White Paper: A helmet for prevention and mitigation of spinal column and spinal cord injuries in Head First Impacts

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

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

Introduction & Background: Head first impacts can cause some devastating injuries to the cervical spine. In the worst cases cervical spine fracture and injury to the spinal cord can occur leaving those afflicted paralyzed for life. Transportation accidents and sporting events are two of the most common situations for head-first impacts and in many of these activities helmets are worn. Some examples are a hockey player going head-first into the boards after a check from behind or a football player hitting another player head-first during a tackle. We think of the helmet as an engineered interface between the head and impact surface and are exploring whether its traditional role of head protection can be extended to include protection of the spine in head-first impacts.

Anecdotally we know there is no shortage of athletes in various sports that continue to suffer head-first impacts and most of these impacts occur with speeds capable of injuring the cervical spine. However, a review of the epidemiology associated with these injuries shows their incidence rates to be extremely low.¹-³

The anatomy and mechanical properties of the cervical spine may explain the low incidence rate. The human cervical spine is extremely flexible in multiple directions. It is in fact very seldom that our cervical spine will behave with a purely axial response to an axial load in a head first impact. Most times, the flexibility and posture of the neck combine with the location and direction of the force being applied to the head to cause the neck to bend and result in the head deflecting off the impact site without a neck fracture. Although fractures do occur in the neutral lordotic posture,⁴,⁵ when the spine is in a straightened posture, and the force to the head is along the axis of the straightened spine, the risk of severe neck injury increases.⁶,⁷ The healthy cervical spine in a normal posture has lordotic curvature. When the head is flexed forward (nodding) approximately 30 degrees this curvature is removed and thus the spine’s response to an axial force delivered to the crown of the head (or helmet) results in more of a compressive response rather than one combined with bending. In the aligned posture, the cervical spine is extremely stiff along its axis. The tolerance to impact in this posture is as low as 3 m/s⁸ and failure can occur at an average 18 mm of displacement.⁷ As this posture is so dangerous, the goal of the Pro-Neck-Tor™ helmet is to induce motion of the head along the impact surface, changing the neck’s posture while creating a more glancing impact between the head and impact surface.

A head-first impact is a dynamic event. The head, neck, and torso have the same initial velocity before the head makes contact. As the neck is situated between the head and torso, if the head stops instantly, the neck is required to bring the much larger mass of the
torso to rest. There is a large body of research showing the influence of different impact conditions between the head and impact surface. In general, the more the impact surface grabs and holds the head/helmet, the worse the risk of injury to the neck. Thus, conditions like high friction, soft stiffness foams, and more perpendicular impact surfaces constrain the head more than low friction, stiffer surfaces that have more oblique angles relative to the incoming velocity and spinal axis. It has also been shown that the head has sufficient mass that once it is stopped in a head-first impact, the neck will fracture before “pushing” the head out of the way. This is true even with a perpendicular, frictionless, and rigid impact surface. There are also some experiments where spinal injuries were avoided or impossible to create with slightly oblique impact angles as the head would simply deflect and not pass load into the neck.

This body of research suggested to us that a helmet could be developed that lowers the constraint on the neck by first lowering constraint on the head by inducing head motion along the impact surface. We hypothesize that by keeping the head in motion during torso deceleration the load passed through the spine will decrease. In order to test this hypothesis, a dedicated mechanical head, neck, and helmet was designed. More information about the surrogate head and neck can be found in the companion white paper on this website here.

**Methods:** The surrogate head was modified to accommodate an experimental helmet. This first prototype helmet has no padding but simulates the effect of having two shells connected by a passive mechanism which allows relative motion between the two shells after a force controlled deployment force is reached. As shown in figure 1, the surrogate head has protrusions on each lateral side. These insert into matching slotted guides on the outer shell. A tab prevents the motion until a specified threshold force has been reached. Our intended long term design incorporates two paths of available motion, either forward or backward, but the prototype described here allows only one motion which can be set up to allow either forward or backward motion.

