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The role of neck muscle co-contraction and postural changes in head kinematics after safe head impacts: Investigation of head/neck injury reduction

Mohammad Homayounpour[a], Nicholas G. Gomez[a], Anita N. Vasavada[b,c,d], Andrew S. Merryweather[a,*]

[a] Department of Mechanical Engineering, University of Utah, Salt Lake City, UT, USA
[b] Gene and Linda Voiland School of Chemical Engineering and Bioengineering, Washington State University, Pullman, WA 99164-6515, USA
[c] Department of Integrative Physiology and Neuroscience, Washington State University, Pullman, WA 99164, USA
[d] Washington Center for Muscle Biology, Washington State University, Pullman, WA 99164, USA

A R T I C L E I N F O

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A B S T R A C T

Concerns surrounding concussions from impacts to the head necessitate research to generate new knowledge about ways to prevent them and reduce risk. In this paper, we report the relative temporal characteristics of the head resulting from neck muscle co-contraction and postural changes following a sudden force applied to the head in four different directions. In the two "prepared" conditions (i.e., co-contraction and postural), participants experienced impulsive forces to the head after hearing a warning. The warning given for the postural condition informed both the direction and timing of the impulsive force. Participants responded to the postural warning by altering their head posture, whereas in the co-contraction warning, the force direction was unknown to them, and they were asked to isometrically co-contract their neck muscles after the warning. Peak angular velocity reduced by 29% in sagittal extension, 18% in sagittal flexion, and 23% in coronal lateral flexion in prepared vs. unwarned conditions. Peak linear acceleration was attenuated by 15% in sagittal extension, 8% in sagittal flexion, and 18% in coronal lateral flexion in prepared vs. unwarned conditions. Changes in peak angular acceleration were not uniform. We also measured a significant delay in the peak angular velocity (22 vs. 44.8 ms) and peak angular acceleration (7 vs. 20 ms) after peak linear acceleration in prepared compared to unwarned conditions. An increase in muscle activation significantly reduced the peak angular velocity and linear acceleration. Gross head movement was significantly decreased with preparation. These findings suggest that a warning prior to impact can reduce head kinematics associated with injury.

1. Introduction

There are approximately 4 million concussions reported in the United States each year that result from a sport-related injury (Leddy et al., 2019). Concussions occur primarily in contact sports such as American football, soccer, hockey, and boxing. Solutions proposed include rule changes, coaching, and neck strengthening (Leddy et al., 2019). Although there is still disagreement on the role of neck muscle strength in concussion (Caswell et al., 2014; Eckner et al., 2018; Eckner et al., 2014; Mansell et al., 2005; Mihalik et al., 2011), there is a consensus that active neck muscles at the time of impact seem to reduce the risk of injury with concussive and subconcussive impacts (Eckner et al., 2014; Fanta et al., 2013; Mortensen et al., 2018; Reynier et al., 2020; Tierney et al., 2005).

Awareness and proper posture have been suggested as factors that affect the head and neck kinematics and risk of injury (Alsalaheen et al., 2018; Eckner et al., 2014; Kuo et al., 2019; Kuramochi et al., 2004; Le Flao et al., 2018; Lincoln et al., 2013; Siegmund et al., 2003a,b). Beusenberg et al. (2001) elaborated on the importance of head and neck coupling in head impact simulation. Keshner (2000) suggested that co-contraction may be the only available neural option when the reciprocal mechanism fails. Lincoln et al. (2013) reported 67% of concussions occurred when the player was unaware or had an awkward posture at the time of impact. Viano et al. (2007) showed that stiffer

* Corresponding author.
E-mail address: a.merryweather@utah.edu (A.S. Merryweather).

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necks reduce linear head displacement, velocity, and acceleration in professional American football. Mortensen et al. (2020) investigated the effect of posture in a simulation study and found leaning the head toward an impact significantly reduced concussion risk metrics compared to no change in posture. Fanton et al. (2019) concluded that leaning toward the direction of impact changes the position of force application and consequently decreases head angular acceleration. Although the importance of posture has been investigated in different simulation studies, few experimental studies investigated the effect of posture changes on head kinematics after a direct head impact. In all the studies where anticipation or muscle co-contraction was tested, the participant knew about the direction of incoming impact beforehand, which may add prior consciousness that affects the kinematics.

Muscle activation and posture can be altered intentionally and unintentionally using a warning system. Efforts have been made in sports (Aston et al., 2020; Homayounpour et al., 2019; Luttmer et al., 2020; Mehrpour Bernety & Schurig, 2020), and motor-vehicle (Siegmund et al., 2003a, 2003b) settings to provide participants with information regarding the impending contact. Happee et al. (2017), in a simulation study, reported that co-contraction has a minor role in dynamic head stabilization compared to vestibulocollic and cervicocollic reflex. Increasing our knowledge about the role of posture and cervical muscle contraction due to such a warning enhances our ability to develop an efficient warning system to reduce the risk of head and neck injury.

