Inducing Head Motion with a Novel Helmet during Head-First Impact Can Mitigate Neck Injury Metrics: An Experimental Proof-of-Concept Investigation using Mechanical Surrogates

T.S. Nelson (B.A.Sc.)[1,2], P.A. Cripton (Ph.D.)[1,2]

[1]Injury Biomechanics Laboratory and Division of Orthopaedic Engineering Research, Departments of Mechanical Engineering and Orthopaedics; [2]International Collaboration on Repair Discoveries (ICORD) University of British Columbia Vancouver, Canada

ABSTRACT

There is a need for cervical spine injury prevention from head-first impacts in many sports and in various transportation contexts. We present an experimental helmet prototype that induces anterior or posterior head motion in a head-first impact as a mitigation strategy. Instrumented mechanical surrogates for the human neck, head, and helmet were tested on a drop tower. Peak lower-neck axial force and moment were used as injury metrics. A factorial experiment examining 3 escapes, 3 platform angles, and 2 platform stiffnesses was performed. The appropriate head-motion "escape" reduced mean peak axial force by up to 56% and moment by up to 72% compared to no-escape.

INTRODUCTION

Head first impacts can cause permanent debilitating cervical spine injuries that devastate the lives of those afflicted and their families. These impacts occur in many different sports where helmets are worn such as football, hockey, equestrian, and the various bicycle and motorcycle disciplines. Many researchers studying cervical spine injuries in head-first impacts have found them to be dependant upon conditions between the head and impact surface. In an investigation using football helmet-clad cadavers, the researchers had to artificially create a fusion at the atlantooccipital joint in order for any significant force to be transmitted through the cervical spine (Hodgson, 1980). Nightingale and McElhaney showed that the cadaveric osseoligamentous cervical spine's axial stiffness increased with increasing constraint on the head (1991). They later showed that in a head-first impact, the head once stopped, had sufficient inertia to provide a constraining end condition for neck injury development. Some specimens avoided spine injury when the head translated and rotated along a frictionless and inclined surface while other more constrained specimens developed a wide array of unstable fractures (1996). A prevention strategy for compressive neck injury might therefore involve keeping the head moving along the impact surface at impact. The only study we are aware of addressing this was performed with a finite element model of a human head and cervical spine in a head-first impact with an experimental roof during rollover (Halldin, 2000). The roof structure had an asymmetrical spring that deflected towards the front of the vehicle during impact causing an occupant's head to translate anteriorly. 27% and 44% reductions in cervical spine axial force for perpendicular and -15º oblique impacts respectively were reported with this roof.

In a head-first impact, we envision the cervical spine as a segmented beam-column trapped between two masses, the head and torso, about to incur an axial load. Oblique motion of the head at impact should provide a less constraining inertial end-condition on the spine that increases eccentricity promoting a spinal posture with lower axial stiffness that is less capable of axial load. Furthermore, it should help promote a bending response over an axial one shielding energy absorption away from the bony spinal column to the musculature and soft tissues over a larger range of displacement and rotation.

As many of these injuries occur in the presence of a helmet, the objective of this work was to evaluate a novel helmet prototype of our design that uses this strategy for neck injury prevention over a range of head-first impacts. It was also desired to assess the efficacy and feasibility of this concept while providing direction for further development and design optimization.

METHODS

A custom mechanical neck, head, and 'helmet' were used with a drop tower. The seven segment aluminum and rubber neck shown in figure 1 has exhibited biofidelic mechanics under sagittal plane bending and axial impact. The helmet contained no substantial padding and was simply an experimental means of inducing head motion at impact. The helmet simulated induced head motion "escape" by use of two shells separated by 1.0" and connected through a passive guide mechanism.

The surrogate head was directly affixed to the passive mechanical guide system. At a threshold force, the head was guided anteriorly {flexion} or posteriorly {extension} relative to the helmet. Only one escape path was possible but could be set up to induce anterior or posterior translation. The motion geometry was the portion of a 4.75” constant radius arc starting 45º from horizontal such that the center point of the head moves through 1.0” horizontally over 0.75” vertically. A conformal pin on each lateral side of the surrogate head protruded into corresponding slots in Delrin Plates on the lateral internal borders of the helmet (Figure 1). This configuration applied relative rotation of 15º between head and helmet during the translation. A replicated full factorial experiment (N=36) was performed consisting of 3 platform angles {0, ±15º}, 2 padding stiffnesses, and 3 escapes {flexion ,extension, none}.

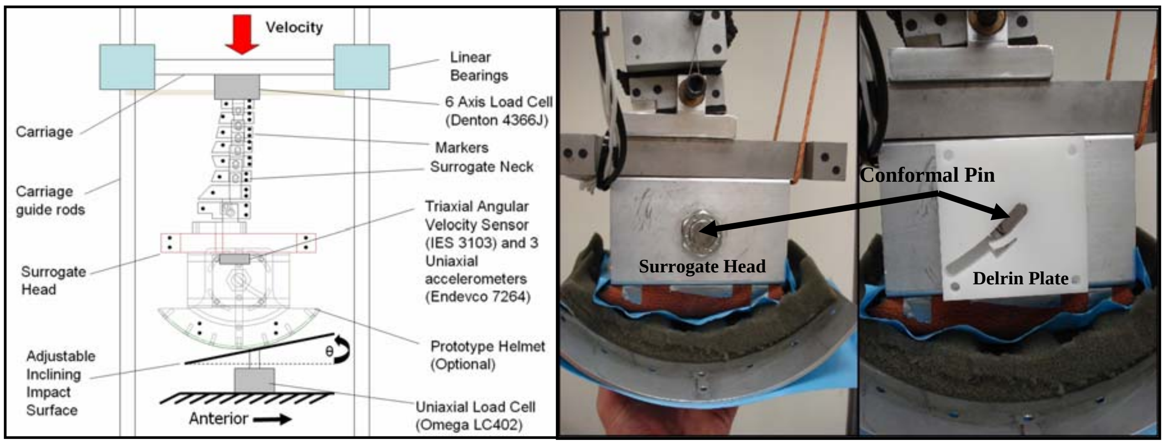

Figure 1: Test apparatus (left) & detailed helmet model with lateral plate removed (right).

Preliminary experiments showed platform angle and stiffness to be the two most influential variables on neck loading. The two stiffnesses were made from representative medium and low density foams used in athletic environments. Each drop was conducted from 60 cm and a Teflon:Teflon friction condition for the head/helmet:surface interface used.

Peak lower-neck axial reaction force and sagittal moment were used as neck injury metrics while head angular and linear accelerations were identified for comparing head injury potential. Each impact was imaged (Phantom V9) at 1000 fps synchronized with instrumentation consisting of: 6-axis load cell (Denton 4366J, accuracy 1% of FS = 133 N for force and 4.5 Nm for moment) at the lower neck, uniaxial load cell (Omega LC 402) under the impact surface, triaxial gyro sensor (IES 3103, accuracy 0.5% of FS = 0.3 rad/s) and 3 uniaxial accelerometers (Endevco 7264C) at the head CoG. All signals were sampled at 78 kHz with pre A/D anti-alias filters to comply with SAE J211b. Fiducial markers on each vertebra, head, and helmet were tracked for calculating 2D kinematics.

Univariate factorial ANOVA was performed separately for both injury metrics with significance α=0.05 and multiple pairwise comparisons were made to isolate differences with Bonferonni correction for 3-level variables angle and escape.

RESULTS

The most interesting result for both axial neck force and sagittal moment was the interaction between platform angle and escape. Figure 2 shows plots of the mean peak values across both platform stiffnesses ± 1 standard deviation. The moment graph shows the peak absolute value thus no directionality is discernable. The head linear acceleration data for all helmeted drops was strongly confounded with ringing during initial contact and deemed unrealistic. The angular velocity channels became mostly intelligible without significant attenuation after a 240 Hz to 150 Hz 4th order low pass Butterworth filter. This 150 Hz digital filter was used to compare peak angular accelerations.

