Evaluation and Application of Brain Injury Criteria to Improve Protective Headgear Design

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Abstract

As many as 3.8 million sports-related traumatic brain injuries (TBIs) occur each year, nearly all of which are mild or concussive. These injuries are especially concerning given recent evidence that repeated concussions can lead to long-term neurodegenerative processes. One way of reducing the number of injuries is through improvements in protective equipment design. Safety standards and relative performance ratings have led to advancements in helmet design that have reduced severe injuries and fatalities in sports as well as concussive injuries. These standards and evaluation methods frequently use laboratory methods and brain injury criteria that have been developed through decades of research dedicated to determining the human tolerance to brain injury. It is necessary to determine which methods are the most appropriate for evaluating the performance of helmets and other protective equipment. Therefore, the aims of this research were to evaluate the use of different brain injury criteria and apply them to laboratory evaluation of helmets. These aims were achieved through evaluating the predictive capability of different brain injury criteria and comparing laboratory impact systems commonly used to evaluate helmet performance. Laboratory methods were developed to evaluate the relative performance of hockey helmets given the high rate of concussions associated with the sport. The implementation of these methods provided previously unavailable data on the relative risk of concussion associated with different hockey helmet models.

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Chapter 1

Introduction and Research Objectives

Concussion has gained increasing recognition as a public health problem. Approximately 87% of the estimated 2.5 million traumatic brain injuries (TBIs) that occur in the US each year are considered mild (mTBI). These incidence rates represent a 45% increase since 2007 [1]. Additionally, these are likely gross underestimates of incidence as they are based on emergency department visits and hospitalizations, which represent a small percentage of the total number of patients that seek health care after a concussion [2]. Accounting for those who do not seek medical care after a concussion, it has been estimated that as many as 3.8 million sports-related TBIs occur each year [3]. While concussions are most frequently studied in athlete populations, they are a major burden for the general public in recreational activities and motor vehicle accidents, as well as military personnel [4-7]. However, sports teams provide a controlled environment in which epidemiologic and biomechanical data can be collected for a population at a higher risk for head injuries [8]. Findings from athlete populations can then be applied to protective equipment design and safety systems for other populations [8].

Of all contact sports, football tends to get the most attention regarding concussions. However, the rate of concussion is higher in both men's and women's ice hockey (Figure 1) [9, 10]. The difference in concussion rates may be even greater in terms of concussions per number of head impacts, given that football players have a much higher exposure to head impacts than hockey players [11, 12]. A survey of collegiate athlete injuries showed that diagnosed concussion rates doubled over a 15 year period, emphasizing the need to mitigate concussions in sports [10]. There are several strategies that can be employed to reduce the number of concussions in sports.

These include rule changes, proper playing technique, and improvements in protective equipment design [13-18]. The focus of the following chapters is on improvements in protective headgear through laboratory evaluation using previously developed brain injury criteria. However, it is important to first understand the pathophysiology and biomechanics of head injury.

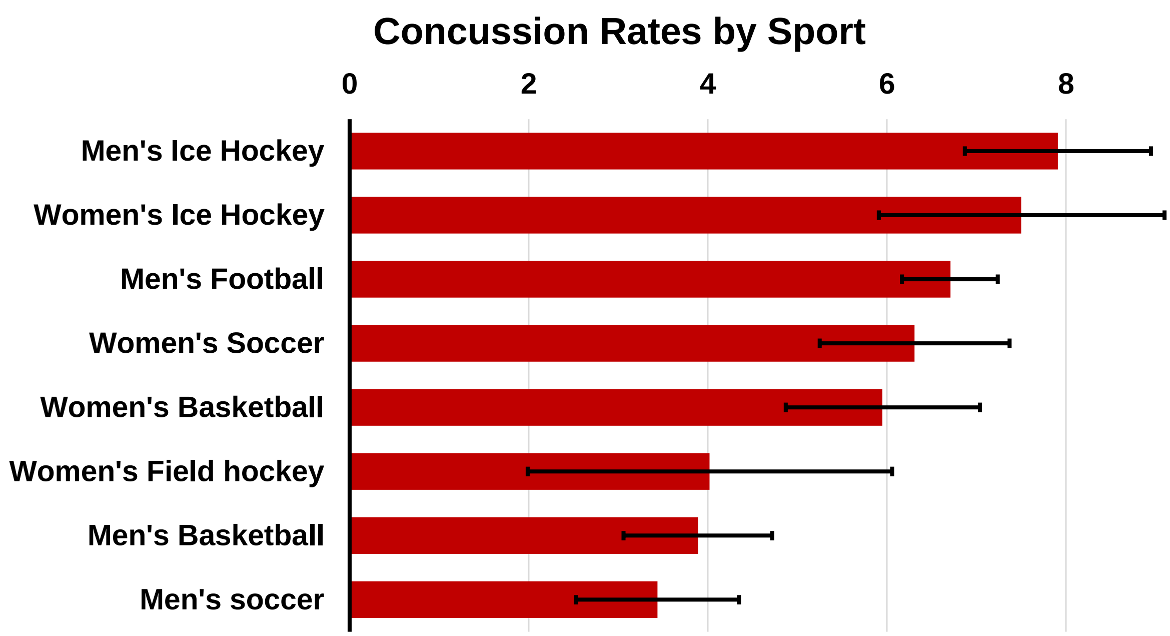

Figure 1: Concussion rates for collegiate athletes per 10,000 athlete-exposures. An athlete-exposure is defined as 1 athlete participating in 1 game or practice. Error bars represent the 95% confidence interval. Both men’s and women’s ice hockey have a higher rate of concussion than football [9].

Concussion Definition and Diagnostics

During a head impact that rapidly accelerates or decelerates the skull, brain motion lags behind that of the rigid skull due to its own inertia. There is relative motion between the brain and skull that causes strains within the brain [19, 20]. This motion is small, on the order of 5-7 mm, but is enough to cause injury. There is some experimental evidence to suggest that in the case of concussions, commonly injured structures in the brain include the centrum semiovale, corpus callosum, and the internal capsule [21]. These are also regions of the brain that are commonly involved in more severe diffuse axonal injuries. The dorsolateral prefrontal cortex has also been indicated as the cause for certain decreases in cognitive function after brain injury [22]. These regions are in agreement with a previously proposed "centripetal theory of cerebral concussion," which states that the strains within the brain are always greater towards the periphery, and only affect deep structures in the brain with more severe impacts when loss of consciousness occurs [23].

Concussion is a difficult injury to define. It is considered a mild subset of traumatic brain injury (TBI) induced by biomechanical forces, with transient characteristic signs and symptoms that may or may not include loss of consciousness [24]. Other potential signs and symptoms are headache, confusion, memory loss, and incoordination [25]. Although there are common features of concussions, they vary greatly depending on both the magnitude of the insult and the regions of the brain most affected [26]. In addition to variations in signs, the altered mental state of the patient can make diagnosis and classification of the injuries challenging [27]. Because the signs of concussion can be subtle, identification often relies on an athlete reporting the injury. Several studies that employed either retrospective surveys or prospective observation methods found that the actual number of concussions ranged anywhere from 2 to greater than 30 times reported rates [28-31]. The most common reason that players cited for not reporting a concussion was that they did not think it was a serious injury [30]. These studies demonstrate the importance of concussion education in athletes starting at an early age when they are first exposed to head injury risks.

Diagnosis of concussion can involve a combination of sideline assessments, symptom scales, balance tests, and neurocognitive tests [32]. The Sport Concussion Assessment Tool 3 (SCAT3) is a commonly used sideline assessment that includes the Glasgow Coma Scale (GCS), Maddocks Score, a symptom checklist, cognitive evaluation, a modified Balance Error Scoring System (BESS), and the Standardized Assessment of Concussion (SAC) [24]. The 2 main types of balance tests are BESS and the Sensory Organization Test (SOT). To administer BESS, the number of errors an athlete makes while maintaining different stances are counted. For SOT, balance disturbances are quantified with a force plate while the athlete's sensory input is varied, [33]. Symptom scales can also be administered longitudinally after an injury to monitor recovery. A number of checklists have been developed for this purpose, including the Graded Symptom Checklist (GSC) and the Post-Concussion Symptom Scale (PCSS) [34, 35]. Neurocognitive tests are also valuable diagnostic tools, and many computerized test batteries have been developed to be easily administered [32]. Computerized neurocognitive tools include the Automated Neuropsychological Assessment Metrics (ANAM), CogSport, the Concussion Resolution Index (CRI), Immediate Post-Concussion Assessment and Cognitive Test (ImPACT), and the NIH Toolbox [36-40]. There is also evidence that these computerized tests can have good sensitivity and specificity compared with traditional neurocognitive tests [40, 41]. While there are many diagnostic tools available for concussions, they still rely on identification of an injurious incident or alteration in the mental status of a player [24].

Another complicating factor in concussion diagnosis is the lack of structural changes in the brain. There are typically no changes on conventional imaging techniques like computed tomography (CT) and magnetic resonance imaging (MRI) [24]. There have been extensive efforts to identify advanced imaging modalities that could assist in diagnosis of concussion. Some of the imaging techniques that have been evaluated include functional magnetic resonance imaging (fMRI), magnetic resonance spectroscopy (MRS), positron emission tomography (PET), and diffusion tensor imaging (DTI) [21, 42-46]. DTI seems especially promising for detecting axonal disruption due to concussion because it measures diffusion of water, which is normally axially oriented in white matter fiber tracts [21, 47]. Other diagnostic tools still in development include blood biomarkers, eye tracking, magnetoencephalography (MEG), and electroencephalography (EEG) [48-53].

Concussions and Long-Term Neurodegenerative Processes

The first description of neuropathology associated with chronic, repeated head trauma was in a study of retired boxers [54]. A similar but milder form of the pathology was later identified in the brain of an NFL player that appeared grossly normal [55, 56]. The term chronic traumatic encephalopathy (CTE) is now widely used for this pathology, and has been described in patients that participated in a variety of contact sports as well as military veterans [55-57]. The neuropathological findings in these studies are classified as a distinct tauopathy [57, 58]. CTE can be distinguished from other neurodegenerative diseases by aggregates of phosphorylated tau (p-tau) around small vessels at the depths of cortical sulci [59]. Neuropathologic diagnostic criteria have been developed for CTE, but unfortunately there is currently no way of diagnosing the disease prior to death [59].

In addition to studies on the long-term effects of repeated head injuries, recent work has been done to determine if clinical changes in individual athletes can be detected after a season of head impacts without a diagnosed concussion. Outcome measures used include neuropsychological screening tests and several advanced imaging techniques. In 2012, McAllister et al. paired changes in cognitive assessment tests with head impact exposure for collegiate athletes over one season of participation, and found associations between higher impact exposure metrics and lower cognitive test scores [60]. Several studies done by Talavage and Breedlove et al. identified a subset of high school football players with functional deficits measured by cognitive assessments and fMRI [61, 62]. This group of athletes also sustained a higher number of impacts throughout the season compared with teammates that did not have functional deficits. Davenport et al. reported correlations between head impact exposure and DTI measures as well as DTI measures and cognitive assessment outcomes [45]. These studies may provide support to the theory that repetitive subconcussive head impacts have cumulative, damaging effects on the brain.

Head Injury Criteria and Safety Standards

Early attempts to quantify human head injury tolerance were based on head kinematics from cadaver experiments [63]. Skull fracture was selected as the failure criterion since functional disturbances cannot be determined from cadavers. Additionally, patients with a linear skull fracture often presented with loss of consciousness, representing a moderate to severe concussion [64]. These tests make up a portion of the Wayne State Tolerance Curve (WSTC), which relates human tolerance to linear head acceleration to impact duration. The curve was later extended for longer durations with a combination of animal and cadaver experimental data, and non-injurious human volunteer data [63, 65-68].

The importance of rotational kinematics in brain injury had previously been theorized [69, 70]. A series of experiments with animal models were performed to elucidate human tolerances to rotational acceleration [71-81]. These experiments showed that both linear and rotational motion were critical factors in determining brain injury severity. In animals subject to purely translational motion, only focal lesions were seen, while diffuse injuries could be reproduced with a combination of translational and rotational motion [23]. Based on finite element analysis of these and other animal models of rotationally-induced head injury, a rotational injury criterion was proposed to supplement the current automotive standards based on linear acceleration alone [82, 83]. These and other proposed brain injury criteria will be discussed in detail in Chapter 2.

The safety standards used today for automotive and sports protective equipment are based on the WSTC [84-87]. The criteria used in these standards have been criticized based on the fact that they only consider linear acceleration [88]. Despite this criticism, they have been very effective at reducing serious injuries and fatalities, because they limit the amount of energy transferred to the head during an impact. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) was formed in 1969 to address the increase in fatalities in sports due to head and neck injuries. After implementation of NOCSAE standards for football helmets, the rate of fatal head injuries was reduced by 74% [89].

Although helmets have been shown to be effective at reducing serious head injuries and fatalities, whether they are effective at reducing concussion risk has been questioned. Studies evaluating differences in concussion rates by helmet model have mixed results [90-93]. These studies define concussion rate as the number of injuries per athletic exposure, which is any game or practice an athlete participates in. Concussions per athletic exposure neglect the level of participation in practices or games as well as the number of hits different players are exposed to. For example, a first string linebacker would be exposed to more head impacts than a third string quarterback in a single game or practice, putting the linebacker at a higher risk for sustaining a concussion. However, both scenarios would be considered a single athletic exposure. Estimates of differences in concussion rates between helmet types are improved when head impact exposure is controlled for. Studies defining concussion rates as the number of injuries per number of head impacts found significant differences in injury rates by helmet model [15, 94].

In addition to on-field studies, laboratory evaluations comparing different helmet models have been performed [11, 12, 95-97]. These studies are based on the fundamental principle that helmets that lower head acceleration reduce the risk of injury. A large body of research has shown that concussive impacts are associated with higher linear and rotational accelerations than impacts that do not result in concussion [12, 98-102]. Several studies compared newer helmet models with thicker padding to older helmet models, and found that in general the newer models performed better when considering peak linear and rotational accelerations [95-97]. Relative performance ratings of football and hockey helmets have also been recently introduced to inform consumers on the ability of different helmets to reduce the risk of concussion [11, 12]. These studies found large differences in relative performance among helmets that pass minimum safety standards. The NFL also recently released a list of relative rankings to assess impact performance of helmets worn by NFL players [103]. Although different methods and injury severity metrics were used, the results of both football helmet rankings were largely in agreement.

From a mechanical standpoint, differences in performance between newer and older helmet models depend on their ability to modulate impact energy transfer to the head. Older models tend to have a smaller offset with less room for padding [96]. Thinner padding is required to be stiffer to manage high energy impacts, with the tradeoff being that it is less effective for lower energy impacts. Thicker padding can be more compliant, while still providing protection for both high and low energy impacts.

Research Objectives

The following chapters address laboratory evaluation of protective headgear using previously developed brain injury criteria. Data collected from instrumented athletes were used to give context to laboratory methods through quantifying magnitudes of head impacts and head impact exposures. The research in this dissertation provides insight to the benefits and limitations of different laboratory methods, and how those methods can be related to head injury severity for injury prevention strategies. The specific objectives of the following chapters are:

1. To provide an analytical review of currently used and previously proposed brain injury criteria by evaluating the predictive capability of each using kinematic data collected from instrumented football players with known clinical outcome.
2. To develop a methodology for evaluating the relative performance of hockey helmets with a custom laboratory impact device.
3. To implement the methodology for evaluating hockey helmets and analyze the relative performance of available helmet models.
4. To assess differences in impact duration in laboratory systems commonly used to evaluate helmet performance.

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Chapter 2

Evaluating the Predictive Capabilities of Brain Injury Criteria: An Analytic Review

Abstract

Despite advances in the understanding of human tolerance to brain injury, the injury criteria used in automotive safety and protective equipment standards have changed little since they were first implemented. Although other criteria have been proposed as improvements over those currently used, evaluating the predictive capability of these criteria is challenging. These criteria are based on head kinematics from experimental work, mechanical models of the head, or finite element models of the head. A review of existing brain injury criteria is presented here, followed by an analysis of the predictive capability of those criteria. Laboratory reconstructions of head impacts from instrumented football players were used to quantitatively compare the predictive capability of all injury criteria. The reconstructions provided kinematic data from head impacts paired with known clinical outcome. Eleven concussive impacts and 44 subconcussive impacts were reconstructed, and all kinematic injury criteria calculated for each impact. Receiver operating characteristic (ROC) curves were used to compare the criteria with the area under the curve (AUC). All criteria were significantly better predictors than random guessing ($p < 0.0007$). Criteria that used a combination of linear and rotational kinematics were better predictors based on AUC. The results of this study can be used to inform selection of injury criteria for laboratory evaluation of protective headgear.

Introduction

Injury criteria are often used in evaluation of automotive safety and protective equipment. There have been a number of proposed criteria for brain injury, but few have actually been implemented in safety standards. Many of these criteria are based on experimental data collected in the 1950's and 1960's used to develop a concussive tolerance curve known as the Wayne State Tolerance Curve (WSTC) [1]. Despite criticisms of this work and the resulting injury criteria, the standards that implement them have been successful in reducing the incidence of head injuries. Shortly after implementation of standards by the National Operating Committee on Standards for Athletic Equipment (NOCSAE), fatal head injuries were reduced in football by approximately 74% [2]. Similar trends were seen with the introduction of mandatory safety standards for motor vehicles, titled Federal Motor Vehicle Safety Standards (FMVSS). Since the implementation of FMVSS, the fatality rate of motor vehicle occupants has been reduced by 81% [3]. Despite the limitations of the experimental data used, these standards have been successful at reducing injury incidence because they limit the amount of energy that is transferred to the head during an impact.

The objective of this chapter is to review existing brain injury criteria, and then analyze the predictive capability of each criterion using laboratory reconstructions of impacts recorded from instrumented football players.

Review of existing head injury criteria

Existing head injury criteria are based on head kinematics from experimental data, simple mechanical system models of the head, or finite element models of the head. More recent efforts have been focused on developing risk curves to determine the probability of head injury for a given input parameter rather than a pass/fail threshold. Each of these categories and their associated criteria will be reviewed here.

Early attempts to quantify human head injury tolerance were based on head kinematics from cadaver experiments. The first iteration of the WSTC in 1960 contained a combination of data from full body cadaver drop tests and detached head drops on to a steel block [4]. Skull fracture was selected as the failure criteria since functional disturbances cannot be determined from cadavers. It had been observed that patients with a linear skull fracture often presented with loss of consciousness, so a linear skull fracture in a cadaver was thought to be representative of a moderate to severe concussion [5]. This initial tolerance curve was composed of 6 data points of head acceleration at the occiput and temporal pressure versus impact duration. These tests make up the first 1-6 ms of the WSTC. The curve was later extended for longer duration impacts (6-10 ms) with a combination of animal and cadaver experimental data [4, 6, 7]. Pressure pulses of varying duration on the dura of anesthetized dogs were related to the cadaver experiments through intracranial pressure (ICP). The animal data provided injury severity, while head acceleration levels causing injurious levels of ICP could be measured with cadavers. The asymptote of the WSTC for long duration head accelerations was set at 42 g based on human volunteer data, but later was revised to 80 g since additional volunteer data had exceeded 45 g [8, 9]. The acceleration values in the tolerance curve are considered "effective" acceleration, but the definition of that term was unclear. Years later, researchers at the Japan Automobile Research Institute proposed a concussive threshold of man through the use of primates and human physical model experiments [10]. From dimensional analysis of the concussive threshold for monkeys, they determined that the WSTC is consistent with a concussive tolerance although the thresholds differed at longer durations.

Due to difficulties in applying the WSTC to automotive safety tests and uncertainties in the meaning of "effective" acceleration, many attempts were made to approximate the tolerance values analytically. Gadd was the first to apply the WSTC to a head injury criterion [11, 12]. He proposed a weighted criterion that would preferentially weight higher magnitude accelerations throughout the duration of an impact pulse. This criterion became known as the Severity Index (SI, Eq. 1).

$$SI = \int a(t)^{2.5} dt \tag{1}$$

Where a is head acceleration as a function of time. The weighting factor of 2.5 was based mainly on the slope of the straight-line approximation of a log-log plot of the animal data used in the WSTC. The original proposed SI tolerance limit for life-threatening head injury was 1000, which was used in FMVSS 208 until it was replaced by the head injury criterion (HIC, described below). Gadd later suggested a limit of 1500 for distributed or non-contact head impacts, which was adopted by NOCSAE [13]. The value was reduced to 1200 when HIC was reduced for automotive safety standards.

In a review of the severity index, Versace argued that the constant used in the SI formula was unjustified [14]. He suggested that an approximation of the tolerance curve could still be used, but the effective acceleration should be defined as average acceleration to better approximate the curve. He proposed the general form for what became known as HIC after it was adopted by NHTSA (Eq. 2).

$$HIC = max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right]^{2.5} (t_2 - t_1) \tag{2}$$

Where $a(t)$ is acceleration as a function of time and t_1 and t_2 are the bounds of the time interval that maximizes the HIC value over the duration of the pulse. HIC was limited to 1000 when it was first adopted by NHTSA. At that time there was no limit to the time interval over the duration of the pulse. NHTSA first proposed to limit the time interval to 36 ms in 1986 because HIC values for long duration accelerations with low magnitude were exceeding the threshold despite volunteer data indicating that should not be the case [15]. Due to the fact that none of the injurious points on the WSTC had HIC durations of greater than 13 ms, a 15-17 ms limit was suggested around the same time. However, using that time limit without changing the max HIC value would result in a drastically reduced failure rate of cars. HIC15 was introduced when the max HIC values were reduced to a value that would result in similar failure rates.

Around the same time as the development of SI and HIC, a number of injury criteria based on mechanical system models were developed. The first one to be developed is known as the J-Tolerance Index (JTI) from the Vienna Institute of Technology [16]. It consisted of a single degree of freedom spring-mass-damper system to model the head. The model was thought to represent the relative displacement of one side of the skull with respect to the other during impact. The maximum displacement of the system could be determined from the solution of the equation of motion (Eq. 3).

$$\ddot{x} + 2D\omega\dot{x} + \omega^2 x = -b(t) \tag{3}$$

Where x, $\dot{x}$, and $\ddot{x}$ are the displacement, velocity, and acceleration of the system respectively, D is the damping coefficient, ω is the angular frequency, and $-b(t)$ is the excitation of the oscillation system, which corresponds to the measured experimental acceleration trace. The model constants and tolerable displacement were determined from the WSTC. The tolerance threshold is surpassed when the maximum displacement determined from an acceleration pulse (x_{max}) exceeds the tolerable displacement (x_{tolr}), or when $J \geq 1$ (Eq. 4).

$$J = \frac{x_{max}}{x_{tolr}} \tag{4}$$

The effective displacement index (EDI) was then introduced building on JTI [17]. An identical spring-mass-damper model was used with the addition of variable damping. It was proposed that the appropriate damping term should be determined from biomechanical data rather than just curve fitting to the WSTC.

A different approach to the mechanical system model involved a series of experiments to determine the mechanical impedance of monkey and human heads over a range of frequencies [18-21]. Multiple iterations of the model developed from these experiments were produced. The first was called the maximum strain criterion (MSC) and was a 2 degree of freedom spring-mass-damper system [19]. The model constants were determined from the experimental mechanical impedance data, and the tolerance curves were determined from a single survival acceleration from previous animal and volunteer data. The tolerance curves were much lower than the WSTC for all acceleration pulse shapes. The model was updated using injury data from 4 primate species to develop tolerable strain and associated tolerance curves for each species [20]. Dimensional analysis was then used to scale to human tolerance. The criterion was referred to as the mean strain criterion (MSC) since it was thought to represent average strain in the brain. The model was then further revised to the new mean strain criterion (NMSC) following difficulties in implementing the method appropriately [18]. The criterion was updated to include 4 directional models, and additional primate and cadaver data were used in the development. The outputs of the model were modified so that level of injury (AIS) could be determined rather than just a pass/fail threshold. The final iteration in the series of these models was the translational energy criteria (TEC) [21]. In this model, additional primate data was again added to further development, but a different approach was taken to determine injury severity by determining the amount of energy dissipated in the part of the model representing brain damping.

A revised brain model (RBM) was proposed on the basis that both MSC and JTI had fundamental inaccuracies [22]. MSC is based on accelerations input to the forehead, but the WSTC accelerations were measured at the occiput making the values not comparable. A transfer function was introduced to relate the frontal and occipital accelerations. The damping coefficient and natural frequency used in JTI were thought to be too high based on measurements from human brains. The RBM used a combination of tolerable velocity for short duration impacts and tolerable deformation for longer duration. This model closely matched the WSTC.

Another similar model was proposed to develop a criterion to replace HIC based on the thought that the injury metric should be based on tissue-level damage rather than acceleration [23]. This model was referred to as the brain compliance model (BCM), with the viscous mechanism of soft tissue injury as its basis. The viscous mechanism states that injury is based on strain and strain rate in the tissue. The BCM was a lumped mass model that simulated the compliance of brain tissue in the skull. The static, elastic compliance of the brain was determined from experiments with dogs to determine the change in ICP with fluid volume added to the cerebrospinal fluid (CSF) space. The brain was modeled as a bilinear spring with increased stiffness with deflection, and an arbitrary damping coefficient chosen to allow more deflection between the brain and skull. The response of the model to selected inputs was compared to the NMSC, and showed more variation in response to skull motion while also allowing restoration of the brain by the elastic elements. The severity of injury for the BCM viscous response and displacement was compared to HIC and linear acceleration. The BCM differed for impacts with complex accelerations where it was thought that the BCM was able to identify the injurious portion of the impact and explain the mechanism of injury.

Using a finite element (FE) model rather than a simple mechanical system model, Ward et al. predicted a brain pressure tolerance (BPT) to injury [24]. The input to the model was linear and rotational acceleration and rotational velocity of the skull from animal and cadaver experiments as well as helmeted aircraft accident reconstructions. The threshold for moderate injury was 25 psi, and severe injury was 34 psi. Curves of linear acceleration versus duration were developed for constant pressure to compare to other injury metrics. The BPT had the best agreement with HIC, but had a lower tolerance for short (< 4 ms) impacts.

Although early work had theorized the importance of both linear and rotational acceleration in brain injury, all of the initially proposed head injury criteria based on head kinematics were based only on linear acceleration [25, 26]. This was largely due to the limited data used to develop the criteria. However, in 1986, the first kinematic-based head injury criterion to include a rotational component was proposed [27]. The criterion was known as a generalized acceleration model for brain injury threshold (GAMBIT), and it was suggested that it was analogous to combined axial and shear stresses used in engineering design. The general form of the equation is shown in Eq. 5.

$$G(t) = \left[\left(\frac{a(t)}{a_c}\right)^n + \left(\frac{\alpha(t)}{\alpha_c}\right)^m\right]^{1/S} \tag{5}$$

Where $a(t)$ is linear acceleration as a function of time, $\alpha(t)$ is rotational acceleration, and a_c and α_c are critical values of linear and rotational acceleration respectively. Two values for the exponents were considered: $n = m = S = 1$ and $n = m = S = 2$. Based on analysis of an automotive accident database and simulations, critical values of 250 g for linear acceleration and 25,000 rad/s^2 for rotational acceleration were recommended [28, 29]. With the exponents set to

$n = m = S = 2$, a threshold of $G = 1$ represented a 50% probability of AIS 3+ head injury [30]. The proposed form of the criterion is shown in Eq. 6, with the maximum value of $G(t)$ used for injury assessment.

$$G(t) = \left[\left(\frac{a(t)}{250}\right)^2 + \left(\frac{\alpha}{10{,}000}\right)^2\right]^{1/2} \leq 1 \tag{6}$$

The next criterion combining linear and rotational acceleration was not proposed until 2000, and was referred to as head impact power (HIP) [31]. HIP was based on the hypothesis that the threshold for head injury would be a critical value of kinetic energy of the head. Since approximations of the WSTC could be rewritten in a form that is proportional to the rate of change in kinetic energy of the head, or power, an expression representing power was chosen for the criterion (Eq. 7).

$$\begin{aligned} HIP = {} & 4.5a_x \int a_x dt + 4.5a_y \int a_y dt + 4.5a_z \int a_z dt + \\ & 0.016\alpha_x \int \alpha_x dt + 0.024\alpha_y \int \alpha_y dt + 0.022\alpha_z \int \alpha_z dt \end{aligned} \tag{7}$$

Where a_x, a_y, and a_z are linear accelerations in the x, y, and z directions, and α_x, α_y, and α_z are rotational accelerations in the x, y, and z directions. The coefficients are equal to the mass and moment of inertia of the human head. The coefficients could be varied based on directional sensitivity of the head to impact. HIP risk curves were developed from NFL reconstructions of concussive impacts in professional football players. For these cases, HIP was a better predictor of mild traumatic brain injury (mTBI) than HIC.

The NFL reconstructions were later used to develop injury predictors through the use of an FE model [32]. Eight different tissue injury predictors were compared with 10 kinematic-based predictors to determine the highest correlations with injury. A linear combination of peak change in rotational velocity and HIC had a high correlation with strain and was suggested as a potential injury predictor (Eq. 8), although several other combinations had similarly high correlations with injury.

$$KLC = 0.004718\Delta\omega_r + 0.000224HIC \tag{8}$$

This metric has been referred to as the Kleiven criterion or Kleiven's linear combination (KLC).

Using a different FE model (SIMon), another combination of head kinematics was correlated with a brain injury predictor from the model [33]. Scaled animal head injury data were used to determine the cumulative strain damage measure (CSDM), which was used to predict injury. A series of anthropomorphic test device (ATD) tests were simulated to determine the best linear fit between CSDM and the kinematic metric, brain injury criterion (BRIC, Eq. 9).

$$BRIC = \frac{\omega_{max}}{\omega_{cr}} + \frac{\alpha_{max}}{\alpha_{cr}} \tag{9}$$

Where ω_{max} is the maximum rotational velocity and α_{max} is the maximum rotational acceleration during the test, and ω_{cr} and α_{cr} are critical values determined through optimization. Critical values were determined for the ATD tests and college football head impact data. The critical values were selected so that BRIC = 1 corresponds to a 30% risk of diffuse axonal injury (DAI) or abbreviated injury scale (AIS) 4+ injury. The critical values were different for the two datasets. It was suggested that BRIC and HIC be used together to better predict injury.

BRIC was then revised using a combination of SIMon and GHBMC FE models with additional data and injury predictors [34]. BrIC as a function of rotational velocity with directionally dependent critical values correlated best with injury, and was recommended for use (Eq. 10).

$$BrIC = \sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2} \tag{10}$$

Here the critical values were selected such that BrIC = 1 corresponds to a 50% probability of an AIS4+ injury.

The THUMS FE model was used to evaluate 2 proposed kinematic injury criteria and compare them to other criteria [35]. The 2 proposed criteria were both modifications of HIC. The first replaced linear acceleration in HIC with rotational (Eq. 11).

$$RIC = max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2}\alpha(t)dt\right]^{2.5}(t_2 - t_1) \tag{11}$$

The second criterion replaced the linear acceleration in HIC with the rotational component of HIP (Eq. 12).

$$PRHIC = max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2}HIP_{rot}dt\right]^{2.5}(t_2 - t_1) \tag{12}$$

Collegiate football head impacts and NFL reconstructions were used to evaluate correlations between the kinematic criteria and FE model injury predictors. The best correlations were between RIC and CSDM 10%, and PRHIC and CSDM 30%. The injury thresholds for RIC and PRHIC were set as a 50% probability of mTBI based on the NFL data.

In some cases, simple acceleration thresholds are used for brain injury prediction and safety standards. Many helmet standards use linear acceleration thresholds to certify new helmet models. For example, the Consumer Product Safety Commission (CPSC) limits the magnitude of linear acceleration during specified impact tests to 300 g. Similarly, ice hockey helmet standards state that the acceleration cannot exceed 275 g (Canadian Standards Association, CSA) or 300 g (American Society for Testing and Materials, ASTM) during certification tests. Some standards use an acceleration threshold with a duration constraint. Based on cadaver head impacts, a linear acceleration threshold that had to be exceeded for at least 3 ms continuously was proposed [36]. Based on this recommendation, FMVSS 201 states that head acceleration from impacts to instrument panels and seat backs cannot exceed 80 g for more than 3 ms. The motorcycle helmet standard FMVSS 218 also includes a duration component, stating that peak acceleration cannot exceed 400 g for any duration, 200 g for more than 2 ms, or 150 g for more than 4 ms during testing.

Rotational acceleration and velocity thresholds have also been proposed based primarily on primate experiments. After performing a series of primate experiments, Ommaya concluded that both translational and rotational motion were critical factors in the resulting brain injury severity [37]. Based on these findings, he recommended the use of 2 injury criteria, one to account for injuries due to direct impact and translation, and one to account for rotation. At that time, he recommended MSC for a translational criteria, and proposed a new set of thresholds for rotational acceleration and velocity (Table 1) based on his scaled primate injury data.

Table 1: Rotational acceleration thresholds with rotational velocity constraints for different abbreviated injury scale (AIS) levels proposed by Ommaya.

≥ 30 rad/s	
< 1700 rad/s^2	AIS 2
< 3000 rad/s^2	AIS 3
< 3900 rad/s^2	AIS 4
< 4500 rad/s^2	AIS 5
< 30 rad/s	
< 4500 rad/s^2	AIS 0 or 1
≥ 4500 rad/s^2	AIS 5

Through a combination of primate experiments, analytical models, and physical models, a tolerance curve for DAI was developed for rotational velocity and rotational acceleration [38]. The physical models were used to scale the animal injury data through equivalent strain values. Tolerance thresholds were derived from the physical models, while the analytical models were used to develop a tolerance curve for continuous combinations of rotational velocity and acceleration. The tolerance thresholds determined from the physical models were 16 krad/s^2 for rotational acceleration and 46.5 rad/s for rotational velocity.

Although a single tolerance value is often used for the purpose of safety standards, there is variation in human tolerance to head injury. Factors such as genetic predisposition and previous concussive history may affect an individual's tolerance to head injury. This variance makes the use of risk curves appealing for head injury prediction, so that probability of injury is determined for a given head impact rather than just a pass or fail. Probability of injury is also useful for comparison of protective equipment and automotive safety features. The first head injury risk curves were developed for HIC based on a series of cadaver experiments with either skull fracture or blood vessel damage as the outcome [2, 39, 40]. The final versions for brain injury and skull fracture risk used a modified median rank method (referred to as the Mertz/Weber method). The predicted risk reduction for fatal head injuries in football before and after helmet certification standards were determined from laboratory helmet evaluation. The predicted risk reduction (78%) matched very closely with the actual reduction in fatal head injuries (74%).

With the addition of head injury kinematics from reconstructions of NFL concussive impacts, a new concussion risk curve was developed [41, 42]. Logistic regressions were fit for HIC, SI, linear and rotational acceleration, change in linear velocity, and rotational velocity. The regression coefficients were also determined for just the struck players, and a combination of struck and striking players (which contained more non-concussive data points). Linear acceleration and its associated metrics had a higher correlation with injury than rotational acceleration.

A similar dataset of NFL reconstructions was used in combination with the WSUBIM FE model to develop risk curves for tissue injury predictors [43]. These injury predictors were also compared with kinematic parameters. Shear stress in the midbrain was the best predictor for mTBI based on this model, which correlated best with resultant rotational acceleration. The injury tolerance for mTBI for 10-30 ms impacts was estimated as 85 g for linear acceleration and 6000 rad/s^s for rotational.

The most recent concussion risk curve was developed using data from instrumented collegiate football players [44]. The data consisted of subconcussive and concussive impacts with defined distributions. A conservative underreporting rate was used to account for undiagnosed concussions. A multivariate logistic regression was fit to the data for a combined prediction of risk. The equation included a term dependent on linear acceleration (a), one dependent on rotational acceleration (α), and a term to account for the interaction between them (Eq. 13).

$$CP = \frac{1}{1 + e^{-(-10.2 + 0.0433a + 0.000873\alpha - 0.000000920a\alpha)}} \tag{13}$$

The combined predictive capability was compared to linear and rotational acceleration individually, which showed that the combined metric was the best predictor, but not significantly different from linear acceleration alone.

Methods

In order to evaluate the predictive capability of all kinematic-based brain injury criteria, head impacts sustained by collegiate football players with a known injury outcome were reconstructed using a pneumatic linear impactor. The reconstructions provided the 6-degree-of-freedom linear and rotational head kinematics necessary to calculate all brain injury criteria evaluated.

Instrumented Football Player Impacts

The head impact data reconstructed were collected from Virginia Tech football players between 2004 and 2015 using the Head Impact Telemetry (HIT) System (Simbex, Lebanon NH). The HIT System consists of 6 single-axis accelerometers mounted on the inner surface of the helmet. The accelerometers are spring-mounted so that they are coupled to the surface of the head, and measure head acceleration rather than helmet acceleration. A novel algorithm is used to calculate linear and rotational acceleration at the center of gravity (CG) of the head, as well as the location of impact [45]. When a specified acceleration threshold is exceeded, 8 ms of pre-trigger data and 32 ms of post-trigger data are recorded at 1000 Hz.

Laboratory Reconstructions

A dataset of 11 concussive impacts recorded by the HIT System were reconstructed. The breakdown of impact locations were: 4 front, 2 side, 3 back, and 2 top impacts (Figure 3). Due to the large number of impacts recorded by the HIT System that did not result in a diagnosed concussion, a linear acceleration threshold was introduced prior to randomly sampling a dataset of 44 subconcussive impacts. A threshold of 40 g was used to ensure that there was overlap between the subconcussive and concussive acceleration values, while still having impacts below the lowest recorded concussion. The region of overlap between the subconcussive and concussive acceleration distributions allows for the best assessment of predictive capability of injury metrics. There were 12 front, 6 side, 15 back, and 11 top impacts.

A pneumatic linear impactor (Biokinetics, Ottawa, Canada) was used reconstruct the player head impact data (Figure 2). The 14 kg ram had a flat, rigid nylon impacting face that was 12.7 cm in diameter. The impactor struck a modified medium NOCSAE headform mounted on a Hybrid III 50th percentile male neck [46, 47]. The headform was instrumented with 3 linear accelerometers (7264B-2000, Endevco, San Juan Capistrano, CA) and 3 angular rate sensors (ARS3 PRO-18K, DTS, Seal Beach, CA). All data were collected at a sampling frequency of 20,000 Hz. The head and neck were mounted on a 5-degree-of-freedom slide table that simulated the effective mass of the torso and allows for adjustment of impact location.

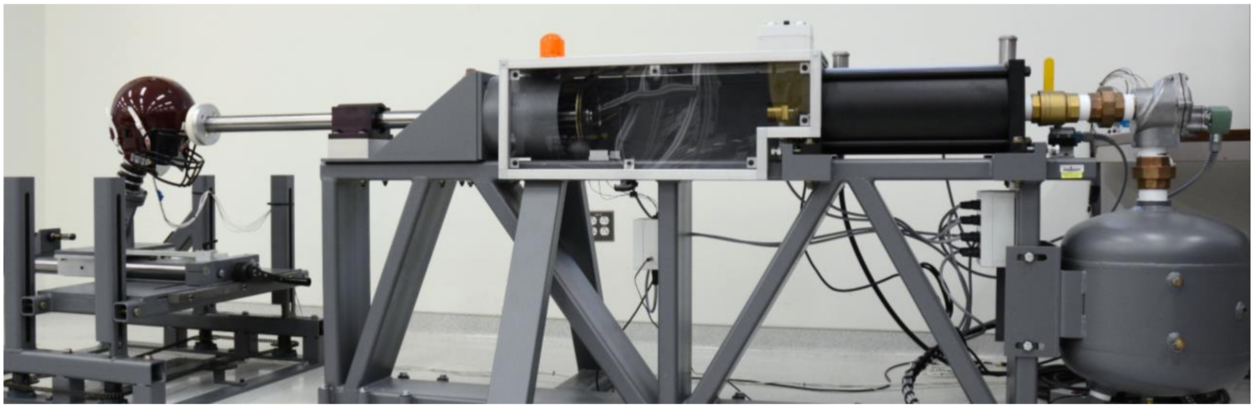

Figure 2: Pneumatic linear impactor used for all reconstructions. The impacting face was rigid, and struck a medium NOCSAE headform modified to be mounted on a Hybrid III 50th percentile male neck. The head and neck were mounted on a 5 degree of freedom slide table that simulated the effective mass of the torso during impact.

Helmet models were matched for each reconstructed impact. The 3 models worn were Riddell VSR4, Riddell Revolution, and Riddell Revolution Speed. The location as output by the HIT System was used as a starting point for each reconstruction. The location and impact speed were varied until the peak resultant linear and rotational accelerations matched the HIT System accelerations within +/- 5%. Linear acceleration data were filtered to channel frequency class (CFC) 1000, and angular rate data were filtered to CFC 155 using a 4-pole phaseless Butterworth low-pass filter. Angular rate data were differentiated to get rotational acceleration for each impact. HIT System and reconstruction linear and rotational acceleration distributions were compared for both concussive and subconcussive datasets.

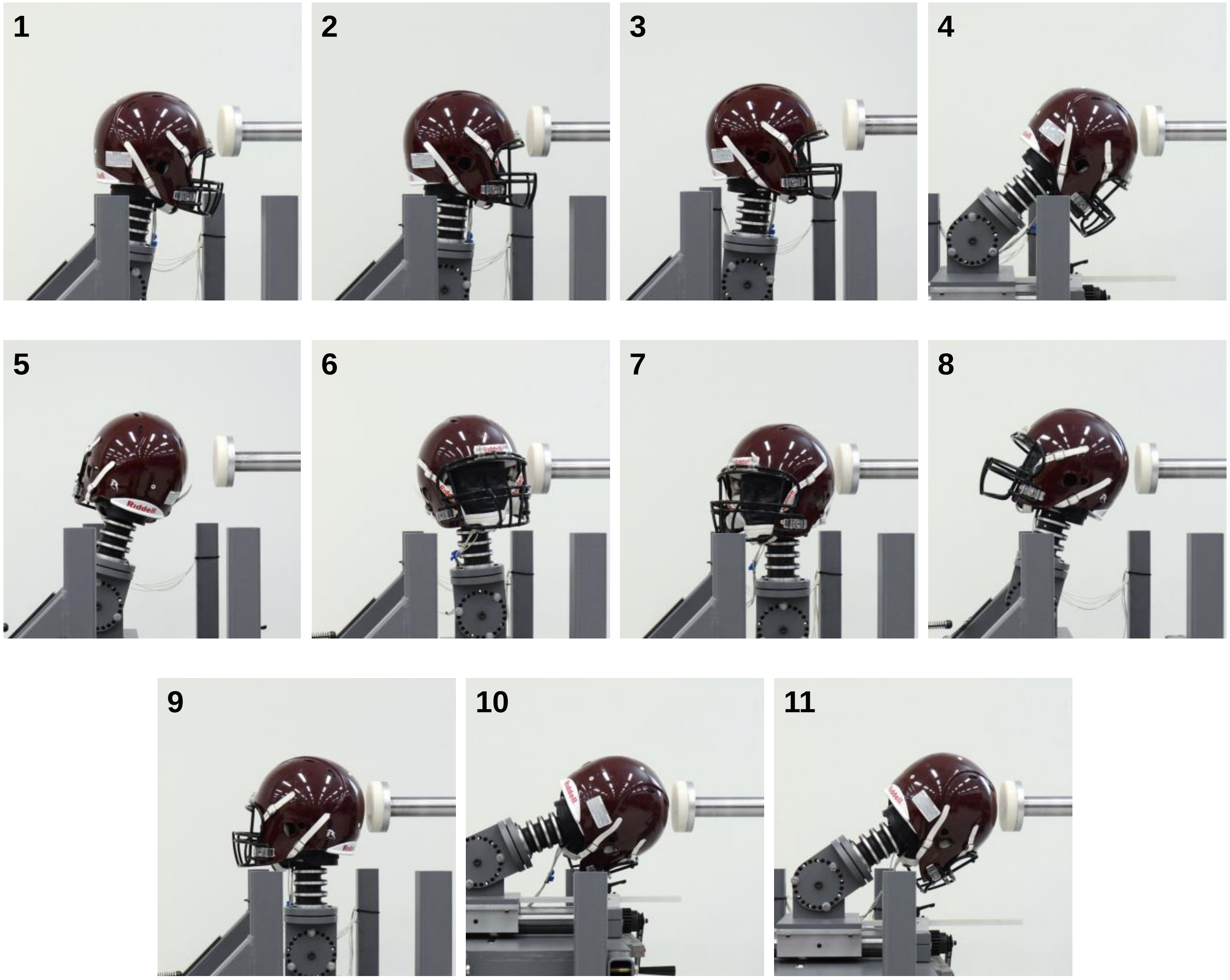

Figure 3: Final impact locations for the 11 concussive reconstructions. There were 4 front (1-4), 2 side (5-6), 3 back (7-9), and 2 top impacts (10-11). The HIT System locations were used as starting points, and adjusted to match the peak linear and rotational accelerations within +/-5%.

Injury Criteria Evaluation

Kinematic data from each reconstruction were used to calculate 10 different kinematic-based brain injury criteria (Table 2). The details of each criterion can be found in the introduction of this chapter. The 10 criteria evaluated were: Severity Index (SI), Head Injury Criterion (HIC), a Generalized Acceleration Model for Brain Injury Threshold (GAMBIT), Head Impact Power (HIP), Kleiven's Linear Combination (KLC), Kinematic Rotational Brain Injury Criterion (BRIC), a New

Kinematic Rotational Brain Injury Criterion (BrIC), Rotational Injury Criterion (RIC), Power Rotational Head Injury Criterion (PRHIC), and the linear portion of the Combined Probability of Concussion risk function (CC).

Receiver operating characteristic (ROC) curves were generated for each of the 10 injury criteria, along with peak resultant linear and rotational acceleration, and peak resultant rotational velocity for comparison. ROC curves represent the true positive rate (sensitivity) versus the false positive rate (1 – specificity) for all possible thresholds over the range of the criteria being evaluated. The area under the curve (AUC) was computed for all ROC curves to compare the predictive capability of all criteria. For AUC values, 1 represents a perfect predictor, and 0.5 no better than random guessing. The AUC for each predictor was compared to random guessing with a significance of $p < 0.10$ [48]. The optimum threshold was also determined for each predictor, with the sensitivity and specificity reported at that value. The optimum threshold represents the cut point in the predictor with the highest accuracy, or highest proportion of correctly classified cases. Pairwise comparisons were carried out to determine if any predictors were significantly different than others ($p < 0.10$) [49, 50]. To assess the correlation between the predictors evaluated, coefficients of determination (R^2) were determined for all pairs. The reconstruction sample size was selected to detect an AUC difference of 0.20 with a power of 0.80 at a significance of $p < 0.10$ using a bootstrap power calculation.

