Joint Helmet-Mounted Cueing System and Neck Muscle Activity During Air Combat Maneuvering

Roope Sovelius; Maunu Mäntylä; Huhtala Heini; Juha Oksa; Rasmus Valtonen; Liisa Tiitola; Tuomo Leino

BACKGROUND: The aim of the study was to determine the characteristics of cervical muscle activity in different head postures when using helmet-mounted display in one fighter vs. two aircraft air combat within visual range (WVR).

METHODS: Cervical EMG was measured with eight F/A-18 pilots using the Joint Helmet Mounted Cueing System (JHMCS) during air combat maneuvering. In-flight G_z acceleration and continuous head position were recorded. EMG activity is divided and presented in a matrix with three-class rotation and five-class flexion-extension postures.

RESULTS: The mean muscle activity in sternocleidomastoids and cervical extensors was 28.9% of maximal voluntary contraction (MVC) and 44.8% MVC, respectively. Cervical flexor and extensor muscles are subjected to loading over MVC during high G_z sorties. Cervical rotation combined with extension exceeded muscle force-producing capacity during high G_z, resulting in a decline in muscle activity.

DISCUSSION: Awkward postures, especially rotational ones, are more prone to increase loading over muscles' capacity. Overloading of muscles increases the risk of muscular and ligamentous injury. In addition, the lack of muscular support potentially leads to the G_z loading being transferred to spinal structures via intervertebral discs and the vertebral column. The JHMCS helmet seems to change the pattern of most loading muscles toward the extensor (posterior) neck muscles.

KEYWORDS: +Gz, EMG, workload, cockpit ergonomics, JHMCS, cervical load.

Sovelius R, Mäntylä M, Heini H, Oksa J, Valtonen R, Tiitola L, Leino T. *Joint helmet-mounted cueing system and neck muscle activity during air combat maneuvering*. Aerosp Med Hum Perform. 2019; 90(10):834–840.

Helmet mounted displays provide situational information and cues from weapon systems for the modern fighter pilot. The integration of technology to enhance information capacity and tactical performance also has disadvantages. It has resulted in increased helmet weight and shifted the head-helmet systems' center of gravity up front, thus leading to a greater torque in the neck.[23] Feedback from pilots is that helmet-mounted cueing systems may have diminished cumulative exposure to high G_z due to shorter air-to-air combat time. At the same time, awkward head postures, especially during the highest G_z forces, have increased.

The Joint Helmet Mounted Cueing System (JHMCS) is not just reported to be the most common cause of flight-related pain.[29] It can also cause significant worsening of pain, especially during flexion of the head or if there is a history of prior neck problems.[6] In Lange's[16] study, the pilots reported combined rotation and extension movements to be the most harmful.

Tactical use of JHMCS forces pilots to use the full range of motion of the head; thus ligaments begin to provide a counteracting force. Coackwell et al.[7] and Snijders et al.[27] have described the high-risk movements as rotations that exceed 35°, extensions that are beyond 30°, and flexions that exceed 15°, as well as all lateral bending. Beyond these limits, the efficiency and force-generating capacity of muscles decrease, and the joint reaction forces tend to increase rapidly.[13,19]

The maximal isometric strength of the neck muscles diminishes downward to the cervical vertebral levels.[24,26,30] Produced extension force or torque is greater than flexion force, and the neck rotator muscles have the least isometric force generating capacity among the functional neck muscle groups.[8] Cervical musculature strength levels apparently vary with head-neck

From the Centre for Military Medicine, Finnish Defence Forces, Pirkkala, Finland; Patria, Tampere, Finland; the University of Tampere, Tampere, Finland; the Institute of Occupational Health, Oulu, Finland; the University of Oulu, Oulu, Finland; and the National Defence University Finland, Helsinki, Finland.

This manuscript was received for review in October 2018. It was accepted for publication in June 2019.

Address correspondence to: Roope Sovelius, P.O. Box 1000, FI-33961 Pirkkala, Finland; roope.sovelius@mil.fi.

DOI: https://doi.org/10.3357/AMHP.5281.2019

position and the direction of contraction. The maximal neck rotation strength is achieved with the head and neck at the highest prerotation angles. However, the neck loads in different postures (flexion, extension, and rotation) under operational conditions are unknown. The aim of the study was to determine definitively the characteristics of cervical muscle activity during one pilot vs. two aircraft air combat maneuvering to identify the conditions when risk limits are achieved in order to get more knowledge for appropriate and effective countermeasures.

