Night Vision Goggle and Counterweight Use Affect Neck Muscle Activity During Reciprocal Scanning

Laura A. Healey; Aaron J. Derouin; Jack P. Callaghan; Duane S. Cronin; Steven L. Fischer

BACKGROUND: Mass, moment of inertia, and amplitude of neck motion were altered during a reciprocal scanning task to investigate how night vision goggles (NVGs) use mechanistically is associated with neck trouble among rotary-wing aircrew.

METHODS: There were 30 subjects measured while scanning between targets at 2 amplitudes (near and far) and under 4 head supported mass conditions (combinations of helmet, NVGs, and counterweights). Electromyography (EMG) was measured bilaterally from the sternocleidomastoid and upper neck extensors. Kinematics were measured from the trunk and head.

RESULTS: Scanning between the far amplitude targets required higher peak angular accelerations (∼7% increase) and neck EMG (between 1.2–4.5% increase), lower muscle cocontraction ratios (∼6.7% decrease), and fewer gaps in EMG (up to a 59% decrease) relative to the near targets. Increasing the mass of the helmet had modest effects on neck EMG, while increasing the moment of inertia did not.

DISCUSSION: Target amplitude, not head supported mass configuration, had a greater effect on exposure metrics. Use of NVGs restricts field-of-view, requiring an increased amplitude of neck movement. This may play an important role in understanding links between neck trouble and NVG use.

KEYWORDS: Muscle, neck pain, rotary-wing, biomechanics, injury.

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Forward positioned head supported technologies, such as night vision or virtual reality goggles, continue to emerge to enhance and augment human vision capabilities. However, the effect that forward added mass has on neck function is not well understood from a biomechanical perspective. For rotary wing aircrew, the addition of night vision goggles (NVGs) is widely accepted as a key contributor to neck trouble[1,17,22] but dissent remains about why. Following the helmet, NVGs are the most commonly used head supported device and are critical for mission effectiveness, however, their use appears to come at a large cost for pilots. In fact, lifetime prevalence of neck pain is 75% for CH 146 Griffon helicopter aircrew,[1] where neck strain may affect 90% of aircrew logging at least 150 h of night flying[1] and where NVG users experiencing a 45% greater chance of head and neck injury compared with non-NVG users.[28] Note that in this paper, 'neck trouble' encompasses discomfort, injury, acute pain or chronic pain.[10]

Despite the large body of evidence suggesting that NVGs have detrimental effects, little is understood about the pathway by which added head supported mass (HSM), in the form of NVGs, may link to underlying injury. Nevertheless, interventions, such as counterweights (CW), have emerged to reduce neck trouble in pilots,[18] without underlying scientific evidence about their effectiveness. A CW is believed to be beneficial as it can counterbalance the forward weight of the NVGs[18] in a neutral static position; however, the effectiveness of this approach is likely limited.[9,23] While in theory a simple CW solution would be effective in balancing the flexion moment caused by NVGs when seated upright and static, in reality, pilots are rarely static and must move their heads through a wide range of postures about all three rotation axes. Importantly, a CW increases the mass and moment of inertia of the helmet

From the Department of Kinesiology, University of Waterloo, Waterloo, Canada.

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Address correspondence to Steven L. Fischer, Ph.D., Associate Professor, Department of Kinesiology, Faculty of Applied Health Sciences, 200 University Avenue West, Waterloo, Ontario N2L 3G1, Canada; steven.fischer@uwaterloo.ca. ORCiD ID https://orcid.org/0000-0002-3347-5403.

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system. Because the pathway by which NVGs may lead to injury are poorly understood, it is not known if a CW is an effective solution, or if in fact it could contribute to neck pain in aircrew.

In order to inform effective interventions, it is important to understand the underlying exposure pathways by which NVG use might relate to injury. We propose a conceptual model to investigate potential injury pathways (**Fig. 1**), which consists of three primary elements: increased mass, increased range of motion, and increased moment of inertia. First, the use of NVGs adds mass, which causes a destabilizing force that requires additional muscle force to support. As a result, it is plausible that the neck extensor muscles would increase their activation, cocontracting with the flexor muscles to stabilize the head in addition to balancing the forward positioned mass. Such an increase and sustained contraction could lead to muscle fatigue. Hagg[14] suggested that, due to an orderly recruitment of muscle fibers, Type I fibers are often recruited first and remain active throughout prolonged contractions, even at a low level. As a result of being the first muscle fibers to work, and the last to turn off, this hypothesis is referred to as the Cinderella Hypothesis (first to work, last to bed).[14] In summary, increased HSM may cause an increase in muscle activation as required to stabilize against the added mass, increasing the number of Type I fibers recruited and forced to sustain a contraction, supporting the plausibility of neck trouble consistent with the well-established Cinderella hypothesis.

Alternatively, NVGs reduce the pilot's field of vision from 140° to 40°,[5] thus requiring pilots to rotate the head through a larger range of motion to actively scan the same visual field. As a result, more extreme neck postures are required of pilots when using NVGs relative to when flying without NVGs.[12] At end ranges of motion, muscle moment arms are altered, and muscle fascicles are lengthened, reducing their force production capability and putting them at a mechanical disadvantage along the force-length curve. Additionally, as a result of more extreme postures when wearing NVGs, there may be an increased relative muscular demand, as more work (the product of torque and angular displacement) is required to move the head when the field-of-view is restricted by NVGs. The combination of increased muscle demand, coupled with a decreased ability to generate force, create a scenario where it is plausible that NVGs can increase the likelihood of suffering from an overexertion injury.[12,30]