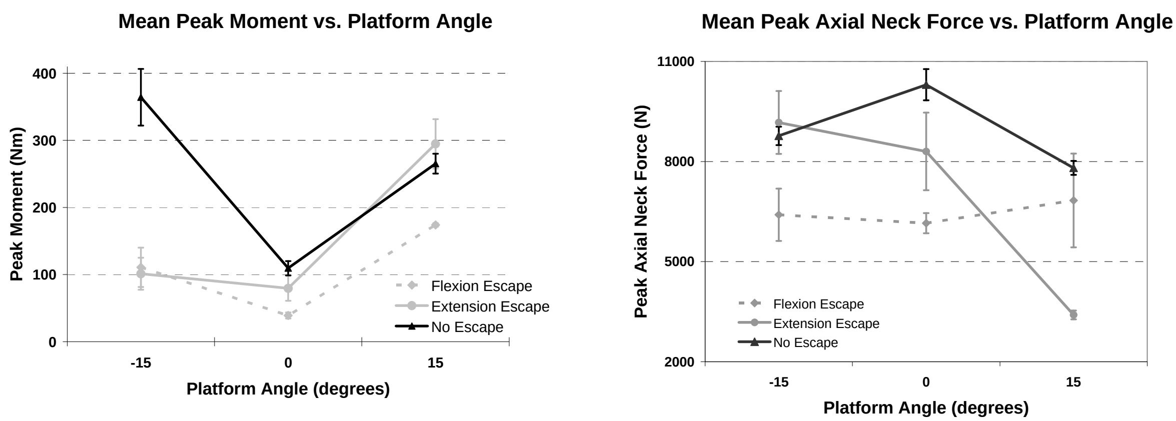

Figure 2: Platform angle and escape interaction for moment (left) and axial force (right).

Figure 3 shows a comparison between extension and no-escape at +15º onto the stiffer surface. With the helmet, the neck impulse develops more slowly reaching lower peak forces. This plot shows the largest reduction in axial force in any of the trials, however a similar trend was observed for all properly chosen escapes.

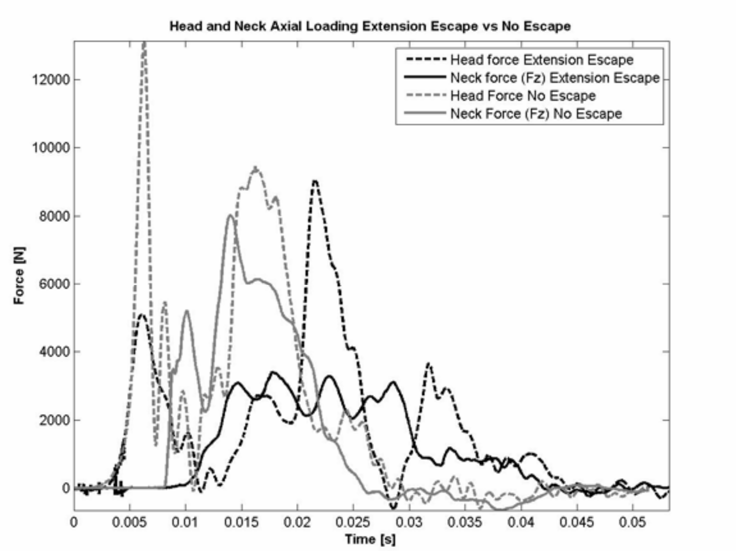

Figure 3: Head and Neck Force with and without Extension Escape at +15 on the stiffer surface.

Axial neck force

ANOVA showed significant results for all main effects and interactions except angle*stiffness ($p = 0.497$). At -15º (positive reference angle shown in figure 1), the mean peak neck force for the flexion escape was reduced by 27% ($p<0.005$) over no escape. Similarly, at +15º the extension escape showed 56% reduction over no escape ($p<0.005$, Figure 3). At 0º, flexion and extension escapes showed 40% ($p<0.005$) and 19% ($p=0.009$) reductions over no escape.

Sagittal neck moment

ANOVA showed all main effects and interactions were significant ($p<0.005$). At -15º, both flexion and extension reduced peak moments by 69% and 72% ($p<0.005$) respectively. At 0º and 15º, only flexion reduced moments by 63 and 34% ($p<0.005$) respectively.

Sagittal angular acceleration

3 pairs of usable data for no-escape and helmeted impacts were studied. Extension escape vs. no-escape for the two drops in figure 3 showed a 56% increase from 3915 to 8874 rad/s^2. Flexion escape vs. no-escape for a medium stiffness at -15º showed a 22% increase from 4293 to 5503 rad/s^2. At 0º with medium stiffness, a flexion escape vs. no escape caused a 75% increase from 1631 to 6360 rad/s^2.