Table 2: Kinematic-based brain injury criteria that were evaluated with reconstructions of head impacts from instrumented football players. Resultant linear acceleration as a function of time: a; Linear acceleration in the x, y, and z directions as a function of time: a_x, a_y, a_z; Resultant rotational acceleration as a function of time: α; Rotational acceleration in the x, y, and z directions as a function of time: α_x, α_y, α_z; Peak resultant linear acceleration: a_m; Peak resultant rotational acceleration: α_m; Peak change in resultant rotational velocity: $\Delta\omega_r$; Peak resultant rotational velocity: ω_m; Peak rotational velocity in the x, y, and z directions: ω_{xm}, ω_{ym}, ω_{zm}; Rotational component of HIP: HIP_{rot}

Criterion	Equation
SI	$\int a^{2.5}\, dt$
HIC_{15}	$max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a\, dt\right]^{2.5} (t_2 - t_1)$
GAMBIT	$\left[\left(\frac{a(t)}{250}\right)^2 + \left(\frac{\alpha}{10{,}000}\right)^2\right]^{1/2}$
HIP	$4.5a_x \int a_x dt + 4.5a_y \int a_y dt + 4.5a_z \int a_z dt + 0.016\alpha_x \int \alpha_x dt + 0.024\alpha_y \int \alpha_y dt + 0.022\alpha_z \int \alpha_z dt$
KLC	$0.004718 * \Delta\omega_r + 0.000224 * HIC_{36}$
BRIC	$\frac{\omega_m}{46.41} + \frac{\alpha_m}{39774.87}$
BrIC	$\sqrt{\left(\frac{\omega_{xm}}{66.25}\right)^2 + \left(\frac{\omega_{ym}}{56.45}\right)^2 + \left(\frac{\omega_{zm}}{42.87}\right)^2}$
RIC_{36}	$max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} \alpha\, dt\right]^{2.5} (t_2 - t_1)$
$PRHIC_{36}$	$max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} HIP_{rot} dt\right]^{2.5} (t_2 - t_1)$
CC	$-10.2 + 0.0433a_m + 0.000873\alpha_m - 0.000000920a_m\alpha_m$

Results

For the 11 concussive impacts that were reconstructed, the median [25th 75th percentiles] peak linear acceleration of the HIT System data was 118 g [77 138], and 121 g [77 136] for the laboratory reconstructions (Figure 4, Figure 5). For the 44 subconcussive impacts reconstructed, the median peak linear acceleration of the HIT System data was 55 g [44 75], and 55 g [45 76] for the reconstructions. The concussive HIT System peak rotational acceleration median was 4871 rad/s^2 [3767 7235], and 4841 rad/s^2 [3766 7167] for the reconstructions. The median peak rotational acceleration for the subconcussive impacts was 2476 rad/s^2 [1846 3504] for the HIT System data, and 2418 rad/s^2 [1851 3383] for the reconstructions. Median peak rotational velocities for the reconstructions were 27 rad/s [17 30] for the concussive impacts, and 15 rad/s [6 20] for the subconcussive impacts. Rotational velocities were not computed by the HIT System algorithm, so they could not be compared to the reconstruction peak rotational velocities.

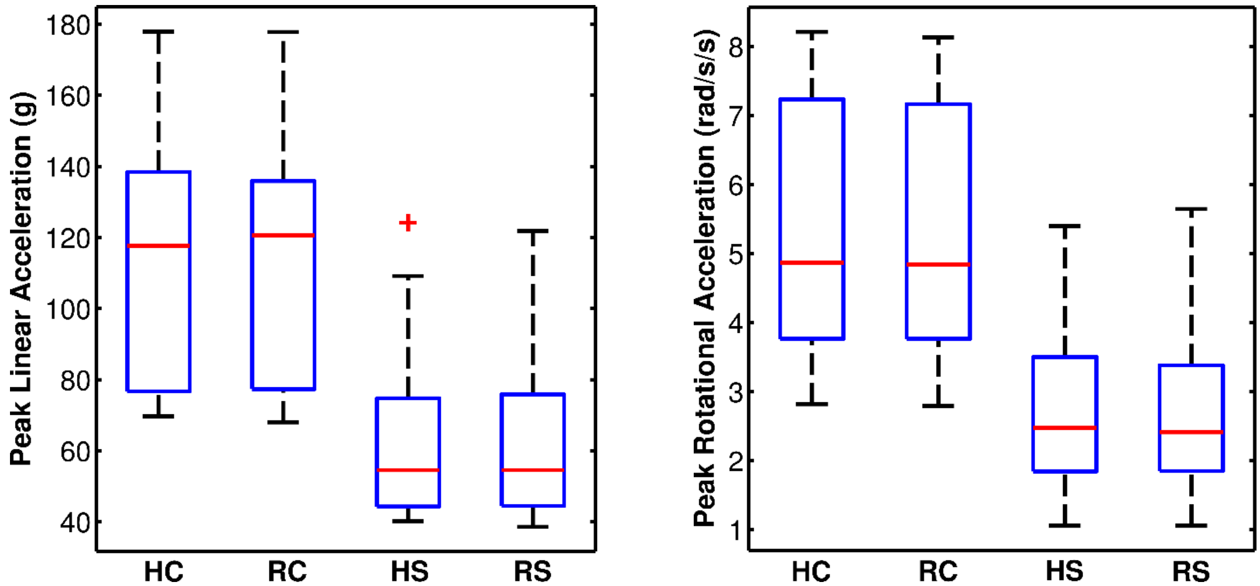

Figure 4: Distributions of peak linear (left) and peak rotational (right) head accelerations for both HIT System data and laboratory reconstructions. There was overlap between the subconcussive and concussive impacts to better assess the predictive capability of different injury criteria and kinematic predictors. The sample of subconcussive impacts had a right-tailed distribution, reflecting the distribution of the underlying population data. HC = HITs concussive; RC = reconstruction concussive; HS = HITs subconcussive; RS = reconstruction subconcussive

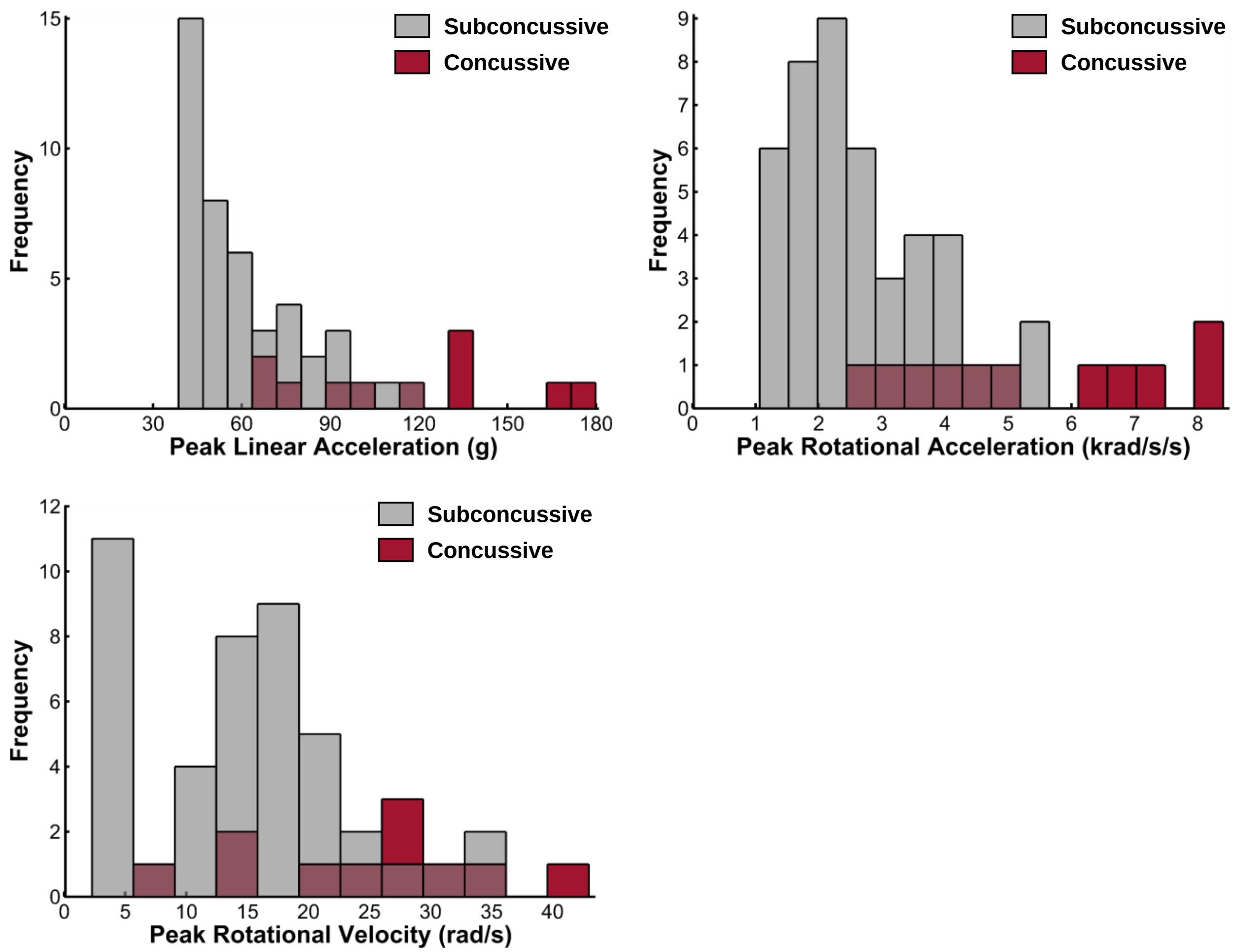

Figure 5: Distributions of peak linear acceleration (top left), peak rotational acceleration (top right), and peak rotational velocity (bottom left) for reconstructed impacts. Distributions are separated by concussive and subconcussive impacts. Subconcussive impacts are lower in acceleration magnitude than concussive, but there is more overlap between the distributions for peak rotational velocity.

The AUCs for all criteria and kinematic predictors evaluated ranged from 0.78 for peak rotational velocity to 0.92 for CC (Figure 6, Table 3). Sensitivity for the optimal threshold was generally low (< 0.60), with high specificity (> 0.90). KLC was the only predictor that had a sensitivity greater than 0.70 and a specificity greater than 0.90 at the optimum threshold. All predictors were significantly different than random guessing (p < 0.0007).

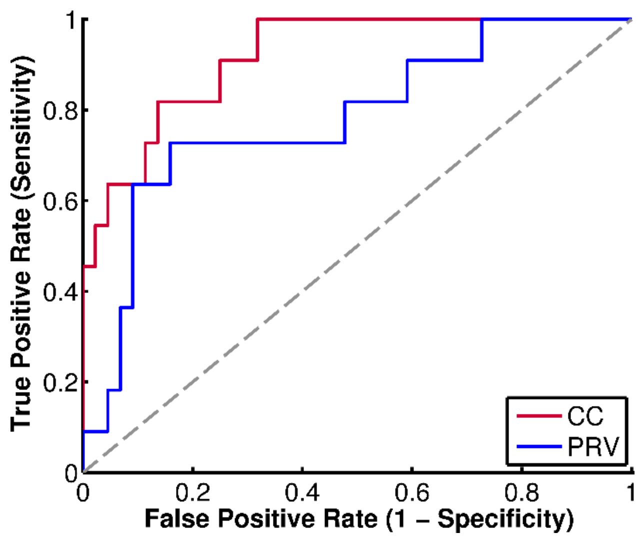

Figure 6: ROC curves for the highest (CC, AUC = 0.92) and lowest (PRV, AUC = 0.78) ranked predictors based on AUC. The difference in AUC for these predictors (0.14) was statistically significant (P < 0.10). The grey dashed line represents AUC = 0.5, or no difference from random guessing for a classifier. Both predictors were significantly different from AUC = 0.5 (p < 0.0007). PRV = peak rotational velocity

Table 3: AUCs with 95% confidence intervals for all predictors evaluated, as well as the optimum threshold for each predictor and the sensitivity and specificity at that point. The optimum threshold is based on the cut point in the predictor with the highest accuracy, or highest proportion of correctly classified cases. P-values for the comparison of each AUC with 0.5 are also listed. All predictors were significantly different than AUC = 0.5 ($p < 0.0007$).

Predictor	AUC [95% CI]	Optimum Threshold	Sensitivity	Specificity	p (AUC=0.5)
CC	0.92 [0.80-1.00]	0.57	0.45	1.00	3.70E-13
GAMBIT	0.91 [0.79-1.00]	0.53	0.55	1.00	1.70E-11
HIP (kW)	0.90 [0.78-1.00]	16.4	0.55	0.98	6.23E-11
PLA (g)	0.90 [0.78-1.00]	131	0.45	1.00	2.05E-10
PRA (rad/s^2)	0.90 [0.78-1.00	6390	0.45	1.00	2.05E-10
SI	0.90 [0.77-1.00]	301	0.64	0.93	3.58E-10
RIC	0.90 [0.77-1.00]	6.42E+06	0.55	1.00	3.58E-10
HIC	0.89 [0.77-1.00]	277	0.55	0.95	1.03E-09
KLC	0.88 [0.74-1.00]	0.16	0.73	0.91	3.39E-08
PRHIC	0.82 [0.66-0.98]	1.14E+06	0.27	1.00	4.41E-05
BRIC	0.81 [0.65-0.97]	0.66	0.64	0.91	8.57E-05
BrIC	0.80 [0.64-0.97]	0.40	0.73	0.86	0.000157
PRV (rad/s)	0.78 [0.61-0.95]	25	0.64	0.91	0.000659

Most significant differences in AUC between pairs of predictors were in combination with peak rotational velocity (Table 4). The peak rotational velocity AUC was significantly different than CC, HIP, peak rotational acceleration, RIC, and KLC ($p < 0.1$). These differences in AUC ranged from 0.10 to 0.14. The only other significant pairwise difference was between KLC and BrIC, with a difference of 0.07 ($p = 0.08$).

Table 4: AUC differences (top half) and p-values (bottom half) for all pairwise comparisons between predictors. Significant differences are highlighted in red ($p < 0.10$). Predictors are rank ordered from highest to lowest AUC on the diagonal. PLA = peak linear acceleration; PRA = peak rotational acceleration; PRV = peak rotational velocity

CC	0.01	0.01	0.02	0.02	0.02	0.02	0.02	0.04	0.10	0.11	0.12	**0.14**
0.52	**GAMBIT**	0.00	0.01	0.01	0.01	0.01	0.01	0.03	0.09	0.10	0.11	0.13
0.55	0.87	**HIP**	0.00	0.00	0.01	0.01	0.01	0.03	0.08	0.09	0.10	**0.12**
0.39	0.41	0.87	**PLA**	0.00	0.00	0.00	0.01	0.02	0.08	0.09	0.10	0.12
0.41	0.79	0.90	1.00	**PRA**	0.00	0.00	0.01	0.02	0.08	0.09	0.10	**0.12**
0.40	0.66	0.69	0.92	0.96	**SI**	0.00	0.00	0.02	0.08	0.09	0.10	0.12
0.37	0.74	0.83	0.95	0.90	1.00	**RIC**	0.00	0.02	0.08	0.09	0.10	**0.12**
0.24	0.47	0.50	0.74	0.86	0.60	0.90	**HIC**	0.02	0.07	0.08	0.09	0.11
0.32	0.48	0.36	0.62	0.54	0.55	0.55	0.64	**KLC**	0.06	0.07	**0.07**	**0.10**
0.16	0.23	0.16	0.29	0.16	0.25	0.15	0.28	0.15	**PRHIC**	0.01	0.02	0.04
0.15	0.22	0.16	0.27	0.15	0.23	0.15	0.26	0.12	0.58	**BRIC**	0.01	0.03
0.12	0.18	0.13	0.24	0.12	0.19	0.12	0.21	**0.08**	0.38	0.30	**BrIC**	0.02
0.09	0.14	**0.10**	0.18	**0.09**	0.15	**0.09**	0.16	**0.07**	0.15	0.15	0.30	**PRV**

Coefficients of determination (R^2) between all pairs of predictors ranged from 0.23 for peak linear acceleration versus peak rotational velocity, to 0.99 for SI versus HIC, PLA versus GAMBIT, PRV versus BRIC, PRV versus BrIC, and BRIC versus BrIC (Figure 7). Most injury criteria were highly correlated ($R^2 \geq 0.84$) with at least one kinematic predictor. SI, HIC, GAMBIT, and HIP were highly correlated with peak linear acceleration ($R^2 \geq 0.89$). CC was highly correlated with both peak linear and rotational acceleration ($R^2 \geq 0.88$). KLC, BRIC, and BrIC were highly correlated with peak rotational velocity ($R^2 \geq 0.84$). RIC was highly correlated with peak rotational acceleration ($R^2 = 0.88$).

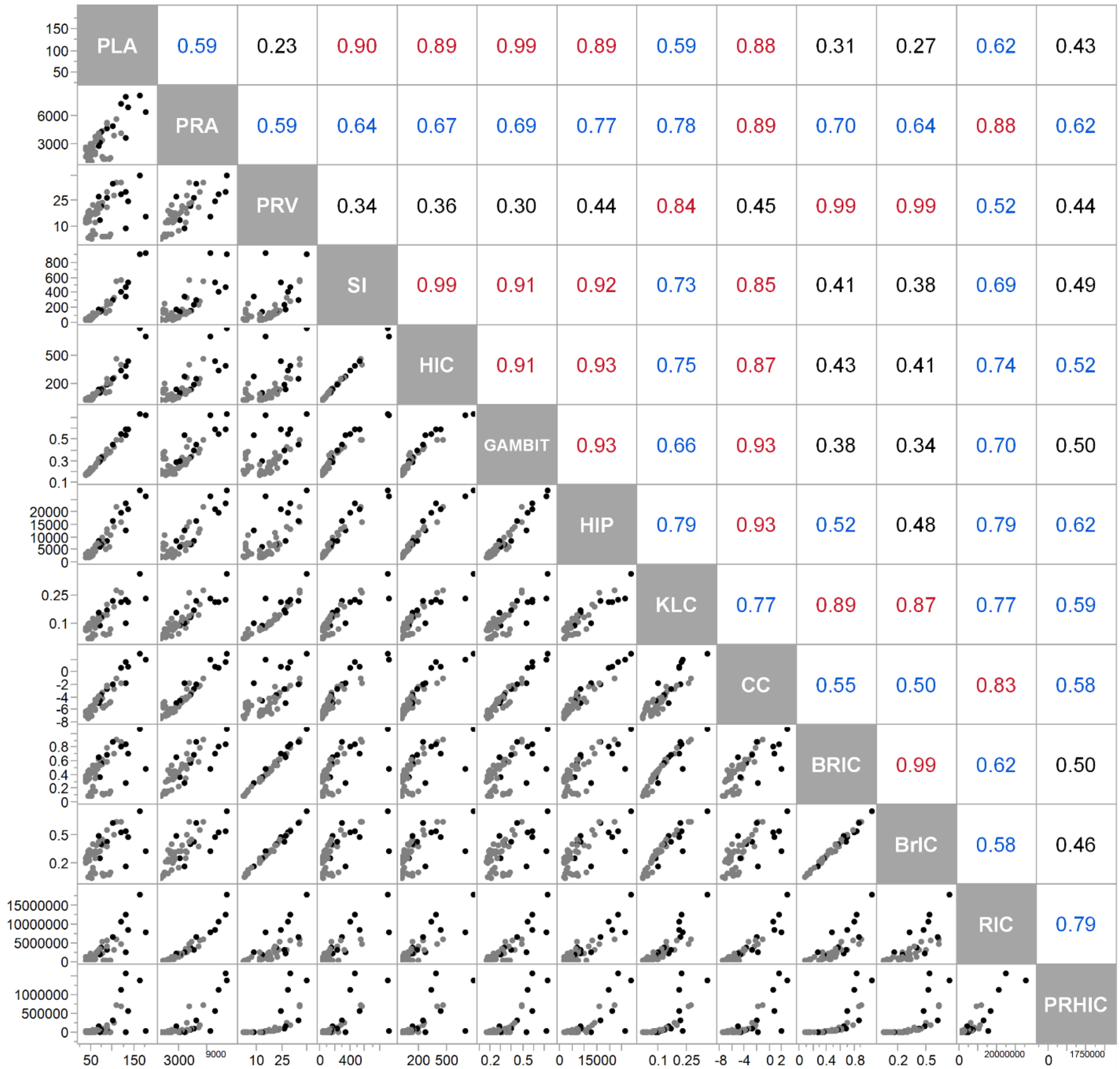

Figure 7: Scatter matrix for all pairwise comparisons between predictors (bottom half) with R^2 value (top half). Subconcussive impacts are denoted with gray markers on the scatter matrix, and concussive impacts with black. The R^2 values ranged from 0.23 (PLA versus PRV) to 0.99 (SI versus HIC, PLA versus GAMBIT, PRV versus BRIC, PRV versus BrIC, and BRIC versus BrIC). R^2 values are categorized into low ($R^2 < 0.50$, black), medium ($0.50 \leq R^2 < 0.80$, blue), and high correlations ($R^2 \geq 0.80$, red).

Discussion

Acceleration distributions for the reconstructed impacts in this study can be compared to previously reported distributions for both concussive and subconcussive impacts. For reconstructions of NFL impacts, the mean +/- standard deviation for concussive impacts was 98 +/- 28 g for peak linear acceleration, and 6432 +/- 1813 rad/s^2 for peak rotational acceleration [51]. The subconcussive impacts had mean accelerations of 57 +/- 22 g and 4028 +/- 1438 rad/s^2. A large dataset of head accelerations from football players wearing the HIT System had mean accelerations of 104 +/- 30 g and 4726 +/- 1931 rad/s^2 for concussive impacts [44]. The subconcussive impacts had mean accelerations of 26 +/- 19 g and 1072 +/- 850 rad/s^2. The mean accelerations for the concussive impacts reconstructed in the current study were 115 +/- 38 g and 5433 +/- 1956 rad/s^2. The subconcussive impacts had mean accelerations of 62 +/- 21 g and 2662 +/- 1153. Based on these comparisons, the concussive impacts reconstructed in this study are similar in severity to those previously reported. For subconcussive impacts, the large dataset of instrumented football player impacts is the most representative of the actual distribution, as the current study and the NFL reconstructions intentionally selected higher severity impacts. The threshold of 40 g used in the current study for subconcussive impacts ensured that there was overlap in the kinematic parameters for subconcussive and concussive impacts. Impacts greater than 40 g represent the top 16% of the impacts that the subconcussive data were randomly sampled from. Without a threshold, these impacts would be unlikely to be randomly selected, and there would be little to no overlap between the subconcussive and concussive impacts. Separation between the two datasets would inflate the predictive capability of all predictors since they are all based on kinematic parameters.

It has been estimated that the true incidence of concussion accounting for underreporting of injuries is approximately 38.8 concussions for every 10,000 head impacts [44]. For the reconstructions in this study to be representative of this true incidence, a total of 2,835 subconcussive impacts would have been necessary. However, 84% of these impacts would be below 40 g. The addition of a large number of low-magnitude impacts would not have improved the predictive capability analysis performed here, because all kinematic predictors would likely be below the lowest recorded concussion. For practicality, a total of 44 subconcussive impacts were reconstructed. This number represents one tenth of the number of subconcussive impacts needed to represent the true incidence of concussion for impacts over 40 g.

AUC provides a metric to compare the performance of all predictors for the impacts reconstructed in this study. In general, predictors that included both linear and rotational kinematics performed better than those dependent on one or the other. Conversely, predictors dependent on rotational velocity alone tended to perform worse, with peak rotational velocity having the lowest AUC. Nine out of the 13 predictors evaluated were dependent on linear and/or rotational acceleration, and had similar AUCs (0.88 – 0.92). There was a gap between the AUCs for those predictors and the 4 dependent only on rotational acceleration (0.78 – 0.82).

The optimum thresholds determined for each predictor based on ROC curves can be compared to thresholds used for current safety standards, and thresholds or risks of injury suggested in the literature. Current NOCSAE standards for football helmets limit SI to 1200, while the optimum threshold for SI in this study was 301. FMVSS 208 limits HIC_{15} for the Hybrid III 50^{th} percentile male to 700, and the optimum threshold was 277. These discrepancies are reasonable given that both standards were implemented to prevent severe head injuries and skull fractures, while the injury data in this study is a form of mTBI (concussion). When GAMBIT was developed, a threshold of 1 was suggested, and thought to represent a 50% probability of an AIS 3+ head injury. Similarly, a BRIC threshold of 0.92 was recommended for automotive safety standards to represent a 30% risk of AIS 3+ injury, and a BrIC value of 1 was set to be equal to a 50% probability of AIS 4+ injury. The optimum thresholds for GAMBIT, BRIC, and BrIC in this study were 0.53, 0.66, and 0.4, again reflecting that the concussive injuries in this study are less severe than the recommended thresholds for safety standards. Risk curves have also been developed for a number of the predictors evaluated [2, 31, 34, 35, 39, 44, 52-54]. Risk of concussion (or AIS 2+ in the case of HIC, BRIC, and BrIC) can be determined for these predictors (Table 5). The optimum threshold varies with respect to risk of injury for each predictor. These predictors were developed based on drastically different datasets, so differences in injury risk are not surprising. CC, peak linear acceleration, and peak rotational acceleration risk were determined from instrumented football players. These risk curves accounted for injury incidence and underreporting of concussions in football. CC used a higher underreporting rate, which resulted in a more conservative risk curve. Injury risk for HIP was based on reconstructions of NFL impacts resulting in concussion, with similar sample sizes for subconcussive and concussive impacts. The similar sample sizes result in a higher injury risk for the same severity when compared to the risk curves developed from instrumented football players. RIC, PRHIC, and BrIC were based on finite element models, and injury criteria developed for those models. The main injury predictors used with these models were based on scaled animal injury data. Finally, HIC risk curves were developed based on cadaver brain injury tests by using dye extravasated from vasculature as an indicator of AIS 4+ injury. HIC risk curves for other AIS levels were scaled from that data.