Finally, the added mass and location of the NVGs increases the moment of inertia of the head-helmet system. The increased moment of inertia of the system increases the resistance to angular motion, where it is likely that higher magnitude muscle forces would be required to stop and start head motion. Increased muscle forces result in increased stress (force per unit area), which over time, may exceed tissue tolerance and possibly result in an overexertion injury or pain due to tissue damage.[21]

The effect of NVGs and CWs on the development of neck trouble is likely multifactorial, making injury pathways difficult to understand. However, isolating potential pathways will provide insight to prioritize what factors may be most important to consider when designing interventions that will mitigate flight related neck trouble. Therefore, the objective of this study was to probe how increased mass, moment of inertia, and range of motion interdependently affect neck function using a novel visual target acquisition task. We

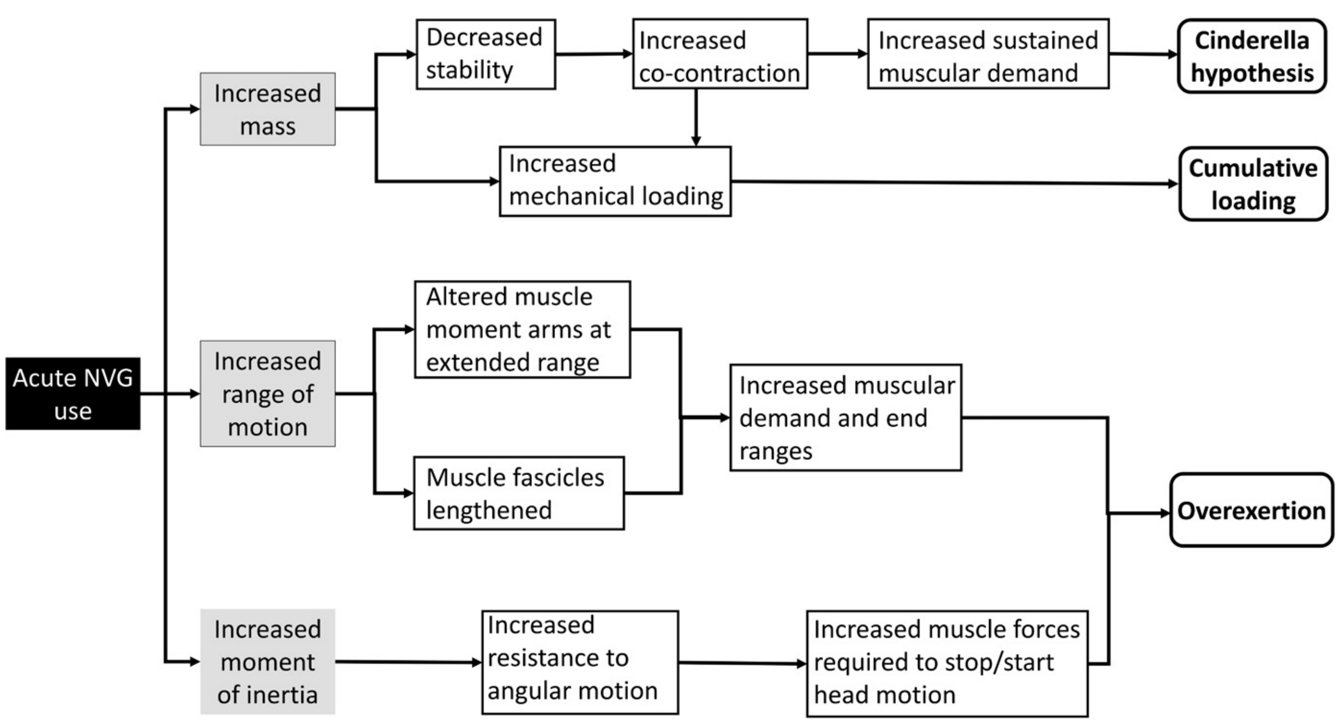


Fig. 1. A conceptual model demonstrating potential pathways of injury caused by NVGs.

hypothesized the following: 1) that increasing mass would increase cocontraction and decrease the number of muscular rest/gaps; 2) that increasing range of motion would increase muscular demand; and 3) that increased moment of inertia would increase peak muscle activity to stop and start head motion.

METHODS

Subjects

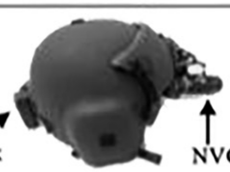
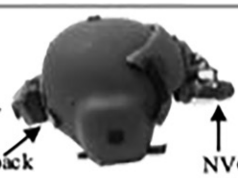
There were 15 male and 15 female subjects ($N = 30$) recruited from a university population (mean ± SD; age: 24 ± 4 yr, height: 144.7 ± 9.7 cm, and body mass: 79.4 ± 18.6 kg). Subjects were excluded if they had any previous history of neck pain, neck injury, concussions, vertigo or dizziness, or fainting during exercise. This study was approved by the University of Waterloo ethics committee, and all subjects gave written informed consent prior to participating in this study.

Equipment and Materials

During all experimental trials, subjects wore a properly sized Gentex HGU-56/P with a head mounted laser pointer. The helmet configuration was manipulated to represent one of four operationally relevant HSM configurations (helmet alone (hOnly); helmet and NVGs (hNVG); helmet, NVGs and traditional CW (hCW); and helmet, NVGs, and a CW liner (hCWL)) (**Table I**). In the hOnly (baseline) condition, light-weight 3D printed tubes (0.04kg) were used to constrain the field of view representative of NVG, without the additional weight of real NVGs. The hNVG and hCW represented operationally relevant configurations, where hNVG caused a forward shift in head-helmet center of mass altering the moment of inertia, and where hCW added additional mass (0.66 kg) posteriorly on the helmet in an effort to restore a more neutral center of mass, again, altering moment of inertia. The hCWL condition was used as an experimental condition that maintained the weight of the hCW condition, but independently altered the moment of inertia by moving the CW into the liner of the helmet. The modified CWL weighed 0.66 kg and was molded and evenly distributed across the posterior-interior aspect of the helmet. In each NVG condition (hNVG, hCW, and hCWL) a battery pack (0.23 kg) was also attached to the back of the helmet, as would be required when wearing NVGs.

