Effect of Neck Musculature on Head Kinematic Response Following Blunt Impact

Christopher P. Eckersley, Roger W. Nightingale, Jason F. Luck, Cameron R. Bass

Abstract Increased neck musculature has been hypothesized to lower the risk of mild traumatic brain injury (mTBI), but this lacks experimental evidence. Here, it was hypothesized that due to low initial coupling between the head and cervical spine and the low moment of resistance supplied by the cervical musculature, increasing strength or activation of cervical musculature will have minimal effect on head kinematics. LS-Dyna was used to model impacts using the Duke University Head and Neck Model (DUHNM) coupled with the National Crash Analysis Center (NCAC) Hybrid III head and torso. Four impact types were tested with relaxed and tensed musculature conditions at eight positions on the head, totaling 64 impacts. To compare differences in mTBI risk, peak resultant linear acceleration, peak resultant angular acceleration, Head Injury Criterion, and Head Impact Power were used. To determine significance, the difference between relaxed and tensed muscle cases was compared to the difference between mild and severe impact metric values derived from literature. None of the injury metrics showed differences between the relaxed and tensed neck condition greater than the effect size.

Keywords Head Kinematics, Mild Traumatic Brain Injury, Modeling, Neck Musculature, Sports Injury Prevention

I. INTRODUCTION

Over the last decade, mild traumatic brain injury (mTBI) has made its way to the forefront of conversation in professional and recreational sports. With an estimated 1.6 to 3.8 million cases of sports-related mTBIs in the United States, many trainers, coaches, physical therapists, and doctors have been searching for ways to mitigate the risk[1]. Traditionally, cervical neck strength has been hypothesized to result in increased resistance to mTBI, which has led many of the above professionals to suggest cervical muscle strengthening exercises as a mTBI prevention strategy[2]. It is believed that by contracting their cervical musculature, athletes are able to increase the coupling between their head and neck. This would in turn increase the effective mass of the head thereby reducing head acceleration and mTBI risk[2]. However this concept lacks experimental evidence[2-4].

A majority of studies that examine this question have utilized human subjects[5-8]. However these studies are only able to expose participants to low peak accelerations and long durations. While this is consistent with inertial loading, it does not represent blunt impacts like those seen on the playing field. Furthermore, each of these studies only examined change in velocity and neck stiffness over relatively large head excursion. Because of this, these studies fail to analyze the influence of musculature on head injury metrics over the short distance and duration where mTBI is believed to occur[9].

Multiple epidemiological studies have been conducted in ice hockey and football populations examining how anticipation and cervical musculature affect head impact response. In 2010, Mihalik *et al*. instrumented youth ice hockey players with the Head Impact Telemetry (HIT) system and found that overall there was no statistically significant difference in linear acceleration, rotational acceleration, or Head Impact Telemetry severity profile (HITsp) between unanticipated and anticipated impacts[10]. In 2011, Mihalik *et al*. again investigated a youth ice hockey population, specifically looking for the effects of neck strength on head impact biomechanics and found that players with greater cervical strength did not experience lower head accelerations[3]. In 2014, Schmidt *et al*. utilized the HIT system to see how cervical muscle characteristics affect head kinematics in a football population. After obtaining cervical muscle characteristics, the study examined impacts over the course of a season and found that neither increased muscle strength nor girth mitigated head impact severity[2].

These epidemiological studies contradict traditional theory, but evidence supports why this is. First, multiple studies indicate there is poor coupling between the head and cervical spine in compression due to a low neutral

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zone cervical spine stiffness[11-13]. While many athletes do not often see compressive loading, work by Liu *et al*. indicates this decoupling translates to other loading modes as well, showing that as little as a 0.5 Nm moment can result in 12 degrees of combined flexion and extension at the O-C1 joint[14]. Furthermore, Vasavada *et al*. found that by utilizing only their cervical muscles, males were able to provide a moment of 30±5 Nm compared to the moment of a 40g impact of approximately 315 Nm[15]. This means that increasing your cervical muscle strength one standard deviation above the mean would provide only 2% added resistance compared to the impact.

The goal of this study was to examine the relationship between cervical muscle strength and head impact kinematics in three athletically relevant scenarios: an impact from a high speed object such as a pitched baseball, a moderate to severe helmeted head impact, and a mild helmeted head impact as seen in football or hockey. The study utilized an anatomically and inertially accurate, validated neck model to analyze the effects of cervical musculature on head kinematics following blunt impacts in a simulated environment. This allowed for the in depth analysis of both injurious and sub injurious simulated on field impacts; controlling for impact position, timing, and cervical muscle response while also providing accurate kinematic measurements. Impacts were assessed using four metrics: Peak Resultant Linear Acceleration (PRLA), Peak Resultant Angular Acceleration (PRAA), Head Injury Criterion (HIC), and Head Impact Power (HIP). It was hypothesized that due to the low inertial coupling between the head and cervical spine, as well as the low moment of resistance supplied by the cervical musculature, increasing strength or activation of cervical musculature will have minimal effect on blunt impact head kinematics.

