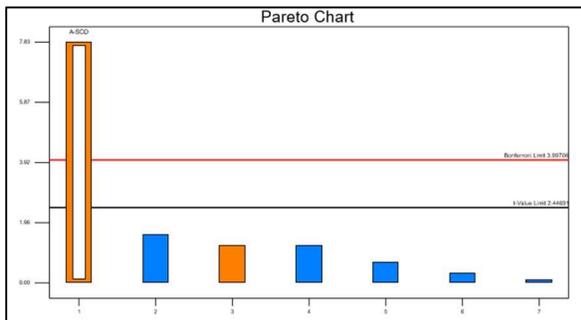


Figure 10. Pareto Chart showing magnitude of SCD(Appendix A).

The analysis of variance (ANOVA) shows the SCD as the most significant factor dictating the lower neck flexion moment. The ANOVA also shows the D-ring height is insignificant in dictating the lower

neck response. The lower neck flexion moment increases as the SCD increases. This relation is shown in Figure 11.

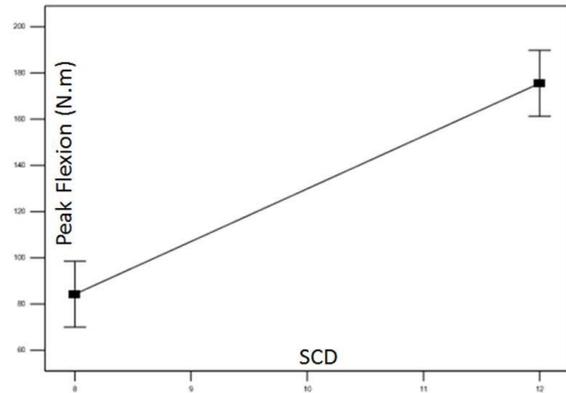


Figure 11. Lower neck peak flexion increases with SCD increase(Appendix A).

DISCUSSION

The ANOVA analysis clearly indicates SCD as the most significant factor affecting the lower neck flexion moment during frontal crash. As the seat position is set backward in a vehicle with a conventional three-point single loop seatbelt system the lower neck flexion moment of a 5th percentile, ATD increased. The tested hypothesis is correct.

The belted thorax kinematics dictates the lower neck flexion moment. The shoulder and lap belt configuration changes with the seat fore-aft position as two points are mounted on the vehicle body. Figure 12 and 13 shows the change in belt configuration as the position is moved backward while increasing the SCD. The belt angle about the horizontal and vehicle longitudinal axis increases with increase in the SCD.

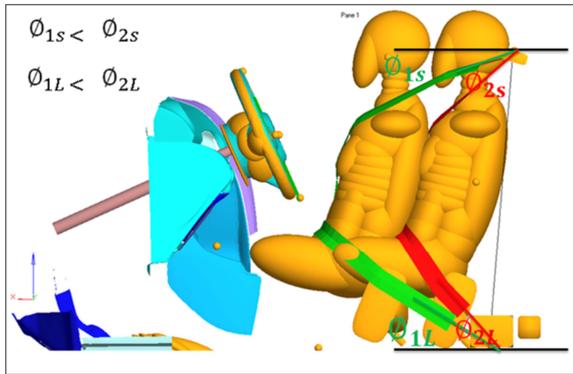


Figure 12. Increase in belt angle increase in SCD.

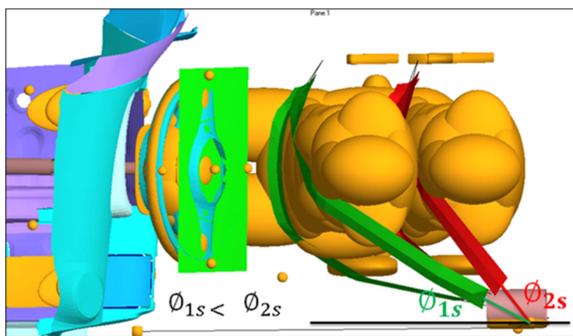


Figure 13. Increase in shoulder belt angle with respect to the longitudinal axis.

The shoulder belt configuration for the higher SCD delays the thorax restraint. This delay causes gain in its kinetic energy before the shoulder belt restrains it. This phenomenon increases the chest deflection and exacerbates the lower neck flexion moment increasing the risk of BFD. This phenomenon is explained in detail elsewhere [14]. The shoulder belt configuration with the smaller SCD tends to restrain upper torso much earlier in the crash allowing gradual dissipation of the thorax kinetic energy. Furthermore, the knee bolster is also effective in restraining the occupant. In contrast, the higher SCD shoulder belt configuration allows upper body velocity build up before the shoulder belt restrains it. This gain in velocity before restraint increases the lower neck flexion moment in small occupants who chose to sit away from the steering wheel.

CONCLUSIONS

The biomechanical response of the MADYMO model correlated very well with the NHTSA test 4419. This model was successfully used to

conduct the sensitivity analysis using factorial design at two levels to test the hypothesis. The conventional seatbelt performance degrades with the higher SCD for small occupants in context to their lower neck biomechanics. Small occupants who chose to sit away (SCD >8 inches) from the steering wheel are at higher risk of lower neck flexion induced injuries in a frontal crash in a vehicle with the conventional three-point belt system.

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APPENDIX A

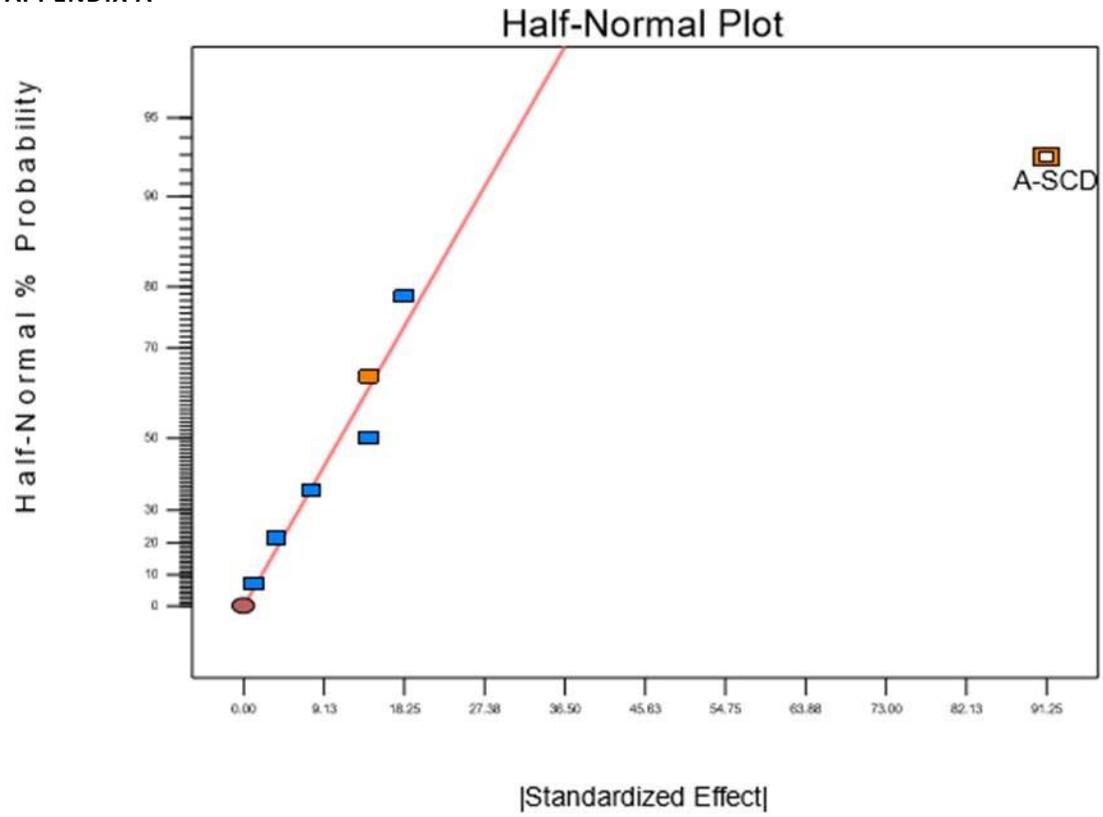


Figure 9. Half Normal plot with effect chosen.

