Effects of Firefighter Helmets on Cervical Intervertebral Kinematics: An OpenSim-Based Biomechanical Study

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Abstract

Wearing head-mounted devices for prolonged time in occupational settings is one of the leading causes of neck pain. In the case of firefighters, the helmets are designed to provide protection against both fire and impact, making it heavier, thereby increasing the risk of neck pain and musculoskeletal injuries. In this study, we evaluated user adaptability and chances of injury with two different firefighter helmet designs, US-style helmet with a comparatively higher center of mass (COM), and European-style helmet with higher mass, based on cervical intervertebral kinematics. We collected motion data of 18 male and 18 female firefighters while performing static and dynamic flexion-extension and lateral bending with and without the helmets and measured the helmets' inertial properties. Musculoskeletal simulations were performed by applying the motion data and helmet inertial properties to a modified OpenSim full-body model to calculate cervical intervertebral kinematics. It was found that the US-style helmet, although lighter in weight, significantly increased the maximum neck and intervertebral angles during static flexion, and lateral bending and decreased static and dynamic maximum extension angles. In addition, both helmets affected cervical lordosis, which could potentially increase the risks of musculoskeletal disorders during long-term usage or firefighting operations. This study revealed that the higher lever arm of the helmet COM and moment of inertia about the axis of rotation, especially in the case of US-style helmets, is the reason for variations in neck kinematics. A low-profile helmet design (lower COM) would improve the safety and musculoskeletal health of the firefighters.

Introduction

Neck pain is a multifactorial musculoskeletal health condition with a high global prevalence. According to the 2019 Global Burden of Disease (GBD) Study, neck pain was among the top four most prevalent musculoskeletal disorders (MSD) and affected 222.7 million people (2.9% of total population) worldwide (GBD, 2019). Various factors, ranging from ergonomic factors in the workplace to daily lifestyle and recreational activities, contribute to the onset and persistence of neck pain. For instances, prolonged static postures (Christensen et al., 2023), repetitive neck motions (Guidotti, 1992), heavy lifting or working in awkward neck postures (Ariens et al., 2000), sedentary lifestyle and poor sleep quality (Peterson and Pihlström, 2021), and chronic stress and certain mental health conditions (Kim et al., 2013) have been associated with work-related neck MSD. In addition to these causes, the usage of helmets and other head mounted devices in medical surgery (Nimbarte et al., 2013), military (Hanks et al., 2018), firefighter (Park et al., 2014; Wang et al., 2021), sports (Kent et al., 2020; Mortensen et al., 2020), and vehicle safety applications (Diyana et al., 2019) were found to increase the chances of neck MSD.

Previous biomechanical studies have linked excessive weight and shifted center of mass (COM) of the helmets and head mounted devices as the cause for cervical spinal disorders and neck muscle fatigue. For instance, Van Dijke et al. (1993) and Newman et al. (2022) showed that jet pilot helmets significantly increases the neck joint forces and moments, increasing the risk of neck injury; Harrison et al. (2016) found that the addition of night vision googles and counterweight increased muscle activation and affected postures, thereby inducing fatigue and cause neck pain; Barrett et al. (2023) showed the COM position, mass, and moment of inertia influence the compressive forces in the cervical spine; Baucher et al. (2022) found that an increase in weight

supported by the head and the restriction of cervical intervertebral motion could be risk factors of degenerative diseases, such as spinal cord spondylosis, degenerative disc disease, and ossification of the ligamentum flavum. In addition, studies on construction (Boschman et al., 2015), mining (Torma-Krajewski et al., 2006), and firefighting (Wang et al., 2021) helmets have emphasized the importance of enhancing ergonomic aspects alongside protective features, as in these domains, the helmets have to be worn for prolonged durations. In comparison to others, firefighter helmets are heavier as they are designed to provide both thermal and impact protection. For instance, in a simple comparison, a firefighter helmet without accessories weighs around 1.7 kg (Bullard UM6WH), while the advanced combat helmet weighs only 1.2 kg (Striker advanced combat helmet) without accessories. Additionally, the modern firefighter helmet serves as a platform for supporting ancillary accessories, such as communication devices, face shield, visor, night-vision goggles, and lighting equipment. The addition of these ancillary devices has two major effects: an increase in the total weight, and a potential shift in the center of mass (COM) of the helmet system (i.e., helmet plus supporting accessories). These changes can further increase the chances of injuries to the cervical spine and fatigue of the neck muscles. It was reported that about 25 % of all cases (15,750 cases out of 63,000 cases) of firefighter injuries are related to the head and neck —out of which, 40% involved strains/overexertion (NFIRS, 2019). A survey by Wang et al. (2021) showed that firefighter helmet is the second ill-fitting equipment among US firefighters and fourth among Chinese firefighters, limiting their head-neck range of motion and causing neck awkward postures, which can eventually lead to the aforementioned cervical intervertebral and disc degenerative diseases. Our systematic literature review indicated that the impacts of firefighter helmet on cervical intervertebral dynamics have remained unexplored, especially the manner in which the inertial properties of existing helmets affect the cervical intervertebral kinematics.

Although musculoskeletal modeling platforms, such as OpenSim (Delp et al., 2007), AnyBody (Damsgaard et al., 2006), and SIMM (Delp and Loan, 1995) can facilitate the estimation of cervical intervertebral kinematics during various human-machine interaction scenarios, only a handful of studies, namely, Barrett et al. (2022a); Barrett et al. (2022b); Mathys and Ferguson (2012); Newman et al. (2022), used OpenSim to study the helmet dynamics on neck biomechanics. Nevertheless, they have mainly focused on the influence of military and aviation helmets on cervical joint loading and muscle activations. To our knowledge, there are no studies on the influence of firefighter helmet on the neck joints. Additionally, there has been limited knowledge in the human adaptability with added mass to the head. Analyzing cervical joint kinematics and posture with helmets can capture human adaptability and provide information on the possibilities of musculoskeletal injuries.

Therefore, this study aimed to investigate the effects of helmet designs and their usage on cervical intervertebral kinematics, especially during various neck dynamic movements to understand how the different firefighter helmet design can influence neck and cervical kinematics. Two types of helmets are frequently used by firefighters: 1) US-style helmets that has a brim (resemblance to hats) and 2) European-style helmets that has no brim (resemblance to aviation helmets). A quantitative understanding of variations in cervical spinal kinematics due to different helmet designs can provide much-needed insights for a deepened understanding of designing a more ergonomic helmet and thereby reducing the risk of neck injury during their prolonged usage.

