Forces applied to the head and their relationship to risk of brain injury

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Preventing concussion and traumatic brain injury (TBI) has been an ongoing mission for protective equipment developers. Unfortunately, the variety of potential injurious events combined with the individual user vulnerabilities make proper design and use of protective gear a constant challenge. Through understanding of the injuries that occur and the loads that cause those injuries, directed improvements to protective equipment can be made to reduce both the incidence and severity of injuries that occur during sporting events, automobile crashes, and even combat. However, current evaluation and rating standards (e.g., the Head Injury Criterion), are inadequate for the proper evaluation of most protective equipment, particularly oral protective appliances built to reduce loads transmitted through the jaw and mandible and into the skull and brain.

A concussion is the application of external forces to the head that initiates a series of biological responses that disrupt the functions of the brain. The disruption of brain function can occur over time or immediately, resulting in either temporary or permanent loss in cognitive ability. The external forces that cause injury can be delivered to the head by mechanisms ranging from a punch in boxing to a head-to-head collision between two football players. For example, a study sponsored by the National Football League (NFL) estimated the relative velocity of two player's heads in a concussive collision is 21 miles per hour (9.3 meters per second).[1] Collisions at this velocity are very likely to result in concussion. However, if the velocity of the struck player's head can be reduced to below 15 miles per hour (7.0 meters per second) the risk of concussion is greatly reduced.[1]

In addition to impact velocity, impact location on the head can also affect the likelihood of a concussion. The same NFL study showed that concussions were primarily observed during impacts to the facemask, side of the head, or during falls where the back of the head struck the ground. But impacts to the top of the head at the same velocity were less likely to result in a concussion. Impacts to the facemask represent additional risk, as a greater percentage of the impact force is transmitted through the mandible. Researchers have estimated the load through the mandible for a potentially concussive 15 mile per hour impact to be on the order of 470 pounds (2100 Newtons).[2] This is in the range established by previous studies which determined that a load of 400 pounds (1780 Newtons) applied to the mandible could fracture the mandible approximately 50% of the time.[3,4,5]

These forces are external loads to the head. Unfortunately, existing instrumentation cannot directly measure the forces transmitted to the brain; the forces are therefore inferred by measuring the global motion of the head or forces contacting the head. Global motions of the head are estimated by recording the linear and/or angular acceleration of a mechanical headform that has the same weight and size of a human head. Acceleration refers to the rate at which the velocity of the head changes. The faster the velocity of the head changes the greater the acceleration. Acceleration is often measured in multiples of gravity, based on the reference acceleration measured at sea level. The acceleration of the headform is typically measured using accelerometers located at the center of gravity of the headform. Contact forces are measured by load cells imbedded into the surface of the headform. Depending on its location, the load cells can be used to measure a variety of forces. Load cells imbedded in the facial region of the

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[1] Pellman EJ (2003) Concussion in professional football: Reconstruction of game impacts and injuries. Neurosurgery, 53(4):799-814.

[2] Viano DC (2011) Effect of mouthguards on head responses and mandible forces in football helmet impacts. Ann Biomed Eng 40(1): 47-69.

[3] Cormier, J.M., Bisplinghoff, J.A., Duma, S.M. (2008) Fracture Tolerance Thresholds of Human Facial Bones Subjected to Blunt Impact. Prepared for United States Army Aeromedical Research Laboratory, CIB Report 2008-030.

[4] Nahum AM (1975) The Biomechanics of Facial Bone Fracture, Laryngoscope., 85(1): 140-56.

[5] Schneider, D.C., Nahum, A.M. (1972) Impact studies of facial bones and skull. 16th Stapp. Car crash conference . SAE Paper 720965 pp. 186.

headform can measure the contact force of the facemask striking the head. When imbedded into the jaw, the load cell will measure the force delivered by a punch in boxing or through the chinstrap of a helmet.

There is not yet sufficient data to support a specific correlation between external force to the head and alterations in brain function. However, cadaver testing and limited data from instrumented players in impact sports indicates that the greater the force, the greater the risk of disrupting the functions of the brain. Therefore, if the external forces delivered to the head can be minimized, the potential risk of brain dysfunction is reduced. To minimize the forces transmitted to the head, equipment such as helmets, faceguards and oral appliances (i.e. mouthguard) are used. Each of these devices provides protection by reducing the forces transmitted to a different region of the head. For instance, helmets use padding to reduce the forces transmitted to the side, rear and top of the skull. The peak force is reduced by spreading the applied force over a longer period of time. Even if the peak force is not reduced, by stretching out the time it takes to reach the peak force, the peak acceleration of the head is reduced. Facemasks reduce the forces applied to the face which have the potential to cause facial fractures. In addition to protecting the teeth and tongue, there is data that shows oral protective appliances have the potential to reduce impact forces applied through the jaw and into the base of the skull.[6]

For evaluating the performance of helmets and facemasks, relationships have been developed between the linear acceleration of the head and the measured external forces. Linear acceleration of the head has been directly correlated to the occurrence of skull fracture and arterial injuries in the region between the skull and brain (e.g., subdural hematoma). The injury threshold for peak linear head acceleration is 98±28g.[1] Researchers also observed that if the linear acceleration was applied over a longer period of time, the peak acceleration which could be tolerated was lower. The Severity Index (SI) is an injury criterion built on this observation and establishes a relationship between linear acceleration, impact duration and observed brain injury.[7] The SI threshold for brain injury in sports is currently 300.[1] SI is applied by the sports community to evaluate all types of head protection equipment. The automotive community uses a similar measure called the Head Injury Criteria (HIC). The difference between them is that the HIC is limited to a duration of no more than 36 milliseconds.