II. METHODS

LS-Dyna and LS-PrePost (Livermore Software Technology Corporation, Livermore CA) were used to complete the modeling analysis in this study. In previous work the Duke Injury Biomechanics Laboratory has constructed the Duke University Head and Neck model (DUHNM). Originally the DUHNM was developed and validated for compressional loading, but has since been improved upon[16-19]. The latest iteration of the model is scalable for age and consists of 23 pairs of active muscles acting on anatomically accurate paths[18,20,21]. In the model, the eight vertebrae (C1-T1) are considered rigid bodies and are connected by seven joints consisting of massless, nonlinear 6-DOF springs in parallel with linear dampers. Joint location was selected in accordance to literature[22-24]. To model the rest of the body, the neck section of the Duke University Head and Neck model was placed between finite element models of the Hybrid III head and torso developed by the National Crash Analysis Center at The George Washington University and distributed by LSTC[25]. The occipital connections on the C1 spinal unit were rigidly connected to the baseplate of the Hybrid III head, while the T1 was rigidly connected to the lower neck bracket in the torso. Three different linear elastic spherical impactors were used in this study, properties are summarized in Table I.

TABLE I
SPECIFICATIONS OF IMPACTORS USED

Impactor	Mass (kg)	Diameter (mm)	Modulus of Elasticity(mPa)
Impactor 1	0.192	73	55.4
Impactor 2	5.507	73	9.4
Impactor 3	5.507	150	1

Impactor 1 modeled a major league baseball. Two baseball material studies were used to determine the stiffness properties of Impactor 1[26,27]. Although a baseball is a nonlinear elastic material it experiences such little deformation upon impact with the Hybrid III head, that the ball acted approximately linear. To further validate the model, impact tests using a bare Hybrid III head and neck were conducted to compare PRLA. Values were 307 G and 382 G for the experimental and modeling trials respectively. The simulation has a higher peak value; however, this is to be expected because the Hybrid III neck is more tightly coupled to the head than the DUHNM which results in a larger effective mass.

Impactors 2 and 3 both model a moderate helmeted impact. Impactor 2 represents similar magnitude and approximately 5 ms shorter duration of the impact, while Impactor 3 has a similar magnitude but approximately 5 ms longer duration of the impact. These impactors were validated using helmeted post-mortem human subject head drop data. A helmeted head equipped with a six degree of freedom sensor package located at the base of the skull posterior to the foramen magnum was dropped from 60 cm to model a moderate impact. Two trials were conducted impacting the front of the helmet. The resultant linear accelerations of the impacts were then averaged. For modeling comparison, Impactors 2 and 3 struck the forehead of an unconstrained Hybrid III head. The elasticities and impact speeds of Impactors 2 and 3 were modulated to achieve the desired relation of resultant linear accelerations to the averaged head drop data. The relationship between the resultant linear acceleration of the head drop average and Impactors 2 and 3 is seen in Figure 1.

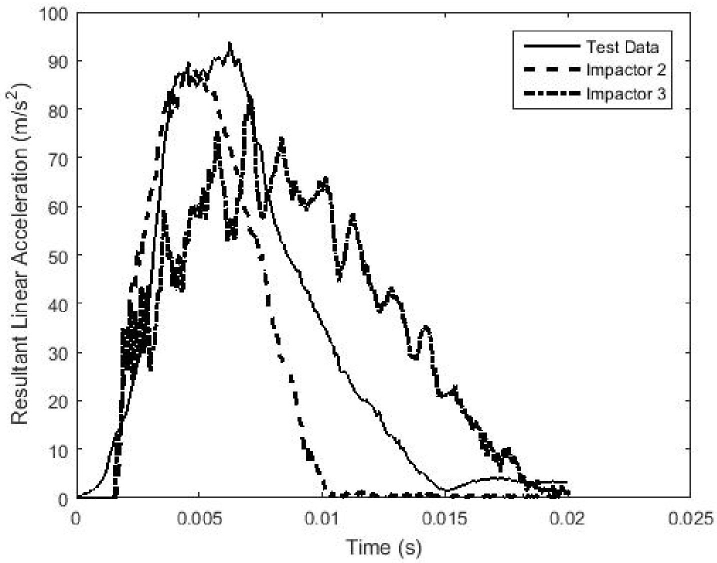

Fig. 1. Experimental validation of Impactors 2 and 3. Resultant linear accelerations of two head drop trials were averaged and plotted as Test Data. Resultant linear accelerations of free floating Hybrid III head following interaction with Impactors 2 and 3 are plotted as Impactor 2 and Impactor 3 respectively.

Four separate impact scenarios were analyzed. The first was simulating a baseball impact where Impactor 1 was given an initial velocity of 30 m/s. The second was to simulate an 80 g helmet to helmet collision where Impactor 2 was given an initial velocity of 5 m/s. The third scenario was to simulate an 80 g helmet to helmet collision with a longer duration than scenario 2, using Impactor 3 with an initial velocity of 8 m/s. The final impact scenario aimed at analyzing a lower 40 g helmet to helmet impact and was achieved using Impactor 2 with an initial velocity of 3 m/s. These impact scenarios are summarized in Table II.

TABLE II
IMPACTORS AND INITIAL VELOCITY FOR IMPACT SCENARIOS

Impact Scenario	Impactor	Initial Velocity m/s	Energy J
Scenario 1	Impactor 1	30	58
Scenario 2	Impactor 2	5	69
Scenario 3	Impactor 3	8	176
Scenario 4	Impactor 2	3	25

For each impact scenario, there were two separate neck conditions; a relaxed neck and a tensed neck. For the relaxed neck, muscles are activated to the minimum level necessary to stabilize the head[20]. With the tensed neck, muscles are activated to the maximum level while still being able to keep the head stable[20].

For each impact scenario and neck condition, eight different positions on the head were impacted shown in Figure 2. The first four are directed towards the center of gravity (CG) of the head approaching from the front of the head, 50 degrees below the head, the back of the head, and the side of the head. The last four are directed off of the CG and approach from the front of the head superior to the CG, back of the head superior to the CG, side of the head superior to the CG, and side of the head anterior to the CG.

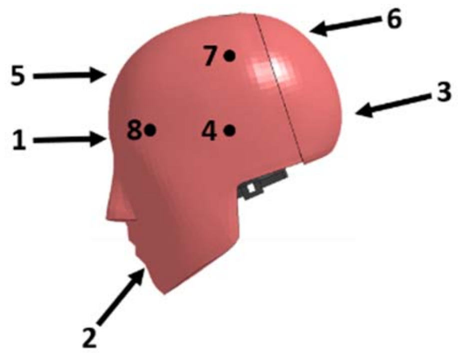

Fig. 2. Sagittal view of the impact locations tested. Impacts 1 – 4 are directed towards the CG of the head. Impacts 1, 2, 3, 5, and 6 impact along the mid-sagittal plane while impacts 4, 7, and 8 impact perpendicular to the sagittal plane.

Each simulation was set to run for 40 ms duration providing enough time for the maximum peak linear and angular accelerations to occur in each simulation. Kinematics were output at 0.0001 second intervals.

In order to compare the results of each simulation, PRLA, PRAA, HIC, and HIP were used. PRLA is a metric easily measured that has been commonly used to analyze head injury severity and describe linear accelerations of an impact[2,4,10,28]. Angular acceleration has been shown to be associated with brain strain and TBI, therefore peak resultant angular acceleration was selected as a representation of angular kinematics[29-39]. HIC was first developed in the automotive industry to assess the risk for skull fracture in severe head impacts[40-42]. HIC incorporates both the integral of linear acceleration and a time normalization and is thought to be a useful metric because it incorporates both impact duration and magnitude. The HIP, which incorporates weighted linear and angular kinematics (Equation 1) was also assessed[43,44]. HIP not only utilizes both linear and angular accelerations and velocities, but also weights each value based on head geometry.

$$HIP = 4.5\left(A_x V_x + A_y V_y + A_z V_z\right) + 0.016\omega_x \alpha_x + 0.024\omega_y \alpha_y + 0.022\omega_z \alpha_z \quad (1)$$

Where A_x, A_y, and A_z are linear accelerations in the x, y, and z directions respectively (m/s^2), V_x, V_y, and V_z are linear velocities in the x, y, and z directions respectively (m/s), $\alpha_x, \alpha_{y,}$ and α_z are angular accelerations about the x, y, and z axis respectively (rad/s^2), and $\omega_x, \omega_{y,}$ and ω_z are angular velocities about the x, y, and z axis respectively (rad/s).

Following each simulation, the kinematic data (linear and angular accelerations and velocities) were used to calculate and compare each injury criterion using MATLAB (MathWorks, USA). To estimate the significance of the muscle effect, the differences between the relaxed and tensed muscle cases were compared to an effect size for each impact metric that was derived from literature. These effect sizes are the differences between previously reported thresholds for mild and severe injury. The effect size for PRLA was found to be 40 m/s^2 using tolerance data from Eiband by subtracting the area of moderate injury from the area of severe injury at 15 ms duration[45]. Eiband's data is found in Figure 3. The effect size for PRAA was found to be 3000 rad/s^2 using data from Margulies and Thibault[39]. This was determined by finding the minimum distance between the 0.1 and 0.15 critical strains for angular accelerations at 250 rad/s of angular velocity found in Figure 4. The effect size for HIC was found to be 400 using the injury risk curves from Mertz and Prasad[46]. Effect size was defined as the difference between 50% risk on the AIS2 and AIS3 curves found in Figure 5. The effect size for HIP was found to be 24 kW based on data from Marjoux[9]. It was found using the difference in 50% risk values of moderate and severe injury curves found in Figure 6.

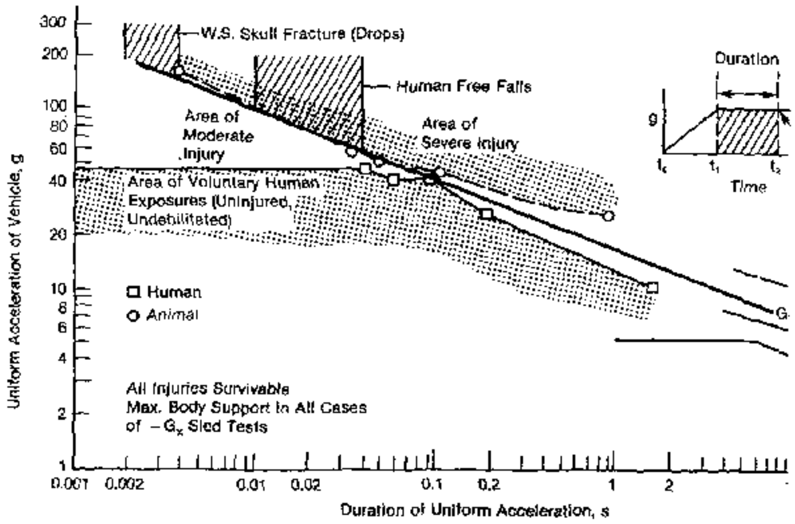

Fig. 3. Linear acceleration risk from Eiband 1959[45]. Effect size for PRLA was determined to be 40 m/s^2 by the difference in maximum and minimum of moderate injury at 15 ms.

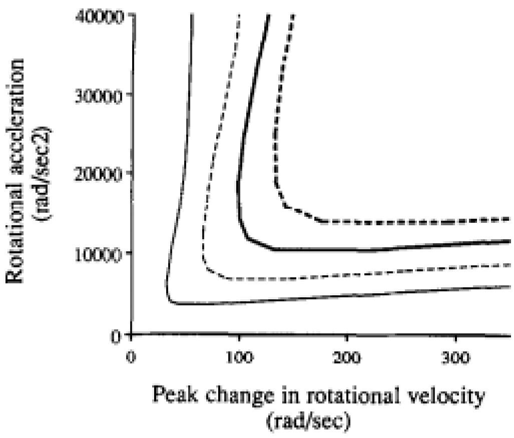

Fig. 4. Critical strain values for rotational velocity and acceleration. Strains: 0.05 (Solid Line), 0.10 (Dashed Line), 0.15 (Heavy Solid Line), 0.20 (Heavy Dashed Line) Margulies and Thibault 1992[39]. Effect size for PRAA was determined to be 2920 rad/s^2 based on the difference between critical strains of 0.1 and 0.15 at a rotational velocity of 200 rad/s.

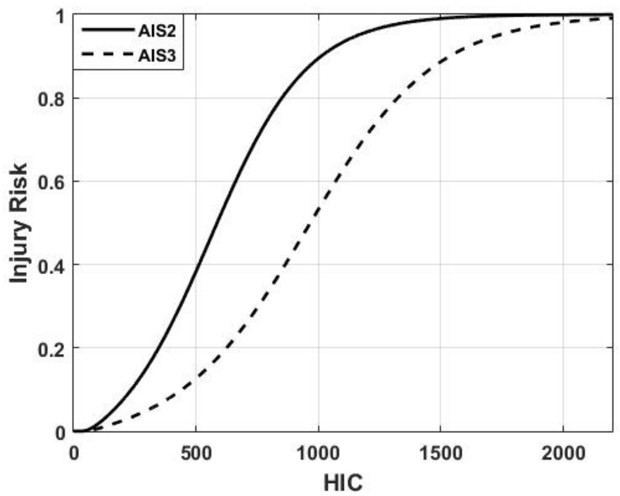

Fig. 5. AIS2 and AIS3 HIC Curves for from Mertz and Prasad 1996[46]. The effective size was found to be 381 based on the difference between the AIS curves at 50% injury risk.

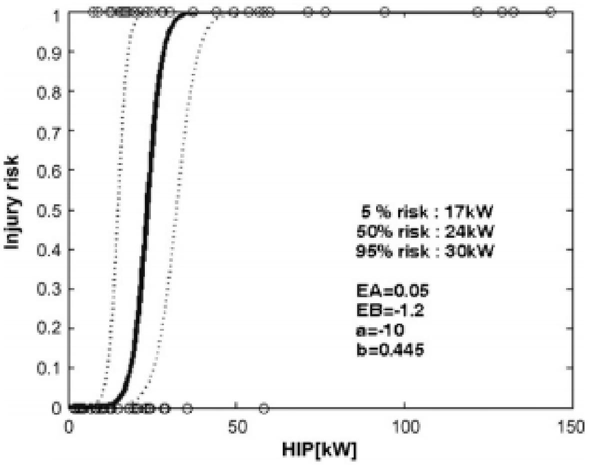

Fig. 6A. Moderate risk HIP curve published by Marjoux 2008[9]. Value used for effect size was at 50% risk.

