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TITLE: Examining the Effects of Head Supported Mass on Cervical Spine Biomechanics and Injury Risk in Special Forces Operators

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14. ABSTRACT

Military personnel face unique challenges and occupational loads under difficult conditions that put them at significant risk for musculoskeletal injury. Improved technology and assistive exoskeletons have increased soldier head supported mass (HSM), and it is believed current HSM requirements exceeds functional occupational limits of what can be borne safely, contributing to cervical spine (C-spine) musculoskeletal injury risk. A subset of these functional limits has been investigated through the lens of helicopter and fixed wing aviation pilot HSM with respect to two specific scenarios, vibrational loading and high G events. These studies used both experimental and modeling approaches to develop HSM design criterion. Our overall objective for this project is to determine the chronic effects of repeated loading with HSM and to assess intervertebral stresses through biomechanics modeling to provide mitigation strategies for the effects of HSM on the modifiable musculoskeletal cervical spine characteristics military personnel. Determining the effects of HSM on musculoskeletal and neuromuscular functional joint stability characteristics will drive the design of chronic injury prevention programs, equipment design guidelines, and other potential injury countermeasures. These targeted injury prevention programs and design guidance can lead to a reduction of musculoskeletal injuries to soldiers from repetitive loading and increase operational effectiveness, lengthening military careers, and improving post-military health.

15. SUBJECT TERMS

Head supported mass; cervical spine; musculoskeletal injury

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1. **Introduction:**

Military personnel face unique challenges and occupational loads under difficult conditions that put them at significant risk for musculoskeletal injury. Improved technology and assistive exoskeletons have increased soldier head supported mass (HSM), and it is believed current HSM requirements exceeds functional occupational limits of what can be borne safely, contributing to cervical spine (C-spine) musculoskeletal injury risk. A subset of these functional limits has been investigated through the lens of helicopter and fixed wing aviation pilot HSM with respect to two specific scenarios, vibrational loading, and high G events. These studies used both experimental and modeling approaches to develop HSM design criterion. Our overall objective for this project is to determine the chronic effects of repeated loading with HSM and to assess intervertebral stresses through biomechanics modeling to provide mitigation strategies for the effects of HSM on the modifiable musculoskeletal cervical spine characteristics military personnel. Determining the effects of HSM on musculoskeletal and neuromuscular functional joint stability characteristics will drive the design of chronic injury prevention programs, equipment design guidelines, and other potential injury countermeasures. These targeted injury prevention programs and design guidance can lead to a reduction of musculoskeletal injuries to soldiers from repetitive loading and increase operational effectiveness, lengthening military careers, and improving post-military health.

2. **Keywords:**

Head supported mass; cervical spine; musculoskeletal injury

3. **Accomplishments:**

What were the major goals of the project?

<u>Specific Aims</u>

- Specific Aim 1: Survey the injury epidemiology, cervical spine strength and flexibility, and current cervical spine radiology of SFCS and physically active non-military personnel.
- Specific Aim 2: Obtain head/neck kinematics and flexion/extension electromyography data of a cohort of SFCS personnel during a number of relevant simulated military scenarios.
- Specific Aim 3: Implement the experimental kinematic and EMG data into a validated hybrid finite element – multibody head/neck model to determine intervertebral stiffnesses during these simulated military scenarios with varying levels and positions of head supported mass.

Major Task 1: University and Military IRB/HRPO Approval – In Progress

What was accomplished under these goals?

Major Activities: During this reporting (June 1, 2020 – May 31, 2021) period we completed transfer of the award to Atrium Health and created a subcontract to Duke University. We have held bi-weekly meetings with all research groups including the Partner PI (United States Army Aeromedical Research Laboratory) and Duke University. We've also have had monthly discussions with military groups at Ft. Bragg (including USASOC) where participants will be recruited and tested. Hiring of two individuals for Atrium Health was completed and doctoral students at Duke University were identified. IRB/Human Subject protection application was initiated, and the IRB was approved at Atrium Health (Advarra, Inc). This IRB will be the master IRB for the project. We have a signed DOD Institutional Agreement between US Army Aeromedical Research Laboratory and Atrium Health. Investigators at Atrium Health have completed initial pilot testing of the musculoskeletal testing that will be employed for cervical spine range of motion and strength testing. We have also completed initial motion analysis and electromyography testing (hardware assessment) at the Duke University laboratory where kinematic and electromyography data of the cervical spine will be assessed. Duke University investigators (Bass) have also conducted additional analysis of preliminary data to continue development of the model that will be utilized in Specific Aim 3 (see Appendix)

Specific Objectives: (1) Transfer of award and subcontract completed; (2) Revised budget based on (1) was completed and approved; (3) Individuals identified and hired for Atrium Health and Duke University completed; (4) Research group meetings conducted bi-weekly; (5) Master IRB was approved at Atrium Health and the agreement for Institution Relying on the IRB Services was signed between Atrium Health (Advarra, Inc) and USAARL; (6) Initial pilot testing of musculoskeletal procedures completed; (7) Preliminary and continued development of model development for Specific Aim 3.

What opportunities for training and professional development has the project provided? *If the*

Nothing to report

How were the results disseminated to communities of interest?

Nothing to report

What do you plan to do during the next reporting period to accomplish the goals?

We plan to complete and secure IRB approval during Quarter 1 of Year 2 including all necessary agreements across USAARL, Duke University, Atrium Health, and USASOC. We will finalize the standard operating procedures and complete internal reliability testing as appropriate for the appropriate test protocols including cervical spine range of motion and strength assessments. We anticipate enrollment of participants by 9/1/2021. This includes enrollment of military and non-military personnel. All goals/objectives/timeline may be affected by continued COVID-19 issues and concerns that have already impacted hiring, administrative procedures, and the ability of the team to conduct in-person activities including protocol testing, on-site training, and pilot work. We have also been notified by USASOC that there is a moratorium on new research. We are working with the appropriate individuals at USASOC to secure permission for testing of the current research project which was approved prior to the moratorium was put in place.

4. Impact:

What was the impact on the development of the principal discipline(s) of the project?

Nothing to report

What was the impact on other disciplines?

Nothing to report

What was the impact on technology transfer?

Nothing to report

What was the impact on society beyond science and technology?

Nothing to report

5. Changes/Problems:

Changes in approach and reasons for change

Nothing to report

Actual or anticipated problems or delays and actions or plans to resolve them

The requested start date was set at 10/1/2020 due to issues caused by COVID-19 that would delay hiring of personnel and completion of administrative tasks. The actual award start date was set 6/1/2020. All institutions included on the project continue to be affected by COVID-19 precautions and administrative changes including research priorities, closure of facilities, modification of research approval processes, human subject testing procedures, and the ability to recruit/test human subjects. Although some of these restrictions are being lifted, we anticipate that COVID-19 will continue to impact all aspects of research including the ability to travel as outlined in the research proposed, in-person meetings such as protocol testing/pilot testing, the ability to recruit/test human subjects, and the ability to utilize research space. We have also been notified by USASOC that there is a moratorium on new research on human subjects at our military research site. This will impact our ability to enroll and test subjects for Specific Aim 1, Major Task 1 and Specific Aim 1, Major Task 2. We will continue to focus on all activities that we can complete based on current restrictions. We will work with the scientific officer assigned to this award regarding any challenges that impact the ability to complete study milestones on schedule.

