Current and Future Concepts in Helmet and Sports Injury Prevention

T. Blaine Hoshizaki, PhD*

Andrew Post, PhD*

R. Anna Oeur, MSc*

Susan E. Brien, MD‡

*School of Human Kinetics, University of Ottawa, Ottawa, Ontario, Canada; ‡Department of Neurosurgery, Centre de Santé et de Services Sociaux (CSSS) Gatineau (Hull site), Quebec, Canada

Correspondence:
Thomas B. Hoshizaki, PhD,
University of Ottawa,
A106-200 Lees Avenue,
Ottawa, ON, Canada,
K1N 7P1.
E-mail: thoshiza@uottawa.ca

Received, August 1, 2013.
Accepted, May 12, 2014.

Since the introduction of head protection, a decrease in sports-related traumatic brain injuries has been reported. The incidence of concussive injury, however, has remained the same or on the rise. These trends suggest that current helmets and helmet standards are not effective in protecting against concussive injuries. This article presents a literature review that describes the discrepancy between how helmets are designed and tested and how concussions occur. Most helmet standards typically use a linear drop system and measure criterion such as head Injury criteria, Gadd Severity Index, and peak linear acceleration based on research involving severe traumatic brain injuries. Concussions in sports occur in a number of different ways that can be categorized into collision, falls, punches, and projectiles. Concussive injuries are linked to strains induced by rotational acceleration. Because helmet standards use a linear drop system simulating fall-type injury events, the majority of injury mechanisms are neglected. In response to the need for protection against concussion, helmet manufacturers have begun to innovate and design helmets using other injury criteria such as rotational acceleration and brain tissue distortion measures via finite-element analysis. In addition to these initiatives, research has been conducted to develop impact protocols that more closely reflect how concussions occur in sports. Future research involves a better understanding of how sports-related concussions occur and identifying variables that best describe them. These variables can be used to guide helmet innovation and helmet standards to improve the quality of helmet protection for concussive injury.

KEY WORDS: Concussion, Head impact protocols, Head protection, Helmet standards, Mechanism of injury

Neurosurgery 75:S136–S148, 2014 DOI: 10.1227/NEU.0000000000000496 www.neurosurgery-online.com

Head injuries in sports were originally defined by traumatic events requiring immediate neurosurgical attention and often involved death. In the late 1800s, the regular occurrence of cranial fractures and death in American football sparked concern for safety.[1] American college football saw 19 deaths in 1905, prompting then President Theodore Roosevelt to hold a meeting at the White House in 1905 to change the rules with the intention of making the game safer.[2] By 1940, the National Collegiate Athletic Association and the National Football League made helmets mandatory. Since this inception of head protection, the incidence of brain injury–related deaths in American football decreased from approximately 150 deaths in 1965 to 1974 to approximately 25 deaths in the years 1985 to 1994.[3] The origin of the Snell Foundation (1957) as a standard for motor sport helmets was established in response to Peter Snell's death from head injuries received while racing cars. In 1979, the National Hockey League followed suit and adopted head protection as required equipment due to the death of professional hockey player Bill Masteron.[4] It was primarily events involving death or severe brain injuries that established the foundation for the development of sport helmet standards and design.

ABBREVIATIONS: EPP, expanded polyprolylene; **EPS,** expanded polystyrene; **HIC,** Head Injury Criterion; **ISO,** International Standards Organization; **NOCSAE,** National Operating Committee on Standards in Athletic Equipment; **TBI,** traumatic brain injury; **VN,** vinyl nitrile; **WSTC,** Wayne State Tolerance Curve

Research investigating the relationship between direct and indirect loading of the head and the resulting brain injury was the primary focus of research on head injury mechanisms in the 1950s.[5-8] Gurdjian et al[8] identified intracranial pressure over time as an effective predictor of severe concussions causing loss of consciousness from direct impacts to the head. Intracranial pressure was impractical to measure on live humans, so head linear acceleration was chosen as a reasonable alternative to represent head trauma, as the 2 were found to be correlated by early cadaver and animal research. From this research, a linear acceleration–time curve (Wayne State Tolerance Curve [WSTC]) was created by Gurdjian et al[5] as a means of defining the tolerance of the human head to an impact. This work was the foundation for many of the current sports standards that use peak linear acceleration or acceleration-time calculations as performance criteria for helmets. These criteria have been successful in the reduction of traumatic brain injury (TBI) in sports. However, sports-related concussions have become a serious problem with associated symptoms that can result in serious and persistent disability. Although both linear and rotational accelerations can contribute to head injury, sports-related concussion is more associated with rotational acceleration.[9-11] The prevalence of concussive injury suggests that the use of performance criteria based on the WSTC and thus linear acceleration may be insufficient, indicating that new performance criteria, such as rotation-based metrics, may be necessary.

Although there are numerous ways to impact the head in sport, the most common mechanism in professional American football is helmet-to-helmet blows, representing 61% of impacts causing concussion.[12] In competitive ice hockey, 88% of reported concussions result from impacts to the head involving the shoulder, elbow, and hands of the opposing players.[13] However, the only type of testing undertaken for the certification of sports helmets uses a drop rig representing a fall to the ground and linear acceleration as the criterion variable. Recent research has provided test methods using high-risk impacts and accurate measures of rotational accelerations to better evaluate concussion.[14] More recent developments involving sophisticated finite-element models of the skull and brain[15-17] have provided a powerful tool defining the relationship between impact characteristics and the resulting brain tissue trauma. Subtle differences in impact location, angle, mass, compliance, and velocity all influence the resulting magnitude, character, and location of brain trauma.[18-20] Understanding the injury mechanisms resulting in concussions involved in specific sports is imperative in developing safer protective helmets.

Helmets do an excellent job for what they were originally designed to do, which is preventing catastrophic brain injuries. Unfortunately, the same cannot be said for the success of helmets in mitigating the risk of concussions, as concussive injuries in helmeted sports such as American football and ice hockey remain a concern at the professional, collegiate, and high school levels.[21-24] This article presents a review of literature of the biomechanical forces related to head injury and the discrepancy between how helmets are designed and tested and how concussions occur in sport.

CRITERION VARIABLES AND CONCUSSION

Research involving brain injury mechanisms is typically designed to identify the most effective criterion variable(s) for predicting the risk of concussion.[25] Once identified, these variables are then used to establish the effectiveness of helmets and other protective technologies.

Linear Acceleration

Linear acceleration describes translational motion of the head and is currently the most common variable used for certification of helmets in the sports industry.[25,26] This variable has been in use since researchers examined the mechanism of TBI in monkeys and cadavers, noting that peak resultant linear acceleration correlated with injurious pressure waves within the skull.[5,27,28] Linear acceleration has also been used as a measurement variable for predicting the risk of skull fracture, with magnitudes between 200 g and 300 g.[29-34] As a result, this research established pass/fail criteria for helmet standards of approximately 250 g to 300 g.[26] The use of this variable in the development of helmet design has led to the reduction of traumatic brain injury (TBI) and skull fracture in sports, but has had limited effect on decreasing the incidence of concussion.[25]

Gadd Severity Index

Although the use of peak resultant linear acceleration was successful in improving the design of helmets to reduce the incidence of TBI, it has been reported to have a low correlation with brain injury when used in predictive research.[35] Researchers proposed that this may be a result of the peak value not accounting for the duration of time that the head was in translational motion from the impact.[5] This led to the development of the Gadd Severity Index (GSI), which is a function based on the WSTC. This curve was developed from direct impacts to animals and cadavers that accounted for the relationship between the duration of impact-induced motion and the magnitude of the translations. The WSTC revealed that high-magnitude translations could be endured for short periods of time and low-magnitude translations for longer periods of time.[36] From these data, the GSI was created for use in the helmet industry and is currently used by the National Operating Committee on Standards in Athletic Equipment (NOCSAE) for the majority of their helmet standards[37] and is represented as:

$$\mathrm{GSI} = \int_{t0}^{t} a^{2.5} dt$$

where a is the response function (acceleration), 2.5 is a weighting factor, and t is time. The value of 2.5 was an approximation of the slope of a log-log plot of the WSTC. A GSI value of 1000 was used as a proposed injury threshold based on data from Wayne

State. Pulses of very long duration (50 ms) were considered inconsequential in this equation because helmeted head impacts are normally less than 15 ms. Although the GSI was interpreted as an improvement over peak resultant linear acceleration, the use of this variable in NOCSAE standards governing football helmets has had a negligible effect on decreasing the risk of concussion because the incidence of injury remains high.[21]

Head Injury Criterion

Head Injury Criterion (HIC) also uses linear acceleration.[38] This measure is commonly used in the motorsport industry and is similar to the GSI with the difference being that it assigns a time limit to the peak resultant linear acceleration–based calculation, those being 36 ms for automotive crashes and 15 ms for direct impacts and helmet development.[39,40] The HIC calculation is represented as:

$$\mathrm{HIC} = (t - t0)\left[\left(\frac{1}{t - t0}\right)\int_{t0}^{t} a(t)dt\right]^{2.5}$$

where t is time and a is peak resultant linear acceleration. In addition, an HIC value of 1000 was suggested as having an equivalency to an abbreviated injury scale rating of 4 (severe).