METHODS

Subjects

There were 13 Finnish Air Force F/A-18 pilots who acted as test subjects. Their mean age was 30 yr (range 27–35). Their mean height was 180 cm (SD ± 4 cm), weight 78 kg (± 5 kg), and body mass index (BMI) $24 \pm 1\ \mathrm{kg \cdot m^{-2}}$. The subjects had no flight limitations in their flying statuses and they did not report any current musculoskeletal symptoms before test flights. They all used the JHMCS (Vision Systems International, San Jose, CA, USA) on a regular basis. Of 13 test flight recordings, 5 were excluded from the study due to technical problems in EMG recordings (3 flights) or with test flight instruments (2 flights).

The subjects were informed of the details of the experimental protocol. The Finnish Defense Forces Medical Research Register, the Finnish Air Force Headquarters, which granted a Research License, and The Ethical Committee of the Central Finland Hospital District approved this study.

Equipment

The test flights were flown in a dedicated F/A-18D Hornet aircraft equipped with test flight instrumentation collecting acceleration of the aircraft in the x, y, and z direction and attitude in pitch, roll, and yaw, and their rates of changes of these variables. Helmet's elevation (flexion or extension angle), azimuth (rotation angle), lateral bending, and the rate of head movements in these planes were also collected as a function of flight time. Each pilots' maximal range of movements in rotation, extension, and flexion were measured during JHMCS calibration before flight. The line of sight of the helmet was used to determine head postures and movements. This was done when strapped in the cockpit with full flight gear on just prior to the start of the sortie.

During the sorties EMG activity of the right and left sternocleidomastoid (SCM), cervical (CES), thoracic, and lumbar erector spinae, as well as trapezoids and oblique abdominal muscles, was measured using bipolar surface electrodes. Measured EMG was compared with EMG recorded during maximal voluntary contraction (MVC) performed prior to the walk to the aircraft, and EMG level and muscle activity were determined as a percentage of MVC (%MVC). Muscular activity was determined using a portable eight-channel EMG device (ME6000P, Mega Electronics Ltd., Kuopio, Finland). Bipolar EMG recordings were obtained using pre-gelled surface electrodes (Medicotest M-OO-S, Olstykke, Denmark) placed longitudinally on muscles with a distance of 2 cm between their measurement surfaces. Ground electrodes were placed on inactive tissues. EMG signals from the skin above working muscles were acquired at a sample rate of 1000 Hz. The measured signal was pre-amplified 2000 times, and the signal band between 20 and 500 Hz was full-wave rectified and averaged with a 100-ms time constant. The mean activity and the peak activity during encounters were studied. EMG signal was averaged with a 1-s time constant for peak EMG signal analysis to filter noise signals and artifacts. The values of EMG activities were read for each subject during different postures and G_z levels.

Procedure

The data was collected during an air combat sortie, one fighter against a two aircraft formation within visual range (WVR). The test subject flew the solo aircraft against two others; this setting was chosen in order to provide as much head movement as possible. The test flight consisted from three to six encounters depending on operational factors. All encounters were recorded from starting point to an end, when a pilot terminated the encounter. All encounters started beyond visual range and pilots were briefed to start them with similar set-ups so that the beginning of maneuvering would be identical. The fight then developed to WVR, dogfighting freely within a tactical situation and without any limiting factors in terms of study settings. The transit flights and time between encounters were excluded from more detailed analysis.

All recorded data via the JHMCS helmet system was linearly interpolated to a 0.02-s time frame during post-processing. Similarly, EMG data was linearly interpolated to the same 0.02-s time frame. EMG and JHMCS data had different time sources and those two data sources needed to be synchronized. The small 0.02-s time frame helped to achieve more accurate synchronization. Time synchronization was determined by comparing JHMCS sensor position values and rates and EMG activity during JHMCS calibration.

Posture matrix head position was classified as either neutral, or at-risk posture (@risk), or beyond voluntary maximal range of motion (ROM) (>max). The posture was classified as @risk if ROM limits were exceeded in accordance with Coackwell[7] and Snijders.[27] If during flight the pilot exceeded his maximal values of ROM in flexion, extension, or left or right rotation measured in calibration, the posture was considered >max. Thus the @risk values were constant over the subjects, but >max values used were set according to each subjects' individual values during ROM calibration. The results are presented in a 5 × 3 position matrix chart with five sagittal positions and three axial rotation positions. Results are determined in the contralateral and ipsilateral muscles (SCM, CES), respectively, with direction of cervical rotation and within different G_z levels. The principle of the posture matrix is presented in **Fig. 1**.

