

**Integration of Muscle Pre-tension and Activation to Evaluate Neck Muscle Strain
Injury Risk during Simulated Rear Impacts Using a Finite Element Neck Model[†]**

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20

21 **Abstract**

22 Prevention of rear-impact neck injuries remains challenging for safety designers due to a
23 lack of understanding of the tissue-level response and injury risk. Soft tissue injuries have
24 been inferred from clinical, cadaveric, and numerical studies; however, there is a paucity
25 of data for neck muscle injury, commonly reported as muscle pain. The goal of this study
26 was to investigate the effect of muscle pre-tension and activation on muscle strain and
27 injury risk resulting from low-severity rear impacts using a detailed finite element head
28 and neck model (HNM).

29 The HNM was extracted from the GHBMC average stature male model and re-postured
30 to match a volunteer study, with measured T1 kinematics applied as boundary conditions
31 to the HNM. Three cases were simulated for three impact severities: the baseline
32 repostured HNM, the HNM including muscle pre-tension, and the HNM with muscle pre-
33 tension and muscle activation. The head kinematics, vertebral kinematics, muscle strains,
34 and three neck injury criteria were calculated to assess injury risk.

35 The kinematic response of the neck model demonstrated an S-shaped pattern, followed
36 by extension in the rear impact cases. The maximum kinetics, kinematics, and muscle
37 strains occurred later in the impact during the extension phase. The distribution and
38 magnitude of muscle strain depended on muscle pre-tension and activation, and the
39 largest predicted strains occurred at locations associated with muscle injury reported in
40 the literature. The HNM with muscle pre-tension and muscle activation provides a tool to
41 assess rear impact response and could inform injury mitigation strategies in the future.

42

43 Keywords: Whiplash, muscle injury, rear impact, finite elements, human body mode,

44 muscle activation

45

1 Introduction

A recent study investigating vehicle crash data from 2001 to 2021 found that 25% of the cases corresponded to rear impacts¹. For these rear impacts, 35% of the incidents required medical treatment and 14% required hospitalization¹. Whiplash-associated disorders (WADs) are of special interest in rear impacts due to the high economic cost involved, approximately \$2.7 billion annually in the United States². Although extensive research has been undertaken to investigate the cause and risk of such injuries in automotive impacts³⁻⁵, a full explanation of the locations and tissues injured, has not yet been established.

Previous sled test studies with both post-mortem human subjects (PMHS) and living human volunteers have shown that during rear impacts, the neck, initially considered to be in a neutral posture (lordotic curvature), quickly transitions into upper cervical spine flexion and lower cervical spine extension (S-shape), followed by the entire neck rotating into extension (C-shape)⁶. Some studies have proposed that injury occurs during the S-shape phase⁶.

The soft tissues of the neck are frequently discussed as potential sources of pain⁶⁻⁸. The facet joints, spinal ligaments, intervertebral discs, vertebral arteries, dorsal root ganglia, and neck muscles were identified in previous studies as possible locations of injury in WAD⁹. Cadaveric, surgical, magnetic resonance imaging and animal studies have correlated damage to the ligaments, intervertebral disc, and facet joint with WADs⁹. However, few investigations of injury in the neck muscles exist even though electromyographic measurements and computational studies indicated the possibility of injury in this tissue due to active contraction against a stretching motion during impacts^{7,10}.

In addition, identifying soft tissue injury thresholds using anthropometric testing devices (ATD) and PMHS has proved challenging as there are large differences between these surrogates and the in vivo tissue behavior.

One possible solution is the use of computational human body models (HBMs) with sufficient detail to assess tissue-level responses. The head and neck of the Global Human Body Model Consortium (GHBMC) model for a 50th percentile young male (M50-O version 5.1) is a good candidate for investigating neck muscle strains during impacts¹¹. The M50 (Figure 1) includes detailed representations of the muscles, skin, ligaments, intervertebral discs, vertebrae, and head. One-dimensional Hill-type elements represent the active muscle contraction and three-dimensional hexahedral elements with visco-hyperelastic response represent the passive muscle tissue response.

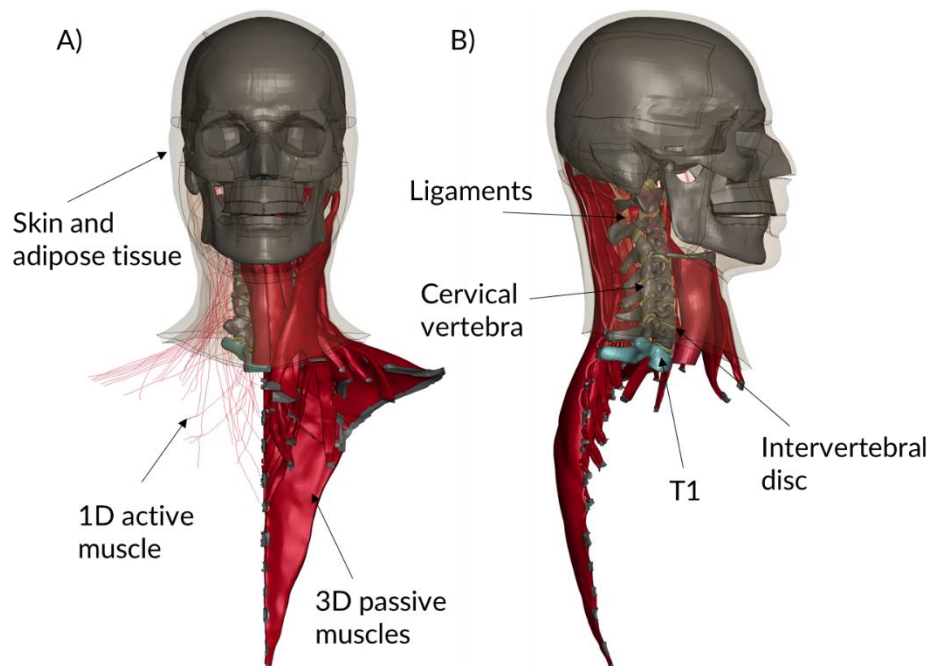


Figure 1: A) Frontal view of the M50 head and neck model with the right three-dimensional passive muscle elements hidden. B) Lateral view of the M50 head and neck model with all the right muscle elements hidden.

85 Recent studies have enhanced the GHBMC head and neck model to improve the
86 muscle tissue response in impact scenarios¹²⁻¹⁴. A novel enhancement to the GHBMC
87 neck musculature was the constrained beam in solid (CBS) methodology, which coupled
88 the one-dimensional Hill-type active elements with the three-dimensional solid passive
89 muscle tissue elements^{11,13}. The CBS method enabled improved prediction of strain in
90 the muscles and no longer required the unphysical attachment elements between the Hill
91 active elements and the vertebra. The CBS method improved the muscle implementation
92 to represent biofidelic muscle strains for the first time^{11,13}. Another study showed that the
93 model presented good correspondence to multidirectional human volunteer impact
94 studies for a physiologically based closed-loop controller based on the vestibulocollic
95 reflex (VCR) and cervicocollic reflex (CCR) mechanisms of the neck¹⁵. The VCR
96 modulates muscle activation by structures in the inner ear that measure the head
97 kinematics, and the CCR modulates muscle activation by specialized cells that measure
98 muscle stretch. The closed-loop controller used proportional-derivative (PD) controllers,
99 as they agree with the equilibrium-point hypothesis¹⁶ and were able to represent reflex
100 mechanisms in previous studies¹⁷. The closed-loop activation used the head center of
101 gravity rotation in the X, Y, and Z axes as input of the PD controllers representing the
102 VCR and the 1D Hill-type element stretches as input to the PD controller representing the
103 CCR.

104 In a recent HBM study¹¹, a muscle pre-tension method was proposed for initiating
105 stresses present in the muscle due to *in situ* pre-stretch, before the impact. The pre-
106 tension stress in the model was obtained by deforming the muscle from the slack length

(relaxed and non-deformed length) to the model *in situ* length (length in an upright neutral posture). Based on estimations from cadaveric measurements, the flexor slack lengths were 8% shorter than the *in situ* muscle lengths, while the extensor slack lengths were 3% shorter than the *in situ* length¹¹. Implementing pre-tension improved the upper cervical spine kinematics, better representing the S-shape reported in the literature for volunteer impacts.

Ideally, the validation of simulated active muscles in HBMs should be achieved by measuring muscle forces *in vivo*, but this remains a large challenge to ethically and safely conduct such experiments in humans¹⁸. Alternatively, human volunteer experimental data reporting vertebral kinematics could help to assess the neck muscle implementation response. Sato et al.¹⁹ conducted a study to measure the neck dynamics of male and female volunteers in low-severity rear impacts. For one of the three rear impact pulses analyzed by Sato et al.¹⁹, the individual vertebral kinematics were recorded with cineradiography during 4g rear impacts and, therefore, could be used for comparison to a neck model at the vertebral level. In addition, a related work²⁰ measured the average posture of males seated in a similar seat from Sato *et al.*¹⁹ Posture has been shown to be an important factor in neck response during impacts²¹. It is important to note that even though studies with volunteers are better suited for the analysis of models with muscle activation, past cadaveric studies also presented vertebral kinematics for rear impacts, like Deng et al. 2000²² with higher severities (>6g) and Stemper et al. 2004²³ for lower severities (4g). In addition, cadaveric studies have also been used to correlate intervertebral rotations to potential soft tissue injury risk²⁴.