Limitations of the GSI and HIC

By impacting cadavers, researchers demonstrated that GSI and HIC were correlated with brain injury.[41] However, the GSI and HIC were later proven to be invalid measures for relating brain injury to the head-impact responses in anthropometric test dummies.[35] They also assumed the skull/brain system was a rigid structure, where anatomically the skull can be considered elastic under many impact conditions, and the brain is anisotropic and nonlinearly viscoelastic in nature. In addition, any relationship between these variables and the abbreviated injury scale was later reported to be unfounded.[35]

Limitations of Linear Acceleration Measures for Predicting the Risk of Concussion

In terms of measuring protection against concussion, current variables used to evaluate the performance of helmets have not been particularly effective in decreasing the incidence of concussion.[23,24] This is linked to the limitations of using variables based on peak resultant linear acceleration as performance metrics for head protection. Peak resultant linear acceleration is a measure linked to TBI[5,28]; however, the mechanism of concussion has been more closely linked to rotations of the head.[7,11,42,43] In addition, peak resultant linear acceleration is a summation of the motion of the head in the x, y, and z directions. As a result, information regarding the direction of the impact and how that would affect the resulting brain injury is compromised. It is also unlikely that 1 criterion variable would be effectively applied across all age ranges because the human brain is known to undergo changes in composition through the natural aging process that would affect how it responds to injurious loading.[44,45]

Rotational Acceleration

All impacts to the head result in both translation and rotation. The peak resultant linear acceleration and other related helmet performance metrics have the same limitation in that they do not measure the resulting head rotation. The common measure of this motion in terms of brain injury is rotational acceleration. This is of particular significance as researchers have identified that the type of diffuse shearing of brain tissue attributed to concussion is related to the severity of the rotation of the head during an impact.[9-11] This shearing of brain tissue associated with rotation is related to the physical characteristics of brain tissue, with a very low resistance to shear forces associated with rotation, but a high resistance to compressive forces associated with translation.[46,47] Current helmet technologies are designed solely based on parameters that measure translation, leaving shear-induced brain injury from rotational acceleration largely unaccounted for. Currently, there are no helmet standards that use rotational acceleration as a measure of helmet performance, and it is likely that until such a standard exists, there will be limited innovation in rotation damping technologies.

Although using rotational acceleration is generally accepted as an important and necessary step in the attempt to mitigate the risk of concussion with regard to helmet design in sports, it has some degree of limitation.[25,48] Linear and rotational accelerations are measures of the motion of the head from an impact, not the brain tissue. This may explain that although there is an association between rotational acceleration and the mechanism of concussion, a definitive threshold for concussion has yet to be identified.[25,48] Concussive injury is undoubtedly linked to the kinematics of the event, but it is the interaction between the brain tissue and those motions that result in injury.[25] This inability to measure rotational accelerations is one of the reasons why the incidence of concussion has not been affected by improvements in helmet designs over the years.

The Use of Brain Tissue Variables in Measuring the Risk of Concussion

Linear and rotational accelerations have not been particularly helpful in predicting the risk of concussion. Researchers use finite-element modeling of the human brain to measure the stresses and strains in brain tissue from simulated impacts.[11,15,17,49] This analysis allows for the interpretation of how the translations and rotations of the head from an impact distort the brain tissues.[48] The most commonly reported measurements of these distortions of brain tissue from an impact include maximum principal strain, von Mises stress, strain rate, product of strain and strain rate, and shear strain/stress (Table 1). Maximum principal strain is a measurement of brain tissue stretch and is used by researchers because it is the closest comparison with anatomic failure testing

TABLE 1. Proposed Brain Response Thresholds for 50% Risk of Concussion

Concussion Threshold Value (50% Chance)	Dependent Variable	Location	Reference
0.21	Maximum principal strain	Corpus callosum	Kleiven[17] (2007)
0.26	Maximum principal strain	Gray matter	Kleiven[17] (2007)
0.19	Strain	Midbrain of the brainstem	Zhang et al (2004)
48.5 s^{-1}	Strain rate	Gray matter	Kleiven[17] (2007)
10.1 s^{-1}	Product of strain and strain rate	Gray matter	Kleiven[17] (2007)
8.4 kPa	von Mises stress	Corpus callosum	Kleiven[17] (2007)
7.8 kPa	von Mises stress	Brain stem	Zhang et al[49] (2004)
18 kPa	von Mises stress	Brain	Willinger and Baumgartner[15] (2003)

and thus allows some degree of comparison with rheological research.[25] Von Mises stress describes the amount of force per unit area that a region of brain tissue undergoes.[50] This parameter is commonly used in engineering for structural failure analysis. Strain rate is used because it has been demonstrated through rheological research that the rate at which the brain tissue is loaded affects its ultimate point of failure.[51,52] Other researchers have created a hybrid measurement involving the product of strain and strain rate under the theory that essentially what drives the failure mechanism of brain tissue is the rate at which it is loaded and the amount of stretch it undergoes.[48,49] Finally, shear stress/strain is measured because it describes the forces acting across the axons of the brain tissue. Some researchers have postulated that it is this shearing that can cause injuries such as concussion and diffuse axonal injury.[53]

Finite-element models of the brain use the dynamic response of the head from an impact to calculate the resulting brain tissue deformation supporting more informed helmet development designed to reduce the stresses and strains in brain tissue. The measurement of brain tissue stress and strain can also allow for the understanding of how the translations and rotations affect different types of brain tissue including white matter.[54] It has also been used to demonstrate how parameters such as impact velocity, mass, and location bring about peak stresses and strains in different regions of the brain.[19,20,55] The additional refinement allows for the design of helmets that manage both linear and rotational accelerations and thus reduce the resulting stress and strain in the brain tissue. The added information provided by the finite-element modeling of the brain facilitates the investigation of how technologies that manage rotations may reduce strains in the brain tissue associated with concussion.

Description of Events and Impact Characteristics

The types of impacts causing concussion will be unique to the sport; however, they can be grouped into falls, collisions, punches, and projectiles. The specific set of impact conditions for each injury event will define how energy is transferred to the head and brain. Falls to the turf or grass are common in soccer, football, and rugby and to the ice in figure skating and ice hockey. Typically, the conditions surrounding a fall are characterized by the weight of the protected or unprotected head onto an immovable and rigid impact surface. Reconstructions of falls typically use a guided drop of an anthropometric dummy head and neck onto a rigid anvil. Figure 1 illustrates a setup of a fall-type impact on ice using a monorail drop system and a Hybrid III head and neck form. These types of impacts cause a large amount of energy to be transferred to the head as rigid surfaces afford little compliance. Research on injuries as a result of falls has found impact velocities of the head to be upward of 6 m/s onto surfaces like ice and concrete.[56-59]

Collisions involving the head are common in contact sports such as American football and ice hockey, where players are required to wear protective helmets. In these types of sports, concussions can occur from impacts with the shoulder, elbow, and helmets of other players.[12,13,60,61] The conditions surrounding each head collision are largely dependent on characteristics of the body part coming into contact with the player's head, such as the equipment worn and the effective striking mass of the impacted

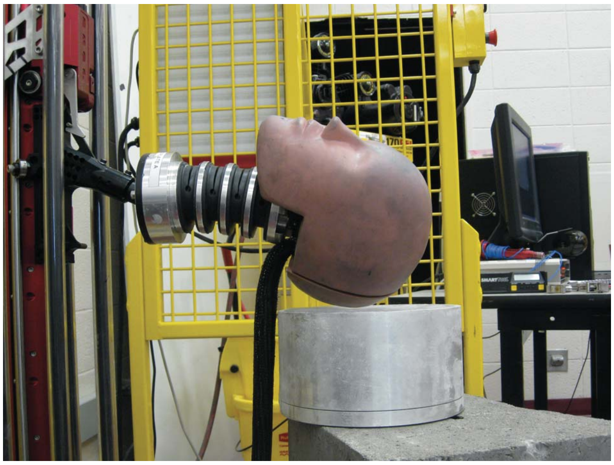

FIGURE 1. *A monorail drop rig system used to reconstruct a fall-type event onto ice.*

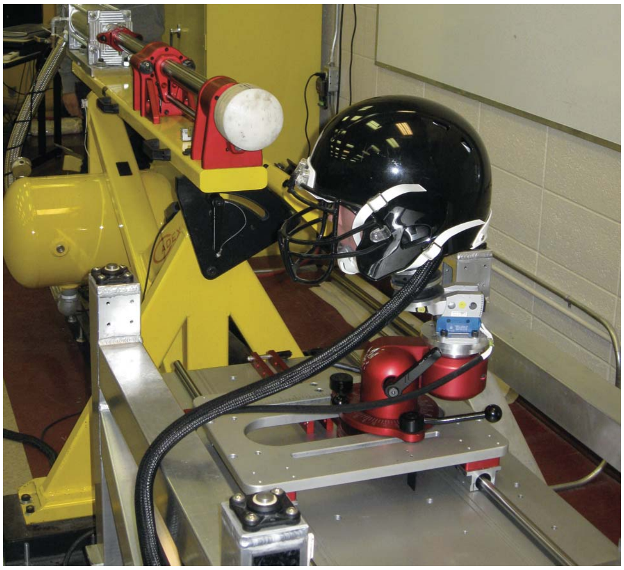

FIGURE 2. *A pneumatic linear impactor system used to reconstruct collision-type events.*

player. To reconstruct player collisions, a linear impactor system uses compressed air to accelerate an impactor arm (13 or 16 kg mass) to deliver blows to a headform. Helmet-to-helmet impacts can be replicated by covering the impactor arm with a vinyl nitrile (VN) nylon cap to represent a helmet and impacting a helmeted headform as shown in Figure 2.[12] Other impact characteristics include impact velocity, location, and angle of the colliding body parts. Reported velocities for player-to-player contact can range from 4 to 11 m/s in American football.[12] Impact mass involved in athlete collisions can reach 15 kg for full-body collisions, which are still much lower compared with impacting the ground.[12]

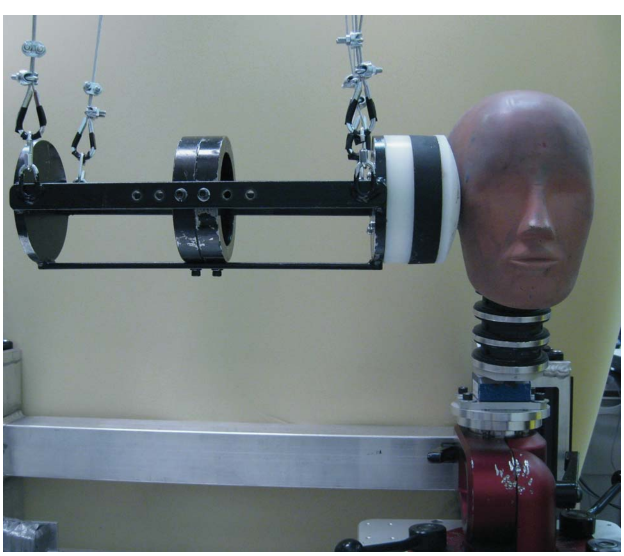

FIGURE 3. *Low-mass pendulum impact used to reconstruct punch-impact events.*

Concussion resulting from a punch to the jaw or temple is common in combative sports. The objective in boxing is to disable the opponent through targeted blows to the head and body, often involving multiple cranial impacts, which result in a concussion. Although punching is the main strategy for creating a concussion in boxing, they also occur in other sports such as fighting in ice hockey.[13] A punch is characterized by the relatively low effective mass of the fist and arm ranging from 1 to 8 kg making contact with the head at velocities ranging from 1 to 12 m/s.[61-63] To reconstruct punch impacts, a lower mass impacting system (compared with the linear impactor) is needed. A pendulum system provides the low mass required to simulate a fist or arm and can be used to deliver temple or jaw impacts to the headform that are common in boxing (Figure 3).

In ice hockey, a slap shot to the helmet or face mask can cause concussion.[64] Similarly in baseball, a player's helmet being struck by a pitch may also cause concussive injury or more severe injuries including skull fractures and intracranial hemorrhage.[65] Head impacts as a result of high-velocity projectiles are characterized by a low-mass object traveling at velocities upward of 30 m/s and 42 m/s for puck impacts in elite ice hockey and pitches in Major League Baseball, respectively.[64,65] In the laboratory, reconstruction of projectile impacts can be done using a ball or puck launcher. This type of system uses compressed air to project a puck or ball at a headform. Figure 4 illustrates the puck launcher system directed at a Hybrid III headform fitted with a hockey helmet to reconstruct a puck impact to the head.

Each of the injury events that cause concussion in sports is defined by unique impact characteristics that influence how mechanical trauma injures the brain. A summary of each injury event and the associated impact characteristics and velocities are presented in Table 2. In reality, it is not likely that 1 mechanism

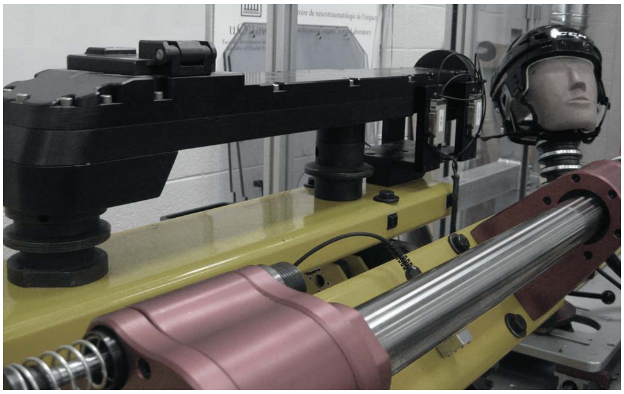

FIGURE 4. *Projectile launcher setup for reconstructing puck-impact events.*

TABLE 2. Summary of Injury Events and the Associated Impact Characteristics

Impact Event	Impact Characteristics
Falls	Mass of the head falling onto rigid surfaces (ice, concrete) at 6 m/s
Collisions	The head colliding with padded shoulders, elbows, or the helmeted head of an opponent (13-16 kg) at 4-11 m/s
Punches[a]	The relatively lower mass of the fist and arm (1-8 kg) impacting the head at 1-12 m/s
Projectiles	Very low mass objects (puck or ball) impacting the head at higher velocities ranging from 30 to 42 m/s

[a]The component of rotational acceleration is dependent on the delivery and direction of the applied force.

of injury can adequately describe all concussions occurring in sports. Thus, it is not appropriate for helmet standards to evaluate head protection with linear drop systems that only represent fall-type events.

Dynamic Response of Each Event

The specific set of impact conditions governing each injury event described will result in characteristic head motions that influence the risk of brain injury.[25] Head motion can be described using dynamic response that includes linear and rotational accelerations.[54] This has an effect on the resulting brain tissue stress and strain calculated using finite-element analysis because tissue deformation values are determined from the *x*, *y*, and *z* linear and rotational acceleration-time histories.[66]

A key aspect to understanding the biomechanics of concussion is characterizing the relationship between impact conditions on head dynamic response and brain tissue deformation and examining their effect on risk of injury. This relationship includes varying the impact parameters such as impact compliance,[67] impact angle,[18] impact mass,[19] and a combination of location and angle resulting in centric and noncentric impacts and has been studied with both unhelmeted[14] and helmeted headforms.[54,68] Centric impacts are those that are through the center of gravity of the headform and noncentric are not through the center of gravity. Research examining specific mechanisms of injury includes ice hockey impacts,[61,64] injuries from falling,[56-59] and collisions in American football.[12,49,69] Because the head can be impacted in a number of different ways, each impact creates a unique head acceleration profile that contributes to varying the level of risk of injury.

