THE STUDY OF HEAD IMPACT BIOMECHANICS IN ADOLESCENT AND YOUTH MINOR ICE HOCKEY PLAYERS

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ABSTRACT

JASON PETER MIHALIK: The Study of Head Impact Biomechanics in Adolescent and Youth Minor Ice Hockey Players
(Under the direction of Kevin M. Guskiewicz)

Mild traumatic brain injuries are one of the most clinically difficult conditions to manage in sports medicine. Better understanding the biomechanics of head impacts will allow clinicians and researchers to better implement interventions designed specifically to reduce the incidence of injury. To date, few studies have looked at the biomechanics of head impacts in the young athlete. The overall objective of this dissertation was to evaluate the biomechanics of head impact severity during participation in youth ice hockey, with a specific evaluation of descriptive factors, and intrinsic and extrinsic factors related to impact biomechanics while playing hockey. We studied a two-year cohort of Bantam and Midget-aged ice hockey players, all of whom participated in all practices and games while wearing specially instrumented helmets capable of measuring head impact measures including linear acceleration, rotational acceleration, and the Head Impact Technology severity profile (HITsp). We also video-recorded every game in the first year of the study and developed an evaluation tool in order to characterize a number of aspects related to relative body positioning and overall anticipation of impending collisions. We also recorded a wide range of information including the number of shifts played, cervical muscle strength, player head and neck anthropometrics, measures of trait aggression, and general aerobic fitness. Our data support the notion that anticipating collisions may play a role in minimizing head impact severity. We also found impacts occurring in the open ice were greater than those occurring

along the playing boards. Further, illegal player infractions occur at a relatively high frequency and typically result in higher measures of head impact severity than legal collisions, especially as it pertains to elbowing, head contact, and high sticking infractions to the head. Based on our data, it does not appear that those with stronger neck muscles are better able to mitigate the forces associated with head impacts. Our data suggest a continued need to educate our players with the necessary technical skills needed to heighten their awareness on the ice. Coaches and athletes should incorporate body collision exercises in practices, and spend time educating young athletes on these proper checking techniques in an attempt to minimize the risk of injury and increase the safety of ice hockey.

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LIST OF ABBREVIATIONS

ASTM	American Society of Testing and Materials
ATC	Certified athletic trainer
AVI	Audio video interleave
BMI	Body mass index
BPAQ	Buss-Perry Aggression Questionnaire
C7	Seventh cervical vertebra
CDC	Centers for Disease Control and Prevention
CHECC	Carolina Hockey Evaluation of Children's Checking
CNS	Central nervous system
COG	Center of gravity
CSA	Canadian Standards Association
CSF	Cerebrospinal fluid
DV	Dependent variable
FAST	Faught Aerobic Skate Test
fMRI	Functional magnetic resonance imaging
HECC	Hockey Equipment Certification Council Inc.
HIT System	Head Impact Telemetry System
HITsp	Head Impact Technology severity profile
IV	Independent variable
NFL	National Football League
NOCSAE	National Operating Committee on Standards for Athletic Equipment
PET	Positron emission tomography

PIM	Penalties in minutes
RAS	Reticular activation system
SPECT	Single photon emission computerized tomography
TBI	Traumatic brain injury
USA	United States of America
USB	Universal serial bus
VO_2max	Maximal oxygen consumption
wPCS	Weighted principal component score

LIST OF SYMBOLS

cm	centimeter
°	degrees
Hz	hertz
kg	kilogram
kHz	kilohertz
g	linear acceleration expressed relative to gravitational acceleration
ml	milliliter
mm	millimeter
ms	millisecond
N	Newton
rad	radian
s	second

CHAPTER I

INTRODUCTION

It has been estimated that between 1.6 and 3.8 million traumatic brain injuries (TBI) result from sports each year in the United States (Langlois, Rutland-Brown, & Wald, 2006). These injuries cost the American health care system approximately $56.3 billion in direct and indirect costs (Langlois, Rutland-Brown, & Thomas, 2004), and make TBI among the most expensive conditions to treat in children (Schneier, Shields, Hostetler, Xiang, & Smith, 2006). While teams of medical professionals surround collegiate and professional athletes, it is often adolescent athletes who are cared for by parent volunteer coaches with very little medical knowledge. As a result, there is a need to study the factors that may contribute to mild TBI in order to minimize the risk of injury in our young athletes.

Much of what researchers know regarding head injury biomechanics is surprisingly quite dated. Historically, Pudenz and Shelden (1946) were among the first to study brain movement in subhuman primates following forces imparted to the head. In their initial studies, the top halves of primate skulls were replaced with transparent plastic domes; and accelerative forces were then delivered and high-speed cinephotography filmed the movement of the brain. Ommaya and Gennarelli (1974) further elucidated our understanding of how linear and rotational accelerations affect injury outcome in the primate model. To this day, it is accepted that linear accelerative forces are more likely related to focal lesions while rotational mechanisms of injury result in diffuse cerebral injuries (Holbourn, 1943, 1945; Ommaya & Gennarelli, 1974). However, research confirming either of these two conclusions has been limited, and controversy remains regarding the relationship between linear and rotational accelerations, and how they contribute to injury mechanism, severity, and type of cerebral insult. Researchers in the automobile industry continued the majority of the work in this area, with very little emphasis as to its direct applicability in the sports arena.

In the past 15 years, researchers have come to study the athletic population injuries extensively as the athletic venue provides investigators an excellent TBI research laboratory. The fact that athletes are exposed to repeated trauma and exhibit high-risk behaviors, coupled with the ability to perform extensive preseason baseline testing and equally comprehensive post-injury testing, are all positive aspects to the sports TBI laboratory setting. During this time, our understanding of head injury biomechanics has improved substantially. The National Football League (NFL) mild TBI committee was among the first to study the biomechanics relating to concussion. In a series of studies published in 2005 and 2006, laboratory reconstruction of NFL video footage presented biomechanical data associated with a number of football-related head impacts (Viano, Casson, Pellman, Bir, et al., 2005; Viano, Casson, Pellman, Zhang, et al., 2005; Viano, Pellman, Withnall, & Shewchenko, 2006); including an analysis of collisions causing concussion, as well as studying the biomechanics of the striking player, and the ability of newer helmets in dissipating some of the accelerative forces experienced by professional football players.

Technological advances in recent years have allowed researchers to study the real-time biomechanical characteristics associated with head impacts in collision sports such as football and, more recently, ice hockey. Our work in this field has identified a number of preliminary findings that we expect to hold true following additional data collection: most severe impacts occur to the top of the head (Mihalik, Bell, Marshall, & Guskiewicz, 2007), concussion injury thresholds proposed in the research literature for helmeted sports do not appear to be supported (McCaffrey, Mihalik, Crowell, Shields, & Guskiewicz, 2007), and no relationship between the magnitude of linear or rotational acceleration seems to exist with clinical measures of concussion such as postural stability, graded symptom reporting, and cognitive testing (Guskiewicz, et al., 2007). Despite these technological advancements, our understanding of head impact biomechanics in the youth athlete remains limited at best. Given the need to understand this injury in order to better address this international health care issue, it seems logical to extend this nature of research to the young athlete. Youth ice hockey, lending to its increasing popularity in the United States, is an excellent venue for this initiative.

Specific Aims

Descriptive factors

1) To study the biomechanics of head impacts sustained during games and practices in Bantam (13- and 14-year-old) and Midget (15- and 16-year-old) youth ice hockey players.

Extrinsic factors

2) To evaluate the effect of game-related exposure (i.e. number of playing shifts by period) on biomechanical measures of head impact severity.
3) To evaluate the effect of illegal player conduct on biomechanical measures of head impact severity.

Intrinsic factors

4) To evaluate the effect of body collision type on biomechanical measures of head impact severity.
5) To evaluate the effect of cervical muscle strength, cervical and head anthropometrics, general aerobic fitness, and player aggression, on biomechanical measures of head impact severity.

Statement of the Problem

Over one million youth participate annually in ice hockey in Canada and the United States; and, arguably, millions more play recreationally every year. Hockey players incorporate legal body collisions into the game at relatively young ages. In some regions of Canada, for example, this can occur as young as 9 years of age. For the most part, 11-year-olds (Peewee) are already playing full body contact ice hockey. At young ages, there is debate as to whether children are being placed at too much risk of injury. This greater risk may be heightened in the Bantam age level (13- and 14-year-old players). Since prepubescent adolescents enter puberty during this time, it is not uncommon for physical maturity between players at this age level to vary considerably.

Further compounding the issue, USA Hockey and Hockey Canada do not require the same level of coaching education at the Bantam level as they do in higher levels of play (i.e. Midget, Junior, collegiate, and professional). As such, coaches at this level are often inexperienced parent volunteers who lack the technical foundation required to instruct young ice hockey players the proper technique to mitigate injury risk while they play contact ice hockey. Second, there is no required mandate by any of the national hockey agencies to have coaches who are trained in first aid; this often leads to difficulty managing obvious player injuries, and an inability to manage more difficult injuries such as mild TBI. Finally, while it is generally accepted that USA Hockey's "Heads Up Hockey" and Hockey Canada's "Play Safe" programs help to reduce the number of catastrophic cervical spine injuries, the evidence behind these programs supporting a reduction in sport-related mild TBI remains anecdotal at best.

Research Questions

In this study, there will be three primary biomechanical measures of head impact severity: linear acceleration (expressed relative to gravitational acceleration, g), rotational acceleration (measured in rad/s^2), and Head Impact Technology severity profile (HITsp). These dependent variables will be analyzed for each of the following descriptive, extrinsic, and intrinsic research questions.

Descriptive factors

1. Are there significant differences in biomechanical measures of head impact severity sustained by youth ice hockey players across position, event type, and location of head impact? Is there a difference in these measures between striking players and those who are struck?

   a. Are there significant differences in biomechanical measures of head impact severity between forwards and defensemen?

   b. Are there significant differences in biomechanical measures of head impact severity between games or scrimmages, and practices?

c. Are there significant differences in biomechanical measures of head impact severity between locations of head impacts (i.e. back, front, sides, and top of head)?
d. Are there significant differences in biomechanical measures of head impact severity between the striking player and the player that is struck?

Extrinsic factors

2. Are there significant differences in biomechanical measures of head impact severity sustained by youth ice hockey players due to game-related exposure?

a. Do significant differences in biomechanical measures of head impact severity exist between head impacts sustained in the first, second, and third playing periods during games and scrimmages?
b. What is the effect of an increase in the number of shifts played per period on biomechanical measures of head impact severity?

3. Is there an association between biomechanical measures of head impact severity sustained by youth ice hockey players and infraction type at the time of the collision?

a. Do significant differences in biomechanical measures of head impact severity exist between head impacts sustained in legal (i.e. “clean”) body collisions and those resulting from a boarding or charging infraction, checking an opponent from behind, or elbowing an opponent or deliberately making head contact?
b. Do significant differences in biomechanical measures of head impact severity exist between head impacts sustained in legal (i.e. “clean”) body collisions and those resulting from a boarding or charging infraction, checking an opponent from

behind, or elbowing an opponent or deliberately making head contact in players who are struck?

Intrinsic factors

4. Is there an effect of body collision type on biomechanical measures of head impact severity sustained by youth ice hockey players?

   a. Are there significant differences between open-ice collisions and those taking place along the playing boards?

   b. Are there significant differences between anticipated and unanticipated body collisions?

   c. In anticipated body collisions, will relative segmental body position affect head impact measures?

5. Are cervical muscle strength, cervical and head anthropometrics, general aerobic fitness, and player aggression, associated with the biomechanical measures of head impact severity sustained by youth ice hockey players?

   a. Is increasing the strength of anterior cervical flexors, anterolateral cervical flexors, cervical rotators, posterolateral cervical extensors, and the upper trapezius, associated with lower biomechanical measures of head impact severity?

   b. Do subject anthropometrics such as player height, mass, head-neck segment length, and other head and neck measurements, affect biomechanical measures of head impact severity?

   c. Is increasing general aerobic fitness, as measured by an on-ice aerobic performance test, associated with lower biomechanical measures of head impact severity?

d. Is increasing player aggression, as measured using the Buss-Perry Aggression Questionnaire and reflected by penalties in minutes (PIM), associated with lower biomechanical measures of head impact severity?

Research Hypotheses

Descriptive factors

1. Are there significant differences in biomechanical measures of head impact severity sustained by youth ice hockey players across position, event type, and location of head impact? Is there a difference in these measures between striking players and those who are struck?

a. There will be no significant differences in biomechanical measures of head impact severity between forwards and defensemen.
b. Head impacts sustained in games will be more severe than those sustained during practices, as indicated by higher linear and rotation accelerations, and HITsp.
c. Impacts imparted to or by the top of the head will demonstrate significantly higher biomechanical measures of head impact severity compared to those of the back, front, and left and right sides. Further, impacts to the front of the head will demonstrate significantly *lower* biomechanical measures compared to those of the back, left, right, and top. No other significant differences are hypothesized to exist.
d. Biomechanical measures of head impact severity will be significantly lower during collisions where the player is the striker, compared to collisions where the player in our sample is being struck.

Extrinsic factors

2. Are there significant differences in biomechanical measures of head impact severity sustained by youth ice hockey players due to game-related exposure?

a. Head impact severity will be significantly greater in the third period compared to the first and second periods. Further, impact severity of collisions in the second period will be greater than those observed in the first.

b. The number of shifts played during the first period will affect the magnitude of head impacts sustained in the second period and, likewise, the number of shifts played in the first and second periods will affect the magnitude of impacts sustained in the third period.

3. Is there an association between biomechanical measures of head impact severity sustained by youth ice hockey players and infraction type at the time of the collision?

a. Body collisions resulting from illegal conduct will result in significantly higher biomechanical measures of head impact severity than those resulting from legal body collisions.

b. Body collisions resulting from illegal conduct will result in significantly higher biomechanical measures of head impact severity in players who are struck compared to players who strike and opponent.

Intrinsic factors

4. Is there an effect of body collision type on biomechanical measures of head impact severity sustained by youth ice hockey players?

a. Open-ice collisions will result in significantly higher biomechanical measures of head impact severity compared to collisions occurring along the boards.

b. Unanticipated collisions will result in significant increases in biomechanical measures of head impact severity when compared to anticipated body collisions.

c. In instances where players anticipate a body collision (delivering or receiving), maintaining a relative body position such that the player's head is facing forward, and the trunk, hips, and knees are flexed, will result in lower biomechanical measures of head impact severity.

5. Are cervical muscle strength, cervical and head anthropometrics, general aerobic fitness, and player aggression, associated with the biomechanical measures of head impact severity sustained by youth ice hockey players?

a. Players with stronger cervical muscles as measured by the strength of anterior cervical flexors, anterolateral cervical flexors, cervical rotators, posterolateral cervical extensors, and the upper trapezius, will experience lower biomechanical measures of head impact severity than those players with weaker cervical muscles.

b. Taller and heavier players will experience lower measures of head impact severity than their shorter and lighter counterparts. There will be no differences in measures of head impact severity as they relate to head-neck segment length, head-neck segment mass, and other head and neck measurements.

c. Increased general aerobic fitness, as measured by an on-ice aerobic performance test, will be associated with lower measures of head impact severity.

d. Increased player aggression, as measured by the Buss-Perry Aggression Questionnaire and reflected in penalties in minutes (PIM) will be associated with higher measures of head impact severity.

Definitions

Boarding: A *minor* or *major* penalty imposed to a player at the discretion of the referee based upon the degree of violence of the impact causing an opponent to be thrown violently into the boards.

Charging: A minor or a major penalty imposed to a player who runs (i.e. takes more than 2 steps or strides) or jumps into or charges an opponent.

Checking from behind: A minor or a major penalty imposed to a player who body checks or pushes an opponent from behind. Minor penalties incur an automatic *misconduct penalty*; major penalties incur an automatic game misconduct penalty. Further, any check from behind resulting in the opponent hitting head first into the end or sideboards, or goal frame, results in an automatic major penalty and accompanying game misconduct penalty.

Elbowing: A minor penalty imposed on or to any player who uses his or her elbow in such a manner as to in any way foul an opponent.

Head contact: A minor or major penalty imposed to any player who intentionally or recklessly contacts a player in the head, including with the stick or with an illegal body check.

Legal ("clean") body collision: A body collision with the intent of the separating the puck carrier from the puck without using any illegal infractions. This is done by colliding into an opponent who is in possession of the puck by using the hip or shoulder from the front, diagonally from the front, or straight from the side, but cannot take more than two strides in executing the check. Current USA Hockey regulations allow body checking at the Peewee level and higher.

Operational Definitions

Anticipated body collision: A body collision such that a player appears to see it approaching and, therefore, knowingly delivers or receives the body collision.

Open-ice collision: A body collision between two players such that neither of the two players involved in the collision hit the playing boards following the collision.

Penalties in minutes: The total number of penalty minutes handed to a player over the entire playing season for infractions they deliver during game participation.

Playing board collision: Any body collision whereby one or both of the players make contact with the playing surface boards during or following a body collision.

Playing shift: An incident where the player steps onto the ice to actively participate in the player and is terminated when the player returns to the team bench. Only one shift is counted in instances where there is a stoppage in play (i.e. offside, icing, penalty, goal) but the player remains on the ice for the start of the next play. The length of a typical shift ranges from 30 seconds to one minute.

Unanticipated body collision: A body collision whereby the receiving player does not appear to see it approaching and, therefore, is unable to prepare him or herself to receive it.

Limitations/Assumptions

The following assumptions will be made in the study:

1. Subjects will complete all testing to the best of their ability and with full effort.

2. Subjects will not knowingly or intentionally alter their style of play at any time while participating in the study.

3. Our sample will be comprised of 13- to 16-year-old male ice hockey players only, and be representative of most travel ice hockey players in the United States playing at the AAA level.

Delimitations

The following delimitations will be made in the study:

1. Participants will consist of male Bantam- and Midget-aged ice hockey players who normally participate in at least three on-ice sessions.

2. Visual evaluation of body collisions is considered sufficient to quantify the player's relative body position.

Significance

This research study addresses a number of important questions pertaining to youth ice hockey safety, specifically as it pertains to injury risk. Ideally, the results provide valuable and applicable information regarding game and practice situations in which youth hockey players are at the greatest risk for sustaining head impacts of higher, and potentially injurious, magnitudes. The study provides a foundation for developing informative teaching techniques for the prevention of mild TBI in youth hockey. The study outcomes have the potential to help create a more positive and safe environment for youth hockey players.

Information obtained from Specific Aim 1 provides data regarding the nature of head impacts sustained on a regular basis by youth ice hockey players, and how these measures relate to those reported in contemporary literature. Accelerative forces were studied with respect to head impact location, and will be valuable to those organizations who set helmet-testing standards such as the American Society for Testing and Materials (ASTM) International, the Canadian Standards Association (CSA), the Hockey Equipment Certification Council Inc. (HECC), the National Operating Committee on Standards for Athletic Equipment (NOCSAE), and to the helmet manufacturers who must meet these standards. Increasing head impact severity in the later stages of the game (Specific Aim 2) underscores the importance of incorporating endurance conditioning programs in youth ice hockey worldwide or limiting the number of shifts played. The knowledge gained by Specific Aims 3 and 4 will directly influence coaching and officiating. By understanding the nature of head impact severity in the context of player anticipation, and collecting data to support an optimal relative body position that minimizes the extent of the head impact, USA Hockey and Hockey Canada can address coaching initiatives and interventions designed with the express purpose of reducing the risk of injury and promoting safe participation of youth athletes in organized ice hockey. While this study will emphasize youth ice hockey, the concepts of collision anticipation extend to all other collision sports. Since the tempo and aggressiveness of ice hockey is a direct reflection on a game's officiating, Specific Aim 3 will serve to better elucidate the effects of player infractions on measures of head impact severity. This information, in addition to educational interventions, will have a targeted purpose of improving the awareness of these infractions, and emphasize the importance of enforcing player infractions likely to result in more severe head impacts. Lastly, the final specific aim addressed another potentially important injury prevention strategy. The study will attempt to gain an understanding of how cervical strength and anthropometrics relates to the severity of head impacts sustained by youth ice hockey players. Little research has substantiated the notion that individuals with weaker neck muscles are at greater risk for sustaining concussions. While this study did not address mild TBI specifically, it will relate measures of neck strength and head-neck anthropometrics to measures of head acceleration collected by a novel real-time accelerometer-based data collection system. These five specific aims were designed with the end purpose of improving the safety of youth ice hockey players, while at the same time providing a scientific foundation for intervention research in the area of youth head injury biomechanics, and the possibility of extrapolating the findings to youth mild TBI research in general.

CHAPTER II

LITERATURE REVIEW

Introduction

There has been a recent surge in the number of publications pertaining to sports-related mild traumatic brain injury (TBI). For example, a recent PubMed search identified that more than half the publications related to “sport mild traumatic brain injury” were published in the past 5 years alone. Notwithstanding the popularity of this topic within the literature, media exposure highlighting high-profile professional athletes forced into early retirement has served to elevate the general public’s awareness of this injury. In fact, the Centers for Disease Control and Prevention (CDC) have stated that the study of TBI, and more specifically its prevention, must continue to be national priorities. With as much attention as youth injury receives, there are few research initiatives focused on injury biomechanics in order to better design and implement prevention initiatives in this population. For reasons that will be made clear throughout this review of the literature, the sample proposed in this dissertation is comprised of young ice hockey players. Through the 2006-07 hockey season, the number of registered members (players, coaches, and officials) with USA Hockey has more than doubled since 1990 (USA Hockey, 2008). Due to the increasing popularity of this sport in the United States, and its continued popularity and growth worldwide including Canada and many countries in Europe, this unique population provides an excellent opportunity to study body collisions in youth sport, and to understand the potential for reducing injury risk by minimizing forces via implementing skills changes. The purpose of this literature review is to provide a comprehensive appraisal of the content matter pertaining to the proposed project, including epidemiological relevance, a review of the neuroanatomy, biomechanics of head injury, and methodological considerations.

Epidemiology of traumatic brain injury

In a CDC-sponsored report on TBI in the United States for the period of 1995-2001, information reported from emergency departments suggested that at least 1.4 million people sustain a TBI from all causes annually. Of these injuries, 1.1 million are treated and released in emergency rooms and as many as 235,000 result in hospitalization. Sadly, as many as 50,000 people die every year in the United States as a result of TBI. Children less than 15 years of age represent the majority of all cases of TBI; during this period of time, they represented as many as 475,000 cases each year. A more interesting, and perhaps speculative, statistic suggests that many more TBIs are sustained annually in the United States for which care is either not sought out in emergency departments, or sought out at all. Adolescents are at an increased risk for secondary injury and may often be allowed to return to activity or full sport participation without appropriate medical supervision. Further, in almost every age group, the rate of TBIs in the United States was higher for males than for their female counterparts (Langlois, et al., 2004).

Adolescent ice hockey traumatic brain injuries

Epidemiological study of youth ice hockey injuries has been limited to a small number of references, consisting of relatively small sample sizes. Despite their limitations, these studies do provide researchers with data pertaining to injury in this population of adolescent athletes. Gerberich et al. studied twelve varsity high school ice hockey programs in Minnesota over the course of the 1982-83 season (Gerberich, et al., 1987). An injury rate of 75 injuries per 100 ice hockey players was reported, including all types of injuries. Additionally, 22% of all injuries sustained were to the head and neck. Subsequent studies used exposure rates (i.e. number of injuries per player-hours) rather than rates (i.e. per player rates) to compute risk of injury. Stuart et al. expanded this line of research to understand the rates of injury across different age and playing levels (Stuart, Smith, Nieva, & Rock, 1995). They included in their sample Squirt (9-10 years of age), Peewee (11-12 years of age), and Bantam (13-14 years of age) ice hockey players. Stuart and his colleagues reported the injury rate in Bantam players to be 4.3 injuries per 1000 player-hours. This was significantly higher than the injury rates reported for Peewee (1.8 per 1000 player-hours) and Squirt (1.0 per 1000 player-hours) ice hockey participants. Stuart et al. were among the first to attempt to identify differences between game- and practice-related injury rates. In this same sample of Bantam ice hockey players, the game injury rate was reported to be as high as 10.9 per 1000 player-hours. In comparison, the practice injury rate was reported to be 2.5 injuries per 1000 player-hours. The Bantam practice injury rate was still higher than overall values reported for the Squirt and Peewee players.

Ice hockey presents a number of factors that predispose the participants to high injury rates compared to other collision sports such as basketball and soccer. First, the playing surface is made of solid ice and uses rigid boards that contain the playing area. Second, players use a stick to manipulate a rigid frozen projectile (the playing puck) that can sometimes exceed eighty miles per hour; both of these objects are often involved in injurious insults. Compounding these two factors, twelve ice hockey players wearing skates with sharp blades taking up position on the ice travel at high speeds, and are encouraged to purposefully collide with any opponent in possession of the puck. All these factors become increasingly more significant as players become bigger, faster, and stronger; this is likely the reason for the increased injury rates among Bantam ice hockey players compared to their younger counterparts. In agreement with these findings are those of Brust et al., where more than half of the injuries they observed occurred at the Bantam level (54%) (Brust, Leonard, Pheley, & Roberts, 1992). This paper further investigates the cause of injuries, and suggests that as many as 15% of all injuries were deemed intentional, and 34% of all injuries occurred in games the authors categorized as "hostile." The authors further describe that of all game-related collision injuries, 39% occurred from illegal checking and substantially less (20%) occurred from legal checking. To summarize Brust et al.'s findings, 86% of all injuries sustained in a game, including every incident of serious injury, resulted from checking and illegal game infractions. While this study represents a relatively small cohort, it presents a number of key findings related directly to the population of interest in this study and how the answers to the current research questions and hypotheses may have significant long-term intervention directives. For one, Specific Aim 3 addresses the key concern of player actions and how illegal game infractions may result in increased head impact forces. Specific Aim 4 seeks to identify a relative body position in which a youth ice hockey player is best able to control head impact forces from impacts delivered and sustained during competition.

Collecting and reporting epidemiological data in youth ice hockey can be a difficult task. This is evidenced by the limited published reports in this area. Most studies have been limited to short-term data collection periods such as tournaments. In one such study, Roberts et al. investigated youth ice hockey tournament injuries in a sample consisting of 695 boys and 112 girls (Roberts, Brust, & Leonard, 1999). The ice hockey players, aged eleven to nineteen years, were participating in five community-sponsored ice hockey tournaments in Minnesota during the 1993-94 winter hockey season. The authors report that body collision was involved in 65% of all injuries and, more specifically, 77% of all boys' injuries deemed significant in this study. This would appear to be in agreement with Bernard et al., reporting that 75% of all observed Bantam-level injuries were caused by body checking (Bernard, Trudel, Marcotte, & Boileau, 1993). Tournaments typically result in higher injury rates compared to regular season play for a number of proposed reasons. First, a loss often leads to a team's early elimination from the competition. Second, teams often play a much higher number of competitive sessions in a shorter time period; for example, it is not uncommon for a team to participate in as many as five games during a single weekend. In the Roberts et al. study, eleven penalties were associated with injuries, including six penalties for checking from behind. Checking from behind, under current USA Hockey regulations, results in immediate expulsion from the contest, and suspension from the ensuing team's match.

Allowing body checking at young ages in ice hockey is not without controversy. Many believe body checking leads to a laissez-faire attitude toward body collisions and an increase in rule infractions (Parayre, 1989). Tator's work in the area of catastrophic cervical spine injury agrees with this statement (Tator, Carson, & Edmonds, 1997), emphasizing a need for strict enforcement of the hit-from-behind rule and the necessity of continued education for coaches and players regarding the risk of head and neck injuries in ice hockey. Cerebral concussion, a form of mild TBI, occurred in each tournament that allowed body checking (Roberts, et al., 1999). The rate of mild TBI ranged from 10.7 to 23.1 per 1000 player-hours in the tournaments observed by Roberts et al. This rate is markedly higher than regular season rates previously reported. Sutherland et al., for example, recorded 0.09 concussions per 1000 player-hours (Sutherland, 1976), while Brust et al. and Stuart et al. estimated 0.75 concussions per 1000 player-hours and no concussions during a season, respectively (Brust, et al., 1992; Stuart, et al., 1995). In contrast, as many as 10% of high school ice hockey players sustained a concussion during the regular season (Gerberich, et al., 1987). Given the reported literature in this area, it is surprising that USA Hockey and Hockey Canada have not done more to educate players, parents, officials, and coaches on this topic.

A Canadian program labeled *Fair Play* was introduced to youth ice hockey in Quebec. *Fair Play* is designed to penalize unnecessary roughness by awarding a fair play point to teams that reduce the number and severity of their penalties. In one study, penalties issued to teams playing under the *Fair Play* program were compared to those teams not using the system (Marcotte & Simard, 1993). The authors reported 30% less major penalties and 25% less game suspensions were issued to the Bantam-level *Fair Play* teams compared to their non-program counterparts. At the Peewee level, *Fair Play* teams averaged 1.3 major penalties per season compared to 6.3 major penalties for non-program teams. Further, among teams using the *Fair Play* system, 71% of them did not receive a single game suspension. This study highlights a number of key points as they relate to injury prevention. First, interventions specifically designed to reward teams' proper behavior appear to result in decreases in illegal conduct and, specifically, severe misconduct more likely to result in injuring an opponent. Secondly, according to this study, it appears as though interventions such as *Fair Play* have a greater impact on younger players. This suggests that targeting younger players may be best suited by these interventions. However, it is unknown whether these behavioral modifications due to these interventions will carry forward into older years (i.e. Bantam, Midget, Junior, etc). While understanding the nature of player behavior on the ice in terms of illegal conduct is a venture worthy of future research, it would be a premature tenet at this time. This dissertation was the first to objectively evaluate several aspects of youth ice hockey, and provide the theoretical basis for which future intervention programs may be designed in order to minimize the severity of head impacts. A more thorough description of head injury biomechanics begins following the discussion of neuroanatomy.

Neuroanatomy

A basic understanding of the neuroanatomical structures is essential to providing a complete understanding of mild TBI. This section will discuss the main regions of the brain, their respective functions and infrastructures, as well as the manifestations that may appear as a result of mild TBI. Although this literature review attempts to demarcate these regions, it is important to note that the different regions of the brain work in tandem to achieve appropriate responses to different stimuli.

Cerebrum

The cerebrum forms the largest portion of the brain, consisting of 2 hemispheres, which occupy the anterior and middle cranial fossae. It accounts for approximately 80 percent of the mass of the brain and is responsible for higher mental functions such as memory and reason (Van De Graaff & Fox, 1999). The two cerebral hemispheres carry out different functions. The left hemisphere controls analytical and verbal skills such as reading, writing, and mathematics. Injuries to this cerebral hemisphere would elicit deficiencies in verbal memory. The right hemisphere controls spatial and artistic kinds of intelligence; right cerebral injuries would result in difficulties with visual and design memory.

The cerebrum consists of two layers. The first layer is referred to as the cerebral cortex and is composed of gray matter approximately 2-4 mm in thickness. Beneath the cerebral cortex is the thick white matter of the cerebrum, which manifests as the second layer of the cerebrum. The thick white matter of the cerebrum consists of dendrites, myelinated axons, and associated neuroglia. These fibers form the billions of connections within the brain by which information is transmitted to the appropriate places in the form of electrical impulses. The key distinguishing characteristic of the cerebral cortex is the many folds and grooves referred to as convolutions. These convolutions effectively triple the area of the gray matter, which is composed primarily of cell bodies of neurons (Van De Graaff & Fox, 1999). The elevated folds are known as the cerebral gyri (singular, gyrus) and the depressed grooves are referred to as cerebral sulci (singular, sulcus). The cerebral gyri and sulci demarcate the different lobes of the cerebrum and, in many cases, house specific areas of the brain responsible for sensory and motor control. Each cerebral hemisphere contains five lobes; four of the lobes appear on the surface of the cerebrum and are named for the overlying cranial bones that protect them. The separate lobes and hemispheres exist due to the specificity of function. This specificity of function is important to note when performing a clinical assessment of mild TBI, as the evaluation should be specific to assess different brain functions. A mild (or more severe) TBI may cause temporary (or permanent) impairment of cerebral functions. Much of what is known about cerebral function, unfortunately, is a result of observing the dysfunctions following trauma to the brain.

Frontal Lobes

The frontal lobes form the anterior portion of the cerebral hemispheres and are the largest of all the lobes. The frontal lobes function to initiate voluntary muscle impulses for the movement of skeletal muscles, analyzing sensory experiences, and providing responses relating to personality. Secondary functions of the frontal lobes include the following: mediation of responses related to memory, emotions, reasons, judgment, planning, and verbal communication. This is evidenced by several studies that have used functional magnetic resonance imaging (fMRI) techniques to identify brain regions associated with working memory (McAllister, et al., 2001). Injuries to the frontal lobe are likely to elicit personality changes, changes in memory, confusion, and disorientation. As previously discussed, the lobes of the brain often work in tandem. Most of the studies involving fMRI techniques have been conducted in healthy controls and have found, with performance of working memory tasks, bilateral frontal and parietal activation (Smith & Jonides, 1998). This is further evidenced by results that illustrate a functional relationship between frontal eye fields and prefrontal and parietal regions of the brain (Calhoun, et al., 2001).

Parietal Lobes

The parietal lobes are separated from the frontal lobes by the central sulcus. The postcentral gyrus can be found just posterior to the central sulcus. This gyrus is designated as a somatesthetic area because it responds to stimuli from cutaneous and muscular receptors throughout the body. In addition to providing somatesthetic stimuli, the parietal lobe functions in understanding speech and in articulating thoughts and emotions. The interpretation of shapes and textures of objects as they are handled is also a function of the parietal lobe. After a mild TBI, some patients may disclose preserved tactile sensations in the hand while, at the same time, they are unable to identify an object they are handling with the eyes closed (Tomberg & Desmedt, 1999). This phenomenon was found to be related to a lesion in the contralateral parietal cortex and is referred to as astereognosia (Caselli, 1991;

Mauguiere, Desmedt, & Courjon, 1983). Mild TBI affecting the parietal lobe would elicit difficulties in extended processing such as design and visual memory. In a 2001 study, a visual perception test was administered to subjects while undergoing fMRI. The researchers reported extended durations of parietal involvement during extended processing which was required in figural and visuospatial selection (Calhoun, et al., 2001). A previous fMRI study revealed that the superior parietal lobe was implicated in mental rotation (Tagaris, et al., 1996). Mental rotation is the task of differentiating between whether an image is simply rotated or is a mirror reflection of the original.

Temporal Lobes

The temporal lobe is located below the parietal lobe and the posterior portion of the frontal lobe, and is separated from them by the lateral sulcus. The temporal lobe functions to receive sensory neurons from the cochlea of the ear since it is in this lobe that the auditory centers are found. The temporal lobe also functions to interpret sensory experiences and stores memories of both auditory and visual events. Patients with mild TBI and postconcussive symptoms reveal, on positron emission tomography (PET) and single photon emission computerized tomography (SPECT) scans, a high incidence of temporal lobe injury (Umile, Sandel, Alavi, Terry, & Plotkin, 2002). Injuries to this area would result in visual memory deficiencies. Abnormal findings on PET and SPECT scans suggest that medial temporal lobe injuries seen in humans may be similar to neuropathologic evidence provided by animal studies following mild TBI (Umile, et al., 2002). In a rare case of refractory reflex sympathetic dystrophy, a complex regional pain syndrome, symptoms were resolved after the patient suffered a traumatic cerebral contusion in the left temporal lobe (Shibata, et al., 1999). This case report suggests two things: functional changes within the central nervous system (CNS) may occur following mild TBI, and the temporal lobe plays some role in nociceptive processing.

Occipital Lobes

The last of the surface lobes, the occipital lobes form the posterior portion of the cerebrum. The lobes lie immediately superior to the cerebellum and are not distinctly separated from the temporal and parietal lobes. Although they are small, the principal function of the occipital lobes concerns vision. The integration of eye movement by directing and focusing the eye is one of the main functions of the occipital lobe. It is also responsible for visual association; that is, correlating visual images with previous visual experiences and other sensory stimuli. The occipital lobe is often injured with a direct blow to the posterior aspect of the head or by contact of the head on the ground following a fall.

Insula

The insula is the deepest lobe of the cerebrum and is not visible on the surface; portions of the frontal, parietal, and temporal lobes cover it. Not much is known about the function of the insula except that it integrates other cerebral activities, and that it is thought to play a role in memory. The insula may be linked to areas located on the medial wall of the cerebral hemispheres when they are recruited to react to signals leading to self-controlled actions (Hulsmann, Erb, & Grodd, 2003). These actions include movement planning and preparation (Paus, 2001).

Cerebellum

The cerebellum is the second largest structure of the brain. It occupies most of the posterior cranial fossa. The cerebellum, too, consists of two hemispheres; it also consists of a midline portion called the vermis. The principal function of the cerebellum is coordinating skeletal muscle contractions by recruiting precise motor units within the muscles. It has also recently been suggested that it performs fundamental operations such as perception and cognition (Hulsmann, et al., 2003). Although the cerebellum is often referred to as the "little brain," it plays a crucial role in our understanding of postural control and proprioceptive feedback. The cerebellum constantly initiates impulses to selective motor units for maintaining posture and muscle tone. This accurate integration of finely-tuned movements have been attributed to the cerebellum since Flourens first published his research experiments on the properties and functions of the nervous system on vertebrate animals in 1824 (Hulsmann, et al., 2003). The cerebellum adjusts to incoming impulses from proprioceptors within muscles, tendons, and joints, and spatial sense organs, to refine learned movement patterns. Proprioceptors are sensory nerve endings that are sensitive to changes in length or tension of a muscle or tendon. The subconscious ability of the cerebellum to regulate and manage motion is evidenced by the muscle spindles' projections that terminate exclusively in the cerebellum (Barnett & Harding, 1955).

Trauma or diseases of the cerebellum frequently cause an impairment of skeletal muscle function. Movements often become jerky and uncoordinated. There is also a loss of equilibrium resulting in a disturbance of gait. The cerebellum has also been involved in motor-related functions such as feedback evaluation, planning, preparation, and response selection (Blakemore, Wolpert, & Frith, 1998; Gao, et al., 1996). Strong suggestions can be made on cerebellar involvement in tasks such as attention, perception, cognition, and consciousness, based on recent observations (Allen, Buxton, Wong, & Courchesne, 1997; Wolpert, Ghahramani, & Jordan, 1995). In fact, complementary analyses of fMRI data suggest the cerebellum may play a significant role in visual perceptual processing (Calhoun, et al., 2001). There is evidence from other studies to substantiate the suggestion that the cerebellum, in addition to frontal eye-field regions, may be involved in primary visual areas (Corbetta, et al., 1998; Nitschke, 2000).

Contrary to the cerebrum, damage in one half of the cerebellum affects the same side of the body. We would be likely to see decreases in reaction time in subjects who have sustained a mild TBI affecting the cerebellum. In addition to increased norepinephrine levels at the lesion site, the contralateral cerebellum also experienced increased levels of norepinephrine (Dunn-Meynell, Hassanain, & Levin, 1998). This suggests that the cerebellum would be affected regardless of the location in the brain of the traumatic insult. Injuries to the cerebellum may be better understood as being involved in diverse cognitive processes (Bloedel & Bracha, 1997).

Reticular formation

The reticular formation is a network of nuclei and nerve fibers found in the brain stem. It functions as the reticular activation system (RAS) and serves to arouse the cerebrum. Portions of the reticular formation can be found in the spinal cord, pons, midbrain, and parts of the thalamus and hypothalamus. The principle function of the RAS is to keep the cerebrum in a state of alert consciousness and to monitor the sensory impulses perceived by the cerebrum. The RAS is extremely sensitive to changes in, and trauma to, the brain. The sleep response is thought to occur because of a decrease in the activity of the RAS. A blow to the head may cause damage to the RAS, resulting in unconsciousness. With respect to neurocognitive function, the RAS is responsible for mental processing speed.

Studies have investigated the sensitivity of the reticular formation in rats. For example, electrical and chemical stimulation of the medullary and pontine reticular formation in decerebrate rats induced muscle atonia (Devor & Zalkind, 2001; Hajnik, Lai, & Siegel, 2000; Taguchi, Kubin, & Pack, 1992). Further supporting the role of the reticular formation in regard to unconsciousness are the results of a study published in 2000 in which functional brain imaging of 11 volunteer subjects under general anaesthesia revealed suppression of reticular formation activity (Alkire, Haier, & Fallon, 2000).

Protective Mechanisms of the Brain

Due to its intricate design, function, and significance, the brain has a number of structures that help protect it from external trauma. The eight cranial bones enclose and protect the brain, the meninges are membranous connective tissue coverings that surround the brain and spinal cord, and the cerebrospinal fluid (CSF) provides a buoyant cushion around the brain and its structures. The eight cranial bones consist of the frontal, two parietals, two temporals, and the occipital, sphenoid, and ethmoid bones. The frontal bone forms the anterior roof of the cranium—the forehead—and the roof of the nasal cavity; it also contains the frontal sinuses, which are connected to the nasal cavity. These sinuses act with others to lessen the weight of the skull. The spinal cord attaches to the brainstem through the foramen magnum located at the base of the skull.

In addition to the eight cranial bones, three membranous connective tissue coverings called the meninges also protect the brain. From outermost to innermost, they are the dura mater, the arachnoid mater, and the pia mater. The separation of the three meninges allow for spaces between the meningeal layers. The dura mater is the outermost and toughest of the membranes covering the CNS. Of special interest is that the attachment of the dura mater to the bones in the floor of the cranial fossae is firmer than its other points of attachment. Thus, a blow to the head at other points of attachment can detach the dura mater without fracturing the bones; whereas, a basal fracture usually tears the dura mater and results in leakage of CSF into the soft tissues of the neck, nose, ear, and nasopharynx. The arachnoid mater is a delicate, transparent membrane and, as its name suggests, is composed of a web-like tissue. Although the pia mater is very thin, it is thicker than the arachnoid. The pia is the innermost of the three layers of meninges and is a highly vascularized, loose connective tissue membrane that adheres closely to the surface of the brain.

The CSF is also a protective barrier, assisting the meningeal layers in sheltering the brain from mechanical injury. This is accomplished since the CSF acts as a buoy for the CNS. The CSF reduces the damaging effects of brain trauma by spreading the force over a larger area. The CSF reduces the effective mass of the brain by 97%. Although up to 800 ml of CSF are produced daily, only 140-200 ml is present at any time (Van De Graaff & Fox, 1999). Leakage of the CSF at the level of the spine or into the middle ear in acute settings should trigger appropriate emergency transport to a medical facility. It has been reported that headaches caused by decreases in intracranial pressure are often due to spontaneous leaks of CSF (Mokri, 2003). Furthermore, a case study was presented whereby computed tomography revealed leakage of CSF in the epidural space, causing postural headaches in a 33-year-old female (Goadsby & Jager).

In addition to the skeletal protection afforded by the cranial bones, the brain's autoregulatory system also serves to provide some form of internal protection. Existing data demonstrate an increase in norepinephrine release following cerebral contusion. These data suggest that this is protective and may act to stabilize the blood-brain barrier in areas surrounding the injury site (Dunn-Meynell, et al., 1998). This protective mechanism does not come without a price. The Dunn-Meynell et al. study also revealed a blockade of norepinephrine function during the first few hours after TBI, suggesting that a return to play in this time period may predispose the brain to further insult. Furthermore, animal studies suggest that these alterations and elevations in norepinephrine and other hormones can be prolonged and may, in some cases, impair catecholaminergic function following brain trauma (Prasad, Tzigaret, Smith, Soares, & McIntosh, 1993).

Biomechanics of traumatic brain injury

The biomechanics of TBI remains an area elusive to many researchers. Investigators in this area are faced with a number of issues as it pertains to understanding head injury impact mechanics. Current ethics standards have made the use of primate and other mammalian animal models very difficult to pursue; animal basic research in this area has been limited to the rat and small mammals in recent years. Second, the use of post-mortem cadavers does not allow researchers the ability to study impact mechanics in the context of everyday activities, including sports participation and work. The lack of muscle tonus and decreased volumes of CSF further make it difficult to replicate an in-vivo sample in the context of this area of study. Given these factors, and the evolutionary nature of head impact mechanics, it is worthwhile to review the historical literature in this area. The review of the literature continues to discuss the notion of linear and rotational acceleration, how impact location may play a role in injury severity, and concludes with a discussion of contemporary head impact mechanics research.

Historical biomechanics research

A number of landmark studies have precipitated our understanding of head impact biomechanics. These studies were initiated in the 1940s by Denny-Brown and his colleagues.

The general theme for Denny-Brown and Russell revolved around TBI biomechanics (Denny-Brown & Russell, 1941). While they used primarily cats—monkeys and dogs were also used—the innovative advance in their line of research was twofold: they used a pendulum hammer to impart the head impact and suspended their subjects such that the head was free to move following an experimental impact. Until that time, impacts had been imparted on animals whose heads were fixed, disregarding entirely the actual dynamics of impact situations such as those occurring in motor vehicle accidents or head impacts sustained on the playing field. Unfortunately, when using animals it is often difficult to assess subtle post-impact cognitive awareness. While loss of consciousness and death are obvious markers in animal subjects, the ability to objectively measure mental status in animals following a given head impact is difficult. Further compounding this issue, most animal studies employed light anesthesia in their animals using substances such as pentobarbital. Pentobarbital is approved for human use to treat seizures and as a preoperative sedative. It functions by depressing the CNS at all levels including the sensory cortex, motor activity, and altered cerebellar function (Deglin & Vallerand, 2009). Pentobarbital has also been used to reduce intracranial pressure and lower cerebral oxygen demands in TBI patients (San Diego Reference Library, 2008) and is the primary ingredient in both veterinary and human euthanasia compounds. Knowing this, it is easy to understand how the use of pentobarbital and other anesthetics masked any cognitive decline that could be observed in the animal model. Complementing this original work and venturing to eliminate the need for the animal model, physical models of the human skull and brain were constructed, and several kinds of impacts were imparted to these models (Holbourn, 1943, 1945). The skull was made of wax while the brain within it was composed of a gelatinous structure. While most of the credit for angular acceleration has been credited to Ommaya and Gennarelli (see later section), it was Holbourn who initially described that rotational motion was likely needed to produce cortical lesions and concussion.

One of the greatest advances in TBI mechanics occurred shortly after the work by Denny-Brown and Holbourn. Pudenz and Shelden (1946) removed the top half of monkey skulls and replaced them with a transparent plastic dome (Pudenz & Shelden, 1946). Using high-speed cinephotography, the researchers were able to capture the movement of the brain following a head impact. Supporting Holbourn's basic tenet, Pudenz and Shelden documented the brain noticeably lags behind the skull upon rotational head movement due to inertia. Almost thirty years later, Ommaya and Gennarelli used animal models and confirmed Holbourn's basic theory (Ommaya & Gennarelli, 1974). In their study, they imparted several rotational impacts to squirrel monkeys; the monkeys suffered a concussion as a result. An important addition to the literature occurred as a result of the studies published by Ommaya and Gennarelli. They observed that a direct impact to the head was not a necessary requirement to result in head trauma; but, rather, inertial non-impact loading resulting from an impulsive force may provide sufficient force to induce a mild TBI following a tackle or with more common whiplash mechanisms associated with car accidents (Letcher, Corrao, & Ommaya, 1973; Ommaya, Corrao, & Letcher, 1973; Ommaya & Gennarelli, 1974; Ommaya, Hirsch, Flamm, & Mahone, 1966; Ommaya, Rockoff, & Baldwin, 1964).

Linear and rotational acceleration

Denny-Brown and Russell were among the first to describe sudden changes in the velocity of the head as an acceleration-deceleration concussion (Denny-Brown & Russell, 1941). While earlier definitions of concussion did not formally exist until the Congress of

Neurological Surgeons published their definition (Committee on Head Injury Nomenclature of the Congress of Neurological Surgeons, 1966), Denny-Brown and Russell struggled to clearly delineate the criteria they used to identify concussion. Further, the use of pentobarbital in their animal subjects made level of consciousness difficult to assess. A strength of their work, however, was that they emphasized the importance of head movements in the context of mild TBI and, more importantly, how these head movements may or may not elicit concussion. Holbourn used these findings to justify his theory that angular acceleration of the head propagated movements of the brain within the skull, generating shear strains most prominent at the surface of the brain (Holbourn, 1945). This usually results in the transient deficits clinicians observe following mild TBI, as opposed to deeper brainstem lesions resulting in more severe forms of TBI. While animals subjected to linear accelerations typically showed no loss of consciousness, many of them did sustain cortical contusions and subdural hematomae. These results all connect to an important role for rotational movements eliciting an episode of mild TBI. Based on their definition of concussion, Ommaya and Gennarelli found that no observable injuries were produced when isolated linear impacts were imparted to twelve monkeys tested (1974). This was contrasted by thirteen monkeys who experienced a loss of consciousness for periods ranging from 2 to 12 minutes when impacted with their device while in the rotational mode. One of these thirteen monkeys never awoke and two others died within one hour of the impact.

In the context of mild TBI, the term *impact* typically denotes an injurious blow that makes direct contact with the head. An *impulse*, on the other hand, refers to a force that sets the head in motion without directly striking it. Examples of impacts range from helmet-to-helmet collisions, striking an opponent's head with a stick, or being struck in the head by a projectile used in the sport (e.g. soccer ball, hockey puck, etc). Impulsive forces are most commonly caused by tackling or body checking, and are the result of abruptly stopping an opponent's body from traveling in the direction in which it was headed. To relate this notion in layperson's terms, it is similar to the effect experienced by passengers when a car quickly accelerates or stops. Impacts and impulses are traditionally linear (translational) or angular (rotational) in nature. In real-world activities, there is usually some combination of both linear and angular accelerations associated with impacts and impulses. The question that still remains elusive to researchers and a matter of contention, for that matter, is "how do the relative contributions of angular and linear accelerations induce mild TBI?" Many factors are thought to play a role in the body's ability to dissipate head impact forces including individual differences in CSF levels and function, vulnerability to brain tissue injury, relative musculoskeletal strengths and weaknesses, and the anticipation of an oncoming impact or impulse.

While the literature review has discussed the phenomena of impacts and impulses, and linear and angular accelerations, there is a need to describe the reasons why not every impact or impulse results in an injurious episode. If the head does not move following a collision, the kinetic energy transferred by the blow should theoretically be transmitted elsewhere leaving the athlete otherwise unharmed. It was this principle that did not allow researchers prior to Denny-Brown and Russell to more fully understand the phenomenon of mild TBI (Denny-Brown & Russell, 1941). Secondly, CSF protects the brain within the cranium. As a result, some impacts or impulses do not exceed a threshold needed to drive the brain to impact the inside walls of the cranium and cause transient lesions and subsequent mild TBI. When an athlete experiences a rotational mechanism, it is thought that rotation of the cerebrum about the brainstem produces shearing and tensile strains. Since activity in the midbrain and upper brainstem are responsible for alertness and responsiveness, rotational mechanisms of TBI are believed to more likely result in loss of consciousness than predominantly linear types of impacts or impulses. Regardless of the type, attribute, or severity of a particular impact or impulse, the end result is as follows: the effective mass of the head has become too large for the body to overcome the acceleration or deceleration forces that have sent it in motion.

Impact locations

While the history of head impact biomechanics has been discussed, and attempts to parse out the literature as it relates to linear and rotational acceleration mechanisms of injury, a discussion of the effect of impact locations on mild TBI remains. Surprisingly, very little data are available to this effect. Hodgson et al. studied reversible cerebral concussion in the context of head impact location (Hodgson, Thomas, & Khalil, 1983). Using an air-propelled striker, impacts were imparted to the frontal (front), temporoparietal (side), occipital (rear), and cranial (top) aspects of rigid protective caps worn by six anesthetized female primates (macca speciosa). Following each impact, the monkey's head was allowed to freely move as much as 8 cm in any one direction before encountering a Styrofoam cushion. Following analysis of high-speed cinematography (4000 frames per second), one observational finding of the study was that all impacts resulted in linear and angular movements. The air-propelled striker delivered its force such that the impacts did not coincide with the head center of gravity (COG) and, as a result, a finding to the contrary would not have been expected. Another reason was that no neck force constraints were present in the monkeys due to partial anesthesia. An interesting finding reported by Hodgson et al. was that impacts to the temporoparietal region (side) produced periods of unconsciousness up to three times longer than loss of consciousness resulting from impacts imparted to the other areas of the head. It is difficult to explain this phenomenon since there are no obvious anatomic reasons in the monkey why this area should be more sensitive than others. Further, since all impacts were imparted at the same level above the head COG, it is plausible that the oval shape of the animal head produces lower mechanical impedance to higher accelerations for side impacts. Hodgson et al. imparted four reversible concussive impacts to each animal within a two-month period. Evidence of neurological deficit or neurogenic dysfunction was not observed in any of the animals. It is unclear based on their report how they purported to determine the extent of neurological deficit.

It is difficult to compare more contemporary literature to Hodgson et al. for two primary reasons: there was no follow-up work to this research question by the Hodgson group and modernized research in this area does not provide substantive data for which comparisons would be deemed meaningful. One such study is that of Guskiewicz et al. (Guskiewicz, et al., 2007). This paper represented the third of three companion papers reporting the results of an ongoing study on injury biomechanics in American football players. This study employed a real-time helmet accelerometer data collection methodology in eighty-eight Division I collegiate football players across three playing seasons. This sample sustained in excess of 104,000 total head impacts resulting in a measurable linear acceleration exceeding 10 g. Of all the impacts collected, thirteen resulted in a clinical diagnosis of mild TBI. The data suggest a higher propensity of top-of-the-head impacts and an increased risk of concussion for impacts to this region. Six of thirteen mild TBI occurred from impacts to the top of the head; this is in contrast to four, two, and one concussions occurring to the front, right, and back, respectively. The data suggest that top-of-helmet impacts may result in larger postural stability deficits following mild TBI. It was speculated that top-of-helmet impacts might result in a coup-contrecoup mechanism occurring in a superior-inferior direction causing the cerebellum to impact the base of the skull and recoil superiorly into the cerebellar tentorium. Interestingly, the data also indicate that top-of-helmet impacts typically result in relatively lower rotational acceleration values compared to injuries following impacts to the other areas of the head. This information brings into question the notion that rotational acceleration is the leading precursor to injury and is suggestive that type of acceleration, in combination with impact location, may be a better determinant for both onset and severity of injury.

Theoretical thresholds and accelerometer-based research studies

Mild TBI research has provided clinicians with useful information as it pertains to individual pieces of the proverbial concussion puzzle including, but not limited to, symptomatology, postural stability, and cognitive function. While these studies have provided us with important information and have changed the way many medical professionals manage injuries, they do very little to help us understand what causes a concussion and how we might best be able to minimize the risk of injury entirely. A number of contemporary studies have investigated impact biomechanics and have sought to shed light on proposed injury thresholds for mild TBI. In the Hodgson et al. study previously discussed, only short-duration impacts (1 to 2 ms) were imparted to the six monkeys (Hodgson, et al., 1983). They report, however, that the linear accelerations of the impacts causing concussion ranged from 2000 to 5000 g. A few years earlier, Ommaya and Gennarelli reported tangential accelerations ranging from 108,000 to 371,000 rad/s^2

(Ommaya & Gennarelli, 1974). These are significantly higher than those values reported in human mild TBI (reported later in this section) and may be explained by a number of factors. First, monkey skulls and musculoskeletal anthropometrics such as bone density, skull thickness, and musculoskeletal strength, are significantly superior to that of humans. Second, the ability of the researchers to accurately measure linear and rotational accelerations more than 25 years ago were quite limited compared to today's standards. Lastly, definitions of concussion employed for research purposes prior to 1997 (American Academy of Neurology, 1997) almost all included some level of loss of consciousness, suggesting they were imparting more serious and severe impacts to the monkey subjects in order to render them "concussed." Unterharnscheidt and Higgens report from their study that rotational accelerations in excess of 200 rad/s^2 produced cerebrovascular injury to most subjects in their sample (Unterharnscheidt & Higgens, 1969).

More recently, Hugenholtz and Richard were among the first in the literature to propose a mild TBI injury threshold in terms of linear acceleration g-forces (Hugenholtz & Richard, 1982). In their report, they proposed that a mild TBI would likely result from a blow to the head exceeding 80 to 90 g, and that these blows be sustained for greater than 4 ms. Over the past six years, the National Football League (NFL) has recently published a sequence of studies in *Neurosurgery* describing many facets of concussion and mild TBI in their league. The initial study from these efforts pertained to the laboratory reconstruction of concussive injuries captured on video (Pellman, Viano, Tucker, Casson, & Waeckerle, 2003); they represented the most sophisticated method of analyzing concussive injuries in professional football players at the time. The studies are not without limitations. For one, the laboratory retrospective reenactments were based on game video footage and important mathematical derivations were extrapolated from relatively low-speed video capture frequencies. Second, only 31 cases out of 182 reviewed were reconstructed in the laboratory. Conclusions were made based on this very small and selective sample of cases. Based on these data, Pellman et al. suggest that mild TBI in helmeted impacts are likely to occur between 70 and 75 g. This contrasts with data we have previously published where only 7 of 1858 (less than 0.38%) head impacts exceeding 80 g resulted in a diagnosed case of mild TBI (Mihalik, et al., 2007). One reason for the discrepancy may be explained by the fact that Pellman et al. studied professional football players while we investigated this phenomenon in collegiate football players. Given similarities in player size, these differences are very unlikely to be explained by the different samples.

Zhang et al. shortly thereafter proposed injury threshold values, employing the use of the Wayne State University Brain Injury Model (Zhang, Yang, & King, 2004). This model replicates a 50th percentile adult male head and includes anatomical structures including the dura, falx cerebri, tentorium and falx cerebelli, the CSF, cerebrum, cerebellum, and brainstem. Prior to analyses, the model was prevalidated against cadaveric intracranial and ventricular pressure data previously published (Nahum, Smith, & Ward, 1977; Trosseille, Tarriere, Lavaste, Guillon, & Domont, 1992). Twenty-four head-to-head impacts sustained in professional football were duplicated using their finite element head model to predict injury thresholds based on brain tissue responses. They reported that resultant linear accelerations of the head COG of 66 g, 82 g, and 106 g, were associated with a 25%, 50%, and 80% probability of mild TBI, respectively. These values are similar to those proposed by Ono et al., who suggested that impacts of 90 g sustained for 9 ms or longer would result in mild TBI (Ono, Kikuchi, Nakamura, Kobayashi, & Nakamura, 1980). The Wayne State Tolerance

Curve, published in 1964 by Gurdjian et al., deemed an 80 g impact noninjurious and an impact greater than 90 g could produce a mild TBI (Gurdjian, Lissner, Hodgson, & Patrick, 1964). Zhang et al. also propose rotational accelerations more in line with those we have collected in our own ongoing work. They associate rotational accelerations of 4600 rad/s^2, 5900 rad/s^2, and 7900 rad/s^2 with a 25%, 50%, and 80% probability of sustaining a mild TBI. While the University of North Carolina at Chapel Hill data suggest that there is far from a 50% probability of sustaining a mild TBI with an impact exceeding 82 g or 5900 rad/s^2, it is important to note that theoretical thresholds are derived primarily from animal models and the direct relation of these theoretical thresholds of injury to the human model remains an area of continued exploration.

Real-time accelerometer data collection is a novel tool recently made available to researchers attempting to better understand the biomechanics of mild TBI. Preliminary data capture techniques were limited in design. For example, Naunheim et al. attempted to study the linear accelerations sustained by high school student-athletes; specifically, an ice hockey defenseman, football offensive lineman, football defensive lineman, and a soccer player. A triaxial accelerometer was inserted within a football and ice hockey helmet and linear acceleration values were recorded during actual play. The data obtained from the soccer player are meaningless to interpret for two main reasons. First, since there was no method of affixing the accelerometer to the player’s head, the soccer player wore an instrumented football helmet. Secondly, game data were not captured; instead, the soccer player was asked to head 23 balls kicked to him or her at a standardized velocity. The mean linear acceleration measured in the football and ice hockey players were 29.2 g and 35.0 g, respectively. The football-related data are higher than what we have observed in our collegiate football sample

(Mihalik, et al., 2007). Further, our preliminary data in youth ice hockey players suggest mean linear accelerations typically do not exceed 19 g (Mihalik, Guskiewicz, Jeffries, Greenwald, & Marshall, 2008). While Naunheim's study represented an important advance toward real-time data collection, they were limited by a very small sample and did not transform the data to render it to be a normal distribution; this tends to overestimate the actual linear acceleration values measured.

Duma et al. were the first to employ acceleration-measuring technology in helmets for large numbers of athletes during normal practice and game situations (Duma, et al., 2005). This technology, the Head Impact Telemetry (HIT) System (Simbex; Lebanon, NH) will be described in more detail later. Duma et al. reported the magnitude of head impacts to be 32 ± 25 g. This contrasts the range of 20 to 23 g in a similar sample of Division I collegiate football players we would later record and report (Mihalik, et al., 2007). A number of explanations exist to account for these differences. Linear acceleration is a highly skewed measure, with the majority of all head impacts yielding low linear acceleration outcomes. Duma et al. calculated the mean and standard deviation of their impacts without first controlling for the highly skewed distribution of their data. Secondly, they alternated eight accelerometer units among their sample, selectively targeting players throughout the course of the season. Our study measured all head impacts sustained by each player in every practice and game throughout the season and alternated between players only to replace an individual who no longer remained in our study due to season-ending musculoskeletal injuries.

Our work continued in this area, attempting to better understand the effect of sustaining head impacts in excess of previously published injury thresholds. In one study, McCaffrey et al. studied how football players performed on clinical measures used to evaluate and diagnose concussion following a game or practice session in which they sustained an impact exceeding 90 g (McCaffrey, et al., 2007). Testing was completed within 16 to 24 hours following the end of the given session. Athletes were tested only in the absence of a concussion diagnosis. Her results found that in a convenience sample of collegiate football players, simply sustaining an impact in excess of 90 g does not result in a clinically observable case of mild TBI.

Others have joined in the effort of implementing the HIT System in the realm of high school and collegiate football. It was not until recently, however, that more extensive study of youth ice hockey started. We have been studying head impact biomechanics in Bantam-aged youth ice hockey players for two complete hockey seasons. Preliminary data in this sample suggest that 13- and 14-year-old ice hockey players sustain head impacts nearing the magnitude of those sustained by collegiate football players (Mihalik, Guskiewicz, et al., 2008). While research continues in this area, it will be important for researchers to use this novel instrumentation to not only better appreciate the nature of head impacts sustained by athletes, but to understand how players can better protect themselves and their opponents from sustaining high-magnitude impacts, hopefully resulting in lower incidences of mild TBI in amateur, collegiate, and professional sports, alike.

The role of neck musculature, and anticipation, on head injury

In reviewing the literature, there remains very little known about the types of forces that cause mild TBI and, perhaps alarmingly, very few suggested methods to reduce head impact forces. Cantu suggests there are five methods that can result in a reduction of mild TBI: changes in rules and coaching technique, improvements in conditioning and equipment, and increasing medical supervision (Cantu, 1996). While the brain is not thought to be able to condition itself to accept repeated blows, it is anecdotally believed the neck can be strengthened and the risk of mild TBI reduced.

The basic tenet of the neck muscle theory for reducing brain injury is that an athlete who anticipates an oncoming collision will be better able to control head movement by contracting (i.e. tensing) their cervical musculature. Using a Newtonian approach, acceleration is the result of force divided by mass. When the cervical musculature is contracted, it is thought to significantly increase the effective mass of the head-neck-trunk segment, resulting in a lower acceleration of the head. When an impact is unanticipated, and the cervical musculature is not tensed and prepared for a collision, the effective mass is reduced to that of the head. Given an equal force from a body collision, the head would experience a substantially greater acceleration and, therefore, more likely to sustain an injury. In theory, this seems rather intuitive; however, research has been ambiguous in this regard partly due to a general lack of research in this area. Studies in this area have focused primarily on a soccer-heading task.

In one study, Tierney et al. investigated the gender differences in head-neck dynamic stabilization during head acceleration (Tierney, et al., 2005). These perturbations were delivered in a known and unknown scenario, and in directions of forced extension and forced flexion. Their results suggest an increase in forced-flexion (males and females) and forced-extension (males) angular acceleration, as well as forced-flexion and forced-extension angular displacement in both males and females when force application was unknown (i.e. unanticipated). Similar trends were observed for electromyographic activity of the sternocleidomastoid and trapezius muscles. A number of limitations exist with this study that make it difficult to relate to our topic. First, a college-aged sample was studied in the Tierney et al. report. Second, subjects were seated and a 50 N force was applied to a pulley system attached at one end to the subject's forehead.

Another study tested the effects of a neck strengthening resistance program on head-neck dynamic stabilization in male and female collegiate soccer players (Mansell, Tierney, Sitler, Swanik, & Stearne, 2005). This study measured electromyographical data of the sternocleidomastoid and upper trapezius muscles. Anticipation of an impact (known vs. unknown) did not significantly affect upper trapezius muscle activity. While this would appear to counter our speculation, this same study demonstrated observable increases in forced-extension head-neck segment kinematic data (acceleration and displacement) when the force application was unknown compared to when it was anticipated. Similar findings were observed for forced-flexion head-neck segment kinematic data. This would suggest that while muscle activity levels appear equal, not anticipating a heading task results in greater kinematic movement (displacement and acceleration). Due to differences between the two samples, it is difficult to draw relationships to this current study. While this work provides interesting data, it relates to head-neck stabilization through cervical muscle activation, as actual strength measures were not recorded.

There is still strong anecdotal support for the role neck musculature may play in reducing the risk of mild TBI that is worthy of investigation in a young, at-risk sample. The question remains: why do humans sustain mild TBI at relatively low loads and other animals (i.e. woodpeckers, rams, buffalo, etc) appear to be unaffected by repetitive high-magnitude loading? The woodpecker, for example, decelerates at a rate of approximately 1000 g each time its beak strikes the tree (May, Fuster, Haber, & Hirschman, 1979). It is believed the powerful muscles of the head and neck act not only to control the rapid head movement of the woodpecker, but also serve to concurrently absorb the energy generated by the collision of their bills with the wood (May, Fuster, Newman, & Hirschman, 1976). Further investigation into this area reveals the woodpecker typically strikes the tree with its bill in a linear fashion. Ommaya and Genarelli report that concussion almost always occurred following angular acceleration, but very few of their primate subjects experienced concussion following a linear mechanism (Ommaya & Gennarelli, 1974).

Player Behaviors

While the purpose of this study is not to understand coaches' and players' aggressive behavior, it is difficult to ignore this aspect of ice hockey; and a brief discussion of this is warranted in the context of the current literature review. Though sportsmanlike conduct is promoted extensively by USA Hockey and Hockey Canada, the culture of the sport among its participants often predicates a mentality among players to ignore injury, play recklessly, and encourages unsportsmanlike conduct such as fighting and illegal checking. In the United States, a study of Peewee-level players reported that fighting broke out in approximately 17 of 52 games observed; and players considered fighting a natural consequence of the game and experienced a certain resignation about fighting (Gerberich, et al., 1987). Another interesting finding reported by Brust et al. is that while 100% of coaches felt sportsmanship was "real important," only 59% of players shared this attitude (Brust, et al., 1992). Parents and coaches, in this sample, viewed the enforcement of rules as being the most important factor in reducing injuries.

Methodological considerations

Subject pool

This dissertation includes Bantam- and Midget-aged ice hockey players. Players in these age levels are typically 13-14 and 15-16 years of age, respectively. Previous work has established anthropometric and biomechanical force profiles for each player in a sample of youth ice hockey players (Bernard, et al., 1993). The height and mass of Bantam players differed by as much as 41 cm (16.14 inches) and 48 kg (105.6 pounds), respectively. Further, when the authors simulated body checking between the smallest and largest players, they observed a 357% difference in the force of impact.

The Canadian Academy of Sports Medicine notes that serious injury in ice hockey begins to appear in the Peewee level, and continues to escalate in Bantam-aged players. They further assert body checking should not be allowed because of the differences in body size between players (Sullivan, 1992). In a study of children's ice hockey injuries (Brust, et al., 1992), more than half of the injuries occurred at the Bantam level (54%) compared to younger players in Peewee (27%) or Squirt (19%). They posit the reason for this trend to be a difference of as much as 53 kg (116.6 pounds) and 55 cm (21.65 inches) between players on Bantam teams in this study. The University of North Carolina at Chapel Hill youth hockey athletes in this study of the Bantam age level had observed differences within Bantam-aged players to be as high as 44.18 kg (97.20 pounds) and 35.56 cm (14 inches).

Head Impact Telemetry (HIT) System

This study used HIT System technology (Crisco, Chu, & Greenwald, 2004) incorporated within the Sideline Response System (Riddell; Elyria, OH). A major component of the HIT System is the installation of six single-axis accelerometers that are custom fitted directly into the foam component of the ice hockey helmet (see Methods section and Figure 3.1). These accelerometers are positioned tangentially to the head in Reebok RBK 6K/8K (Reebok-CCM Hockey, Inc.; St-Laurent, QC, Canada) and Easton Stealth S9 (Easton Sports, Inc.; Van Nuys, CA) helmets. In order for head acceleration data to be recorded, the acceleration of any individual accelerometer must exceed a desired threshold; this threshold is set at 10 g for this study. Precedence for this cutoff threshold has been established from previous work (Guskiewicz, et al., 2007; McCaffrey, et al., 2007; Mihalik, et al., 2007; Mihalik, Guskiewicz, et al., 2008). A question often raised is whether the HIT System measures head acceleration or helmet acceleration. The HIT System has been laboratory validated at multiple test facilities against Hybrid III test dummies, which are considered to be the gold standard in impact biomechanics testing. In all cases, the mean difference in linear acceleration measurements were within 8%, and as low as 2%, of Hybrid III measurements. In short, the HIT System measures head acceleration and not helmet acceleration (Manoogian, McNeely, Duma, Brolinson, & Greenwald, 2006). Further, data collected may be used to compute standard measures of head acceleration such as linear and rotational accelerations, Head Injury Criterion, Gadd Severity Index, and the Head Impact Technology severity profile (HITsp). The telemetry system is capable of transmitting accelerometer data from as many as 100 players over a distance well in excess of the length of a standard international ice surface. In addition, information from 100 separate head impacts can be stored in non-volatile memory built into the accelerometer device (i.e. resides in the helmet proper) in the event communication between the helmet and sideline controller is temporarily unavailable. This system has been successfully employed in the football studies (Brolinson, et al., 2006; Duma, et al., 2005; Guskiewicz, et al., 2007; McCaffrey, et al., 2007; Mihalik, et al., 2007; Schnebel, Gwin, Anderson, & Gatlin, 2007) and has recently been employed in youth ice hockey (Mihalik, Guskiewicz, et al., 2008).

Linear acceleration and rotational acceleration both represent common measures used in the field of head injury biomechanics. Unfortunately, limited injury data has supported linear or rotational acceleration as the gold standard. That is, their respective sensitivity and specificity to diagnosing concussion have been questioned in the literature as a result of previous work performed at The University of North Carolina at Chapel Hill (Guskiewicz, et al., 2007; McCaffrey, et al., 2007; Mihalik, et al., 2007). These latter studies have suggested that variables such as impact location and impact duration may play an equally important role in identifying injuries among athletes. Greenwald et al. (2008), in more recent work, sought to identify a composite variable to do just that. Using collapsed data across a number of institutions that have implemented the HIT System, they performed principal component analyses on 289916 head impacts collected from 449 athletes. They report that all biomechanical measures (linear acceleration, rotational acceleration, and Head Injury Criterion) were more predictive of concussion than simply guessing. When they confine the data to include only the top 1% and 2% of all impacts (of which 82% of all their injuries were accounted for in this subset), only a principal component score containing linear acceleration, rotational acceleration, and impact duration, weighted by impact location (termed wPCS) was found to be more predictive of injury than the other classical measures of head impact severity alone. However, with limited numbers of injuries it is impossible to ascertain fully its superiority, the wPCS appears to stand out as a novel measure in this area of research. In the commercial version of the HIT System, called the Sideline Response

System (Riddell; Elyria, OH), wPCS is referred to as the Head Impact Technology severity profile (HITsp).

Ideally, we would like to instrument both the striking and struck players with an instrumented helmet. This poses a limitation to our study and is the direct result of both funding and geographical issues. However, given the number of collisions we anticipate, we are confident that we will observe a high number of both striking collisions and struck collisions for the players in our sample.

Carolina Hockey Evaluation of Children's Checking (CHECC) List

The CHECC List represents the first attempt by researchers to develop a video analysis-grading rubric designed to specifically evaluate a player's technique while delivering or sustaining a body collision. The objective of the CHECC List is to provide amateur coaches and parents an easy way, without the need for expensive motion capture equipment, to break down a body collision using a simple video feed from a standard video camera. By doing so, it is believed coaches will be able to use this information to modify an individual's ability to more safely minimize head impact forces during a body collision.

CHAPTER III

METHODOLOGY

Study Design

This study employed a prospective quantitative research design in order to address the study's hypothesis-driven specific aims. Data collection occurred through the end of the 2008-09 ice hockey season, with data reduction commencing immediately and continuing through May 2009. Data pertaining to this dissertation were also collected over the course of the 2007-08 season. Testing and data collection occurred at a number of local, state, national, and international venues as the ice hockey teams traveled abroad for their competitions. Data reduction and analysis took place at The University of North Carolina at Chapel Hill. The study timeline is provided in ***Figure 3.1***.

Participants

The study included Bantam- and Midget-level ice hockey players aged 13-14 years and 15-16 years, respectively. These players participated in at least three practice or game sessions each week over the course of the playing season. These players represented a convenience sample of participants from two elite AAA-level ice hockey teams. A detailed explanation of the study was provided for all the athletes, coaches, and parents, prior to the start of the season. While data pertaining to previous history of concussion and years of playing experience were collected, they did not serve as exclusion criteria. Parental permission and minor assent forms approved by the university's institutional review board were signed by each parent and player, respectively, prior to fitting an athlete with an instrumented ice hockey helmet (see Procedures section). The subject recruitment process employed in this study is provided in ***Figure 3.2***.

Instrumentation

Head Impact Telemetry (HIT) System

This study used commercially available Reebok RBK 6K and 8K helmets (2007-2008 cohort; Reebok-CCM Hockey, Inc.; St-Laurent, Quebec, Canada), or Easton Stealth S9 (2008-2009 cohort; Easton Sports, Inc.; Van Nuys, CA) modified to accept the Head Impact Telemetry (HIT) System technology (Simbex; Lebanon, NH). The helmet's foam liner was custom cut to accept six single-axis accelerometers, a battery pack, and the telemetry instrumentation (***Figure 3.3*** depicts an Easton S9 helmet model). The custom helmets passed both ASTM (1045-99) and CSA (Z262.1-M90) helmet standards and were approved by the Hockey Equipment Certification Council (HECC) for use during competition. The HIT System utilized spring-loaded accelerometer holders to maintain contact with the head during an impact event. This method has been shown to successfully decouple accelerometers from the head allowing for measurement of head—not helmet—acceleration (Manoogian, et al., 2006). These accelerometers were positioned tangentially to the head. Linear acceleration of the center of gravity (COG) of the head was computed using a least-squares regression algorithm (Chu, Beckwith, Crisco, & Greenwald, 2006; Crisco, et al., 2004). Data from the six accelerometers were collected at 1 kHz for a period of 40 ms (8 ms pre-trigger and 32 ms post-trigger) following the acceleration of any individual accelerometer exceeding 10 g. The data were time-stamped, encoded, stored locally, and then transmitted in real time to a sideline controller (antenna) incorporated within the Sideline Response System (Riddell; Elyria, OH) via a radiofrequency telemetry link. The sideline controller (***Figure 3.4***) was typically positioned along the playing surface sideboards or in the team's dressing room. Biomechanical measures of head impact severity (see *Data Reduction* section) were computed and stored. The HIT System was capable of transmitting accelerometer data from as many as 100 players over a distance well in excess of the length of a standard international ice surface. In some instances when the real-time transmission of head impact data was unavailable (i.e. signal interruptions, sideline system not set up, etc), information from 100 separate head impacts were capable of being stored in non-volatile memory built into the acceleration monitoring system.

Player shift recording

Appendix A depicts a data collection form used to record the number of playing shifts during games and scrimmages. This form was used by researchers positioned immediately behind the players' bench during games to document the number of shifts played during a competition session. One shift was defined as an incident where the player stepped onto the ice to actively participate in the play during game and scrimmage events, and was terminated when the player returned to the team bench. Only one shift was counted in instances where there was a stoppage in play (i.e. offside, icing, penalty, goal) but the player remained on the ice for the start of the next play. The length of a typical shift ranged from 45 seconds to one minute.

Video recording

This study employed the use of a standard digital video camera (Model: PV-GS35; Panasonic Corporation of North America; Secaucus, NJ) to record live game footage onto 60-minute miniDV tapes (Model: M-DV60ME; JVC Americas Corp.; Wayne, NJ). The video camera was capable of recording video footage at 120 Hz, and had a built-in sports exposure mode that allowed for clear video recording of quick action. It was equipped with a 30X optical zoom and 1000X digital zoom, allowing for close-up and contained images regardless of where on the ice the play was occurring (***Figure 3.5***). Video footage was recorded during games and scrimmages for the primary purpose of addressing Specific Aims 3 and 4, and addressing the descriptive factor of the striking player and player struck (addressed as part of Specific Aim 1).

Strength and anthropometric testing

Cervical muscle strength was measured with isometric "break tests" using a hand-held dynamometer (Model: 01163; Lafayette Instrument Co.; Lafayette, IN). The unit was small and convenient to use, and was similar to handheld dynamometers readily available to clinicians. It weighed 260 grams (10 ounces), was capable of measuring strength up to 136.36 kg (300 pounds) of force, and had an accuracy of ± 0.45 kg (1.0 pound) in the high range and ± 0.23 kg (0.5 pounds) in the low range. Length and diameter anthropometric measurements were recorded using a small anthropometer (Model: 01291; Lafayette Instrument Co.; Lafayette, IN). Circumferential measurements were recorded using a standard clinical tape measure. The procedures employed to measure strength and anthropometrics are described in the *Procedures* section (see below) and illustrated in Figures 3.7 to 3.14.

Faught Aerobic Skate Test (FAST)

The FAST was designed to assess maximum aerobic power using a hockey-specific on-ice testing protocol. Players began at a slow pace and progressed to exhaustion by following cues delivered by the FAST audio track through the arena's public address system. The FAST was designed to promote continuous skating in a counter-clockwise direction without any stops and starts. The FAST took approximately 10 minutes to complete at each administration. Players were instructed to divide into 2 equal groups positioned at opposite corners of the ice. The subjects were required to skate a 160-foot (48.77-meter) distance from one end of the ice surface to the other within the allotted time (***Figure 3.6***). The allotted time for each length began at 15 seconds and decreased by 0.5 seconds every third length (***Table 3.1***). Previous work found no differences between the maximal oxygen consumption (VO_2max) predicted from the FAST and VO_2max measured during a modified Bruce incremental treadmill protocol performed in a laboratory in young ice hockey players (Petrella, Montelpare, Nystrom, Plyley, & Faught, 2007). Good reliability of this instrument has been demonstrated in adult ice hockey players ($r = 0.76$) (Faught, Nystrom, & Montelpare, 2003); as well as in a younger Bantam-aged cohort ($r=0.81$) (Petrella, Montelpare, Nystrom, Plyley, & Faught, 2005).

Carolina Hockey Evaluation of Children's Checking (CHECC) List

The CHECC List (***Appendix B***) was scored on 11 readily observable features of human movement during a body collision. These included relative body positioning including knee position, trunk position, and stance width. The use of shoulders, elbows, hands, or a stick, during the collision was also included. Also included were whether a player was looking ahead in the direction of movement, and if he appeared to see the impending body collision. The overall impression of collision "quality," whether the player was striking an opponent or was the player struck, whether an infraction took place during the collision, and where on the playing surface the collision occurred were all variables of interest included on the CHECC List. The CHECC List was developed along with feedback from a team of USA Hockey-certified coaches directly involved in the coaching of youth ice hockey players. Every member of this group verified the criteria and felt they addressed components of a body collision of interest to youth ice hockey coaches, and those that could be modified through some form of intervention program at the youth/amateur ice hockey level. The intratester reliability of the CHECC List was established as a part of this study. A subsample of body collisions observable on the video footage was re-evaluated by the principal investigator no less than 3 months after the end of the initial video analyses. Intrarater Kappa agreements ranged from 0.40 to 0.92 for the 15-item CHECC List. Interrater agreement suggested moderate to strong agreement between hockey coaches with no scientific experience when reviewing a subsample of 25 collisions independently of each other (***Table 3.2***).

The Buss-Perry Aggression Questionnaire

The Buss-Perry Aggression Questionnaire (BPAQ) consisted of a 29-item, five-point Likert scale ranging from one ("extremely uncharacteristic of me") to five ("extremely characteristic of me"). There are four widely used subscales of the BPAQ established on the basis of factor analyses (Buss & Perry, 1992): physical aggression (9 items), verbal aggression (5 items), anger (7 items), and hostility (8 items). In order to foster an environment whereby participants did not feel pressured in any particular way, the players were asked to complete a paper version of the BPAQ on their own time at home. Test-retest reliability had previously been established for the physical aggression (0.80), verbal aggression (0.76), anger (0.72), and hostility (0.72) subscales of the BPAQ (Buss & Perry, 1992). The BPAQ distributed to the athletes is available in ***Appendix C***.

Procedures

Helmet fitting

Prior to the start of the season, players were measured for helmet and facemask size. They were properly fit with Reebok RBK 6K/8K (2007-2008 cohort) or Easton Stealth S9 (2008-2009 cohort) helmets by a certified athletic trainer (ATC). The ATC instructed each participant to wet his hair to simulate sweating. Facemasks owned and used by the players were secured to the new helmet if they were deemed to be in good condition, and if the facemask was compatible with the new helmet. Otherwise, players were asked to purchase a new facemask or were allocated one by the investigators. They were then fitted with the helmet such that the brim of the helmet rested 3.5 cm (two finger-widths) above the participant's eyebrows. The facemask chinstrap fit tightly under the chin and was securely fastened to the helmet. As a quick test, participants were instructed to hold their head still while the principal investigator attempted to move the helmet. If the investigator was able to move the helmet with no movement of the head, the fitting procedure was repeated. Helmet and facemask fit was verified on a biweekly basis to ensure proper fit throughout the course of the playing season.

Video recording

The principal investigator recorded video games over the course of the 2007-08 ice hockey season. Prior to each game, the video camera date- and time-stamping features were synchronized to the Sideline Response System date and time. In order to maximize video image size, the camera followed movement of the puck in an attempt to isolate body collisions during play and to maximize the capture of these events by the camera. Impacts occurring outside the view of the camera were excluded from our analyses, as there was no way to analyze these collisions using the CHECC List. In order to make the most of the miniDV tapes, video recording began as the players lined up for a faceoff, and was paused when a whistle was blown signaling the end of a play. As such, one tape was often capable of storing video collected during one complete game. In some instances, late body collisions occurring shortly after the officials blew the whistle were not captured and, therefore, were excluded from our analyses. Since the video footage was time-stamped, a short video clip of the scoreboard was taken at the start of each playing period, enabling us to assign HIT System data to a particular playing period.

Cervical strength and anthropometric measurements

Cervical muscle strength was measured as described by Kendall et al. with isometric "break tests" using a hand-held dynamometer (Kendall, McCreary, & Provance, 1993). Strength testing for cervical rotators followed the procedure described by Hislop & Montgomery (Hislop & Montgomery, 1986). Strength of the anterior neck flexors, and bilateral strength measurements of the anterolateral neck flexors, cervical rotators, posterolateral neck extensors, and upper trapezius were recorded. Bilateral peak strength measurements were averaged into a single measure. Two practice trials were performed prior to three test trials for each direction of motion, with a 30-second rest period between trials; each trial lasted 3 seconds. The maximum break force for each of the three test trials were averaged, and then normalized to the player's body mass to facilitate comparisons between individuals of varying body size. Pilot data suggested good to excellent intrasession reliability (all $ICC_{3,1} \geq 0.821$) and precision of the measurements for all the strength tests. These results are presented in ***Table 3.3***.

In order to measure anterior neck flexor strength, the subject was positioned supine with his elbows bent and hands overhead, while resting on a treatment table (***Figure 3.7***). The subject attained the test position by lifting his head from the table with the chin depressed and approximated toward the sternum. Pressure was delivered through the hand-held dynamometer against the forehead in a posterior direction.

The subject was then asked to maintain the same body position, while rotating his head to one side in order to evaluate anterolateral neck flexor muscle strength. The subject lifted his head off the table and the investigator applied pressure against the temporal region of the head in an obliquely posterior direction (***Figure 3.8***).

The patient remained supine with his cervical spine in neutral in order to test the cervical rotators. The head was supported on the table with the face turned as far to one side as possible. The patient attempted to rotate his head slightly toward the neutral (i.e. face up) position against resistance directed towards returning the subject back to the fully rotated position (***Figure 3.9***).

The subject then moved into a prone position with his elbows bent and hands overhead while resting on a table and moved into posterolateral neck extension with his face turned to the side being tested (***Figure 3.10***). The investigator applied pressure against the posterolateral aspect of the head in an anterior direction.

Upper trapezius strength testing was accomplished by placing the subject in a seated position. The subject was then asked to elevate the acromial end of the clavicle and scapula, bring his occiput toward the elevated shoulder with the face turned in the direction opposite the side of testing (***Figure 3.11a***). The head was stabilized and the investigator attempted to depress the scapula (***Figure 3.11b***).

Anthropometric measurements included subject height, mass, head-neck segment length, neck circumference, neck medial-lateral diameter, neck anterior-posterior diameter, head circumference, head medial-lateral diameter, and head anterior-posterior diameter. Height and mass were recorded on a standard medical scale. Head-neck segment length was measured as the vertical distance between the seventh cervical vertebra (C7) spinous process and the top of the head while the subject looked directly in front of them (***Figure 3.12***). Neck circumference and diameter measurements were taken at the level just above the thyroid cartilage (***Figure 3.13***). Head circumference was measured across the middle of the forehead as it would when fitting a player with an ice hockey helmet (***Figure 3.14a***). Head medial-lateral diameter was recorded just above the top of the ears (***Figure 3.14b***), and the anterior-posterior diameter was measured from the middle of the forehead to the middle of the posterior aspect of the head (***Figure 3.14c***). We observed very good intrasession reliability ($ICC_{3,1} \geq 0.782$) and precision for all anthropometric measurements.

Faught Aerobic Skate Test (FAST)

While players required no previous experience with the FAST, they performed a familiarization trial of the FAST during one practice session, and were tested for the purpose of this study in two subsequent on-ice sessions. To avoid a confounding contribution of non-test fatigue, the protocol was administered at the start of practice when it was assumed players were not experiencing fatigue. In order to remain consistent with game situations, players were wearing all of their playing equipment during the testing. The team was separated into two equal groups and positioned in opposite corners of the rink. The test instructions were played over the arena public address system, and the players were then cued with an audible “beep” to begin skating. Players skated counter-clockwise the length of the ice in order to continue to the next FAST length. The automated audio track provided the skaters with a three-second countdown near the end of each length as a warning of time remaining to complete the length. The time required to complete each FAST length decreased by 0.5 seconds every 3 lengths. A subject’s test was deemed completed when they were unable to reach the destination line in the allotted time for two consecutive lengths, or if they voluntarily discontinued owing to fatigue. The subject’s final successfully completed length was recorded as their maximum FAST length.

Evaluation of video footage

Prior to any analysis of game video, the principal investigator met with members of a coaches’ committee and reviewed a select number of body collisions with them to ensure agreement with how the body collisions should be viewed and evaluated. The principal investigator then independently analyzed all game video footage in a quiet environment in order to avoid distractions. As stated earlier, video footage was date and time-stamped to match the data collected by the HIT System. The principal investigator employed the CHECC List to assist in the evaluation of each video-recorded impact.

In order to standardize the evaluation of player collisions based on video recordings, a committee consisting of five USA Hockey-certified coaches (4 head coaches and 1 assistant coach; one with Level 5, three with Level 4, and another with Level 3 coaching certifications) with extensive experience in coaching ice hockey was created. The pre-study expectation was that many collisions would be difficult to evaluate from the video footage and would require an adjudication process to remediate these cases. However, upon completion of all viewable collisions, only three (of 669 evaluated) proved difficult. Due to this low number, these collisions were omitted entirely from our analyses. Instead, the committee independently reviewed 25 collisions selected at random to ensure intertester reliability in the use of the CHECC List. The members of the review committee were able to view this video footage, but were not provided with information related to biomechanical measures of head impact severity associated with the collisions, or to how the principal investigator initially evaluated the collision.

Data reduction

Biomechanical measures of head impact severity

The raw head impact data were exported from the Sideline Response System into Matlab 7 (The Mathworks, Inc.; Natick, MA), where data were reduced to include only those impacts sustained during practices, scrimmages, and games. Impacts occurring outside of team-sanctioned events (i.e. pick-up hockey, impacts imparted to the helmet during handling of equipment or travel, etc) were thus omitted from our analyses. Only impacts registering a linear acceleration greater than 10 g were included for the purposes of our analyses as impacts below this threshold are considered negligible with respect to impact biomechanics and their relationship to head trauma. As each impact was linked to a player enrolled in our study by unique identifiers, we were able to easily associate impacts that belonged to a particular player, and to categorize those impacts based on player position information we had collected at the start of the season. Resultant linear head acceleration, resultant rotational head acceleration, and Head Impact Technology severity profile (HITsp), were the outcome measures of interest and retained for further analysis. These variables were automatically computed by the HIT System. Azimuth and elevation data collected from the accelerometers in the instrumented helmets were used to categorize the location of head impacts. In keeping with our previous work studying Division I collegiate football players (Mihalik, et al., 2007) and youth ice hockey players (Mihalik, Guskiewicz, et al., 2008), we had originally proposed that any impact sustained at an angle greater than 60° in elevation from a horizontal plane through the estimated location of the center of gravity of the head would be categorized as an impact to the top of the head. In developing the wPCS (HITsp as it appears in the HIT System), Greenwald et al. (2008) established the top of the head to represent all impacts sustained at an angle greater than 65° in elevation from horizontal. As such, we have adopted this definition. Front head impacts were defined as those occurring within 45° from either side of the sagittal midline. Similarly, impacts within 45° of either side of the sagittal midline posterior to the head were categorized as an impact to the back of the head. Impacts sustained within 45° of the frontal plane were accordingly categorized as a side impact, regardless of whether the impact occurred to the right or left side of the head (***Figure 3.15***).

Player shift recording

At the end of each playing period, the total number of playing shifts was tabulated and recorded. These data were used in combination with the biomechanical measures of head impact severity to address Specific Aim 2. These data were entered into the statistical spreadsheet and used as part of our analyses addressing the effects of game-related exposure on biomechanical measures of head impact severity.

Video footage

All raw video footage was imported from the video camera connected to a Windowsbased laptop computer using a standard universal serial bus (USB) cable. This was done using MotionDV Studio LE (Version 6.0; Panasonic Corporation of North America; Secaucus, NJ). Once raw video footage was imported, date and time-stamp information were inlayed onto the video, and the video was then exported to a DivX-encoded audio video interleave (AVI) file for storage. Video playback during CHECC List evaluations was performed on a personal desktop (iMac; Apple, Inc.; Cupertino, CA) using QuickTime Player Pro (Version 7.5.5; Apple Inc.; Cupertino, CA).

Strength and anthropometric testing

The peak force generated during each of the strength-testing trials was ensemble averaged and recorded for the purpose of data analysis. In cases where bilateral measurements were recorded for the left and right sides, the ensemble mean of each side was averaged together to result in a single measure of strength for that muscle group. For example, the mean of the trials for the right anterolateral cervical flexors were averaged with the mean of the trials for the left anterolateral cervical flexors into a single measure for anterolateral cervical flexion strength. The strength measures were then normalized to the players' mass. A variable of total neck strength was computed as the sum of the five individual relative strength measures. Head and neck anthropometrics yielded a single measure for each of the following: head-neck segment length, neck circumference, neck medial-lateral diameter, neck anterior-posterior diameter, head circumference, head medial-lateral diameter, and head anterior-posterior diameter. Body mass index (BMI) was also computed.

Carolina Hockey Evaluation of Children's Checking (CHECC) List

Every body collision was assigned a unique event identification number. This identification number was indicated on the CHECC List form used to evaluate a given body collision. Each of the eleven body position criteria (identified in Appendix B) represented a dichotomous response (yes or no). These variables were included in the statistical models used to address Specific Aim 4.

The Buss-Perry Aggression Questionnaire

Data from the BPAQ were compiled and entered into our database. It was originally proposed that these measures of aggression were to be used as covariates of interest designed to control for differing levels of aggression that may manifest in more reckless or aggressive behavior while participating in ice hockey. Preliminary analyses suggested these measures were not useful in this capacity. As such, player aggression was added to Specific Aim 5 and analyzed as an independent variable of interest. The BPAQ was factored into four subscales as follows: physical aggression, verbal aggression, anger, and hostility. In addition to these subscale scores, a total score was also computed and these scores were used in our analyses. Though not a part of the BPAQ, the total number of penalties in minutes (PIM) was also analyzed. This is typically a measure of player aggression used in ice hockey, with those athletes with higher PIM thought to exhibit a more physically aggressive style of play.

Statistical analyses

For our continuous outcomes of playing shift exposures, strength, anthropometrics, general aerobic fitness, and aggression, we categorized our data into three groups (tertiles). For example, the tertiles allowed us to model the differences in biomechanical measures of head impact severity between the strongest players, those with moderate strength, and the weakest players, for our measures of strength. Since head impact data were highly skewed in favor of low-magnitude impacts, data were transformed using a natural logarithmic function in order to meet the assumptions of normality for the proposed parametric analyses described below. All estimates obtained from our analyses were then back-transformed to their original scale for purposes of presentation.

Descriptive analyses (means and 95% confidence intervals) were calculated for the three biomechanical measures of head impact severity (dependent variables): resultant linear acceleration, resultant rotational acceleration, and the HITsp. In order to address our specific aims, separate random intercepts general mixed linear models were employed for each of our dependent variables. Player represented one level in each statistical model as a repeated factor. Independent variables as determined by each specific aim were included in the statistical model when appropriate. Player position, event type, location of head impact, and whether a player was the striker or the player being struck, served as independent variables of interest included in separate statistical models used to address Specific Aim 1. Specific Aim 2 required the inclusion of the number of shifts played as a level in the statistical model in addition to the player repeated factor. Infraction type was included as an independent variable (in addition to player) in the statistical models employed to address Specific Aim 3. Specific Aim 4 required several random intercepts general mixed linear models, each one modeled to include separate independent variables (in addition to player) to address all of the research questions associated with this aim. Lastly, Specific Aim 5 required the inclusion of measures of cervical muscle strength measures, head and neck anthropometrics, general aerobic fitness, and player aggression, as separate independent variables in the statistical model. ***Table 3.4*** represents our general research objectives, dependent and independent variables, and proposed statistical methods.

In addition to the originally proposed statistical methods, we performed a number of additional exploratory analyses by including covariates in our model. As player cervical muscle strength, anthropometrics, aerobic fitness, and player aggression were not evaluated for our Year 1 cohort, we were unable to contrast these variables against the descriptors observed using the CHECC List. We felt that body mass index (BMI) would serve as a potential surrogate for cervical muscle strength in our statistical models and, as such, performed a Pearson bivariate correlation between BMI and total neck strength. We found that those who had larger BMI values typically had lower neck strength; that is, those who were more fit were stronger ($r_{30} = -0.618$; $P < 0.01$). We therefore used BMI as a covariate to represent cervical muscle strength during some of our exploratory analyses. Random intercepts general mixed linear models (PROC MIXED) were performed in SAS/STAT (Version 9.1; SAS Institute, Inc.; Cary, NC). The level of significance was set at $P < .05$ *a priori*.

Study Process	2007-08 Season	2008-09 Season
Subject recruitment		
Bantam AAA sample		
Midget AAA sample		
Data Collection/Procedures		
HIT System data		
Player shift recording		
Competition game footage		
Strength testing		
Head/neck anthropometrics		
Faught Aerobic Skate Test		
CHECC List		
Buss-Perry Aggression Questionnaire		
Adjudication (process		
Data reduction and analysis		
Data reduction		
Statistical analyses		

Figure 3.1. Timeline for subject recruitment, data collection, and data analysis

The shaded areas of the figure represent the timeline for subject recruitment, data collection procedures, and data reduction and analysis.

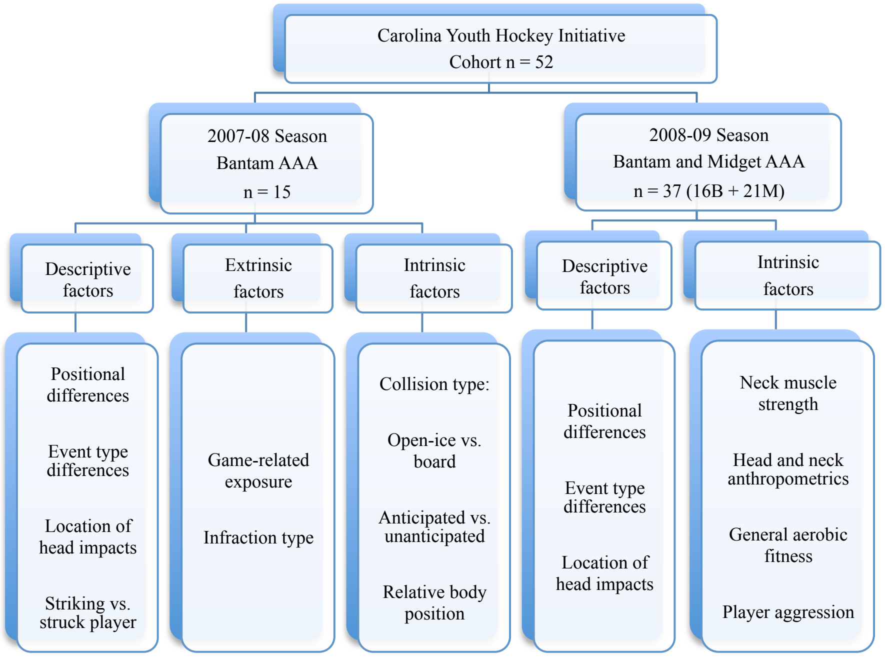

Figure 3.2. Subject recruit process

This figure illustrates the subject recruitment process employed for this project. Note that cervical muscle strength, head and neck anthropometric measurements, general aerobic fitness, and player aggression were only collected on Bantam and Midget AAA players comprising the 2008-09 season cohort.

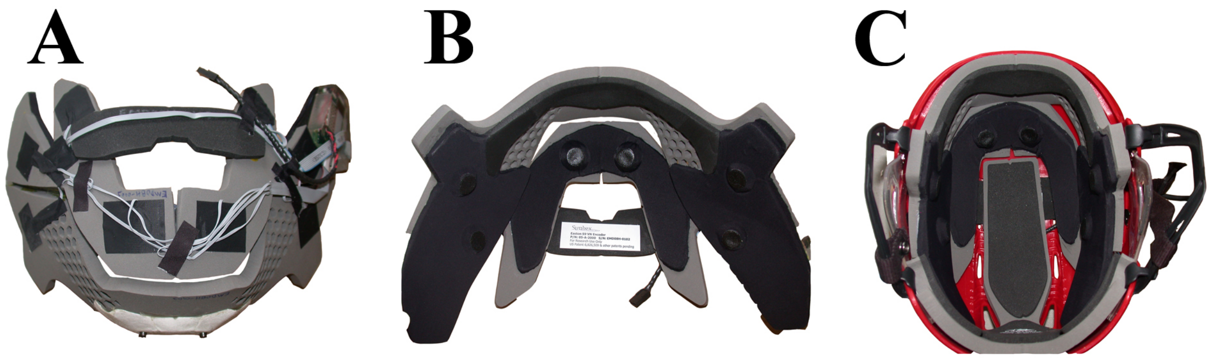

Figure 3.3. Accelerometer installation and setup

The protective foam of the ice hockey helmets were removed from the helmet shell (Easton Stealth S9 model depicted; RBK 6K/8K not depicted). Following this, six single-axis accelerometers were fitted into custom holes cut into the foam. The figure depicts the location of the helmet accelerometers in the protective liner (hard shell removed) in both exterior (A) and interior (B) views. The location of the six accelerometers (two in front, and two on each side) as viewed from the inside of a fully assembled playing ice hockey helmet (C).

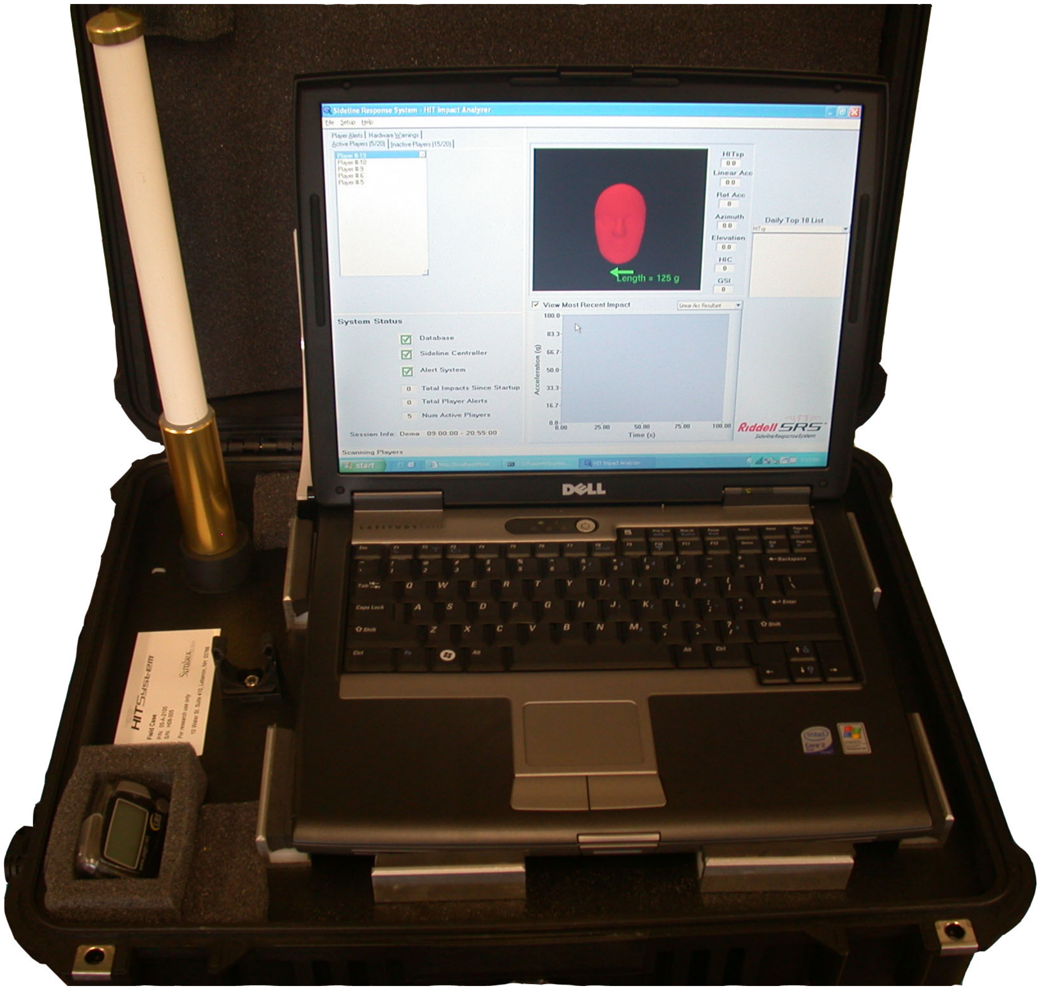

Figure 3.4. Riddell Sideline Response System

The Sideline Response System consisted of a sideline controller (antenna) connected to a laptop computer, and was positioned near the ice hockey playing area. Ice hockey helmets fitted with accelerometers collected head impact data which were then transmitted wirelessly and in real-time to the Sideline Response System.

Figure 3.5. Panasonic PV-GS35 video camera

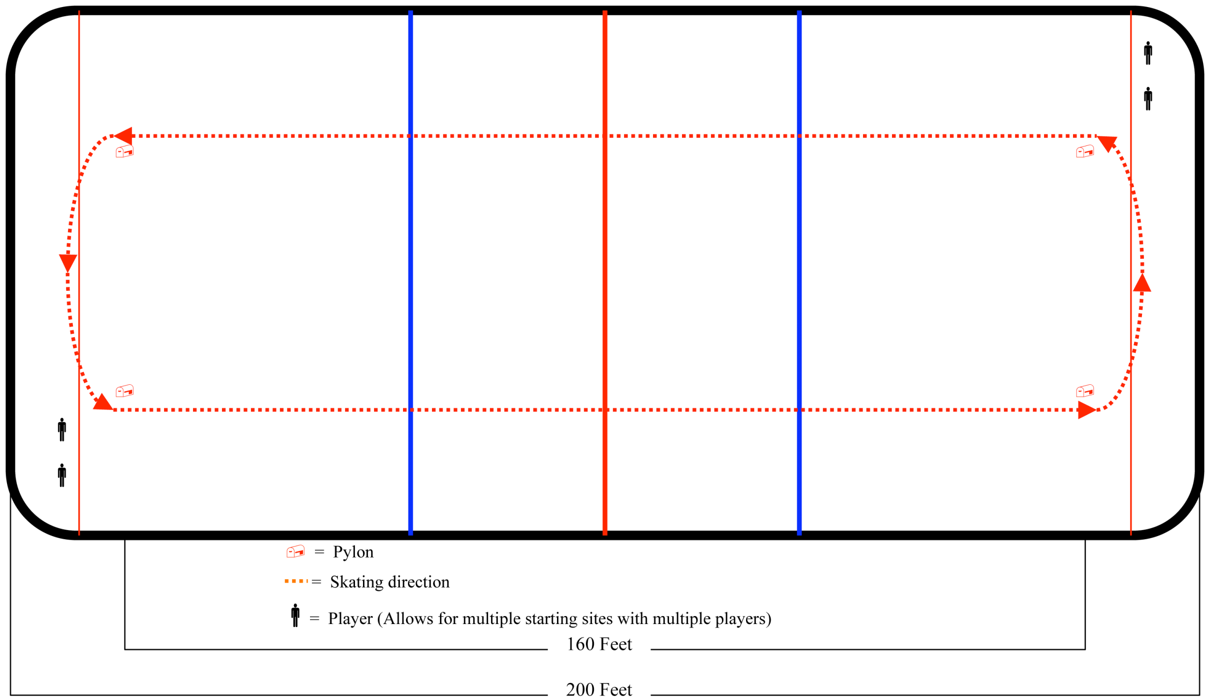

Figure 3.6. The Faught Aerobic Skate Test (FAST)

Figure 3.7. Evaluation of anterior neck flexor muscle strength

The subject was positioned supine with elbows bent and hands overhead, resting on a treatment table. The test position was attained by lifting the head from the table with the chin depressed and approximated toward the sternum. Pressure was delivered through the hand-held dynamometer against the forehead in a posterior direction.

Figure 3.8. Evaluation of anterolateral neck flexor muscle strength

The subject was positioned supine with elbows bent and hands overhead. The subject tucked the chin in towards the sternum, lifted the head off the table, and faced the left side. The investigator applied pressure against the temporal region of the head in an obliquely posterior direction. The procedure was repeated with the subject facing the right side. Illustrated is the position for testing the *right* anterolateral neck flexors.

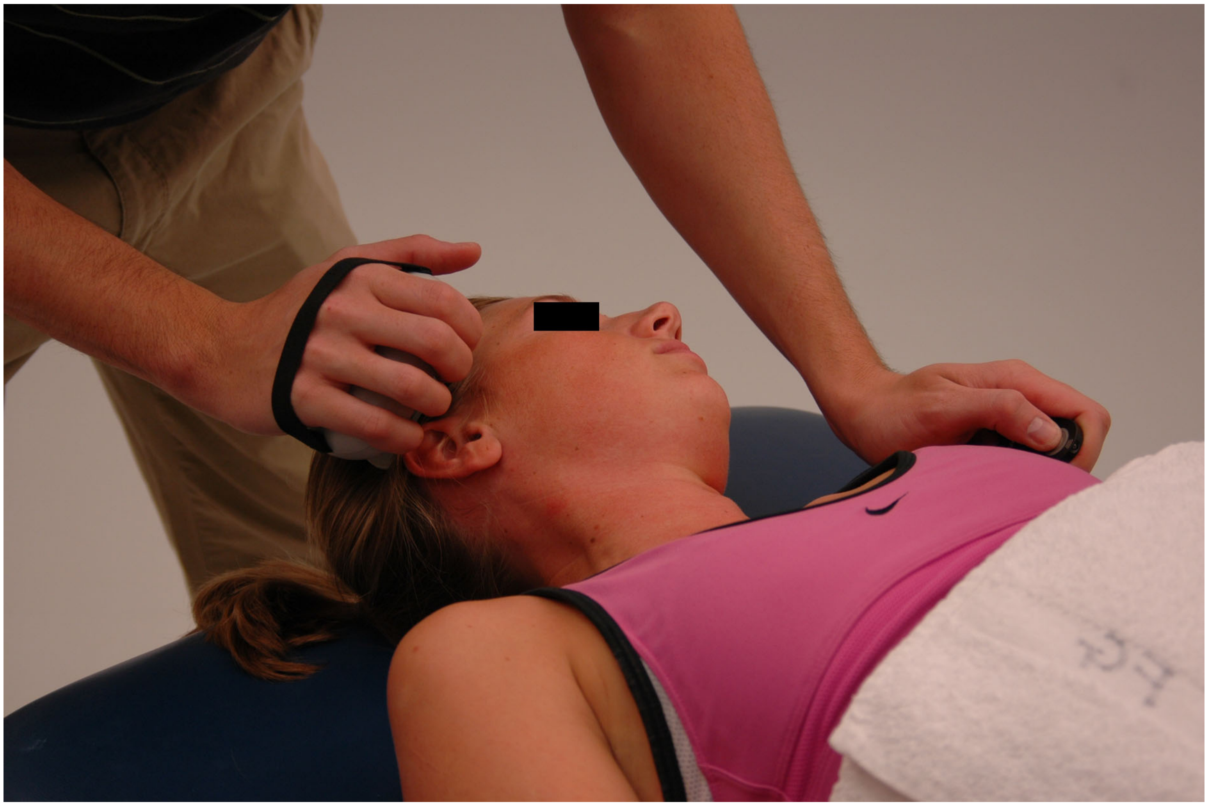

Figure 3.9. Evaluation of cervical rotation muscle strength

The patient was positioned supine with the cervical spine in neutral. The head was supported on the table with the face turned as far to one side as possible. The patient attempted to rotate the head slightly toward the neutral (i.e. face up) position against resistance directed towards returning the subject back to the fully rotated position. Illustrated is the position for testing the *right* cervical rotators.

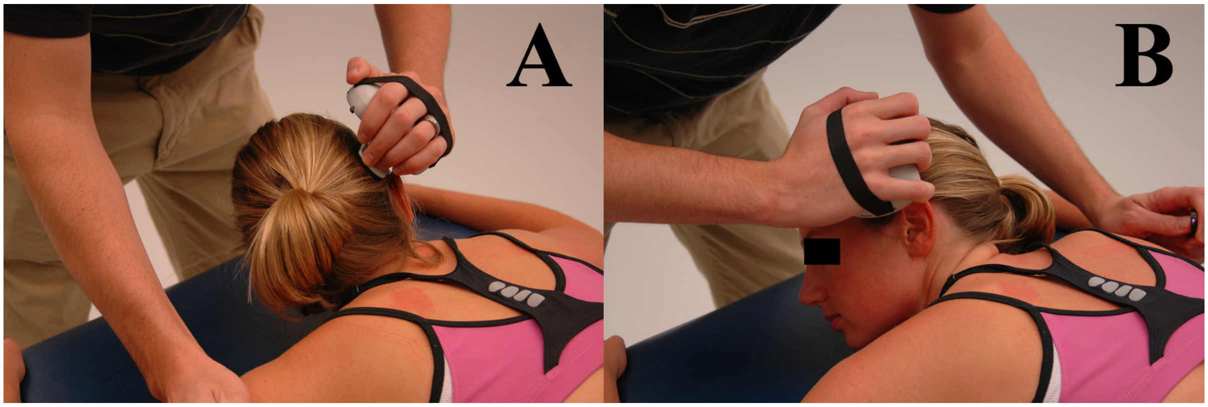

Figure 3.10. Evaluation of posterolateral neck extensor muscle strength

The subject was placed in the prone position with elbows bent and hands overhead while resting on a table. They were asked to move into posterolateral neck extension with their face turned to the side being tested. The investigator applied pressure against the posterolateral aspect of the head in an anterior direction. Illustrated are the test positions for the right (A) and left (B) posterolateral neck extensors.

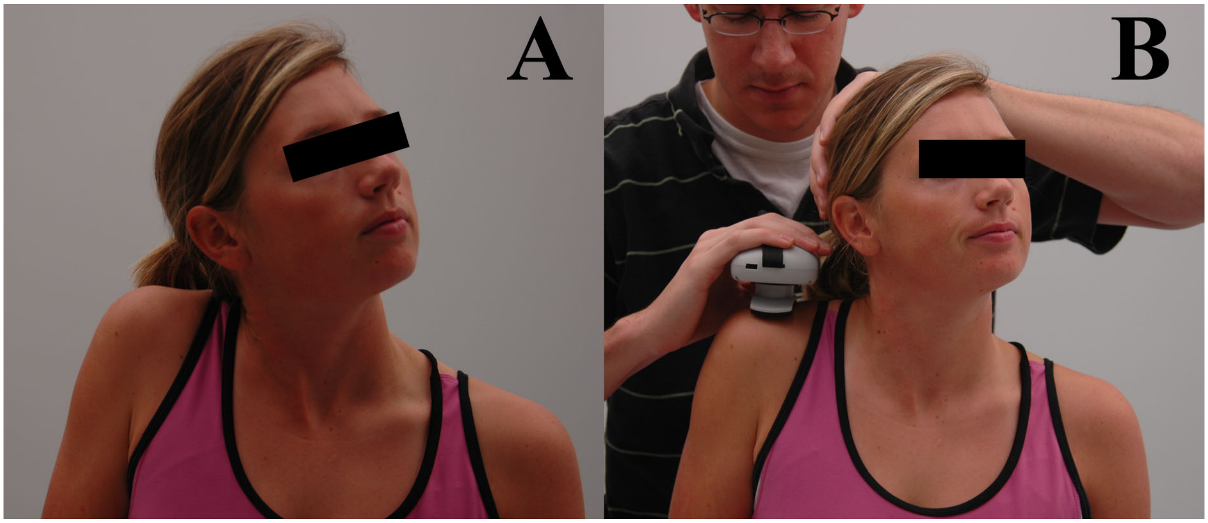

Figure 3.11. Evaluation of upper trapezius muscle strength

Upper trapezius strength testing was performed with the subject in a seated position. The subject was asked to elevate the acromial end of the clavicle and scapula, and to bring the occiput toward the elevated shoulder with their face turned in the direction opposite the side of testing. The head was stabilized and the investigator attempted to depress the scapula. Illustrated are the start (A) and test (B) positions for the right upper trapezius muscle.

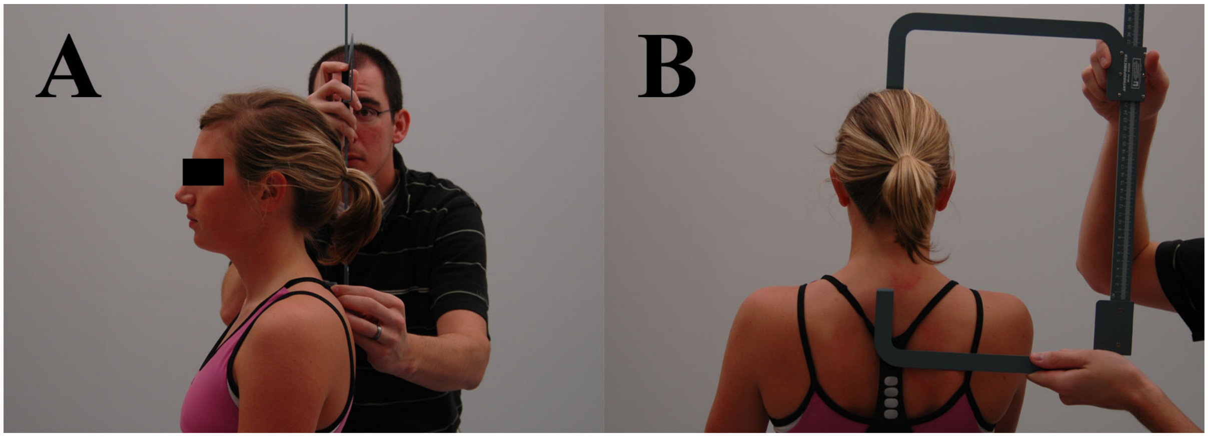

Figure 3.12. Measurement of head-neck length

Head-neck length consisted of the vertical distance between the spinous process of C7 and the top of the head measured while the subject was looking at an object at eye level. Illustrated are the lateral (A) and posterior (B) views of this measurement technique.

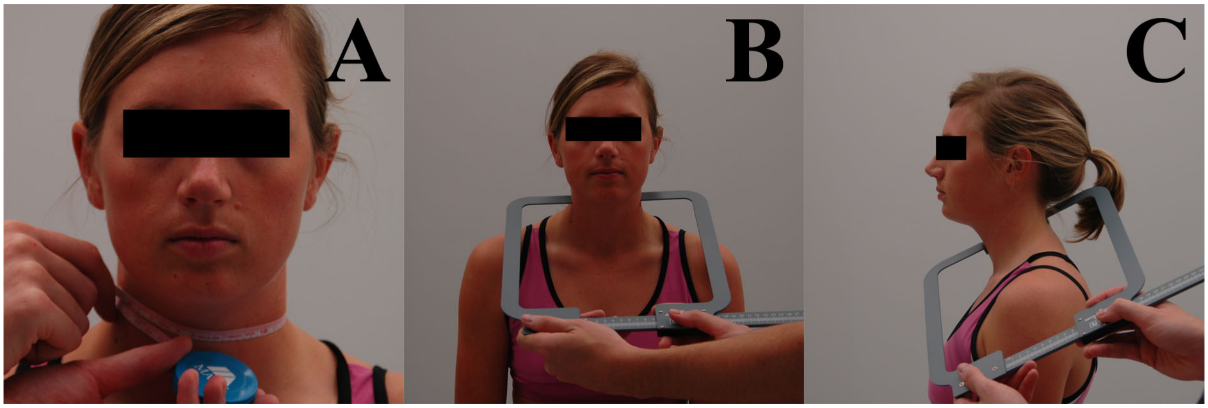

Figure 3.13. Neck anthropometric measurements

Neck circumference (A) was measured using a standard clinical tape measure. Medial-lateral (B) and anterior-posterior (C) diameters were measured using an anthropometer.

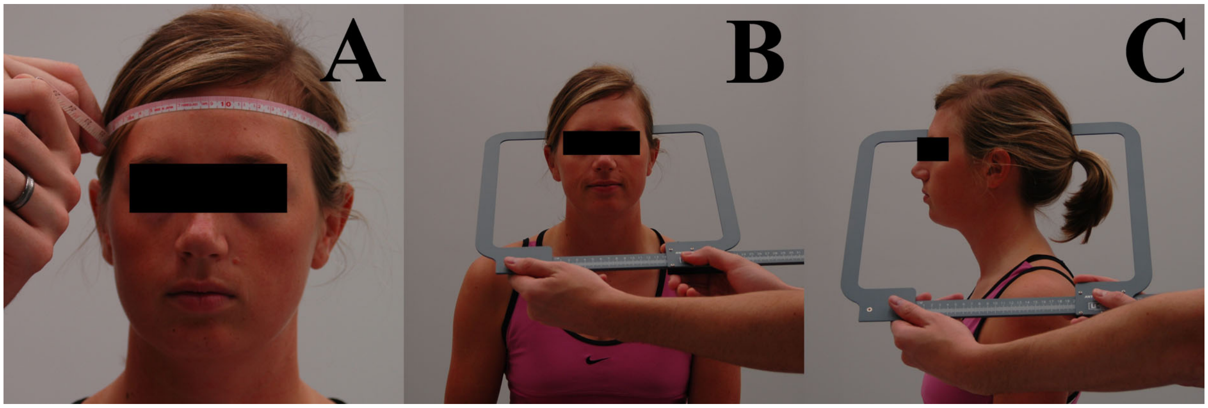

Figure 3.14. Head anthropometric measurements

Head circumference (A) were measured using a standard clinical tape measure. Medial-lateral (B) and anterior-posterior (C) diameters were measured using an anthropometer.

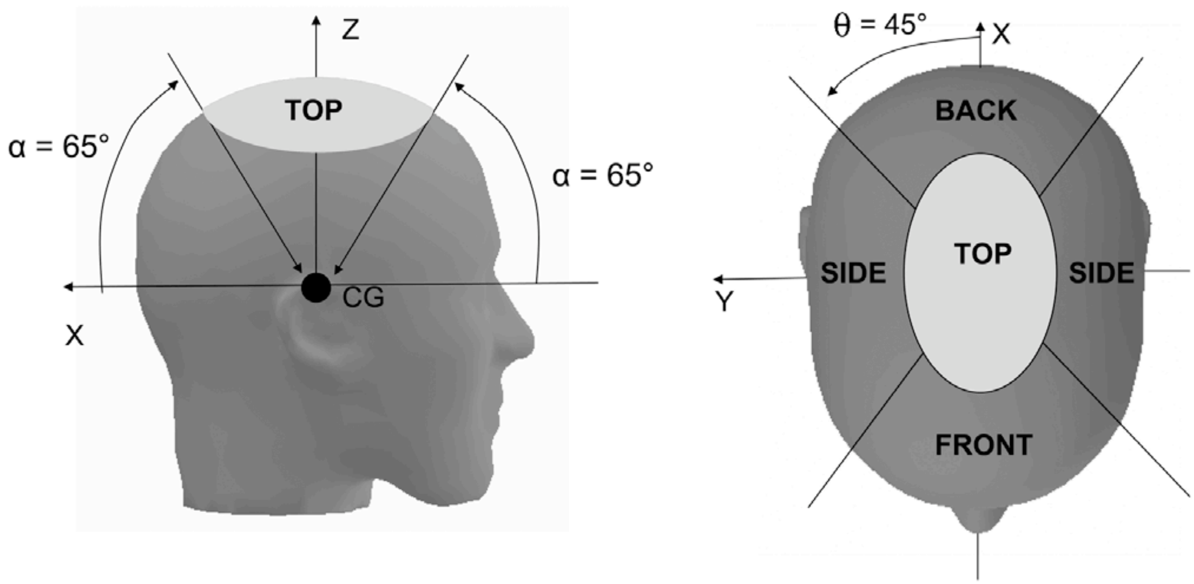

Figure 3.15. Categorization of head impact locations

Head impacts were categorized as back, front, side, or top, as defined by azimuth and elevation data collected at the time of each head impact. Regardless of azimuth position, any impact sustained at an elevation greater than 65° were categorized as an impact to the top of the head. This figure is from earlier work published by Greenwald et al. (2008) (with permission).

Table 3.1. Interval progressions of the Faught Aerobic Skate Test

Length	*Interval (sec)*	*Shuttle Time(min)*
1	15	0:15.0
2		0:30.0
3		0:45.0
4	14.5	0:59.5
5		1:14.0
6		1:28.5
7	14	1:42.5
8		1:56.5
9		2:10.5
10	13.5	2:24.0
11		2:37.5
12		2:51.0
13	13	3:04.0
14		3:17.0
15		3:30.0
16	12.5	3:42.5
17		3:55.0
18		4:07.5
19	12	4:19.5
20		4:31.5
21		4:43.5
22	11.5	4:55.0
23		5:06.5
24		5:18.0
25	11	5:29.0
26		5:40.0
27		5:51.0
28	10.5	6:01.5
29		6:12.0
30		6:22.5
31	10	6:32.5
32		6:42.5
33		6:52.5
34	9.5	7:02.0
35		7:11.5
36		7:21.0
37	9	7:30.0
38		7:39.0
39		7:48.0
40	8.5	7.56.5
41		8:05.0
42		8:13.5
43	8	8:21.5
44		8:29.5
45		8:37.5
46	7.5	8:45.0
47		8:52.5
48		9:00.0
49	7	9:07.0
50		9:14.0
51		9:21.0
52	6.5	9:27.5
53		9:35.0
54		9:40.5
55	6	9:46.5
56		9:52.5
57		9:58.5
58	5.5	10:04.0
59		10:09.5
60		10:15.0

Table 3.2. Interrater agreements for the CHECC List

CHECC Item	Interrater agreements			
	% 6/6 agree[1]	**% 5/6 agree**	**% 4/6 agree**	**% 3/6 agree**
Look ahead	60% (15/25)[2]	80% (20/25)	100% (25/25)	100% (25/25)
See hit	64% (16/25)	92% (23/25)	100% (25/25)	100% (25/25)
Knees flexed	40% (10/25)	72% (18/25)	96% (24/25)	100% (25/25)
Trunk flexed	40% (10/25)	72% (18/25)	96% (24/25)	100% (25/25)
Use shoulders	44% (11/25)	72% (18/25)	92% (23/25)	100% (25/25)
Use elbows	56% (14/25)	80% (20/25)	96% (24/25)	100% (25/25)
Use hands	48% (12/25)	80% (20/25)	96% (24/25)	100% (25/25)
Feet shoulder width apart	4% (1/25)	40% (10/25)	72% (18/25)	100% (25/25)
Use stick	72% (18/25)	80% (20/25)	100% (25/25)	100% (25/25)
Use legs	24% (6/25)	60% (15/25)	96% (24/25)	100% (25/25)
Passing/shooting	84% (21/25)	88% (22/25)	96% (24/25)	100% (25/25)
Overall impression	80% (20/25)	96% (24/25)	100% (25/25)	100% (25/25)
Striker/struck	84% (21/25)	96% (24/25)	100% (25/25)	100% (25/25)
Infraction	44% (11/25)	76% (19/25)	96% (24/25)	100% (25/25)
Board/open-ice	76% (19/25)	96% (24/25)	100% (25/25)	100% (25/25)

[1] Six (6) raters completed a CHECC List evaluation and agreements are demarcated by how many of the six raters agreed on an evaluation.
[2] Twenty-five collisions were evaluated by the six raters.

Table 3.3. Reliability and precision – Strength and anthropometric measurements. Intrasession reliability ($ICC_{3,1}$) and precision (standard error of measurement) for strength and anthropometric measurements

	Reliability ($ICC_{3,1}$)	Precision (SEM)
Strength (kg)		
Anterior neck flexors	0.962	0.849
Anterolateral neck flexors	0.969	0.613
Cervical rotation	0.909	0.790
Posterolateral neck extensors	0.821	1.794
Upper trapezius	0.890	2.113
Anthropometric measurements		
Height (cm)	0.999	0.287
Mass (kg)	1.000	0
Head-neck segment length (cm)	0.782	0.667
Neck circumference (cm)	0.996	0.251
Neck medial-lateral diameter (cm)	0.948	0.301
Neck anterior-posterior diameter (cm)	0.974	0.238
Head circumference (cm)	0.969	0.335
Head medial-lateral diameter (cm)	0.977	0.097
Head anterior-posterior diameter (cm)	0.967	0.139

Table 3.4. Data analysis plan

RQ	Objective	Variables	Statistical Method
1	Test for differences in biomechanical measures of head impact severity across player position, event type, location of head impact, and between the striking player and player struck.	*Dependent*[1]: Linear acceleration Rotational acceleration HITsp *Independent (IV):* Player position Event type Location of impact Striking vs. struck	Separate two-level random intercepts general mixed linear models: Player IV
2	Test for effects of game-related exposure on biomechanical measures of head impact severity.	*Dependent*[1] *Independent:* Shifts played Period of play	Three-level random intercepts general mixed linear model: Player Shifts played Period of play
3	Test for differences in biomechanical measures of head impact severity across infraction type.	*Dependent*[1] *Independent:* Infraction type	Two-level random intercepts general mixed linear model: Player IV
4	Test for differences in biomechanical measures of head impact severity across collision type.	*Dependent*[1] *Independent:* Open-ice vs. boards Level of anticipation Relative body position	Separate two-level random intercepts general mixed linear models: Player IV
5	To determine the effects of cervical muscle strength, head-neck anthropometrics, general aerobic fitness, and player aggression on biomechanical measures of head impact severity	*Dependent*[1] *Independent:* Cervical strength Anthropometrics Aerobic fitness BPAQ[2] PIM[3]	Separate two-level random intercepts general mixed linear model: Player IV

[1] Dependent variables (DV) for all analyses will consist of linear acceleration, rotational acceleration, and Head Impact Technology severity profile (HITsp).
[2] BPAQ includes individual subscales of physical aggression, verbal aggression, anger, and hostility, as well as the total aggression score (sum of all four individual subscales).
[3] PIM refers to measure of penalties in minutes.

CHAPTER IV

RESULTS

Introduction

This chapter will include the results associated with each of the five Specific Aims presented in Chapter I. However, it should be noted that the content related to Specific Aims 3 and 4 will be the focus of Manuscript 1 (Appendix D) and Manuscript 2 (Appendix E), respectively. Over the course of the two-year study, we recruited a total of 52 youth ice hockey players (age = 14.7 ± 1.0 years; height = 172.5 ± 6.1 cm; mass = 65.5 ± 8.8 kg) (***Table 4.1***). Fifteen subjects were from our 2007-2008 Bantam-aged cohort and the remaining 37 participants were from the 2008-2009 Bantam- (N = 16) and Midget-aged (N = 21) cohorts. Three subjects who withdrew from the study during the second season are not included in our results. One subject had an illness at the start of the season (we collected head impact data on a very small number of collisions) and never returned to the team, another subject left to play for another organization out of state, and the third subject quit the team mid-season for personal reasons. We had 35 forwards and 19 defensemen represented in our sample. Four athletes played both positions over the course of the season. Data over the two playing seasons were collected during 151 games and 137 practices. During the 2007-08 season, we collected 4608 head impacts resulting in a linear acceleration greater than 10 g. During the second season (2008-09), we collected 7850 head impacts for both the Bantam and Midget cohorts combined.

Specific Aim 1

The first specific aim was designed to study the biomechanics of head impacts sustained during games and practices in Bantam (13- and 14-year-old) and Midget (15- and 16-year-old) youth ice hockey players. This specific aim served as the foundation of our work in this area, and provided important descriptive information pertaining to the nature of head impacts sustained in youth ice hockey. Data included in the analyses for the positional, event-type, and head location differences included all Bantam impacts sustained over the course of the 2007-08 playing season, in addition to all Bantam and Midget impacts we recorded during the 2008-09 ice hockey season (N = 12392). The data pertaining to the player involvement (striker vs. player struck) included all Bantam impacts sustained over the course of the 2007-08 playing season in which collisions were observable in video footage and for which a collision was assessed using the CHECC List (N = 666). The information provided below includes all omnibus statistical findings in addition to individual means and 95% confidence intervals. All post hoc differences were deemed significant at the $P < 0.05$ level, and P values associated with these post hoc comparisons are omitted from the written results for reasons of clarity. All these post hoc P values, however, are included in **Tables 4.2 to 4.4**.

Player position differences

We did not observe a statistically significant difference in head linear acceleration between defensemen and forwards ($F_{1,4} = 0.13$, $P = 0.738$). Forwards experienced mean head linear accelerations of 18.4 g (95% CI: 17.9-18.9), and defensemen experienced impacts averaging 18.3 g (95% CI: 17.4-19.2). When comparing head rotational accelerations following impacts, no statistically significant differences were observed between defensemen (1433.3 rad/s^2; 95% CI: 1317.5-1559.3) and forwards (1476.8 rad/s^2; 95% CI: 1409.7-1547.1) in our sample ($F_{1,4} = 0.99$, $P = 0.376$). Lastly, the HITsp did not differ significantly between defensemen (13.9; 95% CI: 13.4-14.4) and forwards (14.1; 95% CI: 13.9-14.3) in our sample ($F_{1,4} = 1.20$, $P = 0.336$).

Event type differences

We observed a statistically significant difference when comparing the rotational acceleration of head impacts of youth ice hockey players across games and practices ($F_{1,48} = 19.85$, $P < 0.001$). Head impacts sustained in games (1485.8 rad/s^2; 95% CI: 1420.8-1553.7) were greater than those sustained in practices (1373.8 rad/s^2; 95% CI: 1313.3-1437.1). The HITsp of impacts sustained during games (14.1; 95% CI: 13.9-14.4) were significantly greater than those recorded during practices (13.6; 95% CI: 13.3-13.9) in our sample ($F_{1,48} = 17.40$, $P < 0.001$). There were no significant differences in head linear acceleration between impacts sustained in games (18.4 g; 95% CI: 18.0-18.9) and those sustained in practices (18.3 g; 95% CI: 17.6-19.0) in our sample ($F_{1,48} = 0.40$, $P = 0.531$).

Location of head impact differences

We observed a statistically significant difference when comparing the linear acceleration of head impacts of youth ice hockey players across four different impact locations ($F_{3,145} = 37.09$, $P < 0.001$). Head impacts sustained to the top of the head (21.2 g; 95% CI: 20.0-22.4) were statistically higher than those sustained to the back (19.9 g; 95% CI: 19.3-20.7), front (17.8 g; 95% CI: 17.3-18.3), or sides (17.2 g; 95% CI: 16.9-17.6). Rotational head accelerations also differed across the four impact locations ($F_{3,145} = 38.84$, $P < 0.001$). Contrasting the results we observed for linear acceleration, head impacts sustained to the top of the head (1038.3 rad/s^2; 95% CI: 977.5-1102.9) resulted in significantly lower rotational accelerations compared with those sustained to the back (1443.8 rad/s^2; 95% CI: 1387.5-1502.4), front (1469.2 rad/s^2; 95% CI: 1400.4-1541.3), or sides (1599.2 rad/s^2; 95% CI: 1512.8-1690.6). With respect to the HITsp, we also observed significant differences between location of head impacts ($F_{3,145} = 651.27$, $P < 0.001$), such that impacts to the top of the head (8.5; 95% CI: 8.2-8.9) had a significantly lower HITsp than impacts sustained to the back (11.2; 95% CI: 10.9-11.4), front (15.9; 95% CI: 15.6-16.1), or sides (16.8; 95% CI: 16.5-17.0).

Striking player vs. Player struck

There were no significant differences in head linear accelerations between impacts sustained as a result of striking an opponent compared to those impacts sustained when players were struck by opponents ($F_{1,14} = 0.04$; $P = 0.853$). Striking players sustained mean linear accelerations of 21.3 g (95% CI: 19.7-22.9) compared to 21.4 g for impacts sustained by players struck by opponents (95% CI: 19.8-23.0). Rotational accelerations sustained by striking players (1419.5 rad/s^2; 95% CI: 1351.6-1490.7) and those struck by opponents (1452.3 rad/s^2; 95% CI: 1344.7-1568.4) were not statistically different ($F_{1,14} = 0.68$; $P = 0.423$). Lastly, the HITsp of impacts sustained by striking players (15.9; 95% CI: 15.0-16.8) and players struck (15.6; 95% CI: 14.8-16.5) were not found to be different ($F_{1,14} = 0.45$; $P = 0.513$). We further explored these findings by categorizing impacts in low, moderate, severe, and traumatic quartiles for each of the three outcome measures of linear acceleration, rotational acceleration, and HITsp. We then performed three separate random intercepts linear mixed models including our independent variables for striker/player struck, quartile variable, and the interaction between the two. We did not find a significant interaction between severity of the linear acceleration measure and whether a player struck an opponent or was struck by an opponent ($F_{3,34} = 0.99$; $P = 0.409$). Further, no significant interactions were identified between increasing severity of rotational acceleration and striking nature of a collision ($F_{3,36} = 0.21$; $P = 0.888$). Lastly, we did not observe a significant interaction between the severity of HITsp and whether a player struck or was struck during a collision ($F_{3,36} = 2.15$; $P = 0.111$). Hypothesizing that being struck by an opponent would be associated with higher magnitudes of head impact measures, we then performed a chi-square test of association between increasing magnitude of collision (using quartiles employed above) and striking nature of the collision (striking player vs. player struck). We did not find any association between magnitude of linear acceleration ($\chi^2(3) = 0.48$; $P = 0.922$), rotational acceleration ($\chi^2(3) = 1.98$; $P = 0.577$), or HITsp ($\chi^2(3) = 5.05$; $P = 0.168$), and striking nature of the collision. Further, we performed random intercepts general linear mixed models while selecting only those cases with the head impact measure in the highest tertile, and still found no differences between players who struck opponents, and those who were struck by opponents ($P > 0.05$).

Specific Aim 2

The second specific aim was designed to study the effect of game-related exposure (i.e. number of playing shifts by period) on the biomechanics of head impacts sustained during games in Bantam (13- and 14-year-old) and Midget (15- and 16-year-old) youth ice hockey players. We observed 2978 collisions over the course of the Bantam 2007-08 playing season such that we were able to identify the period in which the collision occurred, and for which we also had recorded the number of shifts the subject played per period in that contest.

The information provided below includes all omnibus statistical findings in addition to individual means and 95% confidence intervals. Since no one has evaluated the nature of playing shift data on measures of head impact severity, we explored four separate methods of accounting for playing exposure, and used these variables in separate random intercepts linear mixed models. **Table 4.5** provides the mean number of shifts played per period, and provides information regarding the covariate data employed during our analyses of game-related exposures (see section below). All post hoc differences were deemed significant at the $P < 0.05$ level, and P values associated with these post hoc comparisons are omitted from the written results for reasons of clarity. All these post hoc P values, however, are included in **Tables 4.6 to 4.8**.

Effect of playing period

Of the 2978 body collisions we observed, 30.8% (918 of 2978) occurred in the first period, 30.8% (916 of 2978) took place in the second period, and the remaining 38.4% (1144 of 2978) were sustained in the third period. We observed a statistically significant difference in head linear acceleration in impacts sustained across the playing periods in a hockey contest ($F_{2,30} = 6.50$, $P = 0.005$). Linear accelerations measured during collisions in the third period (20.7 g; 95% CI: 19.9-21.6) were significantly greater than those sustained during the first (19.5 g; 95% CI: 18.8-20.3) and second (19.4 g; 95% CI: 18.6-20.3) periods, respectively. No significant differences in rotational acceleration were observed across the three playing periods ($F_{2,30} = 0.90$, $P = 0.418$). Likewise, no differences existed in HITsp across the three playing periods ($F_{2,30} = 1.27$, $P = 0.296$).

Effect of game-related exposures

Absolute number of playing shifts: For these analyses, our models analyzed our outcome of interest while controlling for the number of shifts the subject played during the period in which the collision took place. In so doing, however, we did not observe any significant associations between the number of shifts played in a period and increases in the linear accelerations of collisions occurring in those periods ($F_{1,2955} = 0.08$; $P = 0.771$). No associations were found for rotational acceleration ($F_{1,2955} = 0.45$; $P = 0.502$) or HITsp ($F_{1,2955} = 0.56$; $P = 0.456$). Based on our tertile analyses, subjects who participated in a higher number of shifts during the period in which they sustained an impact did not experience higher linear accelerations than athletes who participated in less playing shifts ($F_{2,27} = 0.24$; $P = 0.787$). No differences were noted for rotational acceleration ($F_{2,27} = 1.17$; $P = 0.326$) or the HITsp ($F_{2,27} = 0.24$; $P = 0.789$) for absolute number of playing shifts.

Weighted summation of playing shifts: For these analyses, we included a weighted summation of playing shifts in our statistical models. These were computed as follows for the three playing periods:

Period 1: shifts in Period 1

Period 2: (1*shifts in Period 2) + (0.5*shifts in Period 1)

Period 3: (1*shifts in Period 3) + (0.5*shifts in Period 2) + (0.25*shifts in Period 1)

We observed a positive relationship such that an increase in the weighted summation of playing shifts resulted in higher linear accelerations sustained by subjects in our sample ($F_{1,2955} = 7.80$; $P = 0.005$). No relationships were found for rotational acceleration ($F_{1,2955} = 1.14$; $P = 0.286$) or HITsp ($F_{1,2955} = 1.19$; $P = 0.275$). Based on our tertile analyses, subjects who exhibited a higher weighted summation of playing shifts did not experience higher linear accelerations than athletes who had a lower weighted summation of playing shifts ($F_{2,29} = 1.03$; $P = 0.369$). No differences were noted for rotational acceleration ($F_{2,29} < 0.01$; $P = 0.999$) or the HITsp ($F_{2,29} = 0.27$; $P = 0.763$) for the weighted summation of playing shifts.

Average number of playing shifts: For these analyses, we included the mean of the number of playing shifts in our statistical models. These were computed as follows for the three playing periods:

Period 1: number of shifts in Period 1

Period 2: (shifts in Period 2 + shifts in Period 1) ÷ 2

Period 3: (shifts in Period 3 + shifts in Period 2 + shifts in Period 1) ÷ 3

We did not observe any relationships between the average number of playing shifts and increases in linear acceleration ($F_{1,2955} = 0.77$; $P = 0.379$). Likewise, no relationships were found for rotational acceleration ($F_{1,2955} = 1.56$; $P = 0.212$) or HITsp ($F_{1,2955} = 0.13$; $P = 0.719$). Based on our tertile analyses, subjects who exhibited a higher average number of playing shifts did not experience higher linear accelerations than athletes who had a lower average number of playing shifts ($F_{2,26} = 0.69$; $P = 0.511$). No differences were noted for rotational acceleration ($F_{2,26} = 0.88$; $P = 0.428$) or the HITsp ($F_{2,26} = 2.54$; $P = 0.098$) for average number of playing shifts.

Weighted average number of playing shifts: For these analyses, we included the weighted average number of playing shifts in our statistical models. These were calculated very similarly to the weighted summation of playing shifts above, with some modifications as follows:

Period 1: shifts played in Period 1

Period 2: [(1*shifts played in Period 2) + (0.5*shifts played in Period 1)] ÷ 1.5

Period 3: [(1*shifts Period 3) + (0.5*shifts Period 2) + (0.25*shifts Period 1)] ÷ 1.75

We did not observe any relationships between the weighted average number of playing shifts and increases in linear acceleration ($F_{1,2955} = 0.44$; $P = 0.508$). Similarly, no relationships were found for rotational acceleration ($F_{1,2955} = 1.05$; $P = 0.306$) or HITsp ($F_{1,2955} = 0.01$; $P = 0.942$). Based on our tertile analyses, subjects who exhibited a higher weighted average number of playing shifts did not experience higher linear accelerations than athletes who had a lower weighted average number of playing shifts ($F_{2,27} = 0.23$; $P = 0.798$). No differences were noted for rotational acceleration ($F_{2,27} = 0.65$; $P = 0.528$) or the HITsp ($F_{2,27} = 0.07$; $P = 0.935$) for weighted average number of playing shifts.

Interactions between playing period and the number of playing shift exposures

In an attempt to further identify the effects of playing shift exposures, we endeavored to study the interactions between the playing period in which the collision took place and the number of playing shifts (as measured by the four different methods identified in the previous subsections of Specific Aim 2) on the severity of head impacts sustained by our subjects. First, we did not observe a significant interaction between playing period and the absolute number of playing shifts on measures of linear acceleration ($F_{2,2951} = 1.34$; $P = 0.262$), rotational acceleration ($F_{2,2951} = 0.54$; $P = 0.584$), or the HITsp ($F_{2,2951} = 1.56$; $P = 0.211$). Secondly, we did not observe a significant interaction between playing period and the weighted summation of playing shifts on measures of linear acceleration ($F_{2,2951} = 1.21$; $P = 0.297$), rotational acceleration ($F_{2,2951} = 0.35$; $P = 0.702$), or the HITsp ($F_{2,2951} = 1.86$; $P = 0.156$). Next, we did not observe a significant interaction between playing period and the average number of playing shifts on measures of linear acceleration

($F_{2,2951} = 2.18$; $P = 0.113$), rotational acceleration ($F_{2,2951} = 0.88$; $P = 0.417$), or the HITsp ($F_{2,2951} = 2.54$; $P = 0.079$). Lastly, we did not observe a significant interaction between playing period and the weighted average number of playing shifts on measures of linear acceleration ($F_{2,2951} = 1.74$; $P = 0.176$), rotational acceleration ($F_{2,2951} = 0.69$; $P = 0.499$), or the HITsp ($F_{2,2951} = 2.13$; $P = 0.119$).

Specific Aim 3

The focus of *Manuscript 2* (Appendix E), Specific Aim 3 was designed to address the following research question: Is there an association between biomechanical measures of head impact severity sustained by youth ice hockey players and infraction type at the time of the collision? In addition to a comparison between legal and illegal collisions, we also sought to answer a secondary question that included different infraction types including boarding or charging, checking an opponent from behind, and elbowing an opponent or deliberately making head contact. Data in these analyses included all Bantam impacts sustained over the course of the 2007-08 playing season in which collisions were observable in video footage and for which a collision was assessed using the CHECC List (N = 665). The information provided below includes all omnibus statistical findings in addition to individual means and 95% confidence intervals. All post hoc differences were deemed significant at the $P < 0.05$ level (unless otherwise specified), and are omitted from the written results for reasons of clarity. All these post hoc P values, however, are included in **Tables 4.9 to 4.11.** In addition to understanding the effect of illegal infractions on measures associated with head impact severity, we also sought to identify how infraction types affect those players who are struck by the offending players.

Legal vs. illegal collisions

We observed a total of 665 body collisions for which we were able to complete a CHECC List and assign a level of infraction. Of these collisions, 82.7% (550 of 665) were deemed to be legal body collisions, while the remaining 17.3% (115 of 665) were deemed to be illegal in nature. The specific types of illegal infractions were analyzed and are presented below. Generally speaking, we observed a statistically significant difference in head linear acceleration in impacts sustained from legal collisions compared to those sustained from illegal infractions ($F_{1,13} = 8.46$, $P = 0.012$). Linear accelerations measured during collisions involving illegal infractions (23.0 g; 95% CI: 21.4-24.8) were significantly greater than those sustained during legal collisions (21.0 g; 95% CI: 19.5-22.5). The HITsp measures for illegal infractions (16.8; 95% CI: 15.8-17.9) were significantly greater than those we observed for legal collisions (15.5; 95% CI: 14.7-16.4) in our sample ($F_{1,13} = 6.86$; $P = 0.021$). No significant differences between illegal infractions (1529.9 rad/s^2; 95% CI: 1388.5-1685.8) and legal collisions (1417.5 rad/s^2; 95% CI: 1334.8-1505.3) were observed for measures of rotational acceleration ($F_{1,13} = 2.45$; $P = 0.142$).

Types of illegal infractions

In an effort to better understand head impact biomechanical measures resulting from different types of infractions, we further distinguished the illegal infractions into the following: boarding or charging, checking an opponent from behind, and elbowing an opponent or deliberately making head contact (with their body or playing stick). Of all impacts evaluated using the CHECC List, 82.7% (550 of 665) were legal body collisions, 3.0% (20 of 665) were boarding or charging infractions, 2.9% (19 of 665) were a result of a check from behind, and 11.4% (76 of 665) were a result of elbowing, intentional head contact, or high sticking to the head.

We found a statistically significant difference in head linear acceleration in impacts sustained from legal collisions compared to those sustained from illegal infractions ($F_{3,28}$ = 4.36, P = 0.012). Linear head accelerations due to elbowing, intentional head contact, or high sticking to the head (24.0 g; 95% CI: 21.9-26.2) were significantly greater than those observed in legal collisions (21.0 g; 95% CI: 19.5-22.5). There were no differences between legal collisions, those sustained from boarding or charging (21.2 g; 95% CI: 18.7-24.0), and checking from behind (21.4 g; 95% CI: 18.8-24.3). Rotational head accelerations differed across legal collisions and infraction types ($F_{3,28}$ = 3.53, P = 0.028). Impacts involving elbowing, head contact, or high sticking infractions (1614.3 rad/s^2; 95% CI: 1419.6-1835.7) were significantly greater than those observed for legal collisions (1418.4 rad/s^2; 95% CI: 1335.4-1506.5). Though not statistically significant, we observed a trend in the data to suggest differences between boarding or charging infractions (1575.9 rad/s^2; 95% CI: 1419.2-1749.9), and legal collisions (P = 0.103). Checking from behind (1197.7 rad/s^2; 95% CI: 953.3-1504.9) did not result in any significant differences in head rotational acceleration compared to legal collisions. With respect to the HITsp, the data are suggestive of a significant difference between legal collisions and the different infraction types ($F_{3,28}$ = 2.78; P = 0.059). Further exploring this finding, we observed impacts resulting from elbowing, head contact, or high sticking infractions (17.6; 95% CI: 16.0-19.2) to exhibit higher severity profiles than legal collisions (15.5; 95% CI: 14.7-16.4) (P = 0.010). No significant differences were observed between legal collisions and those sustained as a resulting from boarding or charging infractions (16.8; 95% CI: 14.6-19.3) and those as a result of checking from behind (14.12; 95% CI: 12.0-16.7).

Interactions between infraction types on striking and struck players

While our original interest was to determine whether overall differences existed in biomechanical measures of head impact severity between legal collisions and illegal infractions, we acknowledge the importance of understanding the effects of infraction types in the context of whether a player was delivering a body check or was the recipient of a collision. In so doing, we were testing the hypothesis that players who were struck as a result of illegal infractions would experience more pronounced measures of head impact severity compared to those who instigated the infraction by striking an opponent. We observed a significant interaction between infraction type and whether a player was striking an opponent or was struck by an opponent on measures of rotational acceleration ($F_{3,15} = 4.81$; $P = 0.015$). Surprisingly, players who were checked from behind sustained lower rotational head accelerations (1151.6 rad/s^2; 95% CI: 910.0-1457.3) than those who struck opponents from behind (1395.9 rad/s^2; 95% CI: 1200.5-1623.2). No differences were observed between players who were struck as a result of a boarding or charging infraction (1785.2 rad/s^2; 95% CI: 1480.6-2152.5) and those players who boarded or charged opponents (1609.2 rad/s^2; 95% CI: 1450.4-1785.4) ($P = 0.083$). Elbowing, head contact, or high sticking infractions between players who were struck (1738.0 rad/s^2; 95% CI: 1514.4-1994.6) and those who delivered the collisions (1592.0 rad/s^2; 95% CI: 1392.5-1820.1) were not different from each other ($P = 0.116$). We did not observe any interaction effects between infraction type and whether a player delivered or received an illegal infraction at the time of the body collision for linear acceleration ($F_{3,15} = 0.67$; $P = 0.583$) or the HITsp ($F_{3,15} = 1.07$; $P = 0.391$). The results related to the main effects of infraction type and striking player/player struck in the interaction models did not differ from analyses for these variables previously reported in this chapter.

Specific Aim 4

The focus of *Manuscript 1* (Appendix D), Specific Aim 4 was designed to address the following research question: Is there an effect of body collision type on biomechanical measures of head impact severity sustained by Bantam-aged ice hockey players? In addition to a comparison between anticipated and unanticipated collisions, we also sought to answer a secondary question that included an evaluation of head impact severity between impacts occurring along the playing boards and those occurring in the open ice. In anticipated collisions, we also sought to identify what relative body positions might be most effective in mitigating the severity of head impacts sustained by youth ice hockey players. Data in these analyses included all Bantam impacts sustained over the course of the 2007-08 playing season in which collisions were observable in video footage and for which a collision was assessed using the CHECC List (N = 666). The information provided below includes all omnibus statistical findings in addition to individual means and 95% confidence intervals. All post hoc differences were deemed significant at the $P < 0.05$ level (unless otherwise specified), and are omitted from the written results for reasons of clarity. All these post hoc P values, however, are included in **Tables 4.12 to 4.14**.

Collisions along the boards vs. open-ice collisions

We observed a total of 666 body collisions for which we were able to complete a CHECC List and evaluate whether the collision took place along the boards or in the open ice. Of these collisions, 63.3% (421 of 666) took place along the playing boards, while the remaining 36.8% (245 of 666) occurred in the open ice (percentages add up to 100.01% due to rounding). We observed a statistically significant difference in head linear acceleration in impacts sustained along the playing boards compared to those sustained in the open ice ($F_{1,14} = 5.40$, $P = 0.036$). Linear accelerations sustained from open-ice collisions (22.4 g; 95% CI: 20.6-24.3) were significantly greater than those sustained from collisions along the playing boards (20.7 g; 95% CI: 19.4-22.2). The rotational acceleration measures for open-ice collisions (1564.7 rad/s^2; 95% CI: 1440.3-1699.9) were significantly greater than those we observed for collisions along the playing boards (1367.7 rad/s^2; 95% CI: 1295.6-1443.9) in our sample ($F_{1,14} = 12.75$; $P = 0.003$). With respect to the HITsp, the data suggests a very strong trend towards a significant difference between open-ice collisions and those occurring along the playing boards ($F_{1,14} = 4.38$; $P = 0.055$). Further exploring this finding, we observed impacts during open-ice collisions (16.3; 95% CI: 15.3-17.3) to have a greater HITsp than impacts occurring along the boards (15.5; 95% CI: 14.7-16.3).

Level of anticipation

We observed a total of 666 body collisions for which we were able to complete a CHECC List and evaluate whether the collision was anticipated or unanticipated. Of these collisions, 84.7% (564 of 666) were anticipated while the remaining 15.3% (102 of 666) were deemed to be unanticipated collisions. Though linear accelerations tended to be greater in unanticipated collisions (22.6 g; 95% CI: 20.9-24.5) compared to anticipated collisions (21.1 g; 95% CI: 19.5-22.8), the differences we observed were not statistically significant ($F_{1,14} = 2.52$; $P = 0.135$). A similar trend in the data was observed such that anticipated collisions (1414.3 rad/s^2; 95% CI: 1330.6-1503.3) resulted in lower rotational head accelerations than unanticipated collisions (1550.0 rad/s^2; 95% CI: 1377.4-1744.2). These differences were not statistically significant ($F_{1,14}$ = 2.47; P = 0.138). No significant differences in the HITsp between anticipated and unanticipated collisions were observed ($F_{1,14}$ = 0.10; P = 0.755).

The previous analyses did not take into account the level of anticipation; that is, anticipated collisions were categorized as "anticipated" regardless of whether we deemed an athlete to be in an optimal relative body position for the impending collision or not. For the ensuing analyses, these same collisions were further subcategorized in an "overall impression" variable consisting of the three following levels: anticipated collision (with a good relative body position), anticipated collision (with a poor relative body position), and unanticipated collision. Of these collisions, 47.3% (315 of 666) were anticipated with a good relative body position, 37.4% (249 of 666) were anticipated with a poor relative body position, while the remaining 15.3% (102 of 666) were deemed to be unanticipated collisions. While we observed an increasing trend in the linear accelerations of head impacts sustained during collisions that were anticipated and where the player was in a good position to deliver or sustain the impact (20.7 g; 95% CI: 19.1-22.5), impacts that were anticipated in which the player was not in a good position (21.4 g; 95% CI: 19.6-23.4), and unanticipated collisions (22.6 g; 95% CI: 20.9-24.5), these differences were not statistically significant ($F_{2,28}$ = 1.46, P = 0.249) with a 2,28 degree of freedom mixed model. Since the data suggested the trend just described, we subsequently performed a 1 degree-of-freedom linear trend, observing suggestive evidence of a trend in our data ($F_{1,649}$ = 2.55; P = 0.111). No significant differences in rotational head accelerations ($F_{2,28}$ = 1.24, P = 0.304) or HITsp ($F_{2,28}$ = 0.70, P = 0.503) across anticipation type were observed. These analyses were repeated while including BMI as a covariate and no changes in the results presented above were observed.

Additional analyses

In an attempt to better understand the effect of high-magnitude collisions on collision type and level of anticipation for our three biomechanical measures of head impact severity, we performed additional analyses aimed at determining if impact magnitudes for select comparisons (anticipated vs. unanticipated; open-ice vs. along playing boards) were more disparate in the high-end impact range. Using linear acceleration as our criterion variable, we identified the 75th, 90th, and 95th percentiles. Using each percentile range as a predetermined cutoff value, we performed separate random intercepts general mixed linear models for each of our dependent measures. We repeated this procedure using HITsp as our criterion variable, also identifying the 75th, 90th, and 95th percentiles for this value. We performed separate random intercepts general mixed linear models for our dependent measures. Regardless of our cutoff value, the results did not yield any significant differences in linear acceleration, rotational acceleration, and HITsp, between open-ice collisions and those occurring along the playing boards ($P > 0.05$). We also did not observe any significant differences in linear acceleration, rotational acceleration, and HITsp, between anticipated (good), anticipated (poor), and unanticipated collisions ($P > 0.05$).

Additionally, we explored a number of different impact ranges, particularly those between the 25th to 75th percentile of linear acceleration measures, as well as those between the 50th and 75th percentiles. For the former, we observed a statistically significant difference in linear acceleration ($F_{2,27} = 4.29$; $P = 0.024$), such that anticipated—good collisions (18.7 g; 95% CI: 18.0-19.4) were significantly lower than unanticipated (19.9 g;

95% CI: 19.1-20.7) body collisions (P = 0.007). We evaluated those collisions occurring between the 25th to 75th percentile of HITsp measures, as well as those between the 50th and 75th percentiles. For the latter, we observed a significant difference in rotational acceleration ($F_{2,19}$ = 6.83; P = 0.006), such that impacts from anticipated—good (1215.11 rad/s^2; 95% CI: 1112.6-1327.1) and anticipated—poor (1218.9 rad/s^2; 95% CI: 1107.2-1341.9) collisions were significantly lower than unanticipated collisions (1465.7 rad/s^2; 95% CI: 1240.7-1731.4). We also observed a significant difference in HITsp ($F_{2,19}$ = 4.35; P = 0.028), such that impacts from anticipated—good (15.2; 95% CI: 15.0-15.5) and anticipated—poor (15.3; 95% CI: 15.1-15.5) collisions were significantly lower than unanticipated collisions (15.6; 95% CI: 15.3-15.9). All other analyses did not yield any statistically significant findings (P > 0.05).

Relative body positioning

We performed several analyses comparing the linear accelerations, rotational accelerations, and the HITsp across a number of characteristics we used to describe a player's relative body position at the time of a body collision. These included whether or not a player was looking ahead in the direction of movement, whether they appeared to be looking in the direction of the impending collision, whether the athlete's knees were flexed, and many others (please refer to the CHECC List in Appendix B). For clarity of presentation, subsections delineating each of the eleven separate body collision descriptors can be found below. Analyses pertaining to the remaining four descriptors including overall impression of body collision, player involvement in body collision (striker vs. player struck), infraction type associated with collision, and location of body collision (open-ice vs. along playing boards), have previously been addressed in this chapter. With the exception of a single statistically significant finding (linear acceleration) and a statistical trend (HITsp) with respect to using one's legs to drive into or through a body collision, all other findings were not statistically significant ($P > 0.05$). The detailed statistical findings of these analyses, however, are provided below and can also be found in **Tables 4.15 to 4.17**.

Player looking ahead in direction of movement: No significant differences in linear head acceleration were observed between collisions where the player was looking ahead in the direction of movement and those in which they were not ($F_{1,9} = 0.82$; $P = 0.388$). No differences were noted for rotational acceleration ($F_{1,9} = 0.04$; $P = 0.844$) or the HITsp ($F_{1,9} = 0.04$; $P = 0.845$) for this body collision descriptor.

Player appears to be looking in direction of impending body collision: No significant differences in linear head acceleration were observed between collisions where the player was looking in the direction of the impending body collision and those in which they were not ($F_{1,14} = 0.21$; $P = 0.658$). No differences were noted for rotational acceleration ($F_{1,14} = 0.07$; $P = 0.794$) or the HITsp ($F_{1,14} = 2.74$; $P = 0.120$) for this body collision descriptor.

Knee flexion greater than 30 degrees at the time of body collision: No significant differences in linear head acceleration were observed between collisions where the player was in a knee-flexed position compared to those collisions in which the athlete appeared to be in a more knee-extended position ($F_{1,14} < 0.01$; $P = 0.996$). Again, no differences were noted for rotational acceleration ($F_{1,14} = 0.14$; $P = 0.715$) or the HITsp ($F_{1,14} = 0.04$; $P = 0.846$) for relative knee flexion.

Trunk flexion at time of body collision: Athletes positioned in trunk flexion did not experience lesser linear acceleration than those athletes whose trunks were in a more upright position at the time of collision ($F_{1,14} = 0.23$; $P = 0.639$). No differences were noted for rotational acceleration ($F_{1,14} = 0.09$; $P = 0.771$) or the HITsp ($F_{1,14} < 0.01$; $P = 0.961$) for relative trunk flexion.

Player drives into collision with shoulders: Athletes who drive into a body collision with their shoulders do not experience lower linear accelerations than athletes who fail to do so ($F_{1,14} = 1.05$; $P = 0.323$). No differences were noted for rotational acceleration ($F_{1,14} = 0.58$; $P = 0.460$) or the HITsp ($F_{1,14} = 0.26$; $P = 0.619$) for the use of shoulders during a body collision.

Player uses elbow(s) in body collision: Athletes who drive into a body collision with their elbows, regardless of whether this act would be deemed an elbowing infraction, do not experience lower linear accelerations than athletes who restrain from using their elbows during a collision ($F_{1,12} = 0.02$; $P = 0.901$). No differences were noted for rotational acceleration ($F_{1,12} = 0.42$; $P = 0.531$) or the HITsp ($F_{1,12} = 0.03$; $P = 0.869$) for the use of elbows during a body collision.

Player uses hands in body collision: Youth ice hockey players who drive into a body collision with their hands do not experience lower linear accelerations than athletes who restrain from using their hands during a collision ($F_{1,12} = 0.51$; $P = 0.488$). No differences were noted for rotational acceleration ($F_{1,12} = 0.43$; $P = 0.525$) or the HITsp ($F_{1,12} = 1.93$; $P = 0.190$) for the use of hands during a body collision.

Feet are shoulder width apart at the time of the body collision: Body collisions in which the young player's feet are shoulder width apart do not demonstrate lower linear accelerations than collisions in which this is not the case ($F_{1,14} = 0.21$; $P = 0.656$). Further, no differences were noted for rotational acceleration ($F_{1,14} = 0.02$; $P = 0.877$) or the HITsp ($F_{1,14} < 0.01$; $P = 0.986$) for the use of hands during a body collision.

Player uses stick during collision: Athletes who use their stick during a body collision, regardless of whether this act would be deemed a high-sticking infraction, do not experience lower linear accelerations than athletes who restrain from using their sticks during a collision ($F_{1,9} = 0.28$; $P = 0.608$). No differences were noted for rotational acceleration ($F_{1,9} = 0.04$; $P = 0.841$) or the HITsp ($F_{1,9} = 0.03$; $P = 0.873$) for the use of sticks during a body collision.

Player uses legs to drive into or through a body collision: Athletes who drive into or through a body collision with their legs (20.5 g; 95% CI: 19.2-21.9) experience lower linear accelerations than athletes who do not use their legs (21.7 g; 95% CI: 20.1-23.5) during a collision ($F_{1,13} = 4.67$; $P = 0.049$). No differences were noted for rotational acceleration ($F_{1,13} = 0.62$; $P = 0.446$). A moderate trend observed for the HITsp ($F_{1,13} = 3.47$; $P = 0.085$) suggests that athletes who use their legs to drive through a body collision (15.3; 95% CI: 14.6-16.1) experience lower severity profiles than those instances in which athletes do not use their legs to drive through a collision (16.0; 95% CI: 15.1-16.9).

Player delivering or receiving a pass or shot at the time of body collision: Athletes who are passing or shooting the puck at the time of the collision do not experience lower linear accelerations than athletes who have already delivered the puck prior to the body collision ($F_{1,14} < 0.01$; $P = 0.956$). No differences were noted for rotational acceleration ($F_{1,14} = 0.61$; $P = 0.446$) or the HITsp ($F_{1,14} = 0.56$; $P = 0.466$) for this body collision descriptor.

Specific Aim 5

Specific Aim 5 was designed to evaluate the effect of relative cervical muscle strength, cervical and head anthropometrics, general aerobic fitness, and player aggression, on biomechanical measures of head impact severity during games in Bantam (13- and 14-year-old) and Midget (15- and 16-year-old) youth ice hockey players. This specific aim addressed the following research question: Are cervical muscle strength, cervical and head anthropometrics, general aerobic fitness, and player aggression, associated with the biomechanical measures of head impact severity sustained by youth ice hockey players? Data in these analyses included all Bantam and Midget impacts (N = 7718) sustained over the course of the 2008-09 playing season for which these anthropometric, strength, aerobic fitness, and aggression measures were collected for the players in our sample. The information provided below includes all omnibus statistical findings in addition to individual means and 95% confidence intervals. All post hoc differences were deemed significant at the $P < 0.05$ level (unless otherwise specified), and are omitted from the written results for reasons of clarity. All these post hoc P values, however, are included in tables as identified in the following subsections.

Cervical muscle strength

We performed several analyses comparing the linear accelerations, rotational accelerations, and the HITsp across a number of cervical muscle strength measurements we collected across our sample. These included anterior neck strength, anterolateral neck strength, cervical rotation strength, posterolateral neck strength, and upper trapezius strength. For clarity of presentation, subsections delineating each of the five cervical muscle strength measures can be found below. With the exception of a single significant finding (HITsp) for upper trapezius muscle strength, and two statistical trends (for rotational acceleration) with respect to anterior neck strength and posterolateral neck strength measures, all other findings were not statistically significant ($P > 0.05$). More detailed descriptions of these statistical findings, however, are provided below and are also included in **Tables 4.18 to 4.20**.

Anterior neck strength: Though not statistically significant ($F_{2,29} = 3.10$; $P = 0.060$), we observed rotational accelerations in athletes with the weakest anterior neck muscle strength (1642.9 rad/s^2; 95% CI: 1530.6-1763.4) to be higher than rotational accelerations in the moderate (1482.6 rad/s^2; 95% CI: 1409.0-1560.1) and strong (1581.4 rad/s^2; 95% CI: 1453.3-1720.7) tertiles. No significant differences between the strength groups were observed for linear acceleration ($F_{2,29} = 0.95$; $P = 0.399$) and the HITsp ($F_{2,29} = 0.03$; $P = 0.969$).

Anterolateral neck strength: Athletes who exhibit stronger anterolateral neck muscles do not experience lower linear accelerations than athletes who possess moderate to low anterolateral neck muscle strength ($F_{2,29} = 0.01$; $P = 0.987$). No differences were noted for rotational acceleration ($F_{2,29} = 0.56$; $P = 0.579$) or the HITsp ($F_{2,29} = 0.81$; $P = 0.456$) for anterolateral neck muscle strength measures.

Cervical rotation strength: Athletes who exhibit stronger cervical rotation muscles do not experience lower linear accelerations than athletes who possess moderate to low cervical rotation muscle strength ($F_{2,29} = 2.14$; $P = 0.136$). No differences were noted for rotational acceleration ($F_{2,29} = 1.51$; $P = 0.238$) or the HITsp ($F_{2,29} = 0.98$; $P = 0.389$) for cervical rotation strength measures.

Posterolateral neck strength: A trend observed for rotational acceleration ($F_{2,29} = 2.49$; $P = 0.101$) suggests that athletes with the weakest posterolateral neck muscle strength (1631.2 rad/s^2; 95% CI: 1535.4-1733.0) experienced higher rotational accelerations than those in the moderate (1480.1 rad/s^2; 95% CI: 1383.6-1583.4) and strong (1591.4 rad/s^2; 95% CI: 1464.1-1729.7) tertiles. No significant differences between the strength groups were observed for linear acceleration ($F_{2,29} = 0.13$; $P = 0.883$) and the HITsp ($F_{2,29} = 0.60$; $P = 0.556$).

Upper trapezius muscle strength: There was a significant difference in the HITsp in athletes across three tertiles of upper trapezius muscle strength ($F_{2,29} = 3.71$; $P = 0.037$). Athletes with the strongest upper trapezius muscle strength (14.4; 95% CI: 14.0-14.8) experienced higher HITsp measures than athletes with moderate (14.0; 95% CI: 13.5-14.4) or low (13.6; 95% CI: 13.2-14.0) upper trapezius strength. No differences were noted for linear acceleration ($F_{2,29} = 0.11$; $P = 0.892$) or rotational acceleration ($F_{2,29} = 0.38$; $P = 0.689$) for upper trapezius strength measures.

Additional analyses for cervical muscle strength: We performed three additional random intercepts general linear mixed models (one for each dependent variable) while including all five measures of cervical muscle strength in the same statistical model. While adjusting for the other four measures of neck strength, cervical rotation strength appeared to significantly affect linear acceleration ($F_{2,21} = 8.79$; $P = 0.002$). Those athletes who had the weakest cervical rotators experienced higher linear accelerations (18.6 g; 95% CI: 17.9-19.3) than those who had the strongest cervical rotation strength (17.4 g; 95% CI: 16.9-17.9) ($P < 0.001$). Participants with moderate cervical rotation strength (17.5 g; 95% CI: 17.0-18.1) also experienced higher linear accelerations than those with the strongest cervical rotation strength ($P = 0.018$). While adjusting for the other four measures of neck strength, athletes with the greatest upper trapezius muscle strength (14.4; 95% CI: 13.9-14.8) experienced higher HITsp measures than athletes with moderate (13.9; 95% CI: 13.6-14.3; $P = 0.034$) or low (13.5; 95% CI: 13.1-14.0; $P = 0.003$) upper trapezius strength ($F_{2,21} = 6.62$; $P = 0.006$). We did not observe any changes in our findings of rotational acceleration that differed from what we have presented based on our individual analyses involving only one muscle group at a time.

Cervical, head, and player anthropometrics

We performed several analyses comparing the linear accelerations, rotational accelerations, and the HITsp across a number of cervical and head anthropometric measures we collected across our sample. These included head-neck segment length, head and neck circumference, head and neck medial-lateral diameter, and head and neck anterior-posterior diameter. We also recorded player height and mass as additional anthropometric measurements. For clarity of presentation, subsections delineating each of the nine body, cervical, and head anthropometric measures can be found below. The means, 95% confidence intervals, and associated *P* values for our anthropometric data can be found in **Tables 4.21 to 4.23**.

Player height: Taller athletes do not experience lower linear accelerations than athletes who are of medium height or those who represent the shortest players in our sample ($F_{2,29} = 0.41$; $P = 0.665$). No differences were noted for rotational acceleration ($F_{2,29} = 0.34$; $P = 0.716$) or the HITsp ($F_{2,29} = 0.39$; $P = 0.680$) for player height measures.

Player mass: A significant difference in rotational acceleration ($F_{2,29} = 6.80$; $P = 0.004$) suggests that the heaviest athletes in our sample (1675.8 rad/s^2; 95% CI: 1574.6-1783.4) experienced higher rotational accelerations than the lightest athletes (1467.3 rad/s^2; 95% CI: 1408.5-1528.5). No significant differences were observed for linear acceleration

($F_{2,29} = 0.37$; $P = 0.692$) and the HITsp ($F_{2,29} = 2.17$; $P = 0.133$) between the heaviest, medium, and lightest, players in our sample.

Head-neck segment length: Athletes with the longest head-neck segment length do not experience lower linear accelerations than athletes who have shorter head-neck segment lengths in our sample ($F_{2,29} = 0.52$; $P = 0.601$). No differences were noted for rotational acceleration ($F_{2,29} = 0.02$; $P = 0.982$) or the HITsp ($F_{2,29} = 0.82$; $P = 0.450$) for head-neck segment length measures.

Neck circumference: Athletes with the greatest neck girth do not experience lower linear accelerations than athletes who have smaller neck girths in our sample ($F_{2,29} = 0.82$; $P = 0.452$). No differences were noted for rotational acceleration ($F_{2,29} = 1.73$; $P = 0.195$) or the HITsp ($F_{2,29} = 0.60$; $P = 0.555$) for neck circumferential measures.

Neck medial-lateral diameter: Athletes with the widest neck medial-lateral diameter do not experience lower linear accelerations than athletes who have smaller neck medial-lateral diameters in our sample ($F_{2,29} = 1.85$; $P = 0.175$). No differences were noted for rotational acceleration ($F_{2,29} = 0.37$; $P = 0.697$) or the HITsp ($F_{2,29} = 0.35$; $P = 0.706$) for neck medial-lateral diameter measures.

Neck anterior-posterior diameter: A significant difference in rotational acceleration ($F_{2,29} = 3.53$; $P = 0.043$) suggests athletes with wider neck anterior-posterior diameters in our sample (1695.7 rad/s^2; 95% CI: 1572.9-1828.2) experienced higher rotational accelerations than athletes with moderate neck anterior-posterior diameters (1530.8 rad/s^2; 95% CI: 1417.0-1653.7) and those with the smallest neck anterior-posterior diameters (1506.7 rad/s^2; 95% CI: 1425.7-1592.3). A trend observed for linear acceleration ($F_{2,29} = 2.55$; $P = 0.095$) suggests that athletes with the smallest neck anterior-posterior diameters (17.2 g; 95% CI:

16.8-17.7) experienced lower linear accelerations than those in the moderate (17.4 g; 95% CI: 17.0-17.8) and widest (18.2 g; 95% CI: 17.4-18.9) tertiles. No significant differences were observed for the HITsp ($F_{2,29} = 2.17$; $P = 0.133$) between neck anterior-posterior neck sizes across players in our sample.

Head circumference: A significant difference in rotational acceleration ($F_{2,29} = 20.77$; $P < 0.001$) suggests athletes with larger head circumferences in our sample (1722.7 rad/s^2; 95% CI: 1612.9-1840.0) experienced higher rotational accelerations than athletes with the smallest head circumferences (1409.4 rad/s^2; 95% CI: 1355.5-1465.4). A significant difference was also observed with respect to the HITsp ($F_{2,29} = 9.60$; $P < 0.001$) suggesting athletes with larger head circumferences in our sample (14.2; 95% CI: 13.8-14.6) experienced a higher HITsp than athletes with the smallest head circumferences (13.4; 95% CI: 13.0-13.8). No significant differences were observed for linear acceleration ($F_{2,29} = 0.01$; $P = 0.990$) between head circumferential sizes across players in our sample.

Head medial-lateral diameter: A significant difference in rotational acceleration ($F_{2,29} = 5.00$; $P = 0.014$) suggests athletes with wider head medial-lateral diameters in our sample (1714.6 rad/s^2; 95% CI: 1582.7-1857.5) experienced higher rotational accelerations than athletes with more narrow head medial-lateral diameters (1478.5 rad/s^2; 95% CI: 1402.6-1558.5). A trend in our model ($P = 0.059$) suggested those with moderate head medial-lateral diameters (1552.6 rad/s^2; 95% CI: 1454.8-1657.0) also experienced lower rotational accelerations than those with the widest head medial-lateral diameters. No differences were noted for linear acceleration ($F_{2,29} = 1.47$; $P = 0.248$) or the HITsp ($F_{2,29} = 1.24$; $P = 0.305$) for head medial-lateral diameter measures.

Head anterior-posterior diameter: A significant difference in the HITsp ($F_{2,29} = 3.59$; $P = 0.041$) suggests athletes with wider head anterior-posterior diameters in our sample (14.2; 95% CI: 13.8-14.6) experienced higher HITsp than athletes with more narrow head anterior-posterior diameters (13.4; 95% CI: 13.0-14.0). No significant differences were observed for linear ($F_{2,29} = 0.16$; $P = 0.851$) or rotational ($F_{2,29} = 2.30$; $P = 0.118$) accelerations between head anterior-posterior diameter measures across players in our sample.

Additional analyses for player head and neck anthropometrics: We performed three additional random intercepts general linear mixed models (one for each dependent variable) while including all seven measures of head and neck anthropometrics in the same statistical model. In so doing, we observed significant differences not recognized in our original analyses. After adjusting for the other anthropometric measures, we observed significant differences in linear acceleration for head-neck segment length ($F_{2,17} = 7.16$; $P = 0.006$), neck medial-lateral diameter ($F_{2,17} = 4.55$; $P = 0.026$), neck anterior-posterior diameter ($F_{2,17} = 6.44$; $P = 0.008$), and head medial-lateral diameter ($F_{2,17} = 5.18$; $P = 0.018$). In all cases, those athletes representing the smallest in terms of the respective anthropometric measurement experienced significantly lower linear accelerations than those with the largest measurements ($P < 0.05$).

While adjusting for all other anthropometric measures, we observed that athletes with the shortest (or more narrow) anthropometric measurements experienced significantly lower rotational accelerations than those with the longest (or widest) anthropometric measurements for the following: head-neck segment length ($F_{2,17} = 4.34$; $P = 0.030$), neck circumference ($F_{2,17} = 4.78$; $P = 0.023$), neck medial-lateral diameter ($F_{2,17} = 10.57$; $P = 0.001$), neck anterior-posterior diameter ($F_{2,17} = 6.64$; $P = 0.007$), head circumference ($F_{2,17} = 43.61$; $P < 0.001$), and head medial-lateral diameter ($F_{2,17} = 12.48$; $P = 0.001$). There were no differences from our original analyses for HITsp when all anthropometric measurements were included in the same model.

General aerobic fitness

We performed several analyses comparing the linear accelerations, rotational accelerations, and the HITsp for two measures of general aerobic we collected across our sample. These included the number of lengths completed during the FAST, as well as the gender-, height- and mass-predicted maximal aerobic power as an estimate of volume of maximal oxygen consumption (VO_2max). For clarity of presentation, subsections delineating these two measures of general aerobic fitness can be found below. The means, 95% confidence intervals, and associated *P* values for our general aerobic fitness data can be found in **Tables 4.24 to 4.26**.

Laps attained during the Faught Aerobic Skating Test: A significant difference in linear acceleration was observed ($F_{2,29} = 3.93$; $P = 0.031$) suggesting athletes who were among those attaining the highest levels of the FAST in our sample (18.0 g; 95% CI: 17.4-18.6) experienced higher linear accelerations than athletes who performed the poorest in this aerobic task (17.1 g; 95% CI: 16.7-17.5). A significant difference in rotational acceleration ($F_{2,29} = 3.46$; $P = 0.045$) was also observed suggesting athletes attaining the highest levels of the FAST (1678.6 rad/s^2; 95% CI: 1573.7-1790.4) experienced higher rotational accelerations than athletes who performed in the lowest tertile on the aerobic test (1497.8 rad/s^2; 95% CI: 1406.9-1594.5). No significant differences were observed for the HITsp ($F_{2,29} = 0.52$; $P = 0.600$) between the FAST performance levels in our sample.

Maximal oxygen consumption (VO_2max): While the actual number of laps attained on the FAST provides a quick evaluation for ice hockey coaches at the time of testing, it fails to account for player size (height and mass), gender, and other factors that may mitigate a lower score in one player relative to another. We used the number of laps a player attained in the FAST, in addition to their mass, height, and gender and introduced it into a predetermined prediction model to estimate VO_2max (Petrella, et al., 2007). In so doing, we were able to evaluate the effects of general aerobic fitness while accounting for other mitigating anthropometric measures including player height and mass. A significant difference in linear acceleration was observed ($F_{2,29} = 8.51$; $P < 0.001$) suggesting athletes who were among the most aerobically fit in our sample (18.0 g; 95% CI: 17.7-18.2) experienced higher linear accelerations than those who represented the least aerobic fitness in our sample (17.0 g; 95% CI: 16.6-17.4). No significant differences were observed for rotational acceleration ($F_{2,29} = 0.04$; $P = 0.959$) or the HITsp ($F_{2,29} = 0.30$; $P = 0.744$) between the VO_2max levels in our sample.

Player aggression

We performed several analyses comparing the linear accelerations, rotational accelerations, and the HITsp for six measures of player aggression we collected across our sample. These included subscales of physical aggression, verbal aggression, anger, and hostility derived from the BPAQ. A total aggression score was also tallied representing total player aggression and resulting from the sum of the four BPAQ subscales. Lastly, we recorded PIM, a measure commonly used in ice hockey to identify aggressive players. It is a measure of the total number of penalty minutes assessed to an individual player across the entire season. For clarity of presentation, subsections delineating these six measures of player aggression can be found below. The means, 95% confidence intervals, and associated *P* values for our player aggression data can be found in **Tables 4.27 to 4.29**.

Physical aggression: There was a significant difference in the HITsp between highly physically aggressive players (14.2; 95% CI: 13.7-14.7) and players exhibiting moderate physical aggressive (13.5; 95% CI: 13.2-13.9) tendencies ($F_{2,29}$ = 3.43; *P* = 0.046). No differences were noted for linear acceleration ($F_{2,29}$ = 0.04; *P* = 0.957) or rotational acceleration ($F_{2,29}$ = 1.04; *P* = 0.366) for physical aggression measures.

Verbal aggression: Linear acceleration of head impacts did not differ between athletes who reported strong, moderate, or low verbal aggression tendencies ($F_{2,29}$ = 1.13; *P* = 0.336). No differences were noted for rotational acceleration ($F_{2,29}$ = 0.50; *P* = 0.612) or the HITsp ($F_{2,29}$ = 0.35; *P* = 0.710) for verbal aggression measures.

Anger: Linear acceleration of head impacts did not differ between athletes who reported high, moderate, or low anger tendencies ($F_{2,29}$ = 1.37; *P* = 0.270). No differences were noted for rotational acceleration ($F_{2,29}$ = 0.31; *P* = 0.733) or the HITsp ($F_{2,29}$ = 0.25; *P* = 0.784) for anger measures.

Hostility: Our data suggests a trend such that athlete reporting the highest amount of hostile tendencies (17.7 g; 95% CI: 17.3-18.0) experienced higher linear accelerations than those athletes reporting moderate levels (17.1 g; 95% CI: 16.5-17.6) of hostility ($F_{2,29}$ = 2.59; *P* = 0.092). No differences were noted for rotational acceleration ($F_{2,29}$ = 1.27; *P* = 0.297) or the HITsp ($F_{2,29}$ = 1.27; *P* = 0.296) for hostility measures.

Total aggression score: When combining all subscales of the BPAQ (physical aggression, verbal aggression, anger, and hostility), we noted no significant differences in linear rotation ($F_{2,29}$ = 0.16; *P* = 0.850), rotational acceleration ($F_{2,29}$ = 0.42; *P* = 0.663), or the HITsp ($F_{2,29}$ = 0.23; P = 0.795) across the tertiles representing strong, moderate, and low levels of aggression in our sample.

Penalties in minutes: Athletes who experienced the highest number of PIM experienced higher rotational accelerations (1631.7 rad/s^2; 95% CI: 1503.4-1771.1) than those athletes experiencing lower numbers of PIM (1465.2 rad/s^2; 95% CI: 1382.5-1552.9) in our sample ($F_{2,29}$ = 3.80; P = 0.034). Similarly, athletes with higher PIM (14.3; 95% CI: 13.9-14.7) experienced greater HITsp measures than those with lower PIM (13.6; 95% CI: 13.1-14.0) in our youth ice hockey sample ($F_{2,29}$ = 3.42; P = 0.046). No significant differences were observed in linear acceleration between players with high, moderate, and low numbers of PIM in our sample ($F_{2,29}$ = 0.58; P = 0.566).

Additional analyses for player aggression: We performed three additional random intercepts general mixed linear models (one for each dependent variable) while including all six measures of player aggression in the same statistical model. While adjusting for all other measures of aggression, we were able to identify a significant difference in rotational acceleration for our measure of physical aggression ($F_{2,19}$ = 5.85; P = 0.011) such that those who are most physically aggressive experience significantly lower rotational acceleration (P = 0.003) than those who are least physically aggressive. Also while adjusting for all other measures of aggression, we observed significantly lower measures of HITsp (13.6; 95% CI: 13.2-14.0) in athletes with moderate trait verbal aggression compared to those with the greatest amounts of trait verbal aggression (14.1; 95% CI: 13.6-14.5) ($F_{2,19}$ = 3.51; P = 0.050). No significant differences were observed for linear acceleration ($P > 0.05$), and the results associated with all aggression measures not expanded on in this subsection are the same as presented in the aforementioned subsections.

Table 4.1. Sample demographics.

Cohort	Mean	95% Confidence Interval		Range	
		Lower	Upper	Minimum	Maximum
Bantam (N = 31)					
Age (years)	14.0	13.9	14.2	13.0	14.0
Height (cm)	171.0	168.7	173.3	153.5	183.5
Mass (kg)	63.4	60.3	66.5	53.9	84.4
Midget (N = 21)					
Age (years)	15.8	15.4	16.1	14.0	16.0
Height (cm)	175.7	173.4	178.0	165.9	187.0
Mass (kg)	69.8	66.7	73.0	62.9	90.4
Total sample (N = 52)					
Age (years)	14.7	14.4	15.0	13.0	16.0
Height (cm)	172.5	170.8	174.2	153.5	187.0
Mass (kg)	65.2	62.9	67.6	53.9	90.4

Table 4.2. Descriptive factors – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts sustained by player position, event type, impact location, and whether the player struck an opponent or was struck by an opponent. The 95% confidence intervals and p-values are provided.

	Frequency of impacts	Linear acceleration (g)	95% Confidence Interval Lower	95% Confidence Interval Upper	*P* value[1]
Positional data[2]					
Defensemen (N = 19)	4294 (34.5%)	18.3	17.3	19.3	0.738
Forwards[3] (N = 35)	7969 (64.0%)	18.4	17.9	18.9	—
Event type data					
Games (N = 151)	9343 (75.0%)	18.4	18.0	18.9	0.531
Practices[3] (N = 137)	3115 (25.0%)	18.3	17.6	19.0	—
Location of head impact data					
Top[3]	1200 (9.6%)	21.2	20.0	22.4	—
Back	2878 (23.1%)	19.9	19.3	20.7	0.033
Front	4033 (32.4%)	17.8	17.3	18.3	<0.001
Side	4347 (34.9%)	17.2	16.9	17.6	<0.001
Striker vs. player struck[4]					
Striker[3]	352 (52.9%)	21.3	19.7	22.9	—
Player struck	314 (47.1%)	21.4	19.8	23.0	0.853
Total	12458	18.4	18.3	18.6	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Head impacts sustained by a single goaltender were not included for our positional comparison. This excluded 195 (1.5%) of our analyses
[3] Denotes the reference category used in mixed linear models
[4] A striking player or player struck could only be determined for 666 body collisions using the CHECC List

Table 4.3. Descriptive factors – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts sustained by player position, event type, impact location, and whether the player struck an opponent or was struck by an opponent. The 95% confidence intervals and p-values are provided.

	Frequency of impacts	Rotational acceleration (rad/s^2)	95% Confidence Interval Lower	Upper	*P* value[1]
Positional data[2]					
Defensemen (N = 19)	4294 (34.5%)	1433.3	1317.5	1559.3	0.376
Forwards[3] (N = 35)	7969 (64.0%)	1476.8	1409.7	1547.1	—
Event type data					
Games (N = 151)	9343 (75.0%)	1485.8	1420.8	1553.7	<0.001
Practices[3] (N = 137)	3115 (25.0%)	1373.8	1313.3	1437.1	—
Location of head impact data					
Top[3]	1200 (9.6%)	1038.3	977.5	1102.9	—
Back	2878 (23.1%)	1443.8	1387.5	1502.4	<0.001
Front	4033 (32.4%)	1469.2	1400.4	1541.3	<0.001
Side	4347 (34.9%)	1599.2	1512.8	1690.6	<0.001
Striker vs. player struck[4]					
Striker[3]	352 (52.9%)	1419.5	1351.6	1490.7	—
Player struck	314 (47.1%)	1452.3	1344.7	1568.4	0.423
Total	12458	1461.4	1445.9	1477.1	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Head impacts sustained by a single goaltender were not included for our positional comparison. This excluded 195 (1.5%) of our analyses
[3] Denotes the reference category used in mixed linear models
[4] A striking player or player struck could only be determined for 666 body collisions using the CHECC List

Table 4.4. Descriptive factors – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts sustained by player position, event type, impact location, and whether the player struck an opponent or was struck by an opponent. The 95% confidence intervals and p-values are provided.

	Frequency of impacts	HITsp	95% Confidence Interval Lower	Upper	*P* value[1]
Positional data[2]					
Defensemen (N = 19)	4294 (34.5%)	13.9	13.4	14.4	0.336
Forwards[3] (N = 35)	7969 (64.0%)	14.1	13.9	14.3	—
Event type data					
Games (N = 151)	9343 (75.0%)	14.1	13.9	14.4	<0.001
Practices[3] (N = 137)	3115 (25.0%)	13.6	13.3	13.9	—
Location of head impact data					
Top[3]	1200 (9.6%)	8.5	8.2	8.9	—
Back	2878 (23.1%)	11.2	10.9	11.4	<0.001
Front	4033 (32.4%)	15.9	15.6	16.1	<0.001
Side	4347 (34.9%)	16.8	16.5	17.0	<0.001
Striker vs. player struck[4]					
Striker[3]	352 (52.9%)	15.9	15.0	16.8	—
Player struck	314 (47.1%)	15.6	14.8	16.5	0.513
Total	12458	14.1	14.0	14.2	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Head impacts sustained by a single goaltender were not included for our positional comparison. This excluded 195 (1.5%) of our analyses
[3] Denotes the reference category used in mixed linear models
[4] A striking player or player struck could only be determined for 666 body collisions using the CHECC List

Table 4.5. Descriptive data – shifts played.

Descriptive data pertaining to the number of shifts played per period for our 2007-2008 cohort.

	Mean	**95% Confidence Interval**	
		Lower	**Upper**
Playing shifts per period			
Period 1	5.6	5.6	5.7
Period 2	5.4	5.3	5.5
Period 3	5.4	5.3	5.4
Covariate data accounting for playing shift exposures			
Absolute number of playing shifts	5.6	5.5	5.6
Weighted summation of playing shifts	8.0	7.9	8.1
Average number of playing shifts	5.6	5.5	5.6
Weighted average number if playing shifts	5.6	5.5	5.6

Table 4.6. Game-related exposure – linear acceleration.

Frequency (percentage) of recorded impacts, mean linear acceleration of head impacts sustained by playing period, and by playing shift exposure data type. The 95% confidence intervals and p-values are provided.

	Frequency of impacts	Linear acceleration (g)	95% Confidence Interval Lower	95% Confidence Interval Upper	*P* value[1]
Period of play					
Period 1	918 (30.8%)	19.5	18.8	20.3	0.002
Period 2	916 (30.8%)	19.4	18.6	20.3	0.002
Period 3[2]	1144 (38.4%)	20.7	19.9	21.6	—
Absolute number of playing shifts					
Low	778 (26.1%)	19.8	18.8	20.8	0.755
Moderate	1476 (49.6%)	20.1	19.3	20.9	0.654
High[2]	724 (24.3%)	19.9	19.1	20.8	—
Weighted summation of playing shifts					
Low	1033 (34.7%)	19.7	19.0	20.4	0.215
Moderate	917 (30.8%)	20.1	19.0	21.2	0.812
High[2]	1028 (34.5%)	20.2	19.4	21.0	—
Average number of playing shifts					
Low	875 (29.4%)	19.6	18.6	20.7	0.306
Moderate	1311 (44.0%)	20.1	19.4	20.8	0.998
High[2]	792 (26.6%)	20.1	19.3	21.0	—
Weighted average number of playing shifts					
Low	943 (31.7%)	20.0	18.9	21.2	0.870
Moderate	1174 (39.4%)	19.8	19.1	20.6	0.531
High[2]	861 (28.9%)	20.1	19.3	20.8	—
Total	2978	20.1	19.7	20.4	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.7. Game-related exposure – rotational acceleration.

Frequency (percentage) of recorded impacts, mean rotational acceleration of head impacts sustained by playing period, and by playing shift exposure data type. The 95% confidence intervals and p-values are provided.

	Frequency of impacts	Rotational acceleration (rad/s^2)	95% Confidence Interval		*P* value[1]
			Lower	Upper	
Period of play					
Period 1	918 (30.8%)	1280.5	1219.6	1344.5	0.729
Period 2	916 (30.8%)	1248.0	1184.3	1315.2	0.226
Period 3[2]	1144 (38.4%)	1289.6	1222.5	1360.3	—
Absolute number of playing shifts					
Low	778 (26.1%)	1246.1	1178.5	1317.5	0.191
Moderate	1476 (49.6%)	1282.2	1219.6	1347.9	0.759
High[2]	724 (24.3%)	1290.9	1232.8	1351.8	—
Weighted summation of playing shifts					
Low	1033 (34.7%)	1273.6	1216.2	1333.9	0.980
Moderate	917 (30.8%)	1274.3	1198.6	1354.8	0.997
High[2]	1028 (34.5%)	1274.2	1221.0	1329.7	—
Average number of playing shifts					
Low	875 (29.4%)	1245.6	1170.6	1325.4	< 0.001
Moderate	1311 (44.0%)	1278.6	1217.9	1342.4	< 0.001
High[2]	792 (26.6%)	1300.1	1231.4	1372.7	—
Weighted average number of playing shifts					
Low	943 (31.7%)	1260.9	1188.6	1337.6	0.298
Moderate	1174 (39.4%)	1267.5	1208.4	1329.5	0.316
High[2]	861 (28.9%)	1297.5	1234.8	1363.4	—
Total	2978	1289.0	1264.4	1314.1	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.8. Game-related exposure – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts sustained by playing period, and by playing shift exposure data type. The 95% confidence intervals and p-values are provided.

	Frequency of impacts	HITsp	95% Confidence Interval Lower	95% Confidence Interval Upper	*P* value[1]
Period of play					
Period 1	918 (30.8%)	14.3	13.9	14.7	0.235
Period 2	916 (30.8%)	14.1	13.4	14.47	0.130
Period 3[2]	1144 (38.4%)	14.5	14.1	14.9	—
Absolute number of playing shifts					
Low	778 (26.1%)	14.2	13.6	14.8	0.727
Moderate	1476 (49.6%)	14.3	13.9	14.8	0.662
High[2]	724 (24.3%)	14.3	13.9	14.7	—
Weighted summation of playing shifts					
Low	1033 (34.7%)	14.3	13.9	14.7	0.688
Moderate	917 (30.8%)	14.3	13.8	15.0	0.492
High[2]	1028 (34.5%)	14.2	13.8	14.7	—
Average number of playing shifts					
Low	875 (29.4%)	14.1	13.5	14.7	0.330
Moderate	1311 (44.0%)	14.5	14.1	14.9	0.261
High[2]	792 (26.6%)	14.3	13.8	14.7	—
Weighted average number of playing shifts					
Low	943 (31.7%)	14.3	13.7	14.9	0.938
Moderate	1174 (39.4%)	14.3	13.9	14.7	0.729
High[2]	861 (28.9%)	14.3	13.8	14.7	—
Total	2978	14.4	14.2	14.6	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.9. Infraction type – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts sustained by legality of collision, and infraction type. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts	Linear acceleration (g)	95% CI		*P* value[1]
			Lower	Upper	
Legality of body collision					
Illegal collision	115 (17.3%)	23.0	21.4	24.8	0.012
Legal collision[2]	550 (82.7%)	21.0	19.5	22.5	—
Type of infraction					
Boarding/charging	20 (3.0%)	21.2	18.7	24.0	0.868
Checking from behind	19 (2.9%)	21.4	18.8	24.3	0.722
Elbowing/head contact	76 (11.4%)	24.0	21.9	26.2	0.005
Legal collision[2]	550 (82.7%)	21.0	19.5	22.5	—
Total	665	21.5	20.8	22.3	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.10. Infraction type – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts sustained by legality of collision, and infraction type. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts	**Rotational acceleration (rad/s^2)**	**95% CI Lower**	**95% CI Upper**	***P* value[1]**
Legality of body collision					
Illegal collision	115 (17.3%)	1529.9	1388.5	1685.8	0.142
Legal collision[2]	550 (82.7%)	1417.5	1334.8	1505.3	—
Type of infraction					
Boarding/charging	20 (3.0%)	1575.9	1419.2	1749.9	0.103
Checking from behind	19 (2.9%)	1197.7	953.3	1504.9	0.144
Elbowing/head contact	76 (11.4%)	1614.3	1419.6	1835.7	0.059
Legal collision[2]	550 (82.7%)	1418.4	1335.4	1506.5	—
Total	665	1441.1	1386.6	1497.7	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.11. Infraction type – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts sustained by legality of collision, and infraction type. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts	HITsp	95% CI		*P* value[1]
			Lower	Upper	
Legality of body collision					
Illegal collision	115 (17.3%)	16.8	15.8	17.9	0.021
Legal collision[2]	550 (82.7%)	15.5	14.7	16.4	—
Type of infraction					
Boarding/charging	20 (3.0%)	16.8	14.6	19.3	0.364
Checking from behind	19 (2.9%)	14.1	12.0	16.7	0.199
Elbowing/head contact	76 (11.4%)	17.6	16.0	19.2	0.010
Legal collision[2]	550 (82.7%)	15.5	14.7	16.4	—
Total	665	15.8	15.4	16.3	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.12. Body collision type – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts sustained by body collision type, anticipation, and overall impression. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts[1]	Linear acceleration (g)	95% CI Lower	Upper	*P* value[2]
Body collision type					
Along playing boards	421 (63.3%)	20.7	19.4	22.2	0.036
Open-ice[3]	245 (36.8%)	22.4	20.6	24.3	—
Anticipation					
Anticipated	564 (84.7%)	21.1	19.5	22.8	0.135
Unanticipated[3]	102 (15.3%)	22.6	20.9	24.5	—
Overall impression					
Anticipated—good	315 (47.3%)	20.7	19.1	22.5	0.098
Anticipated—poor	249 (37.4%)	21.4	19.6	23.4	0.279
Unanticipated[3]	102 (15.3%)	22.6	20.9	24.5	—
Total	666	21.5	20.8	22.3	—

[1] Percentages may add up to 100.01% due to rounding
[2] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[3] Denotes the reference category used in mixed linear models

Table 4.13. Body collision type – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts sustained by body collision type, anticipation, and overall impression. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts[1]	Rotational acceleration (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[2]
Body collision type					
Along playing boards	421 (63.3%)	1367.7	1295.6	1443.9	0.003
Open-ice[3]	245 (36.8%)	1564.7	1440.3	1699.9	—
Anticipation					
Anticipated	564 (84.7%)	1414.3	1330.6	1503.3	0.138
Unanticipated[3]	102 (15.3%)	1550.0	1377.4	1744.2	—
Overall impression					
Anticipated—good	315 (47.3%)	1409.4	1303.0	1524.4	0.145
Anticipated—poor	249 (37.4%)	1420.4	1312.4	1537.3	0.184
Unanticipated[3]	102 (15.3%)	1549.9	1377.3	1744.2	—
Total	666	1439.9	1385.5	1496.4	—

[1] Percentages may add up to 100.01% due to rounding
[2] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[3] Denotes the reference category used in mixed linear models

Table 4.14. Body collision type – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts sustained by body collision type, anticipation, and overall impression. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts[1]	HITsp	95% CI Lower	95% CI Upper	*P* value[2]
Body collision type					
Along playing boards	421 (63.3%)	15.5	14.74	16.3	0.055
Open-ice[3]	245 (36.8%)	16.3	15.3	17.3	—
Anticipation					
Anticipated	564 (84.7%)	15.8	15.0	16.6	0.755
Unanticipated[3]	102 (15.3%)	15.5	14.2	17.1	—
Overall impression					
Anticipated—good	315 (47.3%)	15.6	14.6	16.5	0.990
Anticipated—poor	249 (37.4%)	16.1	15.2	17.0	0.491
Unanticipated[3]	102 (15.3%)	15.5	14.2	17.1	—
Total	666	15.8	15.3	16.3	—

[1] Percentages may add up to 100.01% due to rounding
[2] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[3] Denotes the reference category used in mixed linear models

Table 4.15. Relative body position – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts sustained by relative body position as evaluated by the CHECC List. The associated 95% confidence intervals and p-values are provided.