Changes that had a significant impact on expenditures

We had some initial delays in hiring personnel and have modified effort for some personnel on the award to reflect issues outlined above.

Significant changes in use or care of human subjects, vertebrate animals, biohazards, and/or select agents

Significant changes in use or care of human subjects

Nothing to report

Significant changes in use or care of vertebrate animals.

Nothing to report

Significant changes in use of biohazards and/or select agents

Nothing to report

6. **Products:**

Publications, conference papers, and presentations

Nothing to report

Journal publications.

Nothing to report

Books or other non-periodical, one-time publications.

Nothing to report

Other publications, conference papers, and presentations.

Nothing to report

Website(s) or other Internet site(s)

Nothing to report

Technologies or techniques

Nothing to report

Inventions, patent applications, and/or licenses

Nothing to report

Other Products

Nothing to report

7. Participants & Other Collaborating Organizations

What individuals have worked on the project?

Atrium Health

Name	Title	Effort	Person Months
			Cal
Timothy Sell	Principal Investigator	16.9%	2.03
Charles Ralph Reiter	Research Associate II	25.64%	3.08
Ryan Joseph Zerega	Research Associate II	23.5%	2.82

Duke University

Name	Title	Effort	Person Months
			Cal
Cameron 'Dale' Bass	Principal Investigator	18%	2.2
Jason Kait	Research Engineer	27%	3.2
Joost Op T Eynde	Graduate Student	33%	4.0
Concetta Morino	Graduate Student	15%	1.8
Christopher Eckersley	Graduate Student	100.00%	12.00

Has there been a change in the active other support of the PD/PI(s) or senior/key personnel since the last reporting period?

Nothing to report

What other organizations were involved as partners?

8. **Special Reporting Requirements**
   COLLABORATIVE AWARDS:

   QUAD CHARTS:

9. **Appendices:**

Appendix

PRMPR July 1, 2021 Report – Examining the Effects of Head Supported Mass on Cervical Spine Biomechanics and Injury Risk in Special Forces

Problem Statement

Military personnel face unique challenges and occupational loads under difficult conditions that put them at significant risk for musculoskeletal injury[1,2]. Musculoskeletal injuries (MSKIs) pose a significant threat to operational readiness; result in a significant number of lost-duty days and disability[3-7]; and are a leading cause of medical disqualification. They result in a reduction of military readiness and have long-term career and post-military sequelae[2,8] with evidence demonstrating that the problem has worsened[1,2]. Cervical spine (C-spine) injuries and neck pain are particularly problematic, with over 20% of spinal injuries due to neck pain and many individuals incapable of returning to their unit[8,9].

An urgent issue for military personnel who must perform in challenging environments with head supported mass (HSM) is the risk of chronic injury to the C-spine[10-12]. Musculoskeletal characteristics have been shown to degrade over the course of a single deployment. For example, US Army soldiers reported an average load carriage of 22.9 kilograms worn 3.7 days per week for an average of 3.5 hours per day[13]. Further work, in a different military cohort of Special Forces Combat Soldiers (SFCS), has demonstrated that years of military service negatively affects C-spine strength and range of motion (ROM)[14]. Army helicopter pilots are particularly susceptible to neck pain and the effects of chronic neck pain while wearing HSM[15]. A cohort of 27 pilots that reported a history of neck pain 12 months prior to testing showed less active C-spine ROM compared to those without recent reports of neck pain. A direct investigation into the effect that HSM has on musculoskeletal characteristics showed that individuals with the greatest exposure and longest time served demonstrated a decrease in C-spine strength and ROM. [13]. The combination of these results indicates that HSM may lead to C-spine pain which can be objectively measured using C-spine strength and ROM. However, they do not provide guidance on mitigating the effects of HSM on C-spine chronic injury. It is clear from modeling and epidemiological studies[16,17] that HSM, especially with large mass excursions from the center of gravity (CG), increase stress on the C-spine by up to an order of magnitude.

Advanced battlefield head supported technologies and novel exoskeleton suits have bolstered the operational efficiency and battlefield survivability of current warfighters, however these advances are not without their drawbacks. Improved technology has led to an increase in head soldier HSM, but there are functional occupational limits to the amount of HSM that can be borne effectively and safely[18-29]. A subset of these functional limits has been investigated through the lens of helicopter and fixed wing aviation pilot HSM in two specific scenarios, vibrational loading, and high G events (i.e. helicopter crashes and fixed wing ejections)[18-29]. These studies used both experimental (live human and cadaver testing) and modeling approaches to develop HSM design criterion such as the “Knox Box”[30], a plot the recommends helmet weight and longitudinal and vertical weight distributions to ensure safe aircraft ejection with helmet systems. Furthermore, these studies ultimately determined that the magnitude of the impact was the primary driving factor in neck loads, and for these scenarios, the HSM had limited influence on the occurrence of acute neck injuries.

However, these previous studies of acute C-spine injuries under impact do not address chronic injuries. Limited studies have investigated functional limits for vibrational loading with muscle fatigue, and design limitations were selected based on how long it took for kinematic differences to occur for varying magnitudes and locations of HSM[18,23,31]. Studies investigating the role of HSM in high G events only compared modeling and cadaveric experimental results to relevant acute neck injury criterion[19,20]. While these results have largely progressed the area of HSM safety and design, there are still issues that require attention. C-spine injuries are increasingly prevalent within military populations[32,33] outside of the group of helicopter and fixed wing pilots. These injuries occur in SFCSs and groups who also don HSM regularly[14]. Furthermore, injuries that are seen within these populations are largely not acute in nature, but instead are chronic and are associated with repetitive C-spine loading with and without HSM[14,15,18]. It is vital that these issues be addressed, as they are outside the scope of current literature, and are an important limitation on the durability, longevity, and effectiveness of vital military personnel.

To emphasize, HSM (helmet, communications, night vision technology, advanced exoskeleton devices) and its effects on chronic musculoskeletal injury, osteoarthritis (OA), and post-traumatic osteoarthritis (PTOA) is not well understood despite reports of a high incidence of neck pain in military populations[32,33]. Military activities and military service greatly increase the risk for premature disabling OA, a leading cause of disability and medical discharge[34,35]. Premature OA can arise from joint injuries (PTOA)[36] and chronic mechanical overload[37]. Our objective is to determine the chronic effects of repeated loading with HSM and to assess intervertebral stresses through biomechanics modeling to provide mitigation strategies for the effects of HSM on the modifiable musculoskeletal C-spine characteristics of SFCS. Determining the effects of HSM on musculoskeletal and neuromuscular functional joint stability characteristics will drive the design of chronic injury prevention programs, equipment design guidelines, and other potential injury countermeasures. These targeted injury prevention programs and design guidance can lead to a reduction of musculoskeletal injuries to soldiers from repetitive loading and increase operational effectiveness, lengthening military careers, and improving post-military health.

Additional Analysis of Preliminary Data

The Duke University Human Neck Model (DUHNM) consists of an osteoligamentous spine as well as 23 active muscle pairs acting along anatomically accurate paths. We modeled the seven vertebrae (C1C2-T1) as rigid bodies, while we represented the IVDs (OC2-C7T1) with non-linear six degree of freedom springs in parallel with linear dampers and positioned them according to literature. We validated this model with cadaver data in compression, tension, flexion, and extension in the midsagittal plane. We have dynamically validated this model against frontal impact data from the NBDL and CHOP.