A comparison of acceleration-time curves and peak values of acceleration for head impacts as a result of a fall, shoulder, punch, and a puck from Kendall et al[61] and Hoshizaki[70] are presented in Figure 5 for linear acceleration, in Figure 6 for rotational acceleration, and in Table 3 for peak values. The acceleration loading curves obtained from the head impacts were used as input into a finite-element model of the brain in which tissue strain was calculated; peak values are also presented in Table 3.[61,70] When

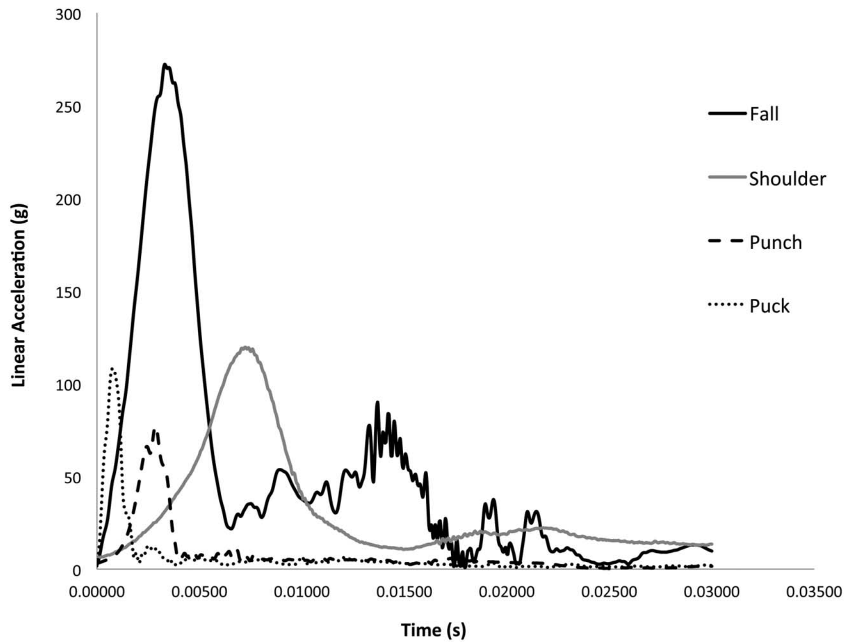

FIGURE 5. *Linear acceleration-time curves of fall, shoulder, punch, and puck impacts.*

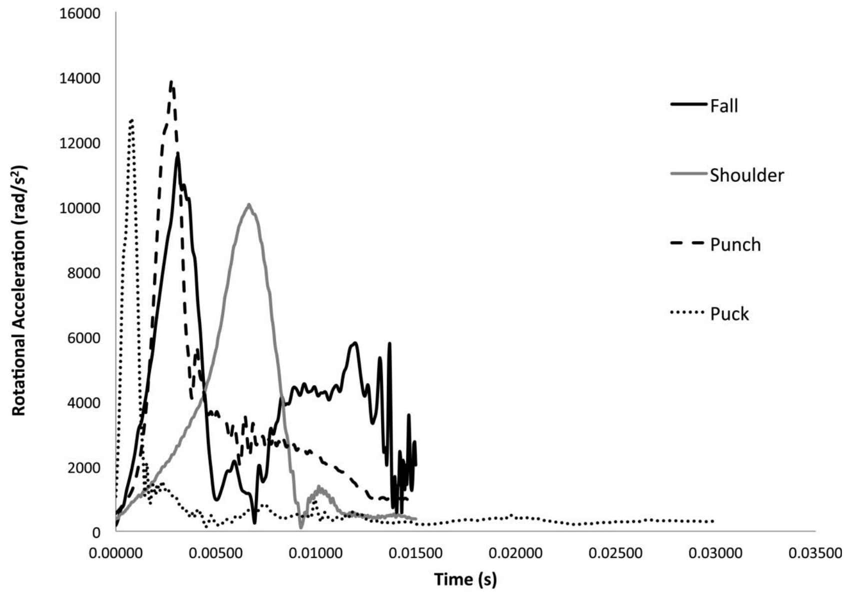

FIGURE 6. *Rotational acceleration-time curves of fall, shoulder, punch, and puck impacts.*

examining headform linear acceleration over time (Figure 5), the highest magnitude was a result of a fall compared with other types of impacts (Table 3). Shoulder and puck impacts produce similar peak values of linear acceleration, with punch impacts being slightly lower (Table 3). The duration of the acceleration impulse as well as the time to peak for each impact type varies considerably (Figure 5). Additionally, the peak, duration of impulse, and time to peak also varied for rotational acceleration (Figure 6). The type of impact with the largest peak rotational value was from a punch, whereas fall and puck impacts tend to have comparable values, and shoulder impacts had the lowest value (Table 3). Although the fall, shoulder, and puck impacts tend to have similar peak rotational acceleration (Table 3), the time to peak and duration of the impulses vary (Figure 6). When comparing different types of impacts, characteristics of the acceleration impulse give a better description of each injury mechanism than peak resultant values alone.

TABLE 3. Mean Peak Linear (*g*) and Rotational (rad/s^2) Accelerations and Maximum Principal Strain With 1 SD in Brackets for Each Mechanism of Injury

Impact Event	Peak Resultant Acceleration: Linear (*g*)	Peak Resultant Acceleration: Rotational (rad/s^2)	Brain Deformation: Average MPS
Fall/Ice	264.4 (33.8)	11.204 (1867)	0.424 (0.019)
Collision/shoulder	112.5 (8.6)	9659 (728.5)	0.305 (0.011)
Punch	87.9 (9.8)	14.001 (1003)	0.229 (0.003)
Projectile (puck)	105.6 (14.6)	12.187 (2104)	0.141 (0.009)

When examining peak strain values for each mechanism (Table 3), it is interesting to observe that a single high magnitude of either linear or rotational acceleration does not necessarily mean a high strain value.[54,71] For example, the fall onto ice has the highest value for strain (0.424) and the highest value of linear acceleration (264.4*g*); however, the rotational acceleration was third highest in magnitude (11.204 rad/s^2). Interestingly, a punch impact is associated with the lowest value of linear acceleration (87.9*g*), but the highest value of rotational acceleration (14.001 rad/s^2) and has a strain value that is similar to a shoulder-to-head collision (0.229). In addition, the magnitudes of strain for falling, shoulder collisions, and punch impacts are at a high risk of concussive injury (0.19-0.26 in Table 1), in which puck impacts are associated with a much lower risk.[17] The differences in risk of brain injury represented by the peak acceleration and strain magnitudes for each type of impact support the notion that there are other factors influencing the occurrence of this type of injury that have yet to be elucidated. It is possible that a single criterion value such as peak linear acceleration, as seen in many standards governing head protection, may not sufficiently describe the risk of concussive injury. Different mechanisms of injury may be better defined by separate injury criteria that are unique to the risks present in a particular sport. The uniqueness of individual impact mechanisms influencing head dynamic responses and brain tissue deformation characteristics speaks to the challenge currently faced by helmet standard organizations and manufacturers of managing the risk of concussion across different sports.

Impact Test Protocols to Measure Risk of Concussion

Currently, impact protocols governing head protection use monorail or free drop systems that measure peak linear acceleration (*g*), GSI, and HIC.[40,72-74] Each standard specifies the use of a particular type, size, and mass of headform depending on the size of the helmet. The Snell standard for motorcycle helmets and the Canadian Standards Association standard for ice hockey helmets both use simplified magnesium headforms as defined in the International Standards Organization (ISO) Standard ISO DIS 6220-1983.[75] The NOCSAE standard for testing American football helmets uses a more compliant Hodgson-WSU headform. The Hodgson-WSU headform has more humanoid features with a soft outer surface and a glycerin-filled cavity. Another headform commonly used in head impact research is the 50th percentile adult male Hybrid III headform.[12,14,68] This headform was primarily designed for high-energy automobile crash testing and is composed of steel with a vinyl rubber outer layer simulating the compliance of skin. Each headform is composed of a specific set of materials that have different masses and geometries. Consequently, these characteristics create unique impact responses that make it difficult to compare results using different headforms.[76] Test headforms are designed to provide reliable impact response values while mimicking the dynamic response of the average human head as best as possible. Considering the wide anatomic variation that exists in the human population, developing headforms of all shapes and sizes is not feasible. Thus, these headforms produce unique responses that are specific to the average male head.