The visual target acquisition system (VTAS) provided a consistent and objective way to elicit reciprocal head movements akin to a scanning task performed by pilots during flight.[16] Similar to a classic Fitts' tasks,[25] subjects performed rapid reciprocal head movements where a small head mounted laser interacted with solar panels (6 V 100 mA, 100 mm diameter; Sundance Solar, Hopkinton, NH) arranged in pairs, such that the changing voltage signal could be used to provide subjects with feedback via LEDs regarding successful target acquisitions. The VTAS has been shown to provide a reliable approach

Table I. HSM Conditions with Corresponding Mass and Moments of Inertia.

Helmet condition		Total Mass (kg)	Relative increase in moments of inertia (%)
hOnly	 3D printed tubes	1.43	Baseline
hNVG	 Battery pack; NVGs	2.21	Lateral bend = 12.17 Rotation = 56.05 Flex/ext = 63.06
hCW	 CW; Battery pack; NVGs	2.81	Lateral bend = 18.23 Rotation = 71.50 Flex/ext = 77.18
hCWL	 CWL; Battery pack; NVGs	2.81	Lateral bend = 14.97 Rotation = 65.33 Flex/ext = 71.70

to assess changes in target acquisition performance under different HSM conditions.[6] For the purpose of our experiment, we oriented the VTAS such that we could evaluate performance, and the corresponding function of neck muscles, in four unique conditions comprised of two movement trajectories (yaw and pitch) and two movement amplitudes (near and far). The four trajectories were: 1) yaw far (70°); 2) yaw near (35°); 3) pitch far (70°); and 4) pitch near (35°). Near amplitudes were deployed to require a neck range of motion akin to the range of motion required during HSM day flight where peripheral vision enables a wider field-of-view in the absence of NVGs. For example, to see a target deviated by 70° from neutral about the yaw axis, under normal field of view, an individual may use a combination of neck and eye motions to see that target. Use of NVG limits compensation by eye motion and requires larger neck motion. The far amplitudes were deployed to require a range of motion akin to the operational configuration of a night flight where field-of-view is restricted and thus greater neck range of motion is required[12] (i.e., a reduced ability to use eye motion in addition to increased neck motion to see the target).

Procedure

Subjects sat in an automobile seat equipped with a 4-point harness, used to represent helicopter seating. Subjects were exposed to HSM conditions in a random order and the presentation of movement conditions were block-randomized within each HSM condition. Subjects performed three repetitions in each movement condition within each HSM condition, such that subjects completed a total of 48 trials (4 HSM conditions $\times$ 4 movement $\times$ 3 repetitions). For each trial, subjects were told which movement condition (trajectory and amplitude) to acquire and were instructed to "acquire as many targets as possible" in a given movement condition in a 20-s time period. As a result, each trial included a series of independent turns (i.e., move from left target to right target, or right target to left target) that could be extracted out for analysis as necessary (described below). Subjects were given at least 30 s rest between trials and at least an 8-min rest between helmet conditions.

Participant head and thorax kinematics were collected at 80 Hz using a 12-camera Vicon motion capture system (Vicon, Centennial, CO, USA). Subjects were instrumented with six reflective markers and one rigid body on their trunk, as required to define and track motion of the trunk segment (which was negligible due to the 4-point harness). Five markers on the helmet system were used to track the head. A neutral seated position was used as a static trial and reference posture for marker calibration. Data were visually inspected, labeled and gap filled in Nexus 2.0. Labeled and filled data were then imported into MATLAB R2018a (Mathworks Inc., Natick, MA) and dual passed through a low pass, second order Butterworth filter with an effective cut-off frequency of 6 Hz.[25] Euler angles were used to determine the motion of the head relative to the trunk. A ZYX rotation matrix (flexion/extension, axial rotation, lateral bend) was applied in accordance with the International Society of Biomechanics recommendations for describing intervertebral motion.[33]

Surface electromyography (EMG) was recorded at 2000 Hz using wireless Trigno mini sensors (Delsys, Natick, MA) placed bilaterally over the sternocleidomastoid (SCM), and upper neck extensors (UNE). Prior to electrode placement the area of interest was shaved and cleansed with 70% isopropyl alcohol. Electrodes were positioned as follows: SCM – two-thirds of the distance between the mastoid process and the suprasternal notch,[2,8] and UNE – at the level of the fourth cervical vertebrae 2 cm from the midline.[13,24] Maximum voluntary contractions (MVCs) were collected for each muscle to allow for normalization. Post collection, EMG data were imported and processed in MATLAB R2018a (Mathworks Inc., Natick, MA). First, EMG was de-trended to remove DC bias. Data were high pass filtered at an effective cut-off of 30 Hz using a dual pass, second order Butterworth filter to remove any contamination from heart rate.[7] Data were then full wave rectified, and filtered using a single pass, second order Butterworth filter with a 4 Hz cut-off.[23] Lastly, linear enveloped signals were normalized to the peak EMG amplitude elicited from each muscle, respectively, during the MVC trials.

