Posture and Helmet Configuration Effects on Joint Reaction Loads in the Middle Cervical Spine

Jeff M. Barrett; Colin D. McKinnon; Clark R. Dickerson; Andrew C. Laing; Jack P. Callaghan

INTRODUCTION: Between 43 and 97% of helicopter pilots in the Canadian Armed Forces report neck pain. Potential contributing factors include the weight of their helmet, night vision goggles (NVG), and counterweight (CW) combined with deviated neck postures. Therefore, the purpose of this investigation was to quantify changes in neck loads associated with posture, helmet, NVG, and CW.

METHODS: Eight male subjects volunteered. They undertook one of five deviated neck postures (flexion, extension, lateral bending, axial rotation) times four configurations (no helmet, helmet only, helmet and NVG, and helmet, NVG, and CW). 3D kinematics and EMG from 10 muscles (5 bilaterally) drove a 3D inverse dynamics, EMG-driven model of the cervical spine which calculated joint compression and shear at C5-C6.

RESULTS: The compression in the neutral posture was 116.5 (5.7) N, which increased to 143.7 (11.4) N due to a 12.7 N helmet. NVGs, weighing 7.9 N, also generated this disproportionate increase, where the compression was 164.2 (3.7) N. In flexion or extension, the compression increased with increasing head-supported mass, with a maximum of 315.8 (67.5) N with the CW in flexion. Anteroposterior shear was highest in the lateral bending [34.0 (6.2) N] condition, but was generally low (< 30 N). Mediolateral shear was less than 5 N for all conditions.

DISCUSSION: Repositioning the center of gravity of the helmet with either NVGs or CW resulted in posture-specific changes to loading. Posture demonstrated a greater potential to reposition the head segment's center of gravity compared to the helmet design. Therefore, helmet designs which consider repositioning the center of gravity may reduce loads in one posture, but likely exacerbate loading in other postures.

KEYWORDS: helicopter, cervical spine, chronic neck pain, musuloskeletal, compression, shear.

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The prevalence of chronic neck pain is alarmingly high among helicopter pilots in the Canadian Armed Forces, where between 43 and 97% of helicopter pilots report chronic neck pain.[12,24,38] Neck pain is partially attributable to mechanical risk factors and has been shown to be chronic and episodic, with sufferers rarely experiencing a full remission from their symptoms once initiated.[9] Accordingly, there is value in understanding neck pain's underlying biomechanical mechanisms toward developing preventive interventions.

For helicopter pilots in the Canadian Military, the risk of chronic neck pain increases when night vision goggles (NVG) are worn during night flights.[13,38] This issue is rooted in the anteriorly positioned mass of the NVG system relative to the vertebral column, inducing a flexion moment on the neck which the neck extensors need to combat.[24] One approach for addressing this issue has been to add a counterweight to the posterior aspect of the helmet to supplement the moment produced by the neck extensors. This solution can have positive effects, provided the operator maintains a neutral posture, as it decreases the moment generation required by the posterior musculature to offset the moment generated by the anterior location of the NVG.[37] However, deviated postures in helicopter pilots are commonplace.[24] This is in part due to the complexity in navigating and controlling a 6-degree of

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freedom aircraft, confounded by the reduced peripheral vision resulting from the NVG and cabin instrumentation.[13] Despite adoption of the counterweight in the Royal Canadian Air Force (RCAF), reports of chronic neck pain have not declined.[7,43]

Investigations into chronic neck pain, helmeted occupations, and electromyography (EMG) of both the superficial and deep musculature of the neck have found modest increases in muscle activity (2–3% maximum voluntary exertion) from donning NVGs.[29,37] However, these authors have yet to contextualize these findings with respect to direct mechanical exposures, such as cervical joint compression or shear forces. To date, there have been few investigations aiming to quantify the loads on the cervical spine with helmet conditions. Snijders et al.[35] used a simplified three-joint model of the cervical spine to explore changes in compression and shear in the upper and lower cervical spine in static postures. They concluded in a follow-up analysis that the loads on the cervical spine were sensitive to helmet design decisions, like the placement of the center of mass or weight of the helmet.[21] Mathys and Ferguson[26] simulated the effects of different helmet configurations in OpenSim using the model of Vasavada et al.[39] for jet fighters during high g force procedures. Using static optimization, they were able to show that the compressive forces in the lower cervical spine increase with added mass to the head, which is further increased in deviated postures and during high G maneuvers.

Ultimately, the purpose of this investigation was to quantify loads in the middle cervical spine in human volunteers as they vary with helmet configuration and posture. Based on previous findings, we hypothesized that joint reaction loads would increase disproportionately with the magnitude of head-supported mass, and that any decline in joint loads would be posture-specific.

METHODS

Subjects

Eight healthy men [mean (SD) age: 21.3 (1.7) yr; height: 177.9 (6.8) cm; body mass: 79.7 (11.5) kg] with no previous history of neck, shoulder, or upper back pain participated. Subjects provided informed consent and the study was approved by the University of Waterloo's institutional review board.

Materials

We used a 24-degree of freedom cervical spine model in OpenSim 4.1 (**Fig. 1**). It is equipped with passive elements representing the intervertebral discs and ligaments, and 216 active muscle elements representing 29 muscles bilaterally in the cervical spine.[6] It was designed to map the total head-trunk angle along with collected EMG channels for compression and shear forces at each vertebral joint in the cervical spine. The overall head-trunk angle was partitioned among the eight intervertebral joints (including the skull-C1 and C7-T12) in proportion to their range of motion. The linear enveloped EMG was used to drive a Hill-type muscle model to estimate the force in each muscle fascicle. An EMG-assisted optimization routine then adjusted each muscle force to ensure that there was agreement between the EMG-derived moment estimations and the net joint moments calculated from inverse dynamics.[8,14]

The helmet (Gentex HGU-56/P; Gentex, Carbondale, PA, USA) was represented as a separate segment in the neck model, securely fastened to the skull segment with a weld joint. For the normal helmet condition, the center of mass of the helmet was assumed to be coincident with the center of mass of the skull. The center of mass of the other helmet conditions were calculated from CAD drawings and represented as offsets from the standard helmet center of mass locations (**Table I**).