The head can get impacted in a variety of locations and angles creating linear or rotationally dominant head impact responses. Research has been undertaken to establish a protocol representative of collision-type impacts using centric and noncentric conditions measuring linear and rotational accelerations.[14,55] A centric impact occurs when the force is through the center of gravity of the headform and the dominant motion is translation of the head and helmet (Figure 7). When a helmet is subject to a centric impact, the helmet liner and shell undergo compression. This type of impact phenomena is characteristic of linear drop systems that are used in current standards to evaluate head protection. In noncentric impacts, the force is applied outside of the center of gravity of the headform and thus causes an initial compression of the shell and liner, after which, the liner will tend to shear to a maximum point where the helmet couples to the headform and the transferred force causes the headform to rotate. This type of impact results in a rotationally dominant response (Figure 8). Current helmet standards have yet to include both centric and noncentric collision-type impacts in the evaluation of head protection in sports.

The NOCSAE standard for baseball helmets and ISO standard for hockey helmets provide a means of evaluating the protection against projectiles. Currently, the NOCSAE standard for baseball helmets calls for a baseball or softball to be launched at a helmeted headform at 55 mph (24 m/s) where peak impact severity is not to exceed 1200 SI units.[77] Although baseball helmets are tested according to how people get injured in the sport, the level of impact may not sufficiently cover the possible range for injury, as one study reports that Major League Baseball players get concussed with baseballs traveling upward to 93 mph (43 m/s).[65] The ISO standard for ice hockey helmets also has a component for projectile impacts. This standard requires pucks to be launched at hockey helmet face protectors at 7 impact sites with velocities ranging from 10 m/s to 36 m/s. The certification criterion only specifies that the puck or helmet is not to touch the no-contact zone outlined on the headform during testing.[78] The range of impact conditions specified in the ISO standard encompasses a range where concussive injury from puck impacts have been reported.[64] Rousseau et al[64] reconstructed puck impacts to the helmeted head using puck velocities between 21 and 30 m/s. To address the risk of concussion in ice hockey resulting from puck impacts, it would be necessary to develop certification criterion that evaluates the ability of the helmet and face protector to mitigate impacts at the concussive level.

All headgear worn in amateur boxing is approved by the corresponding governing body to ensure that the make and materials used in the equipment meet USA Boxing and International Boxing Association requirements. Although organization approval is required, there is no real evaluation of the capacity of boxing headgear to protect against injury. Research has demonstrated that headgear is effective at reducing linear and rotational accelerations of the head from punch impacts.[79,80] Despite these findings, the International Boxing Association has recently banned the use of headgear in competition in an attempt to decrease the incidence of concussion in boxing because preliminary findings have shown a lower rate of concussion for those not wearing headgear compared with those who do.

A novel impact test rig for oblique impacts has been developed by Aare and Halldin.[81] This test rig protocol combines a linear drop with an applied tangential force by dropping a helmeted headform onto a linearly accelerated striking anvil. This type of tangentially applied force is demonstrated schematically in Figure 9. The mechanism of this type of impact can be representative of a subject who falls off a bike while riding and hits his or her head. Current bicycle helmet standards only measure the protective capacity of a helmet using a linear drop system onto a steel anvil.[82] These linear drops may not effectively evaluate the ability of bicycle helmets to protect against concussive injuries because it does not accurately represent the mechanism for concussion in cycling. Because the standard only uses vertical drop velocity to determine severity of impact, it lacks the tangential velocity that would be present for cycling accidents. As a result, using a vertical drop method to assess the risk of brain injury may underestimate the resulting dynamic response because it ignores the influence of the tangentially applied velocity of the rider while biking.

To develop more effective interventions for concussive injury, it is important to understand the mechanism of how concussions occur in sports. Various sports are defined by unique injury mechanisms that present different risks for concussion. The diversity of these injury mechanisms creates a daunting challenge

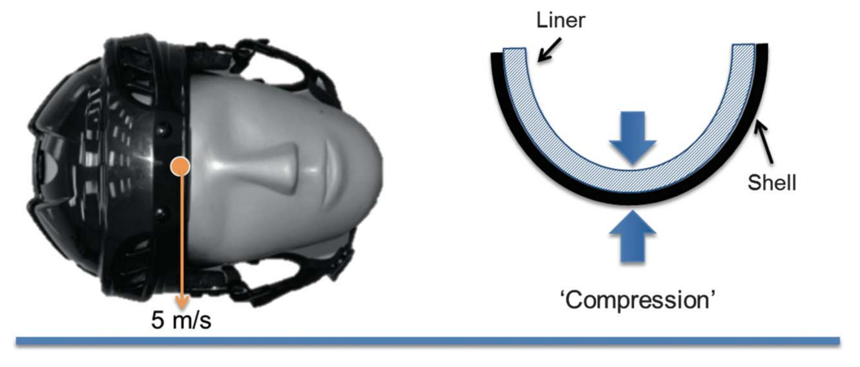

FIGURE 7. *Schematic of a centric impact-inducing linear dominant motion.*

for researchers interested in effectively addressing head injuries in all sports. Current helmet certification testing only involves 1 mechanism of injury, falling to the ground, while neglecting all other types of impacts known to cause concussion. More comprehensive testing protocols linked to the mechanisms of brain injury would aid in the improvement of helmets that are optimized for protection against sport-specific concussive impacts.

Types of Helmet Technologies

There are many types of sport helmets on the market; however, they typically fit into1 of 2 categories: single-impact helmets (or crash helmets) and multiple-impact helmets.[26,83] These 2 distinct categories of helmet types vary in construction; however, they are both assessed using linear acceleration or some integration of the linear acceleration curve such as the GSI or HIC.[26,83] This means that although mechanically the helmets manage the translations induced from an impact differently, they are not designed to manage or mitigate the magnitude of rotation.

Single Impact Helmets

These helmets are typically used in events where the head is likely to be impacted once at high energy (motorsports, bike, alpine ski). The single-impact helmet is typically constructed of an outer shell designed to help distribute the force of the impact and an inner energy absorbing liner.[26,83] The energy absorbing liner is often made of a material such as expanded polystyrene (EPS) that deforms plastically under impact and thus releases energy.[84] This means that after 1 impact, the helmet is compromised and lacks much of its original ability to manage impact energies. In terms of concussion, EPS liners do very little to manage accelerations below concussion thresholds and are particularly poor at reducing rotations associated with concussions. EPS is, however, very effective at reducing linear translations for high-energy impacts relating to the risk of TBI.[83,84]

Multiple Impact Helmets

These helmets are similar in construction to single-impact helmets in that they are composed of an outer shell designed to spread out the force of impact and engage as much of the energy-absorbing liner underneath during an impact. The energy-absorbing liner typically comprises either VN or expanded polyprolylene (EPP) foam. VN foam is excellent at managing low-energy multiple impacts but tends not to perform as well for higher energy impacts.[83,84] This type of foam returns to its original shape after

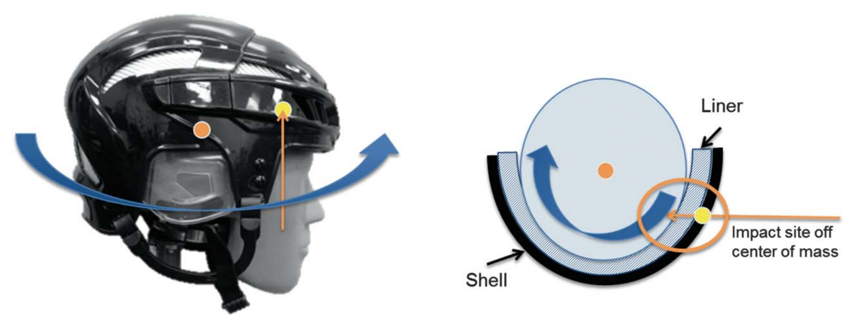

FIGURE 8. *Schematic of a noncentric impact-inducing rotationally dominant motion on a helmeted headform.*