Numerous studies have shown that during an impact where the head is struck the use of protective equipment, in general, reduces the peak linear acceleration of the head. The exception to this observation is with the use of certain types of oral appliances, though there are limited studies that look at oral protective appliances due to a lack of headforms appropriate for such studies. Viano *et al*[2] demonstrated that single arch oral protective appliances appeared to have less of an effect on reducing the linear acceleration of the head compared to dual arch oral protective appliances during impacts to the face shield of a football helmet. Some single arch protective appliances actually increased the SI over the no mouth gear controls. Dual arch devices reduced the SI by 19% from the control for an impact at 15 mph and 13% for the 20 mph impact.[2] Unfortunately SI has only been validated against severe head injuries such as skull fractures and arterial brain injuries such subdural hematomas. SI has not been quantified to the associated reduction in the risk of mild or moderate head injuries typically associated with impacts in sports. Another limitation of using Peak g, SI or HIC is these approaches only take into account linear acceleration during the impact. It is known that angular acceleration and forces applied directly to the head are also a factor in brain injury.[8] Neither SI nor HIC can account for all possible mechanisms of brain injury, so additional injury criteria should be considered in the evaluation of protective equipment, particularly oral protective appliances.

Football helmets, facemasks and dual arch oral protective appliances all reduce the peak force transmitted to the head during a football impact. Oral protective appliances also minimize the forces transmitted to the teeth and the base of the skull for non-helmeted impacts. The data from retrospective

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[6] Hickey JC, (1967) The relationship of mouth protectors to cranial pressure and deformation. JADA 74(3):735-740.
[7] Gadd CW (1966) Use of a weighted-impulse criterion for estimating injury hazard SAE Paper 660793. Society of Automotive Engineers, Warrendale, PA.
[8] Kimpara H (2012) Mild Traumatic brain injury predictors based on angular acceleration during impacts. Ann Biomed Eng 40(1): 114-126.

studies suggest that the use of oral protective appliances may also reduce the risk of brain injury in rugby[9] and boxing[10]. While the exact relationship between the use of oral protective appliances and the reduced risk of brain injury has yet to be quantified for any sport, the boxing community observed long ago that knockouts are often a result of a forceful blow delivered to the jaw.[11]

There are many reasons why the jaw could be more vulnerable than other regions of the head. First a blow to the chin can cause large head accelerations due to the natural lever arm created by the jaw. Additionally, the skull surrounding the brain is thinnest in the region directly over the jaw joint. If this region of the skull is compromised, the resulting injuries could include subdural hematomas.[12] Furthermore, the blood supply to the brain and some major nerve bundles exit the skull in this region of the skull. The temporal lobe of the brain lies directly above the jaw joint. Inside the temporal lobe is the hippocampus, which is essential for memory function, particularly the transfer of thoughts from short to long term memory and control of spatial memory and behavior. The risk of injury to these structures from forces transmitted through the base of the skull cannot be easily estimated with peak acceleration or SI. New criteria which account for the forces transmitted through the jaw joint are needed for the evaluation of the efficacy of oral protective appliances in reducing the risk of injury.

While the exact relevance of all loading mechanisms to brain injury is still unknown there are several methods to estimate the forces applied to the base of the skull and how the various types of oral protective appliances mitigate these forces. The first is through the use of X-ray imaging of athletes who sustained head injuries. The radiological data from these athletes show compression fractures, condylar neck fractures, and degenerative condylar remodeling.[10] Changes in the structure of the mandible such as these are indicators of the application of significant forces. Symptoms associated with mandibular injuries include headaches, nausea, vomiting, vertigo, and sensitivity to light (all of which are considered common symptoms of most brain injuries).

Another method to estimate the forces transmitted through the mandible is with the use of mechanical headforms. Viano (2011) used a customized mechanical headform with an articulating jaw. This custom headform enabled them to evaluate the relationship between the use of different types of oral protective appliances and forces transmitted to the base of the skull.[2] The greatest reductions in force measured with the mechanical headform were achieved with the dual arch oral protective appliances. A dual arch appliance pulled the lower arch forward and locked the upper and lower arches together. In contrast, some single arch oral protective appliance increased the forces in this region compared to the control case where no mouthguard was used.

External forces to the head can initiate a series of biological events that disrupt brain function. To reduce forces and the risk of brain injury, various types of protective equipment have been designed. Protective equipment is designed to minimize forces applied through a variety of directions and mechanisms, including different impact locations, directions, and angles. However, since each region and mechanism causes unique responses, there is a need for multiple injury criteria for the evaluation of the various types of equipment. Currently the sports community only has the ability to voluntarily test protective head gear using head global linear acceleration-based injury criteria. While this approach may be appropriate for some types of helmeted impacts, the application of this one injury criteria is inadequate to evaluate all protective gear against all type of threats. Additional research is required to develop a brain response based injury metric that does not rely on inference like global acceleration of the skull to estimate the risk of injury. Such new criteria can account for rotation acceleration and mandible forces, which are particularly relevant to the evaluation and rating of oral protective appliances.

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[9] Blignaut JB (1987) Injuries sustained in rugby by wearers and non-wearers of mouthguards Br J Sp Med, 21(7):5-7.

[10] Williams ED (1994) Jaw joint disorders in contact sports. Head and neck injuries in sports. ASTM STP 1229 American Society for Testing and Materials, Philadelphia.

[11] Duncanson JG (1903) The knock out blow on the point of the chin. British Medical Journal. April 4; 782-783.

[12] Bhaskar SN (1986) Orban oral and embryology. CV Mosby Co St Louis, MO.