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RESEARCH ARTICLE

Sustained-till-exhaustion effects of firefighter helmets on neck muscle fatigue mechanism

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ABSTRACT
This study investigates how helmet inertial properties—mass and centre of mass (COM)—influence neck muscle fatigue to support the biomechanical design of firefighter helmets. Thirty-six firefighters (18 males, 18 females) performed sustained neck flexion and extension tasks under three conditions: no-helmet, US, and European-style (EU) helmets. Neck angles, endurance time, discomfort ratings, and electromyography (EMG) data from eight neck muscles were collected. Fatigue was assessed as an increase in normalised mean absolute value (NMAV) and decrease in median frequency (MF) of EMG signals, segmented into four intervals (0–25%, 26–50%, 51–75%, 76–100%).Piece-wise regression and ANOVA analyses of NMAV and MF slopes for each interval showed that the US helmet led to greater muscle activation, faster fatigue, and reduced endurance. These findings highlight the importance of optimising COM location—not just the weight—when designing a helmet to reduce neck injury risks.

Practitioner summary: Results indicated that although US-style helmets are lighter than European-style helmets, their high-profile design—with a COM shifted superiorly and anteriorly—resulted in more rapid neck muscle fatigue, suggesting practitioners to prioritize placing the helmet COM location closer to the head COM alongside reducing its weight during helmet design process.

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KEYWORDS
Firefighter helmet; centre of mass; neck flexion and extension; sustained exertion; muscle fatigue mechanism

Introduction

According to the Global Burden of Disease Report, neck pain ranks fourth among the most prevalent musculoskeletal disorders (MSDs), affecting more than 220 million individuals from 1990 to 2016 (2.9% of the total population) globally (GBD 2019). This high occurrence has been associated with various work and recreational activities involving prolonged sustained postures (Lin et al. 2020), awkward neck position (Nourollahi-Darabad et al. 2024), and repetitive neck movements (Falla, Farina, and Graven-Nielsen 2007). One such activity is the prolonged use of helmets in occupations such as firefighting (Lee et al. 2015), military operations (Siyeon and Wonyoung 2020), and airforce piloting (Murray et al. 2016). Especially, any adverse changes in the helmet's inertial properties—mass, centre of mass (COM), and moment of inertia (MOI)—can impose injury-prone loading on the cervical spine, strain neck passive tissues, and accelerate muscle fatigue.

Muscle fatigue is a complex physiological state of a muscle, primarily developed during prolonged voluntary muscle contractions through two mechanisms: 1) the accumulation of metabolites (e.g., lactic acid) within muscle fibers and 2) a reduction in the brain motor drives (Enoka and Duchateau 2008). The progression of muscle fatigue during a task performance alters muscle synergy (aka motor modules) (i.e., coordinated activation of muscle groups for specific movements (Cordo, Bell, and Harnad 1997) and creates proprioceptive deficits in joints, thus impairing the stability of a joint. Consequently, muscle fatigue has been identified as a precursor of inflammatory-type spinal MSDs (Larsson, Søgaard, and Rosendal 2007; Gallagher and Barbe 2022). As helmets impose an added load on the head and cervical spine regions, several previous studies investigated the effects of helmet inertial properties on neck muscle activity and fatigue. For instances, Phillips and Petrofsky (1983) and Bunketorp et al. (1985) found that the greater the helmet's inertia, the higher the rate of muscle fatigue in neck extensors muscles.

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Gallagher et al. (2008) investigated prolonged usages of Airforce helmets, wherein they observed a helmet with an anterior COM displacement to yield more neck discomfort than a regular helmet of similar weight. In contrast, Barrett et al. (2023) observed that helmet weight imposes a greater load on the cervical spine than its COM location. However, they did not study the effects on neck muscle fatigue. As first responders frequently engage in tasks requiring awkward and extreme neck postures, such as ceiling breaching (Dos Santos et al. 2025), bending over and crawling in confined spaces (Davis et al. 2014), and overhead work (Farooq 2022) it is essential to investigate the effects of heavyweight and/or imbalanced helmets on neck muscle fatigue, particularly at those awkward neck postures, in order to understand how such helmets could place substantial strain on the neck muscles.

In addition, modern helmets also serve as a platform for supporting additional functional accessories, such as communication devices, face-shield, lighting equipment, etc., leading to increased helmet weight and potential COM shift (Harrison et al. 2016). Prior studies on helmets (with functional accessories) used in military, law enforcement, and airforce settings reported increased neck musculature strain (Murray et al. 2016; Siyeon and Wonyoung 2020; Healey et al. 2021), adverse changes in neck posture (Forde et al. 2011; Mills, Tvaryanas, and Wade 2019), and muscle fatigue (Barker and Albery 2010; Mills, Tvaryanas, and Wade 2019). Similarly, firefighter helmets are bulkier and heavier (Wang, Chen, and Yu 2022) than aforementioned professional helmets because of the requirement of a two-layered helmet shell to withstand extreme heat and provide impact protection (NFPA 2018). Heavier weight can alter the COM of helmets, leading to increased MOI. This elevated MOI requires greater muscular effort to control head movements during flexion and extension, thereby increasing torque demands on the neck. Over time, these heightened demands can accelerate muscle fatigue and elevate the risk of neck-related injuries. Several survey-based studies on firefighters reported that helmet size and weight primarily contribute to neck discomfort (Lee et al. 2015; Wang, Chen, and Yu 2022) and reduced neck mobility (Park et al. 2014; Wang, Park, and Wang 2021). Between 2016 and 2020, about 23% (27,150 out of 118,070 cases) of firefighting injuries occurred in head and neck regions, with 5.5% linked to sprain-type injuries (Campbell and Molis 2022). Despite this, the biomechanical investigation of firefighter helmets has remained underexplored, with only one previous study addressing the effects on cervical spinal kinematics (Paulon et al. 2024). As firefighters often work in awkward postures and wear heavy and bulky helmets for a prolonged duration (Gentzler and Stader 2010), it is essential to explore how prolonged helmet use affects neck muscle fatigue to uncover the pathomechanism of associated neck MSDs.

Therefore, this study aimed to investigate how the sustained-till-exhaustion use of two firefighter helmets with distinct inertial properties influences neck muscle activation and fatigue mechanisms at maximum flexion and extension positions. We hypothesised that a helmet with heavier mass and a COM further away from the atlantooccipital joint (C0-C1 joint), the center of head rotation, would affect the neck muscle fatigue process. These findings are expected to provide actionable insights to practitioners in order to design a biomechanically optimised helmet, thereby reducing the risk of neck injuries among firefighters.

Materials and methods

Participants

We recruited 36 firefighters (18 males and 18 females) who were in good health and did not have any recent history of musculoskeletal injuries, disorders, or surgeries. Their average age, weight, height, and BMI were respectively as follows (mean ± standard deviation): 35.14 ± 8.91 years (male: 39.18 ± 7.23 years; female: 31.33 ± 8.83 years), 79.27 ± 19.01 kg (male: 93.22 ± 16.68 kg; female: 66.78 ± 10.36 kg), 171.12 ± 9.60 cm (male: 178.82 ± 6.35 cm; female: 164.23 ± 6.10 cm), and 26.84 ± 4.8 kg/m^2 (male: 28.8 ± 5.39 kg/m^2; female: 24.2 ± 2.85 kg/m^2). Before the experimental tasks, they signed a study consent form approved by the local Institutional Review Board (IRB2020-708).

Experimental protocol and instrumentation

Task design

Participants performed two sustained-till-exhaustion tasks in *maximum* neck flexion and extension postures under three different helmet conditions in a randomised order: no-helmet (NH), US traditional firefighter (US) helmet, and European firefighter (EU) helmet.

We first acquired each participant's maximum (static) neck flexion and extension postures for five seconds, by instructing them, respectively, to flex and extend their head-neck system as far as possible without invoking active stretching of passive tissues such as fascia or ligaments. These postures served as the baselines for the main tasks of sustained-till-exhaustion neck flexion and extension exertions. During sustained flexion tasks, participants bent their heads forward until the chin approached their chest in order to adopt the baseline maximum neck flexion position. They were encouraged to hold this position until exhaustion (Figure 1A). Similarly, during maximum extension tasks, they were instructed to extend their heads backward as far as comfortably possible, i.e., adopting the baseline maximum neck extension posture (Figure 1B), and hold the posture until exhaustion. These postures were chosen to simulate extreme yet active cervical positions without entering a fully passive or end-range stretch, which could reduce muscular engagement and confound fatigue-related measurements. Each participant performed two repetitions of the sustained-till-exhaustion task for each helmet condition. To minimise carryover fatigue, rest intervals were set at least twice the duration of the preceding exertion. The subjective discomfort levels were assessed using Borg's CR-10 scale (Borg 1998) before and at the end of each trial. Only when the subjects rated Borg's score of 0 and 1, they were allowed to perform the next trial. All