DISCUSSION

The reductions in lower-neck loading suggest that a helmet can be used to reduce neck injury metrics in a head-first impact. For the angled platform impacts, the largest reductions in axial force were achieved when the induced motion was in the same direction that the angled platform would have caused naturally. This is consistent with the passive nature of the design and shows the importance that we will have to place on designing a selector mechanism capable of choosing the appropriate escape.

The angular accelerations were seen to increase for all three corresponding pairs studied. Zhang estimated the 25, 50, and 80% probability of MTBI from rotational acceleration to be 4600, 5900, and 7900 rad/s^2 (2004) respectively. Therefore, this metric must be carefully evaluated in the future.

The axial force flexion escape reductions at 0º and -15º were similar to those obtained by Halldin (2000). Although our non-frangible metallic spine may suggest higher load reductions than would be present with biological tissue, the mechanics of reducing human spine loading would be similar.

Biofidelity of our neck model is a limitation. While it does have realistic 2D axial and sagittal bending stiffnesses and ranges of motion, the out of plane motions are constrained. In addition, the surrogate head is not sufficiently isolated from natural frequencies in the helmet to allow linear acceleration measurements. A spectral analysis performed only on the linear acceleration channels showed that the helmet adds resonant frequencies around 250 Hz. Subsequent finite element and experimental modal analysis on the helmet confirmed this. Unfortunately, the true signal also has energy content at these frequencies.

It seems clear that beyond some impact velocity this strategy will not suffice for neck injury prevention. However, since the neck's tolerance to head-first impact is equivalent to only a 50 cm freefall (McElhaney, 1979), extending the tolerance by any amount would be beneficial. This system could also compliment performance of prophylactic neck braces aiming to couple the helmet to the shoulders. The induced motion and extra ride-down theoretically encourages a less-stiff neck response that would allow the alternate load path to take more load. The human cervical spine fails axially in an aligned posture at a mean 18 mm of displacement (Maiman, 2002) which highlights the challenge of designing a helmet-torso coupling device with sufficiently high initial-stiffness for neck load shielding that would allow adequate mobility for performance in the sport/activity. This design adds an additional 18 mm vertical ride-down.

CONCLUSIONS

To our knowledge, this is the first study to induce head motion in a head-first impact through the use of a helmet for neck injury mitigation. The results are encouraging and warrant further exploration. Considering the competing interests for reducing neck injury without exacerbating head accelerations, we believe that there exists an optimal path for minimizing neck loading while conforming to head injury tolerance constraints and we are actively working to identify this optimum helmet configuration.

ACKNOWLEDGEMENTS

The authors gratefully acknowledge financial support from the Natural Sciences and Engineering Research Council of Canada and the International Collaboration On Repair Discoveries.

REFERENCES

HALLDIN, P. (2000). Investigation of Conditions that Affect Neck Compression-Flexion Injuries using Numerical Techniques. 44th Stapp Car Crash Conference, SAE.

HODGSON, V. R. AND L. M. THOMAS (1980). Mechanisms of cervical spine injury during impact to the protected head, Warrendale, PA, 24th STAPP car crash conference.

MAIMAN, D. J., N. YOGANANDAN, ET AL. (2002). Preinjury cervical alignment affecting spinal trauma. J Neurosurg Spine 97(1): 57-62.

MCELHANEY, J., R. G. SNYDER, ET AL. (1979). Biomechanical Analysis of Swimming Pool Neck Injuries. Society of Automotive Engineers SP-79: 47-53.

NIGHTINGALE, R. W., B. S. MYERS, ET AL. (1991). The influence of end condition on human cervical spine injury mechanisms. Society of Automotive Engineers Transactions, Paper 912915: 391-399.

NIGHTINGALE, R. W., J. H. MCELHANEY, ET AL. (1996). Dynamic response of the head and cervical spine to axial impact loading. Journal of Biomechanics 29(3): 307-318.

ZHANG, L., K. H. YANG, ET AL. (2004). A proposed injury threshold for mild traumatic brain injury. J Biomech Eng 126(2): 226-36.

AUTHOR LIST

1. Timothy S. Nelson
   828 W 10th Ave, Vancouver, VCHRI, V5Z1L8, BC, Canada
   604 875 4111 x.66298
   tsnelson@interchange.ubc.ca

2. Peter A. Cripton
   828 W 10th Ave, Vancouver, VCHRI, V5Z1L8, BC, Canada
   604 822 6629
   cripton@mech.ubc.ca