During posture analysis, contra- and ipsilateral SCM and CES were separated. When the pilot was looking to the right, the right-side SCM and CES were defined as ipsilateral muscles and left-side muscles as contralateral. When the pilot was looking

Sagittal plane	Muscle & Gz level		
Flexion >max	**No rotation Flexion beyond individual maximum**	**Rotation > 35° Flexion beyond individual maximum**	**Rotation and flexion beyond individual maximum**
Flexion @risk	**No rotation Flexion > 15°**	**Rotation > 35° Flexion > 15°**	**Rotation beyond individual maximum Flexion > 15°**
↑ Neutral ↓	**Head in neutral position**	**Rotation > 35° No flexion/extension**	**Rotation beyond individual maximum No flexion/extension**
Extension @risk	**No rotation Extension > 30°**	**Rotation > 35° Extension > 30°**	**Rotation beyond individual maximum Extension > 30°**
Extension >max	**No rotation Extension beyond individual maximum**	**Rotation > 35° Extension beyond individual maximum**	**Rotation and extension beyond individual maximum**
Rotation →	Neutral	@risk	>max

Contralateral / Ipsilateral muscles are presented separately.
Left and right muscles are combined to contra-/ipsilateral data, depending on the performed direction of rotation

Fig. 1. Reading guide for posture matrix (for Fig. 2 and Fig. 3).

to the left, opposite muscles were defined as ipsi- and contralateral, respectively. Thus, later in the text, ipsi- or contralateral terms include data from both right and left muscles, as the side is dependent on the direction of rotation at the moment. The aim of this approach was not to lose the comparable data between agonist-antagonist or bilateral muscle group activations regardless of direction of rotation.

Statistical Analysis

Descriptive statistics and means with 95% confidence intervals (CI 95%) are used as descriptive parameters. All changes in head postures during test flights were considered as separate test points; subsequently some 156,000 postures were analyzed in all.

Measured outcome (EMG) was the continuous parameter and considered normally distributed. A mixed (random + fixed) model of ANOVA was used to study the effect of given factors on muscle activity. Subjects were considered random factors as there is an intrasubject dependency on measured test points; head postures in different planes (sagittal, horizontal, and frontal plane) and different G_z levels were considered as fixed factors. The level of statistical significance was set at $P < 0.05$. Primarily combined effect of factors was considered. When an effect of an individual factor is presented, it is then stated. Modeling was done as in the posture matrix description above.

The study setting did not allow use of controls during test flights. Control measurements would have needed similar flights with a legacy helmet without head posture data or flying with the JHMCS off. Both options would cause different data to be collected, as well as a lack of available information for pilots that would lead to different cervical loading due to different tactical flying. Thus the study modeling is quantitative in nature.

RESULTS

Pilots' cervical range of movement represents the range of line of sight of the cueing system and vice versa. When pilots were sitting in the cockpit fully equipped, the range of cervical rotation to right or left was on average 97° (range 80–125°). Mean flexion from the neutral posture was 51° (range 30–75°), mean extension 48° (range 35–70°), and in lateral bending mean was 38° (range 30–45°). There was no difference in range between left and right in the rotational axis or in lateral bending. The head was in a neutral position 38% of the time during encounters, at biomechanically assessed risk posture at least in one plane 49% of the time, and beyond preflight measured maximal range of movement in at least in one plane 13% of the time.

Mean muscular activity during flights was highest in cervical muscles SCM and CES. These muscles were primarily affected during at-risk postures. Thus, the results of other measured muscles are excluded in this article.

The mean muscle activity in SCM and CES was 28.9% MVC (CI 95% 21.3–36.5% MVC) and 44.8% MVC (CI 95% 33.2–56.4% MVC), respectively. EMG activity during the whole sortie was between 15% and 50% of MVC 19% of the time, and above 50% MVC 1% of the time in the SCM and CES muscles.

Statistically significant factors for muscular loading were rotations [$F(3,140503) = 4.838$, $P = 0.005$], flexion-extension [$F(3,140503) = 8.263$, $P < 0.001$], and above all acceleration of the aircraft [$F(3,140503) = 6.886$, $P = 0.001$]. Combinations of these factors did not markedly change the level of muscle activity (%MVC in EMG) in general. More detailed results with rotation side difference included are presented in **Table I**. Lateral bending was also analyzed, but no statistically significant effect on SCM [$F(4,112240) = 2.193$, $P = 0.092$] or on CES [$F(4,112240) = 0.191$, $P = 0.942$] was found. It is not used as a separate factor in further analysis. However, there is a strong pattern of coupling in the cervical spine with axial rotation associated with lateral bending and it is difficult to isolate lateral bending from rotation and/or sagittal movements.