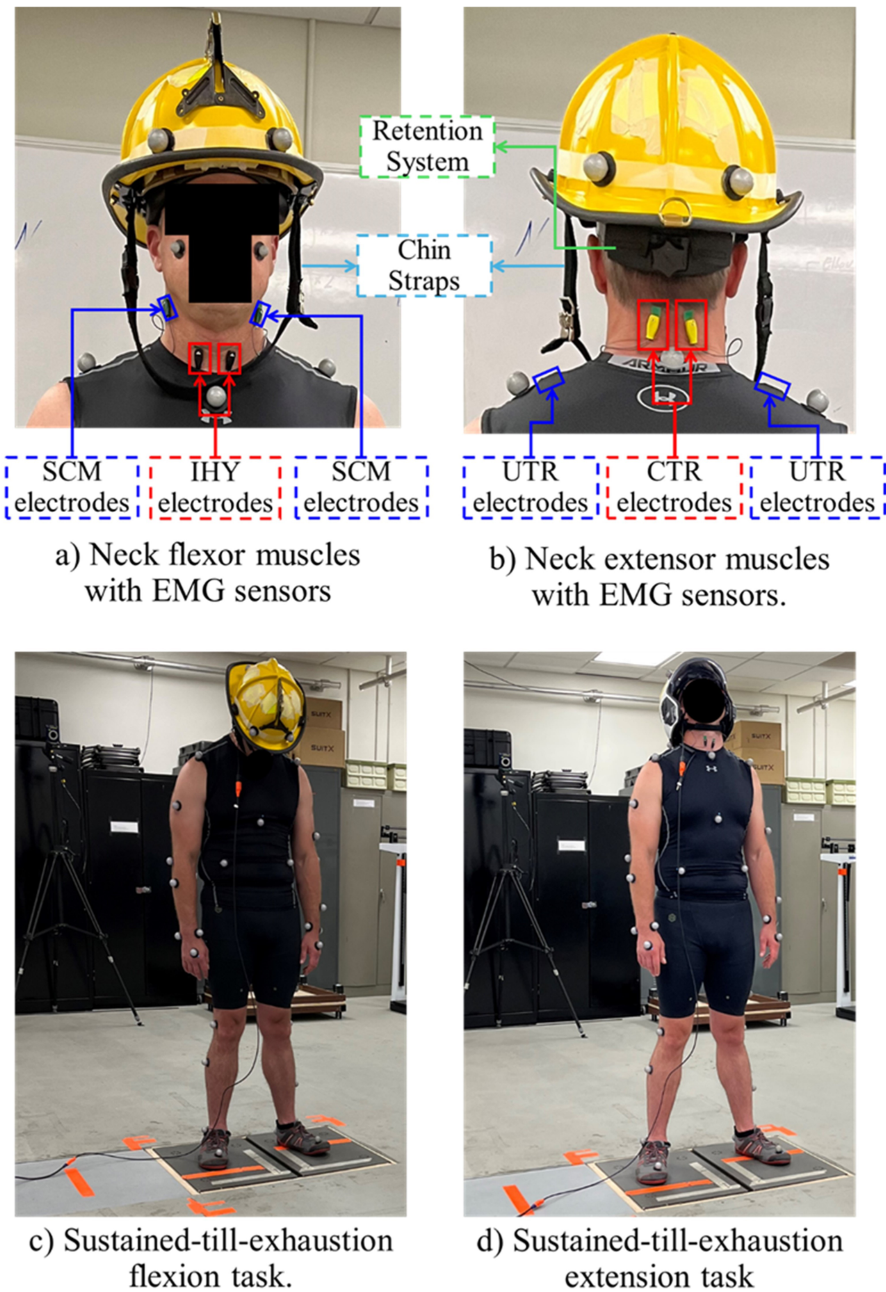

Figure 1. A schematic presentation of data acquisition process: (a) electromyography (EMG) electrodes placed on infrahyoid (IHY) and sternocleidomastoid (SCM) muscles, (b) EMG electrodes placed on upper trapezius (UTR) and cervical trapezius (CTR) muscles, (c) subject performing sustained-till-exhaustion flexion tasks using a US-style firefighter (US) helmet (model: Bullard UM6WH; integrated accessories: visor), (d) subject performing sustained-till-exhaustion extension tasks with a European-style firefighter (EU) helmet (Model: Cairns XF1; integrated accessories: visor, face-shield, communication, and lighting device).

firefighter helmets were securely fastened using both the chin strap and internal retention system to comply with the National Fire Protection Association (NFPA) guidelines. These systems prevent any relative movement between helmet and head as well as ensure consistent alignment across subjects during data collection (Figure 1C and D).

Surface Electromyography (EMG)

We used a surface EMG system (Delsys Inc, Natick, Massachusetts, USA) with Trigno Quattro sensors (12 × 25 × 7 mm bar electrode, 10 mm inter-electrode distance, Ag-AgCl materials; 2222 Hz sampling rate; and 300 gain factor) to acquire muscle activity data from four neck flexor muscles–left and right sternocleidomastoid (SCM) and infrahyoid (IHY)–and four neck extensor muscles–left and right cervical-trapezius at C4–C5 level (CTR) and upper-trapezius (UTR) (Figure 1C and D). These muscles were chosen due to their relevance in facilitating neck mobility and are widely used to study neck strength, endurance, and fatigue (Sommerich et al. 2000; Chowdhury et al. 2022). EMG sensors were positioned on respective muscle bellies in alignment with the muscle fibres, in accordance with previous studies (Vasavada, Peterson, and Delp 2002; Netto and Burnett 2006). Briefly, electrodes were placed over the upper trapezius muscle approximately midway between the acromion and the spinous process of C7, the cervical trapezius muscle along the posterolateral aspect of the neck at the level of C3–C5, the sternocleidomastoid muscle in the anterolateral region of the neck, and the infrahyoid muscle group in the anterior neck region, beneath the hyoid bone and superficial to the thyroid cartilage.

Prior to EMG sensor placement, muscle skins were properly shaved and cleaned with 70% isopropyl alcohol. A total of three five-second maximum voluntary contractions (MVC) for each pair of bilateral neck muscles were performed, with the maximum EMG value of the highest MVC exertion considered for normalisation. To collect MVC data, participants were seated upright in a chair with their arms resting parallel to their trunk in a relaxed position and their feet elevated off the ground and performed forward and backward static neck exertions while their head was securely restrained, in accordance with the MVC protocols mentioned in previous studies (Vasavada, Li, and Delp 2001; Chowdhury et al. 2022). Briefly, the MVC protocols include performing forward head exertions for the neck extensor muscles and maximum backward head exertions for the neck flexor muscles against resistance in neutral head-neck postures. During MVC exertions, verbal encouragement was provided to the participants to exert with their maximum efforts. Each MVC trial was followed by at least a two-minute rest between the MVC trials.

Head-neck kinematics and helmet imaging

We recorded full-body kinematics using a 10-camera Kestrel-1300 motion capture system (Motion Analysis Corporation, Rohnert Park, CA, USA) with a sampling rate of 60 Hz, which was time-synchronized with the EMG system. We used a full-body plug-in gait marker-set protocol (Vicon 2023). During helmet trials, head markers were placed on the outer shell of the helmet. Furthermore, we imaged each helmet using a handheld 3D scanner (EinScan HX, Shining 3D, Hangzhou, China; sampling rate: 20 Hz).

Data analysis

EMG analysis

Data from 32 participants (15 males and 17 females) were analysed, with four participants being excluded due to EMG acquisition issues. EMG analysis focused on the period from maximum neck flexion or extension to the point of task termination. EMG signals were filtered using a band-pass filter (10–500 Hz) and a notch filter to remove the powerline noise (60 Hz) and its harmonics. The filtered signals were then analysed in both frequency and time domains, where a simultaneous increase in amplitude and a decrease in frequency was identified as muscle fatigue biomarkers (Luttmann et al. 1996; Chowdhury et al. 2013; Wei and Chowdhury 2025). We calculated median frequency (MF) using Fast Fourier Transform (FFT) and mean absolute value (MAV) by using a moving window of 125 milliseconds. The MAV of each muscle was then normalised by the maximum amplitude of each muscle's maximum MVC trial, yielding the normalised mean absolute value (NMAV) for each muscle. EMG data from both left and right muscle pairs were pooled due to similar magnitudes and trends.

Kinematics analysis

Neck kinematics were calculated from raw marker data using Cortex-9 software (Motion Analysis Corporation, California, USA). A total of four markers' positions placed on the head—right orbitrate (RORB), left orbitrate (LORB), right back head (RBHD), and left back head (LBHD)—were exported to perform angle calculations during sustained flexions and extension neck postures. Once the markers were exported, we first defined a transverse plane at the initial (neutral) posture that passed through the four head markers.

At each increment of neck flexion or extension, a Frankfurt plane—defined as the plane passing through the RORB, LORB, and the midpoint between RBHD and LBHD—was calculated. Over time, the angles between the Frankfurt plane and the transverse plane were computed to determine flexion (α) and extension (β) angles (Figure 2). For helmet tasks, we positioned the markers of the head on the helmets and the angles were calculated in a similar way.

Furthermore, we reverse-engineered 3D scanned images of both helmets and created their CAD models. We used ANSA finite-element (FE) platform (Beta CAE, Lucerne, Switzerland) to estimate helmet inertial properties, such as mass, COM, and MOI (Figure 3a and b), by inputting their geometric and material properties and mounting each helmet FE model onto an MRI-based FE head model of one of our study participant's head (height: 1.84 m; weight: 121.3 kg). The helmet COM locations were calculated relative to the C0-C1 joint.

Statistical analysis

Previous research showed that the progression of muscle fatigue follows a nonlinear pattern, with a cubic model displaying the fatigue progression with more precision than the linear models, especially the MF drifts occurring more predominantly at 25%, 50%, and 75% of task duration (Chowdhury and Nimbarte 2015). However, it is challenging to interpret and locate the causality of muscle fatigue development using continuous cubic model. To address this, it is essential to discretize the nonlinear trend by approximating linear trends at key time intervals where distinct behaviours emerged. Therefore, this study conducted piecewise linear regression analyses to estimate both MF and NMAV slopes in four-time intervals $-0\sim25\%$ (I_1), $25\sim50\%$ (I_2), $50\sim75\%$ (I_3), and $75\sim100\%$ (I_4) of task completion time (Equation 1):

$$y_i = \beta_0 + \beta_1 x_i + \beta_2 (x_i - 25) D_{25} + \beta_3 (x_i - 50) D_{50} + \beta_4 (x_i - 75) D_{75} \tag{1}$$

where y_i is the dependent measure at each i^{th} (i=1, 2, 3,…100) time point; x_i is the percent task completion time; D_i are dummy variables $(0\, if\, x_i \le i\, and\, 1 if\, x_i > i)$; $\beta_1, \beta_2, \beta_3$, and β_4 are slopes coefficients for 0–25%, 25–50%, 50–75%, and 75–100% time intervals (quartiles) of the task, respectively; and β_0 is the intercept.

We calculated descriptive statistics (mean and standard deviation) of MF, NMAV, MF slope, NMAV slope, and neck angle, in addition to the percentage of total trials displaying positive or negative NMAV and MF slopes at each interval for each muscle and helmet type. As a regression trend can be influenced by outliers or a few extreme cases, we calculated the percentage of trials showing positive NMAV slopes and negative MF slopes to ensure that the observed fatigue trends are consistent across trials as well as to interpret the trends with more distribution-aware inferences. Before statistical tests, we verified data normality and homoscedasticity assumptions with Shapiro-Wilk and Levene's tests (α=0.05), respectively. As some data violated the assumption of homogeneity of variances as assessed by Levene's test, a log transformation was applied to stabilise the variance. Once the data met Levene's test criteria, we performed a repeated

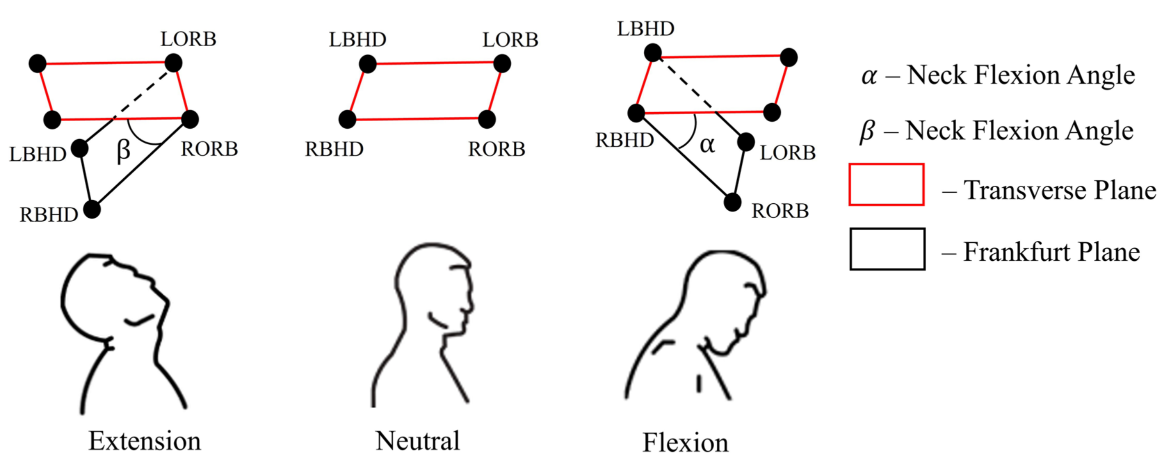

Figure 2. Schematic representation of head marker placement and angle calculation procedure during neutral, neck flexion, and extension postures. Markers were positioned at four locations: right orbitrate (RORB), left orbitrate (LORB), right back head (RBHD), and left back head (LBHD). The transverse plane was defined at the neutral head posture, while the Frankfurt plane was established at each time step, passing through the RORB, LORB, and the midpoint between RBHD and LBHD. The angles between these planes were computed to determine flexion (α) and extension (β) angles.

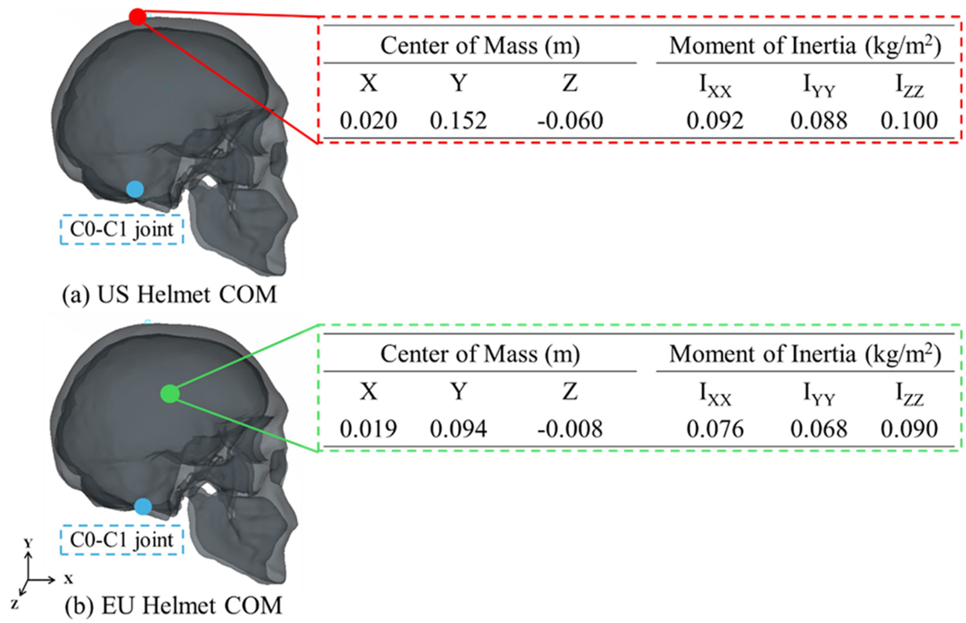

Figure 3. Representation of the COM position and moment of inertia information for (a) the US Helmet and (b) the EU Helmet with respect to the C0-C1 joint modelled in the FE model.

measures analysis of variance (ANOVA) to analyse the effects of *helmet condition* (NH, US, EU), *interval* (I_1; I_2; I_3; I_4), and *helmet*×*interval* interaction on NMAV and MF slopes and average neck angle, MF, and NMAV values for individual muscle groups. We calculated eta squared (η^2) as a measure of effect size to quantify the proportion of variance explained by each factor. Values of η^2 were interpreted as small (0.01), medium (0.06), and large (0.14), based on conventional thresholds (Cohen 2013; Lakens 2013). If significance was reached, we proceeded with post-hoc analysis using Tukey's test. Furthermore, we also performed univariate ANOVAs to test the effect of helmet condition on secondary variables such as subjective discomfort and endurance time (ET). All significance tests were performed at a 95% confidence level (α=0.05), and Bonferroni correction was used for the pairwise comparisons. All statistical analyses were performed at SPSS 29.0 (SPSS Inc., Chicago, Illinois, U.S.A.).

Results

Neck angle

Significant helmet effects on neck angles were observed during flexion and extension tasks across all time intervals (Figure 4). Both US (55.29±1.12°) and EU (55.37±2.12°) helmets resulted in significantly (p-value < 0.01, η^2 = 0.29) greater neck angles—by 17.41% and 17.58%, respectively—compared to the NH condition (47.09±1.35°) during sustained neck flexion tasks. However, no statistically significant difference was found between the US and EU helmet conditions ($p > 0.05$). During neck extension, the NH condition exhibited significantly ($p < 0.01$, η^2 = 0.15) higher neck extension angles (67.17±2.33°), about 12.83% and 12.80% greater than US (59.53±1.26) and EU (59.55±0.92) helmet conditions, respectively. No statistically significant time interval effects were reported. Nevertheless, subtle increasing trends were observed in neck kinematics during both sustained neck flexion and extension tasks. Between the first- and fourth-time intervals during flexion tasks, neck angles increased by 1.93% in the NH condition and 4.63% in the US Helmet condition, whereas they decreased slightly by 0.54% for the EU Helmet condition. Nonetheless, during extension tasks, all conditions exhibited an increase in neck angles between the first and fourth intervals, with corresponding increases of 2.57%, 1.69%, and 0.84% for NH, US, and EU Helmet conditions, respectively.

UTR and CTR muscle activity during neck flexion tasks

NMAV and MF slopes and interval averages of UTR muscle

No significant helmet and interaction effects were observed for either NMAV or MF slopes. However, overall interval effects were highly significant for MF slopes (p-value = 8.49e-5, η^2 = 0.05), with post-hoc tests (Table 1) and the trends in Figure 5 showing decreases in MF across all helmet conditions, particularly between I_1 vs. I_2, I_1 vs. I_3, and I_2 vs. I_4 intervals.

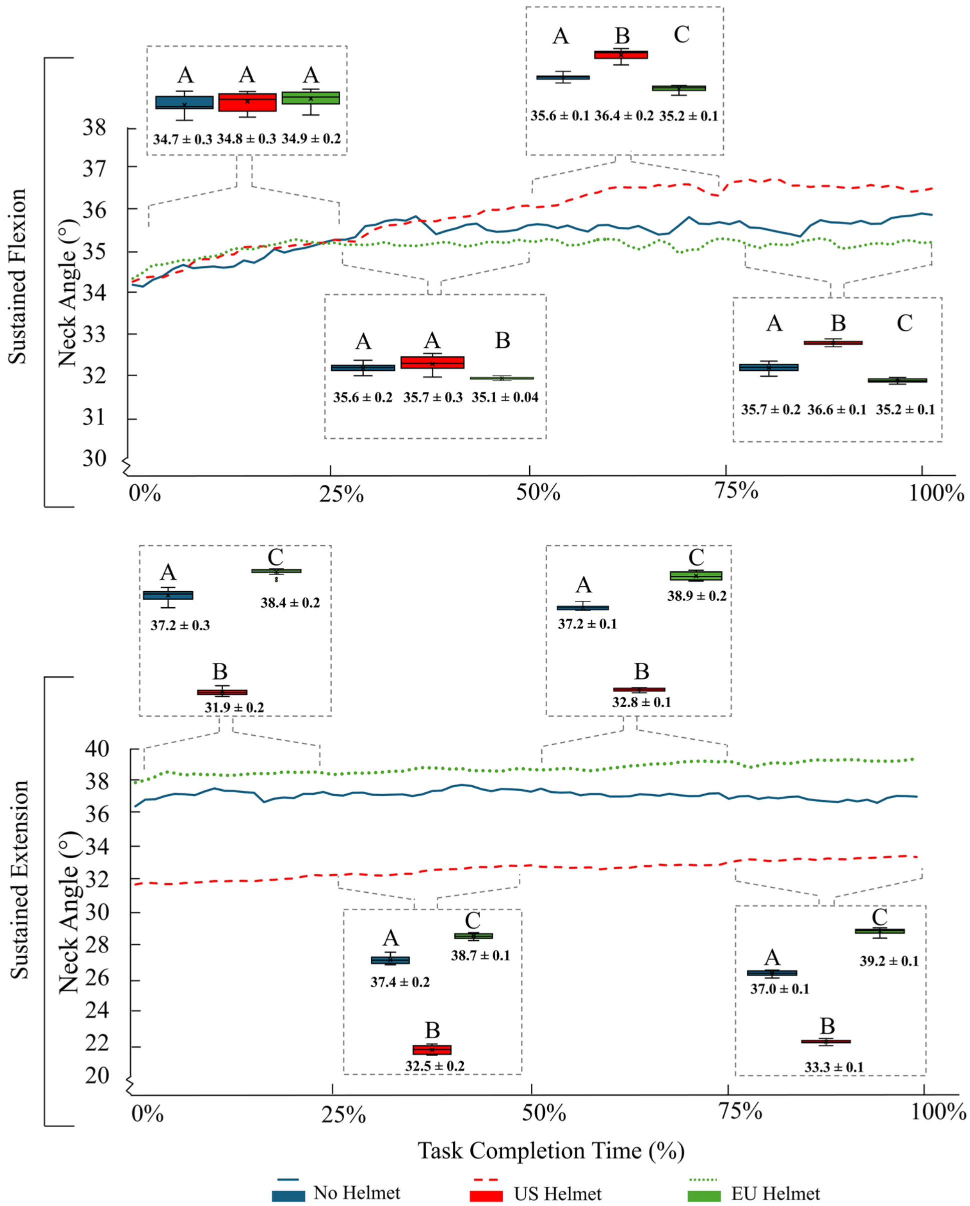

Figure 4. Neck angle during sustained-till-exhaustion flexion (left) and extension (right) tasks for No-helmet, US-style firefighter (US) helmet, and European-style (EU) helmet conditions. Solid lines represent the average neck angle over time, and shaded regions show the standard error.

In contrast, helmet use significantly affected interval averages of NMAV (p-value = 6.64e-5, η^2 = 0.03), with EU helmets displaying lower NMAV values than the NH condition (about I_1 = −8.9%, I_2 = −11.3% I_3 = −12.1%, and I_4 = −15.6% less), while the US helmets showed the highest values, specifically in the I_4 interval (p-value = 0.04, η^2 = 0.02). The interval effect significantly influenced the interval averages of MF, with significant decreases between I_1 vs. I_3 and I_1 vs. I_4.

NMAV and MF slopes and interval averages of CTR muscle

Significant interval effects were observed for NMAV (p-value = 0.01, η^2 = 0.02) and MF (p-value = 9.21e-4, η^2 = 0.03) slopes. Post-hoc results (Table 1) and trends (Figure 6) demonstrated significant decreases in MF slopes and increases in NMAV slopes between I_1 vs. I_2 and I_2 vs. I_4 intervals across all helmet conditions. Helmet × interval interaction effects were also significant for NMAV (p-value = 2.00 e-4, η^2 = 0.05) and MF slopes (p-value = 6.00 e-3, η^2 = 0.03). Specifically, NMAV slopes showed significant increases under the US helmet in the I_1, I_2, and I_4 intervals and under the EU helmet in the I_2 interval. On the contrary, in comparison to the NH condition, helmet use significantly increased interval averages of NMAV values (p-value = 9.55e-16, η^2 = 0.04) under both US (I_1 = 21.2%, I_2 = 27.8%, I_3 = 25.3%, I_4 = 15.9%) and EU (I_1 = 21.2%, I_2 = 28.7%, I_3 =

Table 1. Analysis of variance results (p-values) displaying the effects of helmet, interval, and helmet×interval interaction on NMAV and MF Slopes (top) and NMAV and MF averages (bottom) of neck *extensor* muscles during sustained-till-exhaustion neck *flexion* tasks.

		NMAV Slopes		MF Slopes	
		Upper Trapezius (UTR)	Cervical Trapezius (CTR)	Upper Trapezius (UTR)	Cervical Trapezius (CTR)
Helmet	ANOVA	0.40	0.46	0.25	0.32
	Post-hoc	–	–	–	–
Interval	ANOVA	0.20	0.01*	8.49e-5**	9.00 e-4**
	Post-hoc	–	I_1 vs. I_2 = 0.04* I_2 vs. I_4 = 0.02*	I_1 vs. I_2 = 2.08 e-6** I_1 vs. I_3 = 5.21 e-4** I_2 vs. I_4 = 4.08 e-3**	I_1 vs. I_2 = 0.03* I_2 vs. I_4 = 8.00 e-4**
Helmet × Interval	ANOVA	0.25	2.00 e-4**	0.18	6.00 e-3**
	Post-hoc	–	US1 vs. US2 = 5.00 e-3** NH2 vs. EU2 = 0.01* US2 vs. NH2 = 5.00 e-3** US4 vs. US2 = 0.04*	–	NH1 vs. NH4 = 0.02
		Average NMAV values		**Average MF values**	
		Upper Trapezius (UTR)	Cervical Trapezius (CTR)	Upper Trapezius (UTR)	Cervical Trapezius (CTR)
Helmet	ANOVA	6.64e-5**	9.55e-16**	0.08	2.00 e-3**
	Post-hoc	NH vs. EU = 1.33 e-7**, US vs. EU= 1.63 e-3**	NH vs. EU = 6.70 e-13**, US vs. NH = 1.76 e-11**	–	NH vs. EU = 1.27 e-3**
Interval	ANOVA	0.149	0.757	0.0001**	0.135
	Post-hoc	–	–	I_1 vs. I_3 = 0.05* I_1 vs. I_4=7.23 e-4**	–
Helmet × Interval	ANOVA	0.99	0.99	0.99	0.90
	Post-hoc	–	–	–	–

The p-values below 0.05 and 0.01 are respectively marked with single and double asterisks. The first, second, third, and fourth interval levels are denoted by I_1, I_2, I_3, and I_4, respectively. The no-helmet, US-style firefighter helmet, and European-style helmet conditions are indicated by NH, US, and EU, respectively, and their significance level at any of the four intervals with 1, 2, 3, and 4. For example, EU2 refers to the effect of a European-style helmet in the second interval.

21.4%, I_4 = 11.6%) helmets, while decreased interval averages of MF (p-value = 2.00 e-4, η^2 = 0.02), particularly under US (I_1 = −4.4%, I_2 = −4.3%, I_3 = −2.9%, I_4 =-0.8%) helmet conditions (Table 1; Figure 6).

Percentage of trials with positive NMAV and negative MF slopes

In UTR muscles, about 55% ± 4.4%, 55.75% ± 4.99%, and 55.75% ± 5.68% of trials with positive NMAV slope values and 50.5% ± 8.96%, 50.25% ± 8.89%, and 55.25% ± 4.65% of trials with negative MF slope values were observed across all intervals under NH, US, and EU helmet conditions, respectively (Table 2). Likewise, for CTR muscles, approximately 54.5% ± 11.62%, 53.5% ± 8.35%, and 53.75% ± 11.24% of trials with positive NMAV slope values and 55.25% ± 7.37%, 55.25% ± 6.18%, and 48.25% ± 9.03% of trials with negative MF slope values were respectively found across all intervals for the same helmet conditions.

IHY and SCM muscle activity during neck extension

NMAV and MF slopes and interval averages of IHY muscle

No significant effects of helmet or interaction were detected for either NMAV or MF slopes. However, interval effects were significant for both NMAV (p-value = 9.53e-9, η^2 = 0.07) and MF slopes (p-value = 2.34e-7, η^2 = 0.06), with post-hoc results indicating a significant increase in NMAV slopes and a decrease in MF slopes across the majority of time intervals (Table 3; Figure 7). On the contrary, helmet effects on interval averages of NMAV were significant (p-value = 2.00e-16, η^2 = 0.23) across all intervals (US: I_1 = +144%, I_2 = +152%, I_3 = +133%, I_4 = +73.61%; EU: I_1 = +33.89%, I_2 = +38.60%, I_3 = +33.89%, I_4 = +37.50%). Though no significant interval or interaction effects were detected for the MF interval averages, they trended towards significance (p-value = 0.05, η^2 = 7.34 e-3).

NMAV and MF slopes and averages of SCM muscle

Significant interval effects were observed for NMAV (p-value = 1.03 e-10, η^2 = 0.1) and MF (p-value = 5.00 e-4, η^2 = 0.04) slopes, characterised by increasing NMAV slopes and decreasing MF slopes from I_1 to I_4 intervals (Table 3). The interaction was significant for NMAV slopes (p-value = 1.20 e-3, η^2 = 0.04), with significant increases from I_1 to I_2 for US helmets and I_2 to I_4 for EU helmets. In contrast, interval averages of both NMAV and MF showed significant effects of helmet (p-value = 7.48e-5, η^2 = 0.01) and interval (p-value = 2.00 e-4, η^2 = 0.02), with consistently higher NMAV values (US: I_1 = +24.58%, I_2 = +32.52%, I_3 = +34.73%,

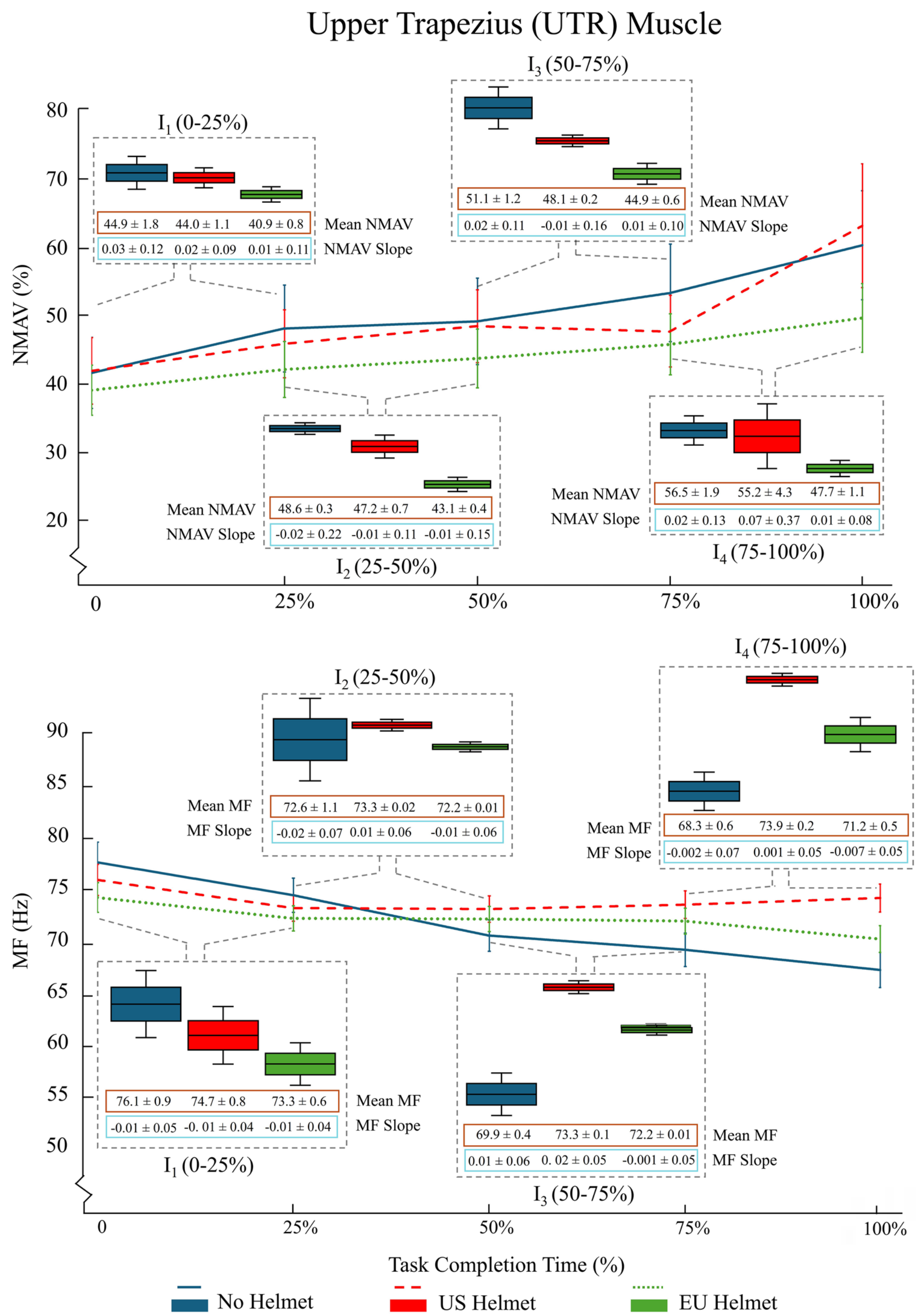

Figure 5. Piecewise linear regression plots of normalised mean absolute value (NMAV) and median frequency (MF) of upper trapezius (UTR) muscles during sustained-till-exhaustion flexion tasks under no helmet, US-style firefighter (US) helmet, and European-style (EU) helmet conditions. Boxplots represent descriptive data (average and standard deviation) of NMAV and MF averages and slope values of the corresponding time intervals. The abbreviations I_1, I_2, I_3, and I_4 represent the sequential time intervals used in the analysis, corresponding to each quartile of the task duration.

I_4 = +18.80%; EU: I_1 = +42.42%, I_2 = +35.63%, I_3 = +34.63%, I_4 = +33.52%) and lower MF values (US: I_1 = −7.54%, I_2 = −6.91%, I_3 = −5.09%, I_4 = −1.25%; EU: I_1 = −2.52%, I_2 = −0.73%, I_3 = −0.47%, I_4 = −1.10%) under helmet conditions than the NH conditions (Table 3, Figure 8).

Percentage of trials with positive NMAV and negative MF slopes

Across all time intervals, approximately 53.75% ± 10.72%, 54.5% ± 7.55%, and 50.0% ± 2.16% of trials exhibited positive NMAV slopes, while 50% ± 13.19%, 52.25% ± 12.23%, and 54.5% ± 13.40% of trials showed

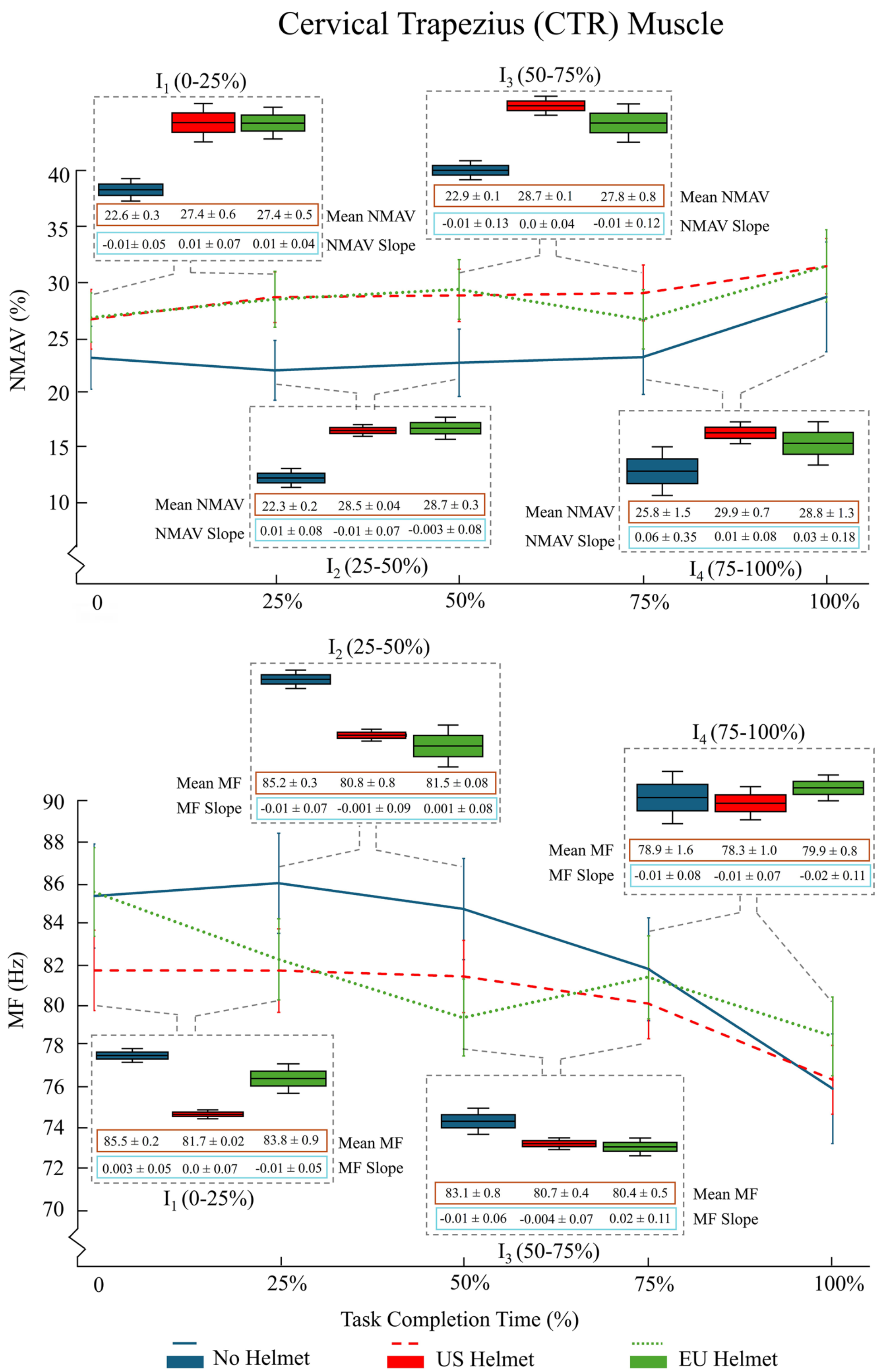

Figure 6. Piecewise linear regression plots of normalised mean absolute value (NMAV) and median frequency (MF) of cervical trapezius (CTR) muscles during sustained-till-exhaustion flexion tasks under no helmet, US-style firefighter (US) helmet, and European-style (EU) helmet conditions. Boxplots represent descriptive data (average and standard deviation) of NMAV and MF averages and slope values of the corresponding time intervals. The abbreviations I_1, I_2, I_3, and I_4 represent the sequential time intervals used in the analysis, corresponding to each quartile of the task duration.

negative MF slopes in SCM muscles for NH, US, and EU helmet conditions, respectively (Table 2). Similarly, for IHY muscles, about 57.25% ± 11.62%, 51.0% ± 8.18%, and 53.75% ± 18.69% of trials demonstrated positive NMAV slopes, and 51.0% ± 2.83%, 53.75% ± 4.57%, and 50.5% ± 8.96% of trials respectively exhibited negative MF slopes across all time intervals for the same helmet conditions.

Table 2. The percentage of trials with positive normalised mean absolute value (NMAV) or median frequency (MF) slopes, %positive (+), and the percentage of trials with negative NMAV or MF slopes, %negative (–), are estimated for all four muscles across all four time intervals.

			%positive (+) / %negative (–) trials per time interval			
Muscle	Helmet Conditions	Parameters	0–25%	25–50%	50–75%	75–100%
Upper Trapezius (UTR)	**No Helmet**	NMAV Slope	(+60%/–40%)	(+57%/–43%)	(+53%/–47%)	(+50%/–50%)
		MF Slope	(+39%/–61%)	(+45%/–55%)	(+60%/–40%)	(+45%/–55%)
	US Helmet	NMAV Slope	(+61%/–39%)	(+49%/–51%)	(+56%/–44%)	(+57%/–43%)
		MF Slope	(+37%/–63%)	(+58%/–42%)	(+52%/–48%)	(+51%/–49%)
	EU Helmet	NMAV Slope	(+60%/–40%)	(+60%/–40%)	(+48%/–52%)	(+55%/–45%)
		MF Slope	(+45%/–55%)	(+56%/–44%)	(+48%/–52%)	(+50%/–50%)
Cervical Trapezius (CTR)	**No Helmet**	NMAV Slope	(+39%/–61%)	(+66%/–34%)	(+53%/–47%)	(+60%/–40%)
		MF Slope	(+50%/–50%)	(+49%/–51%)	(+46%/–54%)	(+34%/–66%)
	US Helmet	NMAV Slope	(+63%/–37%)	(+43%/–57%)	(+52%/–48%)	(+56%/–44%)
		MF Slope	(+40%/–60%)	(+53%/–47%)	(+46%/–54%)	(+40%/–60%)
	EU Helmet	NMAV Slope	(+63%/–37%)	(+39%/–61%)	(+51%/–49%)	(+62%/–38%)
		MF Slope	(+40%/–60%)	(+53%/–47%)	(+62%/–38%)	(+52%/–48%)
Infrahyoid (IHY)	**No Helmet**	NMAV Slope	(+41%/–59%)	(+64%/–36%)	(+57%/–43%)	(+67%/–33%)
		MF Slope	(+45%/–55%)	(+49%/–51%)	(+51%/–49%)	(+51%/–49%)
	US Helmet	NMAV Slope	(+41%/–59%)	(+60%/–40%)	(+49%/–51%)	(+55%/–45%)
		MF Slope	(+45%/–55%)	(+47%/–53%)	(+52%/–48%)	(+41%/–59%)
	EU Helmet	NMAV Slope	(+29%/–71%)	(+71%/–29%)	(+43%/–57%)	(+61%/–39%)
		MF Slope	(+55%/–45%)	(+44%/–56%)	(+59%/–41%)	(+40%/–60%)
Sternocleidomastoid (SCM)	**No Helmet**	NMAV Slope	(+38%/–62%)	(+57%/–43%)	(+58%/–42%)	(+62%/–38%)
		MF Slope	(+34%/–66%)	(+65%/–35%)	(+46%/–54%)	(+55%/–45%)
	US Helmet	NMAV Slope	(+54%/–46%)	(+44%/–56%)	(+60%/–40%)	(+60%/–40%)
		MF Slope	(+30%/–70%)	(+58%/–42%)	(+52%/–48%)	(+51%/–49%)
	EU Helmet	NMAV Slope	(+51%/–49%)	(+47%/–53%)	(+52%/48%)	(+50%/–50%)
		MF Slope	(+36%/–64%)	(+56%/–44%)	(+58%/–42%)	(+32%/–68%)

Table 3. Analysis of variance results (p-values) displaying the effects of helmet, interval, and helmet × interval interaction on normalised mean absolute value (NMAV) and median frequency (MF) Slopes (top) and average NMAV and MF values (bottom) of neck *flexor* muscles during sustained-till-exhaustion neck *extension* tasks.

		NMAV Slopes		MF Slopes	
		Infrahyoid (IHY)	Sternocleidomastoid (SCM)	Infrahyoid (IHY)	Sternocleidomastoid (SCM)
Helmet	ANOVA	0.71	0.49	0.51	0.69
	Post Hoc	–	–	–	–
Interval	ANOVA	9.53 e-9**	1.03 e-10**	2.34 e-7**	5.00 e-4**
	Post Hoc	I_1 vs. I_2 = 0.02*	I_1 vs. I_2= 2.28 e-6**	I_1 vs. I_2 = 3.08 e-3**	I_1 vs. I_2 = 0.02*
		I_1 vs. I_4 = 9.80 e-9**	I_2 vs. I_3 = 3.75 e-7**	I_1 vs. I_4 = 1.97 e-3**	I_1 vs. I_3 = 0.03*
		I_2 vs. I_4 = 8.53 e-3**	I_2 vs. I_4 = 6.55 e-10**	I_2 vs. I_3 = 3.35 e-5**	I_2 vs. I_4 = 7.07 e-3**,
				I_3 vs. I_4 = 1.95 e-5**	I_3 vs. I_4 = 0.01*
Helmet × Interval	ANOVA	0.42	1.22e-3**	0.39	0.23
	Post Hoc	–	US1 vs. US2 = 3.37 e-7**	–	–
			EU2 vs. EU4 = 3.02 e-4**		
			US2 vs. US3 = 2.52 e-7**		
			US2 vs. US4 = 5.36 e-6**		
		Average NMAV values		Average MF values	
		Infrahyoid (IHY)	Sternocleidomastoid (SCM)	Infrahyoid (IHY)	Sternocleidomastoid (SCM)
Helmet	ANOVA	2.00e-16**	7.48 e-5**	0.09	8.97e-9**
	Post Hoc	US vs. EU = 3.55 e-13**	US vs. EU = 5.52 e-3**	–	US vs. EU = 6.23 e-7**
		US vs. NH = 3.30 e-13**			US vs. NH = 1.09 e-7**
Interval	ANOVA	0.54	0.09	0.05	2.00 e-4**
	Post Hoc	–	–	–	I_1 vs. I_3 = 0.03*
					I_1 vs. I_4 = 2.03 e-4**
					I_2 vs. I_4 = 9.19 e-3**
Helmet × Interval	ANOVA	0.99	1.00	0.93	0.96
	Post Hoc	–	–	–	–

The p-values below 0.05 and 0.01 are respectively marked with single and double asterisks. The p-values below 0.05 and 0.01 are respectively marked with single and double asterisks. The first, second, third, and fourth interval levels are denoted by I_1, I_2, I_3, and I_4, respectively. The no-helmet, US-style firefighter helmet, and European-style helmet conditions are indicated by NH, US, and EU, respectively, and their significance level at any of the four intervals with 1, 2, 3, and 4. For example, EU2 refers to the effect of the European-style helmet in the second interval.