In this study, we sought to investigate the extent to which postural changes and cervical muscle activation as a function of different types of audible warnings reduce head kinematics following an applied force to the head. We hypothesized that pre-impact preparation with moving against the direction of pull (postural changes) and isometric cervical muscle contraction (co-contraction) will reduce the head kinematics after the applied force. Head kinematics used to test this hypothesis were peak angular velocity, peak angular acceleration, and peak linear acceleration. We further investigated the importance of the level of postural changes and muscle activation on head kinematics. The results of these preparations on the timing of peak kinematics and head displacement and rotation for each force direction are discussed.

2. Method:

2.1. Participants

Data used in this study are based on the protocol and participants that are described in detail in (Homayounpour et al., 2021). Briefly, ten male participants (age 26.2 ± 3.1 years, height 179.8 ± 5.3 cm, and weight 73.6 ± 7.6 kg) were recruited and gave voluntary consent according to the University of Utah Internal Review Board (IRB: 94138) protocol. Participants were included if they had no history of concussions or neck injuries.

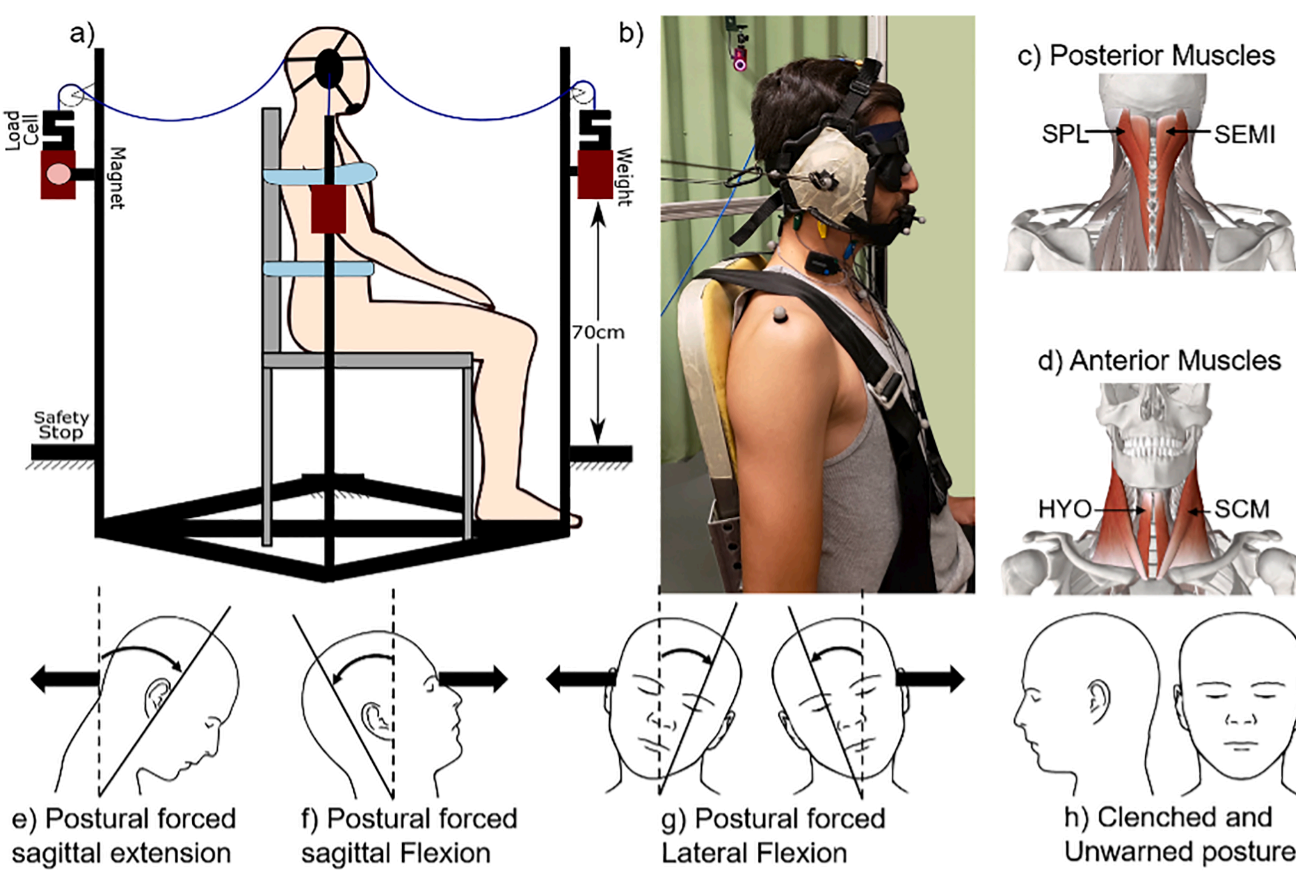

Fig. 1. (a) Schematic of the testbed. The testbed includes four weights, allowing for impulsive loads to be applied in four different directions without modifying the experimental setup. Impulsive forces were applied with a 1.2 kg weight, attached to a head gear using Kevlar cable. The Kevlar cable has slack to allow the mass to free fall on the linear guide for 60 cm and then pull the head. A safety stop was placed 10 cm after the end of the string to make sure the neck would not be overextended. Participants were strapped to the chair to minimize trunk movement, with their hands placed on their laps. (b) Instrumented participant: headgear was sized and sewn for each subject before the test to reduce slipping. The noise isolation with the earbuds and the headband's earpieces and also the blindfold removed environmental acoustic and visual cues for participants in test trials. (c) EMG sensors were placed bilaterally over the posterior muscles, semispinalis capitis (SEMI) and splenius capitis (SPL), and (d) anterior muscles, sternocleidomastoid (SCM) and hyoid (HYO), muscles. Posture in different test conditions: (e) forced sagittal extension, (f) forced sagittal flexion and g) forced lateral flexion to the left and to the right with postural changes and, (h) participant remained in the neutral posture for the co-contracted and unwarned conditions and were not informed about the direction of force. The curved arrows show the positive direction in $Pos_{(Imp)}$ for each force direction, straight arrows.

2.2. Instrumentation

EMG sensors (2222 Hz and 1926 Hz, Delsys Trigno wireless EMG, MA, USA) were placed bilaterally as depicted in Fig. 1. An optical motion capture system (100fps, Flex 3, NaturalPoint, Corvallis, OR) captured marker position. Four masses, 1200 g, could free-fall and apply an impulsive load to the head in four directions. Due to cable compliance, force peaked 21 ± 4 ms after force onset. The pulleys' heights were adjusted to the level of each participant's ear to deliver a horizontal force to the head for each participant. Uniaxial load cells (50 lb, S-Type, PCB, NY) were used in series with the cable and the corresponding mass to measure the applied force (2000 Hz). Participants wore earbuds (ER3XR, ETYMOTIC, IL) to deliver the acoustic warnings.

Kinematic measurements, linear acceleration, angular velocity, and angular acceleration were accomplished using an instrumented mouthguard (Vector, Athlete Intelligence, WA, USA). The mouthguard was fit through the boil-and-bite process and attached to the upper jaw. The mouthguard internally derives the rate gyroscope data to calculate angular acceleration. Accelerometer and gyroscope data were internally low-pass filtered at 400 and 100 Hz cutoff frequencies, respectively (Camarillo et al., 2013). Sensor data were resampled to 5000 Hz. The mouthguard was triggered to start data collection at 1 g linear acceleration and stored data for 13 ms before the 1 g threshold and 80 ms after the threshold.

2.3. Test conditions

Three different types of conditions were tested: postural changes, co-contraction, and unwarned (control). The participant was called "prepared" if they either co-contracted or changed their posture before the applied force based on the given acoustic warning. The acoustic warning played for 500 ms and started 1000 ms before the applied force. In the postural change condition, four unique directional sounds were associated with four pull directions: forced sagittal extension, forced sagittal flexion, forced lateral flexion to the right and left. The participant was asked to move against the pull direction as soon as each sound was heard, Fig. 1. In co-contraction trials, a unique non-directional sound played that was not associated with any pull direction, and the pull direction was unknown to the participant. The participants were instructed to isometrically contract (co-contract) cervical muscles after this sound. The unwarned (control) condition was used as the baseline, and no warning was given prior to the applied force. For postural and co-contraction trials, appropriate movement responses were verified following data collection by evaluating the motion capture and EMG data. Trials were excluded from analyses if the participant moved in the wrong direction or did not move/co-contract before the applied force.

2.4. Protocol

After applying the instrumentation to the participant, a maximum voluntary contraction (MVC) test was performed in four principal directions. Following the MVC trials, non-directional and directional warnings were played for the participants at least three times before delivering any forces to help with acclimation to the warning sounds. The participants received 50 impulsive head forces in four different directions, following a warning, non-directional (co-contraction), directional (postural), startling, or without a warning (unwarned) in random order. Five startle trials for each participant are not included in this manuscript due to the small sample size. There were random 20 to 45 s delays between each applied force.

2.5. Data analysis and normalization

All EMG data were resampled to 1926 Hz for analysis and high-pass filtered at 30 Hz to remove motion artifacts before calculating the 50 ms moving window root-mean-squared (RMS). The EMG activation was then normalized based on this MVC value. The eight measured muscles were combined to make four groups: anterior (bilaterally HYO and SCM), posterior (bilaterally SPL and SEMI), right (R-HYO, R-SCM, R-SEMI, R-SPL), and left (L-HYO, L-SCM, L-SEMI, L-SPL). The EMG activation level before force application was defined as the $EMG_{(Pre\text{-}Imp)}$ for a muscle. The muscle group opposing the direction of pull was averaged together to find the mean $EMG_{(Pre\text{-}Imp)}$ for each trial (i.e. anterior muscles for sagittal extension).