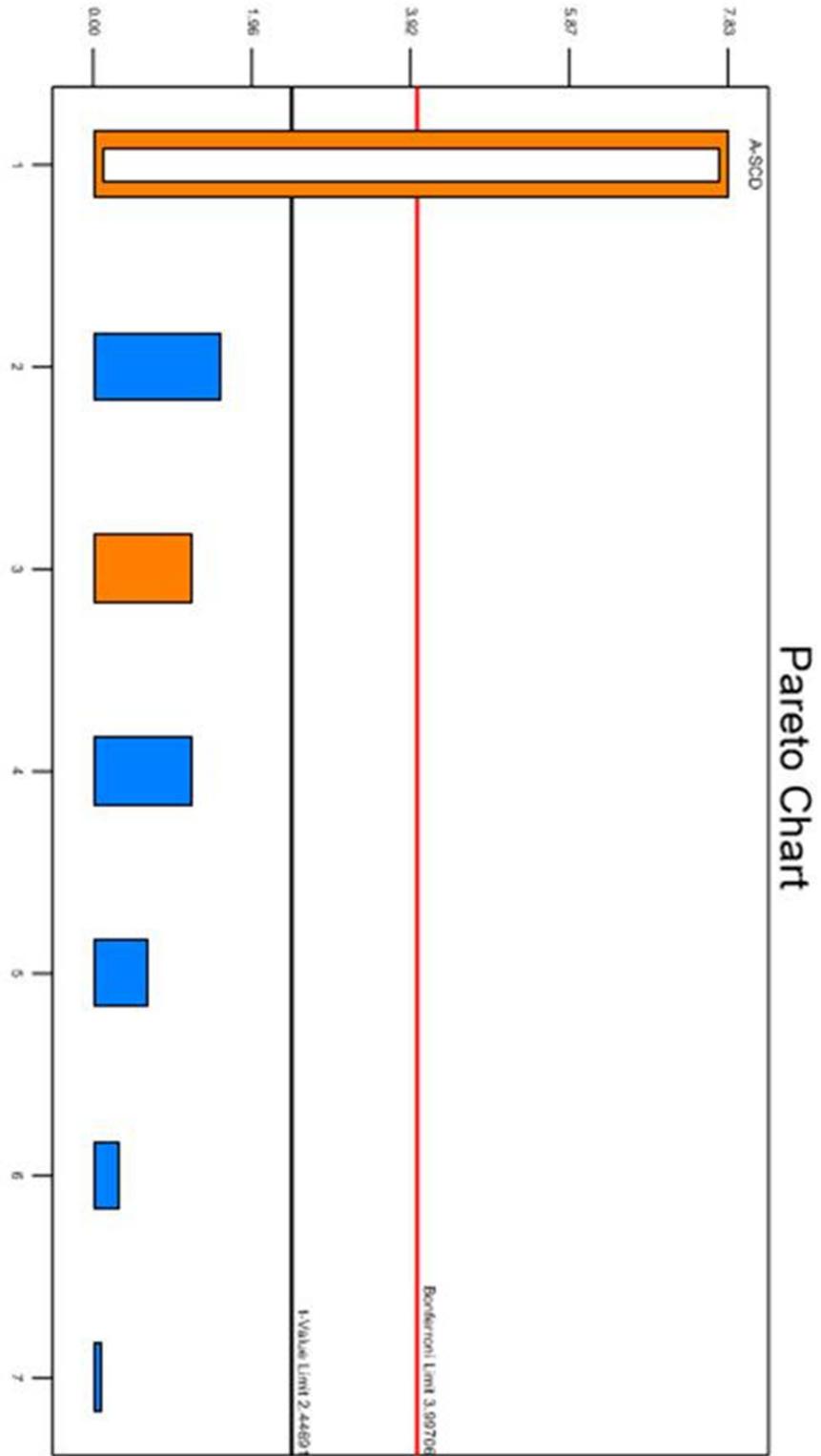


Figure 10. Pareto Chart showing magnitude of SCD.

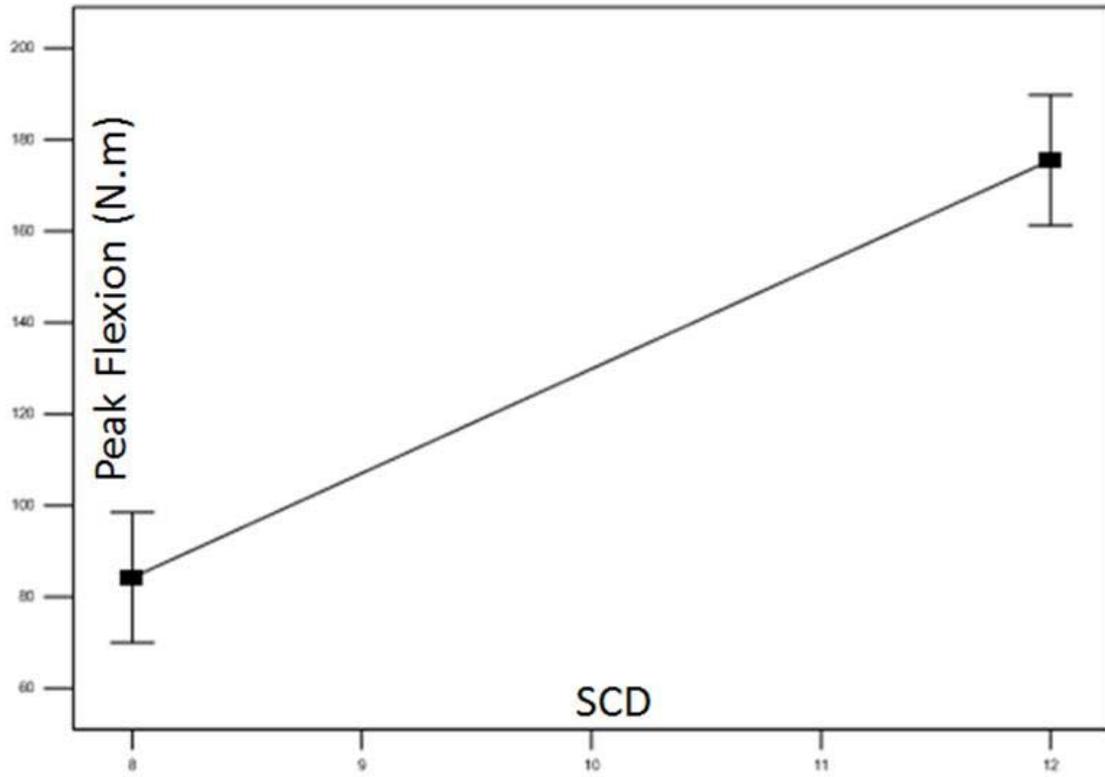


Figure 11. Lower neck peak flexion increases with SCD increase.