Outcome measures were calculated from the third repetition of each trial. Where an error occurred in the third repetition (the same solar panel was acquired twice in a row), the second repetition was used. We avoided use of the first trial to mitigate risks associated with acute familiarization effects.

Neck angle data were differentiated using finite differentiation to get angular velocity (deg/s) and differentiated again to get angular acceleration (deg $\cdot$ s^{-2}). Peak acceleration was extracted for each trial.

Cocontraction was calculated for each turn that occurred within a trial. A turn was operationalized using the VTAS data and was defined from the instant one target was acquired (i.e., the left or bottom target) to the instant the reciprocal target was acquired (i.e., the right or top target). Therefore, yaw trajectories consisted of left and right turns, and pitch trajectories consisted of up and down turns. All turns within a trial were averaged specific to their direction (i.e., left and right, or up and down), before cocontraction was calculated. The cocontraction ratio (CCR) for was calculated using the methods described by Cheng, Lin, and Wang.[4] In brief, muscles were preallocated to agonist and antagonists for each turn and CCR was calculated using Eq. 1 and 2 below.

$$CCR = \frac{\Sigma NAIEMG_{antagonists}}{\Sigma NAIEMG_{total}} \qquad \text{Eq. 1}$$

where: *NAIEMG* is the normalized average integrated EMG, calculated as:

$$NAIEMG = \frac{IEMG}{maxEMG \times t} \qquad \text{Eq. 2}$$

where: *IEMG* is the integration of the filtered EMG signal (not normalized to MVCs), *maxEMG* is the maximum EMG signal as found in the MVCs, and t = the length of the turn, in frames.

A gaps analysis was performed on the normalized, linear enveloped EMG data for an entire trial. Gap frequency was defined as the number of times EMG was below 0.5% MVC for at least 0.5 s during a trial.[15]

Amplitude probability functions[19] were assembled from the normalized, linear enveloped EMG data across the entire trial length. The 10^{th}, 50^{th}, and 90^{th} percentiles were calculated to represent the static (10^{th}), median (50^{th}), and peak (90^{th}) EMG levels for each condition.

Statistical Analysis

Two-factor repeated measures analyses of variance (ANOVAs) ($\alpha = 0.05$, $\beta = 0.08$) were used to assess the potential influence of amplitude (two levels: near and far), and HSM condition (four levels: hOnly, hNVG, hCW, and hCWL). Trajectory was not considered as a factor within the ANOVA, where separate ANOVA models were used for each trajectory (yaw and pitch). Main effects were assessed and pairwise comparisons were made where necessary using Bonferroni corrections. All data were analyzed using SPSS Version 25.0 (IBM Cor, Armonk, NY). Statistical significance was set at $\alpha = 0.05$.

RESULTS

There was a significant main effect of target amplitude on peak acceleration of the head in both the yaw [$F(1,29) = 128.411$, $P \leq 0.001$, $\eta p^2 = 0.816$)] and pitch trajectories [$F(1,29) = 76.550$, $P \leq 0.001$, $\eta p^2 = 0.725$]. Pairwise comparisons revealed that peak head acceleration was significantly higher for far compared to near target amplitudes.

There was no main effect of the HSM condition on CCR in the pitch trajectory; however, there was a significant effect of HSM condition on CCR in the yaw trajectory [$F(3,78) = 8.992$, $P \leq 0.001$, $\eta p^2 = 0.257$]. Pairwise comparisons revealed increased CCR for the counter-weighted conditions (CW and CWL) (**Fig. 2**).

There was a main effect of target amplitude on CCR in the pitch trajectory [$F(1,25) = 38.448$, $P \leq 0.001$, $\eta p^2 = 0.606$] and in the yaw trajectory [$F(1,26) = 110.557$, $P \leq 0.001$, $\eta p^2 = 0.810$]. Pairwise comparisons revealed that CCR was significantly lower for far amplitudes compared to near in both pitch and yaw trajectories (Fig. 2).

The average number of gaps ranged from 0.4 to 2.6 gaps per trial in SCML and SCMR, and 0.1 to 0 gaps for UNEL and UNER. **Table II** shows the average number of gaps in each trajectory for each muscle. There was a main effect of HSM condition on gap frequency for SCMR in the pitch [$F(2.82,67.77) = 3.394$, $P = 0.025$, $\eta p^2 = 0.124$] and yaw trajectories [$F(3.78) = 27.500$, $P = 0.022$, $\eta p^2 = 0.115$], respectively. However, sample-size limitations did not permit sufficient statistical power to detect specific differences post hoc using pairwise comparisons. There were no significant main effects in the SCML, UNEL, and UNER in either trajectory.

There was a main effect of amplitude on the number of gaps in the pitch trajectory for SCML [$F(1,24) = 5.739$, $P = 0.025$, $\eta p^2 = 0.193$] and SCMR [$F(1,24) = 7.233$, $P = 0.013$, $\eta p^2 = 0.232$] (**Fig. 3**). Pairwise comparisons revealed that the near amplitude had significantly more gaps than the far amplitude for both SCML and SCMR. There was no main effect of amplitude on UNEL or UNER in either trajectory.