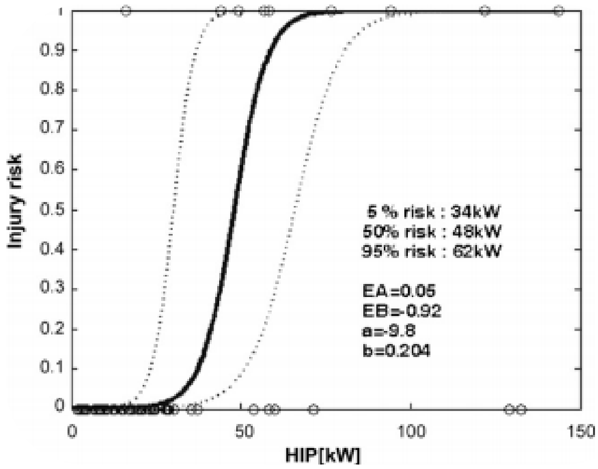

Fig. 6B. Severe risk HIP curve published by Marjoux 2008[9]. Value used for effect size was at 50% risk.

III. RESULTS

Tables 3-6 below list the injury metric values for every scenario simulatied. The higher injury metric value is highlighted with a dark grey.

For PRLA values, the relaxed metric was found to be higher than the tensed metric for 88% of the scenarios. However, none of the scenarios resulted in differences higher than the effect size. The mean ± standard deviation percent difference between the tensed and relaxed conditions was 3.5% ± 2.5%. There appears to be no consistent trend between impact position and PRLA.

For PRAA values, the relaxed condition injury metric was found to be higher in only 28% of the impacts. None of the impacts resulted in a difference between the relaxed and tensed condition that was higher than the effect size. The mean ± standard deviation percent difference between the tensed and relaxed conditions was 9.1% ± 8.1%. Position 7 consistently resulted in the highest PRAA value, while positions 1 and 3 consistently resulted in the lowest. For each impact scenario, position 2 results in a higher relaxed neck condition value.

The results of HIC values are similar to those of PRLA, 84% of the impacts resulted in the relaxed condition having a higher HIC value than the tensed condition. However, none of the impacts yielded a difference between the tensed and relaxed conditions that was higher than the effect size. The mean ± standard deviation percent difference between the tensed and relaxed conditions was 8.2% ± 4.8%. There does not appear to be a pattern between impact position and HIC values.

Similar to HIC and PRLA, HIP results display a high percentage of impacts where the relaxed condition has a higher value than the tensed condition at 78%. Also much like HIC and PRLA, none of the impact scenarios result in a difference between the relaxed and tensed conditions that is higher than the effect size of 24kW. The mean ± standard deviation percent difference between the tensed and relaxed conditions was 5.9% ± 4.3%. There appears to be no consistent trend between HIP values and impact position.

The injury metrics did not exceed the effect size for any of the impact scenarios, muscle conditions, and impact locations. When comparing impacts between scenarios 1-4 across all injury metrics, the relaxed neck condition was higher in 38% of impacts for Scenario 1, 81% of impacts for Scenario 2, 75% of impacts for Scenario 3, and 78% of impacts for Scenario 4. The mean ± standard deviation percent differences between the relaxed and tensed neck conditions were 1.9% ± 1.7%, 7.0% ± 5.2%, 6.8% ± 4.5%, and 8.7% ± 2.1% for Scenarios 1-4 respectively. The mean ± standard deviation percent difference for all impacts was 6.1% ± 5.6%.

TABLE III
RESULTS FOR PEAK RESULTANT LINEAR ACCELERATION (m/s^2) – EFFECT SIZE OF 40

Impact Scenario	Position	Relax Value	Relax Peak Time (ms)	Tense Value	Tense Peak Time (ms)	Absolute Difference	Percent Difference
1	1	382	0.45	386	0.50	3.4	0.88
1	2	322	0.60	313	0.60	8.9	2.79
1	3	390	0.50	391	0.50	0.1	0.03
1	4	382	0.50	378	0.50	4.0	1.02
1	5	343	0.50	342	0.50	1.3	0.38
1	6	337	0.50	340	0.50	2.8	0.84
1	7	333	0.55	335	0.50	2.5	0.75
1	8	418	0.50	415	0.50	2.9	0.69
1	Mean ± SD	364 ± 32	0.51 ± .04	363 ± 33	0.51 ± .03	3.2 ± 2.4	0.92 ± 0.76
2	1	87	3.05	82	3.30	4.7	5.55
2	2	75	3.30	73	3.25	1.8	2.47
2	3	87	3.00	85	3.00	1.6	1.89
2	4	83	3.50	80	3.70	2.1	2.54
2	5	78	3.70	73	2.35	4.8	6.27
2	6	78	3.15	75	3.00	3.5	4.54
2	7	70	3.55	69	3.55	1.2	1.77
2	8	90	3.25	88	3.25	2.2	2.45
2	Mean ± SD	81 ± 6	3.31 ± .23	79 ± 6	3.18 ± .38	2.7 ± 1.3	3.44 ± 1.64
3	1	80	5.60	74	5.60	5.2	6.72
3	2	74	6.05	74	6.15	0.2	0.28
3	3	78	5.90	74	5.90	4.4	5.69
3	4	85	5.75	81	5.75	3.8	4.50
3	5	69	5.35	63	5.55	6.6	9.83
3	6	69	6.65	66	6.60	3.0	4.45
3	7	68	5.65	66	5.60	2.8	4.07
3	8	83	5.70	81	5.70	2.4	2.90
3	Mean ± SD	76 ± 6	5.83 ± .36	73 ± 6	5.86 ± .34	3.5 ± 1.8	4.81 ± 2.61
4	1	47	3.65	44	3.50	3.1	6.65
4	2	39	4.40	38	4.30	0.9	2.41
4	3	50	4.40	48	4.30	1.8	3.68
4	4	47	4.05	46	4.05	1.6	3.32
4	5	42	4.65	39	3.30	3.7	8.88
4	6	43	4.00	40	4.00	3.0	7.19
4	7	39	4.40	38	3.90	1.1	2.86
4	8	50	4.00	48	3.90	2.1	4.22
4	Mean ± SD	45 ± 4	4.19 ± .30	43 ± 4	3.91 ± .33	2.2 ± 0.9	4.90 ± 2.21

TABLE IV
RESULTS FOR PEAK RESULTANT ANGULAR ACCELERATION (rad/s^2) – EFFECT SIZE 3000

Impact Scenario	Position	Relax Value	Relax Peak Time (ms)	Tense Value	Tense Peak Time (ms)	Absolute Difference	Percent Difference
1	1	17738	0.40	17971	0.40	233	1.30
1	2	32572	0.60	31020	0.60	1552	4.88
1	3	15336	0.45	15149	0.45	187	1.23
1	4	18603	0.50	18642	0.55	39	0.21
1	5	19531	0.45	19731	0.45	200	1.02
1	6	20743	0.45	20210	0.45	533	2.60
1	7	36999	0.55	36374	0.55	625	1.70
1	8	30952	0.45	31280	0.45	328	1.05
1	Mean ± SD	24059 ± 7622	0.48 ± .06	23797 ± 7340	0.49 ± .06	462 ± 449	1.75 ± 1.34
2	1	3515	2.45	3905	2.45	390	10.49
2	2	5277	2.85	4112	2.80	1165	24.80
2	3	3083	2.20	3806	2.40	723	20.99
2	4	3619	2.35	4082	2.50	463	12.01
2	5	4500	2.10	4504	2.30	4	0.08
2	6	4568	2.25	5126	2.80	558	11.51
2	7	7333	2.50	7186	2.50	147	2.03
2	8	6535	2.40	6843	2.40	308	4.60
2	Mean ± SD	4804 ± 1404	2.39 ± .21	4946 ± 1258	2.52 ± .17	469 ± 337	10.8 ± 8.2
3	1	3215	5.40	3428	0.70	213	6.41
3	2	5315	1.05	4560	1.10	755	15.28
3	3	3739	1.90	4009	1.90	270	6.97
3	4	3516	5.75	3971	5.75	455	12.15
3	5	3488	3.05	3501	3.05	13	0.38
3	6	4026	3.60	4691	3.60	664	15.25
3	7	6324	5.60	6164	5.60	160	2.56
3	8	4971	4.35	5826	4.60	855	15.84
3	Mean ± SD	4324 ± 1025	3.84 ± 1.6	4519 ± 951	3.29 ± 1.8	423 ± 287	9.36 ± 5.71
4	1	1800	2.10	2139	2.95	339	17.25
4	2	3113	3.15	2622	3.05	491	17.13
4	3	1820	3.20	2397	3.40	577	27.37
4	4	1996	3.80	2559	4.05	563	24.68
4	5	2393	3.20	2480	3.15	87	3.54
4	6	2607	3.50	3027	3.40	420	14.90
4	7	3902	3.75	3970	2.95	68	1.73
4	8	3525	3.25	3817	3.00	292	7.95
4	Mean ± SD	2645 ± 746	3.24 ± .49	2876 ± 632	3.24 ± .35	354 ± 185	14.32 ± 8.73

TABLE V
RESULTS FOR HIC – EFFECT SIZE 400

Impact Scenario	Position	Relax Value	Tense Value	Absolute Difference	Percent Difference
1	1	980	989	9	1.32
1	2	681	631	50	6.58
1	3	1031	1034	3	0.33
1	4	953	922	31	2.83
1	5	771	782	11	3.03
1	6	735	756	21	3.41
1	7	718	736	18	2.59
1	8	1238	1218	20	0.03
1	Mean ± SD	888 ± 182	884 ± 180	20 ± 14	2.52 ± 1.94
2	1	208	186	22	10.08
2	2	109	101	8	5.10
2	3	226	215	11	4.91
2	4	198	184	14	6.77
2	5	156	136	20	11.91
2	6	162	147	15	5.34
2	7	120	115	5	4.41
2	8	223	206	17	6.98
2	Mean ± SD	175 ± 42	161 ± 40	14 ± 5	6.94 ± 2.53
3	1	217	196	21	8.34
3	2	187	162	25	1.58
3	3	228	208	20	0.66
3	4	208	190	18	4.67
3	5	176	157	19	11.43
3	6	173	156	17	4.73
3	7	131	121	10	12.42
3	8	235	211	24	7.86
3	Mean ± SD	194 ± 32	175 ± 29	19 ± 4	6.46 ± 4.02
4	1	53	45	8	10.45
4	2	26	25	1	12.33
4	3	60	54	6	2.04
4	4	51	46	5	4.30
4	5	40	33	7	19.76
4	6	42	36	6	3.70
4	7	31	28	3	6.03
4	8	58	53	5	4.27
4	Mean ± SD	45 ± 12	40 ± 10	5 ± 2	7.86 ± 5.57

TABLE VI
RESULTS FOR HIP (KW) – EFFECT SIZE 24

Impact Scenario	Position	Relax Value	Relax Peak Time (ms)	Tense Value	Tense Peak Time (ms)	Absolute Difference	Effect Size
1	1	24.5	0.65	24.8	0.65	0.33	1.34
1	2	18.7	0.70	17.5	0.70	1.19	6.57
1	3	25.3	0.65	25.4	0.65	0.08	0.32
1	4	23.2	0.65	22.6	0.65	0.65	2.84
1	5	20.0	0.65	20.6	0.65	0.61	3.00
1	6	19.4	0.70	20.1	0.70	0.67	3.39
1	7	19.1	0.70	19.6	0.70	0.50	2.58
1	8	28.4	0.65	28.4	0.65	0.01	0.04
1	Mean ± SD	22.3 ± 3.3	0.67 ± .02	22.4 ± 3.4	0.67 ± .02	0.51 ± 0.35	2.51 ± 1.93
2	1	9.6	4.80	8.7	4.25	0.92	10.05
2	2	5.9	4.45	5.6	5.05	0.29	5.04
2	3	10.4	5.00	9.9	4.90	0.50	4.93
2	4	9.0	5.00	8.4	4.55	0.59	6.78
2	5	7.7	4.85	6.9	4.85	0.87	11.92
2	6	8.0	4.45	7.5	4.50	0.41	5.29
2	7	6.2	5.20	6.0	5.05	0.27	4.43
2	8	9.8	4.60	9.2	4.40	0.66	6.95
2	Mean ± SD	8.3 ± 1.6	4.79 ± .26	7.8 ± 1.4	4.69 ± .29	0.56 ± 0.23	6.92 ± 2.53
3	1	13.5	8.60	12.5	9.75	1.08	8.31
3	2	10.5	8.35	10.7	10.70	0.17	1.60
3	3	12.8	8.70	12.7	10.40	0.08	0.63
3	4	12.9	8.80	12.3	8.70	0.59	4.68
3	5	11.3	9.90	10.1	10.20	1.23	11.50
3	6	11.2	8.10	10.7	8.35	0.52	4.75
3	7	10.5	9.90	9.3	9.90	1.23	12.42
3	8	15.2	8.70	14.0	8.75	1.15	7.88
3	Mean ± SD	12.2 ± 1.5	8.88 ± .62	11.5 ± 1.5	9.59 ± .82	0.75 ± 0.45	6.47 ± 4.03
4	1	3.4	5.65	3.1	5.40	0.34	10.46
4	2	2.4	5.95	2.1	5.75	0.28	12.44
4	3	3.9	5.80	3.8	6.00	0.08	2.08
4	4	3.5	5.55	3.3	5.55	0.15	4.41
4	5	3.0	6.10	2.4	5.65	0.54	20.00
4	6	3.2	5.95	3.0	5.65	0.11	3.55
4	7	2.9	5.70	2.7	5.35	0.17	6.07
4	8	4.3	5.65	4.1	5.10	0.18	4.29
4	Mean ± SD	3.3 ± 0.6	5.79 ± .18	3.1 ± 0.6	5.56 ± .26	0.23 ± 0.14	7.91 ± 5.65

IV. DISCUSSION

The goal of this study was to analyze how cervical musculature effects head kinematics in four impact scenarios. A current theory posits that increasing muscle strength or activating muscles before impact will increase the effective mass of the head thereby reducing head accelerations. It was hypothesized that this theory is not true. Instead, due to the low initial coupling between the head and cervical spine, as well as the low moment of resistance supplied by the cervical musculature, increasing cervical muscle strength or anticipating an impact will have minimal effect on blunt impact head kinematics.

Four common injury metrics used for impact analysis showed that for every single impact, the difference between the relaxed and tensed condition did not exceed the effect size. Furthermore, the mean and standard deviation percent difference of all impacts for all four injury metrics was 6.1% ± 5.6%. These results are consistent with the work by Schmidt *et al*. that found stronger head and neck muscles do not reduce head accelerations[2]. Also, these results support the decoupling between the head and spine seen in compression is applicable to other loading scenarios. However, for the three injury metrics dominated by linear acceleration (PRLA, HIC, and HIP), the relaxed neck condition results in slightly higher injury metrics for most impacts. There are two explanations for this. First, the modest resistance that tensed cervical muscles can provide decreases linear accelerations slightly, but not enough for an effective difference. Second, tensing muscles or increasing strength does increase the effective mass but this increased mass is nearly negligible compared to the head's mass.