Figure 1: Helmet with motion path guides (left), surrogate head with protrusions visible (middle) and with deployment-tab motion path guide overlaid (right).
The surrogate head, neck, and helmet were tested on a custom built drop tower. The drop tower simulates a head-first impact. The carriage had the effective mass of the 50th percentile male torso and was dropped onto an impact platform with adjustable angle as shown in figure 2. The instrumentation used for testing included three uniaxial head accelerometers (Endevco 7264C), a triaxial head angular velocity sensor (IES 3103), a uniaxial load cell underneath the impact platform (Omega LC402), and a 6 axis load cell at the lower neck (Denton 4366J). The analog data signals were sampled according to SAE standards.

![Figure 2: Schematic of Surrogate Head, Neck, and Helmet on Drop Tower.](image)

Preliminary drops without the helmet showed that the peak lower-neck axial force and sagittal moment were most sensitive to the angle of the impact platform and stiffness of the impact surface and these parameters were insensitive to surface friction or postural changes of 10 degrees of head flexion or extension. A series of drop tests were conducted with 3 platform angles, 2 platform stiffnesses, and 3 escape paths using the peak lower-neck force and peak lower-neck moment as measures of injury potential. All drops were conducted with a low friction Teflon interface between the head or helmet and impact surface and with an aligned neck posture. Factorial ANOVA was used for statistical analysis.

**Results:** The results of this study are in press or preparation for peer-reviewed journals but some highlights can be presented here. The most interesting results are seen in the interaction between the direction of escape (anterior or posterior) and the impact platform angle. When the angle of the platform would naturally cause the head to deflect towards the anterior direction, the helmet “escape” which also induces anterior head motion
significantly reduces both peak lower-neck axial force and sagittal moment relative to identical drops without the helmet. The same is true for the reverse case when the plate would naturally cause a posterior head deflection, the preferred helmet escape is the one which also induces posterior motion. Onto a perpendicular impact surface, both escapes significantly reduced lower-neck force and sagittal moment with the flexion escape having a larger reduction. Figure 3 shows this result graphically for the peak axial lower-neck force.

Figure 3: Interaction between helmet “escape” direction and impact platform angle.

A review of the high speed videos showed that with a preferred deployment, the head continues to stay in motion throughout the duration of the carriage (torso) deceleration. This had the effect of lowering the force transmitted through the neck. In figure 4, the lower-neck force is plotted for two impacts, one with the helmet and one without.

Figure 4: Lower-neck axial force with (black) and without (red) a flexion escape at 3.2 m/s onto a perpendicular impact surface with soft foam.
Figure 4 shows the lower-neck force for only two drops, one with and one without, a flexion (anterior) escape onto a perpendicular impact surface with low stiffness. The neck force with the helmet develops more slowly and reaches lower peak values. This trend was seen for all “correct” escapes, i.e. a flexion escape when a flexion escape was desirable.

Discussion: The reductions in neck loading suggest that inducing head motion with a helmet could potentially mitigate or prevent neck injuries in head-first impacts. The results, although preliminary are very encouraging and we believe demonstrate proof of the merit of this concept.

Like any experimental study, ours is not without limitations. The most obvious is the biofidelity of the surrogate neck model. While it was designed to have realistic stiffnesses in sagittal bending and axial compression, out of plane motions representing lateral bending and axial rotation are constrained. Also, the surrogate helmet is currently cylindrical instead of spherical which like the surrogate neck model, limits testing to the sagittal plane. The first helmet prototype simulates our long term design intent which is two shells, an inner connected to the head and an outer. In the prototype, the head is directly attached to the “male” half of the motion mechanism such that realistic helmet retention, i.e. a chin strap, is not included.

All of these concerns are being addressed with the development of a 3D helmet with two shells and a selector mechanism, energy absorbing padding, and realistic helmet retention. In addition, methods of testing cadaveric cervical spines in the presence of muscle force replication with a Hybrid III dummy head have been developed. We are actively working towards developing and testing more advanced prototypes of the helmet.

References: