Original research

The role of cervical muscles in mitigating concussion

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ABSTRACT

Objectives: Increased neck strength has been hypothesized to lower sports related concussion risk, but lacks experimental evidence. The goal is to investigate the role cervical muscle strength plays in blunt impact head kinematics and the biofidelity of common experimental neck conditions. We hypothesize head kinematics do not vary with neck activation due to low short term human head-to-neck coupling; because of the lack of coupling, free-head experimental conditions have higher biofidelity than Hybrid III necks.

Methods: Impacts were modeled using the Duke University Head and Neck Model. Four impact types were simulated with six neck conditions at eight impact positions. Peak resultant linear acceleration, peak resultant angular acceleration, Head Injury Criterion, and Head Impact Power compared concussion risk. To determine significance, maximum metric difference between activation states were compared to critical effect sizes (literature derived differences between mild and severe impact metrics).

Results: Maximum differences between activation conditions did not exceed critical effect sizes. Kinematic differences from impact location and strength can be ten times cervical muscle activation differences. Hybrid III and free-head linear acceleration metrics were $6 \pm 1.0\%$ lower and $12 \pm 1.5\%$ higher than relaxed condition respectively. Hybrid III and free-head angular acceleration metrics were $12 \pm 4.0\%$ higher and $2 \pm 2.7\%$ lower than relaxed condition respectively.

Conclusions: Results from a validated neck model suggest increased cervical muscle force does not influence short term (<50 ms) head kinematics in four athletically relevant scenarios. Impact location and magnitude influence head kinematics more than cervical muscle state. Biofidelic limitations of both Hybrid III and free-head experimental conditions must be considered.

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Practical implications

- Neck strengthening programs and exercises will do little to mitigate concussion risk.
- Implications of choices made in experimental head and neck methodologies need to be considered before applying results to athletic programs.
- Time should be spent further developing programs, rules, and methods to limit overall head exposure in sports instead of neck strengthening programs.

1. Introduction

Professional and recreational sporting communities are increasingly concerned about the risk of athletes experiencing a sports related concussion. Trainers, physical therapists, doctors, and coaches are seeking effective mitigation strategies with the hope of reducing the estimated 1.6–3.8 million cases of sports related concussion in the United States.[1] Stronger neck musculature is widely hypothesized as an effective means to reducing the risk of sports related concussion, leading to neck muscle strengthening exercises being a suggested sports related concussion prevention strategy.[2–4] This hypothesis is based on the idea that neck muscle contraction increases head-to-neck coupling. This increased coupling leads to higher effective mass, especially for anticipated impacts with pretensed musculature, thereby reducing head acceleration and sports related concussion risk.[2]

A number of studies have investigated this hypothesis.[4,7,8] In 2014, Collins et al. used a hand held tension scale to obtain neck strength measurements of high school athletes who were monitored for a concussion. The study found those who received a concussion had significantly weaker necks than those who did not.[4] A number of studies took a series of neck strength and anthropometric measurements, then subjected participants to low impulse loads while measuring head and neck kinematics.[10–12] In each study except Mansell et al. they determined increases in neck

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muscle force led to decreased head kinematics. Mathematical and computational modeling approaches have also been undertaken to examine the neck strength hypothesis.[3,5] Viano et al. utilized a mathematical model and represented the neck as a linear spring with a stiffness determined by neck strength. Jin et al. used a finite element approach with the head and neck from the Global Human Body Modeling Consortium (GHBMC). Much like the human participant studies, the two modeling studies also found that neck muscle force influence head kinematics.

Contradicting evidence to this hypothesis has been presented in epidemiological studies analyzing how cervical musculature and impact anticipation affect head kinematic response in football and ice hockey populations. Mihalik et al. in 2010 instrumented youth ice hockey players with the Head Impact Telemetry (HIT) system, and found no statistically significant evidence that impact anticipation affects linear acceleration, rotational acceleration, or Head Impact Telemetry severity profile (HITsp).[13] For a follow-up study in 2011, Mihalik et al. looked specifically at the effect of neck strength on head impact biomechanics in youth ice hockey players and found athletes with stronger necks did not experience lower head accelerations.[14] In 2014, Schmidt et al. investigated how cervical muscle characteristics influence head kinematics in a football population using the HIT system. After collecting data throughout an entire season, the study found neither increased neck muscle girth nor neck muscle strength provided a statistically significant decrease in head kinematics.[2]

Two lines of biomechanical evidence from previous studies support this contradicting evidence: low short-term inertial head-to-neck coupling, and low moment of cervical muscle resistance compared to the impact moment. Literature indicates loose head/neck coupling in compressive impact loading due to a low neutral zone spine stiffness.[15] While pure spinal compression is not a common athletic loading scenario, Liu et al. finds that a 0.5 Nm moment results in 12° of combined flexion and extension at the O-C1 joint.[16] In tandem with low coupling is the physiologically likely muscle response of living humans. Vasavada et al. determined maximum voluntary cervical muscle contraction force of near 50th percentile adult males generates a moment of 30 ± 5 Nm.[17] 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. Compared to 315 Nm, 30 Nm is only 10%, and increasing cervical muscle force two standard deviations above the mean provides only a 3% increase in resistance.

This contradiction in the available evidence and theoretical biomechanics provides the primary motivation for this study. A second motivation stems from human participants or cadaveric specimens not always being available to study head and neck kinematics and injury risk. In lieu of these participants or surrogates, investigators use a mechanical surrogate – either a Hybrid III (HIII) neck attached to a dummy head or a freely dropped dummy head form (free-head) – when conducting drop or impact tests.[5,18] A potential limitation of this approach is that the HIII neck is overly stiff resulting in a highly coupled head-to-neck complex when compared with humans, particularly in loading scenarios other than those similar to a frontal impact vehicle crash.[19]

The goal of this study was twofold: first, investigate the effect of cervical muscle force on head kinematics following blunt impact in the time period at and just after the impact – denoted as "short term" (<50 ms) – and second, determine the biofidelity of blunt impact head kinematics of a HIII neck and head surrogate, and a free-head surrogate when compared to a validated computational model of the human head and neck. The study examines three athletic scenarios: an impact from a high speed object such as a baseball in flight, an 80 g helmeted head impact, and a 40 g helmeted head impact. The latter are representative of helmeted sports scenarios. For this study, an anatomically and inertially

Table 1
Specifications of impactors used.

Impactor	Mass (kg)	Diameter (mm)	Young's Modulus (MPa)
Impactor 1	0.192	73	55.4
Impactor 2	5.507	73	9.4
Impactor 3	5.507	150	1

accurate and validated 50th percentile human head-neck computational model was used to simulate the response of an adult human head and neck to blunt impacts. The model output provides detailed kinematic analysis of simulated on-field impacts; controlling for impact position, timing, and cervical muscle response.[20,21] Six neck conditions were tested and four metrics were used to assess the resulting kinematics of these impacts under varying neck conditions. There were two hypotheses evaluated in this study. First, increased mass and moment of resistance provided by increased cervical muscle force generating capacity (whether that is through increased strength or activation) will not be large enough to alter short term (<50 ms) head kinematics for any impact scenario tested based on low, short term, initial human head-to-neck coupling. Second, the testing condition of a free-head (no neck) will be more biofidelic compared to a relaxed neck condition than using a stiff HIII neck.

2. Methods

Model simulation was performed with LS-DYNA (V810 double precision, Livermore Software Technology Corporation, Livermore CA) using two previously validated neck models. The first was the neck from 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.[20–22] The second model represents the HIII neck and was developed by the National Crash Analysis Center (NCAC) at The George Washington University and is distributed by LSTC.[23] Both necks were positioned between finite element models of the HIII head and torso developed by NCAC and distributed by LSTC.[23] All contacts and constraints present in the LSTC distributed HIII model remained intact for the HIII neck simulations.

Blunt impacts to the two head-neck models were simulated using three separate linear elastic, spherical impactors depending on the specific impact condition. Their properties are summarized in Table 1. Impactor 1 is modeled on major league baseball stiffness.[24] Both Impactors 2 and 3 were designed to represent helmeted impacts of varying severity and were developed based on human cadaver head experimental drop data conducted in the Duke University Injury Biomechanics Lab. The cadaveric head was instrumented with a six degree of freedom sensor package mounted to the skull posterior to the foramen magnum and fitted with a Riddell Speed football helmet. It was dropped twice from 60 cm impacting the front of the helmet to simulate a moderate impact. Simulated impacts using LS-DYNA were completed for Impactors 2 and 3 where the impactor struck an unconstrained HIII head (NCAC) with the impact velocity determined for the experimental head drops. The elasticities of the impactors were modulated within the computational modeling environment to achieve the desired resultant linear acceleration trace compared to the averaged experimental head drop data. Impactors 2 and 3 have similar peak acceleration, but Impactor 2 has a 5 ms shorter duration than a representative moderate helmeted impact, while Impactor 3 has a 10 ms longer duration than Impactor 2.

Using the three impactors, four impact scenarios were created, the impactors used and initial velocities are summarized in Table 2. The first simulates a baseball impacting a bare head while the second and third simulate 80 g helmet to helmet collisions with shorter

Table 2
Impactors and initial velocities 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

and longer durations respectively. The fourth scenario modeled a 40 g helmet to helmet impact. The values of 40 g and 80 g were selected to be within the ranges presented as mild and moderate impacts by Schmidt et al.[2]

For each impact scenario, six neck conditions were analyzed. The first two are relaxed and tensed neck scenarios.[20] In the relaxed condition, muscles were activated the minimum amount necessary to maintain head stability prior to impact.[20] For the tensed condition, muscles were activated to the maximum amount possible while maintaining head stability prior to impact.[20] The next two conditions model states of fully activating flexor muscles, and fully activating extensor muscles. These conditions represent bounds of muscle activation that are likely unachievable in a living human being while being theoretically possible. The final two conditions model common experimental setups of a free-head and a HIII neck. In the free-head condition, impactors contact an unconstrained HIII head. In the HIII neck condition, the HIII model is placed in-between the head and torso instead of the DUHNM. The three neck types (DUHNM, HIII, and free-head) are illustrated in Fig. 1b–d.

For each of the four impact scenarios and six neck conditions, eight different locations on the head were impacted as illustrated in Fig. 1a. Kinematic data were output every 0.05 milliseconds and simulations were run for 40 ms to ensure linear and angular accelerations reach their peaks and return to zero. The data were filtered at 1650 Hz using a 20th order Butterworth filter that provided a sharp cutoff frequency while remaining stable.

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. Peak resultant linear acceleration is a historically common metric to analyze head injury severity and describe linear acceleration exposure.[2,5,6] Brain strain and sports related concussion

have been shown to be caused by angular accelerations; to represent angular kinematics peak resultant angular acceleration was selected.[25] HIC is believed to be a useful metric because it incorporates both impact duration and linear acceleration magnitude. HIP includes weighted linear and angular kinematics based on the geometry of the head.[26] 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.[27–30]
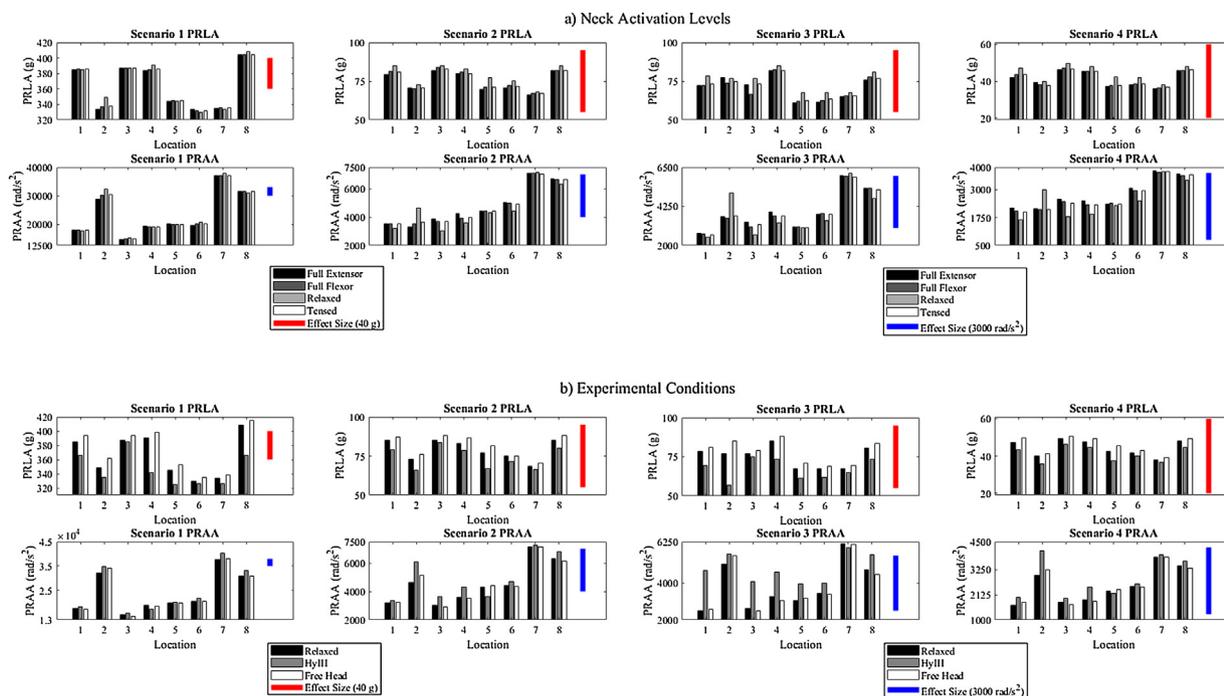
3. Results

A total of 192 simulations were completed spanning the experimental test conditions. Fig. 2a 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. 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 (Fig. 2a). 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 (Fig. 2a).

Impact location variations for peak resultant linear acceleration values (Fig. 2a) displayed no consistent pattern. For peak resultant angular acceleration, location 7 consistently produced the highest
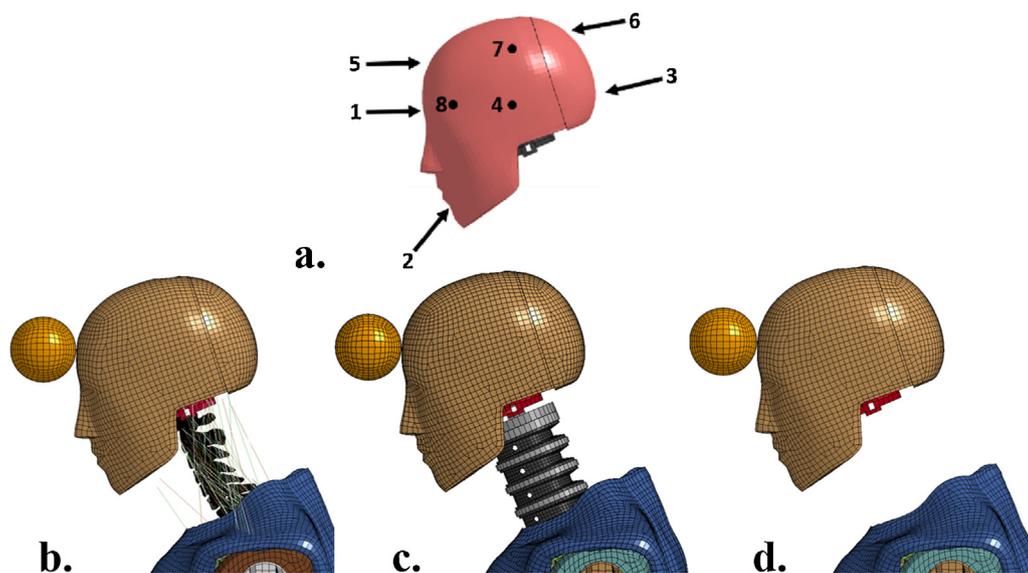
Fig. 1. (a) Sagittal view of the impact locations tested. Impacts 1–4 are directed towards the center of gravity of the head. Impacts 1, 2, 3, 5, and 6 impact along the mid-sagittal plane while impacts 4, 7, and 8 impact perpendicular to the sagittal plane. (b) Sagittal view of the model that implements 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.

Fig. 2. (a) 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. (b) Peak resultant linear acceleration and peak resultant angular acceleration injury metrics for different neck experimental testing conditions separated by location and impact scenario. The influence of experimental neck condition on head kinematics depends on the metric used.

values while locations 1 and 3 consistently had the lowest values as confirmed by the averages of all impacts for a given location.

Neck activation conditions for peak resultant linear acceleration and peak resultant angular acceleration were plotted by impact scenario to determine the biofidelity of simulated experimental blunt impact head kinematics (Fig. 2b). The relaxed neck condition served as representative of a biofidelic 50th percentile male human head–neck response. As in Fig. 2a, HIC and HIP are not included because they trended similar to peak resultant linear acceleration. The HIII neck was $6 \pm 1\%$ (mean $\pm$ SE) lower than the relaxed condition for linear acceleration dominated metrics while the free-head condition were $12 \pm 2\%$ higher. For peak resultant angular acceleration, the HIII neck was $12 \pm 4\%$ higher than the relaxed neck condition while the free-head was $2 \pm 3\%$ lower.

4. Discussion

We found no appreciable improvement in four common sports related concussion short term injury metrics due to increases in cervical muscle generated force. Impact distance from the head center of gravity and magnitude have a larger influence on short term head kinematics than increasing cervical muscle force. This is consistent with Schmidt et al. where stronger neck muscles in a college football population did not reduce head accelerations.[2]

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. Since a majority of impacts are located above the center of rotation, lowering the center of rotation increases the moment arm of the impacts, increasing peak resultant angular acceleration values. This is supported by the relaxed muscle condition having consistently higher peak resultant angular acceleration values than the tensed muscle conditions at location 2, the

one location below the center of rotation. This result shows cervical musculature influences head kinematics differently depending on the metric analyzed. Therefore, based on the current understanding and injury metrics for sports related concussion, no consistent effect can be seen by changing neck muscle force.

These results contradict those in prior studies because a number of limitations prevent the prior studies from accurately capturing the poor head-to-neck coupling.[3–5] Collins et al. only examined whether or not an athlete received a concussion.[4] Confounding variables such as anthropometric size by position, previous head injury, or personality traits, may also lead to concussion. Several studies used human participants who, in a laboratory setting, can only be exposed to non-injurious events with low peak accelerations and long durations.[7–12] These long duration events maximize head/neck coupling and provide a poor surrogate for on field blunt impacts. Also, these studies examine changes in velocity and neck stiffness over large head excursions and durations. Because of this, these studies fail to capture the influence of neck musculature over the short distance and duration where sports related concussion is believed to occur.[30] Finally Viano et al. and Jin et al. utilized a linear spring mathematical model and GHBMC model that oversimplified cervical spine biomechanics and muscle activation Schemes.[3,5] Due to the complex vertebral body and musculature interactions, the cervical spine does not act as a linear spring, and the muscle activation scheme presented in Jin et al. would result in overly tense muscles and an unstable head positioning.[3]

The flexor and extensor muscle activation states are two extremes, and are unrealistic representations of an impact event as they would generate a counterproductive head position prior to the impact. Testing these activation schemes illustrates in even the most extreme scenarios, cervical muscle activation does not improve short term injury metrics.

Results from the second portion of this study indicate performance of common experimental neck conditions compared to a relaxed neck condition depend on the injury metric used. For linear acceleration injury metrics, HIII neck values were lower than the

relaxed neck condition values while free-head values were higher than the relaxed neck condition values. This is expected because a stiff HIII neck provides more linear resistance than a free-head with no neck.[19] For peak resultant angular acceleration, the free-head condition values were lower than the relaxed condition while the HIII values were higher. This may be because the neck provides a tether, translating linear kinetic energy to rotational kinetic energy. With peak resultant angular acceleration, free-head values are closer to relaxed condition values, supporting the hypothesis. The results suggest selecting the experimental neck condition depending on the metric of interest. However, for all neck conditions and injury metrics, except peak resultant angular acceleration with free-head, there is significant deviation from a relaxed neck condition that must be considered when reporting values.

This study has some limitations. First, impactors were modeled with linear elastic material for a nonlinear elastic scenario. However, we hypothesize nonlinear impactors would provide comparable results because of small impactor deformation. Also, DUHNM validation has only been conducted in the mid-sagittal plane. However, because this study compares results across the same model, conclusions should not change. Due to a lack of statistical variation in the modeling environment, only an effect size comparison could be conducted. Finally, while impactors 2 and 3 were able to successfully mimic the experimental head drop data, those experiments have limitations compared to on-field impacts. The drop test impact surface was a rubber covered rigid plate as opposed to a helmeted head, shortening impact duration. Accelerometers for the experimental drop tests were positioned near the center of gravity while model accelerations were determined at the head's exact center of gravity.

5. Conclusion

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 athletically relevant scenarios. This counters statements that strengthening the cervical musculature can lower an athlete's concussion risk. Increasing cervical muscle strength and activation may have different effects depending on the metric used to quantify injury risk, but ultimately impact location and impact magnitude influence short term head kinematics more than the force generating capacity of the cervical musculature. Finally, biofidelic limitations of both Hybrid III and free head experimental conditions must be recognized when reporting results.

References

1. Makdissi M, Darby D, Maruff P et al. Natural history of concussion in sport markers of severity and implications for management. *Am J Sports Med* 2010; 38:464–471.

2. Schmidt JD, Guskiewicz KM, Blackburn JT et al. The influence of cervical muscle characteristics on head impact biomechanics in football. *Am J Sports Med* 2014; 42:2056–2066.

3. Jin X, Feng Z, Mika V et al. The role of neck muscle activities on the risk of mild traumatic brain injury in american football. *J Biomech Eng* 2017; 139:101002.

4. Collins CL, Fletcher EN, Fields SK et al. Neck strength: a protective factor reducing risk for concussion in high school sports. *J Prim Prev* 2014; 35: 309–319.

5. Viano DC, Casson IR, Pellman EJ. Concussion in professional football: biomechanics of the struck player—part 14. *Neurosurgery* 2007; 61:313–328.

6. Broglio SP, Sosnoff JJ, Shin S et al. Head impacts during high school football: a biomechanical assessment. *J Athl Train* 2009; 44:342–349.

7. Eckner JT, Oh YK, Joshi MS et al. Effect of neck muscle strength and anticipatory cervical muscle activation on the kinematic response of the head to impulsive loads. *Am J Sports Med* 2014; 42:566–576.

8. Reid S, Raviv G, Reid Jr S. Neck muscle resistance to head impact. *Aviat Space Environ Med* 1981; 52:78–84.

9. Tierney RT, Sitler MR, Swanik CB et al. Gender differences in head-neck segment dynamic stabilization during head acceleration. *Med Sci Sports Exerc* 2005; 37:272–279.