The HYOID OpenSim model (Mortensen et al., 2019, 2018) was used to calculate the head posture at the time of impact. Models were scaled for each participant. Head orientation and position were extracted from the OpenSim model based on the motion capture data. Posture was reported as the pre-impact head orientation in degrees, $Pos_{(Pre\text{-}Imp)}$, in the principal plane of movement with respect to the lab coordinate frame prior to force onset while initial posture represents zero angle (Fig. 1). The positive direction was defined as the opposite direction of pull (i.e., sagittal extension movement for sagittal flexion pull, Fig. 1). In addition to reporting the pre-impact head posture (Table 2), we also reported the maximum movement of the head from its maximum pre-impact posture to its maximum post impact posture (Table 4). Since the delay between the sound initiation and the impact was 1 s, the participants had enough time to move and remain stationary at the time of impact. For trials with postural changes, we confirmed the head stopped moving away from the direction of pull using motion capture data.

Kinematics data were filtered inside the mouthguard as explained by Camarillo et al. (2013) prior to exporting for analysis. Only angular acceleration was low-pass filtered (12 Hz, zero-lag 4th order Butterworth) after the export. The resultant linear acceleration, filtered angular acceleration, and angular velocity were analyzed. Data from left & right exposures were averaged after sign correction. Due to the participant's posture changes during the test and the warnings, the free-fall height was not constant for all the trials. Since the applied force is a function of the square-root of free-fall height, and as mean free-fall height measured was $h = 63$cm, we normalized all the kinematic results by multiplying them by $\sqrt{\frac{.63}{\Delta h_{mes}}}$.

2.6. Statistics

Linear mixed models were fit to describe each kinematic outcome, including peak angular velocity (PAV), peak angular acceleration (PAA), peak linear acceleration (PLA), retraction, and time of the PAV and PAA compared to PLA (all continuous) across all head motion directions. Tables 1–4 reported estimated means and standard errors (SE) from the linear mixed model. Preparation types (co-contraction, postural, and unwarned) were defined as the fixed effect for all the models except one, in which $EMG_{(Pre\text{-}Imp)}$ and $Pos_{(Pre\text{-}Imp)}$ (both continuous variables) were used as the main effects. All models designated participants as a random effect. Model assumptions were validated by examining the normality of residuals (Fino, 2016). Pairwise comparisons were performed on the model to check for the significance between the postural and co-contraction preparations. Multiple comparisons were controlled using the Benjamini–Hochberg method (Benjamini &and Hochberg, 1995), resulting in an adjusted *P-value*<0.032 as statistically significant in the linear mixed model in Matlab 2020a (MathWorks, Natick, MA, USA).

3. Results

Angular velocity and angular acceleration in the unwarned condition show a noticeable lag compared to the prepared conditions (postural change and co-contraction) in all force directions. The head's kinematics in the principal movement axis is depicted in Fig. 2 based on preparation type. Coronal lateral flexion values were combined due to no significant difference found between left and right for all participants. Peak angular velocity (PAV) with both muscle co-contraction (4.19, 3.63, and 3.25 rad/s) and postural changes (4.06, 4.23, and 4.29 rad/s) were

Table 1
Peak angular velocity (PAV), peak angular acceleration (PAA), and peak linear acceleration (PLA) changes based on the preparation type, co-contraction and postural compared to Unwarned trials in three different force directions, sagittal extension (Sag. Ext.), sagittal flexion (Sag. Flex.), and coronal lateral flexion (Lat. Flex.). P-value are reported in co-contraction and postural trials compared to Unwarned (Control). Significant values (p-value < 0.032) are bolded.

Side	Type	PAV (rad/s)		PAA (rad/s^2)		PLA (m/s^2)		Force (N)	
		Mean(SE)	*p-value*	Mean(SE)	*p-value*	Mean(SE)	*p-value*	Mean(SE)	*p-value*
Sag. Ext.	Unwarned	5.78(0.25)		226.9(12.7)		45.37(1.87)		187.2(4.7)	
	Co-contract	**4.19(0.17)**	<0.0001	231.4(10.4)	0.6673	**38.32(1.87)**	0.0003	188.0(5.5)	0.8830
	Postural	**4.06(0.13)**	<0.0001	211.5(8.2)	0.0647	**38.56(1.47)**	<0.0001	189.6(4.4)	0.5924
Sag. Flex.	Unwarned	4.79(0.19)		183.8(9.6)		43.01(2.13)		188.5(6.4)	
	Co-contract	**3.63(0.31)**	0.0004	**213.1(12.4)**	0.0203	**36.93(2.65)**	0.0243	193.0(7.1)	0.5307
	Postural	**4.23(0.24)**	0.0252	**213.1(9.6)**	0.0029	**36.85(2.05)**	0.0035	**162.6(5.5)**	<0.0001
Lat. Flex.	Unwarned	4.88(0.16)		323.7(9.2)		42.63(1.92)		179.1(3.9)	
	Co-contract	**3.25(0.15)**	<0.0001	**239.1(14.2)**	<0.0001	**34.89(1.98)**	<0.0001	185.8(4.8)	0.1638
	Postural	**4.29(0.11)**	<0.0001	**262.3(10.5)**	<0.0001	43.53(1.47)	0.5389	**190.6(3.6)**	0.0014

Table 2
Mean and standard error (SE) of neck muscle activation level, $EMG_{(Pre\text{-}Imp)}$ and head posture, $Pos_{(Pre\text{-}Imp)}$, before the applied force in different force direction and type of preparation. The positive angle in $Pos_{(Pre\text{-}Imp)}$ is the head orientation in the opposite direction of pull.

Side	Type	$EMG_{(Pre\text{-}Imp)}$ (%MVC) Mean(SE)	$Pos_{(Pre\text{-}Imp)}$ (deg) Mean(SE)
Sag Ext	Control	0.9(2.2)	1.9(2.9)
	Co-contraction	27.2(2.8)	5.0(3.3)
	Postural	18.1(2.2)	22.9(2.7)
Sag Flex	Control	1.6(2.9)	−2.2(3.1)
	Co-contraction	35.6(3.2)	−7.7(3.8)
	Postural	19.0(2.4)	17.5(3.0)
Lat Flex	Control	1.7(3.0)	0.6(2.3)
	Co-contraction	30.5(3.0)	0.8(1.5)
	Postural	19.0(2.2)	25.5(1.1)

significantly lower compared to unwarned conditions (5.78, 4.79, 4.88 rad/s) in forced sagittal extension, sagittal flexion, and lateral flexion, respectively (Table 1). PAV in co-contraction trials was also significantly lower than in postural changes trials in the lateral force direction ($p < 0.0001$). Peak linear acceleration (PLA) with both muscle co-contraction (38.3 and 36.9 m/s^2) and postural changes, (38.6 and 36.9 m/s^2) were significantly attenuated compared to the unwarned trials (45.4, 43.0 m/s^2) in sagittal extension and sagittal flexion, respectively. PLA significantly decreased in the co-contraction condition (34.9 m/s^2) compared to the unwarned (42.6 m/s^2) and postural changes trials (43.5 m/s^2) in the lateral force direction. Peak angular acceleration (PAA) with co-contracted muscles (213.1 rad/s^2) and postural changes (213.1 rad/s^2) were significantly higher compared to the unwarned condition (183.8 rad/s^2) in sagittal flexion; however, PAA with co-contracted muscles (239.1 rad/s^2) and postural changes (262.3 rad/s^2) were significantly lower compared to the unwarned condition (323.7 rad/s^2) in the lateral direction. There was no significant difference between the warned conditions or in any condition in sagittal extension in PAA. For the trials where participants changed their posture prior to the impact, the mean cervical flexion (22.9 degrees), sagittal extension (17.5 degrees), and lateral flexion (25.5 degrees) were significantly different based on the type of preparation (Table 2). We investigated the impact of both posture and muscle activation on PAV, PAA, and PLA (Table 3). This table describes how much each kinematic variable changes per degree of posture change and per %MVC of muscle activation. For example, in sagittal extension, the PAV decreases by 0.04 rad/s for each % increase in EMG activity, and the PAV decreases by 0.03 for each increase in posture angle from neutral.

4. Discussion

We hypothesized that preparation with postural changes and isometric neck muscle co-contraction would reduce the kinematic response after the impulsive force to the head compared to the unwarned (control) condition. PAV, PAA, and PLA are reported as the main contributing factors in head and neck injury metrics (Alsalaheen et al., 2019; Camarillo et al., 2013; Eckner et al., 2014; Le Flao et al., 2018; Reynier et al., 2020; Siegmund et al., 2016). This hypothesis was mostly supported for PAV and PLA though not all force directions reached significance. PAV with any preparation shows a 29% reduction in sagittal extension, 18% reduction in sagittal flexion, and a 23% reduction in the lateral force direction compared to unwarned trials. This finding is consistent with what has been reported in the literature (Eckner et al., 2014; Kuo et al., 2018; Reynier et al., 2020; Simoneau et al., 2008). To provide more resolution apart from just the presence of co-contraction or postural change, we investigated the level of changes (i.e. $EMG_{(Pre\text{-}Imp)}$ and $Pos_{(Pre\text{-}Imp)}$) and reported them in Table 2. This approach was taken throughout each test for kinematic responses regardless of the

Table 3
Changes in PAV, PAA, and PLA with respect to each % increase in MVC, $EMG_{(Pre\_Imp)}$, and respect to each change of degree of posture, $Pos_{(Pre\_Imp)}$, with respect to the control condition. The control rows show the mean kinematic value with zero neck muscle activation, $EMG_{(Pre\_Imp)}$, and zero degrees head posture, $Pos_{(Pre\_Imp)}$, (initial posture). $EMG_{(Pre\_Imp)}$ and $Pos_{(Pre\_Imp)}$ rows show the coefficient of changes for every %MVC and degrees of change respectively, compared to the associated control row. As an example, beta in sagittal extension for EMG, −0.04, shows that for every %MVC increase in muscle activation, mean PAV, 5.62 rad/s, will decrease −0.04. Significant values (p-value < 0.032) are bolded.