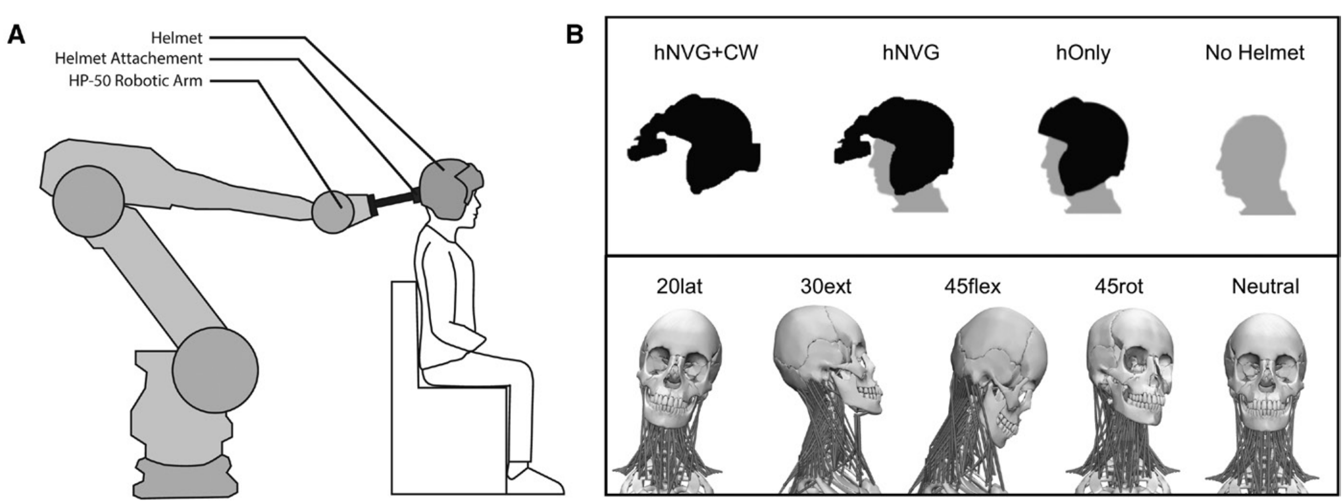

Fig. 1. A) Experimental set-up for MVE trials. The robotic arm was attached to the posterior aspect of the helmet and subjects performed exertions in flexion, extension, rotation, and lateral bending from this position. B) (Top row) The helmet conditions examined in this study. From right to left, No Helmet, helmet only (hOnly), helmet and night vision goggles (hNVG), and the helmet, night vision goggles, and counterweight (hNVG + CW). (Bottom row) The OpenSim cervical spine model demonstrating the five posture conditions used in the study. There is no representative geometry for the helmet, but it is represented mathematically.

Table I. Helmet Mass Properties for Each of the Helmet Conditions.

HELMET CONDITION	MASS (kg)	CENTER OF MASS OFFSET FROM SKULL CENTER OF MASS	
		ANTERIOR-POSTERIOR (MM)	SUPERIOR-INFERIOR (MM)
hOnly	1.296	0.0	0.0
hNVG	2.093	26.0	−9.0
hNVG + CW	2.723	−22.0	−14.0

hOnly: helmet only; hNVG: helmet and night vision goggles; hNVG + CW: helmet, night vision goggles, and counterweight.
For the hOnly condition, the center of mass was assumed to be coincident with the skull center of mass. Positive numbers represent anterior and superior displacements for anterior-posterior and superinferior positions. For comparison, the mass of the skull-segment is 4.6 kg.

Procedure

Following a brief warm up, subjects were fixed to a chair with Velcro straps to isolate cervical spine movement while wearing a Canadian Armed Forces CH146 Griffon helicopter helmet. The helmet was mounted on a robotic arm (HP50 with NX100 Controller, Yakasawa Motoman Robotics, Mississauga, Ontario, Canada). Maximum voluntary exertions (MVE) were performed isometrically against the resistance of the robot in three directions: pressing forward (flexion), backward (extension), or to the right (lateral bending). The protocol is illustrated in Fig. 1A, but the interested reader is directed to McKinnon et al.[29] for additional details.

Kinematic data from a passive motion capture system (Vicon MX, Vicon Motion Systems Ltd., Los Angeles, CA, USA) were sampled at 50 Hz from a total of nine markers. Markers from the xiphoid process, acromion process, and C7 spinous process were used to make the local coordinate system (LCS) for the trunk, while the markers over the ears and the most superior point on the helmet were used to make the LCS for the head/helmet system. The relative rotation between these two LCSs was used to quantify the overall head-trunk angle used as an input into the cervical spine model.[6] A self-selected neutral posture was obtained by instructing participants to "look straight ahead" during a seated calibration trial.