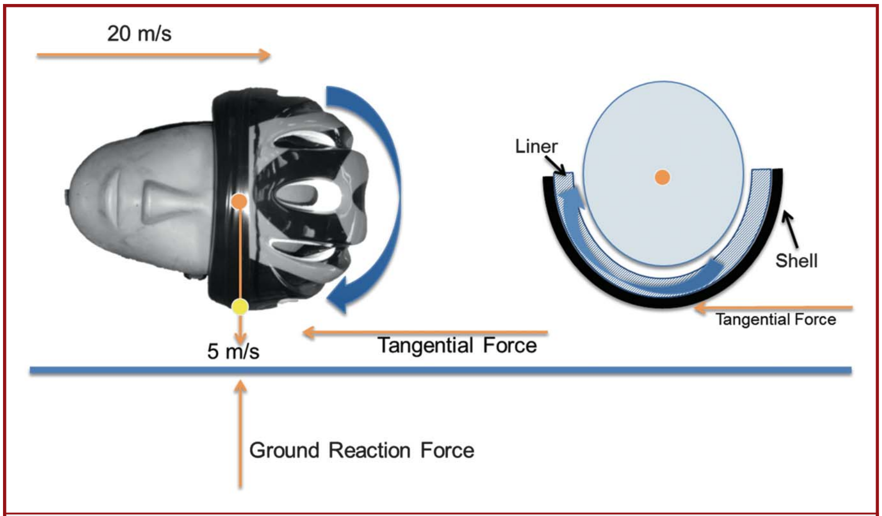

FIGURE 9. *Schematic of a fall with a high-velocity tangential component.*

an impact. EPP is similar to EPS in its manufacturing process but has more elastic properties.[26,83] As a result, the EPP deforms under the impact, dissipating the impact energy returning to its original shape. As it can manage higher energies more effectively than VN foam, it is thought to be better for multi-impact applications. It should be noted that EPP tends to degrade faster than its VN counterpart.[83,84] More recent research examined the ability of these 2 foams to manage rotational acceleration, reporting that VN has an improved capacity to manage this type of impact-induced motion.[54,64,85,86]

Energy-Absorbing Structures

The energy-absorbing protective capacity of VN, EPS, and EPP foams in helmets are primarily related to their density and thickness.[26,84,87] Typically, the thicker the material is, the better the protection, and the denser the material is, the higher the impact energy that it is designed to manage.[88] Energy-absorbing structures are a more recent innovation in helmet design because they use a combination of material structural characteristics such as geometry, material stiffness, and type to mitigate impact energy. These energy-absorbing structures add an additional dimension to optimize helmet performance as they allow the rate of material compression to be controlled.[83,87,89-91] The increased number of design parameters creates an opportunity to engineer structures that potentially manage a wider range of impact energies.[90,91] Engineered structures are a recent development in helmet technology, with only a few helmets currently using this kind of technology. In American football, the Xenith X1 series helmets (Xenith, LLC, Lowell, Massachusetts) use an air-filled 3-dimensional structure to manage impacts, and the Schutt DNA series helmets (Schutt Sports Inc, Litchfield, Illinois) also use a type of engineered structure. In ice hockey and lacrosse helmets, Cascade helmets (Cascade, Liverpool, New York) uses 3-dimensional type structures. Engineered structures provide an increased opportunity to innovate systems designed to manage the motions more closely linked to the mechanism of concussion.

Future Innovations in Sport Helmets to Manage Concussion

There have been a few technologies developed purporting to manage rotational acceleration in motocross, alpine, and bike helmets. In motocross helmets, 6D helmets (6 Degrees of Freedom, Brea, California) have developed the ATR-1 helmet that manages rotational acceleration through a series of rubber dampers installed between EPS foam liners. Also in motocross and alpine and bike helmets, MIPS AB (Stockholm, Sweden) has developed the Multi-Directional Impact Protection System, a suspension system that helps to dissipate the tangential impact forces that produce rotational acceleration from falls. In ice hockey, Bauer Performance Sports Ltd (Exeter, New Hampshire) has developed a helmet that is described to reduce the rotational acceleration from impacts using a suspension system similar to the one developed by MIPS AB. All of these companies claim to have innovated mechanisms designed to reduce rotational acceleration incurred from impacts. However, currently, there is no standard method of testing these new technologies to determine helmet performance in terms of reducing rotational acceleration. Until standards for sport helmets measure the performance of the rotation damping technologies, the effectiveness of these innovations will lack objective validation.

SUMMARY

Concussion has been shown to be complex in nature and is not easily described using engineering parameters. There are many types of injury mechanisms associated with concussion depending on the impact conditions; therefore, a single mechanism cannot effectively describe the risk of injury for all concussions occurring in sports. Concussive impacts to the head not only create a range of dynamic responses, but likely involve different parts of the brain. Suggesting that a single measurement criterion is adequate to predict the risk of all types of concussions is wishful thinking because the symptoms of concussion are wide ranging and the ability to define the prognosis of the resulting disability complicated. We can, however, conclude that present-day helmets are not specifically designed to protect against concussions because helmet safety standards are also not developed for this purpose.

Early research describing the dynamic response of the head resulting from direct impacts identified both linear and rotational accelerations as important to measuring the risk of injury.[28,53] More recent investigations have expanded the original research of Gurdjian et al[8] and continue to develop more refined and precise measures of concussion. It is essential that head protection organizations take the initiative by developing standards designed to address the mechanisms of concussive injury in specific sports. It is also important for helmet manufacturers to support this initiative through improvements in the safety of helmets through specific innovations that meet the unique demands of each sport. There have been recent innovations in helmet technology claiming to improve protection against the risk of concussion that involves systems designed to manage rotational acceleration. Research describing the relationship between rotational acceleration and the risk of concussive injuries reinforces the value of this strategy.[11,42] In the absence of a standard designed to evaluate the ability of a helmet to manage rotational acceleration, these innovations claiming to protect against concussions cannot be validated.

Future efforts for reducing concussive injuries in sports include advances in testing methods, more appropriate measurement variables for risk, and improved concussion diagnosis. Finite-element analysis also shows potential as a tool for designing safer helmets. It must be recognized, however, that each finite-element model is unique in geometry and material characteristics with data specific to the model used to calculate brain trauma values. Similarly, the dynamic response data obtained are specific to the impact characteristics of that particular headform. Efforts to improve identification of injured athletes include helmets with concussion-warning systems using accelerometers to provide an alert relating the level of risk of head impacts. Challenges still exist in accounting for the interaction of the head and helmet-imbedded accelerometers during an impact that results in inconsistent acceleration values. It is important to understand that helmets provide excellent protection for catastrophic injuries, and although some helmets provide some protection against concussive injuries, to date, they have not been particularly effective. Although there is much research to be done targeting sports-related concussive injuries, recent advances in measurement techniques and innovative energy management technologies provide an optimistic future for managing the risk of injury.

Disclosure

The authors have no personal, financial, or institutional interest in any of the drugs, materials, or devices described in this article.