Table 5: Risk of concussion or AIS 2+ injury for the optimum thresholds determined from ROC curves for each predictor. There is a wide range of risk values, but the risk curves were developed based on very different datasets including human volunteers, reconstructions of injuries, cadavers, and scaled animal injury data [2, 31, 34, 35, 39, 44, 52-54].

Injury Criterion	Optimum Threshold	% Probability of Concussion/AIS 2+
CC	0.57	64
GAMBIT	0.53	92
HIP (kW)	16.45	79
PLA (g)	131	4
PRA (rad/s^2)	6390	56
RIC	6.42E+06	6
HIC	277	13
PRHIC	1.14E+06	89
BRIC	0.66	3
BrIC	0.40	26

Sensitivity for the optimum threshold was substantially lower than specificity for nearly all predictors. Since the optimum threshold was determined based on the maximum number of correct classifications, the weighting of 4:1 for subconcussive:concussive impacts favored higher specificity. If the threshold was decreased to increase sensitivity, a larger number of subconcussive impacts would become false positives than the number of concussive that would become true positives, thus decreasing the overall number of correct classifications.

Pairwise comparisons between predictor AUCs were generally not significant. Peak rotational velocity had the largest number of significant differences because it had the lowest AUC. The only other significant difference was between KLC and BrIC. Since paired data were used (same injury and non-injury cases used for all predictors), the correlation between predictors was a factor in the standard error of the difference between AUCs. Predictors with higher correlations had lower standard errors, making significant differences more likely. The low number of significant differences can be partially attributed to the similarities between predictors. All predictors evaluated were based on kinematic parameters, and while there is some variation in predictive capability, all parameters generally increase with increasing impact severity. However, there would likely be more significant differences with a larger sample size.

Correlations between predictors were largely intuitive. Most predictors were highly correlated with at least one of the kinematic predictors they were dependent on. GAMBIT, HIP, KLC, and CC were combinations of linear and rotational kinematics, however, CC was the only predictor highly correlated with both linear and rotational acceleration. GAMBIT and HIP were more heavily weighted towards linear acceleration due to the coefficients used in their equations. KLC was more heavily weighted towards rotational velocity. Additionally, many of the criteria were highly correlated with other criteria. SI, HIC, GAMBIT, HIP, and CC were all highly correlated with each other. KLC, BRIC, and BrIC were highly correlated as they were mostly dependent on rotational velocity. CC and RIC were highly correlated since both were highly correlated with peak rotational acceleration. The top performing predictors were combinations of linear and rotational kinematics, with the highest AUC (CC) being the only predictor highly correlated with both linear and rotational kinematics.

There were several limitations in this study. The small sample size, particularly the low number of concussions available for reconstruction, decreased the likelihood of detecting differences between predictor AUCs. The optimum thresholds were also limited to the values of cases reconstructed, which would be more continuous with a larger dataset. Additionally, the lower sample size made ROC curves more dependent on individual data points. While the HIT System has low average error, the individual data points are less reliable [55]. However, the average of concussive data points used in this study was not drastically different form previously reported averages of larger datasets [41, 44].

Conclusion

This study provides a unique dataset of 6-degree-of-freedom linear and rotational head kinematics for impacts reconstructed from events with a known injury outcome. The predictive capability of head injury criteria and kinematic parameters were evaluated using ROC curve AUCs. All predictors were significantly different from random guessing (AUC = 0.5). Predictors that combined linear and rotational kinematics performed better than those dependent on a single parameter. Additionally, predictors dependent only on rotational velocity performed worse than those dependent on linear and rotational acceleration. Correlations between predictors showed that many of the predictors were correlated with one another, suggesting that they do not provide any unique information on impact severity that would help predict injury.

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Chapter 3

Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets

Abstract

Optimizing the protective capabilities of helmets is one of several methods of reducing brain injury risk in sports. This paper presents the experimental and analytical development of Hockey STAR (Summation of Tests for the Analysis of Risk). The formula combines head impact exposure with brain injury probability over the broad range of 227 head impacts that a hockey player is likely to experience during one season. These player exposure data are mapped to laboratory testing parameters using a series of 12 impact conditions comprised of three energy levels and four head impact locations, which include centric and non-centric directions of force. Injury risk is determined using a multivariate injury risk function that incorporates both linear and rotational head acceleration measurements. All testing parameters are presented along with exemplar helmet test data. The Hockey STAR methodology provides a scientific framework for manufacturers to optimize hockey helmet design for injury risk reduction, as well as providing consumers with a meaningful metric to assess the relative performance of hockey helmets.

Introduction

Football is often the focal point of concussion research because of its popularity and the high incidence of concussions associated with it; however, the rate of concussion is higher in ice hockey [1, 2]. Moreover, it is the most common injury for women's collegiate ice hockey, and the second most common for men's [3, 4]. The current helmet safety standards for hockey helmets have changed little over the past 50 years when they were created to reduce the incidence of serious head injuries and deaths [5]. The first hockey helmet standards were instituted by the Swedish Ice Hockey Association (SIA) in 1962. Shortly thereafter, US and Canadian organizations developed similar standards. Today, most hockey helmets bear stickers representing certification by 3 different organizations: the Hockey Equipment Certification Council (HECC), the Canadian Standards Association (CSA), and the International Organization for Standardization (ISO) represented by a CE marking. These standards all have similar pass/fail criteria that were implemented to reduce the risk of catastrophic head injuries.

Recently, concussion has gained national attention and become a research priority as the incidence of injury rises and concerns about the long-term effects of repeated mild injury are brought to light [2, 6-10]. Many strategies have been employed in attempts to decrease the incidence of concussion, such as rule changes, education programs, legislation, and improvements in protective equipment [11, 12]. Examples of rule changes designed to reduce injuries include fair-play and body-checking rules, which are implemented in some ice hockey leagues. Studies have shown a reduction in the incidence of more serious injuries including concussions when these rules are in place [13, 14]. Education programs such as the Centers for Disease Control and Prevention's "heads up" on concussion initiative and the Hockey Concussion Education Project (HCEP) were developed to help educate coaches, players, and their parents on preventing, identifying, and responding appropriately to concussions [15-22]. Although all states and the District of Columbia now have concussion laws in place, it is unclear at this time how effective they are [11, 59]. These laws usually focus on education, removal from play, and approval required for return to play.

There is currently no objective information available to consumers on which hockey helmets provide better protection against serious, as well as milder, head injuries like concussions. Prior to the development of the Football STAR (Summation of Tests for the Analysis of Risk) Evaluation

System in 2011, this information was not available for football helmets either [23]. Since the first set of helmet ratings using this evaluation system were released, the number of helmets receiving the highest rating possible of 5 stars has risen from just one to a total of 12 helmets in 2014 [24]. In the past, there were no conclusive studies on the effectiveness of different helmet types in reducing concussions on the field [11, 12]. However, recent research has demonstrated that the risk of concussion on the field is lowered with a helmet that better reduces head accelerations upon impact [25].

Football STAR was developed based on two fundamental principles. The first is that the tests performed are weighted based on how frequently a similar impact would occur on the field during one season of play [23]. The second is that helmets that decrease acceleration decrease the risk of concussion. There are a number of concussion risk functions that have been developed to define probability of concussion as a function of linear head acceleration, angular head acceleration, or both [23, 26-31]. Debates over the mechanisms of brain injury and the ability of metrics that include linear or angular head acceleration to predict injury risk are long-standing [27, 32]. Numerous studies have attempted to differentiate the effects of linear and angular head accelerations on brain injury and determine if one or the other is more likely to result in concussion [33-35]. Current metrics for head injury safety standards use only linear head acceleration, and are based on human cadaver skull fracture and animal data [36-38]. However, more recently it has been shown that the combination of linear and angular head acceleration is a good predictor of concussion, and that helmets reduce both linear and angular acceleration [29, 31, 39]. Given the fact that all head impacts have both linear and rotational acceleration components, future helmet evaluation should quantify injury risk using both linear and rotational head kinematics.

The objective of this study is to describe the development of a new evaluation system for hockey helmets. The evaluation system will provide a quantitative measure of the ability of individual helmets to reduce the risk of concussion. Building on the framework of Football STAR, Hockey STAR will define laboratory test conditions weighted to represent how often hockey players experience similar impacts.

Methods

Hockey STAR Equation

The Football STAR equation was developed to identify differences in the ability of football helmets to reduce concussion risk [23]. The equation represents the predicted concussion incidence for a football player over one season. This predictive value is determined from laboratory tests with a helmeted headform to simulate head impacts at different locations and energy levels. Each laboratory condition is associated with the number of times that type of impact would occur over one season (exposure), and the probability that a concussion would occur due to the resultant head acceleration during each test (risk). In the Football STAR equation (Eq. 1), *L* represents the impact location of front, side, top, or back; *H* represents the drop height of 60, 48, 36, 24, and 12 in; *E* represents the exposure as a function of location and drop height, and *R* represents risk of concussion as a function of linear acceleration (*a*).

$$Football\ STAR = \sum_{L=1}^{4} \sum_{H=1}^{5} E(L,H) * R(a) \tag{1}$$

A similar equation is presented for Hockey STAR, with several important modifications (Eq. 2). The risk function now incorporates both linear and rotational acceleration since all head impacts result in both, and the combination has been shown to be predictive of concussion [29, 31]. The exposure component was modified to reflect data collected from hockey players that consisted of both linear and rotational acceleration. In the Hockey STAR equation, *L* represents the head impact locations of front, side, top, and back; *θ* represents different impact energy levels defined by the angle of the pendulum arm used to impact the head; *E* represents exposure, or the number of times per season a player is expected to experience an impact similar to a particular testing condition as a function of location and impact energy; and *R* is the risk of concussion as a function of linear (*a*) and angular (*α*) head acceleration. The exposure and risk components of the equation are described in later sections.

$$Hockey\ STAR = \sum_{L=1}^{4} \sum_{\theta=1}^{3} E(L, \theta) * R(a, \alpha) \tag{2}$$

The laboratory testing matrix includes 3 impact energy levels and 4 impact locations, for a total of 12 testing conditions per helmet. In practice, two helmets of every model will be purchased. Each of these helmets will be tested in the 12 conditions twice for a total of 48 tests per helmet model. Acceleration values for each helmet's test conditions will then be averaged for each impact condition prior to using the risk function to determine probability of concussion. Concussion risks will then be multiplied by the exposure values for each impact condition to determine incidence values. All incidence values are then aggregated to calculate a Hockey STAR value for each helmet. The Hockey STAR values for each helmet will then be averaged to determine a helmet's overall Hockey STAR value.

Hockey Head Impact Exposure

Head impact exposure is defined here as the number of impacts a player experiences over one season of play. Data from two different studies were utilized to determine the median number of impacts per season over a broader population of males, females, and youth ice hockey players. Wilcox et al. collected data from both male and female National Collegiate Athletic Association (NCAA) ice hockey teams over three seasons from 2009-2012 using helmet-mounted accelerometer arrays [40]. Using the same instrumentation, Mihalik et al. collected data from a population of male Bantam (13-14 years old) and Midget (15-16 years old) players over two years [41]. These accelerometer arrays have previously been described in detail, but briefly, each helmet contains six single-axis linear accelerometers that are oriented tangentially to the head and integrated into foam inserts which allow the sensors to maintain contact with the head during impact [42]. The median number of head impacts per player per season experienced by collegiate athletes was 287 for males and 170 for females [40]. The median number of impacts per player per season for youth athletes was 223 [41]. The median values for each population were averaged to determine an overall exposure of 227 impacts. This value was used to represent the total number of impacts for one player over one season. The exposure value was further defined by impact location and severity as described below.

Data collected with the helmet-mounted accelerometer arrays was used to map on-ice player impact exposure to lab conditions [40, 43]. Data from two male and two female NCAA ice hockey teams as well as one male and one female high school team were included. The data were scaled to reduce measurement error using a relationship determined from correlating resultant head accelerations calculated from the helmet instrumentation to a reference measurement in an instrumented dummy headform during controlled laboratory impact tests [44].

The helmet data were then stratified by impact location. The locations are defined by the azimuth and elevation of the impact vector and are generalized into bins representing the front, right, left, back and top of the head [45]. The front, right, left, and back consist of impacts with an elevation less than 65 degrees, and are divided equally into 4 bins that are centered on the intersection of the midsagittal and coronal planes, but offset by 45 degrees. The remaining impacts greater than 65 degrees in elevation are grouped as top impacts. The exposure for each impact location was weighted by how often they occur in data collected in the literature [40-43]. The front, side (left and right combined), and back were approximately 30% each, with the remaining 10% of impacts to the top of the head. These values were used to weight exposure by impact location.

Hockey Helmet Impact Device

The next step in defining exposure was to transform on-ice player head acceleration data distributions to impact conditions in the lab. To do this, a series of impact tests were performed over a range of input energies using a custom impact pendulum to map laboratory-generated head accelerations to those measured on-ice directly from hockey players. The impact pendulum system used for these tests, impact locations evaluated, and methods for the acceleration transformation are described in detail below.

A pendulum was chosen due to increased repeatability and reproducibility when compared with other head impact methods [46]. The pendulum arm is composed of 10.16 x 5.08 cm rectangular aluminum tubing with a 16.3 kg impacting mass at its end. The length of the pendulum arm from the center of its pivot point to the center of its impacting mass is 190.5 cm. The pendulum arm has a total mass of 36.3 kg and a moment of inertia of 72 kg-m^2. The impacting mass accounts for 78% of the total moment of inertia. The nylon impactor face has a diameter of 12.7 cm, which is flat and rigid in an effort to maximize repeatability and reproducibility of the tests. Furthermore, a rigid impacting face was chosen due to rigid surfaces in hockey, and to avoid impactor compliancy masking differences between helmets in comparative testing [47].

The pendulum impactor strikes a medium NOCSAE headform, which is mounted on a Hybrid III 50th percentile neck (Figure 8). The NOCSAE headform was used to provide the most realistic fit between helmet and headform [48]. A custom adaptor plate was used to mate the NOCSAE headform to the Hybrid III neck while keeping the relative locations of the occipital condyle pin and headform center of gravity (CG) as close as possible to that of the Hybrid III 50th percentile male head and neck assembly. Material was removed from the underside of the headform to optimize the position of the occipital condyle and accommodate the neck. The adaptor plate's mass was equal to the material removed. Although these distances matched exactly in the anterior-posterior and medial-lateral directions, the NOCSAE CG was 22 mm superior relative to the Hybrid III CG. The head and neck assembly are mounted on a sliding mass intended to simulate the effective the mass of the torso during impact. This sliding mass is part of a commercially available linear slide table that is commonly used for helmet impact testing (Biokinetics, Ottawa, Ontario, Canada). Contrary to most helmet drop test rigs, this system allows for linear and rotational motion to be generated during impact. To measure the kinematics resulting from impact, the headform was instrumented with a 6 degree of freedom sensor package consisting of 3 accelerometers and 3 angular rate sensors (6DX-Pro, DTS, Seal Beach, CA).

The front, side, back, and top of the headform were chosen to impact in laboratory tests (Figure 9). In order to account for a wider array of impact types, two of the locations were centric, or aligned with the CG of the headform (front and back), and two were non-centric (side and top). These locations resulted in some impacts with higher rotational components for a given linear acceleration than others, which was quantified by the effective radius of rotation at each condition. Effective radius of rotation was defined as the quotient of peak linear acceleration and peak rotational acceleration. Table 6 specifies the impact locations using measurement markings provided on the commercially available linear slide table.

Figure 8: The custom impact pendulum device was used to strike a NOCSAE headform mounted on a Hybrid III 50th percentile neck. The head and neck were mounted on a sliding mass that simulates the effective mass of the torso during impact. The slide table had 5 degrees of freedom so that any location on the helmet could be impacted: translation along the x axis, translation along the y axis, translation along the z axis, rotation about the y axis, and rotation about the z axis.

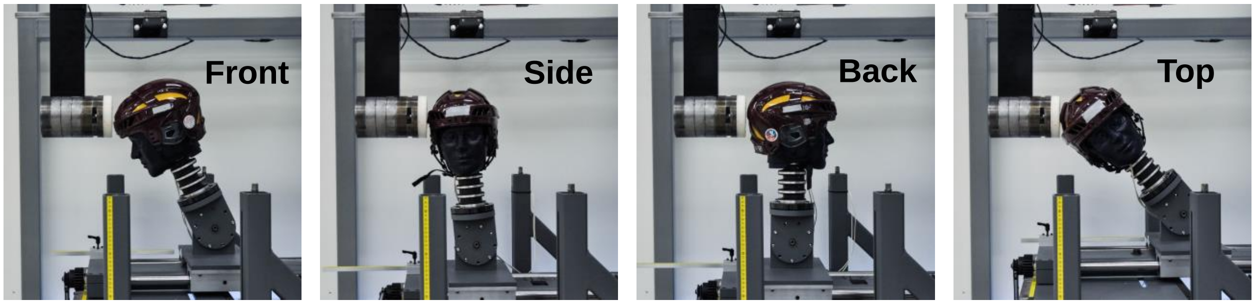

Figure 9: Photographs of the front, side, back, and top impact locations used to assess helmet performance. The side and top impact locations are non-centric, meaning the direction of force is not aligned with the CG of the headform; while the front and back impact locations are centric.

Table 6: Measurement markings and angles of rotation on the linear slide table for each impact location tested. The position on the x axis was defined as 1.25 cm before the pendulum was vertical for all locations. The center of the slider table is aligned with the centerline of the pendulum.

	Y Translation (cm)	Z Translation (cm)	Y Rotation (°)	Z Rotation (°)
Front	40.3	8.9	25	0
Side	36.9	3.5	5	80
Top	42.7	13.5	40	90
Back	40.3	4.9	0	180

Mapping Exposure Data to Laboratory System

A series of tests were performed to map the on-ice helmet data to laboratory pendulum impacts. For these tests, the NOCSAE headform was fitted with a size medium CCM Vector V08 helmet (Reebok-CCM Hockey, Inc., Montreal, Canada). The V08 model was chosen because it was one of the helmet types worn by instrumented players to generate head impact exposure data [40]. The linear accelerometer and angular rate data were collected at a sampling rate of 20,000 Hz.

Linear acceleration data were filtered to CFC 1000 Hz according to SAE J211, while angular rate data were filtered to CFC 155. Angular acceleration was calculated by differentiating the angular rate data. All data were then transformed to the CG of the headform. Three V08 helmets were tested, with each impacted from pendulum arm angles of 20, 30, 40, 50, 60, 70, 80, and 90 degrees at each of the four locations defined above, resulting in 96 impact tests.

After determining the total impact exposure per player per season and stratifying the on-ice helmet data by impact location, the data were transformed to laboratory impact conditions. To do this, the on-ice data for each location were reduced to include only impacts with effective radii of rotation in the range of corresponding laboratory impacts. Within these constraints, the on-ice head acceleration distributions were related to impact conditions in the lab. Bivariate empirical cumulative distribution functions (CDF) comprised of peak linear and peak rotational head accelerations were computed for on-ice data within each impact location's constraints. The CDFs were defined by determining the percentage of impacts less than or equal to each impact's peak linear and peak rotational acceleration. Using the location-specific CDFs, the percentile impact for each pendulum impact energy was determined by relating peak linear and peak rotational acceleration average values generated from each laboratory condition. Through this process, location-specific impact energy CDFs were determined for each population (male collegiate, female collegiate, male high school, and female high school). The 4 resulting impact energy CDFs were then averaged between populations for equal weighting between populations.

Low, medium, and high impact energy conditions were set prior to computing the weighting used in the Hockey STAR formula. These conditions were chosen to be representative of a span of impacts severities that encompass both sub-concussive and concussive head impacts, and are defined by pendulum arm angles of 40° (low), 65° (medium), and 90° (high). Weightings to be used for the Hockey STAR test configurations were determined by setting bounds on the impact energy CDFs midway between each test angle. For each location, the percentage of impacts below 52.5° was defined for the low energy condition, the percentage of impacts between 52.5° and 77.5° was defined for medium energy condition, and the percentage of impacts greater than 77.5° was defined for the high energy condition. The weightings for each test configuration were then computed by multiplying these percentages by the total number of head impacts that the average hockey player sustains at each location.

Injury Risk Function

The risk function used in Hockey STAR was updated to incorporate both linear (a) and rotational head acceleration (α) components (Eq. 3). Development of the combined risk function for concussion has previously been described [29].

$$R(a, \alpha) = \frac{1}{1 + e^{-(-10.2 + 0.433 * a + 0.000873 * \alpha - 0.000000920 * a\alpha)}} \qquad (3)$$

In short, the risk function was developed using data collected from high school and collegiate football players. A multivariate logistic regression analysis was used to model risk as a function of linear and angular head acceleration. There is an interaction term because linear and rotational acceleration are correlated. This risk function is unique in that it accounts for the under-reporting of concussion in the underlying data used to develop the curve [49, 50]. The predictive capability of the risk function was found to be good using NFL head impact reconstructions in addition to the impacts used to generate the function.

Exemplar Hockey Helmet Tests

Three exemplar helmets are used to demonstrate Hockey STAR. Each helmet was tested in 12 impact conditions: 4 locations with 3 impact energies per location. Pendulum arm angles of 40°, 65°, and 90° were tested, which equate to impact velocity of 3, 4.6, and 6.1 m/s. These illustrative tests differ from actual Hockey STAR tests in that only one helmet per model was tested, and each test configuration was only tested once. In practice, each test condition would be tested twice for each helmet, and acceleration values in each condition would be averaged before calculating risk. Hockey STAR values for the two helmets of each model are averaged to determine a helmet model's overall Hockey STAR value. For demonstrative purposes, two hockey helmets and one football helmet were tested under these conditions and Hockey STAR values calculated.

Results

Mapping Exposure Data to Laboratory System

Bivariate CDFs for linear and rotational accelerations experienced by male collegiate hockey players are shown in Figure 10 for each impact location. Peak linear and rotational head acceleration values generated during the pendulum tests are overlaid on the CDFs to illustrate how the laboratory tests relate to the on-ice head impact distributions. Constant impact energies varied in percentile by impact location. For example, releasing the pendulum arm from 40° was representative of the 88.2 percentile impact to the front location, 90.4 percentile impact to the side location, 81.4 percentile impact to the back location, and 80.7 percentile impact to the top location. This demonstrates that higher head accelerations were more commonly associated with back and top impact locations in the on-ice helmet data. The tails of these right-skewed distributions exhibited similar trends. Releasing the pendulum arm from 70° was representative of the 98.2 percentile impact to the front location, 98.6 percentile impact to the side location, 95.5 percentile impact to the back location, and 98.9 percentile impact to the top location.