For the models to have biofidelic representation, we determined a baseline muscle activation scheme to support the head with additional HSM using an optimization analysis. Our test matrix for static optimization analysis used masses of 0, 3, and 5 kg located at combinations of 0.0, 0.05 and 0.1 m from the head CG as seen in Figure 1. We selected these values to cover the range of potential HSM magnitudes and locations and serve as the basis for future more granular design envelopes. To begin the optimization, we set muscle activation states to a relaxed neck position. We determined this activation state in previous work [38] without HSM and show it reflected EMG data determined from literature[39]. To reduce the sample space and overall computation time, we modulated only the main six flexor and extensor muscle activation levels (Longus Capitis, Sternocleidomastoid, Splenius Capitis, Splenius Cervicis, and Trapezius) [40]. We used a Latin Hypercube design to sample the parameter space iteratively. The cost function was to minimize overall muscle fatigue as defined by Pedotti *et al.* [41], and the center of gravity of the head was constrained to ±5 mm of translation in the x direction and ±0.09 rad of rotation about the y axis. Each simulation was only under gravitational force with T1 fixed, and ran for 0.3 s. This allowed the head to reach equilibrium following the initial muscle contractions.

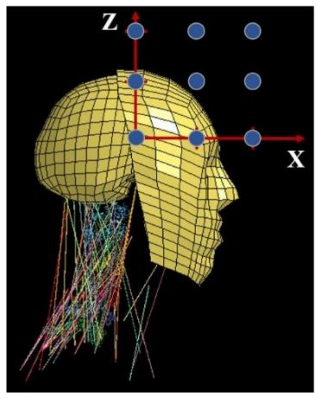
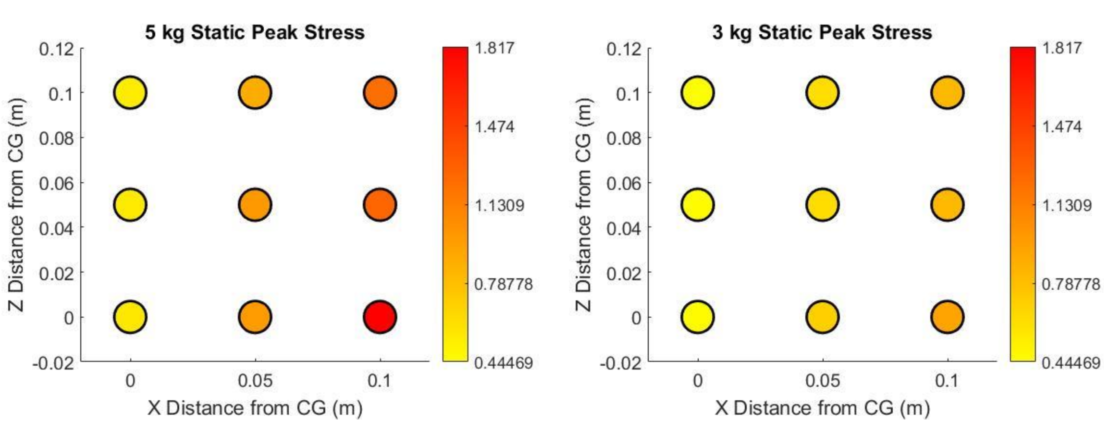


Figure 1. Illustration of the head and neck model used for simulation and the locations where additional mass was added. The axis originates at the CG of the head and the tick marks are 0.05 and 0.1 m from the CG. These values were selected to cover the range of potential head supported masses and locations.

We were interested in the increase in additional stress from HSM in static loading conditions. Once the optimal neck activation levels were determined, intervertebral forces and moments were extracted with LS-PrePost, and combined using Equation 1, a combined loading criterion derived from PMHS testing that normalizes axial loading and bending moments to values corresponding to 50% risk of cervical spine AIS ≥ 2 injuries [21]. CL is the combined loading value, F_z is the peak intervertebral axial force in Newtons, and M_y is the peak intervertebral anterior/posterior moment in Newton-meters. We calculated peak effective intervertebral stress by dividing the peak intervertebral force by an IVD cross sectional area of 0.000358 m^2. We determined this number by averaging the IVD cross sectional area of C2-C3 through C7-T1 for 22 PMHS MicroCT images.

$$CL = \frac{F_z}{5430} + \frac{M_y}{141} \quad (1)$$

Figure 2 illustrates the effective stress and combined loading results of the static optimization tests. There was no trend between mass or mass location for the resulting force, moment, and combined loading values in the osteoligamentous spine. The values with the additional HSM are higher than when no mass is applied. Values increase with increasing X distance from the CG, but a more subtle trend appears with increasing Z distance from the CG. Remaining at the X-coordinate of the CG while increasing the Z distance has little effect on the loading values. However, as the X-distance from the CG increases, increasing the Z-distance from the CG decreases the loading values. The range of stress is 1.37 MPa and the range of combined loading is 0.112. The work on these optimizations continues in preparation for HSM design criterion development. Currently we are running optimizations with increased granularity with intervals of 0.2 kg for the mass and 0.01 m for the superior and anterior directions from the CG.