For PRAA, contrary to linear injury metrics, a majority of impacts displayed slightly higher metrics for the tensed condition than the relaxed condition. This occurs because the increased effective mass is nearly negligible compared to the head's mass, but lowers the center of rotation (COR) for the head/neck system. This lowered COR results in a larger moment arm increasing the PRAA metric in all positions except position 2 where the relaxed neck condition had a consistently higher injury metric. Position 2, unlike the others, impacts the head below the relaxed head and neck's COR; therefore when muscle activation lowers the COR, it decreases the impact moment arm and reduces the PRAA metric.

By comparing the results of linear and angular acceleration metrics, it is seen that increasing neck muscle strength or activation affects the two metrics differently. Distinction between the results of linear and angular acceleration dominated metrics is vital because injury prevention strategies vary based on which metric holds primary importance. Research has provided strong evidence that angular acceleration is the primary mechanism of mTBI[29-38]. Based on this evidence, while increasing muscle strength or muscle activation does not generate an effective difference in head kinematics, it may still have a negative effect on athlete safety.

Results for PRAA depict how the moment arm about the COR affects head kinematics. Impact positions further from the COR such as position 7 resulted in higher PRAA values than those closer such as positions 1 and 3. Distance from the COR may not be the only factor in the relationship between impact position and head kinematics. Evidence has shown that cervical musculature strength also varies with impact direction[15,47]. However, much like the increase in muscle tension due to increased muscle activation or strength, changes in muscle strength with respect to impact direction appear to have little effect. Positions 1, 3, and 4 impact the head from different directions, but all are directed towards the head's center of gravity. The difference between these three positions does not exceed the effect size for any impact scenario, injury metric, or neck condition. As this was not the focus of this study, further work needs to be conducted to analyze this relationship because it may have strong implications for mTBI. It has been shown that brain strain is sensitive to impact direction, therefore if the differences in cervical musculature result in similar sensitivities to impact direction, it could magnify the risk of mTBI[48].

For impact scenario 1, the tensed muscle case resulted in higher injury metrics for 62% of impacts. However, the mean ± standard deviation percent difference between the two neck conditions was only 1.9% ± 1.7%. Based on this evidence, it is believed that similar to the other scenarios the impact force to the system is so large compared to the resistance cervical musculature can provide, therefore, cervical musculature is not playing a significant role in the difference in head kinematics.

Only two muscle states were modelled in this study – relaxed and tensed. A state of maximum resistance to the direction of the applied impact was not modeled, however this condition is not practical. In an athletic scenario the maximum resistance that can be supplied prior to an impact is the tensed condition tested, otherwise the athlete's head would not be stable. This means that any additional resistance must be applied following the initiation of impact. The acceleration from the impact has concluded 10 – 20 ms following impact initiation. However, it has been shown that the onset of voluntary neck muscle contraction occurs 65 – 90 ms following the event[49,50]. Therefore, in any impact scenario, additional resistance to the impact can only be applied well after the acceleration event has concluded. Because of this, it is believed that the low coupling between the head and neck prevents neck musculature from influencing head accelerations in any impact scenario.

There are a number of limitations present in this study. First, the impactors used were modeled with a linear elastic material while in general, head accelerations on the playing field will be caused by nonlinear impactors. However, based on the validation conducted for the impactors, it is hypothesized that using nonlinear impactors would result in outcomes similar to this study. Furthermore while Impactors 2 and 3 were able to successfully mimic a moderate helmeted head drop impact, the head drop testing has limitations when compared to an on field impact. The impact surface of the testing is a rigid plate coated with rubber instead of another helmeted head, which shortens the impact duration. Also, the location of the accelerometers during testing is near the head's CG but not exact. The point of measured acceleration in the modeled Hybrid III is located on the head's CG. Since this was a modeling study, there are limitations in the statistical rigor of analysis in that instead of comparing the neck conditions using statistical variation, an effect size was used.

V. Conclusions

The hypothesis that muscle activation does not significantly affect head injury risk was supported based on the results described. While increasing cervical musculature activation or strength does increase the effective mass of the head-neck system, the weak coupling between the head an neck renders this increase in effective mass negligible. Therefore, increasing cervical musculature activation or strength does not result in a kinematic difference larger than the effect size for high speed object impacts, moderate to severe helmeted head impacts, or mild helmeted head impacts. This contradicts a current theory that increasing neck strength or muscle activation will lower head kinematics because the weak coupling between the head and neck results in an effective mass increase that is much lower than the theory anticipates. One important consideration of this increased effective mass is that it also lowers the center of rotation of the system. This may lead to larger peak resultant angular accelerations during the tensed neck condition, although they are not larger than the effect size. It was also determined that differences in cervical resistance based on the direction of impact do not generate kinematic differences larger than the effect size.

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VII. References

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