Postures with a combination of rotation and extension already cause very high muscle activity (>MVC) during low acceleration of +2–3 G_z. When the pilot's head is further from neutral position in rotation and/or flexion or extension, more lateral flexion additionally increases muscular loading. Above +4 G_z pilots avoid flexion postures (empty columns in **Fig. 2** and **Fig. 3**). When cervical flexion took place during higher G_z it resulted in high muscle activity.

Contralateral SCM is subjected to high demands during cervical rotation postures. Ipsilaterally higher rotation angles resulted in lower muscle activity as the muscle's load-bearing capacity is exceeded either by G_z load or due to an awkward posture where the muscle is unable to work efficiently, or as a combination of these conditions. A similar pattern is seen in the CES on the extensor side during moderate acceleration (+4–5 G_z). A higher G_z load causes very high muscle loading on extensors during flexion postures (Fig. 2).

The highest peaks in EMG activity were well above preflight maximal voluntary contraction levels. However, there was a tendency both in the SCM and CES for decreased muscular activity in higher rotation angles in all sagittal plane postures and at all G_z levels (Fig. 3).

DISCUSSION

Peak muscle activities in our study were all very high (>MVC). This shows that pilots are easily subjected to higher loading compared to those recorded before flight. Cervical muscles and ligaments are at risk for injury in most awkward postures, e.g., check six and high G_z levels. Results of this study confirm the need for preflight preparations and warming up. Of course, the accuracy of MVC can be discussed and presented values of muscle activity must be considered with care. Normalization of EMG activity has its bias,[4,28] but on the other hand, all previous in-flight EMG measurements have been performed with the same methodology, so results are comparable with previous reports.

Previous studies have presented a correlation of acceleration and muscle activity.[11,12,20] The results of this paper support this with some notes. The independent effect of G_z was statistically significant both in the SCM and CES. These muscles act as load-bearing columns during postures with rotation. Contralateral SCM has an important role also as a rotator muscle. When the head is rotated beyond risk limits and combined with extension, the contralateral SCM is stretched and concentric muscle activation changed to eccentric.[17] In extensor muscles, the more G_z loading there is, the more contralateral CES is activated. Ipsilateral CES activation is diminished when maximal rotation takes place. Load-bearing function is more clearly seen in contralateral muscles during rotations as ipsilateral SCM and CES are shortened and they are not able to perform optimally.

Maximal EMG activity is lower in eccentric conditions compared to concentric or isometric actions. During this high-tension loading condition, the neural drive to the activated muscles is reduced, despite maximal voluntary effort.[31] Thus, during the maneuvering at +6–7 G_z, the EMG activity of muscles

Table I. Summary of Independent Variables of Muscular Loading (%MVC).

	SCM		CES	
	CONTRALATERAL	IPSILATERAL	CONTRALATERAL	IPSILATERAL
	MEAN (SD)	MEAN (SD)	MEAN (SD)	MEAN (SD)
Rotation				
neutral	25.4 (29.8)	22.0 (26.7)	30.6 (37.0)	37.7 (41.3)
@risk	45.7 (37.8)	42.9 (41.0)	51.6 (41.6)	51.5 (46.4)
>max	61.2 (39.8)	75.2 (49.4)	74.0 (45.6)	58.4 (42.7)
F_{df}	$F(2,129893) = 32.1$	$F(2,127277) = 15.9$	$F(2,127007) = 25.1$	$F(2,124410) = 15.2$
P	<0.001	<0.001	<0.001	<0.001
Flexion-extension				
flexion > max	22.2 (27.2)	19.7 (22.3)	36.6 (39.3)	35.3 (35.5)
flexion @risk	17.7 (25.3)	15.2 (24.8)	26.6 (35.2)	27.1 (33.8)
neutral	23.9 (27.7)	24.7 (31.3)	29.2 (35.6)	37.0 (40.9)
extension @risk	51.0 (39.5)	40.9 (37.8)	47.8 (37.7)	50.7 (46.6)
extension > max	56.6 (37.8)	46.1 (39.6)	68.0 (45.7)	62.8 (45.7)
F_{df}	$F(4,129894) = 32.5$	$F(4,127277) = 11.7$	$F(4,127007) = 14.1$	$F(4,124410) = 10.4$
P	<0.001	<0.001	<0.001	<0.001
Acceleration (+G_z)				
1	20.2 (24.6)	22.0 (29.9)	25.9 (33.8)	33.8 (39.3)
2–3	44.1 (36.8)	36.8 (36.7)	52.3 (44.1)	51.5 (45.2)
4–5	54.8 (40.4)	43.9 (39.2)	56.6 (40.7)	55.3 (47.1)
6–7	54.7 (36.6)	42.8 (36.1)	61.3 (42.0)	65.0 (44.0)
F_{df}	$F(3,129895) = 53.4$	$F(3,127278) = 15.4$	$F(3,127008) = 20.8$	$F(3,124411) = 10.0$
P	<0.001	<0.001	<0.001	<0.001