Methods

Participants

We recruited eighteen male (weight: 88.5 ± 18.9 kg; height: 1.77 ± 0.09 m; BMI: 28.8 ± 5.39) and eighteen female (weight: 69.5 ± 12.2 kg; height: 1.64 ± 0.05 m; BMI: 24.2 ± 2.85) firefighters from the local fire departments for this study. The inclusion criteria required all participants to be healthy and without any type of musculoskeletal, degenerative, or neurological disorders and did not have any recent history of neck, shoulder, and back injury or pain. Prior to their participation, they signed an informed consent form approved by the Institutional Review Board of Texas Tech University (IRB2020-708).

Experiment

Participants were asked to perform four different tasks: 1) holding a static head-neck position in full flexion, extension, and left and right lateral bending for five seconds (Figure 1), 2) performing self-paced dynamic movements of full flexion-extension and lateral bending, each separately for 3 seconds controlled using a digital metronome (Figure 1), 3) sustaining in full head-neck flexion and extension until fatigue, and 4) walking in two different velocities controlled by a digital metronome with the frequencies 1 Hz and 2 Hz. These tasks were performed while using three different helmet conditions: 1) no-helmet condition, 2) Bullard UM6WH US-style helmet, and 3) Caims XF1 European-style helmet in a random order. Static posture data was collected before each task, and all tasks were randomized and repeated twice.

We collected full-body movements, including the head-neck movements, using motion capture system at 60 Hz (Krestel 1300 motion capture systems; Motion Analysis Corporation, Rohnert Park, California, USA). Plug-in gait marker protocol (Vicon, 2023), consisting of 44 markers, was used. In addition, four wireless surface EMGs Delsys Trigno® (Delsys, Inc, Natick,

Massachusetts, USA) to record activations from two flexor (infrahyoid and sternocleidomastoid) and two extensor neck (cervical and upper trapezius) muscles at a sampling rate of 2222 Hz, two force plates (Bertec Columbus, Ohio, USA) to measure ground reaction forces at a sampling rate of 600 Hz , and a handheld, rotating 3D scanner (EinScan HX, Shining 3D, Hangzhou, China) to image individual helmets at a speed of 20 Hz frame rate (1,200,000 points/s) were used.

Figure 1: Subject performing a) dynamic flexion-extension and dynamic lateral bending tasks; and b) static flexion, extension, left-, and right-bending tasks for each helmet conditions.

As we were focused only on cervical kinematics, the data from task 3 (sustaining head-neck movements) and 4 (walking in two different velocities) were not included here. Besides, some of the data from task 1 and 2 was excluded due to poor quality. We have observed that the pelvis markers of some subjects were moved during the trials, while for some others the force plate data was poor. Therefore, only 20 subjects', 10 males (weight: 90.6 ± 19.1 kg; height: 1.77 ± 0.066 m; BMI: 28.8 ± 2.85 kg/m^2) and 10 females (weight: 67.4 ± 8.28 kg; height: 1.67 ± 0.055 m; BMI:

24.2 ± 2.85 kg/m^2), data was found suitable for this study. The selected motion capture data were preliminary processed in Cortex-9 software (Motion Analysis Corporation, Rohnert Park, California, USA) and converted to Track Row Column format. Then a custom MATLAB code was used to transform the coordinate frames of the data to use in OpenSim. Force plate data was exported from Cortex-9 software as C3D format and then converted to Motion files using C3Dtools (Mokhtarzadeh and Bagheri, 2023) web platform to input in OpenSim.

OpenSim Modeling

The MASI (Musculoskeletal model for the Analysis of Spinal Injury) model (Segments: 35, Joints: 34), a validated model developed by Cazzola et al. (2017), was used as a basis for our study. The MASI model is a full-body model that comprehends 35 rigid body segments and 34 joints. The model has 78 muscles on the neck region and 23 torque actuators that accounted for the upper and lower limbs. Moreover, it also inherited head and neck model of Vasavada et al. (1998), which implements three rotational degrees of freedom (row, pitch, and yaw) to C0-C1(atlanto-occipital), C1-C2 (atlanto-axial), C2-C3, C3-C4, C4-C5, C5-C6, C6-C7, and C7-T1 joints, totalizing 24 joints for the head and cervical spine. The inertial properties of head and neck components of the modified model were taken from Ivancic et al. (2007). Since the base model does not have infra-hyoid muscles, we incorporated these muscles from (Mortensen et al., 2018) to the base model to enhance the moment generating capabilities and to produce more biofidelic results. The modified model has a total of 98 Hill's type muscles in the neck. We also added markers to the model that match the experimental marker protocol for inverse kinematic analysis. For easy distinction, the markers on the model are referred to as model markers, while those from the experiment are called experimental markers in this study.

Three different versions of modified models corresponding to the no-helmet, US-style , and jet-style helmets were created to simulate static and dynamic neck movements of this study. In addition to the ergonomic aspects, the helmet inertial properties can also affect the cervical kinematics. Therefore, we initially estimated the helmets' inertial properties by modelling individual components in finite element pre-processor ANSA (BETA CAE Systems SA, Greece) and applying their respective mass measured using a weighing scale. Another modelling software NMSBuilder (Valente et al., 2017), was then used to generate the helmets model in OpenSim and the estimated inertial properties were applied. It was observed that COM of the US-style helmet is 5.8 cm superior and 1.7 cm anterior to the European-style helmet. However, the jet-style helmet is 250 g heavier than the US-style helmet. Additionally, we calculated the combined moment of inertia of the head and the helmet with respect to the C0-C1 joint by applying the parallel axis theorem.

Table 1: Mechanical properties of each helmet condition that were used in the simulations.

	Mass (kg)	COM* (m)			Moment of Inertia ($kg.m^2$)			Moment of Inertia combined with the head ($kg.m^2$)		
		X	Y	Z	I_{xx}	I_{yy}	I_{zz}	I_{xx}	I_{yy}	I_{zz}
US-style helmet	1.77	-0.002	0.152	-0.006	0.017	0.028	0.022	0.092	0.088	0.10
European-style helmet	2.02	-0.019	0.094	-0.008	0.023	0.03	0.031	0.076	0.068	0.090

*Calculated with respect to the basis of the occiput.