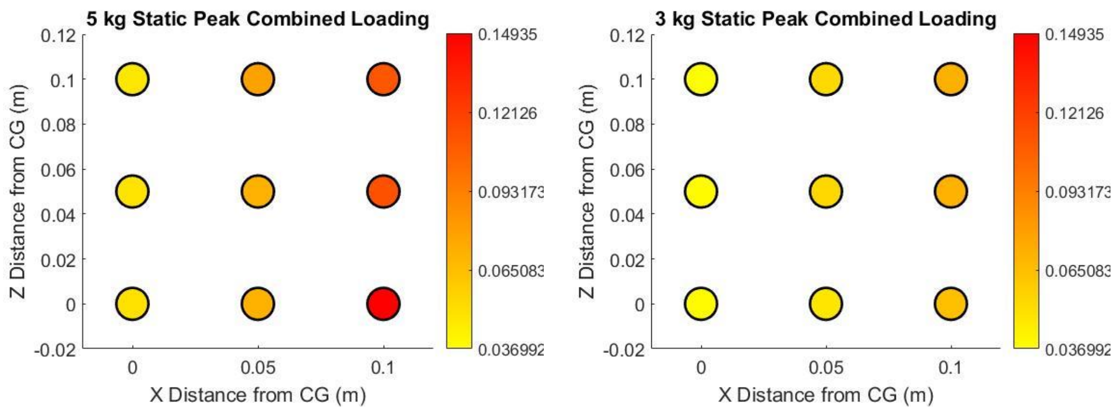


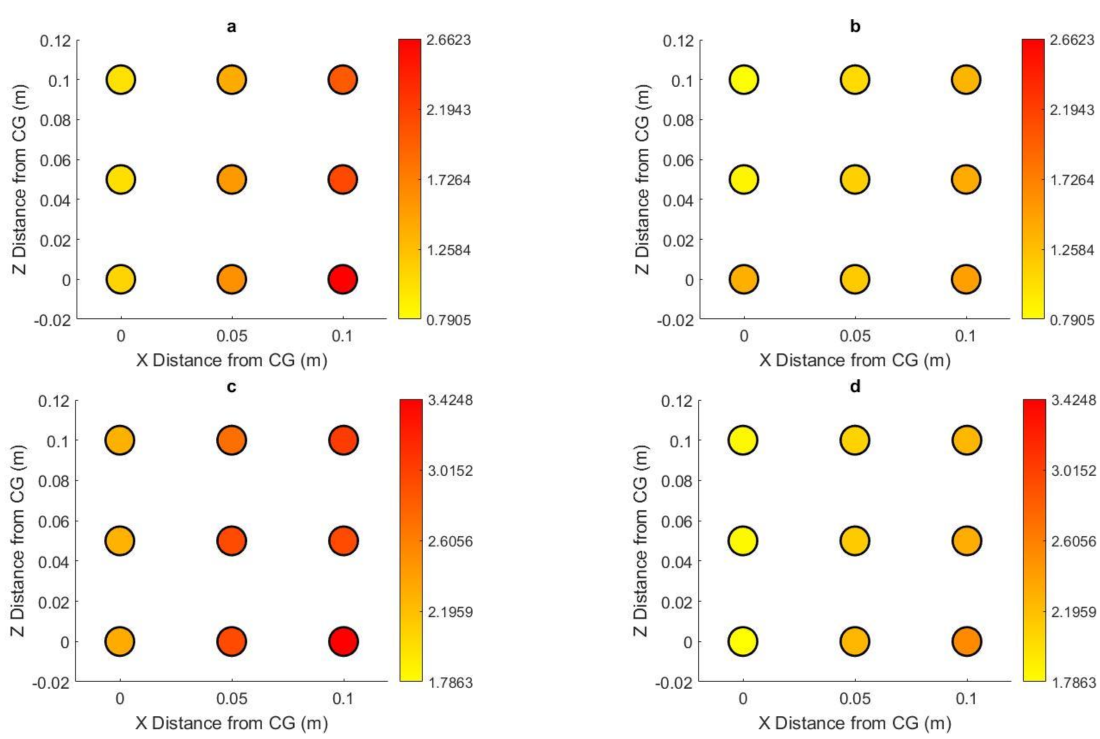
Figure 2. Heat map illustrations of the peak intervertebral effective stresses (MPa) amd peak effective combined loading values for statically supporting the head mass under gravity. Stresses generally increase with increasing X-distance of head supported mass from the head CG.

In addition to the static optimizations, we have also conducted further analysis on the dynamic simulations presented in the initial proposal. We simulated three dynamic scenarios summarized in Table 1 were simulated for each location in Figure 1 and masses of 0, 3, and 5 kg. We used muscle activations derived from the static optimizations for each scenario to provide realistic conditions before impact. We used three militarily relevant scenarios to simulate a range of kinematic exposures. First, to simulate running the first dynamic scenario provided a vertical 1 g half sine acceleration pulse over 500 ms to T1. Second, jumping from an elevated platform such as a helicopter or ledge was simulated by providing a vertical 4 g half sine acceleration pulse over 100 ms to T1. Third, a free fall parachute opening was simulated by providing a horizontal 10 g acceleration pulse over 50 ms to T1.

Table 1. Summary of Dynamic Simulation Inputs

Scenario	**Peak Accel. (g)**	**Duration (ms)**	**Direction**
Running	1	500	+Z
Jumping	4	100	+Z
Parachute Opening	10	50	-X

Figure 3 illustrates the maximum effective intervertebral stresses by location for the three scenarios and two masses. In each plot, we changed the minimum and maximum value to line up with that particular scenario in order to highlight the impact of location. There is an apparent trend for all actions that moving HSM further in the X direction from the head CG results in higher intervertebral stresses. However, for parachute openings, there appears to be an additional trend that HSM further in the Z direction decreases effective stresses. The peak effective stress values for the 5 kg mass are 2.66, 3.42, and 3.30 MPa for running, jumping, and parachute openings respectively, and for the 3 kg mass are 1.49, 2.54, and 2.66 MPa for running, jumping, and parachute openings respectively. The maximum stresses of the dynamic scenarios without HSM were 0.66, 1.06, and 0.84 MPa for the running, jumping, and parachute opening scenarios respectively.



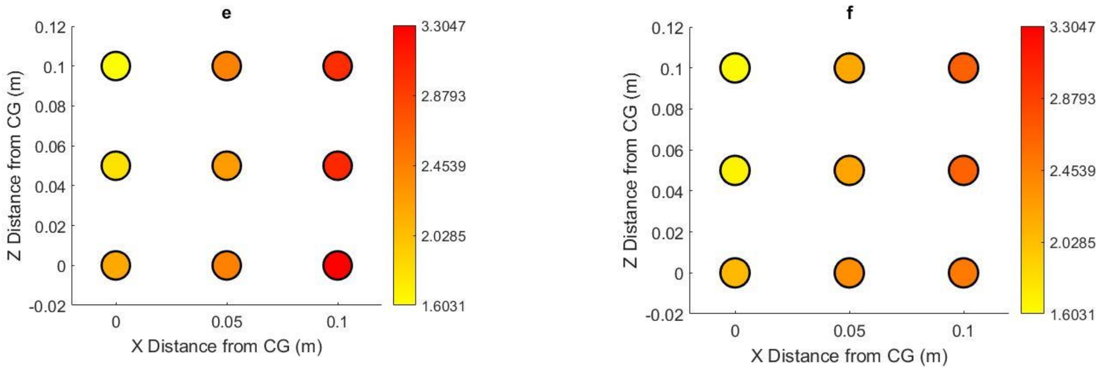


Figure 3. Heat map illustrations of the maximum intervertebral effective stresses (MPa) of the six combinations of mass and dynamic scenarios: a) 5 kg running b) 3 kg running c) 5 kg jumping d) 3 kg jumping e) 5 kg parachute opening f) 3 kg parachute opening. Stresses generally increase with increasing X-distance of HSM from the head CG.

Figure 4 illustrates combined loading, which adds peak force and moment normalized by 50% injury risk values, by location for the three scenarios and two masses. Similar to Figure 3, each plot has the minimum and maximum value optimized to highlight the impact of location. Also, in a manner similar to Figure 3, the combined loading value increases as the X-distance from the head CG increases. Furthermore, for parachute openings, there again appears to be an additional trend that moving HSM further in the Z direction decreases effective stresses. Unlike with peak effective stress, the combined loading values for the parachute opening are higher than jumping. The peak combined loading values for the 5kg mass are 0.25, 0.34, and 0.38 for running, jumping, and parachute openings respectively, and for the 3kg mass are 0.15, 0.24, and 0.32 for running, jumping, and parachute openings respectively. The maximum combined loading values for the dynamic scenarios without HSM are 0.05, 0.10, and 0.13 for the running, jumping, and parachute opening scenarios.