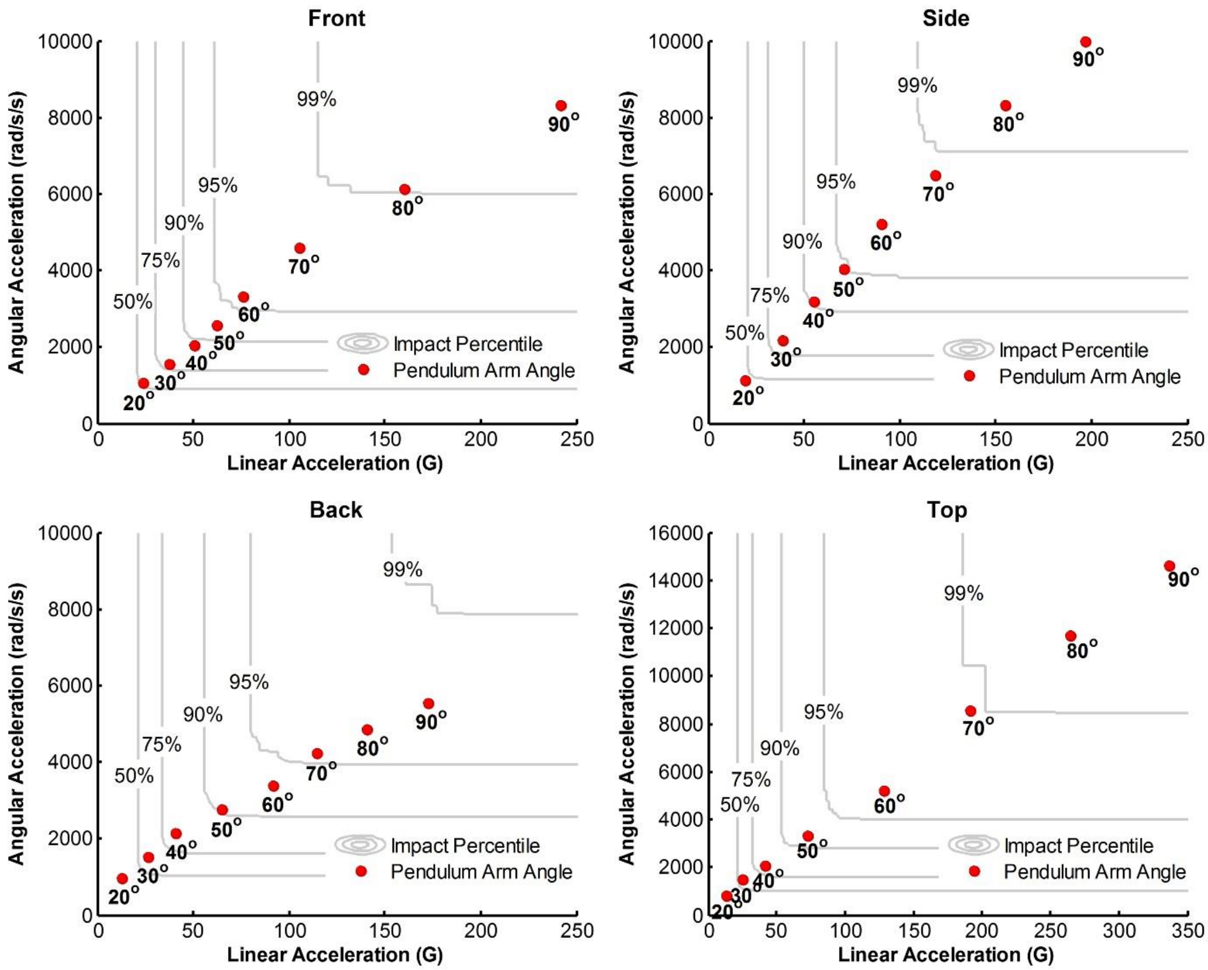

Figure 10: Peak linear and rotational head acceleration values generated during the pendulum tests are overlaid on the bivariate CDFs for each impact location. These plots relate laboratory impact energies to on-ice head impact data and were used to define head impact distributions as a function of impact energy. Where a given impact energy (pendulum arm angle) fell within the distributions varied by impact location. While these plots only illustrate this for male collegiate hockey, this was done for each of the 4 hockey player populations in which on-ice data were previously collected.

On-ice head acceleration distributions were transformed to impact energy distributions (represented by pendulum arm angle) by determining the percentage of on-ice data that fell below each energy for each impact location. This process was done for each population (male and female collegiate, male and female high school). Resulting impact energy CDFs were then averaged to determine an overall impact energy CDF that gave equal weighting to each population (Figure 11). The impact energy CDFs were related to generalized impact energy conditions: a low energy condition (40° pendulum arm angle), a medium energy condition (65° pendulum arm angle), and a high energy condition (90° pendulum arm angle). For all locations, the low energy condition accounts for greater than 90% of head impacts. The medium energy condition ranged between 3.2% and 6.8% of impacts for each condition. The high energy condition generally accounted for less than 1% of impacts for each location, with the exception of the back location. From this analysis, weightings were determined for each laboratory impact condition based on how frequently a player might sustain a similar impact (Table 7). Summating these laboratory condition-specific exposure values results in the 227 head impacts that the average player experiences throughout a season of hockey.

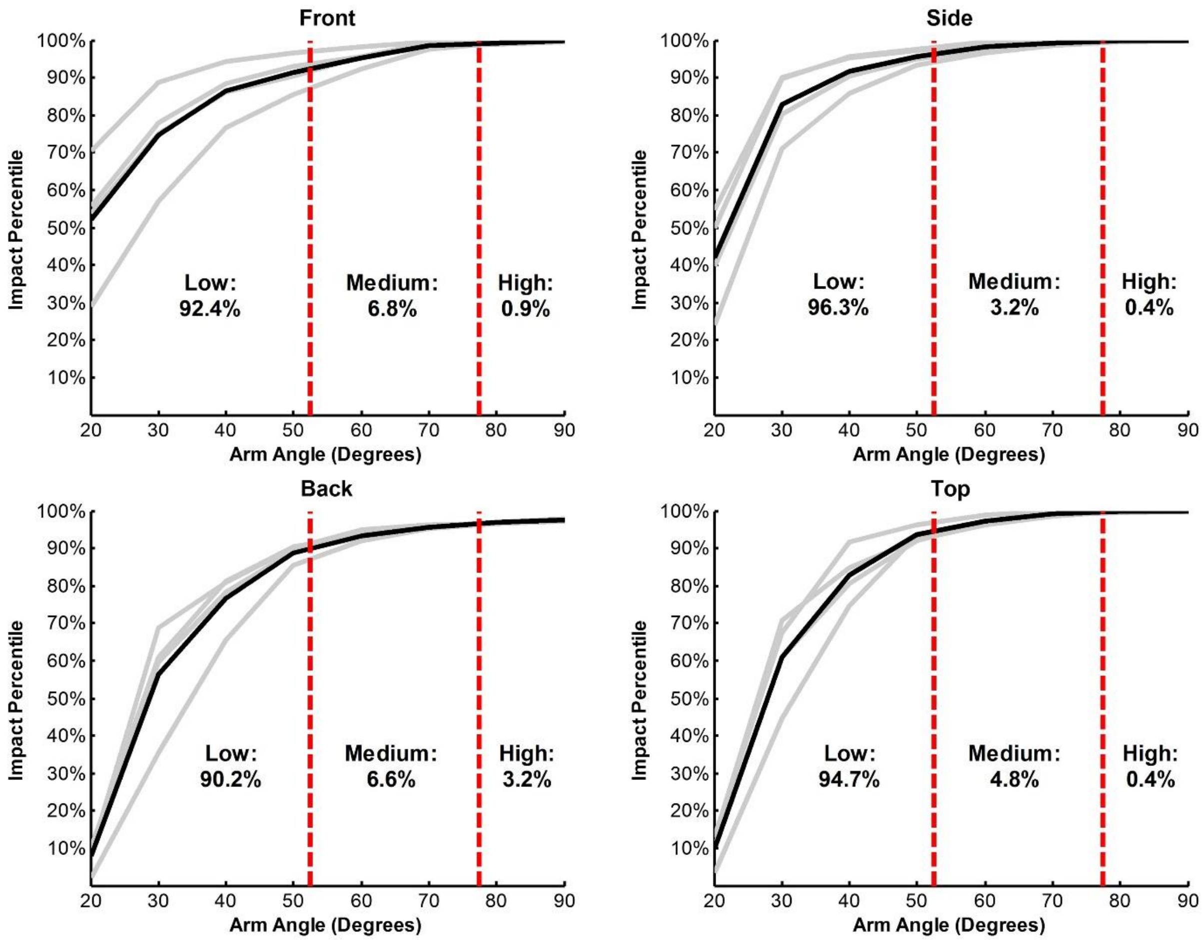

Figure 11: Impact energy CDFs for each impact location resulting from the transformation of on-ice data to laboratory impact conditions. The gray lines represent impact energy CDFs for each population and the black line is the equal-weight average of the four populations. The dashed red lines show the bounds used to determine the percentage of impacts at each location associated with the low, medium, and high energy impact conditions. This analysis was used to define the exposure weightings for each impact configuration in the Hockey STAR formula.

Table 7: Mapping of on-ice head impact exposure to generalized laboratory test conditions. Each impact configuration was related to a number of impacts that the average player experience during a season of play. These numbers represent the exposure weightings for each test condition in the Hockey STAR formula.

	40°	**65°**	**90°**	**Total**
Front	62.9	4.6	0.6	**68.1**
Side	65.6	2.2	0.3	**68.1**
Top	21.5	1.1	0.1	**22.7**
Back	61.4	4.5	2.2	**68.1**
Total	**211.4**	**12.4**	**3.2**	**227**

Exemplar Hockey Helmet Tests

Three helmets were evaluated with the Hockey STAR evaluation methods described above: two hockey helmets and one football helmet. The detailed results for each helmet are shown in Tables 8-10. Hockey STAR values were 7.098 for hockey helmet A, 12.809 for hockey helmet B, and 1.213 for the football helmet. Lower STAR values equate to lower risk of concussion. Given the assumptions that all players experience an identical head impact exposure to that which was modeled and had the same concussion tolerance to head impact, these STAR values suggest that the concussion rate for players in hockey helmet A would be 44.6% less than that of players in hockey helmet B. Comparing the hockey helmets to the football helmet, players in the football helmet would experience concussions rates 82.9% less than players in hockey helmet A and 90.5% less than players in hockey helmet B.

Table 8: Hockey STAR evaluation of hockey helmet A helmet with resultant peak linear (a) and angular (α) acceleration, corresponding risk of injury, and season exposure for each condition to calculate incidence. The resulting Hockey STAR value is 7.098.

Impact Location	Angle (degrees)	Peak a (g)	Peak α (rad/s^2)	Risk of Injury	Exposure per Season	Incidence per Season
Front	40	64	2154	0.34%	62.9	0.213
Front	65	108	3591	5.94%	4.6	0.273
Front	90	168	6680	86.57%	0.6	0.519
Side	40	71	4220	2.39%	65.6	1.568
Side	65	124	7149	64.74%	2.2	1.424
Side	90	176	9370	98.34%	0.3	0.295
Top	40	37	2590	0.16%	21.5	0.035
Top	65	103	6061	26.23%	1.1	0.289
Top	90	263	12666	99.99%	0.1	0.100
Back	40	41	2020	0.12%	61.4	0.072
Back	65	111	4345	11.43%	4.5	0.514
Back	90	169	6076	81.60%	2.2	1.795
					STAR	**7.098**

Table 9: Hockey STAR evaluation of hockey helmet B with resultant peak linear (a) and angular (α) acceleration, corresponding risk of injury, and season exposure for each condition to calculate incidence. The resulting Hockey STAR value is 12.809.

Impact Location	Angle (degrees)	Peak a (g)	Peak α (rad/s^2)	Risk of Injury	Exposure per Season	Incidence per Season
Front	40	64	2570	0.48%	62.9	0.299
Front	65	87	3819	3.21%	4.6	0.148
Front	90	164	6333	81.58%	0.6	0.489
Side	40	74	5037	5.04%	65.6	3.305
Side	65	115	8254	75.17%	2.2	1.654
Side	90	155	10189	98.12%	0.3	0.294
Top	40	66	3869	1.47%	21.5	0.315
Top	65	124	7001	61.60%	1.1	0.678
Top	90	163	9548	97.72%	0.1	0.098
Back	40	56	3448	0.71%	61.4	0.435
Back	65	135	6647	65.27%	4.5	2.937
Back	90	178	9073	98.07%	2.2	2.158
					STAR	**12.809**

Table 10: Hockey STAR evaluation of a football helmet with resultant peak linear (a) and angular (α) acceleration, corresponding risk of injury, and season exposure for each condition to calculate incidence. The resulting Hockey STAR value is 1.213.

Impact Location	Angle (degrees)	Peak a (g)	Peak α (rad/s^2)	Risk of Injury	Exposure per Season	Incidence per Season
Front	40	37	1787	0.08%	62.9	0.052
Front	65	76	2679	0.84%	4.6	0.039
Front	90	115	3646	8.21%	0.6	0.049
Side	40	35	2210	0.11%	65.6	0.072
Side	65	64	3940	1.47%	2.2	0.032
Side	90	122	7120	61.95%	0.3	0.186
Top	40	32	1965	0.08%	21.5	0.017
Top	65	67	3554	1.20%	1.1	0.013
Top	90	100	4622	9.28%	0.1	0.009
Back	40	44	2177	0.16%	61.4	0.096
Back	65	78	3886	2.37%	4.5	0.107
Back	90	109	5644	24.60%	2.2	0.541
					STAR	**1.213**

Discussion

The purpose of this paper is to introduce a new evaluation system for hockey helmets that can provide information to consumers on the relative performance of different helmets. Hockey STAR is in no way meant to diminish the importance of, or replace, the current ASTM standards enforced by HECC. Since the introduction of these standards and other rule changes in the game, the rate of catastrophic head injuries has greatly decreased [51]. The standards also require important specifications regarding the elongation of the chin strap and appropriate area of coverage of helmets. The Hockey STAR evaluation system intends to only test hockey helmets that have already been certified by HECC. HECC and other helmet certifications are analogous to the Federal Motor Vehicle Safety Standards (FMVSS) and regulations which have pass/fail standards. These standards provide baseline safety requirements that are crucial for protecting drivers. The New Car Assessment Program (NCAP) developed by the National Highway Traffic Safety Administration (NHTSA) augments the existing standards by providing consumers with a rating system to help guide their selections [52, 53]. Hockey and Football STAR serve the same purpose as NCAP: to provide additional information to consumers after the minimum safety requirements have been met through certification.

Advances from Football STAR

Like Football STAR, Hockey STAR is based on two fundamental principles: 1) helmets that lower head acceleration reduce concussion risk and 2) each test is weighted based on how often players experience similar impacts. An Institute of Medicine (IOM) report on sport-related concussion in youth reviewed Football STAR and characterized it as a theoretically grounded approach to evaluating helmet protection that is based on sound principles [54]. However, the report also noted that adding rotational acceleration to the methodology would increase its wide-spread application. Considering this recommendation, Hockey STAR was developed to evaluate helmets using both linear and rotational head acceleration. This addition contributed to the unique head impact exposure analysis in Hockey STAR. The exposure distributions used to weight each impact configuration included both linear and rotational head acceleration from collegiate hockey players [40]. The total number of impacts over one season was also an average of impacts experienced by youth boy's and collegiate men's and women's hockey, since the same helmet models are used for all ages and gender with variations only in helmet size [40, 41]. This is one of two key differences between Football STAR and Hockey STAR.

The second key difference is that Hockey STAR accounts for a higher underreporting rate of concussion than Football STAR. The bivariate risk function was developed with the assumption that only 10% of concussions sustained by players are diagnosed by physicians [29, 49]. In contrast, the Football STAR risk function assumes that 50% of concussions sustained by players are diagnosed by physicians [23, 50]. Recent studies have suggested that the underreporting rate may be much greater than 50%, and have even suggested that structural changes as a result of cumulative head impact exposure in the absence of diagnosed concussion [55-57]. Because the risk function utilized by Hockey STAR assumes that 90% of concussion go unreported, the Hockey STAR values are not anticipated to be predictive of the number of diagnosed concussions sustained by hockey players, but rather the total number of injuries sustained, diagnosed and undiagnosed.

Biofidelity of Impact Model

The biofidelity of the impact model used for Hockey STAR was ensured through appropriate headform selection and comparison of acceleration traces with other data collected from hockey players. The NOCSAE headform was chosen because of its superior helmet fit at the base of the skull, and around the jaw, cheeks, and chin compared to that of the Hybrid III headform [48]. A helmet that does not fit properly can shift on the head during tests, and if the contact area of the helmet padding with the headform varies from what is realistic, the effective stiffness of the padding will vary, potentially resulting in a mischaracterization of a helmet’s energy management capabilities.

The headform responses generated from pendulum impacts in the lab were compared to on-ice data by generating corridors from both on-ice player data and ice rink testing with a Hybrid III head (Figure 12, Figure 13). The lab impacts fell within the response corridors generated from both datasets with the exception of the top impacts in the lab compared with the top impacts from ice rink testing. There are two reasons for this difference. The first is that the top impacts for the ice rink testing were pure axial loading to the top of the headform, while the Hockey STAR top location is non-centric and meant to generate rotational acceleration. The second reason is that the ice condition was not tested for the top location on the ice rink, so only boards and glass responses are averaged. These impacts are longer in duration and not representative of the full spectrum of impacts seen by ice hockey players. Overall, this analysis provides further evidence that the laboratory testing is representative of head impacts in hockey.

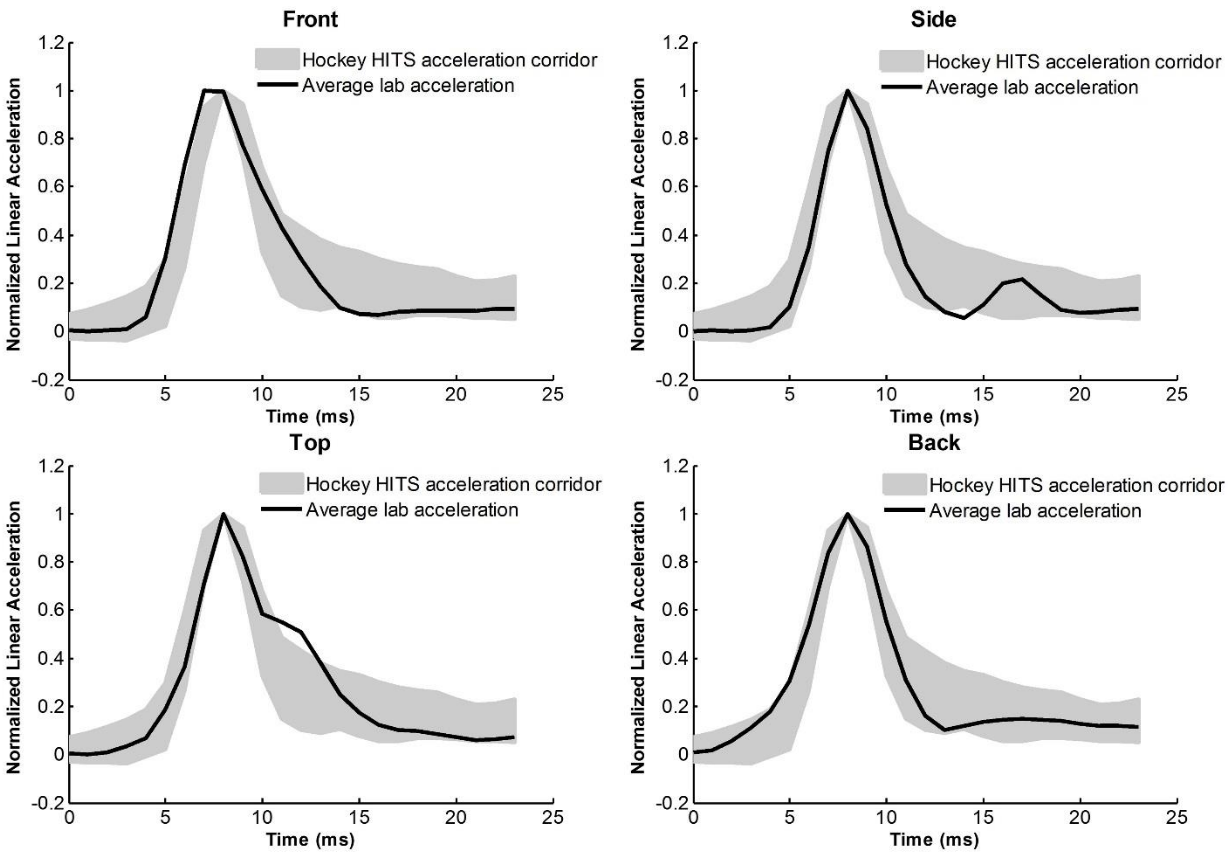

Figure 12: Average acceleration traces from the laboratory pendulum tests were compared to corridors developed from on-ice volunteer data by impact location. The head impact response of the laboratory tests closely matches that which was measured directly from hockey players, suggesting the impact system generates a biofidelic response.

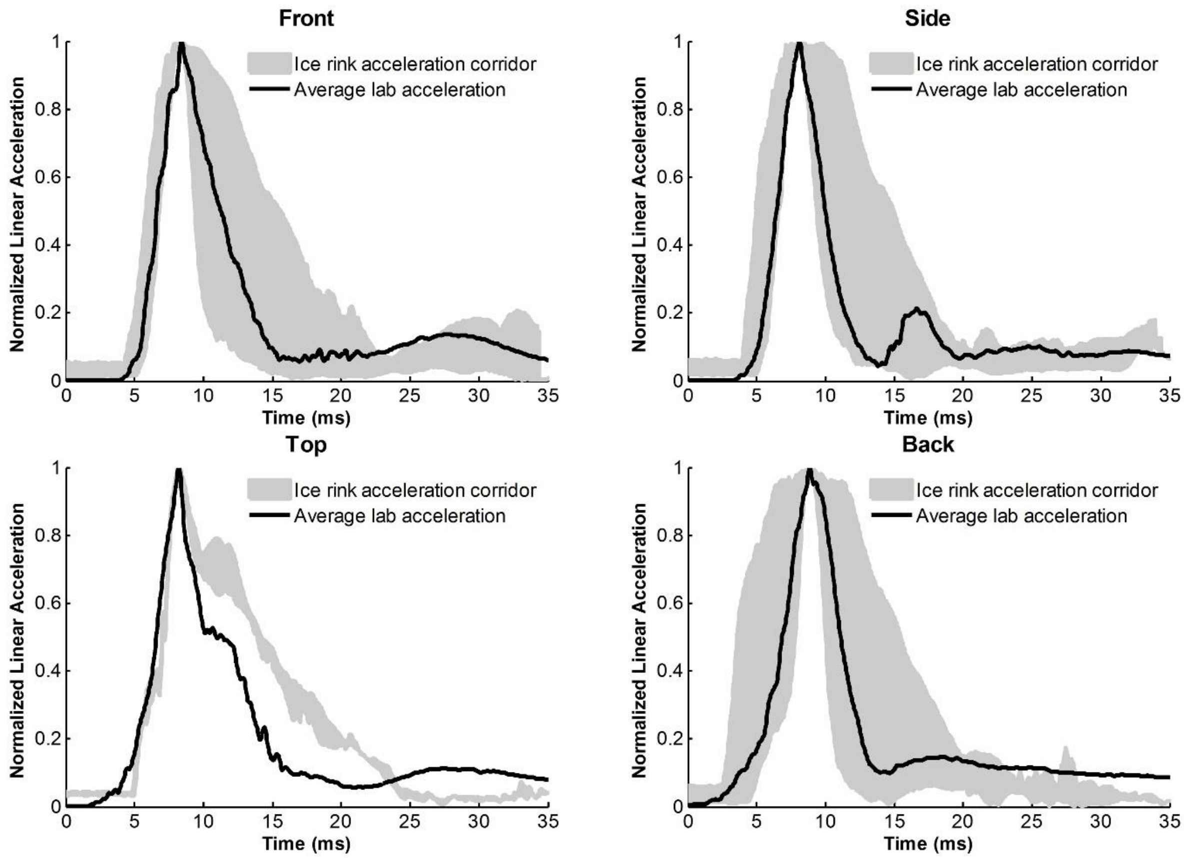

Figure 13: Head impact responses generated in the lab were also compared to dummy head impacts collected at an ice rink. Here, average acceleration traces from the laboratory pendulum tests were compared to corridors developed from controlled dummy head impacts to the boards, glass, and ice at an ice rink. The head impact response of the laboratory tests closely matches that which was measured at the ice rink, which further suggests that impact system generates a biofidelic response.