Borg's perceived discomfort

Helmet use led to significantly higher subjective effort than the NH condition (p-value <0.01). During flexion, discomfort was respectively 25.81% and 38.71% higher for EU (3.9 ± 1.8) and US (4.3 ± 1.9) helmets than NH (3.1 ± 1.6) conditions. During extension, discomfort increased by 16.67% and 33.33% for EU (4.2 ± 2.0) and US (4.8 ± 2.0) helmets, respectively, compared to NH (3.6 ± 1.8) conditions.

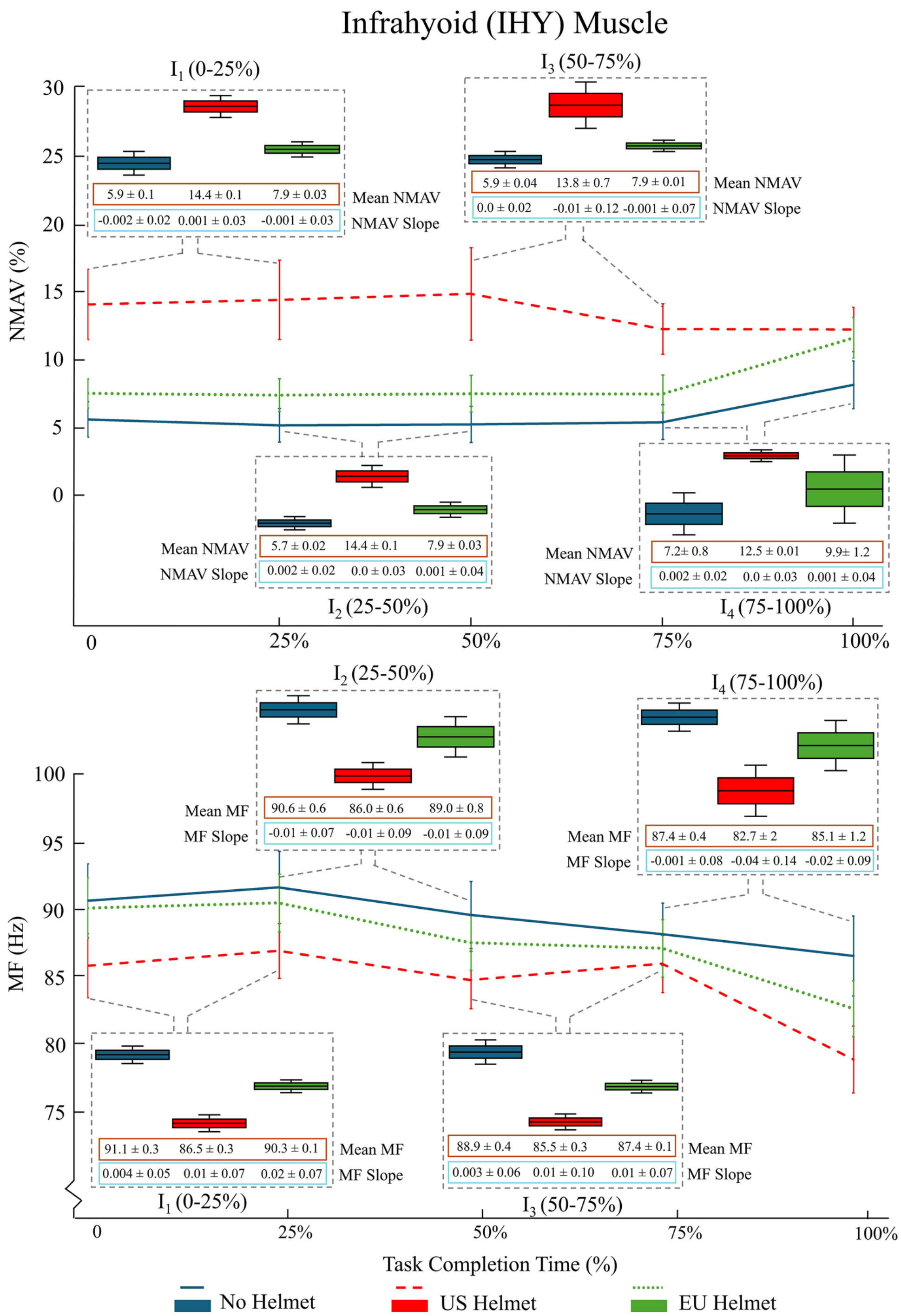

Figure 7. Piecewise linear regression plots of normalised mean absolute value (NMAV) and median frequency (MF) of infrahyoid (IHY) muscles during sustained-till-exhaustion extension tasks under no helmet, US-style firefighter (US) helmet, and European-style (EU) helmet conditions. Boxplots represent descriptive data (average and standard deviation) of NMAV and MF averages and corresponding slope values for the individual time intervals. The abbreviations I_1, I_2, I_3, and I_4 represent the sequential time intervals used in the analysis, corresponding to each quartile of the task duration.

Endurance time (ET)

ET data showed no significance for helmet uses in flexion (p-value = 0.61) or extension (p-value = 0.10). Participants respectively endured about 22.28% and 15.10% longer with NH conditions (182.54 ± 138.02s) than US (149.28 ± 110.67s) and EU (158.58 ± 109.02s) helmets during flexion, while about 58.51% and 33.34% longer with NH conditions (144.93 ± 109.11s) than US (91.43 ± 70.45s) and EU (108.69 ± 81.37s) helmets. EU helmets produced 6.23% and 18.88% longer ET than US helmets for

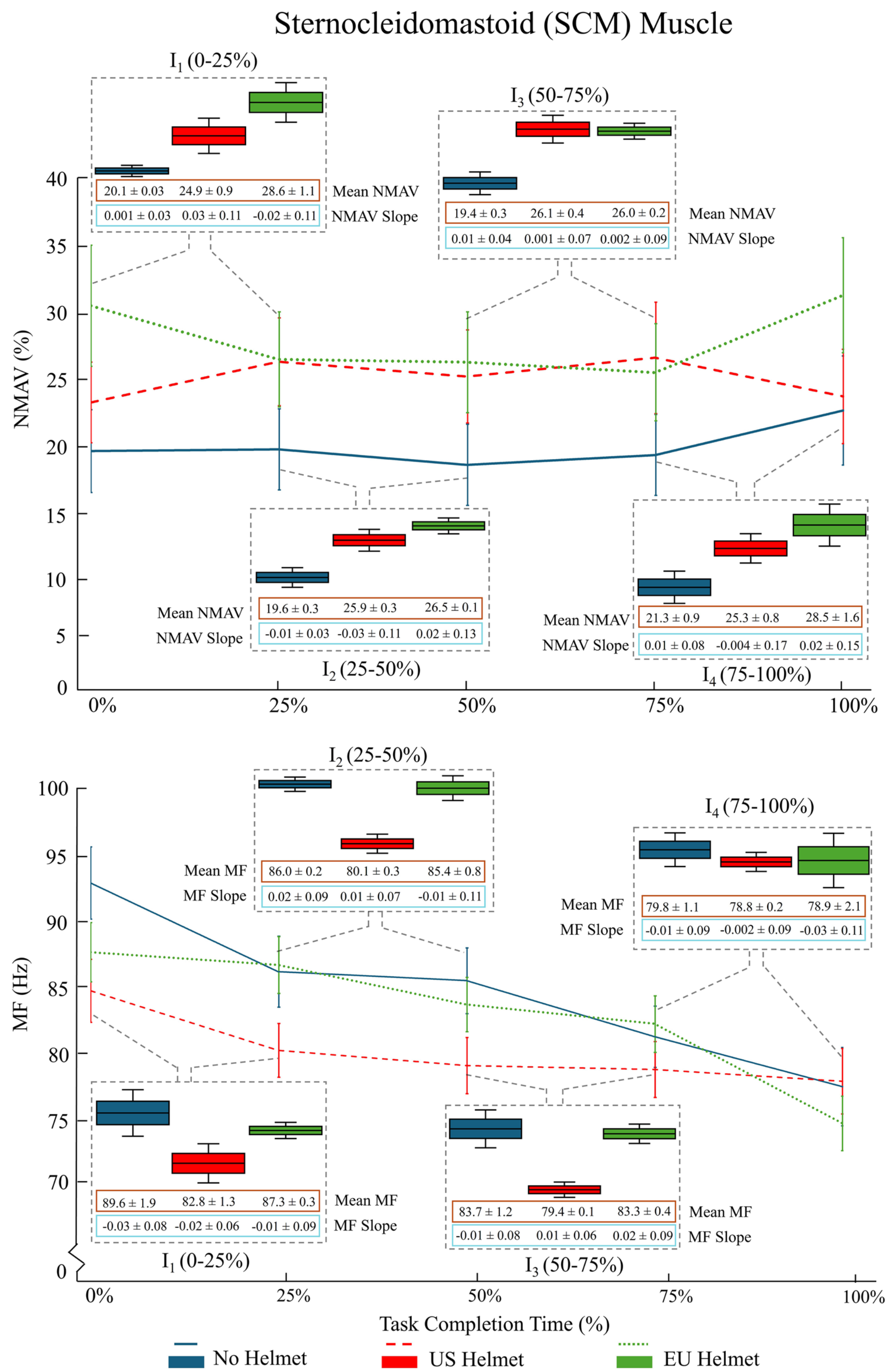

Figure 8. Piecewise linear regression plots of normalised mean absolute value (NMAV) and median frequency (MF) of sternocleidomastoid (SCM) muscles during sustained-till-exhaustion extension tasks under no helmet, US-style firefighter (US) helmet, and European-style (EU) helmet conditions. Boxplots represent descriptive data (average and standard deviation) of NMAV and MF averages and corresponding slope values for the individual time intervals. The abbreviations I_1, I_2, I_3, and I_4 represent the sequential time intervals used in the analysis, corresponding to each quartile of the task duration.

flexion and extension tasks, respectively. Across all helmet conditions, flexion trials showed longer ET than extension trials (NH: 25.95%, US: 63.27%, EU: 45.90%).

Discussion

This study examined how the sustained-till-exhaustion use of two firefighter helmets with diverse inertial properties—1) a US helmet with far superiorly-shifted COM location relative to the C0-C1 joint (i.e., high-profile design) and 2) an EU helmet with COM location closer to the C0-C1 joint (i.e., low-profile design)—affect neck muscle fatigue, endurance, and discomfort during maximum neck flexion and extension positions. The results demonstrated that a lower-profile helmet (i.e., EU helmet), which has a centre of mass (COM) located closer to the C0-C1 joint and thus a lower moment of inertia (MOI), resulted in lower neck muscle activity, less fatigue, improved perceived discomfort, and higher endurance time than a lighter-weight helmet (US helmet).

The fatigue progression, characterised by an increase in NMAV and a decrease in MF, was evident across all neck muscles. Notably, the US helmet induced faster fatigue despite being lighter (12.4%; 250 g) than the EU helmet. This may be attributed to its COM location—15.2 cm superiorly away from the C0–C1 joint—which is approximately 38.2% more superior than that of the EU helmet (9.4 cm). Consequently, the US helmet imposed about 17.4% greater MOI on the cervical spine, thereby increasing muscular demand. Our findings align with previous research demonstrating that high-profile helmets impose greater biomechanical loads on the cervical spine compared to low-profile helmets (Harms-Ringdahl et al., 1986; Paulon et al. 2024). Similarly, Phillips and Petrofsky (1983) reported a greater increase in NMAV and a decrease in MF during prolonged use of Air Force helmets compared to no helmet use, further supporting our results. Despite these findings, we observed a seemingly contradictory result—lower muscle activation during helmet conditions compared to the no-helmet condition. This outcome is likely due to the fact that the helmet further flexed the head-neck system, forcing participants to achieve extreme neck flexion angles. Because participants were required to maintain maximum neck flexion and extension postures to induce muscle fatigue, passive resistance structures of the cervical spine (e.g., ligaments and joint capsules) became engaged to relieve the load, thereby shifting part of the support from active muscle contraction to passive elements. Harms-Ringdahl et al. (1986) similarly found that EMG activities of both flexor and extensor muscles were reduced during extreme neck positions, suggesting a shift towards passive support. Nevertheless, maintaining these positions for an extended period required more than minimal activation from both agonist and antagonist muscle groups, indicating that active muscular engagement remained essential despite the contribution of passive elements.

Muscle fatigue was more pronounced in the first and second intervals, reflecting an initial phase of increased muscle involvement (and thus muscle fatigue) before the load shifted to passive structures such as ligaments, cervical spine, and intervertebral discs (Panjabi 1992; Colloca and Hinrichs 2005). As muscles become fatigued, neural drive to muscles decreases, and the passive tissues assume a greater load (McGill and Kippers 1994; Choi et al. 2020). Consequently, muscle fatigue was not consistent in the third interval. However, in the fourth interval, muscle involvement partially resumed, likely due to afferent feedback from muscles and cervical joints to the brain to adopt compensatory strategies to maintain spinal stability and mitigate the risk of impairment to the spine (Norasi et al. 2023). These findings reinforce the nonlinear time-specific progression of muscle fatigue (Chowdhury and Nimbarte 2015), and the appropriateness of piece-wise regression analysis to understand time-specific progression of muscle fatigue and passive-active tissue load-sharing dynamics. Additionally, as fatigue progressed, our results showed a slight graduate increase in neck angle over time, indicating increased viscoelastic deformation of passive components due to a decrease in agonist muscle activity and their fatiguing contractions (Chen, Chan, and Alexander 2024). Interestingly, the results demonstrated that helmet use significantly reduced the total neck extension angle compared to the no-helmet condition. This reduction can likely be attributed to the physical constraints imposed by helmet structures—particularly those with bulkier designs, such as firefighting helmets. During extension, the posterior portion of the helmet can come into contact with the upper back, cervical region, or shoulder, thereby physically impeding further neck range of cervical extension. In contrast, during sub-maximal neck flexion, there is no anatomical or structural obstruction limiting the motion. The additional mass and elevated centre of mass introduced by the helmet increased the flexion torque, imposing a greater load that drives the head downward. As a result, neck flexion angles were found to be greater under helmet conditions.

Interestingly, the UTR muscles exhibited a distinct muscle activation pattern. Unlike other muscles, it showed lower NMAV values under both helmets compared to the NH condition. This may result from a myoelectric inhibition mechanism during full neck flexion (Troiano et al. 2008; Nimbarte, Zreiqat, and Chowdhury 2014). The increased moment at the C0–C1 joint under helmet use likely compelled further neck flexion, intensifying the UTR inhibition further. This is consistent with a previous study that reported higher neck muscle activation under NH condition than the helmet condition during neck flexion tasks (Callaghan, Laing, and Dickerson 2014). In contrast, the CTR muscle demonstrated higher NMAV values with helmet use, possibly due to additional loading on the nuchal ligament, which both counterbalances moments on the neck joint (Wang et al. 2014) and limits the range of motion during flexion tasks (Takeshita et al. 2004). Additionally, when the nuchal ligament overstretches, it transfers more load to the CTR, thereby increasing CTR muscle activity. The IHY and SCM muscles, being involved in neck extension, displayed increased NMAV and greater fatigue under helmet use. Not to mention, neck angles during extension were smaller than during its flexion under all helmet conditions. This can be attributed to the fact that neck flexors (IHY and SCM) are smaller and weaker than neck extensors (Uhlig et al. 1995; Chowdhury et al. 2022). Additionally, they aremore fatigue-prone (Miller et al. 1993; Billeter and Hoppeler 2003). To mitigate injury risks during extension, posterior passive tissues trigger a compensatory mechanism by increasing their stiffness, which can result in further reduction in neck extension angle (McGill, Seguin, and Bennett 1994; Niewiadomski et al. 2019). ET and discomfort data further supported that sustained extension tasks under helmet use were more physically demanding than sustained flexion tasks.

This study has several limitations. The assessment of neck muscle fatigue under helmet use was limited to neck flexion and extension tasks. Since unintentional compensatory movements involving neck lateral bending or rotation may occur during daily operational tasks (Niewiadomski et al. 2019), future research should explore helmet effects under these movements. Additionally, deeper neck muscles were not analysed due to the use of surface EMG. Age and sex, which influence neck muscle endurance and fatigue (Hunter 2014), were not considered. Future research incorporating these variables may offer broader insights into the underlying physiological mechanisms driving fatigue progression across diverse populations. Lastly, despite acquiring EMG data from both flexor and extensor muscles, we did not include antagonist muscle activity and coactivation between agonist-antagonist muscle groups in this study. We planned to report them in future studies, focusing on the head-neck flexion-extension phenomenon and visco-elastic dynamics of neck passive tissues during sustained helmet use. In summary, though minimising helmet weight is essential to reduce the risk of developing neck injuries, our findings suggest that a low-profile helmet design, characterised by a COM close to C0-C1 joint, should be equally prioritised, especially when a helmet is used for prolonged duration in awkward postures.

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Author contributions

Leonardo H. Wei: Conceptualisation, Methodology, Data Preparation, Writing. **Gustavo M. Paulon:** Literature Review, Writing, Data Analysis, **Pramiti Sarker, Ph.D.:** Reviewing, Writing, and Editing. **Suman K. Chowdhury, Ph.D.:** Conceptualisation, Supervision, Funding Acquisition, Investigation, Reviewing, and Editing.

Disclosure statement

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