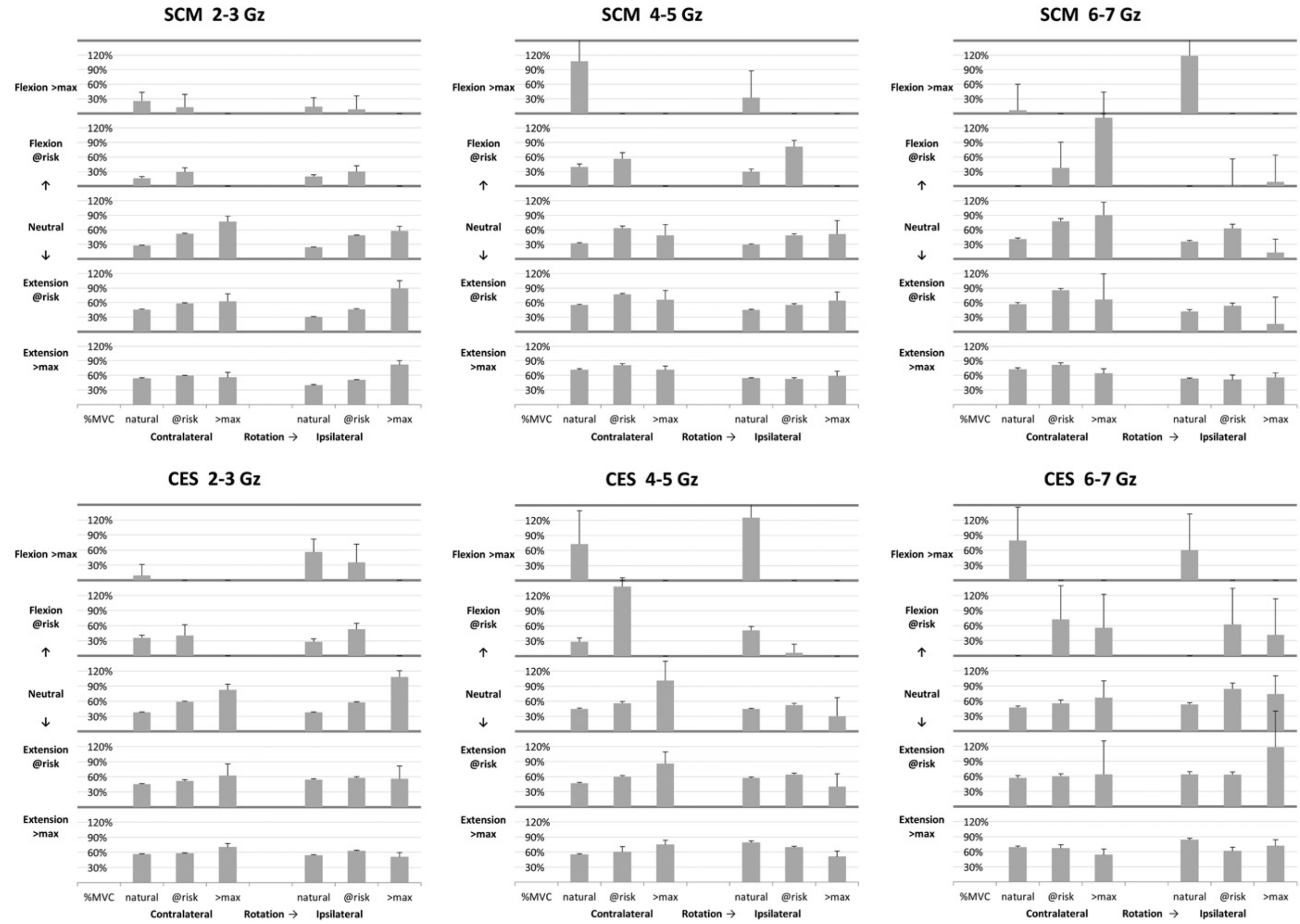

Fig. 2. Averaged muscle activity during encounters in different postures and acceleration levels. %MVC = proportional maximal voluntary contraction activity; SCM = sternocleidomastoids; CES = cervical erector spinae; @risk = head posture at biomechanically assessed risk posture in given plane; >max = head posture beyond calibrated maximal ROM; Ipsilateral = muscle toward rotation; contralateral = muscle away from rotation. Error bars present 95% confidence intervals. Absence of bar indicate that given posture at that G_z-level did not take place during test flights.

did not increase prominently compared to +4–5 G_z. Additionally, a decreasing tendency in EMG activities from neutral to >max in different muscles, sagittal planes, and acceleration levels was observed (Fig. 3).