Implementing Hockey STAR

Given that there are 32 helmets currently on the market, a total of 1536 impact tests are required to evaluate all hockey helmets using the proposed protocol. While this methodology proposes a reasonable number of tests to evaluate helmets, there are practical limitations to the number of tests that can be run. For this reason, there are other variables that have been considered and researched. For example, helmet temperature is not varied in this protocol. We performed a study investigating the temperature inside football helmets during games [58]. When a player wears a helmet, the temperature of the padding will approach that of the head. For this reason, and that fact that testing multiple temperatures could double or triple the number of tests, helmet temperature is not varied in the Hockey STAR protocol. Additionally, Hockey STAR does not evaluate helmets with a facemask on. There are a number of facemask configurations that can be used on a helmet. These include full cage facemasks and clear visors. Testing in the lab demonstrated that the facemask does not significantly affect either linear or rotational head acceleration, with differences less than 2%. This suggests that hockey helmet performance is not influenced by the presence of a facemask, and that testing with and without facemasks is not necessary. In short, there are a near-infinite number ways to test a helmet, but there are practical limitations to the number of tests used to evaluate products.

Star Rating Thresholds

The Hockey STAR methodology will ultimately be used to apply star ratings to hockey helmets, which allows consumers to easily compare overall helmet performance between models. While this is already being done with football helmets, the STAR value thresholds used to determine the star ratings of football helmets cannot simply be applied to hockey helmet evaluations due to a number of key differences in the Hockey STAR and Football STAR formulas. The impact exposure weightings are specific to each sport, the test conditions differ, and a more conservative risk function is used in the Hockey STAR. Current football helmet ratings were re-analyzed using a similarly conservative risk function for linear head acceleration [24]. The differences in test conditions were also accounted for by comparing results of the exemplar football helmet tested under Hockey STAR conditions to the results of the same helmet tested with Football STAR. These equivalent Hockey STAR thresholds would hold hockey helmets to the same performance criteria that football helmets are held to.

An objective of Hockey STAR is to inform consumers of relative differences between existing helmets. There are major differences between current hockey helmet and football helmet designs, so holding hockey helmets to the same STAR performance criteria would likely not be informative to consumers. For this reason, the proposed thresholds for Hockey STAR will better identify differences among current hockey helmets, while still providing incentive to manufacturers to advance helmet design. Table 11 compares these proposed Hockey STAR thresholds to the current thresholds used in Football STAR and the equivalent Hockey STAR thresholds. As hockey helmets advance in future years, these star rating thresholds may be modified to be more sensitive to future helmet designs, similar to how NCAP rescales the star rating criteria for automobiles as crash performance increases.

Table 11: Comparison of the proposed Hockey STAR rating thresholds to the current thresholds used in Football STAR and Hockey STAR thresholds that are equivalent to current Football STAR thresholds using the proposed methodology. To earn a number of stars, a helmet's STAR value must be below the specified threshold. The Proposed Hockey STAR thresholds will better identify relative differences among existing hockey helmets, as most current hockey helmets are likely to.

Star Rating	Current Football STAR	Equivalent Hockey STAR	Proposed Hockey STAR
5	0.300	1.463	2.000
4	0.400	2.069	4.000
3	0.500	2.676	6.000
2	0.700	3.889	8.000
1	1.000	5.708	10.000

Exemplar Hockey STAR Results

For the three helmets tested using the Hockey STAR methodology, the Hockey STAR values were 7.098, 12.809, and 1.213 for helmet A, helmet B, and the football helmet, respectively. These values are related to the relative risk of concussion, such that a player wearing helmet A would be 44.6% less likely to sustain a concussion than a player wearing helmet B if both players had the same head impact exposure over one season. Similarly, if a player wore the football helmet and also had the same head impact exposure, that player would be 82.9% less likely to sustain a concussion than a player wearing helmet A, and 90.5% less likely than a player wearing helmet B. Again, it is important to note that these STAR values are not representative of the number of diagnosed concussions players will experience, but rather an overall estimate of undiagnosed and diagnosed injuries combined. While these values are tied to concussion risk, ultimately the rating system identified helmets that best reduce head acceleration throughout the range of head impacts that hockey players experience.

Given the proposed thresholds outlined in Table 11, helmet A would be rated as a 2 star helmet, helmet B would not be recommended, and the football helmet would receive a 5 star rating. The disparity in performance between the football and hockey helmets can be attributed to the differences in padding and design for energy attenuation. Specifically, the football helmet has a greater offset, which allows more compression during impact when modulating the impact energy. This enables the padding system to compress on lower severity impacts and not bottom out on higher severity head impacts. The hockey helmets' padding systems are much thinner, restricting the ability to reduce head acceleration throughout the full range of head impacts experienced by players.

Conclusions

This paper presents a novel methodology for comparing the performance of different hockey helmets. The methods are comparable to the existing Football STAR rating system, however the equation has been updated to include both linear and rotational acceleration. The exposure and testing conditions were also modified to represent the number and type of head impacts experienced by hockey players. A new impact pendulum was designed and built for laboratory testing, and the biofidelity of the system was ensured by comparison with on-ice player data and other testing methods. Given that Hockey STAR will be used to rate hockey helmets, exemplar tests of existing helmets were performed to evaluate and compare of the ability of a small sample of helmets to reduce risk of concussion.

Similar outcomes to those resulting from Football STAR are anticipated for Hockey STAR. Consumers will use the hockey helmet evaluations as a purchasing tool, which will drive manufacturers to advance hockey helmet design to reduce concussion risk. This reduction in concussion risk measured in the lab will translate to hockey players because the laboratory evaluations are representative of head impacts experienced by hockey players.

Finally, it is important to note that no helmet can completely protect a player from all head injuries, and there are always risks associated with playing the sport. The analysis presented here is based on trends and probabilities, but an individual's risk of concussion may vary with a number of factors such as prior history of head injury or genetic predispositions.

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Chapter 4

Biomechanical Performance of Hockey Helmets

Abstract

Ice hockey is a fast-paced sport with a high rate of concussion. Helmet safety standards introduced over 40 years ago have greatly reduced the number of serious and fatal head injuries in hockey, however, until recently there was no objective measure of the relative impact performance of hockey helmets that pass the standard. The Hockey STAR methodology was developed to address the relative performance of hockey helmets and their ability to reduce the risk of concussion. The objective of this study was to analyze the kinematic parameters and overall performance of currently available helmet models using the Hockey STAR methodology. Thirty-seven helmet models were evaluated with an impact pendulum at 4 locations and 3 impact energies, resulting in a total of 1,776 impact tests. The results showed that there are large differences in impact performance among currently available helmet models, with the best performing helmet reducing concussion risk by approximately 70% compared with the worst performing helmet. These findings support the need for an objective measure of relative performance for hockey helmets.

Introduction

Ice hockey participation is rapidly increasing in the United States, with the number of amateur athletes nearly tripling over the past 25 years. Ice hockey has one of the highest rates of concussion among all sports [1-3]. Concussions and more severe head injuries are a primary safety concern in ice hockey given the high rate of injuries and potential long-term neurocognitive effects [4-7]. There are a number of strategies that can be used to reduce the incidence of concussion in sports, one of which is optimizing the performance of protective equipment through design changes [8, 9].

Early hockey helmets were not widely used, and most consisted of strips of leather lined with felt. In the mid-1960s, helmet companies started using injection-molding techniques to manufacture plastic shells from either high-density polyethylene or polycarbonate, which are still used today [10]. Successful helmet models from this time period look similar to what players wear today. Most had a 2-piece adjustable plastic shell with a thin layer of foam lining. Following the death of NHL player Bill Masterton in 1968 from a head injury on the ice, more players began to voluntarily wear helmets, and many leagues started mandating their use. That same year, two teenage players in Canada died as a result of head injuries while wearing helmets, which prompted the Canadian Amateur Hockey Association to request that the Canadian Standards Association (CSA) develop safety standards for helmet performance [11]. The first CSA standard was not published until 1975, and many helmets did not meet the minimum criteria for impact performance at that time. When the first CSA standard was introduced, manufacturers had to make a shift to adding thicker, higher-density foams for better impact attenuation. For some helmet models, extra padding was added to the outside of the shell to pass the standard without drastically altering the manufacturing process. The NHL did not begin mandating certified helmet use until 1979, and even then, the rule only applied to new players [12]. Helmet designs have changed little since the first safety standard was introduced. The shells have essentially remained the same, and the foam liners have been made from a variety of materials which usually measure around 1.5 – 2.5 cm in thickness.

Safety standards have saved countless lives by requiring a minimum impact performance for protective equipment and other products. Effects of these standards have been quantified for applications such as football and automotive safety. For example, after implementation of helmet standards by the National Operating Committee on Standards for Athletic Equipment (NOCSAE), fatal head injuries were reduced in football by approximately 74% [13]. Similar trends were seen with mandatory safety standards for motor vehicles, the Federal Motor Vehicle Safety Standards (FMVSS), introduced in the mid-1960s. In 1978, the National Highway Traffic Safety Administration (NHTSA) began evaluating the relative safety of vehicles through the New Car Assessment Program (NCAP) to inform consumers and provide incentive to manufacturers to improve safety performance [14]. Through the combination of minimum safety standards and relative performance ratings, the fatality rate of motor vehicle occupants had been reduced by 81% as of 2012 [15]. Ultimately, these standards have been effective because they limit the amount of energy transferred to the head during impact.

While helmet safety standards have been very effective at eliminating nearly all catastrophic head injuries in sports, they do not address the ability of helmets to reduce concussion risk. Until recently, no consumer information existed on the relative impact performance of helmets, partly due to challenges associated with summarizing the results of a large number of tests. The star ratings implemented by NHTSA through NCAP were used as a model to summarize the impact performance of helmets because of the effectiveness of that program. Football helmets were the first to be evaluated using a five star rating system [16]. The number of five star football helmets has increased each year since the introduction of the ratings. Additionally, the reduction in injury risk predicted by the evaluation system has been supported by on-field research [9, 17, 18]. The football helmet rating system was later adapted to evaluate hockey helmets [19]. The objective of this study was to describe the overall impact performance of currently available hockey helmets by quantifying and comparing various metrics of impact severity.

Methods

A total of 1,776 impact tests were performed to evaluate 37 hockey helmet models (Table 12) using the Hockey STAR methodology [19]. All hockey helmet models available for purchase at the time of the study were tested.

Table 12: Hockey helmets evaluated and analyzed for the current study, grouped by manufacturer. These helmets represent all models available for purchase at the time of the study.

Bauer	**CCM**	**Easton**	**Reebok**
2100	Fitlite	E300	3K
4500	Fitlite 40	E400	4K
5100	Fitlite 60	E600	5K
7500	Fitlite 80	E700	7K
9900	Resistance		8K
IMS 7.0	Resistance 100		11K
IMS 9.0	Resistance 300		
IMS 11.0	Vector V04		
Re-Akt	Vector V06		
Re-Akt 100	Vector V08		

Mission	**Tour**	**Warrior**
Inhaler	Spartan GX	Krown 360
M15	Spartan ZX Pro	Krown LTE
		Krown PX3

Hockey STAR Test Methods

The hockey helmet evaluation system (Hockey STAR) used for this study was based on two fundamental principles. The first is that each laboratory test is weighted based on how frequently similar impacts occur on the ice, and the second is that helmets that lower head acceleration reduce the risk of concussion. The exposure weightings for different locations and impact severities were determined from instrumented hockey players, while the risk of concussion was calculated with a previously published bivariate risk function [19, 20]. To evaluate each helmet, the appropriate size was selected and fit to a medium NOCSAE headform. The headform was mounted on a Hybrid III 50th percentile neck after several modifications. Material was removed from the headform to better position the neck with respect to the CG of the head, and a custom neck mount was made for attachment [21, 22]. The headform was instrumented with a 6 degree of freedom sensor package that consisted of 3 linear accelerometers and 3 angular rate sensors (6DX-Pro, DTS, Seal Beach, CA). Data were collected at a sampling rate of 20,000 Hz. The headform and neck were attached to a slide table that simulated the effective mass of the torso (Figure 14). The slide table had 5 degrees of freedom to adjust the impact location and direction of force on the helmet.

The helmeted head and neck assembly was struck with an impact pendulum designed to simulate head impacts in sports. The length of the pendulum arm from the pivot point to the center of the impacting face was 190.5 cm, with a 16.3 kg impacting mass attached. The flat, rigid nylon impacting face was 12.7 cm in diameter. Two helmet samples were evaluated per helmet model. Each sample was impacted twice at 3 energies (40°, 65°, and 90° pendulum arm angles) and 4 locations (front, side, top, and back), for a total of 24 impacts per sample. The pendulum arm angles equate to impact velocities of 3.0, 4.6, and 6.1 m/s. The front and back locations were centric, meaning the direction of force was aligned with the CG of the headform, while the side and top locations were non-centric (Figure 14, Table 13).

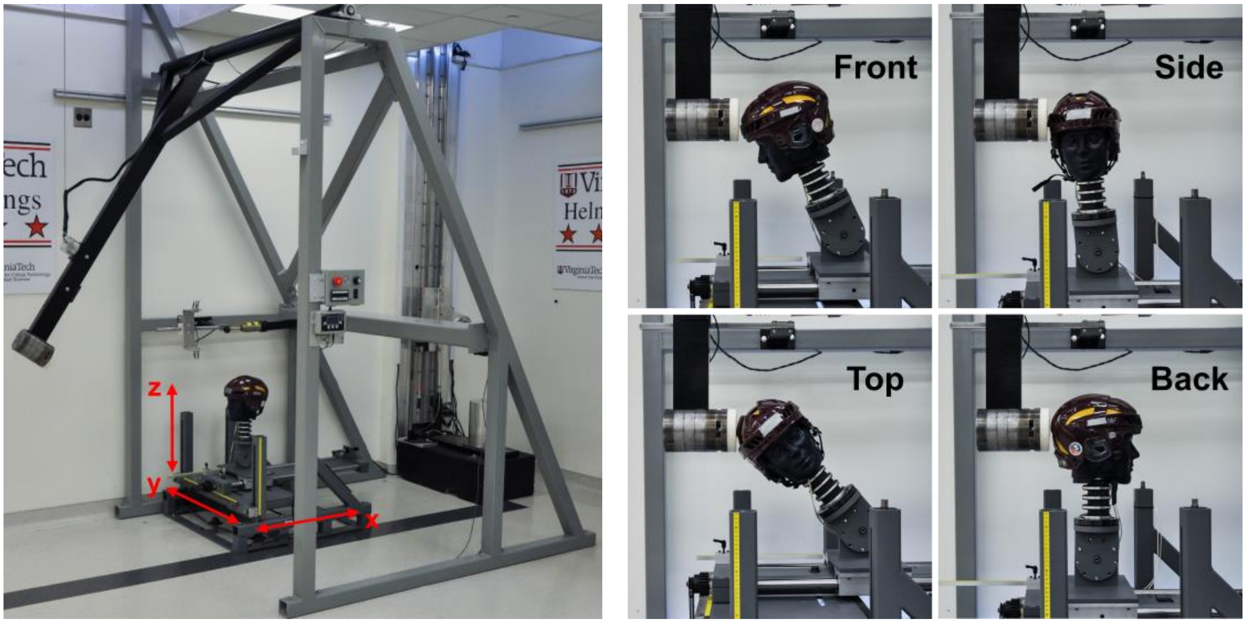

Figure 14: [Left] The impact pendulum device used for helmet evaluation struck a NOCSAE headform mounted on a Hybrid III 50th percentile neck. The head and neck were mounted on a sliding mass that simulated the effective mass of the torso during impact. The slide table had 5 degrees of freedom so that any location on the helmet could be impacted: translation along the x axis, translation along the y axis, translation along the z axis, rotation about the y axis, and rotation about the z axis. [Right] Front, side, top, and back impact locations used to assess helmet performance. The side and top impact locations were non-centric, meaning the direction of force was not aligned with the CG of the headform; while the front and back impact locations were centric [19].

Table 13: Translation and rotation measurements for headform initial positions for each impact location. Y- and Z-axis translation measurements were made with respect to the headform in a position where the median (midsagittal) and basic (transverse) plane intersection was aligned with the center of the impactor with 0° Y- and Z-axis rotation, using the SAE J211 coordinate system. The position on the X-axis was defined as +1.25 cm before the helmeted headform contacted the pendulum in a neutral vertical position for all locations.

	Y Translation (cm)	Z Translation (cm)	Y Rotation (°)	Z Rotation (°)
Front	0.0	+1.9	-25	0
Side	+3.1	+7.3	-5	-80
Top	-2.7	-2.7	-40	-90
Back	0.0	+5.9	0	-180

After completing all impact tests, acceleration data were filtered to channel frequency class (CFC) 1000 and angular rate data were filtered to CFC 155 using a 4-pole phaseless Butterworth low-pass filter. All data were transformed to the same coordinate system at the center of gravity of the headform. Rotational acceleration was calculated by differentiating angular rate. Peak linear and rotational acceleration values were recorded for each impact. A bivariate risk function was used to calculate risk of concussion for each test condition using the average peak linear (a) and rotational (α) accelerations from repeated tests (Equation 1) [20].

$$R(a,\alpha) = \frac{1}{1+e^{-(-10.2+0.433*a+0.000873*\alpha-0.000000920*a\alpha)}} \quad (1)$$

The Hockey STAR value for each sample tested was then calculated using Equation 2 and the weighting values for each condition determined from head impact exposure (E) in hockey players as a function of impact location (L) and energy (θ) (Table 14). The Hockey STAR value represents predicted injury incidence for a player that experiences the same impact exposure used in the equation. Incidence (exposure multiplied by risk) was calculated for each impact condition, and then summed across all conditions for a single Hockey STAR value for each helmet sample.

$$Hockey\ STAR = \sum_{L=1}^{4}\sum_{\theta=1}^{3} E(L,\theta) * R(a,\alpha) \qquad (2)$$

Table 14: Weighting values for each test condition that represent how often the average player experiences similar impacts. These numbers were the head impact exposure values used in the Hockey STAR equation [19].

	40°	**65°**	**90°**	**Total**
Front	62.9	4.6	0.6	**68.1**
Side	65.6	2.2	0.3	**68.1**
Top	21.5	1.1	0.1	**22.7**
Back	61.4	4.5	2.2	**68.1**
Total	**211.4**	**12.4**	**3.2**	**227.0**

Helmet Performance Analysis

To describe the distribution of impact performance among all hockey helmets, peak linear and rotational accelerations were summarized by the median [5^{th}-95^{th} percentile] for each impact severity. Accelerations were also summarized by impact location and severity with box plots to show the median, interquartile range (IQR), and spread of data (excluding outliers).

To describe differences in overall impact performance between helmet samples, the distribution of Hockey STAR values was quantified and fit to a log-normal probability density function. The values were also summarized with the median, full range, and IQR. Variability in Hockey STAR values for samples of the same model was assessed by dividing the range of the two values by the average, and was expressed as a percent. Regression analyses were performed to investigate the relationship between the Hockey STAR value for each sample tested and 4 different variables: linear acceleration, rotational acceleration, risk of concussion, and predicted incidence of injury. The variables were stratified by impact severity to evaluate correlations for different types of impacts. The coefficient of determination (R^2) was used to compare correlations.

To assess the relationship between kinematic parameters and various brain injury criteria, a scatter plot matrix was generated. The kinematic parameters evaluated were: peak linear acceleration, peak rotational acceleration, and peak rotational velocity. The brain injury criteria evaluated were (Table 15): concussion correlate (CC), severity index (SI), head injury criterion (HIC), brain injury criterion (BrIC), and head impact power (HIP). CC is the linear portion of the bivariate risk function (Equation 1) used with the Hockey STAR evaluation system [20]. The risk function was developed using on-field data from high school and collegiate football players in a multivariate logistic regression analysis to model risk of concussion as a function of linear (a) and rotational (α) head acceleration. SI and HIC are both functions of linear acceleration (a) and time (t), and are based on the Wayne State Tolerance Curve (WSTC) [23-25]. The WSTC was the result of a combination of cadaver skull fracture data, animal models, and human volunteer data to determine the maximum tolerable levels of linear acceleration for different impact durations [26-29]. Acceleration in these functions is weighted to a greater degree than impact duration. SI is commonly used as a metric for helmet safety standards, while HIC is used most notably in automotive safety standards and performance ratings. BrIC was proposed by NHTSA as a rotational brain injury metric to supplement HIC in automotive and other safety standards [30]. The criterion was developed using finite element models of the human brain, and a combination of scaled animal injury data and a variety of anthropomorphic test device (ATD) data. BrIC is a function of rotational velocity in each direction (ω_x,ω_y,ω_z) normalized by critical values (ω_{xC},ω_{yC},ω_{zC}). HIP is a function of linear (a_x,a_y,a_z) and rotational (α_x,α_y,α_z) acceleration about each axis, and time (t), and represents the rate of change of kinetic energy of the head. It was developed using reconstructions of concussive impacts in professional football players [31]. R^2 values were used to compare the relationships between the kinematic parameters and brain injury criteria outlined above.

Table 15: Equations for all brain injury criteria used to evaluate the relationships between kinematic parameters and various criteria for all Hockey STAR tests.

Criterion	Equation
CC	$-10.2 + 0.0433a + 0.000873\alpha - 0.000000920a\alpha$
SI	$\int a^{2.5}dt$
HIC	$max\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right]^{2.5}(t_2 - t_1)$
BrIC	$\sqrt{\left(\frac{\omega_x}{\omega_{xC}}\right)^2 + \left(\frac{\omega_y}{\omega_{yC}}\right)^2 + \left(\frac{\omega_z}{\omega_{zC}}\right)^2}$
HIP	$4.5a_x\int a_x dt + 4.5a_y\int a_y dt + 4.5a_z\int a_z dt +$ $0.016\alpha_x\int \alpha_x dt + 0.024\alpha_y\int \alpha_y dt + 0.022\alpha_z\int \alpha_z dt$

Results

Between all helmet models and impact locations, median [5th-95th percentile] peak linear accelerations were 49 [33-63] g for low severity impacts, 103 [74-160] g for medium severity impacts, and 211 [131-371] g for high severity impacts (Figure 15). Median peak rotational accelerations were 2745 [1866-4139] rad/s^2 for low severity impacts, 5210 [3301-8159] rad/s^2 for medium severity impacts, and 8732 [5504-14568] rad/s^2. Median peak rotational velocities were 18 [15-21] rad/s for low severity impacts, 28 [22-32] rad/s for medium severity impacts, and 37

[29-41] rad/s for high severity impacts. Acceleration distributions varied by impact location and severity (Figure 16). Variance increased with increasing severity for all locations, and was generally greater for the top location than other locations. The wide range in acceleration values shows the substantial differences in impact performance among currently available helmet models.

Hockey STAR values for individual samples varied greatly, with a range of 2.67 to 10.85. The median STAR value was 6.12, with IQR of 5.08 to 8.23. The distribution of STAR values was right-skewed, and best fit by a log-normal probability density function (Figure 17). The median variability in STAR values between samples of the same model was 5.0%, with an IQR of 2.0% to 8.5%.

The relationships between Hockey STAR values and different impact severity measures (linear acceleration, rotational acceleration, risk, and incidence) were quantified with linear regressions (Figure 18). The parameters were averaged across all locations, but stratified by impact severity for each helmet sample tested. Lower STAR values were associated with lower accelerations (both linear and rotational), which resulted in lower risk and incidence values. For current helmet models, the medium impact severity exhibited the highest correlation with overall performance. Risk, and therefore incidence, was saturated at the high severity condition for most samples with a Hockey STAR value greater than 6. As helmet performance decreased (higher STAR values), variance increased for all parameters.

The correlations between 3 different kinematic parameters and 5 brain injury criteria emphasize at a high level which parameters provide similar information on impact severity (Figure 19). All injury criteria were highly correlated with one another ($R^2 \geq 0.89$), with the exception of BrIC. The highest correlation was between SI and HIC ($R^2 = 1.00$), and the lowest was between BrIC and peak rotational acceleration (R^2 = 0.45). Peak linear acceleration was highly correlated with all criteria except for BrIC, however, CC was the only criteria highly correlated with both linear and rotational acceleration ($R^2 \geq 0.91$). BrIC was highly correlated with peak rotational velocity (R^2 = 0.97), which was not highly correlated with other injury criteria.

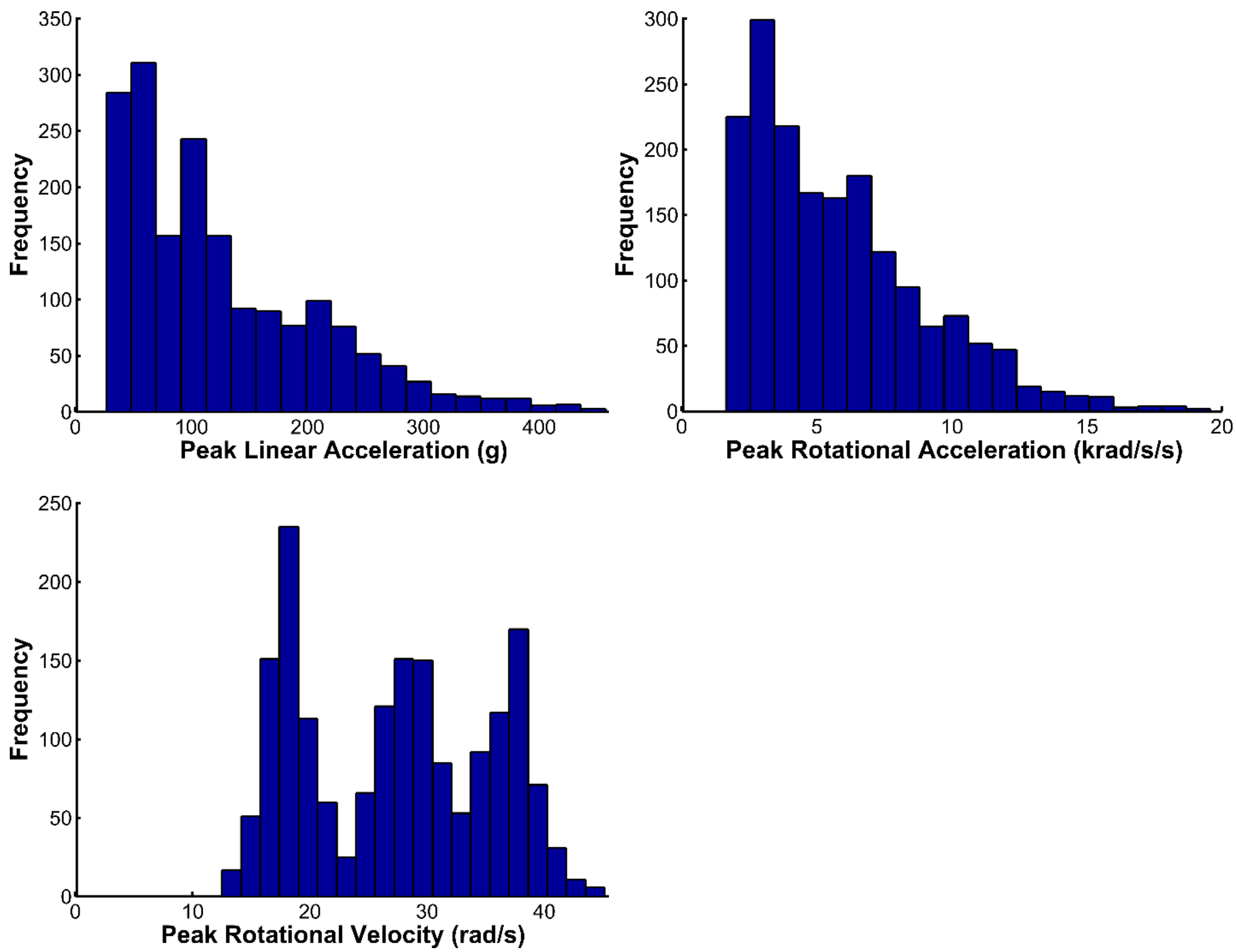

Figure 15: Distributions for peak linear acceleration (top left), peak rotational acceleration (top right), and peak rotational velocity (bottom left) for all impacts to evaluate hockey helmets. Acceleration distributions are right-tailed due to increasing variance with increasing impact speed. Rotational velocity values were less variable for each impact speed, resulting in a multimodal distribution shape.

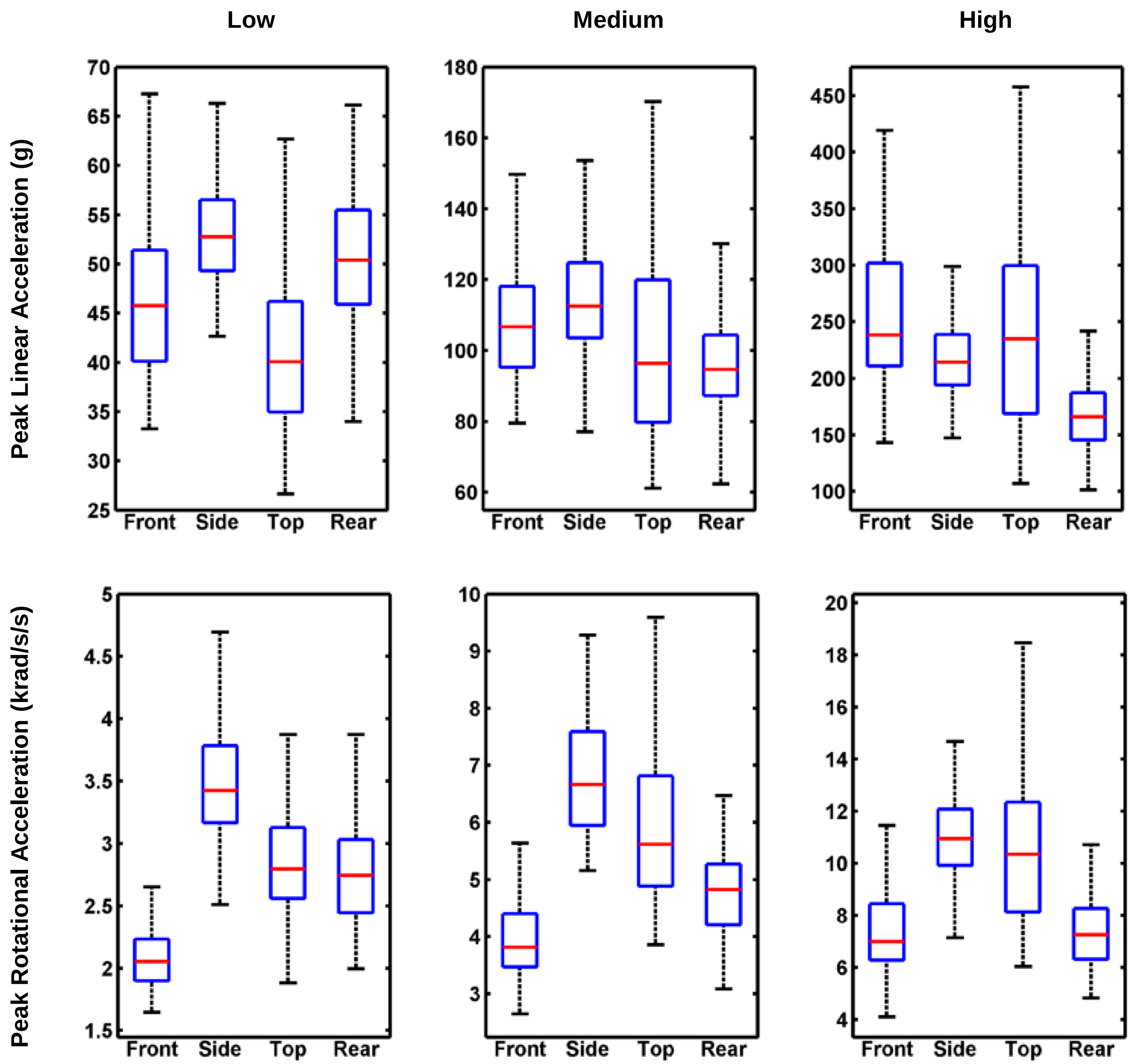

Figure 16: Box plots for peak linear acceleration (top) and peak rotational acceleration (bottom) for each impact condition in the Hockey STAR methodology. The median (red line), interquartile range (blue box), and full range excluding outliers (black dashed line) are displayed for each condition. Acceleration distributions varied by impact location and severity. Variance in acceleration values increased with increasing impact severity, and was generally greater for the top impact location. The large amount of variance in acceleration values shows the wide range of impact performance for currently available helmet models.

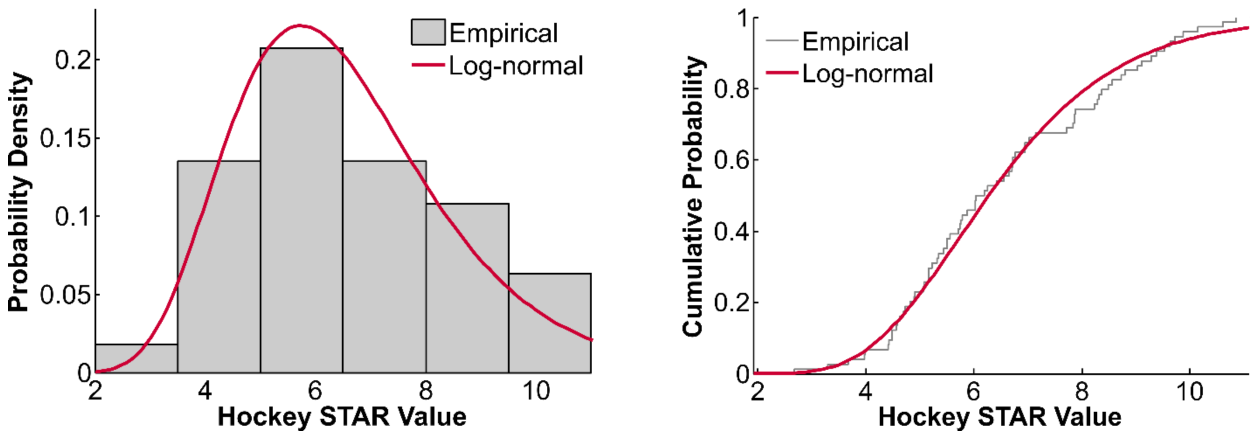

Figure 17: [Left] The discrete probability density function of Hockey STAR values for all helmet samples was fit with a log-normal probability density function. [Right] The empirical cumulative distribution function of Hockey STAR values with the best fit log-normal cumulative distribution function overlaid. Overall, the Hockey STAR values varied greatly between all helmet models, showing a wide range of impact performance for currently available helmets.

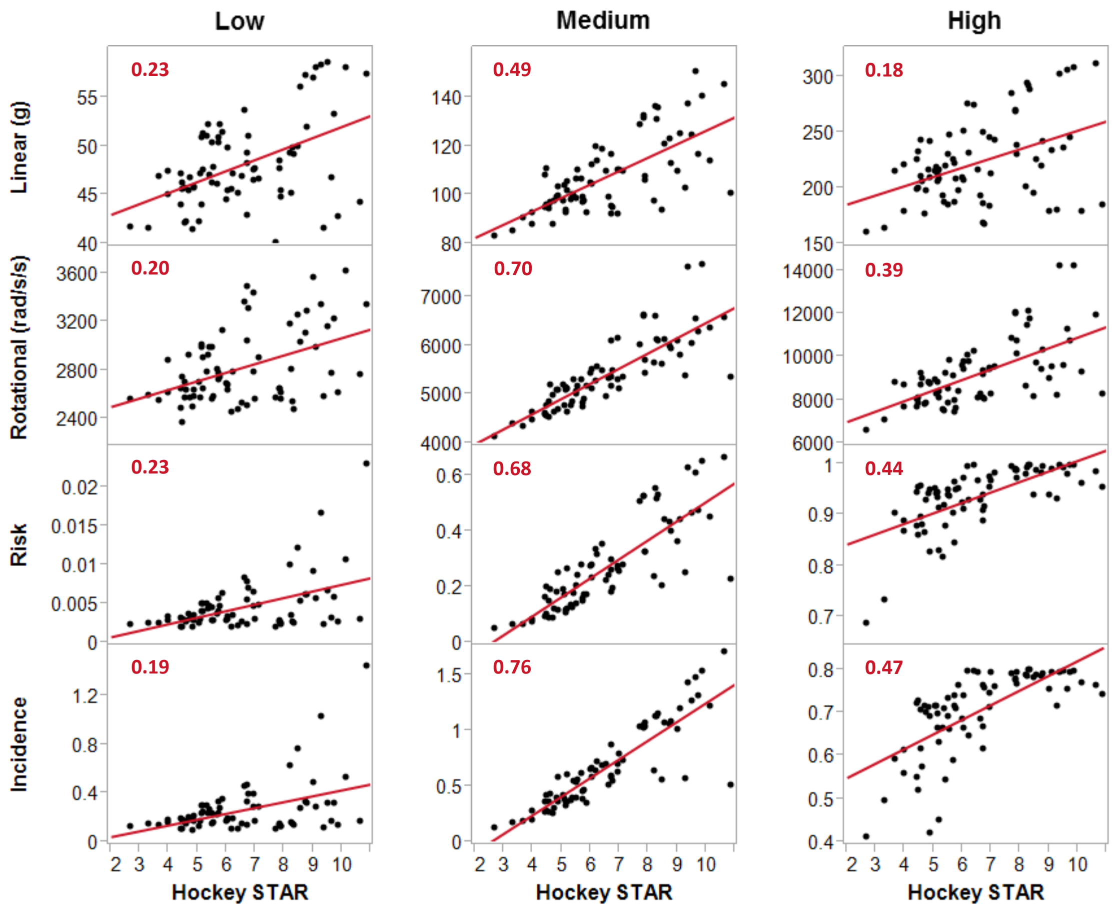

Figure 18: Linear regressions for Hockey STAR values as a function of 4 different parameters: peak linear acceleration (row 1), peak rotational acceleration (row 2), risk of concussion (row 3), and predicted incidence of concussion (row 4). The regressions were stratified by low (left), medium (middle), and high (right) severity impacts. The regression lines are shown in red, and the associated R^2 value for each relationship is shown at the upper left corner of each plot. Lower STAR values were associated with lower accelerations, which translated to lower risk and predicted incidence. Risk, and therefore incidence, were saturated for many helmet samples at the high severity impact conditions.

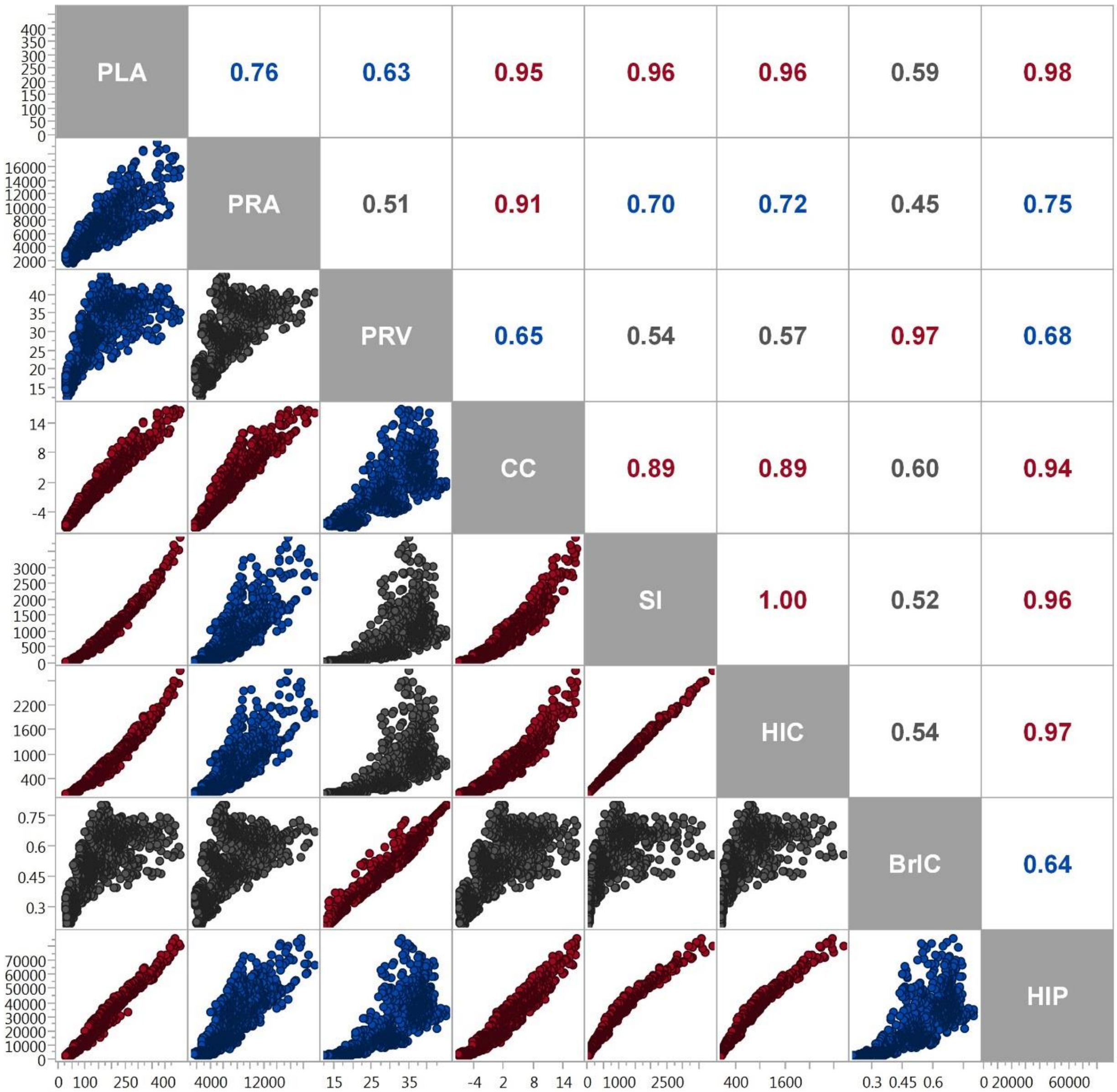

Figure 19: Correlations between kinematic parameters and brain injury criteria. The diagonal has parameter labels for each row and column. The bottom portion shows correlations between parameters with scatter plots, while the upper portion quantifies those relationships with R^2 values. Gray = $R^2 < 0.6$; blue = $0.6 \leq R^2 < 0.8$; red = $R^2 \geq 0.8$; PLA = peak linear acceleration (g); PRA = peak rotational acceleration (rad/s/s); PRV = peak rotational velocity (rad/s); CC = concussion correlate; SI = severity index; HIC = head injury criterion; BrIC = brain injury criterion; HIP = head impact power. All injury criteria with the exception of BrIC were highly correlated with

one another ($R^2 \geq 0.89$). All injury criteria other than BrIC were also highly correlated ($R^2 \geq 0.95$) with linear acceleration, however, CC was the only criterion highly correlated with both linear and rotational acceleration ($R^2 \geq 0.91$).

Discussion

The main finding in this study was that there is a wide range of impact performance among currently available hockey helmets, which supports the need for an objective measure of performance to inform consumers. Acceleration and Hockey STAR value distributions were used to quantify the large amount of variation in hockey helmet performance. Correlations between Hockey STAR values and different measures of impact severity showed that lower STAR values were associated with lower linear and rotational accelerations, as well as lower risk and predicted incidence of concussion. Finally, the large dataset used here provided a unique opportunity to evaluate the relationships between a variety of kinematic parameters and brain injury criteria. These relationships showed that all brain injury criteria evaluated were highly correlated to one another with the exception of BrIC, indicating that they provide similar information on impact severity.

Acceleration distributions exhibited a large amount of variation in impact performance between different helmet models, despite the fact that they all passed the minimum safety standard. As an example of some of the drastic differences seen, for the same impact condition (front location, medium severity), a top performing helmet had a 2% risk of concussion (89 g, 3,157 rad/s^s), while a poor performing helmet had an 84% risk of concussion (183 g, 5,573 rad/s^2). The top performing helmet had over a 50% reduction in linear acceleration, and over a 40% reduction in rotational acceleration for identical impact conditions. The increases in acceleration variance seen with higher impact severities were likely due to certain helmets bottoming out for those impacts. If the foam in a helmet bottomed out during an impact, a larger percent of the impact energy would be transferred to the headform rather than being modulated by the helmet. Hockey helmet standards limit peak linear acceleration to 275-300 g for their test methods. The high severity impacts used for Hockey STAR are more severe than those used for standards, given that some helmets exceeded 400 g for certain impact conditions. However, it is possible to limit the amount of energy transferred to the head even for those high severity impacts if there is a greater offset in the helmet. For example, the football helmet tested for development of the Hockey STAR methodology did not exceed 125 g for any of the high severity impacts [19]. Top rated football helmets generally have around a 4 to 5 cm offset from padding, compared to hockey helmets, which have substantially less at around 1.5 to 2.5 cm.

The distribution of Hockey STAR values shows the wide range of impact performance among currently available helmets. The drastic differences in helmet performance are most notable when comparing a sample of the best performing helmet (Helmet A) with a sample of the worst performing helmet (Helmet B) (Table 16). Linear accelerations for Helmet B were up to 3.6 times Helmet A (top impact, high severity), while rotational accelerations were up to 2.2 times Helmet A for the same condition. The Hockey STAR value is intended to summarize these differences in performance into a single number that can be easily interpreted by consumers. The Hockey STAR values represent the predicted incidence of all diagnosed and undiagnosed concussions for a player that has the same head impact exposure used with the evaluation system. The numbers presented here seem particularly high because the risk function used in Hockey STAR has a conservative estimate of underreporting, assuming that the actual number of concussions is 10 times that of what is reported. If the Hockey STAR values are adjusted to be more representative of reported concussions, the values would range from 0.27 to 1.09. These values are still representative of a player with a specific head impact exposure, which not all players experience in one season, but allows for direct comparison of reduction in injury risk.

Table 16: Comparison of linear and rotational accelerations, and risk of concussion from a sample of the top performing helmet (Helmet A, Hockey STAR = 2.67) with a sample of the worst performing helmet (Helmet B, Hockey STAR = 10.62). The accelerations are averaged for two repeated tests per impact condition. The low severity impacts were similar in acceleration values and risk of injury, but the differences are pronounced for the medium and high severity conditions.

		PLA (g)		**PRA (rad/s^2)**		**Risk (%)**	
Impact Location	Impact Speed (m/s)	Helmet A	Helmet B	Helmet A	Helmet B	Helmet A	Helmet B
Front	3.0	36	49	1833	2210	0.08	0.20
Front	4.6	88	183	3258	5573	2.21	83.65
Front	6.1	239	396	6724	11522	98.92	100.00
Back	3.0	49	37	2314	2835	0.21	0.20
Back	4.6	83	62	3495	5083	2.16	3.38
Back	6.1	126	153	5347	8507	33.53	93.48
Side	3.0	50	53	3187	3437	0.45	0.61
Side	4.6	96	137	5288	7749	13.07	81.85
Side	6.1	158	276	7362	12859	87.88	99.99
Top	3.0	32	38	2906	2595	0.17	0.17
Top	4.6	66	201	4509	7955	2.42	98.14
Top	6.1	118	423	6916	14882	55.16	100.00

Hockey STAR values are meant to summarize the overall performance of hockey helmets by considering linear and rotational accelerations generated for a range of impact locations and velocities. The correlations of different parameters with Hockey STAR values demonstrated how well impact performance was summarized by those values. Lower STAR values were associated with lower accelerations, and as the STAR values increased, accelerations increased as well. These correlations suggest that Hockey STAR is a good metric of overall performance, and is able to differentiate between helmets that generate higher or lower accelerations during impact. The medium severity impacts had the highest correlations for all parameters for the helmets evaluated in this study. Most helmets had less than 1% risk on average for the low severity condition, resulting in a small contribution to the overall STAR value. Additionally, risk, and therefore incidence, were saturated at the high impact condition for many of the helmet samples. Therefore the medium severity impacts were the most influential in differentiating helmet performance, with average risk ranging from 5-67%. This wide range of risk resulted in a wide range of incidences contributing to the overall STAR value. It is important to note that while risk was saturated at the high severity for many of the samples, some of the better performing samples did limit the risk of concussion even for severe impacts. If future helmet designs are able to minimize risk of injury for all high severity impacts as well as the more common lower severity impacts, they would outperform currently available helmets. It can be seen that, in general, there was a greater amount of variance among all parameters for higher Hockey STAR values (poorer performing helmets). This variation was introduced when a helmet performed much worse in a particular condition, and increased the STAR value. The better performing helmets were more consistent across all impact conditions.

The 1,776 tests that resulted from evaluating all currently available hockey helmets provided a large dataset of laboratory head impact tests with hockey helmets, and a unique opportunity to evaluate relationships between a variety of kinematic parameters and brain injury criteria. There was a moderately high correlation between peak linear and peak rotational accelerations (R^2 = 0.76). That correlation would increase if the data were separated by impact location, since there are both centric and non-centric locations (non-centric had higher rotational acceleration relative to linear). There were several interesting relationships to be noted with different brain injury criteria. All criteria except BrIC are highly correlated with linear acceleration ($R^2 \geq 0.95$). These correlations make sense intuitively, given that all criteria other than BrIC are functions of linear and rotational acceleration. BrIC is a function of rotational velocity alone, which does not correlate as well with either linear or rotational acceleration. Both CC and HIP are dependent on linear and rotational acceleration, however CC is the only criteria that is highly correlated to both linear and rotational acceleration. This is likely due to the fact that CC uses peak values for both linear and rotational acceleration, while HIP is the maximum of a function computed over time, and peak linear and rotational accelerations do not occur at the same time. The high correlation between HIP and peak linear acceleration suggests that the function may be maximized closer to the peak linear acceleration. There is also a nearly perfect correlation between SI and HIC because all impacts are similar in duration (approximately 8-15 ms), and both criteria are a function of linear acceleration. BrIC has lower correlations with all kinematic parameters and criteria evaluated here, with the exception of peak rotational velocity, indicating that it is providing different information than what it is being compared to. These relationships do not provide insight as to which parameters or criteria are predictive of injury, only how they relate to one another.

Conclusion

The Hockey STAR methodology is an evaluation system to identify differences in hockey helmet performance, and rate them to summarize a large amount of data in a way that can be easily interpreted by consumers. Since the publication of the methodology, 37 helmet models have been tested and rated. This study presents the results of those tests and analyzes the relationships between various measured and calculated parameters. The large amount of variation seen in the results presented here support the need for an objective measure of impact performance for hockey helmets. All of the helmets tested passed the required safety standards, but the top performing helmet reduced the risk of concussion by approximately 70% when compared to the worst performing helmet. Hockey STAR values demonstrated the ability to differentiate between helmets that generated higher or lower head accelerations, and seem to describe overall impact performance well.

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Chapter 5

Quantifying Head Impact Duration: Analysis of Laboratory Helmet Evaluation Systems

Abstract

Although head impact duration is thought to contribute to head injury severity, it is rarely quantified. Additionally, a number of different laboratory impact systems have been used to evaluate protective headgear performance. The objective of this study was to determine if impact durations from different commonly used laboratory impact systems were different, and how they compare to previously reported durations. Four different laboratory systems were evaluated using 2 different helmet types (football and hockey), 3 impact locations, and 2 impact speeds. Differences in duration were evaluated between helmet types and laboratory systems. Both helmet type and impact system had a significant effect on impact duration ($p < 0.0003$). Although there were significant differences in duration, these differences were small, and similar to previously reported values for helmeted head impacts.

Introduction

Head impact duration has been shown to contribute to injury severity since the earliest experimental work to determine human head injury tolerance. Cadaver drop tests and animal brain injury studies demonstrated a decreasing tolerance to head acceleration or pressure with increasing impact duration [1-3]. These studies along with human volunteer data were used to develop the Wayne State Tolerance Curve (WSTC) [4]. This curve represented human tolerance for moderate to severe head injury with acceleration magnitude as a function of time. The WSTC has been used as the basis for a number of proposed head injury criteria [5-8]. The dependence of head impact tolerance on impact duration has also been supported by experimental work with primates [9, 10].

Despite the theorized importance of impact duration on head injury tolerance, it is often not quantified. For laboratory simulations of real-world impacts, it is important to ensure that the impacts are representative of what they are simulating in both magnitude and duration. A number of different systems have been developed to evaluate headgear performance for sports. Some safety standards account for impact duration by evaluating the Severity Index (SI), while others use only peak acceleration tolerances [5]. Regardless of the criterion used to evaluate headgear, the impact durations for these systems have not been quantified and compared to impacts that occur on the field. The purpose of this study was to quantify impact duration for 4 different laboratory systems used to evaluate helmet performance. The durations were objectively quantified, compared to detect differences between systems and helmet types, and compared with previously reported values from laboratory and field studies.

Methods

A total of 126 impact tests were performed to compare impact durations between different laboratory systems and helmet types. Four different laboratory impact systems were evaluated: a NOCSAE drop tower, an ISO drop tower, a pneumatic linear impactor, and a pendulum impactor (Figure 20). On each system, 2 helmet types, 3 impact locations, and 2 impact speeds were tested. Only the lower impact speed was evaluated for the ISO drop tower, as the high speed impact resulted in accelerations that exceeded the range of the sensor. The 2 helmet types were a Riddell Revolution Speed football helmet and a CCM Resistance 100 hockey helmet. The 3 impact locations were front, side, and back, with impact speeds of 4 m/s and 6 m/s at each condition.

Laboratory Impact Systems

1: NOCSAE Drop Tower

The NOCSAE-style drop tower consisted of a medium NOCSAE headform attached to a rigid carriage on twin guide wires. The headform impacted an anvil with a 15.2 cm diameter, 1.27 cm thick MEP pad mounted on top. Three linear accelerometers were located at the CG of the headform (PCB-356A66, PCB Piezotronics, Depew, NY). Acceleration data were collected at 20,000 Hz.

2: ISO Drop Tower

The ISO-style drop tower consisted of an ISO magnesium half headform (Cadex, Quebec, Canada) attached to a rigid carriage on twin guide wires. The impacting surface was a 13 cm diameter flat, rigid steel anvil. A ball joint connected the carriage to the headform, and allowed for adjustment of impact location. A single uniaxial linear accelerometer was located in the ball joint at the CG of the headform with the sensing axis always normal to the impact surface regardless of impact location (PCB-353B18, PCB Piezotronics, Depew, NY). Acceleration data were collected at 20,000 Hz.

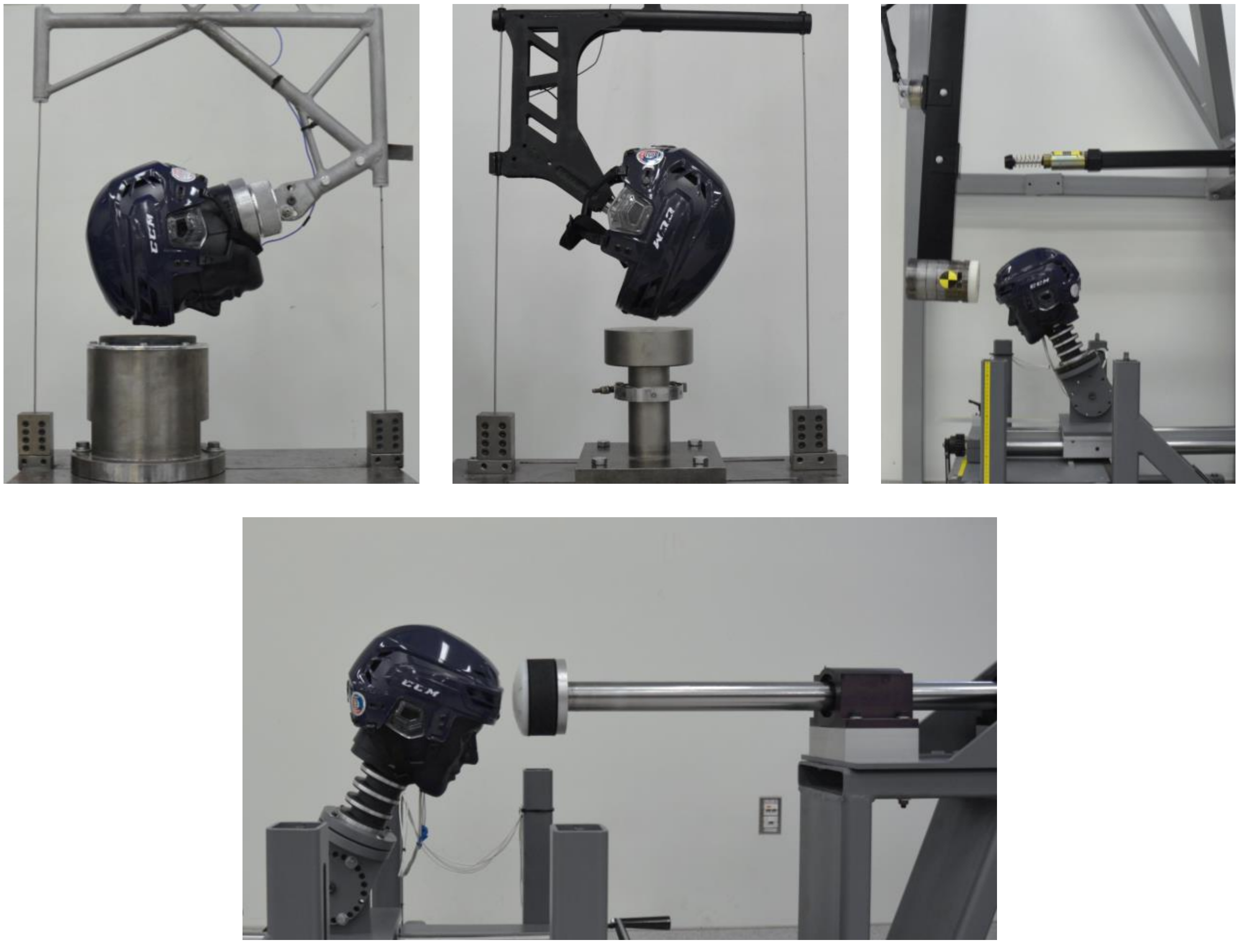

Figure 20: The 4 laboratory systems evaluated in this study were: a NOCSAE drop tower (top left), an ISO drop tower (top middle), a pendulum impactor (top right), and a pneumatic linear impactor (bottom). Each system is set to the front impact location with the CCM Resistance 100 hockey helmet.

3: Linear Impactor

The pneumatic linear impactor (Biokinetics, Ottawa, Canada) had a 14 kg impacting ram with a 12.7 cm diameter impacting face. The impacting face had a 4 cm thick layer of VN600 vinyl nitrile padding with a convex nylon cap to simulate a football helmet. The impactor struck a medium NOCSAE headform (Southern Impact Research Center, Rockford, TN) modified to be mounted on a Hybrid III 50th percentile male neck (Humanetics, Plymouth, MI) [11, 12]. The head and neck assembly was mounted on a 5 degree of freedom slider table that simulated the effective mass of the torso during an impact. Three linear accelerometers were located at the CG of the headform (7264B-2000, Endevco, San Juan Capistrano, CA). Acceleration data were collected at a sampling frequency of 20,000 Hz.

4: Pendulum Impactor

The pendulum arm length was 190.5 cm, with a total mass of 36.3 kg [13]. The impacting mass at the end of the arm was 16.3 kg and accounted for 78% of the total moment of inertia of the arm. The impacting face was flat, rigid nylon with a diameter of 12.7 cm. The impactor struck a medium NOCSAE headform modified to be mounted on a Hybrid III 50th percentile male neck [11, 12]. The head and neck assembly was mounted on a 5 degree of freedom slider table that simulated the effective mass of the torso during an impact. Three linear accelerometers were located at the center of gravity (CG) of the headform (7264B-2000, Endevco, San Juan Capistrano, CA). Acceleration data were collected at a sampling frequency of 20,000 Hz.

Impact Duration

Linear acceleration data from all laboratory systems were filtered to channel frequency class (CFC) 1000 using a 4 pole phaseless Butterworth low pass filter. Impact duration was quantified for all tests. Duration was defined as the difference between the time of the first axis-specific acceleration to cross +/- 10 g and the time of the last axis-specific acceleration to decrease below +/- 10 g after the peak resultant linear acceleration. Durations were then compared with two-factor ANOVAs by impact speed ($\alpha = 0.05$). The factors evaluated were impact system and helmet type. Post hoc Tukey's HSD tests were performed for any significant factors or interactions with multiple comparisons.

Results

Mean durations (+/- standard deviation) ranged from 7.0 ms (+/- 0.8 ms) for the hockey helmet impacted at 6 m/s on the pendulum, to 12.5 ms (+/- 3.6 ms) for the football helmet impacted at 4 m/s on the NOCSAE drop tower (Figure 21). The linear impactor was consistently associated with longer durations (> 11 ms) for both helmet types and impact speeds. The NOCSAE and ISO drop towers also had longer durations for the low speed football helmet impacts. The largest difference in impact duration between helmet types was associated with the ISO drop tower at 4 m/s (2.9 ms), while the smallest difference was associated with the linear impactor at 4 m/s (0.2 ms).

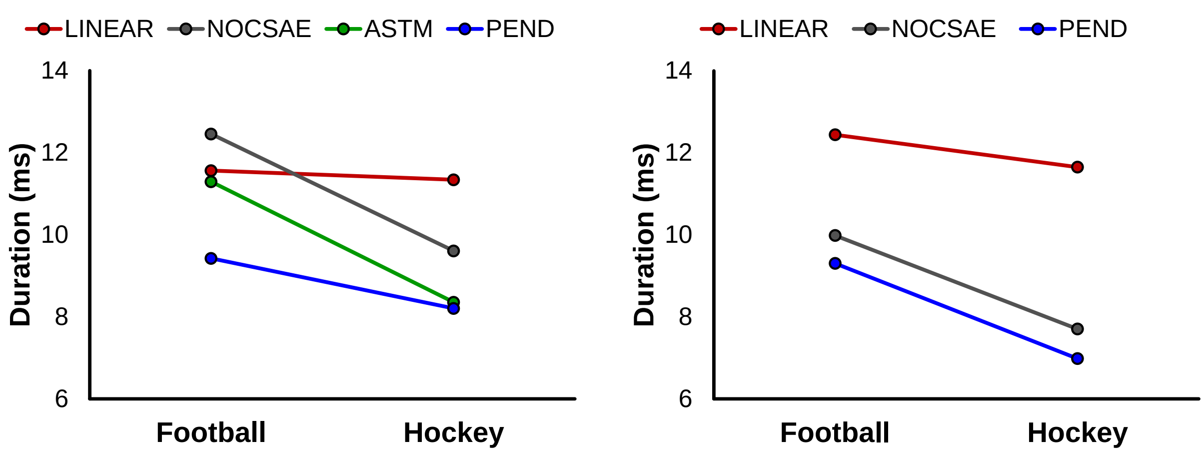

Figure 21: Mean durations for each impact system by helmet type, for 4 m/s (left) and 6 m/s (right) impacts. The linear impactor most frequently had the longest impact durations, while the pendulum impactor had the shortest. All hockey helmet impacts were shorter than football helmet impacts.

At 4 m/s, significant effects were associated with both impact system ($p = 0.0003$) and helmet type ($p = 0.0001$). For impact system, the pendulum had significantly shorter durations than both the linear impactor ($p = 0.0005$) and the NOCSAE drop tower ($p = 0.0046$). There were no significant interactions between impact system and helmet type at 4 m/s.

At 6 m/s, significant effects were also associated with both impact system ($p < 0.0001$) and helmet type ($p = 0.0021$). The linear impactor had significantly longer durations than both the pendulum ($p < 0.0001$) and the NOCSAE drop tower ($p < 0.0001$). There were no significant interactions between impact system and helmet type.

Discussion

A variety of impacting systems have been designed to evaluate helmet performance or simulate real-world impacts to further the understanding of sports head injury biomechanics. The systems evaluated in this study are used for different helmet standards or evaluation methods. The NOCSAE drop tower is used to certify football helmets in accordance with the NOCSAE standard. The standard requirements were developed to prevent skull fractures and severe brain injuries during play based on the same research used to develop the WSTC, and have proven to be very effective at doing so [1-4, 14]. Similarly, the ISO drop tower is used to certify ice hockey helmets to prevent severe head injuries. The pendulum impactor was designed to evaluate the relative performance of hockey helmets [13]. The pendulum has increased repeatability of impact speed compared with other impact devices [15]. A flat, rigid impacting face was also used to increase repeatability of the system, and better assess the relative performance of helmets. When a padded impacting face is used, the padding modulates some of the impact energy and can mask differences in helmet performance. Additionally, the rigid impacting face was more representative of the rigid surfaces encountered in hockey like the boards, glass, and ice. The linear impactor was designed for a proposed NOCSAE football helmet standard to supplement the current drop tests [15]. The design specifications for the linear impactor were based on reconstructions of impacts resulting in concussion in the NFL [16]. The padded impactor face was meant to simulate impact durations seen in the reconstructed impacts.

For the impacts in this study, duration was generally related to the amount of padding in both the helmet and impacting face. Mean impact durations on all systems and impact speeds were shorter for the hockey helmet, which had approximately 1.5 cm of padding compared to 3 cm in the football helmet. The linear impactor had the only padded impacting face, which resulted in consistently longer mean impact durations for all conditions (11.3 – 12.4 ms). The NOCSAE and ISO drop towers were also associated with longer durations (12.5 and 11.8 ms), but only for low speed impacts with the football helmet. Most of the football helmet impacts at 4 m/s were longer in duration because the padding in the helmet did not bottom out, distributing the impact force and prolonging duration. The padding on the linear impactor face also similarly modulated impact energy, masking differences between the football and hockey helmets.

The impact speeds selected for this study represent a wide range of impact severities depending on the system used (Table 17). The lowest peak linear accelerations were associated with the linear impactor, while the highest were associated with the ISO drop tower. The amount of energy transferred to the head during impact varied depending on the impact mechanism. For drop tests, all energy was transferred to the head as it was brought to a stop by the impacting anvil. The MEP pad on the NOCSAE drop tower modulated some of the impact energy, while the rigid steel anvil on the ISO drop tower resulted in much higher accelerations. On the linear impactor and the pendulum, the impactors continued to translate after impact, so not all of the impact energy was transferred to the headform. The padded linear impactor face modulated some of the impact energy, masking differences between helmet types, while the rigid face and greater mass of the pendulum resulted in more severe impacts for the same impact speeds.

Table 17: Mean peak linear accelerations for all combinations of impact system, speed, and helmet type. Acceleration values are reported in g's. The linear impactor was associated with the lowest magnitude accelerations, and the ISO drop tower was associated with the highest.

Speed	NOCSAE Drop		ISO Drop		Linear Impactor		Pendulum	
(m/s)	Football	Hockey	Football	Hockey	Football	Hockey	Football	Hockey
4	71	107	96	225	45	48	65	84
6	151	262	-	-	67	81	111	234

The impact durations in this study can also be compared to previous studies that used various types of laboratory and field data. Some of the earliest studies to report impact durations associated with head injury used cadaver heads impacted on rigid surfaces [3]. Skull fracture was associated with 1-6 ms acceleration durations. More recently, laboratory reconstructions of NFL impacts resulting in concussion were performed to quantify biomechanics associated with injury [16]. Most of these reconstructions involved head to head impacts of 2 helmeted anthropomorphic test devices (ATDs). The resulting concussive impacts averaged approximately 15 ms. Head acceleration durations have also been characterized in the field by instrumenting helmets of football players [17]. The average duration of these impacts was 9 ms (+/- 3 ms). For comparison, head impacts in motor vehicle accidents are usually less than 6 ms for impacts with rigid structures, and greater than 40 ms for airbag impacts [16].

The impact durations reported in this study are similar to those previously reported for head impacts in football players. There may be differences in the way duration was quantified in previous studies, however, criteria for quantifying duration are not normally reported. The method used to quantify impact duration in this study eliminates complications of defining duration based on resultant head acceleration by inspecting axis-specific accelerations (Figure 22). Forces from the impacting surface and neck both act on the head during impact. Accelerations due to neck forces can oscillate throughout the impact and continue to act on the head after the impact is over, making duration difficult to quantify based on resultant acceleration. The threshold for defining duration in this study was 10 g, which is reasonable given that normal activities like jumping can result in accelerations up to 10 g, and are not likely to contribute to injury [18].

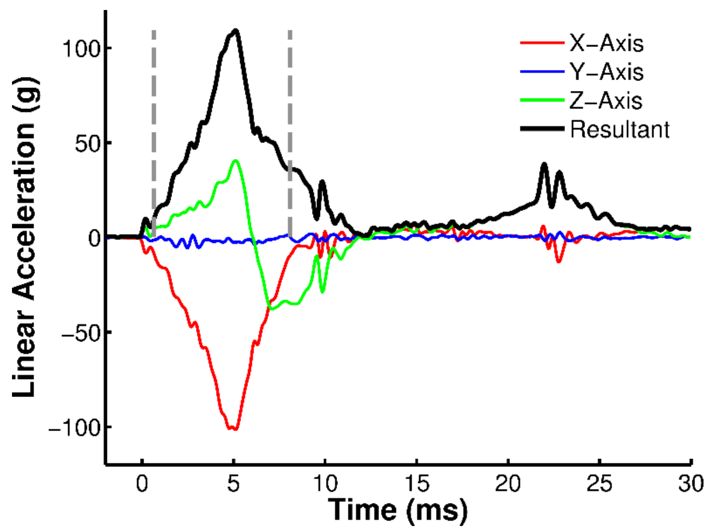

Figure 22: Example of a pendulum front impact at 4 m/s for a hockey helmet. The gray dashed lines represent the start and end of the impact as determined by the axis-specific method used in this study. It can be seen that if the resultant acceleration was used, duration would be prolonged by accelerations due to neck forces (z-axis) that continue after the impact is over.

Conclusion

This study evaluated differences in impact duration for 4 laboratory impact systems and 2 helmet types. Both laboratory system and helmet type had a significant effect on impact duration. The linear impactor had longer impact durations while the pendulum had the shortest. The football helmet also consistently had longer impact durations than the hockey helmet. Although there were significant differences in impact durations, most were small, with a maximum range of 4.7

ms within helmet type. These durations were also similar to previous studies using both laboratory and field data. Since all impact systems are within the range of previously reported laboratory and field impact durations, selection of a system depends on the desired application.

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Chapter 6

Research Summary and Publications

Research Summary

The research presented here provides an evaluation of different brain injury criteria and laboratory methods, as well as an application of those methods for evaluating the relative performance of headgear. While there have been many advances in the understanding of human tolerances to head injury through decades of research, direct applications to improving the safety of protective headgear are rarely made. By applying advances in injury criteria to laboratory methods for evaluating the relative performance of headgear, improvements in design can lead to reduction in frequency and severity of concussions in sports. The specific outcomes of the research in this dissertation were:

1. A review and evaluation of existing kinematic brain injury criteria, demonstrating that injury criteria using a combination of linear and rotational kinematics tend to be better predictors of injury than those using one or the other.
2. A methodology to evaluate the relative performance of hockey helmets, which serves to inform consumers while also providing manufacturers with a scientific framework to improve helmet designs.
3. An analysis of performance for existing hockey helmet models using the same methodology, showing that there are differences in relative risk of concussion for different helmet models.

4. A comparison of impact durations for different laboratory systems. While significant differences were present, the differences were small and within ranges of previously reported impact durations for helmeted head impacts.

Expected Publications

The chapters in this dissertation will be submitted for publication in a journal, or have already been published (Table 18). All chapters are presented here in their planned publication form.

Table 18: Expected or previous publications resulting from the chapters in this dissertation.

Chapter	Title	Journal
2	Evaluating the Predictive Capabilities of Brain Injury Criteria: An Analytic Review	Annals of Biomedical Engineering
3	Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets*	Annals of Biomedical Engineering
4	Biomechanical Performance of Hockey Helmets	Journal of Sports Engineering and Technology
5	Quantifying Head Impact Duration: Analysis of Laboratory Helmet Evaluation Systems	Journal of Sports Engineering and Technology

* *Published*