Following MVEs, the helmet was decoupled from the robotic arm and the experimental trials were completed. Four helmet conditions were tested (Fig. 1B): no helmet (No Helmet), helmet only (hOnly), helmet with the NVGs deployed (hNVG), and helmet with the counterweight and NVG (hCW + NVG). Starting from a neutral posture, subjects moved to one of five target postures, which they held for 15 s before returning to the neutral posture. The target postures were: 45° of flexion (45flex), 30° of extension (30ext), 45° of axial rotation (to the left), 20° of lateral bending (to the left), and a neutral posture (no movement) (Fig. 1B). Real time Euler-Cardan (XYZ sequence) angles were calculated in accordance to ISB standards,[42] and shown to the participant for kinematic feedback on their posture. These postures were selected as they tend to be some common ones undertaken by flight crew.[13]

Pregelled (2 cm interelectrode distance) Ag/AgCl surface EMG electrodes recorded the muscular activity of 10 muscles (5 bilaterally): splenius capitis, sternocleidomastoid, levator scapulae, the splenii and semispinalis (cervical erectors), and the upper trapezius. Briefly, the upper trapezius was palpated against light resistance to shrugging and the sternocleidomastoid to axial rotation. These two muscles form the borders to the neck's posterior triangle situated on the lateral aspect of the neck, where the superficial portions of splenius capitis and levator scapulae are accessible for surface EMG. Finally, cervical erector electrodes were placed roughly 2 cm on either side of the midline at approximately the C2 level. The interested reader is referred to more detailed documentation on this protocol.[23,33] A 16-channel Noraxon Telemyo 2400T G2 Telemetry electromyography system (Noraxon USA Inc., Scottsdale, AZ, USA) amplified the signals which were sampled at 1500 Hz through a 16-bit analog to digital card. The resulting signal was linear-enveloped by de-trending, full-wave rectifying, and low-pass filtering with a single pass low-pass critically damped filter with cutoff frequencies chosen to mimic the electromechanical delay of the underlying muscle, which were adapted from Almosnino et al.[2] (**Table II**). Muscles with no documented electromechanical delay were assigned a cutoff frequency of 4 Hz, as this has been used in previous cervical spine investigation.[25] Each muscle was normalized to the maximum from the MVE trials.

Table II. Electromechanical Delays That Were Available for the Model Adapted from Almosnino et al.[2]

MUSCLE	ELECTROMECHANICAL DELAY (ms)	CUTOFF FREQUENCY (Hz)
Splenius Capitis	32.2 (5.1)	4.9
Upper Trapezius	38.1 (3.3)	4.2
Sternocleidomastoid	70.3 (4.6)	2.3
Other muscles	N/A	4.0

Statistical Analysis

Average joint compression, anteroposterior shear (positive indicating posterior shear) and mediolateral shear (positive indicating to the left), and the flexion-extension moment at C5-C6 were obtained over the middle 2 s of the trial, when the subject was holding the posture. The C5-C6 level was chosen as it has been reported to be the most common site for disc injury in the cervical spine.[27]

A general linear model with repeated posture and helmet factors (5 postures, 4 helmet conditions with interaction effects) was used to evaluate the effect that each had on the compression and shear forces at C5-C6 (R version 4.0, the R Foundation). An a priori level of significance was set at 0.05, with a Tukey HSD pairwise *t*-test for post hoc analyses. The flexion-extension moment at this joint level was analyzed for the neutral, 30ext, and 45flex conditions using a repeated measures general linear model with helmet and posture (3 posture conditions, 4 helmet conditions with interaction effects).

RESULTS

There was a significant interaction effect between posture and helmet configuration (Wald χ^2: 25.388; df = 12; $P = 0.01309$). Deviated postures tended to increase compression, but the effect was not consistent across helmet conditions, indicating a

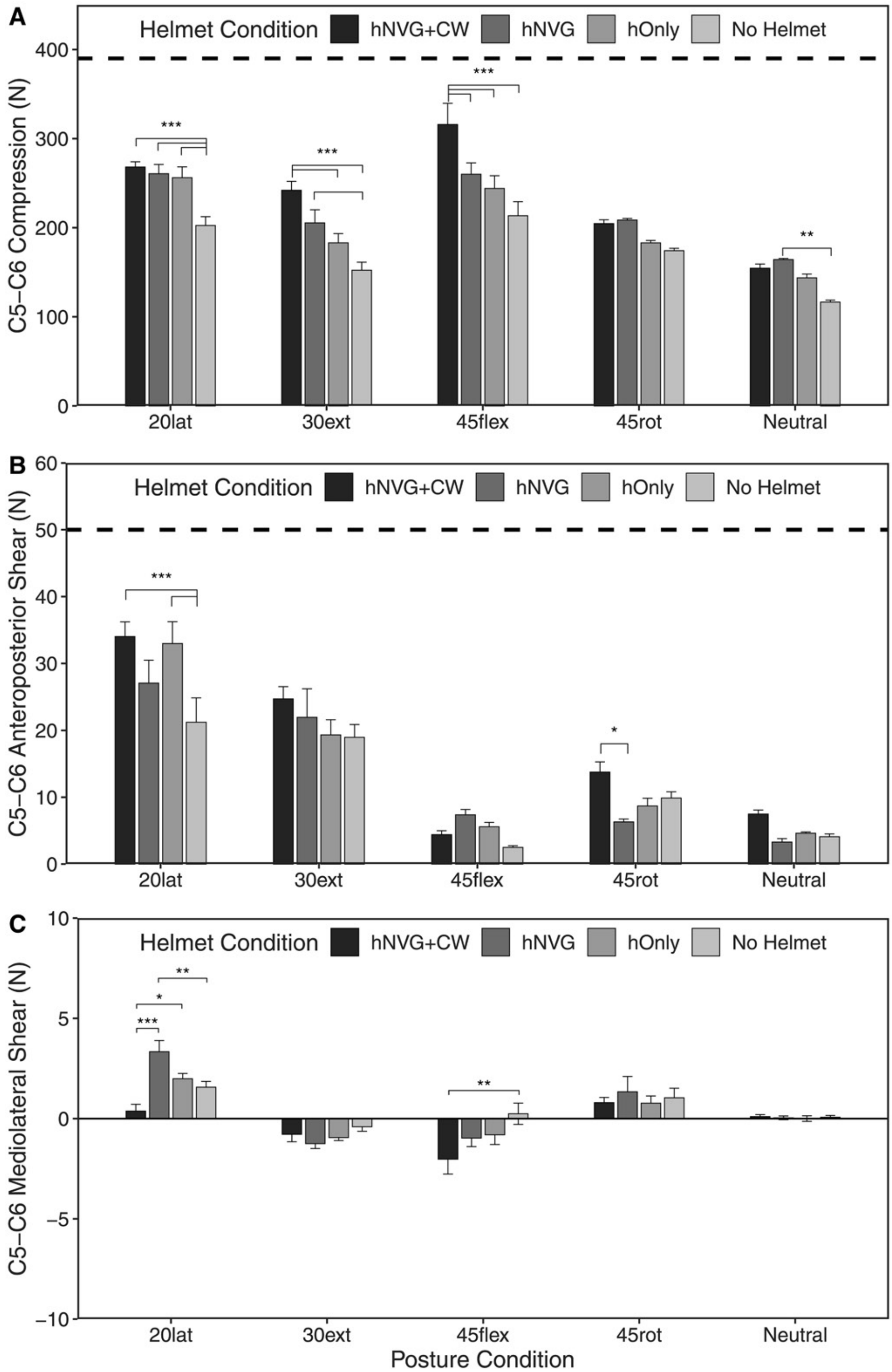

Fig. 2. Joint reaction forces across all postures and helmet conditions. A single asterisk denotes significant differences at $P < 0.05$, double asterisks $P < 0.01$, and triple asterisks $P < 0.001$. A) Compression. The dashed horizontal line indicates 10% of the estimated compressive tolerance (390 N) of the C5-C6 functional spinal unit (3.9 kN). B) Anteroposterior shear acting on C5. These can be thought of as the required anteriorly directed force sustained by the joint to prevent C5 from accelerating posteriorly. The bold dashed line (50 N) indicates 10% of the estimated shear tolerance (500 N) of the C5-C6 functional spinal unit. C) Mediolateral shear.

nonlinear relationship between added head mass and joint compression (**Fig. 2A**). Compression was highest for the 45flex posture with both night vision goggles and counterweight (315.8 ± 67.5 N), which was significantly higher than the night vision goggles (260.0 ± 33.4 N; Cohen's $d = 1.89$), helmet (244.1 ± 40.4 N; Cohen's $d = 2.43$), and no helmet conditions (213 ± 44.7 N; Cohen's $d = 3.47$) for that posture ($P < 0.05$). A similar trend was observed in the extended posture, where the counterweight

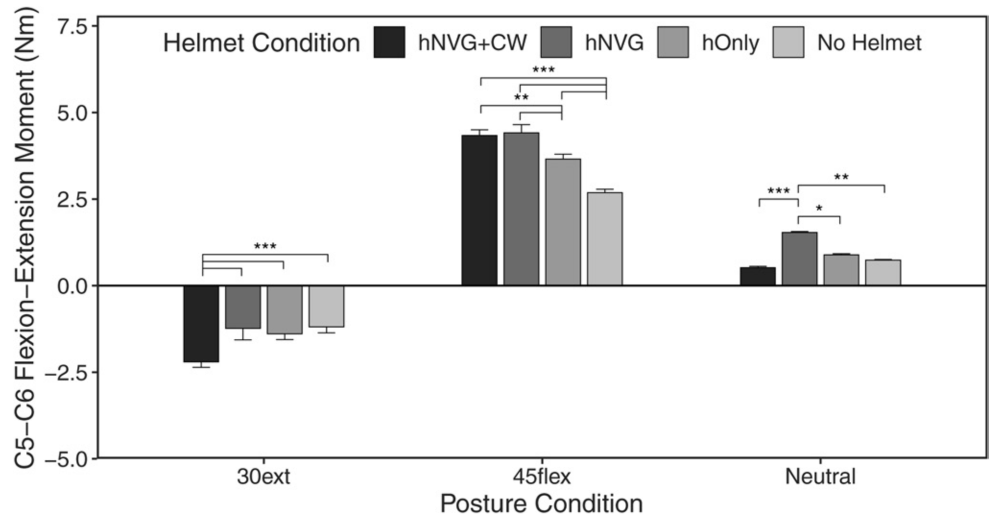

Fig. 3. Changes in the flexion-extension moment with posture and helmet condition. Significance is annotated as indicated in Fig. 1, with a single asterisk denoting $P < 0.05$, double asterisks $P < 0.01$, and triple asterisks $P < 0.001$.

(242.1 ± 26.2 N) resulted in significantly more compression compared to the helmet (183.0 ± 24.1 N; Cohen's $d = 1.99$) and no helmet conditions (152.2 ± 24.1 N; Cohen's $d = 3.05$). For extension, there was a trend for the counterweight to increase compression compared to the night vision goggles (205.4 ± 41.7 N), but this trend was not statistically significant ($P = 0.082$). In lateral bending, all helmet conditions resulted in increased compression from no helmet (202.5 ± 28.2 N) by an average of 58.8 N (Cohen's $d = 1.82$, 1.98, 2.23 for hOnly, hNVG, and hNVG+CW, respectively). In a neutral posture, the compression from the night vision goggles (164.2 ± 3.7 N) was significantly greater than the no helmet condition (116.6 ± 5.7 N; Cohen's $d = 1.62$).

There was a significant interaction effect between posture and helmet condition on anteroposterior shear (Wald $\chi^2 = 26.14$, $df = 12$, $P = 0.01026$). In all cases, the net force acting on the joint was directed posteriorly, indicated by the positive forces in **Fig. 2B**. In a neutral posture, anteroposterior shear is on average 4.9 ± 1.2 N, which was similar to a flexed posture (5.0 ± 1.6 N). In the lateral bending condition, both the helmet only and NVG+CW conditions resulted in significantly more posterior shear (32.9 ± 9.2 N and 34.0 ± 6.2 N, Cohen's $d = 0.40$ and 0.43, respectively) compared to the no-helmet condition (21.2 ± 10.3 N).

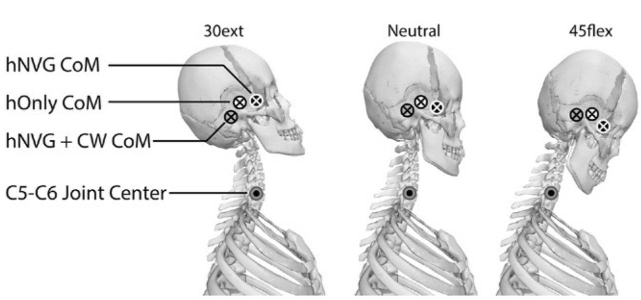

Fig. 4. Center of mass (CoM) positions for the helmet only condition (hOnly) compared to the night vision goggles (hNVG) without and with (hNVG + CW) conditions.

Like the previous two outcome measures, there was a significant interaction effect between posture and helmet condition on mediolateral shear (Wald $\chi^2 = 38.3$, $df = 12$, $P = 0.00014$; **Fig. 2C**). In a neutral posture, for all helmet conditions, the mediolateral shear was almost identically zero (within 1 N). In the lateral bending condition, the counterweight significantly reduced the magnitude of mediolateral shear from 3.3 ± 1.6 N to 0.4 ± 1.0 N compared to the NVG condition (Cohen's $d = -2.79$). Overall, the magnitudes of mediolateral shear tended to be low, below 10 N.

The flexion moment also exhibited a significant interaction effect between helmet and posture configuration (Wald $\chi^2 = 87.3$, $df = 6$, $P < 2 \times 10^{-16}$). The night vision goggles induced a 0.83 ± 0.15 Nm increase in the flexion moment over the helmet alone (0.89 ± 0.08 Nm; Cohen's $d = 1.87$) in a neutral posture, which was entirely negated by the adoption of the counterweight (**Fig. 3**). In a flexed posture, the NVGs and counterweight induced a larger flexion moment (4.4 ± 0.6 Nm and 4.3 ± 0.5 Nm, Cohen's $d = 1.85$ and 1.62, respectively) over just the helmet alone (3.7 ± 0.4 Nm). The counterweight accounted for the largest change in extensor moment in extension, amounting to −2.2 ± 0.4 Nm compared to no helmet (−1.2 ± 0.6 Nm, Cohen's $d = 2.51$).

DISCUSSION

This study examined changes in the magnitude of joint reaction forces in the middle cervical spine with respect to postural and helmet configurations. The aim was to quantify the loads experienced by helicopter pilots in deviated postures while wearing a helmet, and night vision goggles with and without a counterweight. Overall, we found a nonlinear relationship between head-supported mass and resulting joint compression that interacted with posture. Added head-supported mass led to a disproportionate increase in joint loads. For instance, donning a 12.7 N helmet resulted in an increase of 27.2 N of compression—more than twice the weight of the helmet. This effect compounds with posture, for example, the largest magnitude of compression was with the NVG and counterweight system in flexion for a grand total of 315.8 N.

This investigation highlights that posture has a substantial influence over the joint loads, more so than the helmet center of mass location. To understand why, consider the displacement of the helmet's center of gravity with the addition of night vision goggles and with the counterweight (Table I). These amount to 2–3 cm of either anterior or posterior displacement, in a neutral posture, relative to the C5-C6 joint center. For comparison, the same anterior-posterior displacements can be achieved with only 15° of flexion or extension, respectively. Therefore, for the helmet designs explored in this investigation, there exists postures for which the helmet center of mass is displaced to a greater extent than the initial helmet design. More to this point, in a neutral posture the night vision goggle weight induced a 0.83 Nm neck flexor moment which vanished in 30° of extension (Fig. 3). Conversely, in line with its design philosophy, the helmet-counterweight system induced an extensor moment in a neutral posture. In 45° of flexion, however, the center of mass of the system moves anterior to the C5/6 joint center (**Fig. 4**), inducing a 4.4 Nm flexor moment, the same moment as the NVG condition. This is due to a tradeoff between increased helmet weight and decreased moment arm in this posture. However, for extension, the addition of the counterweight increased both the moment arm to the center of gravity of the helmet and the weight of the helmet system, resulting in a 2.2 Nm extensor moment. These moment changes translated to increased compression at this joint level with the addition of night vision goggles or the counterweight in both flexion and extension. This result highlights a challenging issue in helmet design: the mobility of the human neck. Helmet designs based on repositioning the center of mass of the helmet system may reduce the gravitational moment in one posture but may exacerbate it in another.

Of the postures evaluated, a neutral position elicited the smallest joint reaction loads, approximately 116.5 ± 5.6 N of compression and 4.1 ± 1.1 N of posterior shear. Addition of the helmet, of weight 12.7 N, correspondingly increased the compression to 143.7 ± 4.3 N; a difference more than twice the actual weight of the helmet. This amplified increase in compression is due to muscle activity required to balance the gravitational moment. To this point, the NVG system, an added weight of 7.8 N, increased the compression further to 164.2 ± 1.3 N, a difference almost three times the weight of the NVGs. Counterintuitively, addition of a 6.2-N counterweight decreased the compression to 154.5 ± 4.6 N, simply due to the repositioning of the center of mass. These trends are similar in the lateral bending conditions, where donning a helmet with or without NVGs significantly increased joint compression from 202.4 ± 9.4 N to 261.3 ± 12.2 N. Overall, deviations from a neutral posture correspondingly increased the joint load, both in terms of compression and shear. This is in agreement with previous studies of helmeted individuals, where a more flexed posture has been repeatedly associated with a larger restorative moment requirement.[13,35,37] Interestingly, cervical spine flexion, a known risk factor for the development of chronic neck pain,[3,4] resulted in the largest magnitudes of compression: 315.8 N with a counterweight and night vision goggles—the highest compression examined in this study. This compression was 102.3 N higher than the no-helmet condition, nearly four times the magnitude of the total added weight of 26.7 N. Coincidentally, pilots frequently adopt flexed postures when they are scanning the ground for possible obstructions or the instrumentation clusters.[24,38]

The results of this study agree well with other literature on the topic. Since the counterweight did not reduce the mechanical parameters evaluated here, it is not surprising that its adoption into the military did not result in a corresponding decrease in reports of neck pain between 2004 and 2014.[7,43] Forde et al.[13] used a simple inverse dynamics model to estimate the cumulative change in joint reaction forces and moments in helicopter pilots over flights. They found that pilots wearing NVG spent significantly more time in a flexed posture than those without, which translated into cumulative load values that were significantly higher on night flights than day flights. Similarly, in this study, a flexed posture was generally associated with greater loads that were magnified with the inclusion of NVG and counterweight. Over an entire night flight this would have manifested itself as an increase in all measures of cumulative load exposure. The compressive forces reported here with the NVG deployed are three times that reported in Forde et al.,[13] while the anteroposterior and mediolateral shear values are slightly lower; this is attributable to the inclusion of muscular forces in this model.

Dibblee et al.[11] designed and tested a novel spring at the back of the helmet to produce a supplementary extension moment without increasing the mass of the head-system. They found a significant decrease in measures of fatigue, decreased median power frequency, and ratings of perceived exertion when their intervention was used, and an increase in these measures when the counterweight was used over just the helmet alone. However, these investigators did not measure joint mechanics in their evaluation. Intuitively, the loading scenario of a posteriorly mounted pretensioned spring should be similar to the counterweight as both intend to provide a downward force at the head to counter the weight of the NVG. Future investigations would be needed to determine how this device changes joint mechanics. The compression and shear values reported here in a nonhelmeted condition agree very closely with an early model by Snijders et al.[35] This was a relatively basic model of the cervical spine, partitioning it into three segments and including only 35 muscle elements, compared to the 218 used in this study. They later applied their model to the ergonomics of F-16 pilots in the Dutch Air Force and came to a similar conclusion: the helmet mass contributes significantly to the compression and shear loads in the cervical spine.[21]

It should be noted that the magnitudes of compression and shear observed here are only 10% of what is required to acutely fail a functional spinal unit: it takes 3.9 kN of compression or 500 N of shear for failure to occur.[32,34] Because these loads are so low in an acute scenario, it is rational to suspect that one source of the pain is muscular. Indeed, reducing muscular demands seems to be the current direction the Canadian Armed Forces is taking.[43] However, muscular pain would not explain why reports of neck pain persisted after the introduction of the counterweight, as the counterweight has been shown to reduce the metabolic demand of the neck extensors.[1,20] If the joint loads are responsible for the chronic neck pain experienced by most helicopter pilots, then it obviously implicates a mechanism of cumulative load rather than an acute loading event. This chronic pathway also explains why RCAF pilots with more flight time report more neck pain.[1] Gooyers and Callaghan[16] used an in-vitro porcine cervical spine model to show that loads as low as 10% of the ultimate compressive tolerance are enough to significantly alter the surface geometry of the intervertebral disc in response to a 0.5-Hz loading regime in a flexed posture over a 2-h loading protocol. Coincidentally, compression in helmeted conditions and flexed postures are close to the 10% of failure strength (Fig. 2), potentially implicating the intervertebral disc as a source of pain in helicopter pilots. Future studies are necessary to fully determine if these low-magnitude loads sustained over an extended duration are sufficient for the development of osteoligamentous injury, or if the reduction of load in the helmet is an effective means of mitigating the pain experienced by helicopter pilots.

Chronic neck pain is a multifaceted issue attributable to psychosocial, physiological, and mechanical factors.[10,20,31] Nevertheless, this study documented that the presence of a helmet and NVG significantly alter the mechanical exposure of helicopter pilots, which the counterweight fails to address comprehensively across the range of postures adopted during flight.

The most significant limitation in this investigation was the omission of whole-body vibration (WBV), which helicopter aircrew endure. Exposure to WBV, especially near the resonant frequency of the spine, is very strongly associated with the development of low back and neck pain[30,40,41] and microscopic tissue damage.[18] Further, the posture of operators, an attribute explored extensively in this investigation, is known to influence the mechanical response of individual functional spinal units to vibration.[5,17] Additionally, work with in vivo participants has shown that individuals exposed to WBV co-contract their spinal musculature, which stiffens and stabilizes their spines;[19] naturally, the elevated muscle activity would likely increase the joint loads. Theoretically, since the vibration modes of a mechanical system are related to both its mass and stiffness matrices,[15] we hypothesize that the magnitude of head-supported mass from a helmet would interact with the stiffness from co-contraction to alter the cervical spine's vibration behavior. Unfortunately, this analysis fell outside of the scope of the current investigation.

The study was carried out on a healthy male population, which may not be representative of the helicopter pilots who work for the Canadian military. Additionally, subjects carried out static postural holds which are also likely not representative of the ballistic movements helicopter pilots need to undertake as they control their aircraft, nor all the static postures assumed by operators. Combined flexion with axial rotation, for example, is also a very common posture which was not explored in this investigation.[13,36] These intermittent bursts during dynamic tasks would plausibly produce compression and shear forces greater than those reported here, and it is unclear how the presence of the counterweight would either mitigate or exacerbate those forces. Similarly, subjects replicated these postures in a controlled laboratory environment, which is different from the noninertial reference frame that is the cockpit of an active CH-146. Finally, no attempt was made to quantify the resulting inertial forces that would be acting on the head: the Coriolis, centrifugal, or Euler forces, which are proportional to the mass of the head and helmet system.

The model suffers from limitations as well. In terms of kinematics, the model linearly partitions the overall head-neck angle among the intervertebral joints. This is a necessary model assumption but simplifies the number of different configurations the cervical spine can adopt while maintaining the same external head-neck angle.[28,39] In terms of kinetics, there are a number of limitations germane to the Hill-type muscle model, most notably its failure to characterize force production capacity of submaximal eccentric contractions.[22,44]

In conclusion, this study quantified the loads on the middle cervical spine in a variety of posture and helmet conditions. We hypothesized that joint reaction loads would increase proportionately with the magnitude of head-supported mass, and the proportionality would be posture-specific. Supporting our hypothesis, compression at C5-C6 increased proportionately with the weight of the helmet and night vision goggles. For instance, donning a 12.7 N helmet correspondingly increased the compression by 27.2 N. Further, addition of the NVG system, an added weight of 7.8 N, increased the compression another 20.5 N. This amplification can be exacerbated by posture; for instance, a flexed posture with helmet, NVG, and counterweight systems with a total weight of 26.7 N, exhibited a 102.3 N increase in compression compared to the same posture without a helmet. These findings supported our hypothesis that repositioning the center of gravity leads to posture-specific responses, with beneficial responses in neutral head-neck positions, but detrimental loading in nonneutral postures. This result may be of interest to designers of helmets, night vision goggles, and counterweights. The increased moment demands generally translated into greater compression and anteroposterior shear. For every static posture and helmet conditions, the compression and anteroposterior shear magnitudes were generally lower than 10% of the ultimate tolerance, which we hypothesize to implicate a cumulative loading pathway for the development of chronic neck pain.

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REFERENCES

1. Adam J. Results of NVG-induced neck strain questionnaire study in CH-146 Griffon aircrew. Toronto (Canada): DRDC Toronto; 2004.
2. Almosnino S, Pelland L, Pedlow SV, Stevenson JM. Between-day reliability of electromechanical delay of selected neck muscles during performance of maximal isometric efforts. Sports Med Arthrosc Rehabil Ther Technol. 2009; 1(1):22.
3. Ariëns GA, Bongers PM, Douwes M, Miedema MC, Hoogendoorn WE, et al. Are neck flexion, neck rotation, and sitting at work risk factors for neck pain? Results of a prospective cohort study. Occup Environ Med. 2001; 58(3):200–207.
4. Ariëns GA, van Mechelen W, Bongers PM, Bouter LM, van der Wal G. Physical risk factors for neck pain. Scand J Work Environ Health. 2000; 26(1):7–19.
5. Barrett JM, Gooyers CE, Karakolis T, Callaghan JP. The impact of posture on the mechanical properties of a functional spinal unit during cyclic compressive loading. J Biomech Eng. 2016; 138(8):081007.
6. Barrett JM, McKinnon CD, Dickerson CR, Callaghan JP. An electromyographically driven cervical spine model in OpenSim. J Appl Biomech. 2021; 37(5):481–493.
7. Chafé G, Farrell PSE. Royal Canadian Air Force CH-146 Griffon aircrew 2014 spinal musculoskeletal trouble survey. Toronto, Ontario (Canada): Defence Research and Development Canada; 2016.
8. Cholewicki J, McGill SM. EMG assisted optimization: a hybrid approach for estimating muscle forces in an indeterminate biomechanical model. J Biomech. 1994; 27(10):1287–1289.
9. Côté P, Cassidy DJ, Carroll LJ, Kristman V. The annual incidence and course of neck pain in the general population: a population-based cohort study. Pain. 2004; 112(3):267–273.
10. Côté P, Kristman V, Vidmar M, Van Eerd D, Hogg-Johnson S, Beaton D, Smith PM. The prevalence and incidence of work absenteeism involving neck pain. A cohort of Ontario lost-time claimants. J Manipulative Physiol Ther. 2009; 32(2, Suppl.):S219–S226.
11. Dibblee J, Worthy P, Farrell P, Hetzler M, Reid S, et al. Evaluating a prototype device designed to alleviate night vision goggle induced neck strain among military personnel. Ergonomics. 2015; 58(12):2067–2077.
12. Farrell PSE, Shender BS, Goff CP, Crowley JBJ, Davies M, et al. Aircrew neck pain prevention and management. Brussels (Belgium): North Atlantic Treaty Organization; 2020.
13. Forde KA, Albert WJ, Harrison MF, Neary JP, Croll J, Callaghan JP. Neck loads and posture exposure of helicopter pilots during simulated day and night flights. Int J Ind Ergon. 2011; 41(2):128–135.
14. Gagnon D, Arjmand N, Plamondon A, Shirazi-Adl A, Larivière C. An improved multi-joint EMG-assisted optimization approach to estimate joint and muscle forces in a musculoskeletal model of the lumbar spine. J Biomech. 2011; 44(8):1521–1529.
15. Goldstein H, Poole C, Safko J. Classical mechanics, third edition. Flushing (NY): Addison-Wesely; 1950.
16. Gooyers CE, Callaghan JP. Exploring interactions between force, repetition and posture on intervertebral disc height loss and bulging in isolated porcine cervical functional spinal units from sub-acute-failure magnitudes of cyclic compressive loading. J Biomech. 2015; 48(13): 3701–3708.
17. Gooyers CE, McMillan RD, Howarth SJ, Callaghan JP. The impact of posture and prolonged cyclic compressive loading on vertebral joint mechanics. Spine. 2012; 37(17):E1023–E1029.
18. Gregory DE, Callaghan JP. Does vibration influence the initiation of intervertebral disc herniation? An examination of herniation occurrence using a porcine cervical disc model. Spine (Phila Pa 1976). 2011; 36(4):E225–E231.
19. Hansson T, Magnusson M, Broman H. Back muscle fatigue and seated whole body vibrations: an experimental study in man. Clin Biomech (Bristol, Avon). 1991; 6(3):173–178.
20. Harrison MF, Neary JP, Albert WJ, Veillette MDW, Forcest C, et al. Physiological effects of night vision goggle counterweights on neck musculature of military helicopter pilots. Mil Med. 2007; 172(8): 864–870.
21. 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.
22. Joyce GC, Rack PM. Isotonic lengthening and shortening movements of cat soleus muscle. J Physiol. 1969; 204(2):475–491.
23. Keshner EA, Campbell D, Katz RT, Peterson BW. Neck muscle activation patterns in humans during isometric head stabilization. Exp Brain Res. 1989; 75(2):335–344.
24. 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.
25. Lu WW, Bishop PJ. Electromyographic activity of the cervical musculature during dynamic lateral bending. Spine. 1996; 21(21):2443–2449 .
26. 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.
27. Matsumoto M, Fujimura Y, Suzuki N, Nishi Y, Nakamura M, et al. MRI of cervical intervertebral discs in asymptomatic subjects. J Bone Joint Surg Br. 1998; 80-B(1):19–24.
28. McGill SM, Norman RW. Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine. 1986; 11(7):666–678.
29. McKinnon CD, Dickerson CR, Laing ACT, Callaghan JP. Neck muscle activity during simulated in-flight static neck postures and helmet mounted equipment. Occup Ergon. 2016; 13(3–4):119–130.
30. Paddan GS, Griffin MJ. The transmission of translational seat vibration to the head. I. Vertical seat vibration. J Biomech. 1988; 21(3):191–197.
31. Pelham TW, White H, Holt LE, Lee SW. The etiology of low back pain in military helicopter aviators: prevention and treatment. Work. 2005; 24(2):101–110.
32. Pintar FA, Yoganandan N, Voo L. Effect of age and loading rate on human cervical spine injury threshold. Spine (Phila Pa 1976). 1998; 23(18):1957–1962.
33. Schüldt K, Ekholm J, Harms-Ringdahl K, Németh G, Arborelius UP. Effects of changes in sitting work posture on static neck and shoulder muscle activity. Ergonomics. 1986; 29(12):1525–1537.
34. Shea M, Edwards WT, White AA, Hayes WC. Variations of stiffness and strength along the human cervical spine. J Biomech. 1991; 24(2):95–107.
35. 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.
36. Tack DW, Bray-Miners J, Nakaza ET, Osborne A, Mangan B. Griffon helicopter neck strain project. Toronto, Ontario (Canada): DRDC Toronto; 2014.
37. Thuresson M, Äng B, Linder J, Harms-Ringdahl K. Mechanical load and EMG activity in the neck induced by different head-worn equipment and neck postures. Int J Ind Ergon. 2005; 35(1):13–18.
38. van den Oord MHAH, De Loose V, Meeuwsen T, Sluiter JK, Frings-Dresen MHW. Neck pain in military helicopter pilots: prevalence and associated factors. Mil Med. 2010; 175(1):55–60.
39. Vasavada AN, Li S, Delp SL. Influence of muscle morphometry and moment arms on the moment-generating capacity of human neck muscles. Spine. 1998; 23(4):412–422.

40. Wei L, Griffin MJ. Mathematical models for the apparent mass of the seated human body exposed to vertical vibration. J Sound Vibrat. 1998; 212(5):855–874.
41. Wilder DG, Pope MH, Frymoyer JW. The biomechanics of lumbar disc herniation and the effect of overload and instability. J Spinal Disord. 1988; 1(1):16–32.
42. Wu G, Siegler S, Allard P, Kirtley C, Leardini A, et al. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion. Part I: ankle, hip, and spine. J Biomech. 2002; 35(4):543–548.
43. Xiao A, Farrell PSE. Royal Canadian Air Force neck-and back-trouble research. Toronto, Ontario (Canada): DRDC Toronto; 2016.
44. Zahalak G. Modeling muscle mechanics (and energetics). In: Winters JM, Woo SL-Y, editors. Multiple muscle systems. Biomechanics and movement organization. New York: Springer Verlag; 1990: 1–23.