When neuromuscular support fails as a result of high G_z and overloaded eccentric muscle activation due to high G_z, more load is transferred through the vertebral columns and intervertebral discs. The role of the facet joints in carrying compressive loads is increased, especially in torsional postures and increased intervertebral disc pressure produces more load via annuli than nuclei.[1,2] These changes in load pathways are reported to result in degenerative changes in the facet joints and intervertebral discs.[5,9] Thus the described risky conditions most likely play a part in premature degenerative changes of the spine in fighter pilots.

Pilots use aircraft structures as head support in order to diminish cervical loading during high G_z.[21] This paper emphasizes how different head postures affect cervical muscle activity, but the presented results do not take into consideration whether the pilot has used the seat's headrest or canopy as a support during high G_z. This may have an effect on results, especially when the head is rotated maximally and during extension postures. Also, the nominal scale for postures includes undetermined different ROM angles within the scale and this may cause some bias in results.

In terms of musculoskeletal loading, what is novel with the JHMCS compared to the flying with conventional helmet system? Due to the tactical efficiency of JHMCS, the length of time for sustained high G_z is reduced, diminishing the cumulative exposure of high G_z.[10,25] But laboratory studies have shown that the extra mass of a helmet and the shift of center of gravity,[3,14] as well as enhanced head movements,[15,22] all increase the workload of the cervical muscles. The mass of the JHMCS results in an over 70% increase in moments of inertia.[18] There is a change of most loading muscles from cervical flexors to extensor side when comparing our results to some earlier in-flight EMG studies with conventional helmets.[20,25] However, the muscle activation patterns presented in this study are generally comparable with a report by Green and Brown.[10]

New technology allowed collecting more detailed information during sorties and this helped to determine some earlier unreported characteristics of muscle activity during air combat maneuvering. The results of this study help to understand the