REFERENCES

1. Bennett T, ed. *The NFL's Official Encyclopedia History of Professional Football*. 2nd ed. New York, NY: Macmillan; 1977.
2. Roper WW. *Football: Today and Tomorrow*. New York: Duffield and Co.; 1927: pp. 155-156.
3. Cantu RC. *Neurologic Athletic Head and Spine Injuries*. Philadelphia, PA: WB Saunders Company; 2000:pp. 124-131.
4. Clement L, Jones D. Research and development of hockey protective equipment: a historical perspective. Safety in Ice Hockey. Philadelphia, American Society for Testing and Materials. 1989;164-186.
5. Gurdjian ES, Roberts VL, Thomas LM. Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury. *J Trauma.* 1966;6(5):600-604.
6. Ommaya AK. Trauma to the nervous system. *Ann R Coll Surg Engl.* 1966;39(6): 317-347.
7. Gennarelli TA, Thibault LE, Adams JH, Graham DI, Thompson CJ, Marcincin RP. Diffuse axonal injury and traumatic coma in the primate. *Ann Neurol.* 1982;12(6):564-574.
8. Gurdjian ES, Webster JE, Lissner HR. Observations on the mechanism of brain concussion, contusion, and laceration. *Surg Gynecol Obstet.* 1955;101(6):680-690.
9. Bain AC, Meaney DF. Tissue-level thresholds for axonal damage in an experimental model of central nervous system white matter injury. *J Biomech Eng.* 2000;122(6):615-622.
10. Kleiven S. Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure. *Int J Crashworthiness.* 2006;11(1):65-79.
11. Forero Rueda MA, Cui L, Gilchrist MD. Finite element modelling of equestrian helmet impacts exposes the need to address rotational kinematics in future helmet designs. *Comput Methods Biomech Biomed Engin.* 2011;14(12):1021-1031.
12. Pellman EJ, Viano DC, Tucker AM, Casson IR, Waeckerle JF. Concussion in professional football: reconstruction of game impacts and injuries. *Neurosurgery.* 2003;53(4):799-812.
13. Hutchison MG. *Concussions in the National Hockey League (NHL): The Video Analysis Project [PhD]*. Ann Arbor, MI: University of Toronto (Canada); 2011.
14. Walsh ES, Rousseau P, Hoshizaki TB. The influence of impact location and angle on the dynamic impact response of a hybrid III headform. *Sports Eng.* 2011;13(3): 135-143.
15. Willinger R, Baumgartner D. Human head tolerance limits to specific injury mechanisms. *Int J Crashworthiness.* 2003;8(6):605-617.
16. Horgan TJ, Gilchrist MD. The creation of three-dimensional finite element models for simulating head impact biomechanics. *Int J Crashworthiness.* 2003;8(4):353-366.
17. Kleiven S. Predictors for traumatic brain injuries evaluated through accident reconstructions. *Stapp Car Crash J.* 2007;51:81–114.
18. Oeur A. *An Analysis of Head Impact Angle on the Dynamic Response of a Hybrid III Headform and Brain Tissue Deformation [Master's]*. Ottawa, Canada: Human Kinetics, University of Ottawa; 2012.
19. Karton C. *The effect of inbound mass on the dynamic response of the Hybrid III headform and brain tissue deformation* [Master's]. Ottawa, Canada: Human Kinetics: University of Ottawa; 2012.
20. Post A, Oeur A, Hoshizaki B, Gilchrist MD. The influence of velocity on injury risk in American football. Paper presented at: Helmet Performance and Design; 2013; London, England.
21. Daneshvar DH, Nowinski CJ, McKee AC, Cantu RC. The epidemiology of sport-related concussion. *Clin Sports Med.* 2011;30(1):1-17.
22. Gessel LM, Fields SK, Collins CL, Dick RW, Comstock RD. Concussions among United States high school and collegiate athletes. *J Athl Train.* 2007;42 (4):495-503.
23. Wennberg RA, Tator CH. National Hockey League reported concussions, 1986-87 to 2001-02. *Can J Neurol Sci.* 2003;30(3):206-209.
24. Casson IR, Viano DC, Powell JW, Pellman EJ. Twelve years of national football league concussion data. *Sports Health.* 2010;2(6):471-483.
25. Post A, Hoshizaki TB. Mechanisms of brain impact injuries and their prediction: a review. *Trauma*. 2012;14(4):327-349.
26. Hoshizaki TB, Brien SE. The science and design of head protection in sport. *Neurosurgery*. 2004;55(4):956-966; discussion 966-967.
27. Haddad BF, Lissner HR, Webster JE, Gurdjian ES. Experimental concussion; relation to acceleration to physiologic effect. *Neurology*. 1955;5(11):798-800.

28. Thomas LM, Roberts VL, Gurdjian ES. Experimental intracranial pressure gradients in the human skull. *J Neurol Neurosurg Psychiatry.* 1966;29(5):404-411.
29. Got C, Patel A, Fayon A, Tarriere C, Walfisch G. Results of experimental head impact on cadavers: the various data obtained and the relation to some measured physical parameters. Paper presented at: 22nd Stapp Car Crash Conference; 1978; Ann Arbor, Michigan.
30. Hodgson VR. Tolerance of the facial bones to impact. *Am J Anat.* 1967;120(1): 113-122.
31. Nahum A, Ward C, Raasch F, Adams S, Schneider D. Experimental studies of side impact to the human head. Paper presented at: 24th Stapp Car Crash Conference; 1980; Troy, Michigan.
32. Schneider D, Nahum A. Impact studies of facial bones and skull. Paper presented at: 16th Stapp Car Crash Conference; 1972; Detroit, Michigan; 186-203.
33. Stalnaker R, Melvin J, Nuscholtz G, Alem N, Benson J. Head impact response. Paper presented at: 21st Stapp Car Crash Conference; 1977; New Orleans, Louisiana; 305-335.
34. Yoganandan N, Pintar FA, Sances A Jr, et al. Biomechanics of skull fracture. *J Neurotrauma.* 1995;12(4):659-668.
35. Newman JA. Head injury criteria in automotive crash testing. Paper No. 801317. Paper presented at: 20th Stapp Car Crash Conference: Society of Automotive Engineers Technical; 1980; Troy, Michigan.
36. Ono K, Kikuchi A, Nakamura M, Kobayashi H, Nakamura N. Human head tolerance to sagittal impact reliable estimation deduced from experimental head injury using subhuman primates and human cadaver skulls. Paper presented at: 24th Stapp Car Crash Conference; 1980; Troy, Michigan; 101-160.
37. Gadd CW. Use of a weighted impulse criterion for estimating injury hazard. Paper presented at: 10th Stapp Car Crash Conference; 1966; Hollomon Air Force Base, New Mexico.
38. Versace J. A review of the severity index. Paper presented at: 15th Stapp Car Crash Conference; 1971; Coronado, California.
39. National Highway Traffic Safety Administration, Department of Transportation (DOT). *49 CFR Parts 552, 571, 585 and 595 Federal Motor Vehicle Safety Standards*. Occupant Crash Protection; 2011; Washington, DC.
40. Department of Transportation (DOT). *Federal Motor Vehicle Safety Standard No. 218.49 CFR 571.218.* Washington, DC: US Department of Transportation, National Highway Traffic Safety Administration; 1988.
41. Nahum AM, Smith R. An experimental model for closed head impact. Paper presented at: 20th Stapp Car Crash Conference: SAE Paper No. 760825; 1976; Dearborn, Michigan.
42. Adams JH, Graham DI, Gennarelli TA. Acceleration induced head injury in the monkey. II. Neuropathology. *Acta Neuropathol Suppl.* 1981;7(Suppl. VII):26-28.
43. Gennarelli TA, Adams JH, Graham DI. Acceleration induced head injury in the monkey.I. The model, its mechanical and physiological correlates. *Acta Neuropathol Suppl.* 1981;7(Suppl. VII):23-25.
44. Goldsmith W. Current controversies in the stipulation of head injury criteria. *J Biomech.* 1981;14(12):883-884.
45. Prange MT, Meaney DF, Margulies SS. Defining brain mechanical properties: effects of region, direction and species. Paper presented at: 44th Stapp Car Crash Conference; 2002; Atlanta, Georgia.
46. Ommaya AK. Mechanical properties of tissues of the nervous system. *J Biomech.* 1968;1(2):127-138.
47. Ommaya AK, Hirsch AE. Tolerances for cerebral concussion from head impact and whiplash in primates. *J Biomech.* 1971;4(1):13-21.
48. King AI, Yang KH, Zhang L, Hardy W, Viano DC. Is head injury caused by linear or angular acceleration. Paper presented at: IRCOBI conference; 2003; Lisbon, Portugal.
49. Zhang L, Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury. *J Biomech Eng.* 2004;126(2):226-236.
50. Silva VD. *Mechanics and Strength of Materials*. New York, NY: Springer Verlag; 2006.
51. Rashid B, Destrade M, Gilchrist MD. Mechanical characterization of brain tissue in tension at dynamic strain rates. *J Mech Behav Biomed Mater.* 2013;28:71-85.
52. Rashid B, Destrade M, Gilchrist MD. Mechanical characterization of brain tissue in compression at dynamic strain rates. *J Mech Behav Biomed Mater.* 2012;10:23-38.
53. Ommaya AK, Gennarelli TA. Cerebral concussion and traumatic unconsciousness. Correlation of experimental and clinical observations of blunt head injuries. *Brain*. 1974;97(4):633-654.
54. Post A, Oeur A, Hoshizaki B, Gilchrist MD. Examination of the relationship between peak linear and angular accelerations to brain deformation metrics in hockey helmet impacts. *Comput Methods Biomech Biomed Engin.* 2013;16(5): 511-519.
55. Post A, Oeur A, Hoshizaki B, Gilchrist MD. An examination of American football helmets using brain deformation metrics associated with concussion. *Mater Des.* 2013;45(0):653-662.
56. O'Riordan K, Thomas PM, Phillips JP, Gilchrist MD. Reconstruction of real world head injury accidents resulting from falls using multibody dynamics. *Clin Biomech (Bristol, Avon).* 2003;18(7):590-600.
57. Doorly MC, Gilchrist MD. The use of accident reconstruction for the analysis of traumatic brain injury due to head impacts arising from falls. *Comput Methods Biomech Biomed Engin.* 2006;9(6):371-377.
58. Post A. *The influence of dynamic response characteristics on traumatic brain injury* [PhD Dissertation]. Ottawa, Canada: Human Kinetics: University of Ottawa; 2013.
59. Oeur A, Karton C, Post A, et al. An examination of headform dynamic response for concussive and traumatic brain injuries. Paper presented at: Helmet Performance and Design; 2013; London, England.
60. Fréchède B, McIntosh AS. Numerical reconstruction of real-life concussive football impacts. *Med Sci Sports Exerc.* 2009;41(2):390-398.
61. Kendall M, Walsh ES, Hoshizaki TB. A comparison of dynamic impact response and brain deformation metrics of head impact reconstructions for three mechanisms of head injury in ice hockey. Paper presented at: International Research Council on the Biomechanics of Impact (IRCOBI); 2012.
62. Walilko TJ, Viano DC, Bir CA. Biomechanics of the head for Olympic boxer punches to the face. *Br J Sports Med.* 2005;39(10):710-719.
63. Atha J, Yeadon MR, Sandover J, Parsons KC. The damaging punch. *Br Med J (Clin Res Ed).* 1985;291(6511):1756-1757.
64. Rousseau P, Hoshizaki B, Gilchrist MD. A reconstruction of a puck impact to the head resulting in concussion in ice hockey. Paper presented at: Symposium on the Mechanics of Concussion in Sports, ASTM International; 2012; Atlanta, Georgia.
65. Athiviraham A, Bartsch A, Mageswaran P, et al. Analysis of baseball-to-helmet impacts in major league baseball. *Am J Sports Med.* 2012;40(12):2808-2814.
66. Post A, Hoshizaki B, Gilchrist MD. Finite element analysis of the effect of loading curve shape on brain injury predictors. *J Biomech.* 2012;45(4):679-683.
67. Dawson L, Post A, Hoshizaki B. The influence of material compliance on the dynamic response of a hybrid III headform. Paper presented at: Annual Conference and Committee Week; 2013; Calgary, Canada.
68. Post A, Oeur A, Walsh E, Hoshizaki B, Gilchrist MD. A centric/non-centric impact protocol and finite element model methodology for the evaluation of American football helmets to evaluate risk of concussion. *Comput Methods Biomech Biomed Engin.* 2014;17(16):1785-1800.
69. Zanetti K, Post A, Marshall M, Karton C, Hoshizaki B, Gilchrist MD. Identifying injury characteristics for three player positions in American football using physical and finite element modeling reconstructions. Paper presented at: International Research Council on the Biomechanics of Injury (IRCOBI); 2013; Gothenburg, Sweden.
70. Hoshizaki TB. The use of technology for the study of research in head trauma and understanding chronic traumatic encephalopathy. Include the hits system, the impact signal systems, impact reconstruction (estimating impact parameters, using dummy heads,) FEA modeling. Paper presented at: National conference on Chronic Traumatic Encephalopathy; October 2012; Las Vegas, USA.
71. Nishizaki K, Marino W, Hoshizaki B, Post A, Oeur A, Walsh ES. Evaluation of dynamic response and brain deformation metrics using a centric/non-centric impact protocol using a monorail impacting system. In: Ashare, A. & M. Ziejewski (eds). The Mechanics of Concussion in Sports, STP 155. West Conshohocken, PA: ASTM International. doi: 10.1520/STP155220120169.
72. National Operating Committee on Standards for Athletic Equipment (NOCSAE). Standard Performance Specification for Newly Manufactured Football Helmets. NOCSAE DOC (ND)002-12m-13. Overland, Kansas; 2013.
73. Canadian Standards Association (CSA). *CAN/CSA-Z262.1–M90 Ice Hockey Helmets*. Toronto, Canada: A National Standard for Canada; 2009.
74. Snell Memorial Foundation. *2010 Standard for Protective Headgear for Use With Motorcycles and Other Motorized Vehicles*. North Highlands, CA: Snell Memorial Foundation, Inc.; 2010.
75. International Standards Organization (ISO). *DIS 6220-1983. Headforms for use int he testing of protective helmets*. 1983; Geneva, Switzerland.
76. Kendall M, Walsh ES, Hoshizaki TB. Comparison between Hybrid III and Hodgson–WSU headforms by linear and angular dynamic impact response. Paper presented at: Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology; March 1, 2012; Dublin, Ireland.

77. National Operating Committee on Standards for Athletic Equipment (NOCSAE). *Standard Performance Specification for Newly Manufactured Baseball/Softball Batter's Helmets*. NOCSAE DOC (ND)002-10m-12. 2012; Overland, Kansas.
78. International Standards Organization (ISO). *ISO/EN 10256: 2006-03-10 (E). Head and face protection for use in ice hockey*. 2006; Geneva, Switzerland.
79. Dau N, Chien HC, Sherman D, Bir C. Effectiveness of boxing headgear for limiting injury. Paper presented at: American Society of Biomechanics; 2006; Virginia, USA.
80. Bartsch AJ, Benzel EC, Miele VJ, Morr DR, Prakash V. Boxing and mixed martial arts: preliminary traumatic neuromechanical injury risk analyses from laboratory impact dosage data. *J Neurosurg.* 2012;116(5):1070-1080.
81. Aare M, Halldin P. A new laboratory rig for evaluating helmets subject to oblique impacts. *Traffic Inj Prev.* 2003;4(3):240-248.
82. Consumer Product Safety Commission (CSPC). Part II: 16 CFR Part 1203: Safety Standard for Bicycle Helmets; Final Rule 1998. Bethesda, Maryland.
83. Post A, Hoshizaki B, Brien S. *Head Injuries, Measurement Criteria, and Helmet Design. Handbook of Ergonomics in Sport and Exercise*. New York: Routledge Publishers; 2013.
84. Hoshizaki B. Engineering head protection. In: Cantu RC, ed. *Neurologic Athletic Head and Spine Injuries*. Philadelphia, PA: WB Saunders; 2000.
85. Hoshizaki TB, Walsh E, Post A, et al. The application of brain tissue deformation values in assessing the safety performance of ice hockey helmets. Paper presented at: Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology; 2012; Dublin, Ireland.
86. Walsh ES, Post A, Rousseau P, et al. Dynamic impact response characteristics of a helmeted Hybrid III headform using a centric and non-centric impact protocol. *Proc Inst Mech Eng P.* 2012;226(3-4):220-225.
87. Gimbel G, Hoshizaki TB. A comparison between vinyl nitrile foam and new air chamber technology on attenuating impact energy for ice hockey helmets. *Int J Sports Sci Eng.* 2008;2(3):154-161.
88. Gimbel GM, Hoshizaki TB. Compressive properties of helmet materials subjected to dynamic impact loading of various energies. *Eur J Sport Sci.* 2008;8(6):341-349.
89. Lamb L, Hoshizaki TB. Deformation mechanisms and impact attenuation characteristics of thin-walled collapsible air chambers used in head protection. *Proc Inst Mech Eng H.* 2009;223(8):1021-1031.
90. Lamb L, Post A, Hoshizaki B. Development of a new methodology capable of characterizing the contribution of air flow velocity to impact attenuation in chamber structures. *J ASTM Int.* 2009;6(1). doi: 10.1520/JAI101884.
91. Hoshizaki B, Post A. Impact attenuation characteristics of thin walled collapsible air chambers for use in protection helmets. *J ASTM Int.* 2009;6(4). doi: 10.1520/JAI101883.