The assessment of neck injury risk in ATDs and models is achieved using injury risk criteria. Over the years, different criteria have been proposed^{9,25}: the Neck Injury Criterion (NIC), based on pressure gradients relation to the spinal cord damage, is calculated using accelerations and velocities of C1 and T1 vertebrae; the intervertebral Neck Injury Criterion (IV-NIC), is defined as the intervertebral motion divided by the physiological range of motion; the Nij injury criterion, based on a combination of the axial force and the bending moment in frontal impacts; the N_{km} , based on a combination of the axial force and the bending moment in rear impacts; the Lower-Neck Load Index (LNL), based on the loads at the T1 vertebra related to facet joint injury; and the Neck Displacement Criterion (NDC), based on graphs of the angular velocity and displacement of the head relative to the T1 vertebra.

An alternative approach to evaluating injury risk in detailed HBMs is a tissue-level assessment²⁶ directly comparing the values of muscle strains from the model and the experimental data on muscle tears or pain. The range of reported deformations over which muscles rupture, observed in experiments with excised specimens, is relatively large. Experimental data reported engineering strain at failure ranging from 0.23 to 0.42²⁷ for active stretch, and failure Green strain between 0.50 and 0.62 when no activation was present^{28,29}.

In the present study, the GHBMC M50-O head and neck model was used to assess how the combination of muscle activation and pre-tension alters the distribution of 3D muscle strains in rear impacts. The strains were compared with experimental data on muscle tissue failure, and the head and neck response with traditional neck injury criteria, to assess the relation of muscle strain to injury risk.

2 Methods

The head and neck regions were extracted from the GHBMCM50-O v5.1 (M50) model, representing a 50th percentile male. The head and neck posture was changed¹¹ to match the volunteers sitting in a sled in rear impact experiments²⁰. The muscles of the model were updated by implementing the CBS methodology, and the CBS was integrated with muscle pre-tension¹¹ and open-loop active muscle control¹⁵.

Rear impacts were simulated by applying measured T1 kinematics from 4g rear impact experiments with volunteers from the literature¹⁹. The boundary conditions of higher severity impacts (7g and 10g²²) from PMHS testing were also simulated.

To assess the effects of the different muscle implementations, the rear impact scenarios were simulated for three different versions of the model: the M50 extracted from the full body model; the M50 model with muscle pre-tension, but no muscle activation (M50_P); and the M50 model with pre-tension and closed-loop muscle activation (M50_{PC}). Finally, the head kinematics, vertebral kinematics, muscle strains, and three neck injury criteria were calculated to quantify response and injury risk in the simulated scenarios.

2.1 Neck muscle pre-tension implemented in the M50_P and M50_{PC} head and neck model

The neck muscle pre-tension was defined in this work as the muscle stretch corresponding to a neutral, relaxed posture before the simulated impact or muscle activation. The objective of the muscle pre-tension was to better replicate stresses present in the muscle tissue *in vivo*. The pre-tension stress in the model was obtained by deforming the muscle from the shorter muscle slack length (muscle length with no

stresses) to the longer *in situ* length (muscle length in a neutral position prior to the impact simulation).

The simulations were carried out in 32 cores using a single precision commercial FE solver (LS-DYNA, R.9.3.0). The stress initialization for muscle pre-tension was conducted in two steps as reported in a recent study¹¹. First, a dynamic relaxation simulation was conducted to obtain the deformations of the muscle from the slack length to the *in situ* length. Second, an explicit simulation was run for 200 ms with T1 fixed and gravity loading to allow the passive pre-stretched musculature to reach a quasi-equilibrium state. The two steps and the subsequent impact were conducted directly one after the other in the same simulation. The 200 ms stabilization period was run for the three models and led to initial strains (pre-stretch) in the muscles. The postures of the three models were not the same after the stabilization.

2.2 Closed-loop neck muscle activation implemented in the M50_{PC} head and neck model

The closed-loop muscle activation was based on a controller scheme from a previous publication¹⁵ based on the physiological reflex mechanisms of the neck (Figure 2).

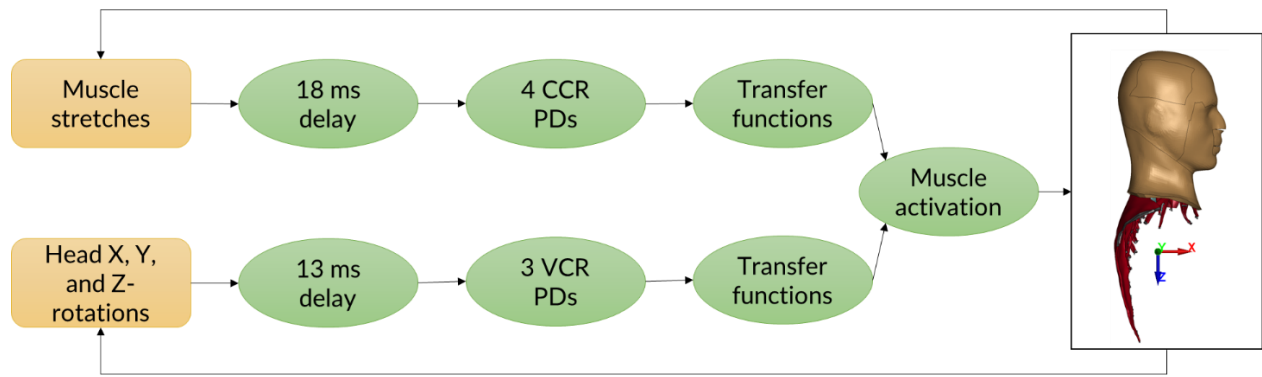


Figure 2: Closed-loop controller schematic of the muscle activation. The head rotation and muscle stretches are the inputs of the controller. The CCR and VCR delays for each of the four muscles were based on experimental data. The delays are applied to the inputs using PD controllers. The outputs of the PDs are combined and transformed into an activation level to be applied to the muscle through transfer functions.

For the current study, the controllers were updated with a time delay of 13 ms for the VCR and 18 ms for the CCR, based on experimental data³⁰. The 13 ms delay was the time needed for the signal generated by the head rotation to affect muscle activation. The 18 ms delay was the time needed for the signal generated by the muscle stretch to affect the muscle activation. The input of the CCR was simplified as the changes in length between the origin and insertion point of the Hill-type beam elements at the medial portion of the trapezius and the sternocleidomastoid muscles (Figure 3). In addition, each one of the four muscle groups (right extensors, left extensors, left flexors, and right flexors) was controlled separately¹⁵.

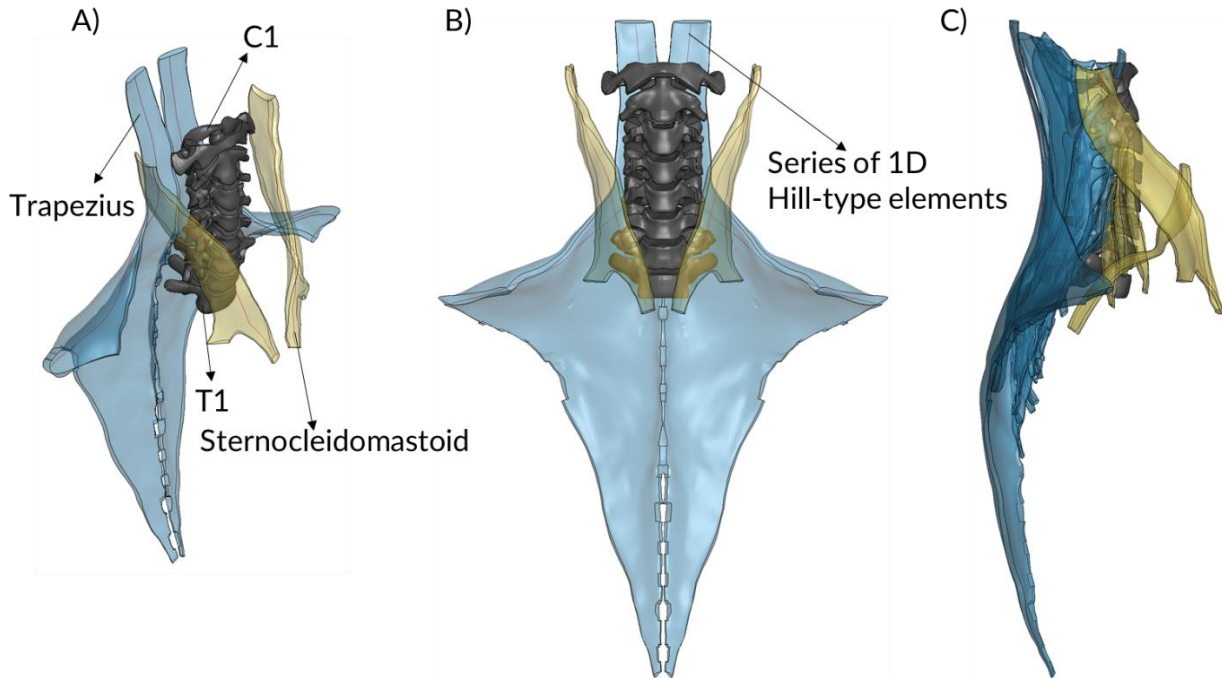


Figure 3: A) Isometric view of the sternocleidomastoid, trapezius, and cervical spine. B) Frontal view of the series of 1D Hill-type elements (red lines) used as input for the closed-loop controller. C) Lateral view of the flexors (red) and extensors (blue).