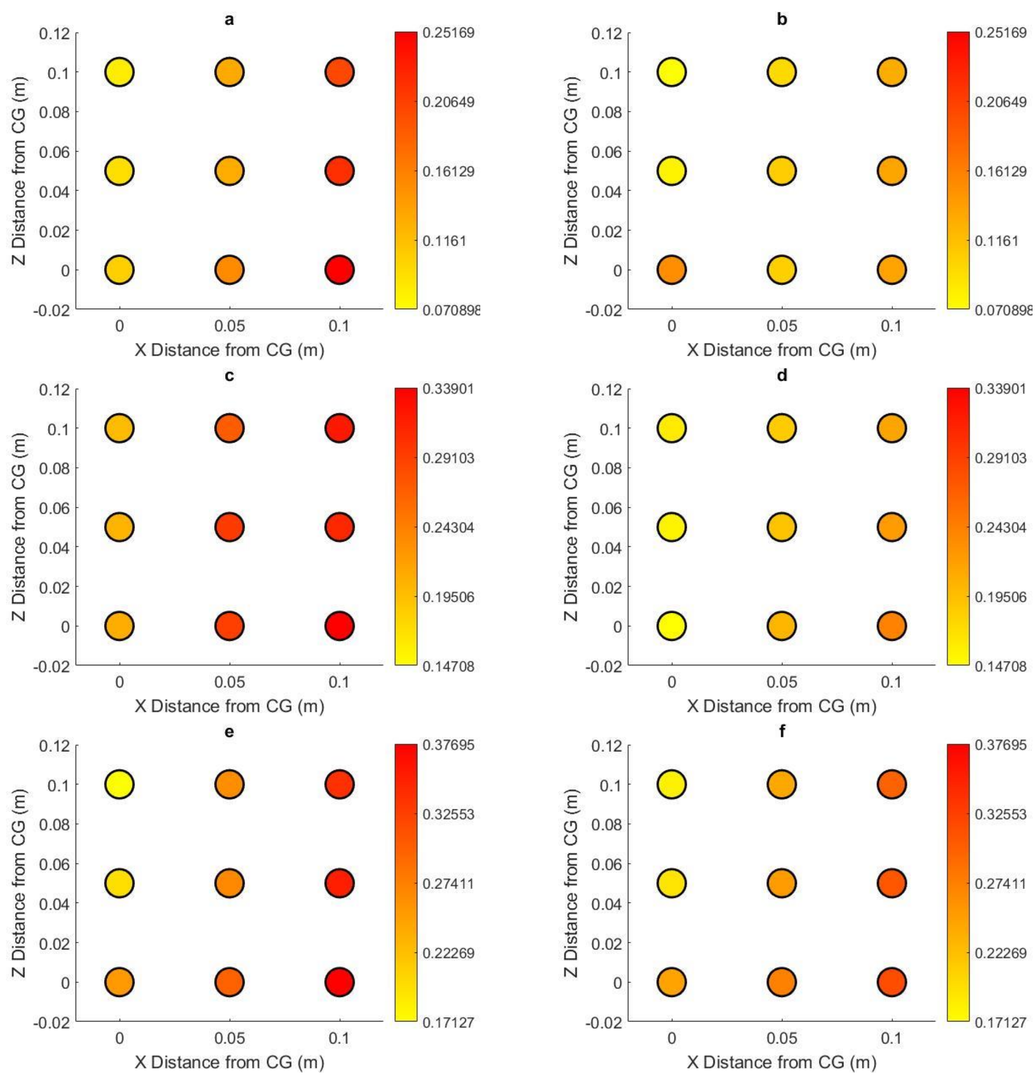


Figure 4. Heat map illustrations of the maximum intervertebral combined loading values of the six combinations of mass and dynamic scenarios: a) 5 kg running b) 3 kg running c) 5 kg jumping d) 3 kg jumping e) 5 kg parachute opening f) 3 kg parachute opening. Combined loadings generally increase with increasing X-distance of head supported mass from the head CG.

Current helmet masses range from 1.22-1.62 kg and have a CG located near the CG of the head [42,43]. Night vision systems can range from 0.75-0.9 kg and a similar mass counter weight can be added to bring the system CG back to zero [44,45]. As seen in Figure 7, these values are well contained within the sample space of this study. However, as with the static optimization work, we are currently conducting dynamic simulations to increase the granularity of the sampling space to provide increased value of the design criterion and lay the groundwork for the HSM design criterion section proposed in Aim 3.

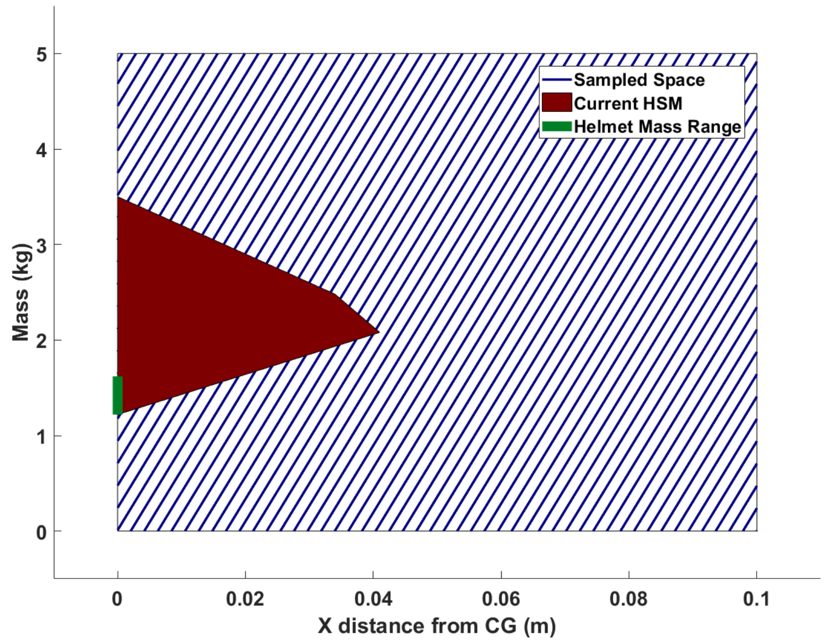


Figure 7. Parameter sample space vs. true head supported mass magnitudes and CG locations. Values are determined using LWACH helmet (Small – X-Large) and L-3 GPNVG-18 night vision system with and without a counterbalance. NVG systems assumed to have CG 0.1 m from head CG.

Comparing the static and dynamic simulations, forces generated by the cervical musculature in the static loading scenarios would be a significant portion of the forces experienced in the dynamic scenarios. The static stresses account for 30 – 70% of the peak dynamic stresses. This makes sense because the increased stresses within the neck are due to three key factors: increased load under gravity, increased muscle activation to support the load, and increased dynamic inertial loading. These results show that inertial loading results in only a fraction of the intervertebral stresses, and the static loading may be a key contributing factor to chronic injury.

In the data, we see a correlation between increases in the intervertebral stress and combined loads and increases in the mass location distance from the CG in the X-direction but not the Z-direction even for the parachute openings where the acceleration was transverse to the cervical spine. These results are illustrated in the field, where anecdotal evidence shows it is much easier for Special Operations Combat Soldiers prefer to carry night vision systems directly above their head CG when not in use. We attribute this result to two likely aspects. The first is increasing X-distance of the HSM increases the moment about the OC joint since it is perpendicular to the loading vector. This not true for increasing the Z-distance, because it is parallel to the loading vector. The second is the non-linearity of muscle loading curves. Mass added further from the CG in the X-direction, unlike the Z-direction, requires higher extensor activation to balance the increased moment. This increased activation moves the extensor muscles further up the loading curve, resulting in higher generated forces for a given displacement. Therefor higher dynamic intervertebral forces will be generated by increasing the HSM X-distance from the head CG but not Z-distance. The results also showed that increasing the Z-distance of the HSM for a given X-distance away from the CG decreased the loading values. This is because increasing the Z-distance of the HSM decreases the X-distance of the overall CG for the head/HSM system, allowing the phenomenon described above to take effect.

Current Work on Model Development

We are currently implementing numerous additions to the DUHNM in preparation for the experimental data we will use for aim three. We positioned between finite element models of a Hybrid III head and torso that will provide the proper inertial response in dynamic simulations. We are creating a custom Matlab suite that seamlessly integrates with LS-dyna that will enable us to prescribe easily the kinematic data collect from motion capture to the head and torso. We are also creating a post-processing suite that will upon model completion, automatically calculate peak intervertebral forces and stresses at each IVD as well as provide traces as to how those values change in time through the simulation. In preparation for dynamic cervical muscle EMG data, we are working to create an automated system that translates and scales experimental measurements to the baseline values determined in our static optimization trials. We are validating this process with EMG data from literature, and it will ensure EMG values are biofidelic for the anatomy of the DUHNM despite the anatomical characteristics of the test subject.

Closing Statements

As we continue working towards completion of this proposal, we maintain awareness of the work on HSM in the military and civilian biomechanics community. Recently there has been an increase in the funding interest and research output around chronic HSM injuries. We are excited about this progress, not just, because it will provide additional tools and resources for collaboration to tackle this complicated problem. Upon looking at the landscape around us, we feel our research team has a synergistic advantage allowing vertical integration of data collection with direct access to military personnel as research subjects all the way to personnel conducting end result finite element model analysis. Furthermore, the incomplete understanding of pain and causes of chronic injury makes it difficult for many research groups to contribute useful biofidelic conclusions to this problem. However, we feel we have developed a novel, creative research plan that enables us to generate valuable risk factors, injury risk curves, and design criterion that can revolutionize chronic cervical spine injury prevention, treatment, and device design. We are grateful and incredibly enthusiastic for the opportunity to continue the march down our research path in pursuit of healthier, stronger, more operationally effective military personnel.

References

[1] Jones, B.H. and Hansen, B.C. An armed forces epidemiological board evaluation of injuries in the military. *American journal of preventive medicine*, 2000. 18(3): p. 14-25

[2] Jones, B.H., Canham-Chervak, M., Canada, S., Mitchener, T.A., and Moore, S. Medical surveillance of injuries in the US military: descriptive epidemiology and recommendations for improvement. *American journal of preventive medicine*, 2010. 38(1): p. S42-S60

[3] Songer, T.J. and LaPorte, R.E. Disabilities due to injury in the military. *American journal of preventive medicine*, 2000. 18(3): p. 33-40

[4] Litow, F.K. and Krahl, P.L. Public health potential of a disability tracking system: Analysis of US navy and marine corps physical evaluation boards 2005–2006. *Military medicine*, 2007. 172(12): p. 1270-1274

[5] Lauder, T.D., Baker, S.P., Smith, G.S., and Lincoln, A.E. Sports and physical training injury hospitalizations in the army. *American Journal of Preventive Medicine*, 2000. 18(3): p. 118-128

[6] Popovich, R.M., Gardner, J.W., Potter, R., Knapik, J.J., and Jones, B.H. Effect of rest from running on overuse injuries in army basic training. *American journal of preventive medicine*, 2000. 18(3): p. 147-155

[7] Kelley, P., "Military Preventive Medicine: Mobilization and Deployment". 2003, Washington, D.C.: Walter Reed Army Medical Center Borden Institute.

[8] Cohen, S.P., Gallagher, R.M., Davis, S.A., Griffith, S.R., and Carragee, E.J. Spine-area pain in military personnel: a review of epidemiology, etiology, diagnosis, and treatment. *The Spine Journal*, 2012. 12(9): p. 833-842

[9] Cohen, S.P., Kapoor, S.G., et al. Neck pain during combat operations: an epidemiological study analyzing clinical and prognostic factors. *Spine*, 2010. 35(7): p. 758-763

[10] Abt, J.P., Sell, T.C., et al. Injury epidemiology of US Army special operations forces. *Military medicine*, 2014. 179(10): p. 1106-1112

[11] Lovalekar, M., Abt, J.P., Sell, T.C., Wood, D.E., and Lephart, S.M. Descriptive epidemiology of musculoskeletal injuries in naval special warfare sea, air, and land operators. *Military medicine*, 2016. 181(1): p. 64-69

[12] Lovalekar, M.T., Abt, J.P., et al. Descriptive epidemiology of musculoskeletal injuries in the Army 101st Airborne (Air Assault) Division. *Military medicine*, 2016. 181(8): p. 900-906

[13] Nagai, T., Abt, J.P., et al. Effects of deployment on musculoskeletal and physiological characteristics and balance. *Military medicine*, 2016. 181(9): p. 1050-1057

[14] Eagle, S.R., Nagai, T., Morgan, P., Hendershot, R., and Sell, T.C. Naval Special Warfare (NSW) crewmen demonstrate diminished cervical strength and range of motion compared to NSW students. *Work*, 2017. 58(2): p. 111-119

[15] Nagai, T., Abt, J.P., et al. Neck proprioception, strength, flexibility, and posture in pilots with and without neck pain history. *Aviat Space Envir Md*, 2014. 85(5): p. 529-535

[16] Eckersley, C.P., Cox, C., et al. A Real Pain in the Neck: Design Limits on Magnitude and Location of Head Supported Mass, in *Personal Armour Systems Symposium 2018*. 2018: Washington D.C., USA.

[17] Shivers, B.L., Madison, A.M., et al. Preliminary Characterization of Head-Supported Mass Exposure in a Simulated Dismounted Operating Environment.

[18] Butler, B.P. Long-Duration Exposure Criteria for Head-Supported Mass. 1996, ARMY COMMAND AND GENERAL STAFF COLL FORT LEAVENWORTH KS.

[19] Manoogian, S.J., Kennedy, E.A., and Duma, S.M. A Literature Review of Musculoskeletal Injuries to the Human Neck and the Effects of Head-Supported Mass Worn by Soldier. 2005, ARMY AEROMEDICAL RESEARCH LAB FORT RUCKER AL.

[20] Manoogian, S.J., Kennedy, E.A., Wilson, K.A., Duma, S.M., and Alem, N.M. Predicting neck injuries due to head-supported mass. *Aviat Space Envir Md*, 2006. 77(5): p. 509-514

[21] Bass, C., Donnellan, L., et al. A new neck injury criterion in combined vertical/frontal crashes with head supported mass. *Proceedings of 2006 international IRCOBI conference on the biomechanics of impact*, 2006.

[22] Parr, J.C. A Method To Develop Neck Injury Criteria To Aid Design And Test Of Escape Systems Incorporating Helmet Mounted Displays. 2014, AIR FORCE INSTITUTE OF TECHNOLOGY WRIGHT-PATTERSON AFB OH GRADUATE SCHOOL OF ….

[23] Alem, N.M., Meyer, M.D., and Albano, J.P. Effects of Head Supported Devices on Pilot Performance During Simulated Helicopter Rides. 1995, ARMY AEROMEDICAL RESEARCH LAB FORT RUCKER AL.

[24] McEntire, B.J. and Shanahan, D.F. Mass requirements for helicopter aircrew helmets. 1997, ARMY AEROMEDICAL RESEARCH LAB FORT RUCKER AL.

[25] Barazanji, K.W. and Alem, N.M. Effects of Head-Supported Devices on Female Aviators During Simulated Helicopter Rides. Part I: Biomechanical Response. 2000, ARMY AEROMEDICAL RESEARCH LAB FORT RUCKER AL.

[26] LaFiandra, M., Harman, E., et al. The effects of the personal armor system for ground troops (PASGT) and the advanced combat helmet (ACH) with and without PVS-14 night vision goggles (NVG) on neck biomechanics during dismounted soldier movements. 2007, ARMY RESEARCH INST OF ENVIRONMENTAL MEDICINE NATICK MA MILITARY PERFORMANCE DIV.

[27] Mobasher, A.A., Brozoski, F.T., McEntire, B., and Alem, N.M. Effects of Seat Stroke Distance on the Allowable Mass of Head Supported Devices. 1998, ARMY AEROMEDICAL RESEARCH LAB FORT RUCKER AL.

[28] Settecerri, J.J., Mckenzie, J., Privitzer, E., and Beecher, R.M. Mass Properties and Inertial Loading Effects of Head Encumbering Devices. 1986, SYSTEMS RESEARCH LABS INC DAYTON OH.
[29] Halldin, P., Hedenstierna, S., Brolin, K., and Holst, H.v. Finite element analysis of the effects of head-supported mass on neck responses. 2006, United States Army: London, England.
[30] Knox, F., Buhrman, J., Perry, C., and Kaleps, I. Interim head/neck criteria. *Consultation Report. Wright-Patterson Air Force Base, OH: Armstrong Laboratories, Escape and Impact Protection Branch*, 1991
[31] McEntire, B.J. U.S. Army Aircrew Helmets: Head Injury Mitigation Technology. *Proceedings of AGARD AMP Specialists' Meeting on "Impact Head Injury: Responses, Mechanisms, Tolerance, Treatment, and Countermeasures"*, 1996. Mescalaro, New Mexico
[32] Salmon, D.M., Harrison, M.F., and Neary, J.P. Neck pain in military helicopter aircrew and the role of exercise therapy. *Aviat Space Envir Md*, 2011. 82(10): p. 978-987
[33] van den Oord, M.H., De Loose, V., Meeuwsen, T., Sluiter, J.K., and Frings-Dresen, M.H. Neck pain in military helicopter pilots: prevalence and associated factors. *Military medicine*, 2010. 175(1): p. 55-60
[34] Brown, T.D., Johnston, R.C., Saltzman, C.L., Marsh, J.L., and Buckwalter, J.A. Posttraumatic osteoarthritis: a first estimate of incidence, prevalence, and burden of disease. *Journal of orthopaedic trauma*, 2006. 20(10): p. 739-744
[35] Rivera, C.J.D., Wenke, J.C., Buckwalter, J.A., Ficke, C.J.R., and Johnson, L.A.E. Posttraumatic osteoarthritis caused by battlefield injuries: the primary source of disability in warriors. *The Journal of the American Academy of Orthopaedic Surgeons*, 2012. 20(0 1): p. S64
[36] Keays, S.L., Newcombe, P.A., Bullock-Saxton, J.E., Bullock, M.I., and Keays, A.C. Factors involved in the development of osteoarthritis after anterior cruciate ligament surgery. *Am J Sports Med*, 2010. 38(3): p. 455-463
[37] Coyle, C.H., Henry, S.E., Haleem, A.M., O'Malley, M.J., and Chu, C.R. Serum CTXii correlates with articular cartilage degeneration after anterior cruciate ligament transection or arthrotomy followed by standardized exercise. *Sports Health*, 2012. 4(6): p. 510-517
[38] Eckersley, C.P., Nightingale, R.W., Luck, J.F., and Bass, C.R. The role of cervical muscles in mitigating concussion. *Journal of Science and Medicine in Sport*, 2019
[39] Harrison, M.F., Forde, K.A., Albert, W.J., Croll, J.C., and Neary, J.P. Posture and helmet load influences on neck muscle activation. *Aerospace medicine and human performance*, 2016. 87(1): p. 48-53
[40] Dibb, A.T., Cox, C.A., et al. Importance of muscle activations for biofidelic pediatric neck response in computational models. *Traffic injury prevention*, 2013. 14(sup1): p. S116-S127
[41] Pedotti, A., Krishnan, V., and Stark, L. Optimization of muscle-force sequencing in human locomotion. *Math Biosci*, 1978. 38(1-2): p. 57-76
[42] ArmorSource. AS-505 Product Infromation Sheet. ArmorSource.
[43] Rash, C.E., Mozo, B.T., McEntire, B.J., and Licina, J.R. Comanche Health Hazard and Performance Issues for the Helmet Integrated Display and Sighting System. 1996, U.S. Army Aeromedical Research Laboratory: Fort Rucker, Alabama.
[44] Oldham, T.W. Night vision and night vision goggles. 1990, AIR WAR COLL MAXWELL AFB AL.
[45] McLean, B., Shannon, S., McEntire, J., and Armstrong, S. Counterweights Used with ANVIS. 1996, U.S. Army Aeromedical Research Laboratory: Fort Rucker, Alabama.

Role of Neck Musculature in Head Kinematics

Two lines of biomechanical evidence suggest that neck musculature has limited effect on short term kinematics in direct contact scenarios: low short-term inertial head-to-neck coupling, and low moment of cervical muscle resistance compared to the impact moment. Previous work emphasizes loose head/neck coupling in compressive impact loading due to a low neutral zone spine stiffness (Nightingale, McElhaney et al. 1996). While pure spinal compression is not a common athletic loading scenario, Liu *et al.* finds that a 0.5 Nm moment results in 12 degrees of combined flexion and extension at the O-C1 joint (Liu, Krieger et al. 1982). Physiologically appropriate muscle response of humans works in tandem with low structural stiffness to reduce overall coupling of the head and torso through the neck. Vasavada *et al.* determined maximum voluntary cervical muscle contraction force generates a moment of 30±5 Nm in near 50th percentile adult males (Vasavada, Li et al. 2001). For comparison, an impact involving a 40 g linear head acceleration using a head mass of 5 kg and a neck length of 0.175 m results in a moment of 315 Nm. For a 40 g blunt impact, increasing cervical muscle force two standard deviations above the mean provides only a 3% increase in resistance.

The apparent contradiction between theoretical biomechanics and certain experimental results reported in literature provides the primary motivation for this study. The goal of this study was to investigate the effect of cervical muscle force on head kinematics following blunt impact in the time period at and immediately following the impact - denoted as “short term” (<50 ms). Six neck conditions were tested, and four metrics used to assess the resulting kinematics of these impacts under varying neck conditions. The principal hypothesis was that increased mass and moment of resistance provided by increased cervical muscle force generating capacity will not substantially alter short term (<50 ms) head kinematics.

Methods

Model simulation was performed using the Duke University Head Neck Model (DUHNM) developed by the Duke Injury Biomechanics Laboratory consisting of an osteoligamentous cervical spine, and 23 active muscle pairs and the HIII neck National Crash Analysis Center (NCAC). Four realistic direct contact scenarios were created ranging in energy from 25 to 176 J in energy and 3 m/s to 30 m/s in velocity. Six neck conditions were analyzed, stably relaxed, stably fully tensed, fully activated flexors, fully activated extensors, no neck, Hybrid III dummy neck (Figure 1), and 8 contact locations were assessed.

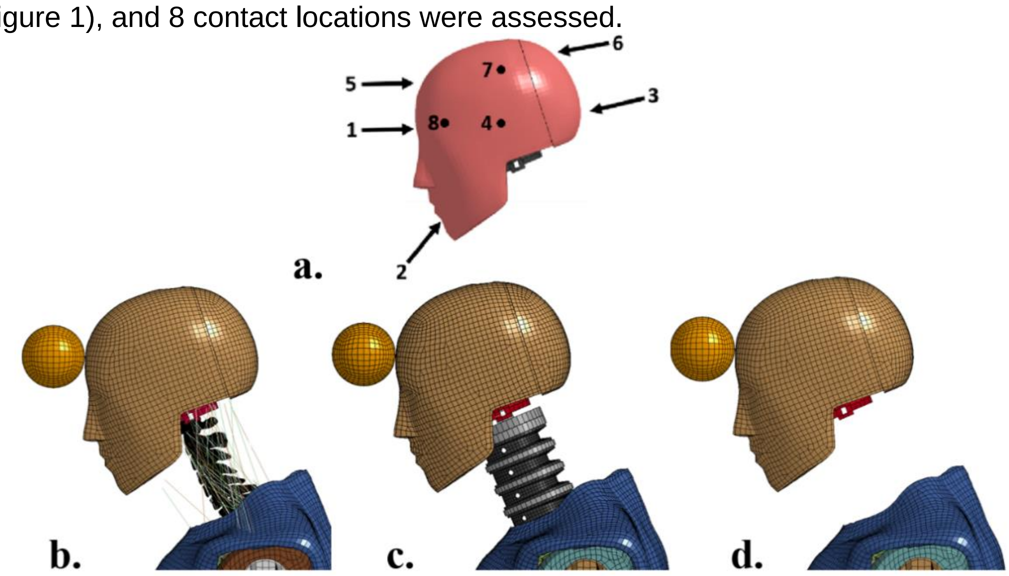


Figure 1: (a) Sagittal view of the impact locations tested. Contacts 1 – 4 are directed towards the center of gravity of the head. Contacts 1, 2, 3, 5, and 6 impact along the mid-sagittal plane while impacts 4, 7, and 8 impact perpendicular to the sagittal plane. (b) Sagittal view of the model that implements the Duke University Head Neck Model between a Hybrid III head and torso. (c) Sagittal view of the model that implements the Hybrid III neck between a Hybrid III head and torso. (d) Sagittal view of the free-head condition where the Hybrid III head remains unconstrained.

To compare the kinematic responses from each simulation, Peak Resultant Linear Acceleration, Peak Resultant Angular Acceleration, Head Injury Criterion (HIC), and Head Impact Power (HIP) were used. To determine significance between the results of the different neck conditions, the range of values for all neck conditions in each impact scenario and location were compared to critical effect sizes for each injury metric. These were determined from literature and defined as the difference between mild and severe injury thresholds.

Results

One hundred and ninety-two simulations were completed spanning the experimental test conditions. Figure 2 plots kinematic results for the various neck activation conditions and impact locations for peak resultant linear acceleration and peak resultant angular acceleration separated by impact scenario. HIC and HIP are not depicted because they follow similar patterns as the peak resultant linear acceleration results.

None of the maximum differences between the four neck activation conditions for a given impact scenario and location exceeded the critical effect size based on injury reference values. The extensor neck condition exhibited the lowest linear acceleration dominated metrics (peak resultant linear acceleration, HIC, and HIP) for 57% of impacts while the relaxed neck condition showed the highest linear acceleration dominated metrics for 80% of impacts (Figure 2). The relaxed neck condition showed lowest peak resultant angular acceleration values for 65% of impacts while the extensor neck condition showed highest peak resultant angular acceleration values for 56% of impacts. Impact location 2 shows a different trend compared to the global peak resultant angular acceleration results, as the relaxed neck condition shows the highest values while the tensed neck conditions produce lower values. The influence of impact location and scenario on linear and rotational injury metrics can be an order of magnitude larger than changes in muscle activation alone (Figure 2).

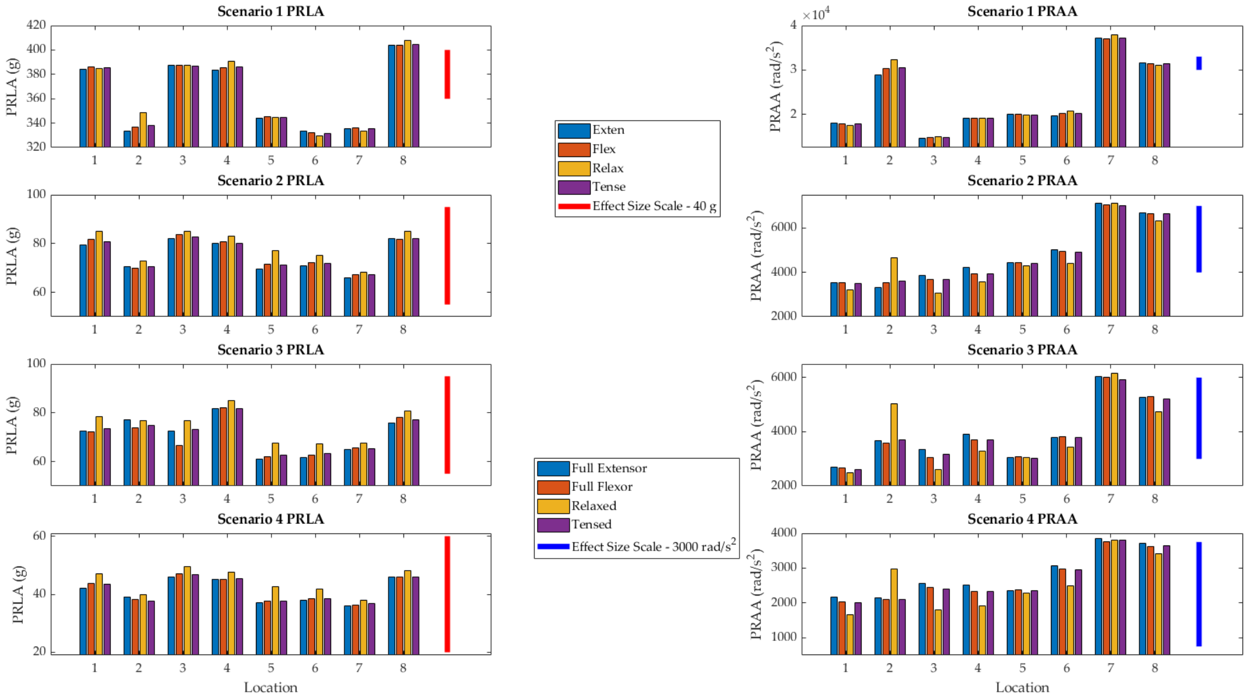


Figure 2: Peak resultant linear acceleration and peak resultant angular acceleration injury metrics for different neck activation states separated by location and impact scenario. Compared to the effect size color bar and differences in location and scenario, neck condition has little influence on metrics.

No appreciable change in four common short term injury metrics were found due to increases in cervical muscle generated force. D from the head center of gravity and magnitude have a larger influence on short term head kinematics than increasing cervical muscle force. Peak resultant angular acceleration, in contrast with linear acceleration metrics, shows tensed activation states result in higher values than relaxed activation states. This is due to small effective mass increase. Additional mass is recruited below the head-neck center of rotation, lowering the center of rotation relative to the initial free mass. These results have implications in assessing the contribution of short vs long term kinematics in contributing to both neck and head short- and long-term injuries and will be further explored in this study.

Current results using a validated biofidelic head-neck model suggest increased cervical muscle force does not influence short term (<50 ms) head kinematics in four relevant scenarios. Without a strong influence on head kinematics, neck muscle strengthening cannot strongly influence response to these scenarios' impacts. It remains to be seen how neck strengthening will influence long term kinematics and the potential for cervical injury. This will be investigated in the next project period.