10. Simoneau M, Denninger M, Hain TC. Role of loading on head stability and effective neck stiffness and viscosity. *J Biomech* 2008; 41:2097–2103.

11. Eckner JT, Goshtasbi A, Curtis K et al. Feasibility and effect of cervical resistance training on head kinematics in youth athletes: a pilot study. *Am J Phys Med Rehab* 2018; 97:292–297.

12. Mansell J, Tierney RT, Sitler MR et al. Resistance training and head-neck segment dynamic stabilization in male and female collegiate soccer players. *J Athl Train* 2005; 40:310.

13. Mihalik JP, Blackburn JT, Greenwald RM et al. Collision type and player anticipation affect head impact severity among youth ice hockey players. *Pediatrics* 2010; 125:e1394–e1401.

14. Mihalik JP, Guskiewicz KM, Marshall SW et al. Does cervical muscle strength in youth ice hockey players affect head impact biomechanics? *Clin J Sport Med* 2011; 21:416–421.

15. Nightingale RW, McElhaney JH, Richardson WJ et al. Dynamic responses of the head and cervical spine to axial impact loading. *J Biomech* 1996; 29:307–318.

16. Liu YK, Krieger K, Njus G et al. *Cervical spine stiffness and geometry of the young human male. DTIC document*, 1982.

17. Vasavada AN, Li S, Delp SL. Three-dimensional isometric strength of neck muscles in humans. *Spine* 2001; 26:1904–1909.

18. Beckwith JG, Greenwald RM, Chu JJ. Measuring head kinematics in football: correlation between the head impact telemetry system and hybrid iii headform. *Ann Biomed Eng* 2012; 40:237–248.

19. Dibb AT, Nightingale RW, Chancey VC et al. Comparative structural neck responses of the thor-nt, hybrid iii, and human in combined tension-bending and pure bending. *Stapp Car Crash J* 2006; 50:567.

20. Dibb AT, Cox CA, Nightingale RW et al. Importance of muscle activations for biofidelic pediatric neck response in computational models. *Traffic Inj Prev* 2013; 14:S116–S127.

21. Dibb AT, Cutcliffe HC, Luck JF et al. Pediatric head and neck dynamics in frontal impact: analysis of important mechanical factors and proposed neck performance corridors for 6-and 10-year-old atds. *Traffic Inj Prev* 2014; 15: 386–394.

22. Chancey VC, Ottaviano D, Myers BS et al. A kinematic and anthropometric study of the upper cervical spine and the occipital condyles. *J Biomech* 2007; 40:1953–1959.

23. Mohan P, Marzougui D, Kan C-D. *Development and validation of hybrid iii crash test dummy. SAE technical paper*, 2009.

24. Hendee SP, Greenwald RM, Crisco JJ. Static and dynamic properties of various baseballs. *J Appl Biomech* 1998; 14:390–400.

25. Cullen DK, Harris JP, Browne KD et al. A porcine model of traumatic brain injury via head rotational acceleration, In: *Injury models of the central nervous system: methods and protocols.*, 2016. p. 289–324.

26. Newman JA, Shewchenko N. *A proposed new biomechanical head injury assessment function-the maximum power index. SAE technical paper*, 2000.

27. Eiband AM. *Human tolerance to rapidly applied accelerations: a summary of the literature*, 1959.

28. Margulies SS, Thibault LE. A proposed tolerance criterion for diffuse axonal injury in man. *J Biomech* 1992; 25:917–923.

29. Mertz HJ, Prasad P, Nusholtz G. Head injury risk assessment for forehead impacts. *SAE Tech. Paper* 1996.

30. Marjoux D, Baumgartner D, Deck C et al. Head injury prediction capability of the hic, hip, simon and ulp criteria. *Accid Anal Prev* 2008; 40:1135–1148.