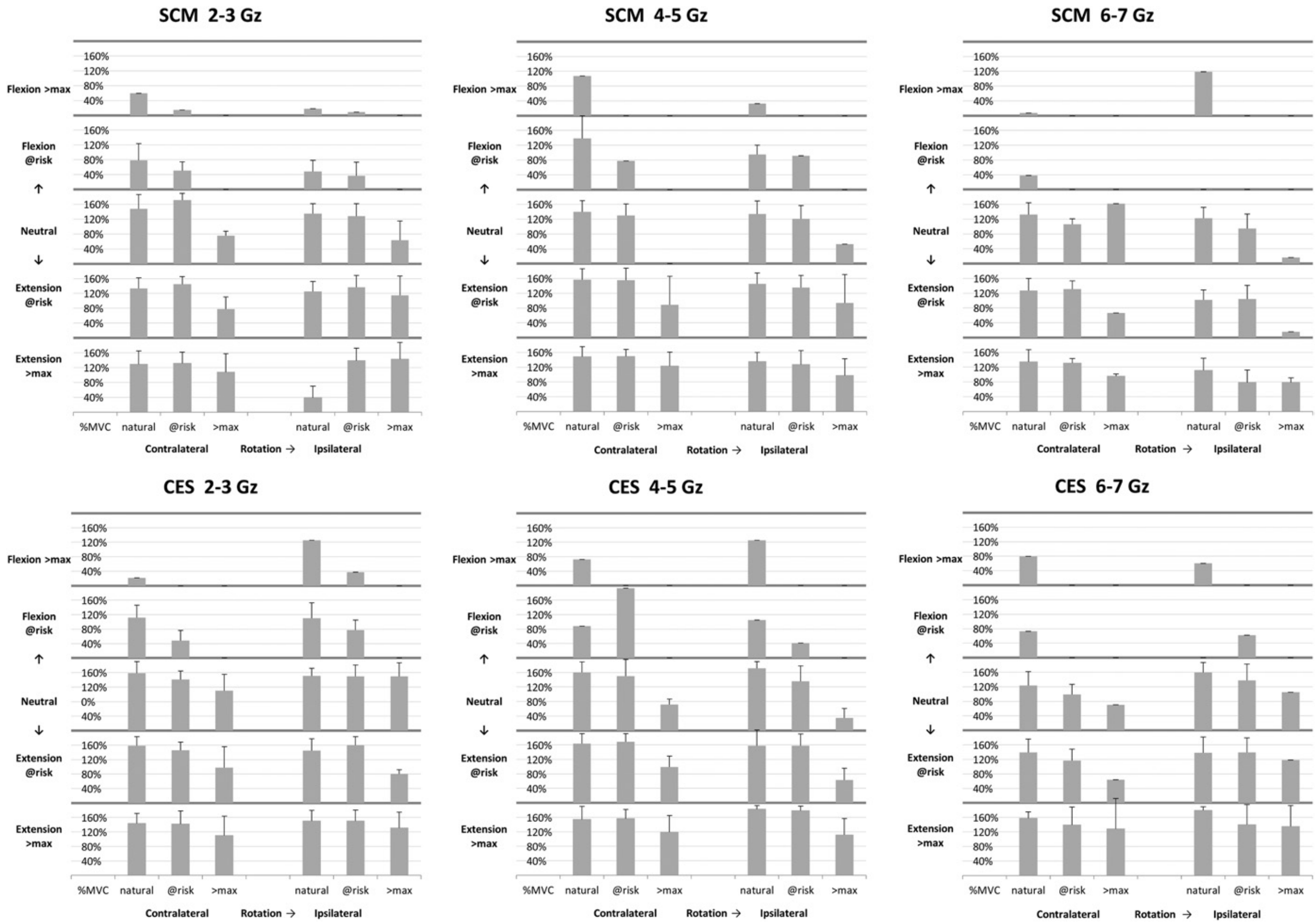

Fig. 3. Peak values of muscle activities in different postures and acceleration levels. %MVC = proportional maximal voluntary contraction activity; SCM = sternocleidomastoids; CES = cervical erector spinae; @risk = head posture at biomechanically assessed risk posture in given plane; >max = head posture beyond calibrated maximal ROM; ipsilateral = muscle toward rotation; contralateral = muscle away from rotation. Error bars present 95% confidence intervals. Absence of bar indicates that a given posture at that G_z level did not take place during test flights.

majority of loading conditions of the cervical spine in the cockpit of a modern fighter and to develop more accurate training programs and cockpit ergonomic training.

ACKNOWLEDGMENTS

This study was supported by a grant from the Centre for Military Medicine, Finnish Defence Forces.

This work was completed within the reinforcing spirit of NATO STO HFM 252.

Authors and affiliations: Roope Sovelius, Ph.D., and Liisa Titola, B.Sc., Centre for Military Medicine, Finnish Defence Forces, Pirkkala, Finland; Maunu Mäntylä, M.Sc., Patria, Tempere, Finland; Huhtala Heini, M.Sc., University of Tampere, Tampere, Finland; Juha Oksa, Ph.D., Institute of Occupational Health, Oulu, Finland; Rasmus Valtonen, M.D., University of Oulu, Oulu, Finland; and Tuomo Leino, Ph.D., National Defence University Finland, Helsinki, Finland.

REFERENCES

1. Adams MA, Dolan P, McNally DS. The internal mechanical functioning of intervertebral discs and articular cartilage, and its relevance to matrix biology. Matrix Biol. 2009; 28(7):384–389.
2. Adams MA, Freeman BJ, Morrison HP, Nelson IW, Dolan P. Mechanical initiation of intervertebral disc degeneration. Spine. 2000; 25(13):1625–1636.
3. Ashrafiuon H, Alem NM, McEntire BJ. Effects of weight and center of gravity location of head-supported devices on neck loading. Aviat Space Environ Med. 1997; 68(10):915–922.
4. Baratta RV, Solomonow M, Zhou BH, Zhu M. Methods to reduce the variability of EMG power spectrum estimates. J Electromyogr Kinesiol. 1998; 8(5):279–285.
5. Boden SD, Riew KD, Yamaguchi K, Branch TP, Schellinger D, Wiesel SW. Orientation of the lumbar facet joints: association with degenerative disc disease. J Bone Joint Surg Am. 1996; 78(3):403–411.
6. Chumbley EM, Stolfi A, McEachen JC. Risk factors for cervical pain in F-15C pilots. Aerosp Med Hum Perform. 2017; 88(11):1000–1007.
7. Coakwell MR, Bloswick DS, Moser R Jr. High-risk head and neck movements at high G and interventions to reduce associated neck injury. Aviat Space Environ Med. 2004; 75(1):68–80.
8. Conley MS, Meyer RA, Bloomberg JJ, Feeback DL, Dudley GA. Noninvasive analysis of human neck muscle function. Spine. 1995; 20(23):2505–2512.
9. Fujiwara A, Tamai K, Yamato M, An HS, Yoshida H, et al. The relationship between facet joint osteoarthritis and disc degeneration of the lumbar spine: an MRI study. Eur Spine J. 1999; 8(5):396–401.
10. Green ND, Brown L. Head positioning and neck muscle activation during air combat. Aviat Space Environ Med. 2004; 75(8):676–680.
11. Hämäläinen O. Flight helmet weight, +Gz forces, and neck muscle strain. Aviat Space Environ Med. 1993; 64(1):55–57.