Side	Variable	PAV (rad/s)		PAA (rad/s^2)		PLA (m/s^2)	
		beta(SE)	*p-value*	beta(SE)	*p-value*	beta(SE)	*p-value*
Sag Ext	Control	5.62(0.28)		234.7(12.5)		43.96(1.89)	
	$EMG_{(Pre\text{-}Imp)}$ (%MVC)	**−0.04(0.00)**	<0.0001	0.2(0.3)	0.5135	**−0.20(0.05)**	0.0002
	$Pos_{(Pre\text{-}Imp)}$ (Deg)	**−0.03(0.00)**	<0.0001	**−1.0(0.3)**	0.0001	−0.39(0.39)	0.3163
Sag Flex	Control	4.61(0.20)		198.1(10.9)		40.34(1.91)	
	$EMG_{(Pre\text{-}Imp)}$ (%MVC)	**−0.02(0.01)**	0.0106	0.5(0.3)	0.0934	−0.08(0.08)	0.2736
	$Pos_{(Pre\text{-}Imp)}$ (Deg)	0.02(0.01)	0.0426	0.5(0.3)	0.1584	−0.20(0.44)	0.6438
Lat Flex	Control	4.76(0.18)		290.9(11.4)		42.08(2.05)	
	$EMG_{(Pre\text{-}Imp)}$ (%MVC)	**−0.02(0.00)**	<0.0001	**−0.8(0.3)**	0.0112	**−0.09(0.04)**	0.0320
	$Pos_{(Pre\text{-}Imp)}$ (Deg)	0.00(0.01)	0.9935	0.8(0.8)	0.3530	**0.12(0.05)**	0.0149

Table 4
Time of peak angular velocity (PAV) and peak angular acceleration (PAA) compared to the peak linear acceleration (PLA) in response to different force directions, forced sagittal extension (Sag. Ext.), sagittal flexion (Sag. Flex.) and lateral flexion (Lat. Flex) with different types of preparation and in different impact direction. Peak skull displacements, angle and displacement, were reported with different warnings. The differences between the Directional and Non-Directional warnings were not significant. Significant values (p-value < 0.032) are bolded.

Side	Type	Delay after PLA (ms)				Peak Displacement			
		PAV Time		PAA Time		Translation (mm)		Angle (deg)	
		Mean(SE)	*p-value*	Mean(SE)	*p-value*	Mean(SE)	*p-value*	Mean(SE)	*p-value*
Sag Ext	Unwarned	51.5(2.5)		16.7(1.7)		71.2(3.4)	<0.0001	19.8(1.3)	
	Co-contraction	**24.7(2.1)**	<0.0001	**6.7(1.8)**	<0.0001	**38.7(3.8)**	<0.0001	**8.9(1.3)**	<0.0001
	Postural	**21.4(1.7)**	<0.0001	**4.0(1.5)**	<0.0001	**36.5(3.2)**	<0.0001	**9.3(1.1)**	<0.0001
Sag Flex	Unwarned	45.1(2.0)		22.9(3.3)		69.8(4.2)	<0.0001	19.2(1.5)	
	Co-contraction	**22.2(4.8)**	0.0005	7.4(7.9)	0.0532	**50.3(4.9)**	0.0002	**11.3(1.7)**	<0.0001
	Postural	**22.3(2.7)**	<0.0001	**3.6(4.4)**	<0.0001	**48.8(3.8)**	<0.0001	**12.9(1.3)**	<0.0001
Lat Flex	Unwarned	37.8(1.4)		20.0(0.6)		68.1(3.4)	<0.0001	11.1(0.8)	
	Co-contraction	**24.8(2.2)**	<0.0001	**12.0(1.0)**	<0.0001	**47.9(2.9)**	<0.0001	**5.9(0.8)**	<0.0001
	Postural	**20.7(1.7)**	<0.0001	**6.4(0.7)**	<0.0001	**44.4(2.1)**	<0.0001	**8.7(0.6)**	<0.0001

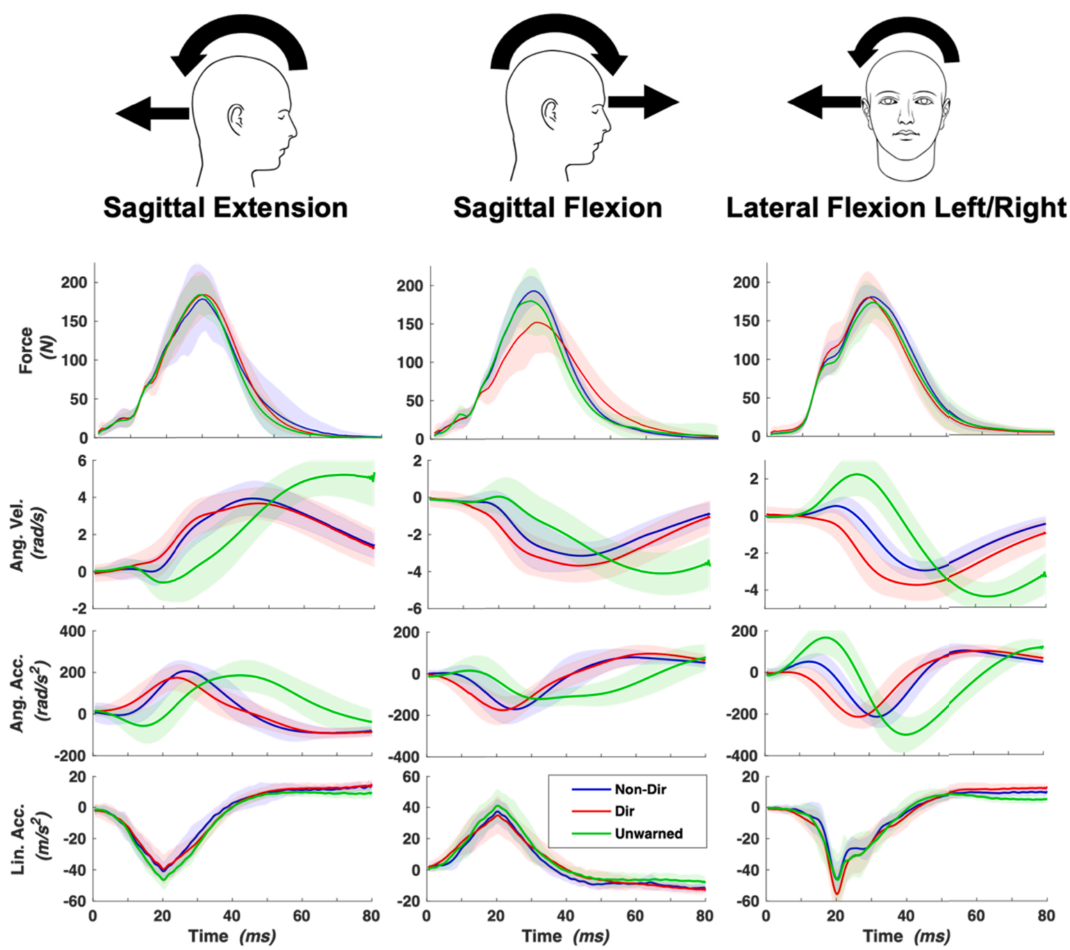

Fig. 2. Mean and standard error (shades) of applied force to the head, head angular velocity, angular acceleration, and linear acceleration in sagittal extension, sagittal flexion, and coronal lateral flexion in the main axis of movement. Data were synchronized based on the peak linear acceleration.

type of preparation (see Table 3). We found that for PAV, there was a strong inverse correlation between $EMG_{(Pre\text{-}Imp)}$ and the PAV for all force directions, indicating that the intensity of the isometric contraction is as important as its presence. Simoneau et al. (2008) investigated forced muscle activation and found a greater reduction in angular velocity in sagittal extension when preloading on the neck muscles was increased. We had also expected the $Pos_{(Pre\text{-}Imp)}$ to have a similar relationship, but our results showed that PAV was only reduced significantly in forced sagittal extension by an increase in $Pos_{(Pre\text{-}Imp)}$. The interaction term, $EMG_{(Pre\text{-}Imp)}$*$Pos_{(Pre\text{-}Imp)}$, was significant in sagittal extension and the lateral flexion force direction for PAV, however the model coefficients were<0.001, and were therefore ommited. Muscle co-contraction and postural changes also had a significant effect on the PAV time (Table 4). In the unwarned condition, the PAV occurred 45 ms following PLA. That time was reduced to 24 ms with postural changes and 22 ms with the co-contraction.

The results for peak linear acceleration (PLA) were similar to PAV but with some minor differences. PLA decreased significantly by 15% and 8% in sagittal extension and flexion with any preparation. PLA decreased by 18% only in the co-contraction condition, and there were no differences between the postural change and the unwarned trial in the lateral flexion force direction (Table 1). Reynier et al. (2020) reported a PLA reduction in lateral impact when the participant unilaterally activated their neck muscle but not during uniform muscle co-contraction. This difference can be due to the difference in the level and applied forces and their method since they used a direct head impactor, and the mean PLA was 12gs vs. 4gs in this study. An inverse relationship between $EMG_{(Pre\text{-}Imp)}$ and PLA and PAV was found (Table 3). However, unlike the PAV relationship, the PLA actually increased in the coronal flexion directions with an increased head angle.

Unlike PAV and PLA, changes in peak angular acceleration (PAA) were not consistent between the directions. The PAA decreased, increased, and did not change in forced lateral flexion, sagittal flexion, and sagittal extension, respectively. We measured a 23% significant reduction when the participants were prepared compared to the control scenario during the lateral flexion force direction (Table 1). PAA increased by 16% in sagittal flexion force direction with preparation and did not show any significant change in the sagittal extension force direction. These findings partially support what Tierney et al. (2005) previously reported that PAA decreased when participants were prepared compared to scenarios where they were not prepared in the sagittal force direction. Reynier et al. (2020) could not find any significance in PAA with a higher level of lateral head impacts with preparation. The PAA reduction rate was only significant with $Pos_{(Pre\_Imp)}$ in sagittal extension and $EMG_{(Pre\_Imp)}$ in lateral direction; however, the coefficients compared to the mean were very small (Table 2). That is to say, PAA is not dependent on either change in posture or muscle activation, and it changed with anticipation of the applied force, which activates the muscles sooner (Alsalaheen et al., 2018). Despite the inconsistencies in amplitude reduction, the PAA showed a similar reduction in delay after PLA, as seen with the PAV, when either isometric muscle contraction or postural changes were adopted compared to the control condition (Table 4). In the control condition, the PAA occurred 20 ms following the PLA, and that time decreased to 9 and 5 ms when the participants were co-contracted or changed posture, respectively. One possible reason for the inconsistency in PAA changes with and without preparation in different directions can be explained by the movement of the head's instantaneous center of rotation (HICOR). Kuo et al. (2018) reported that HICOR significantly moved inferior in the transverse plane by co-contracting neck muscles in coronal lateral flexion.

HICOR did not change significantly in the forced sagittal extension compared to the unwarned condition (Kuo et al., 2018), and consequently, we did not measure any changes in PAA in sagittal extension. Since PAA increased in sagittal flexion, we may speculate that HICOR would have moved up, superior in the transverse plane, with muscle co-contraction in this direction.

PAV, PLA, and PAA are useful for characterizing the head's movement during an impact and therefore providing information that can help researchers better understand the risk of injury. Our study found that PLA, PAA, and peak retraction did not change significantly between the postural change and neck co-contraction conditions. However, in trials with postural change, we measured lower muscle activation and shorter length in the opposing muscles by moving against the pull direction compared to the co-contraction trials. Muscle injuries, specifically contraction-induced strains, reduce by increasing pre-activation, the magnitude of strain, and initial muscle length. (Lovering et al., 2005) We measured less pre-activation, similar head angular changes, and postures that imply shorter initial muscle lengths before impact with postural changes compared to the co-contraction condition. These factors, in combination, may reduce the risk of muscle injury from contraction-induced strains. Based on our results, we propose that providing an individual with a warning that results in a postural change toward the impending impact would have the greatest effect on potentially reducing the risk of injury. This warning may reduce the reaction time (Bella & Silvestri, 2017; Homayounpour et al., 2021; Zhang et al., 2015) and help players alter their posture and avoid neck muscle overextension thus providing greater range of motion for the muscles. Additional study to investigate this finding and the effect of these postural changes on resulting brain strain during impacts to the head is warranted as future work.

This study has several limitations. The most prominent is its applicability to higher-load conditions. The forces and kinematics presented in this study are significantly lower than those that would be observed during a concussive-level event. However, since it is not ethical to perform such a test on a human participant and the effect of muscle activation may not accurately be tested on a cadaver, this study still serves to increase our knowledge about head kinematics in response to postural and muscle activation changes and helps improve computer simulation. Another limitation of this study is that by changing posture, the effective moment arm of the applied force may change. Due to the unique design of this testbed, the exact change in moment arm could not be calculated and therefore we were restricted to only reporting gross posture changes and resultant kinematics. For this purpose, we published our dataset, including all forces, muscle activation, head kinematics from the mouthguard, and MoCap data. Another limitation was our use of the Vector Mouthguard to quantify head kinematics. Though there are more accurate instruments available, studies confirmed the accuracy of the instrumented mouthguards (Bartsch et al., 2014; Camarillo et al., 2013; Liu et al., 2020), and they are widely used in field studies, though there could be errors associated with their results (Siegmund et al., 2016) that can add or subtract a bias to PAV, PAA, or PLA. We believe that such a bias would not affect the relative differences between the conditions. Lastly, although gender differences are reported in head kinematics (Collins et al., 2014; Gutierrez et al., 2014; Tierney et al., 2005), we only recruited male participants in this study phase, but we plan to elaborate on these differences in a future study with both males and females.

In conclusion, we measured reductions in PLA and PLV with isometric cervical muscle contraction, co-contraction, and postural changes in all force directions and reductions in PAA in the lateral direction compared to the control condition. Isometrically contracting the neck muscle would be a more important contributing factor to affect the head kinematics than changing the head posture in sub-concussive-level impulsive force. We did not measure a major difference between isometric contraction and postural changes in the resultant head kinematics; however, we expect to lower the risk of muscle damage with the postural changes compared to muscle co-contraction alone. Future studies are needed to investigate the effect of gender on postural changes as well as concussive level forces on head kinematics in simulation studies using these data as the first step toward validation.

Link to the dataset: http://bit.ly/Smart_Helmet_Data.

CRediT authorship contribution statement

Mohammad Homayounpour: Conceptualization, Methodology, Software, Formal analysis, Investigation, Resources, Data curation, Writing – original draft, Writing – review & editing, Visualization, Supervision, Project administration. **Nicholas G. Gomez:** Conceptualization, Methodology, Software, Formal analysis, Investigation, Resources, Data curation, Writing – original draft, Writing – review & editing, Visualization, Supervision. **Anita N. Vasavada:** Conceptualization, Software, Writing – original draft, Writing – review & editing, Visualization, Supervision. **Andrew S. Merryweather:** Conceptualization, Methodology, Formal analysis, Investigation, Resources, Writing – original draft, Writing – review & editing, Visualization, Supervision, Project administration, Funding acquisition.

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2021.110732.

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