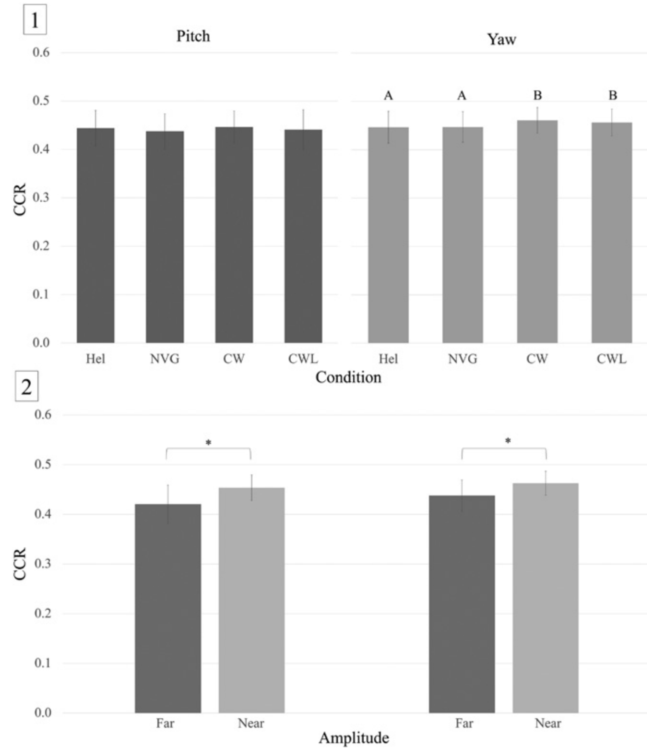


Fig. 2. 1) The effect of HSM condition on mean CCR in both the pitch and yaw trajectories. A and B indicate there is a significant difference between conditions ($P < 0.05$). 2) The effect of amplitude on average CCR in both the pitch and yaw trajectories. *Indicates significant differences ($P < 0.05$). Error bars represent 1SD.

In the pitch trajectory, HSM condition had a main effect on muscle activity in SCMR [$F(3,75) = 2.972$, $P = 0.037$, $\eta p^2 = 0.106$], UNEL [$F(3,75) = 5.400$, $P = 0.002$, $\eta p^2 = 0.178$] and UNER [$F(1.766,44.158) = 4.658$, $P = 0.018$, $\eta p^2 = 0.157$] in the 10^{th} percentile, UNER [$F(2.221,55.530) = 8.194$, $P = 0.001$, $\eta p^2 = 0.193$] in the 50^{th} percentile, and UNER [$F(3,75) = 5.353$, $P = 0.002$, $\eta p^2 = 0.176$] in the 90^{th} percentile. In the yaw trajectory, HSM condition had a main effect on muscle activity in SCMR [$F(3,75) = 5.400$, $P = 0.002$, $\eta p^2 = 0.178$] and UNER [$F(3,75) = 5.400$, $P = 0.002$, $\eta p^2 = 0.178$] in the 10^{th} percentile, UNER [$F(3,75) = 5.400$, $P = 0.002$, $\eta p^2 = 0.178$] in the 50^{th} percentile, and UNER [$F(3,75) = 5.400$, $P = 0.002$, $\eta p^2 = 0.178$] in the 90^{th} percentile. Pairwise differences are shown in **Fig. 4**.

In the pitch trajectory, target amplitude had a main effect on muscle activity in all muscles in each percentile. Similarly, in the yaw trajectory, target amplitude had a main effect on all muscles in each percentile, with the exception of SCML in the 10^{th} percentile. Pairwise comparisons revealed that EMG was significantly higher in far trajectories compared to near.

Table II. Average (± 1 SD) Number of Gaps for Each Condition in the Yaw and Pitch Trajectories for Each Muscle.

	PITCH				YAW			
	hOnly	**hNVG**	**hCW**	**hCWL**	**hOnly**	**hNVG**	**hCW**	**hCWL**
SCML	2.0 ± 3.4	1.3 ± 2.6	1.4 ± 3.0	1.0 ± 2.0	2.8 ± 3.2	2.7 ± 2.9	2.2 ± 3.1	2.5 ± 2.8
SCMR	1.5 ± 2.7	1.3 ± 2.6	0.9 ± 2.2	0.9 ± 2.0	2.3 ± 2.7	2.4 ± 3.0	1.5 ± 2.0	1.9 ± 2.4
UNEL	0.0 ± 0.2	0.0 ± 0.1	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0
UNER	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0	0.0 ± 0.0

DISCUSSION

The objective of this study was to probe how increased mass, moment of inertia, and range of motion interdependently affected neck function using a novel visual target acquisition task. The main findings were threefold: 1) increased mass resulted in a modest increase in cocontraction for the monitored muscles; 2) increased amplitude of motion generally increased neck muscle activity as measured using static (10th percentile), median (50th percentile), and peak (90th percentile) EMG levels; and 3) increasing moment of inertia had minimal effect on any of the outcome measures evaluated in this study (as determined by assessing CW vs. CWL conditions). On average, upper neck extensor muscles had zero rest during the scanning task regardless of HSM condition or range of motion. Overall, the current results suggest that increased range of motion (movement amplitude), and in turn, increased muscular demand, may be a dominant causal pathway by which NVGs lead to neck trouble. However, as a secondary pathway, measured data also demonstrated that increased mass resulted in increased cocontraction. To further support increased mass as a secondary pathway, an average of zero gaps, or no muscular rest, was found in the upper neck extensors. Sustained, increased cocontraction requirements combined with little to no rest have implications with respect to the Cinderella hypothesis-based injury model.

Our first hypothesis postulated that an increase in mass would increase cocontraction and decrease the number of muscular gaps or rest. It was proposed that an increase in mass would cause a destabilizing force, thereby requiring increased cocontraction to stabilize the head across the entire scanning task (Fig. 1). Significant differences in cocontraction were found between counter-weighted conditions (hCW and hCWL) and non-counter-weighted conditions (hOnly and hNVG) in the yaw trajectory, partially supporting this hypothesis. While the results indicate an increase in cocontraction with increased mass, these results should be interpreted with caution as mean differences were no greater than 0.015 ± 0.004, or a 0.15% change in CCR, of which the clinical significance is not known. Comparatively, Callaghan[3] reported muscle coactivation during sustained static postures, under different helmet conditions, and found no helmet mass main effects. One explanation for small changes

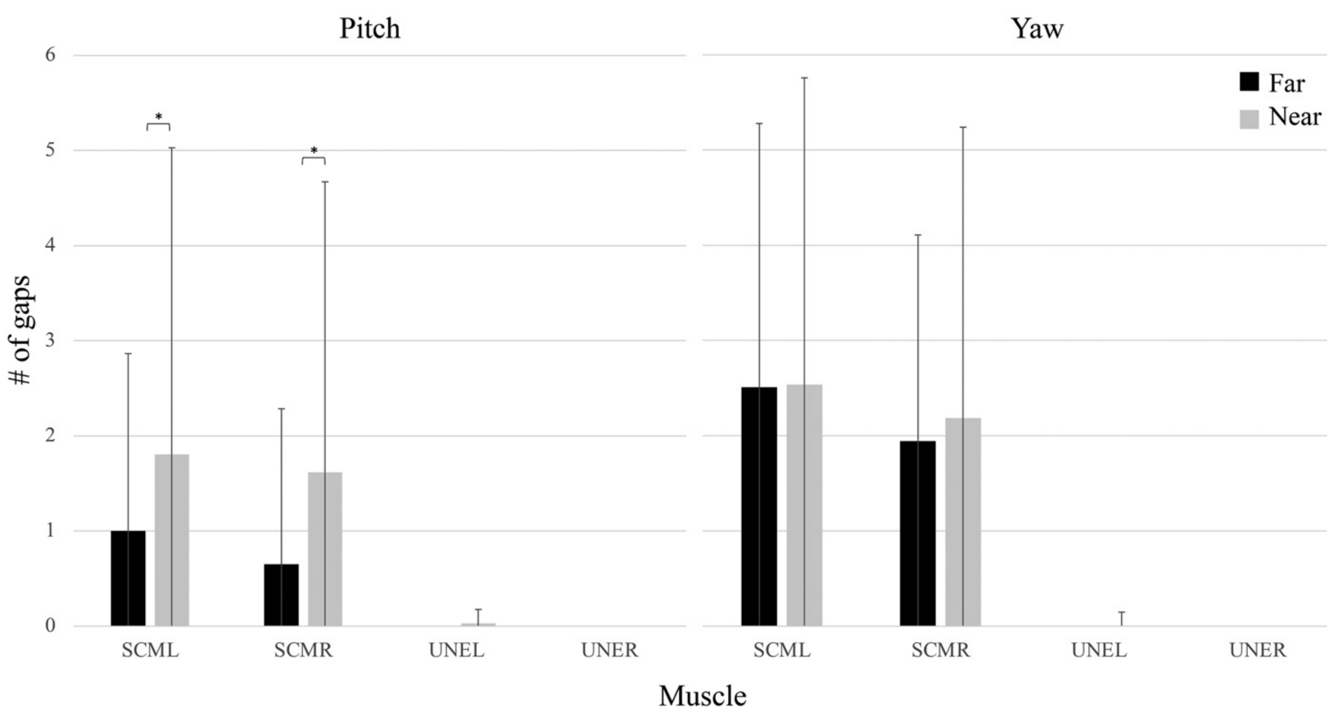


Fig. 3. The effect of amplitude on mean number of gaps for each muscle in the pitch and yaw trajectories. *Indicates statistical significance ($P < 0.005$). Error bars represent 1 SD.

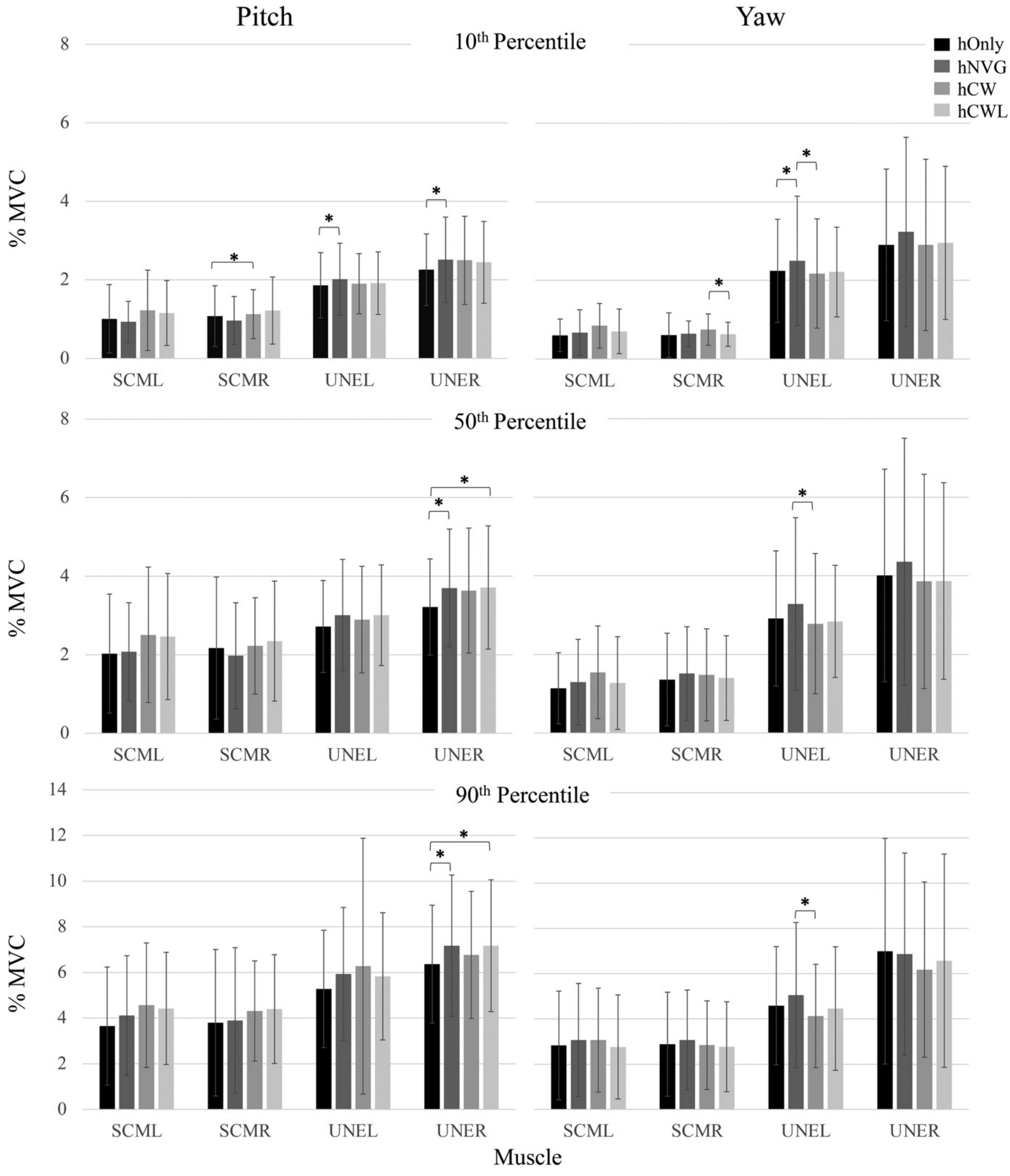


Fig. 4. The effect of HSM condition on average 10th, 50th, and 90th percentile EMG values (%MVC) for the pitch and yaw trajectories. *Indicates statistical significance ($P < 0.05$), error bars represent 1 SD.

despite increased load may be due to the complexity of the neck musculature and load sharing.[24,31] As only two muscles were assessed in this study, it is unknown what the contributions of the deep cervical stabilizers are. It is possible that this study has underestimated the increase in cocontraction due to the inability to access a number of deep neck muscles. However, although the changes seen in this study were small, these changes may be amplified when also exposed to vibration consistent with an operational environment. Therefore, the data suggest the plausibility of a mass-related destabilization effect resulting in increased cocontraction, where the influence of this pathway may be amplified in real flight scenarios, under vibration.

In work with sustained, low-level contractions, the number of short breaks, or muscular 'gaps' has been shown to predict neck/shoulder disorders.[32] For this reason, a gaps analysis was performed to provide further insight into the effects of increased mass and its potential effect on type one fiber fatigue or the Cinderella hypothesis.[14] Interestingly, we did not find an effect of HSM condition on the number of muscular gaps, with the exception of SCMR; however, the effect size was small and therefore pairwise differences could not be detected. Importantly, across all the muscles and conditions, the number of muscular gaps was very small, with an average of zero gaps in the UNEs. This suggests that the helmet mass alone may be enough to require sustained muscle contraction as required to initiate damage via a Cinderella hypothesis-like pathway.

Consistent with our second hypothesis, an increase in muscular activation and demand was associated with the increase in range of motion between the near and far target amplitudes. In fact, static (10th percentile), median (50th percentile), and peak (90th percentile) were all significantly higher for far compared to near amplitudes, with the exception of the 10th percentile SCML in the yaw trajectory. Posture and range of motion have long been understood as influential factors when considering the effects of helmets and HSM on neck trouble.[20,23,31] In fact, a 2004 report offering recommendations to reduce flight-related neck pain suggested moving the control display unit to a point further up to reduce extreme forward flexion.[1] Further, Forde et al.[12] determined that a key difference between day flying (without NVGs) and night flying (with NVGs) was time spent in extreme postures at night. They determined that loading was increased in part by the mass of the helmet and NVG system, but more significantly by time spent in nonneutral postures. Interestingly, while Harms-Ringdahl et al.[16] agreed that extreme postures increase the load moment at the C7-T1 joint and are likely a causal factor in neck trouble, they found no significant increase in muscle activity during sustained extreme flexion and extension. They proposed that this finding may suggest that, when holding very extreme flexion positions, the induced moment is balanced by passive connective tissue structures such as joint capsules and ligaments. Given that the subjects quickly moved in and out of extreme postures, it was not surprising that our results identified a greater magnitude of change in EMG than Harms-Ringdahl et al.[16] Our findings support previous work suggesting that range of motion is a critical risk factor in the injury pathway toward neck trouble. However, more detailed musculoskeletal modeling is required to probe how the load moment might be balanced via active and passive tissues.

When discussing the effects of target amplitude, it is important to take into consideration the change in head and trunk kinematics. While not part of our original hypothesis, it was found that peak head acceleration was significantly higher for far target amplitudes compared to near target amplitudes. This may be an important influential factor as higher muscle forces were required to achieve higher angular acceleration and to bring the head to a stop as well. This change in acceleration between amplitudes may also help to explain the change in CCR with amplitude. An increase in angular acceleration and concurrent decrease in CCR (observed in the far amplitude) is aligned with findings from Cheng et al.,[4] who assessed the effect of speed on neck muscle cocontraction using the CCR. They determined that CCR was also significantly lower in fast speeds (23.0-32.1 $° \cdot s^{-1}$) compared to slow and medium speeds (3.0-13.1 $° \cdot s^{-1}$). They attribute these differences to control strategies, one being a feedback loop for slow and controlled movements, and the second being a feed-forward loop for fast movements. Future studies should investigate the role of range of motion versus speed on neck muscle activity to determine the influence of both factors.

In our third hypothesis, we postulated that an increase in moment of inertia would increase peak muscular activity required to stop and start the head. As the CW and CWL conditions had the same mass, but different moments of inertia, statistical difference between these conditions would indicate that moment of inertia is an important factor. However, few changes were seen in the 90th percentile EMG results, which represent the peak muscle activation (Fig. 4). This suggests that changes in moment of inertia from the addition of a CW may not affect muscle activation requirements to control dynamic motion of the head to a magnitude that is important with respect to plausible injury pathways. It is interesting that despite increasing helmet mass by over 50%, and changing the moment of inertia of the helmet, peak neck muscle activity to stop and start did not significantly change during the rapid reciprocal scanning. Few authors have assessed the effect of HSM on neck muscle activity during dynamic movements in the laboratory or in flight, but our results continue to support emerging findings. For example, Callaghan[3] assessed the effects of no helmet, helmet only, helmet with NVGs, and helmet with CW on neck muscle activity during static and slow-moving tasks. Of the 315 statistical comparisons done to determine the effect of helmet condition on muscular activation, including mean, median, peak root mean square (RMS), and amplitude probability distribution function (APDF), only 6 main effects of head supported mass were found. They concluded that helmet condition had little effect on neck muscular responses. Murray et al.[24] found similar results when recording EMG during a cruising flight and concluded that added NVGs resulted in less than a 1% difference in mean muscle activity. Mounting evidence increasingly supports that altered postural requirements and not mass may be a dominant factor influencing the pathway toward neck injury.

It remains curious that increased mass and moment of inertia of the helmet system seem to have limited effect on neck muscle activity within the context of flying. A number of factors may influence this phenomenon. First, some authors suggest it is due to the nonlinear relationship between force and muscle activity.[24,31] This was demonstrated by Schuldt and Harms-Ringdahl[26] who demonstrated that a force up to 40% of maximum could be produced by a muscle activity level between 10–15% MVC. They determined that muscle activity required to produce the same force can differ based on neck position, which data from our study support. Further, it is likely that the load is shared among a number of muscles in the neck acting synergistically. Because only two muscles were assessed in this study, it is unknown what the contributions of the deep cervical stabilizers are. It is possible that other muscles in the neck are contributing more to stop and start the head; however, we did not capture it with the UNE and SCM muscles.

Limitations must be taken into consideration when interpreting the results of this study. First, surface EMG of the neck is susceptible to large amounts of cross-talk due to the small size and number of muscles in the neck.[31] Care was taken in placement of electrodes to try to mitigate these effects. Further, only two muscles were assessed in this study, SCM and UNE. While these are among the most commonly observed in HSM studies and were determined to be the primary movers in the yaw and pitch trajectories assessed, deeper cervical stabilizing muscles may also have a very important role in rapid neck movements and stabilization of different HSM conditions.

Despite attempts to maximize the external validity of this study, select operationally relevant characteristics were difficult to capture. This study tried to simulate a number of flight-like characteristics, such as amplitudes for near and far scanning,[12] the helmet and NVGs, and the chair and harness. However, there are characteristics that differ from a real flight and may be important to consider. First, our study population was healthy, young adults. Active aircrew are noted to have degenerative spine changes and therefore their response to a scanning task may differ from a healthy population.[24] Secondly, these results should be interpreted relative to the length of a military sortie, which can be up to 3.5 h.[17,24] In the present study subjects were given ample rest time to prevent fatigue from occurring; however, many authors have noted that fatigue may be a factor leading to neck trouble in pilots.[18,30,31] It remains unknown whether the results from this study would be consistent over a longer duration. Lastly, we did not consider vibration or macroscopic loads caused by helicopter acceleration and deceleration, a factor that many authors have suggested may be contributing to neck pain and degraded performance.[11,27,29] It is hypothesized that vibration could amplify the results we observed here, such that 10^{th} and 50^{th} percentile EMG values could approach risk threshold limit values. It remains important to determine how vibration influences muscle activity and performance under these different target amplitudes and HSM conditions.

In conclusion, this study probed the effects of mass, moment of inertia, and range of motion on neck muscle activity during the performance of a dynamic reciprocal scanning task, as related to rotary wing aircrew tasks. The most important outcome from this study was that target amplitude, which required an increased range of motion (akin to the increased range of motion required when wearing NVGs), had the greatest influence on neck muscle activity, relative to added mass or altered moment of inertia. Increased mass increased cocontraction requirements, reinforcing the continued pursuit of existing strategies aimed at reducing head supported mass, but as a secondary pathway to modulate neck trouble. In conclusion, when considering how to reduce neck pain and injury in helicopter pilots, the current study points to increasing the field of vision of NVGs to reduce the more extreme ranges of neck motion associated with NVG flying as well as new helmet designs that reduce mass and inertia.

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Authors and affiliations: Laura A. Healey, M.Sc., Jack P. Callaghan, Ph.D., and Steven L. Fischer, Ph.D., Department of Kinesiology, and Duane S. Cronin, Ph.D., and Jack P. Callahan, Ph.D., Department of Mechanical and Mechatronics Engineering, University of Waterloo, Waterloo, Canada; and Aaron J. Derouin, Ph.D., School of Kinesiology and Health Studies, Queen's University, Kingston, Canada.

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