The elements for the CCR input were selected from the muscles presenting the highest force and deformation after observing frontal and rear impact simulations without muscle activation. Due to the muscle pre-tension, the controller parameters had to be optimized for this new scenario. A Python code was implemented to optimize the proportional and derivative gains of the PD controllers. The optimization fitted the activation curves of the closed-loop controller to the co-contraction activation curves of an open-loop controller¹⁵. The optimization used the head rotation and muscle stretches from the open-loop controller simulation as the input signals. For the VCR, the proportional gain obtained through this methodology was 0.44 and the derivative gain was equal to 14.9. For the CCR, the proportional gain was 0.063 and the derivative gain was equal to 0 for the extensors, and 0.28 and 4.5, respectively, for the flexors. The VCR

maintained an activation ratio of 1:5 between the extensors and flexors to maintain the cocontraction ratio from a previous study¹⁵. The VCR gains were positive for the antagonistic muscle groups, relative to the head rotation, and negative for the agonist muscles. The CCR gains were positive for lengthening and zero for shortening of the muscles relative to the resting length. It is important to note that, during the simulation, Massively Parallel Processing (MPP) decomposition was used so the 3D muscles and 1D Hill elements calculations occurred in the same core. The MPP decomposition was necessary to ensure numerical consistency among the two muscle parts.

2.3 Rear impact simulation

Low-acceleration rear impact simulations were run using three different models: M50, M50_P, and M50_{PC} to calculate the muscle deformations during the impact. The average X-displacement, Y-rotation, and Z-displacement of the T1 vertebra of the experiment with 4 male volunteers from Sato *et al.* 2014¹⁹ were applied to the T1 vertebra of the models. The X-rotation, Z-rotation, and Y-translation of T1 were free to move. The muscle, skin, and flesh ends were constrained to T1 to represent the boundary condition of the torso (Appendix Figure A1). These boundary conditions represented a rear impact with a peak acceleration of approximately 4g and were selected as the recorded vertebral kinematics were available for comparison with the model results. The applied X-displacement was obtained by double integrating the X-acceleration presented in the literature. Also, two additional 7g and 10g impacts were simulated with the M50_{PC} for a subsequent analysis of the effect of the severity on the possibility of injury. The M50_{PC} was selected for the severity analysis because it presented the highest muscle strains and therefore the most injurious scenario. For these high-severity impacts, the X-

acceleration, Y-rotation, and Z-acceleration from Deng et al. 2000²² study with PMHS were applied to the T1 of the M50_{PC}.

The resultant head and vertebral kinematics of the simulations were extracted and compared to the experiments to verify the simulations. The 95th percentile of the maximum principal strains (MPS) from the muscles were extracted and compared with muscle failure data. The strain time histories and strain distribution were inspected for the 3D passive elements in all simulations. The highest spurious values were associated with the muscle-tendon interface, and the 95th percentile criterion consistently reported the relevant muscle strains occurring in the muscle while excluding the high strains at the tendon-muscle interface. (Figure A4, Appendix). Furthermore, the Neck Injury Criterion (NIC), the N_{km} injury criterion, and the intervertebral Neck Injury Criterion (IV-NIC) were calculated for the simulations to compare the capabilities of these criteria to identify possible muscle injury. Although N_{km} is usually measured at the occiput (close to C0/C1), this criterion was used in all vertebral joints for comparison to the IV-NIC. The forces and moments at the cervical spine required to calculate the injury criteria were obtained following a similar approach to the cross-section method used by Johnson *et al.* 2020³². No filtering was applied to the force and moment histories, and the data was sampled at 200 Hz. In addition, the physiological ranges of motion of each vertebral level were based on the averages found by a study with cadaveric specimens³³. The intercept values necessary to calculate the injury criteria were obtained from the literature (Table 1).

Table 1: Intercept values for the calculation of the injury criteria.

Criterion	Intercept values	Injury threshold
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NIC	-	NIC=10 ⁹ (Pain threshold)
N _{km}	$F_{int} = 845 \text{ N}^{34}$ $M_{int} = 47.5 \text{ N m/ } 81.1 \text{ N m}^{34} \text{ (extension/}$ flexion)	N _{km} = 1 ³⁴

3 Results

3.1 Low-severity 4g impact simulations with the M50, M50_P and M50_{PC} head and neck models

All the obtained results were zeroed to the 0 ms time at the onset of the impact. The kinematics of the center of gravity (CoG) of the head were extracted from the rear impact simulations. The CoG X-displacement of the M50 and M50_P were within the corridors of one standard deviation of the experiments¹⁹ (Figure 4). However, the M50_{PC} X-displacement was slightly higher than the corridors. The CoG Y-rotational displacement presented a similar trend to X-displacement initially. The M50 and M50_P models were inside of the experimental corridors up to the rebound phase at 170 ms, while the M50_{PC} were inside during rebound up to 220 ms. The contrasting M50_{PC} response to the other models was a result of the high forces generated by the small activation of the extensors during the stabilization phase of the model. Although this effect could be reduced by modifying the muscle activation parameters, for consistency with the previously published model, the activation parameters were kept constant throughout the study. The extensors are larger muscles, even a low activation was enough to increase the head rearward excursion in the M50_{PC} model (Figure 5). In addition, the developed M50_P and M50_{PC}

models took 3 days to run on 32 CPUs, mainly due to the additional time required for the dynamic relaxation.

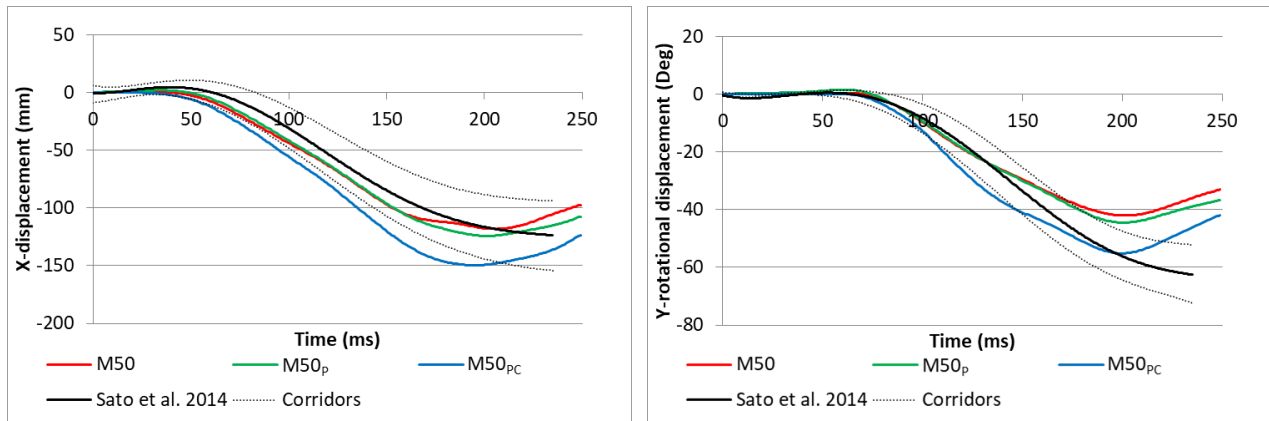


Figure 4: Kinematics of the head CoG for the 4g rear impact simulations. The X-displacement was calculated relative to T1. The corridors represent one standard deviation from the experimental average from Sato et al 2014¹⁹.