OpenSim Simulation

With the collected data, we performed the inverse pipeline in OpenSim to understand the effects of the different helmet types. The simulation workflow is illustrated in Figure 2. The inverse

pipelines include subject-specific model scaling to match anthropometric measurements, inverse kinematics for joints angle determination, inverse dynamics to calculate net joint moments, and static optimization for muscle force estimation. We used the marker dataset of a baseline static posture trial for scaling and making the model subject-specific. The acceptable scaling error was set to less than two centimeters. In the helmet models, the interface between the helmet model to the skull of the scaled model with a welded joint and applied their corresponding inertial properties.

We then used task-specific motion capture data to perform Inverse Kinematics (IK) simulations for all the subject-specific models. The IK used a optimization routine of experimental marker data (Task 1 and 2) to calculate each joint motion. The optimization routine minimizes virtual marker positions to match the experimental marker positions at each time frame to calculate the kinematics during each task. In some cases, adjustments were made to the model markers and their weightage to ensure both the RMS and maximum marker errors remained below the recommended thresholds (RMS: < 2 cm; Maximum: < 4 cm) for the IK analysis (Figure 2).

Both total neck angle and cervical kinematics were required to analyze the contribution of each cervical joint on the total neck movement. OpenSim IK provides only cervical spinal kinematics corresponding to each of the degrees of freedom in the model. Hence, we calculated the total neck angle by summing individual cervical kinematics. In addition, since each subject took different time to complete each task, we have downsampled the IK results to 100 data points with a MATLAB routine to compare the results between subjects.

The IK data was then used as input, together with the ground reaction forces, to perform OpenSim Static Optimization. It uses the IK body motion in an optimization routine to calculate the muscle activation by minimizing a cost function described by the summation of the activation of all the

98 muscles. The static optimization constraints the muscles activation by matching the summation of moments caused by the muscles with the moment of a given joint during a specific task. However, this study focused on the effects of the usage of helmets on the cervical spine and neck kinematics and the results of joint and muscle forces will be presented in a different study.

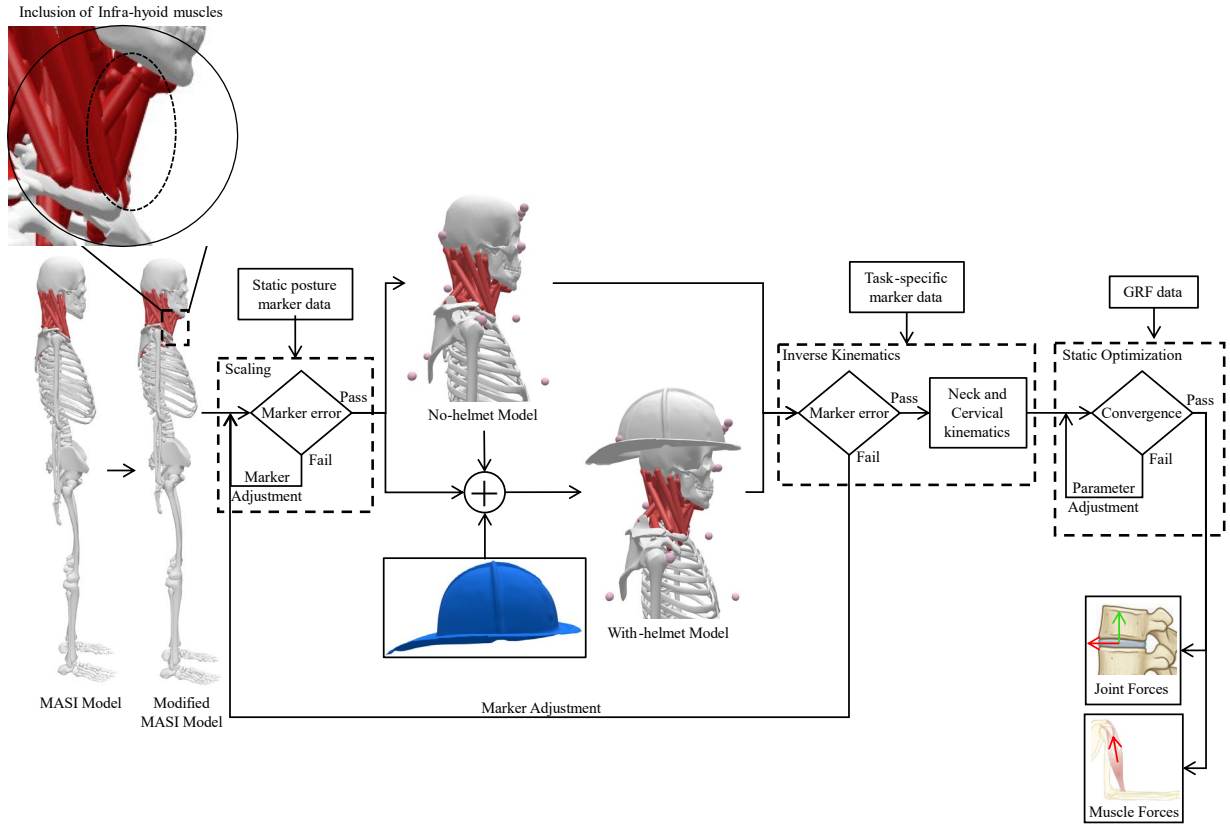

Figure 2: OpenSim workflow to create anthropometric subject-specific models and calculate task-specific neck and cervical kinematics and dynamics data with our model

Statistical Analysis

We calculated the descriptive statistics (mean and standard error) of the maximum flexion, extension, left-, and right-bending angles across all subjects—males and females separately—for all static and dynamic neck exertion trials. In order to investigate the cervical spine kinematics

under the effects of helmets, sex, and BMI and their interactions, we first considered a generalized linear regression model, however the data failed to follow normal distribution and presented skewness, even after data transformation. Therefore, we performed Kruskal-Wallis non-parametric analysis, where the kinematics of cervical joints and total neck angle were the dependent variables, and the independent variables were set to be helmet condition, sex, and BMI, which were considered as factors. For p-values lower than $\alpha = 0.05$ the effect is considered significant.

Results

Neck angle

The maximum neck angles observed for different tasks, helmet conditions and gender are presented in Table 2. For all subjects, helmets increased flexion and lateral bending angles and decreased extension angles. Additionally, neck angles during dynamic extension trials with helmets were reduced even more than the static extension trials (3° for European-style helmet and 6° for US-style helmet). It was also observed that the US-style helmet resulted in a greater increase in flexion and lateral bending angles while causing a more reduction in neck angle.

Between females and males, maximum neck angles without helmet for different tasks were observed to be higher in females. Except for dynamic extension with the US-style helmet, the differences in maximum neck angles among genders were reduced for all conditions. Females had comparatively lower maximum extension angles with the US-style helmet, indicating that the US-style helmet affects neck extension motion more adversely for females. Also, Kruskal-Wallis results showed that gender as a significant factor for maximum neck angles during all static postures and during dynamic left-bending. In the case of helmet condition, it was a significant factor only during static flexion, while BMI is not significant in any.

The mean time to complete the flexion-extension and lateral bending tasks were found to be, respectively, 3.03 s ± 0.59 s and 3.76 s ± 1.31 s for no-helmet; 3.24 s ± 0.76 s and 3.27 s ± 0.65 s for US-style helmet; and 3.23 s ± 0.38 s and 3.21 s ± 0.46 s. Interestingly, with the US-style helmet, the cervical joint and neck angles reached their maximum quicker than other conditions except during left-bending (**Figure 3Figure 4**). The European-style helmet also showed a very similar

time needed to reach the maximum angle during flexion-extension trials, but greater time to achieve maximum angle during both left- and right-bending.

In addition, when compared with the no-helmet condition, the helmets showed differences regarding the trend of the flexion-extension trials. The results suggested that the helmets removed an extended initial condition that is present mainly in the antlanto-occipital joint, by slightly inclining the head to a flexed condition. Also, the final position of the neck showed to be closer to zero for the no-helmet and European-style helmet conditions only, while the US-style condition had a flexed final position.

Table 2: Maximum flexion, extension, right-bending, and left-bending neck angles while holding static neck postures and performing dynamic neck movements.

		Static neck posture		Dynamic neck movement	
Maximum Flexion (°)	No Helmet	M: 33.1 ± 5.22 F: 38.0 ± 2.22	35.6 ± 1.01	M: 36.3 ± 4.48 F: 37.7 ± 4.05	35.62 ±1.02
	US-style Helmet	M: 38.7 ± 1.32 F: 39.3 ± 0.18	39.0 ± 0.21	M: 37.8 ± 3.08 F: 38.2 ± 1.71	38.04 ± 0.53
	European-style Helmet	M: 37.3 ± 3.03 F: 38.0 ± 3.92	37.7 ± 0.75	M: 36.3 ± 4.48 F: 37.7 ± 4.05	37.02 ± 0.92
Maximum Extension (°)	No Helmet	M: 45.6 ± 8.45 F: 51.7 ± 6.60	48.7 ± 1.75	M: 43.3 ± 10.8 F: 47.3 ± 10.9	45.32 ± 2.34
	US-style Helmet	M: 41.0 ± 11.2 F: 46.0 ± 6.58	43.5 ± 2.02	M: 39.2 ± 13.6 F: 35.8 ± 8.98	37.47 ± 2.48
	European-style Helmet	M: 44.8 ± 11.1 F: 50.0 ± 6.44	47.4 ± 2.00	M: 41.6 ± 11.4 F: 44.6 ± 7.66	43.06 ± 2.08
Maximum Right Bending (°)	No Helmet	M: 24.6 ± 4.25 F: 28.2 ± 3.74	26.4 ± 0.94	M: 27.2 ± 4.02 F: 29.3 ± 3.32	28.26 ± 0.82
	US-style Helmet	M: 27.3 ± 2.30 F: 29.0 ± 2.88	28.1 ± 0.58	M: 27.7 ± 3.30 F: 27.4 ± 3.19	27.58 ± 0.69
	European-style Helmet	M: 26.8 ± 3.69 F: 27.8 ± 3.07	27.3 ± 0.73	M: 26.0 ± 3.33 F: 27.0 ± 3.50	26.5 ± 0.74

Maximum Left Bending (°)	No Helmet	M: 24.8 ± 2.87 F: 28.2 ± 3.41	26.5 ± 0.77	M: 24.3 ± 7.49 F: 28.2 ± 3.49	26.27 ± 1.31
	US-style Helmet	M: 27.4 ± 2.85 F: 30.2 ± 2.28	28.8 ± 0.63	M: 27.4 ± 4.21 F: 29.2 ± 2.84	28.28 ± 0.79
	European-style Helmet	M: 27.2 ± 3.36 F: 27.9 ± 3.52	27.5 ± 0.73	M: 24.9 ± 6.10 F: 27.6 ± 2.50	26.24 ± 1.04

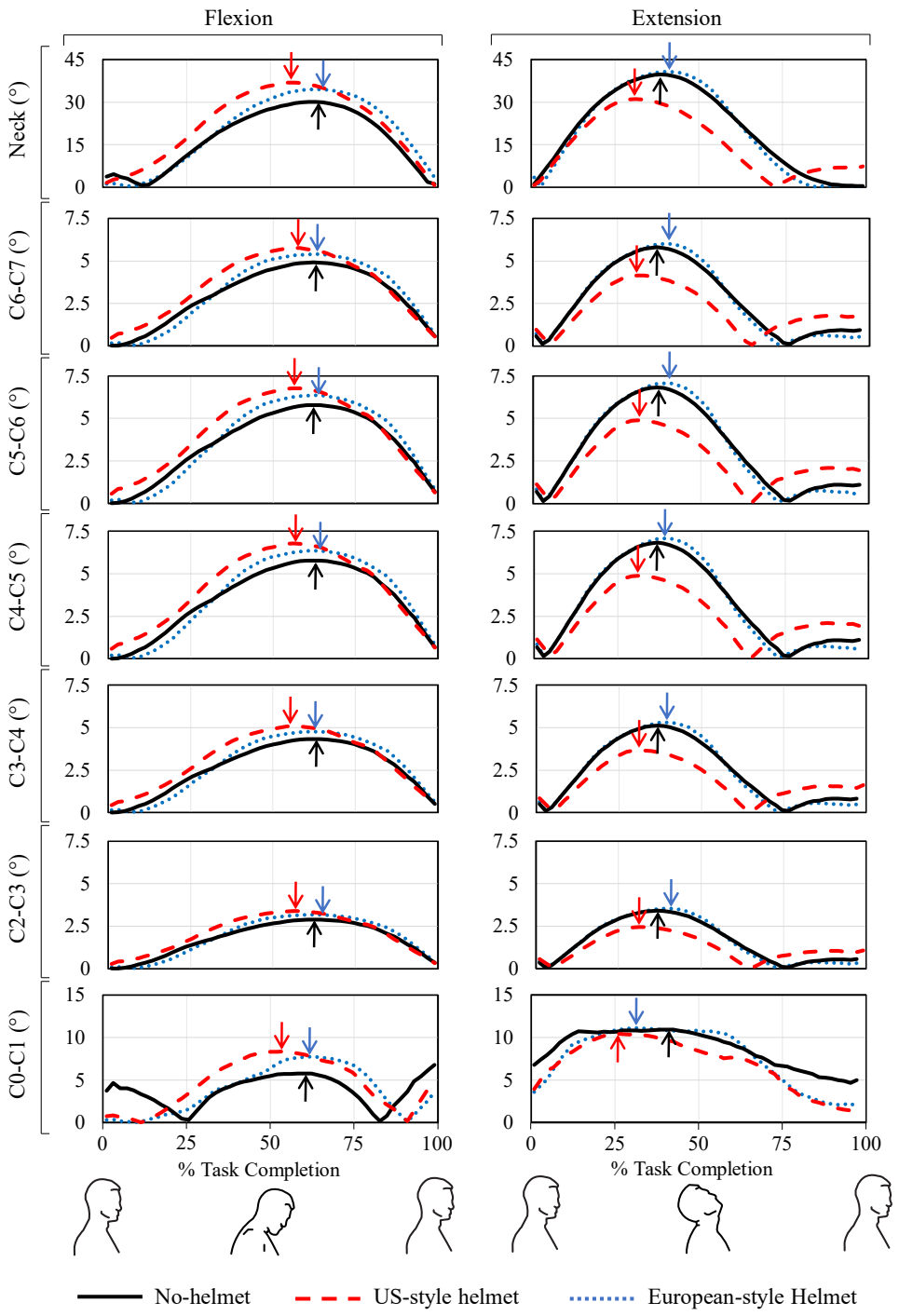

Figure 3: Flexion and extension movements of cervical spine joints for different helmet conditions. The C1-C2 and C7-T1 joints were not included as their movements were insignificant. The time-to-maximum flexion and extension angles is indicated with arrow sign.

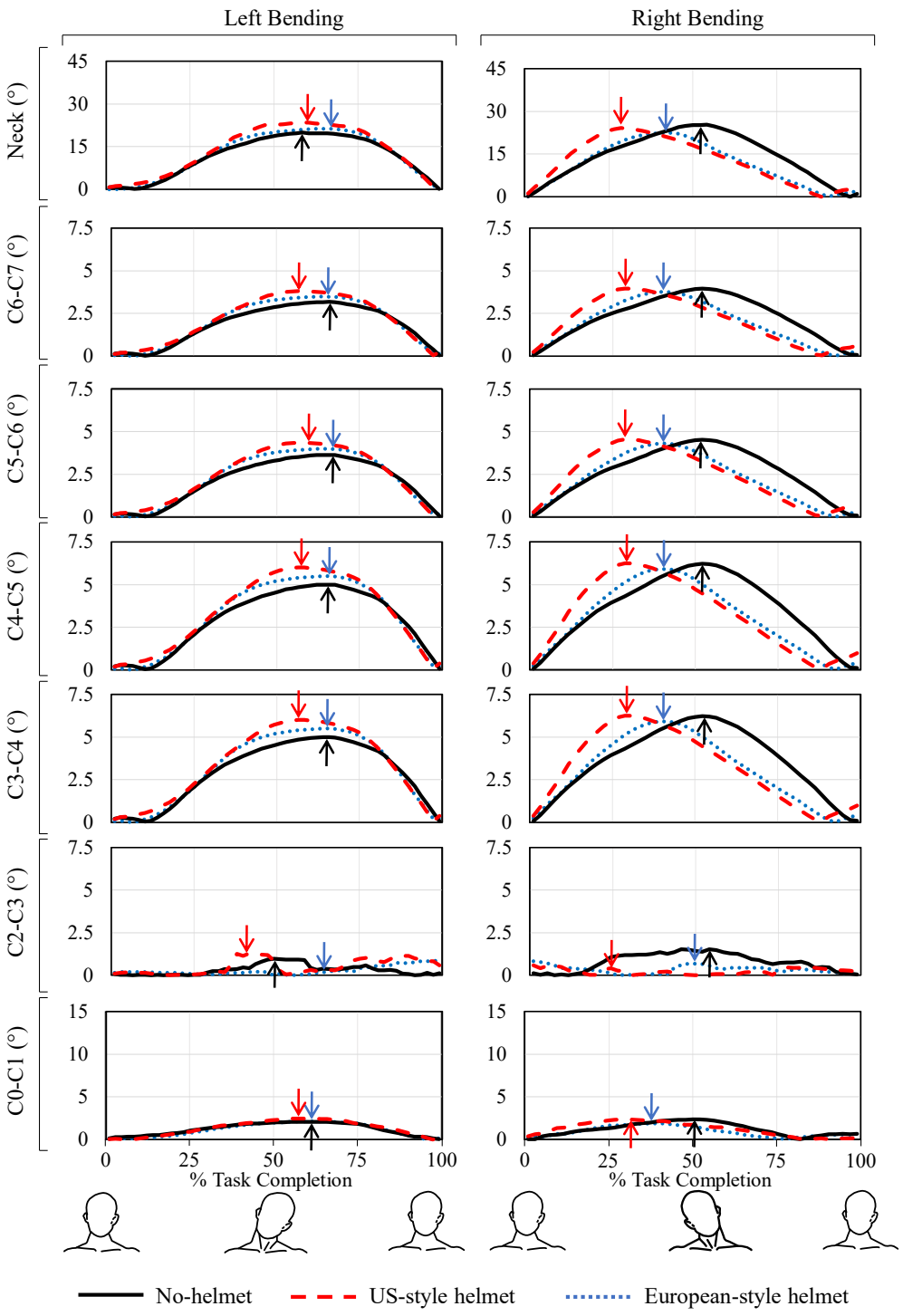

Figure 4: Left- and right-bending movements of cervical spine joints for different helmet conditions. The C1-C2 and C7-T1 joints were not included as their movements were insignificant. The time-to-maximum flexion and extension angles are indicated with arrow sign.

Maximum intervertebral angles

A comparison of maximum intervertebral joint angles showed that the atlanto-occipital joint contributed the most to neck flexion and extension, accounting for over 18% of overall neck movements during both static and dynamic trials (Figure 3). In contrast, the atlanto-occipital joint exhibited a lower range of motion during lateral bending tasks. Lateral bending motion is predominantly achieved by the middle cervical joints (C3 to C7), with C3-C4 compensating for the reduced range of motion in the other cervical spine joints. The least motion was observed in C1-C2, followed by C7-T1, for all cases of helmet conditions and tasks.

The Kruskal-Wallis test showed that wearing a helmet significantly affects the joint angles during flexion and lateral bending tasks. From the pairwise comparison of different helmet conditions, it was then revealed that only the US-style helmet is significantly influencing the neck kinematics. Unlike for the flexion, the C0-C1 joint angles during extension tasks did not vary for different helmet conditions. Instead, the helmet conditions only affected the C2-C3 to C6-C7 joint motions. BMI did not show any significance for both dynamic and static trials.

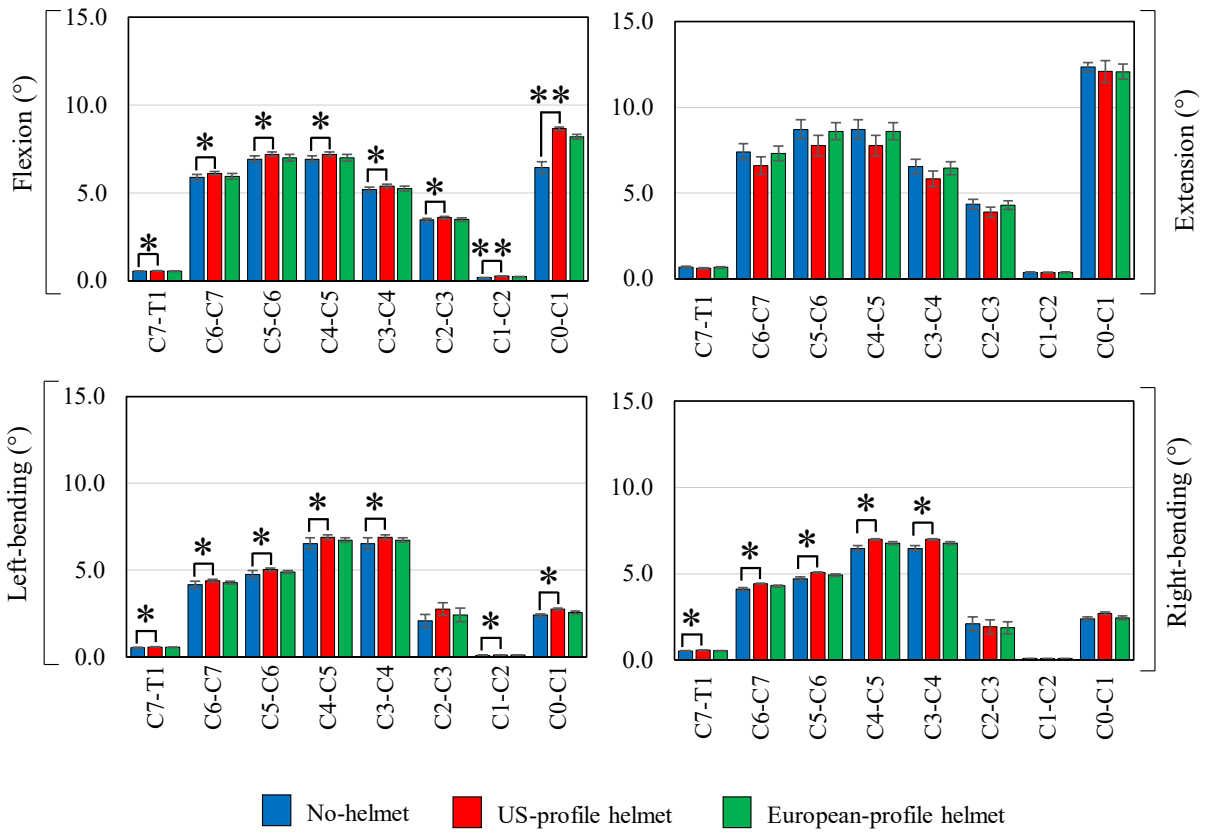

Figure 5: The maximum angles of individual intervertebral joints (C0-C1, C1-C2, C2-C3, C3-C4, C5-C6, and C7-T1) in maximum flexion, extension, and left and right lateral bending positions for all helmet conditions: no-helmet, US-style helmet, and European-style helmet. If the helmet condition was statistically significant for a joint, it is indicated with asterisks (*: p-value < 0.05; **: p-value < 0.01) symbol.

Discussion

Firefighter helmet designs have generally focused on historical or cultural aspects, often resulting in traditional and iconic designs while overlooking ergonomic aspects related to musculoskeletal health. The results of this study showed that the US-style helmet caused significant deviations in neck kinematics during static flexion, left-, and right bending, while the European-style helmet did not show any significance (Figures 3-5, Table 2).

At extreme positions, the neck undergoes eccentric contractions and thus the flexor muscles become more active during maximum neck extension, while extensor muscles become more active during maximum neck flexion. Also, the flexors are weaker and more sensitive than extensors (Garces et al., 2002; Jordan et al., 1999), which could be reason for reduced neck extension and increased neck flexion with the US-style helmet. Moreover, a high-profile design with 5.8 cm superiorly shifted COM of the US-style helmet (Table 1) increased the moment of inertia about C0-C1 joint axis (axis about which the head pivots during flexion-extension) by 12.4% that could significantly demand more muscle effort. Consequently, a discomfort due to extra exertion during extension trials might have affected neck kinematics (Lee et al., 2023). These findings indicate that an increase in the effective inertial properties of the head (head + helmet) could affect the extension movement of the head.

The contrasts in kinematics observed among different genders for different helmet conditions and tasks are due to the biological and anthropometric factors. Studies have shown that females usually have a greater proportion of muscles fibers type I (Miller et al., 1993), which provide greater endurance resistance; smaller vertebra (Stemper et al., 2008), resulting in higher cervical spinal flexibility; and weaker neck muscles (Vasavada et al., 2008). These factors may explain why

283 females attained a higher range of motion without the helmets, but a comparable range as that of
284 males with helmet conditions (Table 2). In other words, this suggests that females are more affected
285 by the helmet inertial properties.

286 The significant reduction in the C0-C1 joint moment during lateral bending tasks from flexion-
287 extension tasks is due to the shape of the joint. The occiput (C0) exhibits a convex shape, while
288 the atlas (C1) has a concave shape, primarily facilitating flexion-extension movement of the head
289 (Bogduk and Mercer, 2000). The reduction in the range of movement of the C0-C1 in the frontal
290 plane is compensated by an increase in movement of the C3-C4 joint movement during lateral
291 bending tasks (Figure 5). The C1-C2 joint did not contribute for flexion-extension and lateral
292 bending as it allows only axial rotation due to odontoid process (Bogduk and Mercer, 2000). By
293 observing the helmeted conditions, for both high- and European-style helmets (Figure 3: C0-C1
294 kinematics), it is evident that the helmet affected the cervical lordosis (natural curvature) at neutral
295 position (0 % of flexion task). The loss of cervical lordosis is associated with reduction in extensor
296 muscle strength (Alpayci et al., 2016) and degenerative process of the cervical spine (Choi et al.,
297 2021; Gao et al., 2019). The posteriorly shifted COM of the helmets could have caused a mass
298 imbalance about C0-C1 and thereby affected the cervical lordosis. A helmet with the COM lies
299 closer to the C0-C1 might improve the cervical lordosis. Additionally, although the time to
300 complete each task was comparable for US-style and European-style helmets, with the US-style
301 helmet attained maximum dynamic angles faster than with the European-style helmet. The faster
302 reach of maximum could be influenced by the greater inertia of the head-helmet about the atlanto-
303 occipital axis.

304 Even though both helmets provide protection from fire and falling objects, the effects of their
305 inertial properties must be considered to mitigate the chances of MSD and injuries. The US-style

helmet with brim design accounts for a wider range of protection against falling objects at the expense of increased effective moment of inertia of the head-helmet. On the other hand, the European-style helmet provides occipital and temporal coverage to protect the full head. This additional coverage increased the weight but lowered the COM of the helmet. Overall, this study's findings indicate that a lower COM of the helmet is more beneficial than the reduced weight of the helmet, in terms of achieving near normal kinematics. However, this does not necessarily mean that the European-style helmet would reduce spinal compression in comparison to US-style helmet as it is heavier. Only a dynamic analysis could provide the joint reaction forces to understand their effects on spinal compression, which will be the focus of future studies.

Conclusion

Firefighter helmet designs often overlook the ergonomic aspects, and their impact on cervical kinematics has not been thoroughly studied. This study showed that the helmet with a superiorly shifted COM, such as that of US-style helmet, significantly increased the neck flexion and lateral bending angles over a comparatively heavier European-style helmet with lower COM due to the increased lever arm and moment of inertia about the cervical joint axis. In addition, it is revealed that a superiorly shifted COM of US-style helmet made it difficult to control the neck rotational motion and affected the neutral posture (cervical lordosis), which could potentially increase the chance of neck injury during prolonged usage as well during on-ground operations. Therefore, both weight and COM location should be considered during the design of new firefighter helmets to enhance musculoskeletal health and reduce the fatigue rate while wearing them.

Conflicts of Interest

None of the authors have conflicts of interest to declare.

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References

Alpayci, M., Şenköy, E., Delen, V., Şah, V., Yazmalar, L., Erden, M., Toprak, M., Kaplan, Ş., 2016. Decreased neck muscle strength in patients with the loss of cervical lordosis. Clinical biomechanics 33, 98-102.

Ariens, G.A., Van Mechelen, W., Bongers, P.M., Bouter, L.M., Van Der Wal, G., 2000. Physical risk factors for neck pain. Scandinavian journal of work, environment & health, 7-19.

Barrett, J.M., Healey, L.A., Fischer, S.L., Callaghan, J.P., 2022a. Cervical spine motion requirements from night vision goggles may play a greater role in chronic neck pain than helmet mass properties. Human factors, 00187208221090689.

Barrett, J.M., Healey, L.A., McKinnon, C.D., Laing, A.C., Dickerson, C.R., Fischer, S.L., Callaghan, J.P., 2023. Head supported mass, moment of inertia, neck loads and stability: A simulation study. Journal of Biomechanics 146, 111416.

Barrett, J.M., McKinnon, C.D., Dickerson, C.R., Laing, A.C., Callaghan, J.P., 2022b. Posture and helmet configuration effects on joint reaction loads in the middle cervical spine. Aerospace medicine and human performance 93, 458-466.

Baucher, G., Taskovic, J., Troude, L., Molliqaj, G., Nouri, A., Tessitore, E., 2022. Risk factors for the development of degenerative cervical myelopathy: a review of the literature. Neurosurgical Review, 1-15.

Bogduk, N., Mercer, S., 2000. Biomechanics of the cervical spine. I: Normal kinematics. Clinical biomechanics 15, 633-648.

Boschman, J.S., Frings-Dresen, M.H., Van Der Molen, H.F., 2015. Use of ergonomic measures related to musculoskeletal complaints among construction workers: a 2-year follow-up study. Safety and health at work 6, 90-96.

Cazzola, D., Holsgrove, T.P., Preatoni, E., Gill, H.S., Trewartha, G., 2017. Cervical spine injuries: a whole-body musculoskeletal model for the analysis of spinal loading. PloS one 12, e0169329.

Choi, H., Varghese, V., Baisden, J., Braza, D.W., Banerjee, A., Yoganandan, N., 2021. Upright magnetic resonance imaging study of cervical flexor/extensor musculature and cervical lordosis in females after helmet wear. Military Medicine 186, 632-638.

Christensen, S.W.M., Palsson, T.S., Krebs, H.J., Graven-Nielsen, T., Hirata, R.P., 2023. Prolonged slumped sitting causes neck pain and increased axioscapular muscle activity during a computer task in healthy participants–A randomized crossover study. Applied Ergonomics 110, 104020.

Damsgaard, M., Rasmussen, J., Christensen, S.T., Surma, E., De Zee, M., 2006. Analysis of musculoskeletal systems in the AnyBody Modeling System. Simulation Modelling Practice and Theory 14, 1100-1111.

Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE transactions on biomedical engineering 54, 1940-1950.

Delp, S.L., Loan, J.P., 1995. A graphics-based software system to develop and analyze models of musculoskeletal structures. Computers in biology and medicine 25, 21-34.
Diyana, M.A., Karmegam, K., Shamsul, B., Irniza, R., Vivien, H., Sivasankar, S., Syahira, M.P.A., Kulanthayan, K., 2019. Risk factors analysis: Work-related musculoskeletal disorders among male traffic policemen using high-powered motorcycles. International Journal of Industrial Ergonomics 74, 102863.
Gao, K., Zhang, J., Lai, J., Liu, W., Lyu, H., Wu, Y., Lin, Z., Cao, Y., 2019. Correlation between cervical lordosis and cervical disc herniation in young patients with neck pain. Medicine 98.
Garces, G.L., Medina, D., Milutinovic, L., Garavote, P., Guerado, E., 2002. Normative database of isometric cervical strength in a healthy population. Medicine & Science in Sports & Exercise 34, 464-470.
GBD, 2019. Global Burden of Disease Study 2019 (GBD 2019) Results., in: Seattle, U.S.I.f.H.M.a.E.I., 2020. (Ed.).
Guidotti, T., 1992. Occupational repetitive strain injury. American Family Physician 45, 585-592.
Hanks, M.M., Sefton, J.M., Oliver, G.D., 2018. Neck kinematics and electromyography while wearing head supported mass during running. Aerospace medicine and human performance 89, 9-13.
Harrison, M.F., Forde, K.A., Albert, W.J., Croll, J.C., Neary, J.P., 2016. Posture and helmet load influences on neck muscle activation. Aerospace medicine and human performance 87, 48-53.
Ivancic, P.C., Ito, S., Panjabi, M.M., 2007. Dynamic sagittal flexibility coefficients of the human cervical spine. Accident Analysis & Prevention 39, 688-695.
Jordan, A., Mehlsen, J., Bülow, P.M., Østergaard, K., Danneskiold-Samsøe, B., 1999. Maximal isometric strength of the cervical musculature in 100 healthy volunteers. Spine 24, 1343.
Kent, R., Forman, J., Bailey, A.M., Funk, J., Sherwood, C., Crandall, J., Arbogast, K.B., Myers, B.S., 2020. The biomechanics of concussive helmet-to-ground impacts in the National Football League. Journal of biomechanics 99, 109551.
Kim, M.G., Kim, K.-S., Ryoo, J.-H., Yoo, S.-W., 2013. Relationship between occupational stress and work-related musculoskeletal disorders in Korean male firefighters. Annals of occupational and environmental medicine 25, 1-7.
Lee, Y., Park, D., Kim, Y.M., 2023. The effect of wearing a head-mounted display on the boundaries of the cervical range of motion based on perceived comfort in a static posture. Virtual Reality 27, 815-828.
Mathys, R., Ferguson, S., 2012. Simulation of the effects of different pilot helmets on neck loading during air combat. Journal of biomechanics 45, 2362-2367.
Miller, A.E.J., MacDougall, J., Tarnopolsky, M., Sale, D., 1993. Gender differences in strength and muscle fiber characteristics. European journal of applied physiology and occupational physiology 66, 254-262.
Mokhtarzadeh, H., Bagheri, S., 2023. Streamlining C3D File Processing and Visualization: A User-Friendly Approach Using Google Colab and Open-Source Python Packages.
Mortensen, J.D., Vasavada, A.N., Merryweather, A.S., 2018. The inclusion of hyoid muscles improve moment generating capacity and dynamic simulations in musculoskeletal models of the head and neck. PloS one 13, e0199912.
Mortensen, J.D., Vasavada, A.N., Merryweather, A.S., 2020. Sensitivity analysis of muscle properties and impact parameters on head injury risk in American football. Journal of biomechanics 100, 109411.
Newman, P., Riches, A., Mara, J., Spratford, W., 2022. The effect of helmet mass and aircraft acceleration on cervical spine loads during typical fast jet aircraft pilot head motions. Journal of science and medicine in sport 25, 855-860.
NFIRS, 2019. Fire-Related Firefighter Injuries Reported to the National Fire Incident Reporting System (2015-2017), Topical Fire Report Series. U.S. Fire Administration, Emmitsburgh, MD.

Nimbarte, A.D., Sivak-Callcott, J.A., Zreiqat, M., Chapman, M., 2013. Neck postures and cervical spine loading among microsurgeons operating with loupes and headlamp. IIE Transactions on Occupational Ergonomics and Human Factors 1, 215-223.

Park, H., Park, J., Lin, S.-H., Boorady, L.M., 2014. Assessment of Firefighters' needs for personal protective equipment. Fashion and Textiles 1, 1-13.

Peterson, G., Pihlström, N., 2021. Factors associated with neck and shoulder pain: a cross-sectional study among 16,000 adults in five county councils in Sweden. BMC Musculoskeletal Disorders 22, 1-9.

Torma-Krajewski, J., Steiner, L.J., Lewis, P., Gust, P., Johnson, K., 2006. Ergonomics and mining; charting a path to a safer workplace.

Valente, G., Crimi, G., Vanella, N., Schileo, E., Taddei, F., 2017. nmsBuilder: Freeware to create subject-specific musculoskeletal models for OpenSim. Computer methods and programs in biomedicine 152, 85-92.

Van Dijke, G.H., Snijders, C., Roosch, E., Burgers, P., 1993. Analysis of biomechanical and ergonomic aspects of the cervical spine in F-16 flight situations. Journal of biomechanics 26, 1017-1025.

Vasavada, A.N., Li, S., Delp, S.L., 1998. Influence of muscle morphometry and moment arms on the moment-generating capacity of human neck muscles. Spine 23, 412-422.

Vicon, 2023. Plug-in Gait Reference Guide. Vicon Motion Systems.

Wang, S., Park, J., Wang, Y., 2021. Cross-cultural comparison of firefighters' perception of mobility and occupational injury risks associated with personal protective equipment. International journal of occupational safety and ergonomics 27, 664-672.