12. Hämäläinen O, Vanharanta H. Effect of Gz forces and head movements on cervical erector spinae muscle strain. Aviat Space Environ Med. 1992; 63(8):709–716.
13. Helleur C, Gracovetsky S, Farfan H. Tolerance of the human cervical spine to high acceleration: a modelling approach. Aviat Space Environ Med. 1984; 55(10):903–909.
14. Hoek van Dijke GA, Snijders CJ, Roosch ER, Burgers PI. Analysis of biomechanical and ergonomic aspects of the cervical spine in F-16 flight situations. J Biomech. 1993; 26(9):1017–1025.
15. Kumar S, Narayan Y, Amell T, Ferrari R. Electromyography of superficial cervical muscles with exertion in the sagittal, coronal and oblique planes. Eur Spine J. 2002; 11(1):27–37.
16. Lange B, Torp-Svendsen J, Toft P. Neck pain among fighter pilots after the introduction of the JHMCS helmet and NVG in their environment. Aviat Space Environ Med. 2011; 82(5):559–563.
17. Linnamo V, Moritani T, Nicol C, Komi PV. Motor unit activation patterns during isometric, concentric and eccentric actions at different force levels. J Electromyogr Kinesiol. 2003; 13(1):93–101.
18. Mathys R, Ferguson SJ. Simulation of the effects of different pilot helmets on neck loading during air combat. J Biomech. 2012; 45(14):2362–2367.
19. Mayoux-Benhamou MA, Revel M. Influence of head position on dorsal neck muscle efficiency. Electromyogr Clin Neurophysiol. 1993; 33(3):161–166.
20. Netto KJ, Burnett AF. Neck muscle activation and head postures in common high performance aerial combat maneuvers. Aviat Space Environ Med. 2006; 77(10):1049–1055.
21. Newman DG. Head positioning for high +Gz loads: an analysis of the techniques used by F/A-18 pilots. Aviat Space Environ Med. 1997; 68(8): 732–735.
22. Newman DG. Helmet-mounted equipment in the high +Gz environment. Aviat Space Environ Med. 2002; 73(7):730–731.
23. Newman RG. Helmet-mounted display design guide K. San Marcos (TX, USA): NASA, C. Systems; 1997.
24. Nightingale RW, Winkelstein BA, Knaub KE, Richardson WJ, Luck JF, Myers BS. Comparative strengths and structural properties of the upper and lower cervical spine in flexion and extension. J Biomech. 2002; 35(6):725–732.
25. Oksa J, Hämäläinen O, Rissanen S, Myllyniemi J, Kuronen P. Muscle strain during aerial combat maneuvering exercise. Aviat Space Environ Med. 1996; 67(12):1138–1143.
26. Seng KY, Lee Peter VS, Lam PM. Neck muscle strength across the sagittal and coronal planes: an isometric study. Clin Biomech (Bristol, Avon). 2002; 17(7):545–547.
27. Snijders CJ, Hoek van Dijke GA, Roosch ER. A biomechanical model for the analysis of the cervical spine in static postures. J Biomech. 1991; 24(9):783–792.
28. Staudenmann D, Roeleveld K, Stegeman DF, van Dieen JH. Methodological aspects of SEMG recordings for force estimation–a tutorial and review. J Electromyogr Kinesiol. 2010; 20(3):375–387.
29. Thoolen SJ, van den Oord MH. Modern air combat developments and their influence on neck and back pain in F-16 pilots. Aerosp Med Hum Perform. 2015; 86(11):936–941.
30. Vasavada AN, Li S, Delp SL. Three-dimensional isometric strength of neck muscles in humans. Spine. 2001; 26(17):1904–1909.
31. Westing SH, Cresswell AG, Thorstensson A. Muscle activation during maximal voluntary eccentric and concentric knee extension. Eur J Appl Physiol Occup Physiol. 1991; 62(2):104–108.