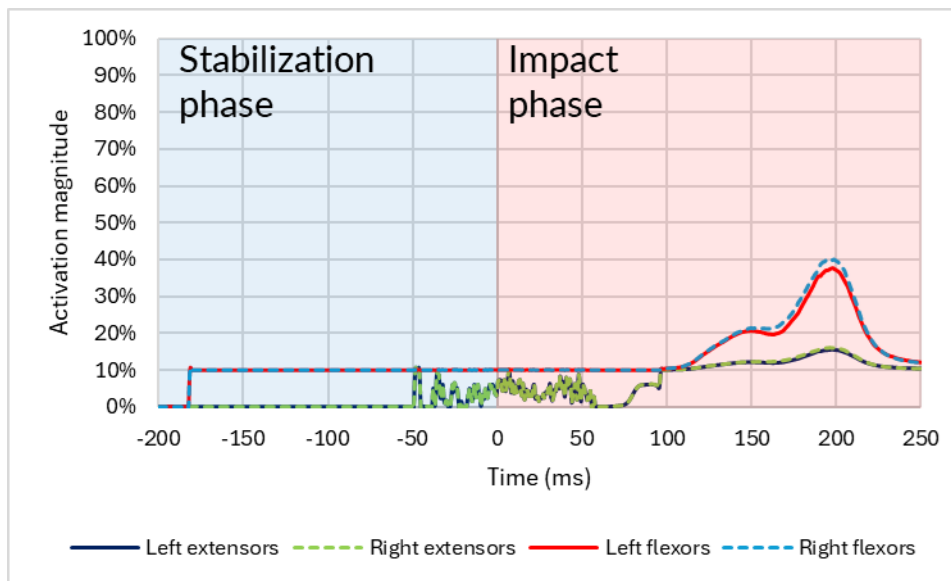


Figure 5: Muscle activation level for the M50PC in the 4g impact simulation. The muscles presented a baseline activation of around 10% before the impact. The timing of the muscle reflex activation was around 96 ms, inside the expected EMG values between 55 and 99 ms³⁵. In addition, the activation magnitude of the extensors was lower than the flexors as expected. The flexors quickly deformed enough to achieve full CCR activation

(10%) while the extensor oscillated around $t=0$. The oscillation was due to the neck achieving the final posture while the extensors presented some small variation on their stretch. The oscillation stopped after the impact reduced the deformation of the extensors.

The intervertebral rotations were extracted from the simulations, from C0 (skull) to C7 (Figure 6). The upper cervical flexion was more pronounced for the M50_{PC} during the S-shape phase and the neck extension was more pronounced during the C-shape phase. The extension of the vertebra of the M50_{PC} was outside the corridors due to the activation of the extensors. Overall, the model joint kinematics followed trends evident in the experimental response.

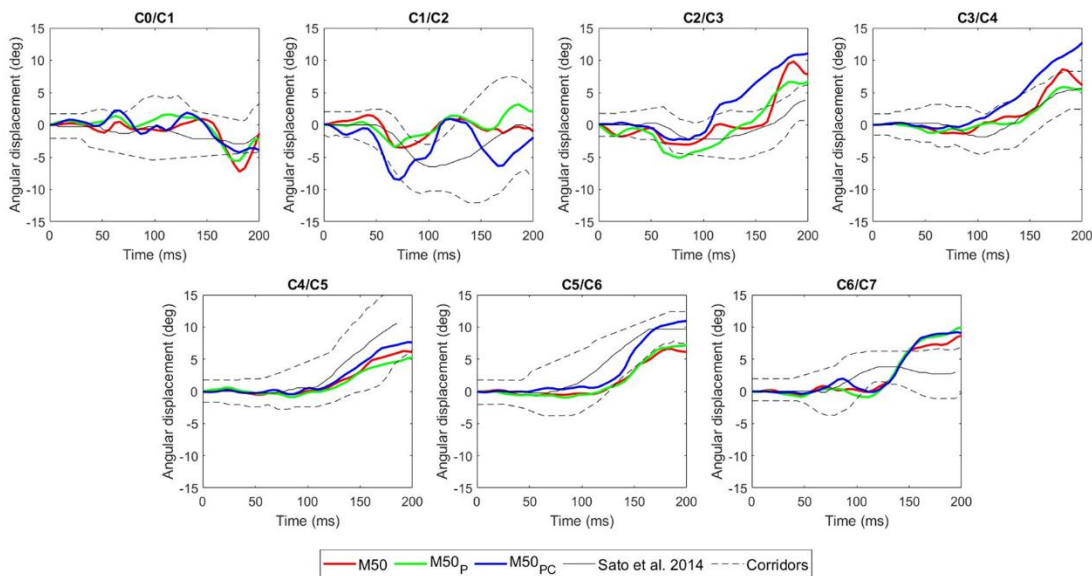
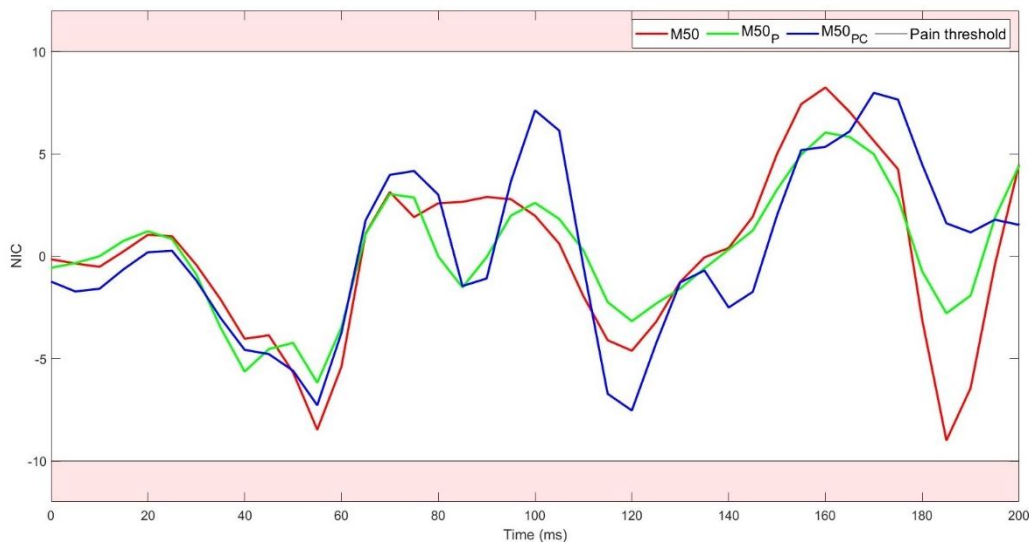


Figure 6: Intervertebral rotations from C0 (skull) to C7 for the three models. The corridors represent one standard deviation of the experimental data. The negative direction indicates flexion, and the positive direction indicates extension.

All three models presented similar trends concerning the injury criteria. The NIC (Figure 7) of all the models were below the pain threshold. The N_{km} (Figure 8) of the

313 models were low during the S-shape phase (50 to 100 ms) but increased at the C-shape
 314 phase (above 100 ms). The IV-NIC (Figure 9) of the models also presented values higher
 315 than 1 during hyperextension of the C-shape phase. However, the M50_P crossed the
 316 value of IV-NIC=-1 at the C2-C3 joint during flexion. Overall, the model with pre-tension
 317 and without muscle activation obtained the lowest values for all injury criteria. In addition,
 318 the LNL was calculated for the models and indicated the same trends from the other injury
 319 criteria.

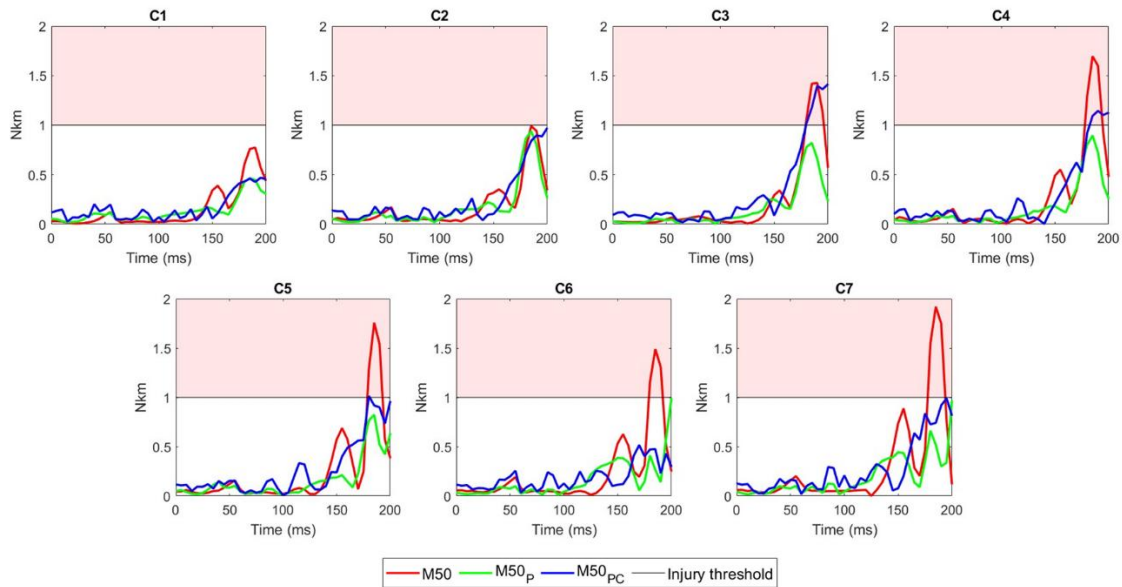


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321 *Figure 7: NIC values for the rear impact simulation for all three models were below the*
 322 *experimental pain threshold (NIC=10)*

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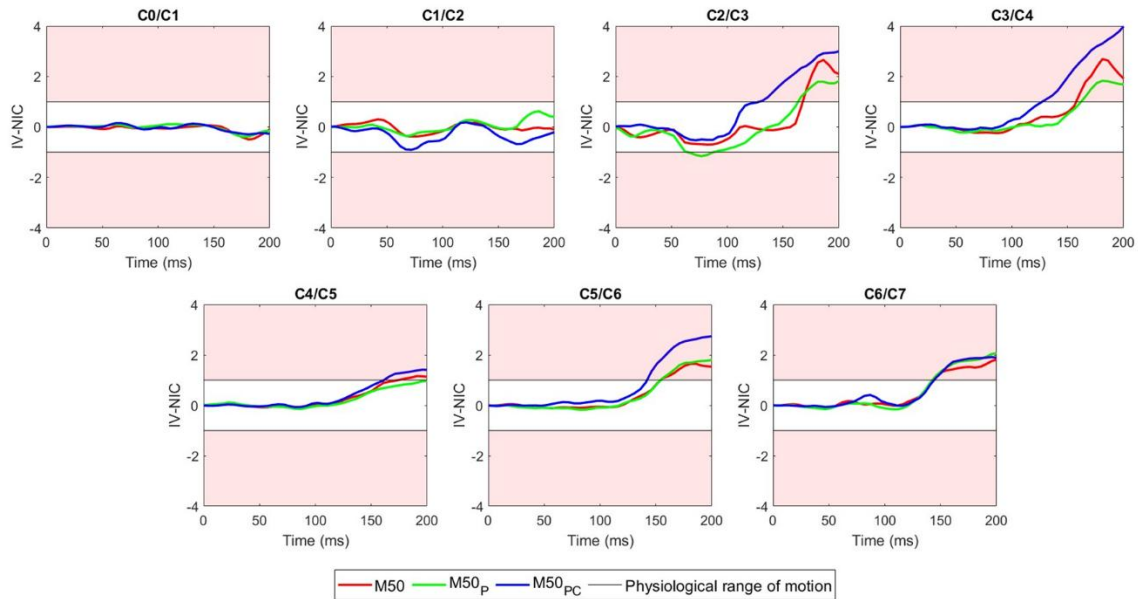
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326 *Figure 8: N_{km} values for the rear impact simulation for all three models. When calculating*
327 *the N_{km} , the value of the M_{int} in flexion was used for the vertebral flexion and the M_{int} in*
328 *extension was used for the vertebral extension.*

329



330

331 *Figure 9: IV-NIC values for the rear impact simulation for all three models. When*
 332 *calculating the IV-NIC, the value of the physiological range of motion in flexion was used*
 333 *for the vertebral flexion (negative values) and the physiological range of motion in*
 334 *extension was used for the vertebral extension (positive values).*

335 The distributions of strains in the muscles were different among the models (Figure
 336 10). For example, the sternocleidomastoid MPS was higher than the splenius capitis for
 337 the M50_P model, while the opposite was true for the M50 and M50_{PC}. Also, the M50_{PC}
 338 decreased the sternocleidomastoid MPS while it increased the longus capitis MPS
 339 compared to the M50_P. Additionally, the average MPS increased with the addition of pre-
 340 tension and increased again with the addition of muscle activation. The average 95th
 341 percentile MPS were 0.12, 0.13, and 0.15 for the M50, M50_P, and M50_{PC}, respectively.
 342 The three muscles presenting the highest strains were all extensors: the rectus capitis
 343 lateralis (strains from 0.29 to 0.32), the trapezius (strains from 0.20 to 0.27) and the
 344 semispinalis cervicis (strains from 0.18 to 0.23).

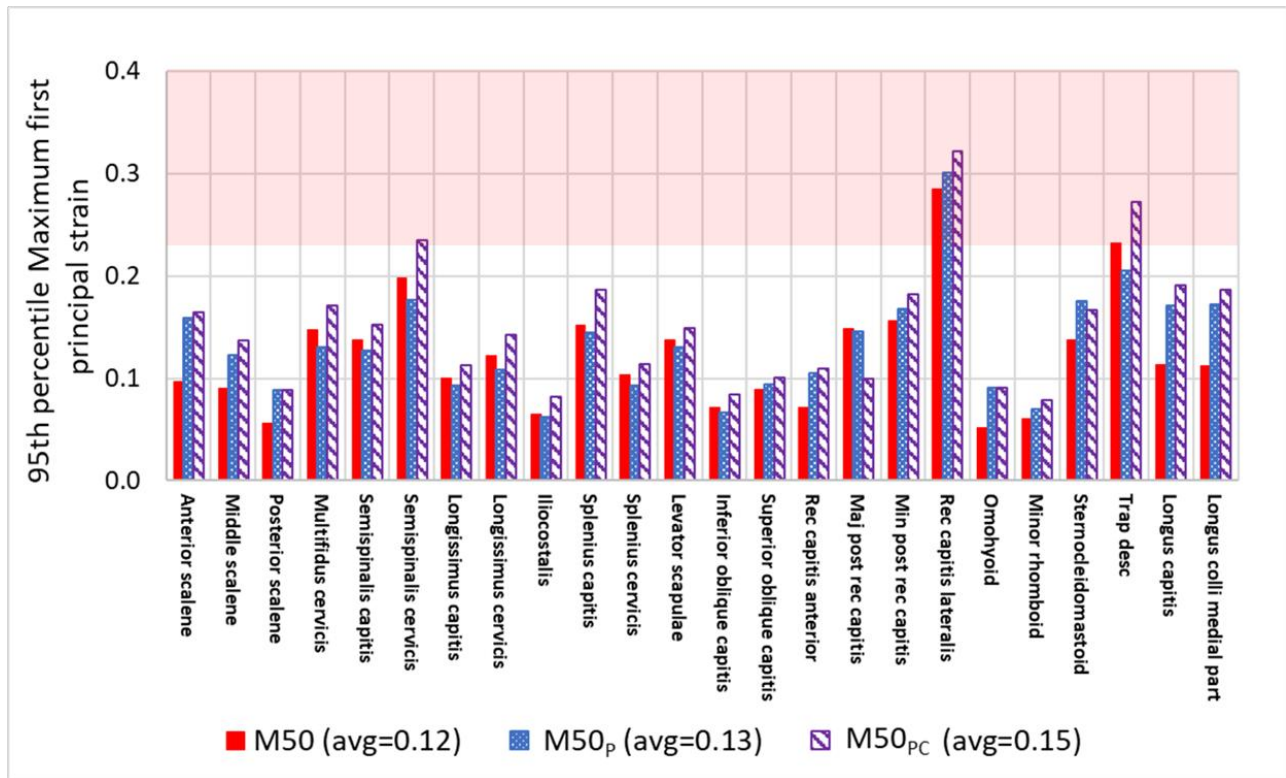


Figure 10: 95th percentile maximum first principal strain for each muscle in the 4g rear impact simulation for all three models. The average strains (avg) are shown inside parentheses. The red region shows values above the minimal injurious strain of 0.23.

3.2 High-severity 7g and 10g impact simulations with the M50_{pc} head and neck model

For the subsequent comparisons of increasing severity impacts, the intervertebral rotations were also extracted from the simulations, from C0 (skull) to C7 (Figure 11). As expected, the 10g impact generated the highest intervertebral angular displacements in flexion and extension. However, the 7g impact presented slightly lower intervertebral angular rotations compared to the 4g impact. The 10g simulation provided useful data up to a simulation time of 175 ms and then terminated due to numerical instabilities in the tendon elements at the level of C1. Based on kinematic observation of the tendon elements, the model works without issues up to 175 ms after the impact. The large rotations of the vertebrae were a result of the impact severity.

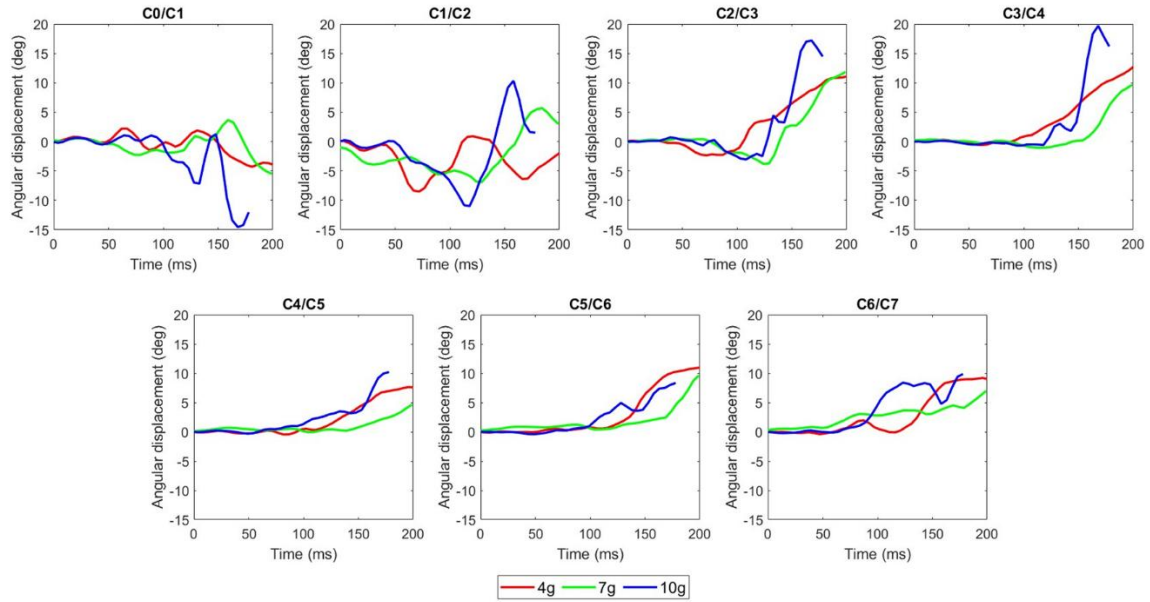


Figure 11: M50_{PC} intervertebral rotations from C0 (skull) to C7 for 4g, 7g, and 10g impact severities. The negative direction indicates flexion, and the positive direction indicates extension.

The calculated injury criteria values increased with impact severity except for the IV-NIC of the 7g case compared to the 4g case. The N_{km} (Figure 12) was low during the S-shape phase for all severities but increased at the C-shape phase. The IV-NIC (Figure 13) also presented values higher than 1 during hyperextension of the C-shape phase for all the severities. However, the 10g case crossed the value of IV-NIC=-1 at the C1-C2 joint during flexion.

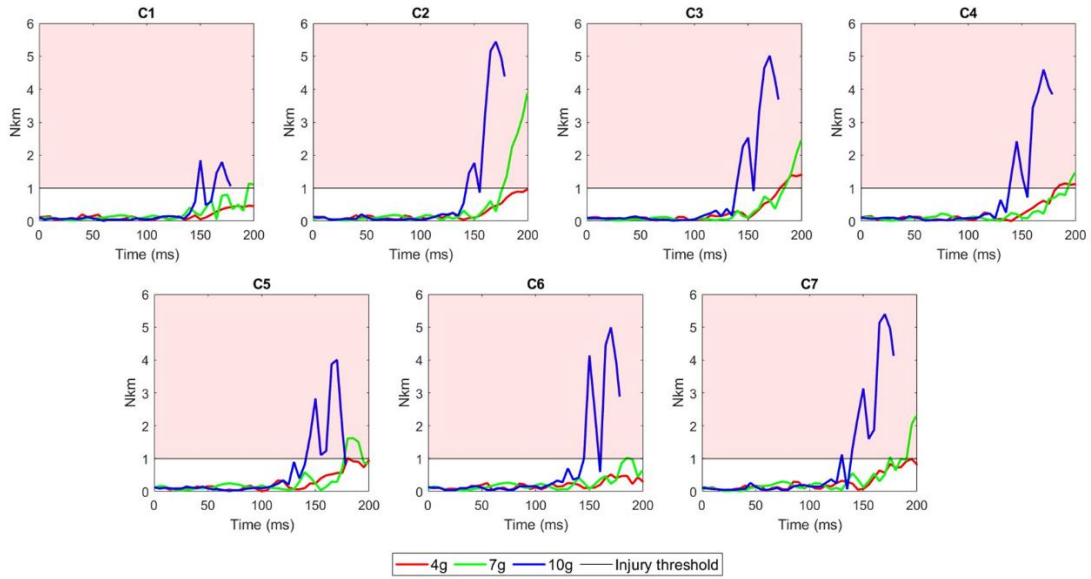


Figure 12: $M50_{PC}$ N_{km} values for the 4g, 7g, and 10g rear impact simulations. When calculating the N_{km} , the value of the M_{int} in flexion was used for the vertebral flexion and the M_{int} in extension was used for the vertebral extension.

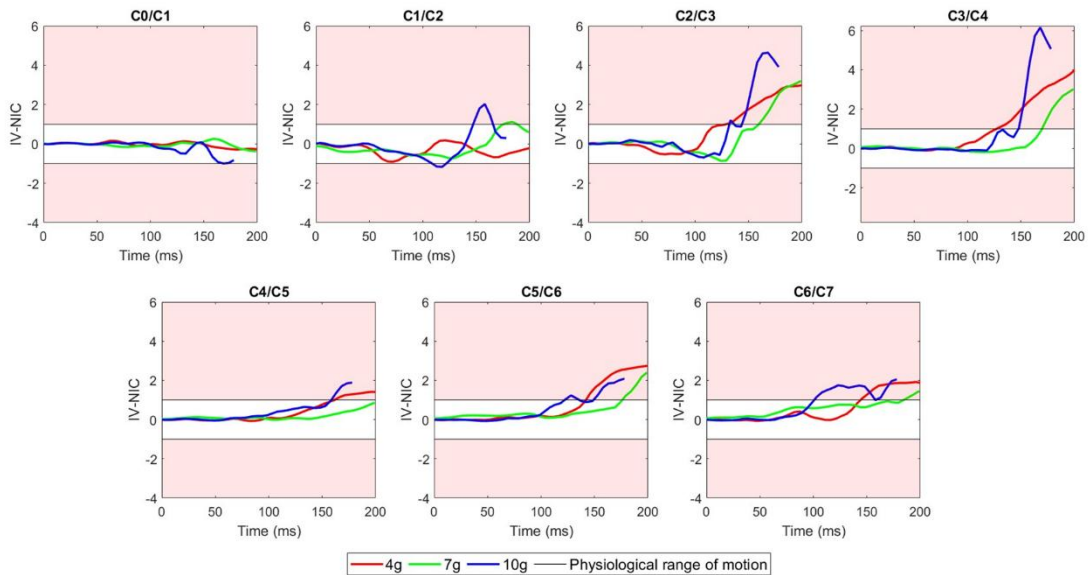


Figure 13: $M50_{PC}$ IV-NIC values for the 4g, 7g, and 10g rear impact simulations. When calculating the IV-NIC, the value of the physiological range of motion in flexion was used for the vertebral flexion (negative values) and the physiological range of motion in extension was used for the vertebral extension (positive values).

The highest average muscle strain (0.22) was found in the 10g impact while the 4g and 7g cases had the same average strain (0.15) (Figure 14). For all the severities, the three extensors presenting the highest strains were: the rectus capitis lateralis (strains from 0.25 to 0.44), the trapezius (strains from 0.27 to 0.41), and the semispinalis cervicis (strains from 0.21 to 0.33). However, for the 10g impact, a flexor, the rectus capitis lateralis, also presented a high strain (0.43) in contrast to the lower severities (0.11 to 0.15) in which only the extensors presented the highest strains.

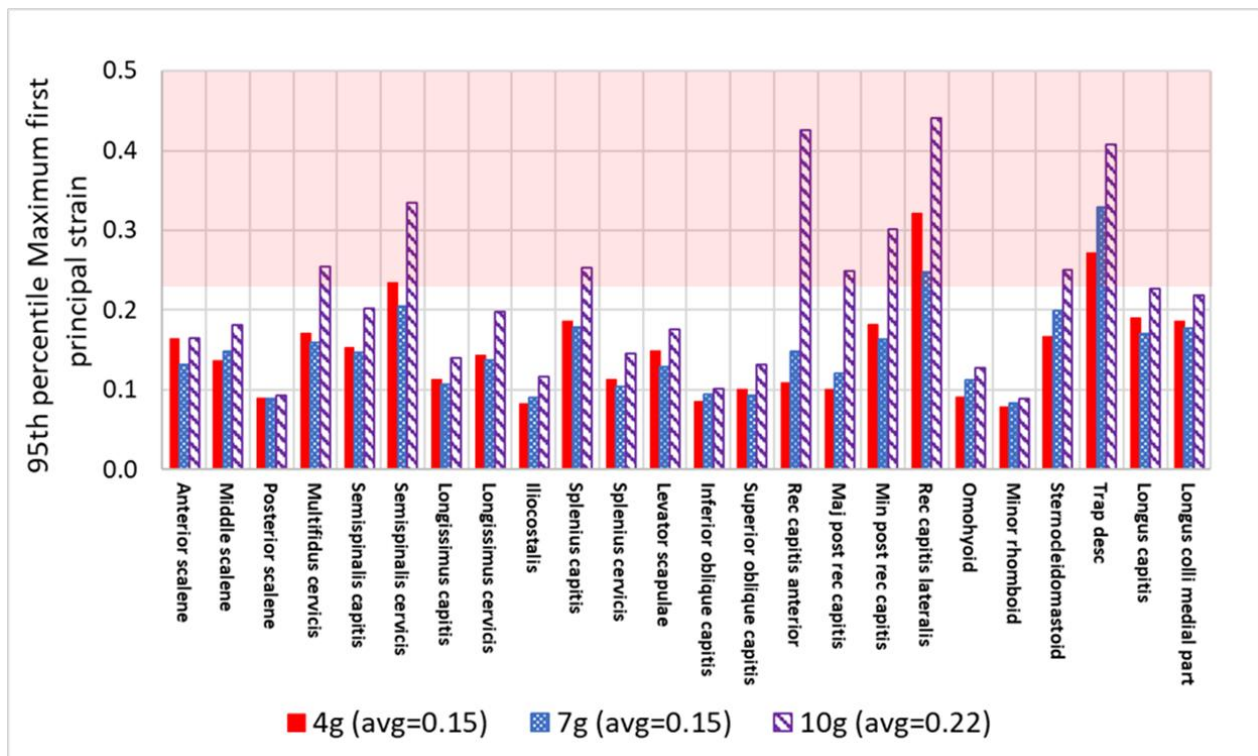


Figure 14: 95th percentile maximum first principal strain for each muscle in the M50_{PC} for the 4g, 7g, and 10g rear impact simulations. The average strains (avg) are shown inside parentheses. The red region shows values above the minimal injurious strain of 0.23.

4 Discussion

The integration of the muscle pre-tension and a closed-loop muscle activation controller was successfully implemented in the GHBMC head and neck model, resulting

in head and vertebral kinematics from the model that were within or close to the response corridors of the human volunteers for a 4g impact. The M50_{PC} obtained slightly higher head displacements and rotations, which is expected due to the additional force produced by the muscles, especially the larger extensors.

Compared to the M50, the M50_P increased the upper cervical spine flexion from 7.5° degrees to 8.9° degrees, and the M50_{PC} increased it to 11.1°. Both M50_P and M50_{PC} approximated peak C1-C4 flexion to the experimental average of 10.3° degrees as expected for a better representation of the muscle forces. Nevertheless, the additional loads of the muscle activation of the M50_{PC} were required to achieve C1-C2 peak flexion similar to the experiments. However, the M50_{PC} resulted in a higher extension at the C2-C3 level during the C-shape phase, indicating a tradeoff between these two phases in the current model. The simulated vertebral joints presented negligible shear (Appendix Figure A2 and A3) compared to the experiments from Sato et al. 2014¹⁹. The low shear possibly indicates differences in the intervertebral disk implementation and bony geometry that may have led to the trade-offs observed in the model.

The distributions of the muscle strains were different among models (Figure 10) even though the differences in head kinematics were small. This contrast shows that obtaining precise values for usually neglected aspects of the muscles such as pre-tension is essential to quantify soft tissue injurious strains in low severity impacts. The highest values of MPS in all models were in the trapezius, rectus capitis lateralis, and semispinalis cervicis. These three muscles are extensors. For the M50, the trapezius and rectus capitis lateralis MPS surpassed the lowest strain in which injury was identified in the literature (0.23)²⁷. For the M50_P, only the rectus capitis lateralis surpassed the lowest strain in which

injury was identified in the literature. For the M50_{PC}, the trapezius, rectus capitis lateralis and semispinalis cervicis MPS surpassed the lowest strain in which injury was identified in the literature. The muscle strains indicated a potential risk of injury in the posterior region of the lower cervical spine and at the posterior region near the base of the skull. The injury regions identified agreed with MRI studies^{36–38} of whiplash patients that identified a higher increase of fatty tissue (a signal of injury) at the semispinalis cervicis, multifidus, trapezius, and rectus capitis posterior major and minor (both in close proximity to the rectus capitis lateralis) after the impact. The possibility of injury also agreed with the reported pain felt by the volunteers during the sled tests¹⁹ used for the boundary conditions of the simulations.

Importantly, even with the higher flexion of the M50_{PC} the IV-NIC, and N_{km} indicated a higher risk of injury during the C-shape phase. The trend indicated by these injury criteria was in agreement with the timing of higher strains in the muscles. However, the pain threshold was not crossed by any of the models for the NIC, indicating that this metric may be insensitive to such low-severity impacts. The IV-NIC showed potential injury from C2 to C7 for all models. The N_{km} showed potential injury in all vertebral levels for the M50, no potential injury for the M50_P, and only potential injury from C2 to C4 for the M50_{PC}. The reduced sensitivity of the IV-NIC compared to N_{km} to the different models is expected due to the smaller changes in vertebral displacement compared to the larger changes in load distribution. For the IV-NIC, the C2-C3 values for the M50_P were slightly beyond the injury threshold, which may indicate that for higher severities of similar impacts, the S-shape may become increasingly injurious. This data also agrees with experimental data that indicated injury at C2-C3 and C5-C6 levels, showing the importance of muscle pre-

tension to the injury analysis at low-severity impacts. Nevertheless, the hyperextension phase still presented higher values of the injury criteria, which is in agreement with the analysis of tissues strains during neck extension conducted in the rectus capitis posterior minor⁸. Overall, the pre-tension reduced the injury criteria values even though it increased average muscle strains, indicating that the investigated injury criteria may not be adequate for understanding the soft tissue response. The ligaments, vertebrae and discs did not present injurious loads in any of the simulations, which was also in contrast to what was identified by the injury criteria. In addition, a new injury criterion relating force and activation of 1D Hill-type elements to muscle injury was proposed by Nölle *et al.* 2022³⁹, and could be adapted to the GHBMC model in the future to further investigate the muscle injury possibility.

The overall analysis of the increase in impact severity indicated that the vertebral angular displacements and calculated injury criteria were highest for the 10g case as expected. The 4g and 7g presented comparable values for the intervertebral kinematics, but not for the N_{km} . One possible explanation is the vertebral kinematics being maintained between these two severities may be due to muscle activation, which would explain the increase in loads for the 7g case. This would agree with previous studies⁴⁰ that indicated that muscle activation could help protect the neck against injury. Another possible explanation is the use in the simulations of boundary conditions from two different studies for the different severities; therefore, the 7g response of the model may not follow a progressive increase compared to the 4g pulse. However, the 7g and 10g cases were from the same study and therefore presented the expected increase in vertebral kinematics with severity. The distribution of muscle strain also increased for the 10g

impact with the extensors displaying the highest strain. In contrast, the M50_{PC} had one flexor, rectus capitis anterior, presenting one of the highest strains. This flexor is a small muscle at the base of the skull which explains the large deformations only when the muscle was actively contracted. The 10g impact IV-NIC also surpassed the physiological range of motion during the S-shape in opposition to the 4g or 7g cases. The high severity needed to onset the possibility of injury during the S-shape phase agrees with the literature that uses this mechanism to justify males being less prone to whiplash injury than females due to anthropometric differences^{21,41}. The 10g simulation terminated prematurely due to numerical instability in a tendon element connected to a muscle beam element. The small beam elements for the active muscle presented extremely high relative stretches near the tendons in the 10g impact severity generating instabilities during the Hill-type element muscle force calculation. The simulation could be run to completion if longer elements were implemented in the future.

The current study includes some limitations. The scarcity of *in vivo* force and strain data for muscle tissue and vertebral kinematics during rear impacts makes further validation of the simulated results challenging. Recent ultrasound studies^{42,43} were able to measure volumetric changes in large muscles during voluntary contraction and may be adapted to help validate the simulated deformations during impacts. For comparison with the IV-NIC, the N_{km} was used to observe the loads in all vertebral joints although it was developed for measuring loads at the occiput. It was assumed the N_{km} intercepts of all vertebral levels were the same as the ones of the occiput. However, obtaining N_{km} intercepts for each vertebral joint would be ideal. The simplified grouping of the muscles into flexors and extensors may not capture all the complexities of pre-tension and

activation of individual muscles. The sternohyoid and sternothyroid-thyrohyoid muscles were not considered in the strain analysis as they reported high localized strains in the T1 region, especially for the M50_{PC}. The localized high strains were deemed artifacts due to the muscles, adipose tissue and skin being fixed to T1 for the extracted head and neck model. In addition, the M50_{PC} model generated a higher head extension than the experiments, indicating the need to further adapt the closed-loop controller for a model with pre-tension. One parameter that could be modified would be the cocontraction ratio between flexors and extensors that were not updated for the model with pre-tension.

5 Conclusions

The current study was the first to analyze the effect of muscle pre-tension and muscle activation on head and vertebral kinematics, and the possibility of injury, using the M50 head and neck model.

The more pronounced upper cervical S-shape of the M50_{PC} better represented the experimental peak flexion but had a more pronounced C-shape and head kinematics than the tests with volunteers.

The muscle strain increased progressively with the inclusion of pre-tension and muscle activation. The distribution of muscle strains among the muscles varied for the three models. The M50_{PC} strains indicated the possibility of injury in muscles that better represented the regions of pain from experiments. Furthermore, only the model with muscle activation showed possible muscle injury in the deep muscles (semispinalis cervicis) as expected from the literature. In addition, the strains increased with impact severity as expected.

The predicted maximum muscle strains, and injury metrics in all but one model configuration identified that the C-shape was the main injurious phase for the impact severities investigated. Only the M50_{PC} indicated injury for the IV-NIC criterium at the C2-C3 level during the S-shape, as was previously suggested in the literature. It is important to note that the current study focused on the muscle response although other tissues such as the facet joints were raised as potential regions of WAD injury.

The current study was the first to analyze the relationship between muscle strain during rear impacts and neck injury criteria. Although further experiments are still needed to validate the current study results, the proposed models can help inform regions of interest for injury analysis and are a step toward improving understanding of soft-tissue injury mechanisms.

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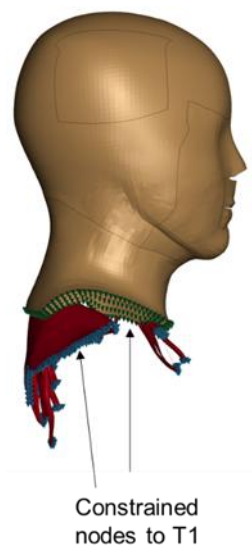
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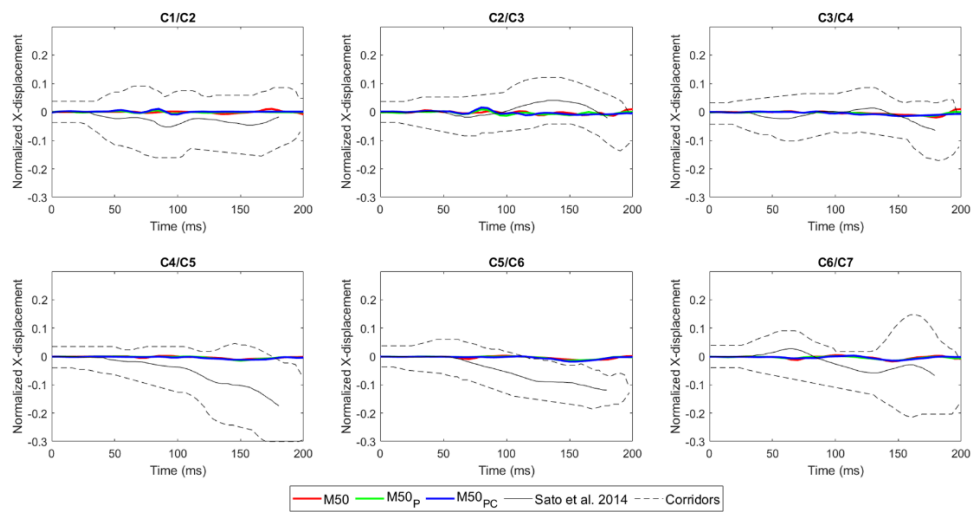
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654 *Figure A1: Muscle, skin, and flesh nodes rigidly constrained to T1.*
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657 *Figure A2: Intervertebral X-displacement from C0 (skull) to C7 for the three models. The*
658 *corridors represent one standard deviation of the experimental data.*

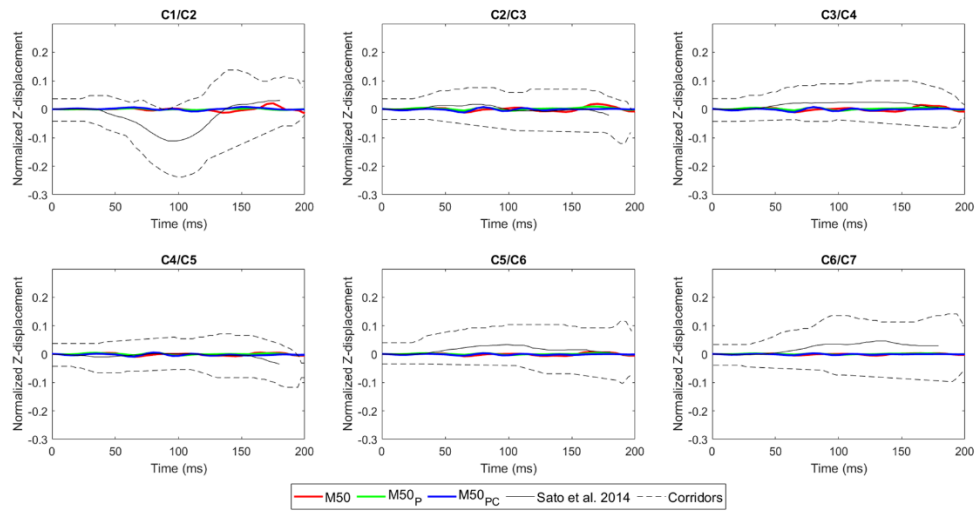


Figure A3: Intervertebral Z-displacement from C0 (skull) to C7 for the three models. The corridors represent one standard deviation of the experimental data.

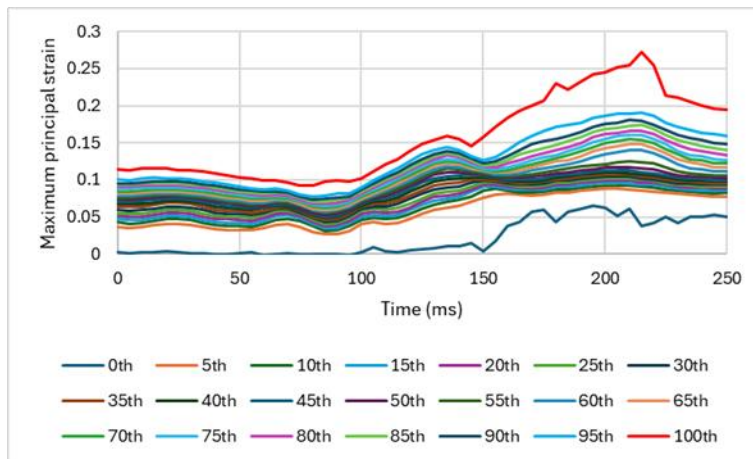
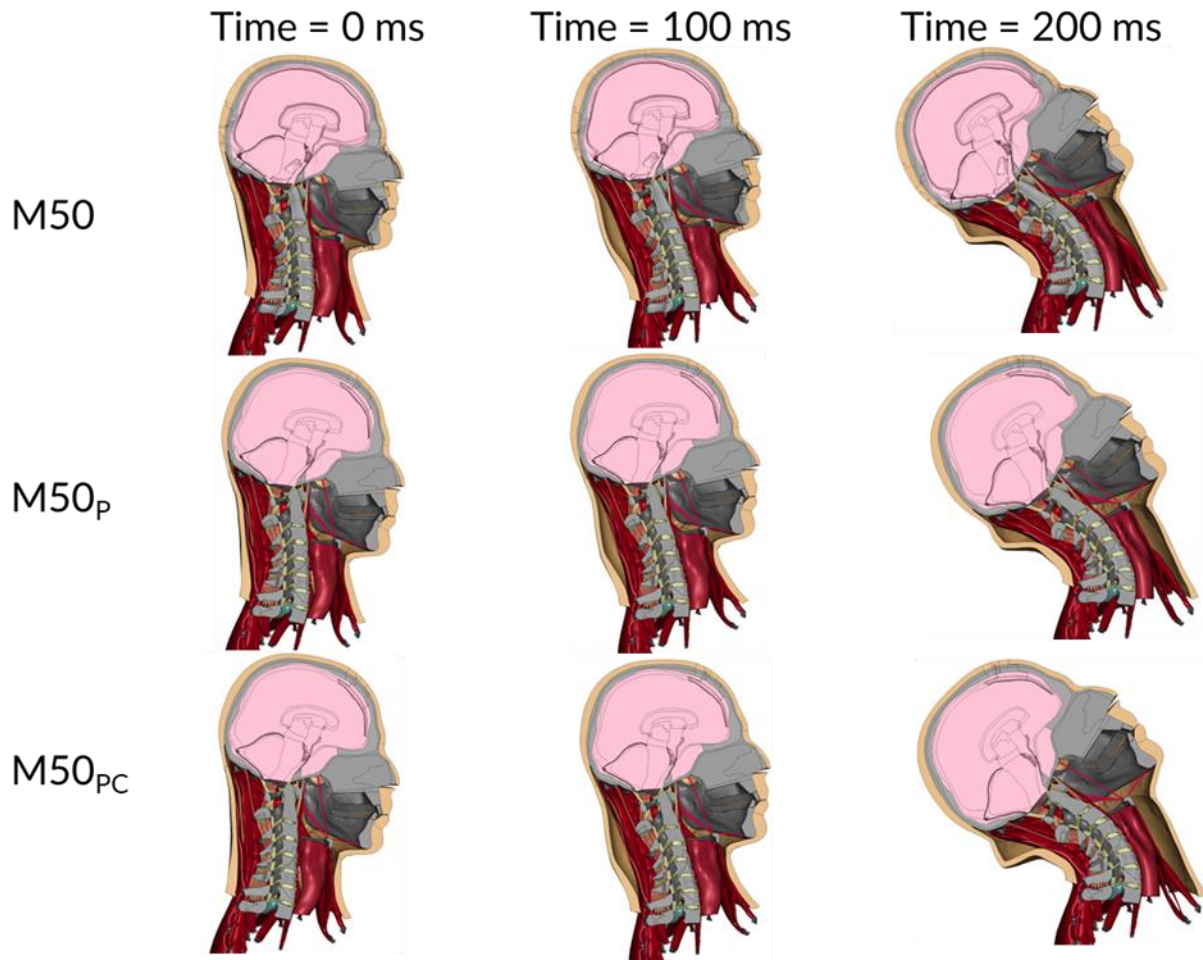


Figure A4: The MPS of the longus capitis are shown for the M50PC in the 4g impact for all percentiles. The 95th percentile included strains in the main portion of the muscle, while the 100th percentile strain corresponded to the muscle-tendon attachment.



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668 *Figure A5: Cross-section of the M50, M50P, and M50PC models in the 4g rear impacts*
 669 *at 0 ms, 100 ms, and 200 ms. The red rectangle shows the higher M50P and M50PC*
 670 *upper cervical flexion during the S-shape phase compared to the M50.*