CHECC List item	Frequency of impacts	Linear acceleration (g)	95% CI Lower	95% CI Upper	*P* value[1]
Player looking ahead in direction of movement?					
No	20 (3.0%)	22.9	19.2	27.2	0.388
Yes[2]	646 (97.0%)	21.3	19.8	22.8	—
Player appears to be looking in direction of impending body collision?					
No	139 (20.9%)	21.6	20.1	23.2	0.658
Yes[2]	102 (15.3%)	21.2	19.6	22.9	—
Knee flexion greater than 30 degrees at the time of body collision?					
No	235 (35.3%)	21.3	19.8	23.0	0.996
Yes[2]	431 (64.7%)	21.3	19.8	23.0	—
Trunk flexion at time of body collision?					
No	315 (47.3%)	21.7	19.7	23.8	0.639
Yes[2]	249 (37.4%)	21.1	19.3	23.0	—
Player drives into collision with shoulders?					
No	315 (47.3%)	21.8	20.1	23.6	0.323
Yes[2]	249 (37.4%)	20.9	19.4	22.6	—
Player uses elbow(s) in body collision?					
No	315 (47.3%)	21.3	19.8	22.9	0.901
Yes[2]	249 (37.4%)	21.5	18.3	25.2	—
Player uses hands in body collision?					
No	315 (47.3%)	21.2	19.6	22.8	0.488
Yes[2]	249 (37.4%)	22.0	19.7	24.5	—
Feet are shoulder width apart at the time of the body collision?					
No	315 (47.3%)	21.1	19.2	23.2	0.656
Yes[2]	249 (37.4%)	21.4	20.0	22.9	—
Player uses stick during collision?					
No	315 (47.3%)	21.3	19.9	22.9	0.608
Yes[2]	249 (37.4%)	20.3	16.2	25.3	—
Player uses legs to drive into or through a body collision?					
No	315 (47.3%)	21.7	20.1	23.5	0.049
Yes[2]	249 (37.4%)	20.5	19.2	21.9	—
Player delivering or receiving a pass/shot at the time of body collision?					
No	315 (47.3%)	21.3	19.9	22.9	0.956
Yes[2]	249 (37.4%)	21.3	19.6	23.1	—
Total	666	21.5	20.8	22.3	—

[1] *P* values reflect significant differences relative to reference category[2] used by random intercepts general mixed linear model analyses

Table 4.16. Relative body position – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts sustained by relative body position as evaluated by the CHECC List. The associated 95% confidence intervals and p-values are provided.

CHECC List item	Frequency of impacts	Rotational accel. (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[1]
Player looking ahead in direction of movement?					
No	20 (3.0%)	1419.3	1274.4	1580.8	0.844
Yes[2]	646 (97.0%)	1435.7	1354.6	1521.6	—
Player appears to be looking in direction of impending body collision?					
No	139 (20.9%)	1446.2	1335.7	1565.9	0.794
Yes[2]	102 (15.3%)	1432.1	1348.0	1521.5	—
Knee flexion greater than 30 degrees at the time of body collision?					
No	235 (35.3%)	1448.4	1330.3	1577.0	0.715
Yes[2]	431 (64.7%)	1427.9	1347.5	1513.0	—
Trunk flexion at time of body collision?					
No	315 (47.3%)	1450.2	1329.4	1581.9	0.771
Yes[2]	249 (37.4%)	1426.9	1324.4	1537.4	—
Player drives into collision with shoulders?					
No	315 (47.3%)	1463.9	1350.0	1587.5	0.460
Yes[2]	249 (37.4%)	1411.8	1314.2	1516.7	—
Player uses elbow(s) in body collision?					
No	315 (47.3%)	1430.8	1344.8	1522.3	0.531
Yes[2]	249 (37.4%)	1516.5	1283.7	1791.6	—
Player uses hands in body collision?					
No	315 (47.3%)	1422.9	1329.4	1522.9	0.525
Yes[2]	249 (37.4%)	1487.8	1318.0	1679.5	—
Feet are shoulder width apart at the time of the body collision?					
No	315 (47.3%)	1428.3	1316.7	1549.3	0.877
Yes[2]	249 (37.4%)	1438.1	1347.2	1535.1	—
Player uses stick during collision?					
No	315 (47.3%)	1434.0	1349.4	1523.9	0.841
Yes[2]	249 (37.4%)	1476.5	1098.0	1985.4	—
Player uses legs to drive into or through a body collision?					
No	315 (47.3%)	1424.8	1326.6	1530.3	0.446
Yes[2]	249 (37.4%)	1455.6	1393.3	1520.8	—
Player delivering or receiving a pass/shot at the time of body collision?					
No	315 (47.3%)	1420.3	1358.5	1485.0	0.446
Yes[2]	249 (37.4%)	1456.0	1334.6	1588.4	—
Total	666	1439.9	1385.5	1496.4	—

[1] *P* values reflect significant differences relative to reference category[2] used by random intercepts general mixed linear model analyses

Table 4.17. Relative body position – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts sustained by relative body position as evaluated by the CHECC List. The associated 95% confidence intervals and p-values are provided.

CHECC List item	Frequency of impacts	HITsp	95% CI Lower	95% CI Upper	*P* value[1]
Player looking ahead in direction of movement?					
No	20 (3.0%)	15.9	14.1	18.0	0.845
Yes[2]	646 (97.0%)	15.7	15.0	16.6	—
Player appears to be looking in direction of impending body collision?					
No	139 (20.9%)	15.2	14.3	16.2	0.120
Yes[2]	102 (15.3%)	15.9	15.1	16.7	—
Knee flexion greater than 30 degrees at the time of body collision?					
No	235 (35.3%)	15.8	14.9	16.8	0.846
Yes[2]	431 (64.7%)	15.7	14.9	16.6	—
Trunk flexion at time of body collision?					
No	315 (47.3%)	15.8	14.9	16.7	0.961
Yes[2]	249 (37.4%)	15.7	14.8	16.8	—
Player drives into collision with shoulders?					
No	315 (47.3%)	15.8	15.1	16.7	0.619
Yes[2]	249 (37.4%)	15.7	14.7	16.6	—
Player uses elbow(s) in body collision?					
No	315 (47.3%)	15.8	14.9	16.6	0.869
Yes[2]	249 (37.4%)	15.5	13.4	18.0	—
Player uses hands in body collision?					
No	315 (47.3%)	15.5	14.7	16.4	0.190
Yes[2]	249 (37.4%)	16.6	15.1	18.2	—
Feet are shoulder width apart at the time of the body collision?					
No	315 (47.3%)	15.7	14.8	16.7	0.986
Yes[2]	249 (37.4%)	15.7	14.9	16.6	—
Player uses stick during collision?					
No	315 (47.3%)	15.7	14.9	16.6	0.873
Yes[2]	249 (37.4%)	15.8	14.4	17.5	—
Player uses legs to drive into or through a body collision?					
No	315 (47.3%)	16.0	15.1	16.9	0.085
Yes[2]	249 (37.4%)	15.3	14.6	16.1	—
Player delivering or receiving a pass/shot at the time of body collision?					
No	315 (47.3%)	15.9	15.0	16.7	0.466
Yes[2]	249 (37.4%)	15.6	14.7	16.5	—
Total	666	15.8	15.3	16.3	—

[1] *P* values reflect significant differences relative to reference category[2] used by random intercepts general mixed linear model analyses

Table 4.18. Cervical muscle strength – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts across the tertile measures of cervical muscle strength. The associated 95% confidence intervals and p-values are provided.

Cervical muscle group	Frequency of impacts	Linear acceleration (g)	95% CI Lower	95% CI Upper	*P* value[1]
Anterior neck					
Weak	2679 (34.5%)	17.3	16.8	17.9	0.217
Moderate	2585 (33.3%)	17.4	17.0	17.9	0.264
Strong[2]	2506 (32.3%)	17.9	17.3	18.5	—
Anterolateral neck					
Weak	2514 (32.4%)	17.6	16.9	18.4	0.895
Moderate	2695 (34.7%)	17.5	17.2	17.9	0.978
Strong[2]	2561 (33.0%)	17.5	17.1	18.0	—
Cervical rotation					
Weak	2487 (32.0%)	17.9	17.1	18.7	0.115
Moderate	2736 (35.2%)	17.6	17.3	18.0	0.084
Strong[2]	2547 (32.8%)	17.2	16.8	17.6	—
Posterolateral neck					
Weak	2572 (33.1%)	17.7	16.9	18.4	0.687
Moderate	2574 (33.1%)	17.5	17.1	17.8	0.950
Strong[2]	2624 (33.8%)	17.5	17.0	18.0	—
Upper trapezius					
Weak	2590 (33.3%)	17.6	16.9	18.3	0.758
Moderate	2540 (32.7%)	17.6	17.2	18.0	0.637
Strong[2]	2640 (34.0%)	17.4	16.8	18.1	—
Total	7770	17.5	17.3	17.7	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 4.19. Cervical muscle strength – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts across the tertile measures of cervical muscle strength. The associated 95% confidence intervals and p-values are provided.

Cervical muscle group	Frequency of impacts	Rotational accel. (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[1]
Anterior neck					
Weak	2679 (34.5%)	1642.9	1530.6	1763.3	0.485
Moderate	2585 (33.3%)	1482.6	1409.0	1560.1	0.191
Strong[2]	2506 (32.3%)	1581.4	1453.3	1720.7	—
Anterolateral neck					
Weak	2514 (32.4%)	1597.0	1496.9	1703.9	0.827
Moderate	2695 (34.7%)	1516.8	1401.7	1641.4	0.459
Strong[2]	2561 (33.0%)	1579.7	1461.8	1707.2	—
Cervical rotation					
Weak	2487 (32.0%)	1641.6	1536.8	1753.5	0.268
Moderate	2736 (35.2%)	1513.5	1406.7	1628.5	0.613
Strong[2]	2547 (32.8%)	1553.7	1441.7	1674.3	—
Posterolateral neck					
Weak	2572 (33.1%)	1631.2	1535.4	1733.0	0.627
Moderate	2574 (33.1%)	1480.1	1383.6	1583.4	0.177
Strong[2]	2624 (33.8%)	1591.4	1464.1	1729.7	—
Upper trapezius					
Weak	2590 (33.3%)	1527.1	1421.7	1640.2	0.455
Moderate	2540 (32.7%)	1579.2	1483.0	1681.6	0.853
Strong[2]	2640 (34.0%)	1595.8	1451.2	1754.8	—
Total	7770	1587.7	1565.4	1610.2	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 4.20. Cervical muscle strength – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts across the tertile measures of cervical muscle strength. The associated 95% confidence intervals and p-values are provided.

Cervical muscle group	Frequency of impacts	HITsp	95% CI Lower	95% CI Upper	*P* value[1]
Anterior neck					
Weak	2679 (34.5%)	14.0	13.5	14.5	0.992
Moderate	2585 (33.3%)	13.9	13.5	14.3	0.840
Strong[2]	2506 (32.3%)	14.0	13.5	14.5	—
Anterolateral neck					
Weak	2514 (32.4%)	13.9	13.5	14.3	0.401
Moderate	2695 (34.7%)	13.8	13.2	14.3	0.228
Strong[2]	2561 (33.0%)	14.2	13.7	14.6	—
Cervical rotation					
Weak	2487 (32.0%)	14.1	13.6	14.6	0.965
Moderate	2736 (35.2%)	13.7	13.2	14.2	0.218
Strong[2]	2547 (32.8%)	14.1	13.7	14.5	—
Posterolateral neck					
Weak	2572 (33.1%)	14.0	13.7	14.4	0.941
Moderate	2574 (33.1%)	13.7	13.2	14.3	0.329
Strong[2]	2624 (33.8%)	14.1	13.6	14.5	—
Upper trapezius					
Weak	2590 (33.3%)	13.6	13.2	14.0	0.011
Moderate	2540 (32.7%)	14.0	13.5	14.4	0.140
Strong[2]	2640 (34.0%)	14.4	14.0	14.8	—
Total	7770	14.0	13.9	14.1	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 4.21. Head, neck, and player anthropometrics – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts across the tertile measures of cervical, head, and player anthropometrics. The associated 95% confidence intervals and p-values are provided.

Anthropometric measure	Frequency of impacts	Linear acceleration (g)	95% CI Lower	95% CI Upper	*P* value[1]
Player height					
Short	2595 (33.4%)	17.5	17.0	18.0	0.744
Moderate	2930 (37.7%)	17.7	17.2	18.3	0.388
Tall[2]	2245 (28.9%)	17.4	16.8	18.0	—
Player mass					
Light	2566 (33.0%)	17.4	16.9	17.9	0.466
Moderate	2481 (31.9%)	17.6	17.3	17.9	0.796
Heavy[2]	2723 (35.0%)	17.7	17.0	18.4	—
Head-neck segment length					
Short	3057 (39.3%)	17.6	17.2	18.0	0.835
Moderate	1798 (23.1%)	17.3	16.7	17.9	0.373
Long[2]	2915 (37.5%)	17.7	17.0	18.3	—
Neck circumference					
Short	2870 (36.9%)	17.4	16.9	17.8	0.666
Moderate	2357 (30.3%)	17.9	17.2	18.5	0.415
Long[2]	2543 (32.7%)	17.5	17.0	18.1	—
Neck medial-lateral diameter					
Narrow	2679 (34.5%)	17.9	17.6	18.1	0.100
Moderate	2494 (32.1%)	17.4	16.8	18.2	0.808
Wide[2]	2597 (33.4%)	17.3	16.8	17.9	—
Neck anterior-posterior diameter					
Narrow	2917 (37.5%)	17.2	16.8	17.7	0.033
Moderate	2419 (31.1%)	17.4	17.0	17.8	0.070
Wide[2]	2434 (31.3%)	18.2	17.4	18.9	—
Head circumference					
Short	2787 (35.9%)	17.6	17.0	18.1	0.983
Moderate	2266 (29.2%)	17.5	16.9	18.1	0.891
Long[2]	2717 (35.0%)	17.6	17.0	18.1	—
Head medial-lateral diameter					
Narrow	2505 (32.2%)	17.6	16.9	18.3	0.502
Moderate	2957 (38.1%)	17.3	16.9	17.8	0.098
Wide[2]	2308 (29.7%)	17.9	17.4	18.4	—
Head anterior-posterior diameter					
Narrow	2360 (30.4%)	17.6	17.0	18.3	0.596
Moderate	2965 (38.2%)	17.6	17.1	18.1	0.692
Wide[2]	2445 (31.5%)	17.4	16.9	17.9	—
Total	7770	17.5	17.3	17.7	—

[1] *P* values reflect significant differences relative to reference category[2] used by random intercepts general mixed linear model analyses

Table 4.22. Head, neck, and player anthropometrics – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts across the tertile measures of cervical, head, and player anthropometrics. The associated 95% confidence intervals and p-values are provided.

Anthropometric measure	Frequency of impacts	Rotational accel. (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[1]
Player height					
Short	2595 (33.4%)	1537.5	1426.7	1656.9	0.421
Moderate	2930 (37.7%)	1562.4	1460.7	1671.3	0.589
Tall[2]	2245 (28.9%)	1607.7	1479.4	1747.2	—
Player mass					
Light	2566 (33.0%)	1467.3	1408.5	1528.5	0.001
Moderate	2481 (31.9%)	1569.3	1422.3	1731.5	0.258
Heavy[2]	2723 (35.0%)	1675.8	1574.6	1783.4	—
Head-neck segment length					
Short	3057 (39.3%)	1573.5	1448.3	1709.5	0.852
Moderate	1798 (23.1%)	1564.0	1471.7	1662.0	0.932
Long[2]	2915 (37.5%)	1557.9	1454.2	1669.1	—
Neck circumference					
Short	2870 (36.9%)	1500.8	1418.3	1588.1	0.091
Moderate	2357 (30.3%)	1586.2	1466.4	1715.7	0.576
Long[2]	2543 (32.7%)	1637.9	1504.0	1783.7	—
Neck medial-lateral diameter					
Narrow	2679 (34.5%)	1546.5	1438.9	1662.2	0.995
Moderate	2494 (32.1%)	1602.5	1501.4	1710.5	0.501
Wide[2]	2597 (33.4%)	1545.9	1418.8	1684.5	—
Neck anterior-posterior diameter					
Narrow	2917 (37.5%)	1506.7	1425.7	1592.3	0.015
Moderate	2419 (31.1%)	1530.8	1417.0	1653.7	0.062
Wide[2]	2434 (31.3%)	1695.7	1572.9	1828.2	—
Head circumference					
Short	2787 (35.9%)	1409.4	1355.5	1465.4	< 0.001
Moderate	2266 (29.2%)	1638.2	1565.8	1713.9	0.208
Long[2]	2717 (35.0%)	1722.7	1612.9	1840.0	—
Head medial-lateral diameter					
Narrow	2505 (32.2%)	1478.5	1402.6	1558.5	0.004
Moderate	2957 (38.1%)	1552.6	1454.8	1657.0	0.059
Wide[2]	2308 (29.7%)	1714.6	1582.7	1857.5	—
Head anterior-posterior diameter					
Narrow	2360 (30.4%)	1487.7	1367.9	1617.9	0.046
Moderate	2965 (38.2%)	1561.5	1473.3	1655.1	0.170
Wide[2]	2445 (31.5%)	1659.1	1552.9	1772.6	—
Total	7770	1587.7	1565.4	1610.2	—

[1] *P* values reflect significant differences relative to reference category[2] used by random intercepts general mixed linear model analyses

Table 4.23. Head, neck, and player anthropometrics – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts across the tertile measures of cervical, head, and player anthropometrics. The associated 95% confidence intervals and p-values are provided.

Anthropometric measure	Frequency of impacts	HITsp	95% CI Lower	95% CI Upper	*P* value[1]
Player height					
Short	2595 (33.4%)	13.8	13.3	14.3	0.384
Moderate	2930 (37.7%)	14.0	13.6	14.4	0.683
Tall[2]	2245 (28.9%)	14.1	13.6	14.6	—
Player mass					
Light	2566 (33.0%)	13.8	13.3	14.2	0.069
Moderate	2481 (31.9%)	13.8	13.2	14.4	0.146
Heavy[2]	2723 (35.0%)	14.3	13.9	14.7	—
Head-neck segment length					
Short	3057 (39.3%)	14.1	13.6	14.6	0.687
Moderate	1798 (23.1%)	13.7	13.4	14.1	0.433
Long[2]	2915 (37.5%)	13.9	13.5	14.4	—
Neck circumference					
Short	2870 (36.9%)	13.8	13.4	14.3	0.287
Moderate	2357 (30.3%)	13.9	13.4	14.4	0.534
Long[2]	2543 (32.7%)	14.1	13.7	14.6	—
Neck medial-lateral diameter					
Narrow	2679 (34.5%)	14.1	13.7	14.5	0.414
Moderate	2494 (32.1%)	13.9	13.4	14.4	0.753
Wide[2]	2597 (33.4%)	13.8	13.3	14.3	—
Neck anterior-posterior diameter					
Narrow	2917 (37.5%)	13.8	13.4	14.3	0.232
Moderate	2419 (31.1%)	13.8	13.5	14.2	0.226
Wide[2]	2434 (31.3%)	14.2	13.7	14.7	—
Head circumference					
Short	2787 (35.9%)	13.4	13.0	13.8	0.006
Moderate	2266 (29.2%)	14.5	14.1	14.8	0.251
Long[2]	2717 (35.0%)	14.2	13.8	14.6	—
Head medial-lateral diameter					
Narrow	2505 (32.2%)	13.7	13.4	14.1	0.133
Moderate	2957 (38.1%)	14.0	13.5	14.5	0.507
Wide[2]	2308 (29.7%)	14.2	13.7	14.7	—
Head anterior-posterior diameter					
Narrow	2360 (30.4%)	13.4	13.0	14.0	0.026
Moderate	2965 (38.2%)	14.2	13.9	14.6	0.876
Wide[2]	2445 (31.5%)	14.2	13.8	14.6	—
Total	7770	14.0	13.9	14.1	—

[1] *P* values reflect significant differences relative to reference category[2] used by random intercepts general mixed linear model analyses

Table 4.24. General aerobic fitness – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts across the tertile measures of general aerobic fitness. The associated 95% confidence intervals and p-values are provided.

Aerobic fitness measure	Frequency of impacts	Linear acceleration (g)	95% CI Lower	95% CI Upper	*P* value[1]
Laps complete during the FAST					
Least	2480 (31.9%)	17.1	16.7	17.5	0.013
Moderate	2653 (34.1%)	17.7	17.1	18.3	0.501
Most[2]	2637 (33.9%)	18.0	17.4	18.6	—
Maximal oxygen consumption ($\boldsymbol{VO_2max}$ ***)***					
Least	2703 (34.8%)	17.0	16.6	17.4	< 0.001
Moderate	2493 (32.1%)	17.8	17.1	18.6	0.731
Most[2]	2574 (33.1%)	18.0	17.7	18.2	—
Total	7770	17.5	17.3	17.7	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 4.25. General aerobic fitness – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts across the tertile measures of general aerobic fitness. The associated 95% confidence intervals and p-values are provided.

Aerobic fitness measure	Frequency of impacts	Rotational accel. (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[1]
Laps complete during the FAST					
Least	2480 (31.9%)	1497.8	1406.9	1594.5	0.015
Moderate	2653 (34.1%)	1552.8	1435.1	1680.1	0.128
Most[2]	2637 (33.9%)	1678.6	1573.7	1790.4	—
Maximal oxygen consumption (VO_2max)					
Least	2703 (34.8%)	1566.3	1469.4	1669.6	0.860
Moderate	2493 (32.1%)	1577.5	1460.9	1703.4	0.773
Most[2]	2574 (33.1%)	1551.8	1424.2	1690.9	—
Total	7770	1587.7	1565.4	1610.2	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.26. General aerobic fitness – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts across the tertile measures of general aerobic fitness. The associated 95% confidence intervals and p-values are provided.

Aerobic fitness measure	Frequency of impacts	HITsp	95% CI Lower	95% CI Upper	*P* value[1]
Laps complete during the FAST					
Least	2480 (31.9%)	13.8	13.3	14.3	0.557
Moderate	2653 (34.1%)	14.1	13.7	14.6	0.587
Most[2]	2637 (33.9%)	14.0	13.6	14.3	—
Maximal oxygen consumption (VO_2max)					
Least	2703 (34.8%)	13.8	13.4	14.2	0.505
Moderate	2493 (32.1%)	14.0	13.5	14.5	0.906
Most[2]	2574 (33.1%)	14.0	13.5	14.6	—
Total	7770	14.0	13.9	14.1	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 4.27. Player aggression – linear acceleration.

Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts across the tertile measures of player aggression. The associated 95% confidence intervals and p-values are provided.

Aggression measure	Frequency of impacts	Linear acceleration (g)	95% CI Lower	95% CI Upper	*P* value[1]
Physical aggression					
Least	2664 (34.3%)	17.5	17.0	18.0	0.772
Moderate	2525 (32.5%)	17.6	17.0	18.2	0.922
Greatest[2]	2581 (33.2%)	17.6	17.0	18.2	—
Verbal aggression					
Least	2272 (29.2%)	17.7	17.3	18.3	0.810
Moderate	2949 (38.0%)	17.3	16.9	17.7	0.369
Greatest[2]	2549 (32.8%)	17.6	17.0	18.3	—
Anger					
Least	2719 (35.0%)	17.9	17.4	18.3	0.176
Moderate	2319 (29.8%)	17.3	16.6	18.0	0.750
Greatest[2]	2732 (35.2%)	17.4	17.0	17.9	—
Hostility					
Least	2917 (37.5%)	17.9	17.3	18.5	0.533
Moderate	2273 (29.3%)	17.1	16.5	17.6	0.065
Greatest[2]	2580 (33.2%)	17.7	17.3	18.0	—
Total aggression score					
Least	2676 (34.4%)	17.6	17.2	18.1	0.579
Moderate	2669 (34.4%)	17.5	16.8	18.2	0.913
Greatest[2]	2425 (31.2%)	17.5	17.0	17.9	—
Penalties in minutes					
Least	2395 (30.8%)	17.4	17.0	17.8	0.732
Moderate	2538 (32.7%)	17.8	17.1	18.5	0.483
Greatest[2]	2837 (36.5%)	17.5	17.0	18.1	—
Total	7770	17.5	17.3	17.7	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 4.28. Player aggression – rotational acceleration.

Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts across the tertile measures of player aggression. The associated 95% confidence intervals and p-values are provided.

Aggression measure	Frequency of impacts	Rotational accel. (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[1]
Physical aggression					
Least	2664 (34.3%)	1581.6	1493.8	1674.6	0.596
Moderate	2525 (32.5%)	1505.0	1394.7	1624.1	0.173
Greatest[2]	2581 (33.2%)	1622.7	1498.6	1757.2	—
Verbal aggression					
Least	2272 (29.2%)	1568.3	1458.2	1686.9	0.604
Moderate	2949 (38.0%)	1605.5	1497.8	1721.0	0.326
Greatest[2]	2549 (32.8%)	1526.3	1413.6	1648.0	—
Anger					
Least	2719 (35.0%)	1602.0	1486.0	1727.0	0.502
Moderate	2319 (29.8%)	1548.3	1450.5	1652.8	0.961
Greatest[2]	2732 (35.2%)	1544.5	1425.6	1673.3	—
Hostility					
Least	2917 (37.5%)	1631.4	1520.4	1750.4	0.124
Moderate	2273 (29.3%)	1564.4	1469.5	1665.5	0.419
Greatest[2]	2580 (33.2%)	1502.1	1386.5	1627.2	—
Total aggression score					
Least	2676 (34.4%)	1603.5	1502.3	1711.6	0.623
Moderate	2669 (34.4%)	1539.8	1442.7	1643.4	0.827
Greatest[2]	2425 (31.2%)	1559.1	1417.3	1715.1	—
Penalties in minutes					
Least	2395 (30.8%)	1465.2	1382.5	1552.9	0.037
Moderate	2538 (32.7%)	1622.2	1520.1	1731.1	0.910
Greatest[2]	2837 (36.5%)	1631.7	1503.4	1771.1	—
Total	7770	1587.7	1565.4	1610.2	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses
[2] Denotes the reference category used in mixed linear models

Table 4.29. Player aggression – HITsp.

Frequency (percentage) of recorded impacts, mean HITsp of head impacts across the tertile measures of player aggression. The associated 95% confidence intervals and p-values are provided.

Aggression measure	Frequency of impacts	HITsp	95% CI Lower	95% CI Upper	*P* value[1]
Physical aggression					
Least	2664 (34.3%)	14.2	13.7	14.6	0.877
Moderate	2525 (32.5%)	13.5	13.2	13.9	0.037
Greatest[2]	2581 (33.2%)	14.2	13.7	14.7	—
Verbal aggression					
Least	2272 (29.2%)	13.9	13.5	14.4	0.723
Moderate	2949 (38.0%)	14.1	13.7	14.5	0.421
Greatest[2]	2549 (32.8%)	13.8	13.3	14.3	—
Anger					
Least	2719 (35.0%)	14.0	13.5	14.4	0.642
Moderate	2319 (29.8%)	14.0	13.6	14.5	0.496
Greatest[2]	2732 (35.2%)	13.8	13.4	14.3	—
Hostility					
Least	2917 (37.5%)	14.2	13.7	14.6	0.408
Moderate	2273 (29.3%)	13.7	13.3	14.1	0.581
Greatest[2]	2580 (33.2%)	13.9	13.4	14.4	—
Total aggression score					
Least	2676 (34.4%)	14.1	13.6	14.5	0.725
Moderate	2669 (34.4%)	13.8	13.4	14.3	0.793
Greatest[2]	2425 (31.2%)	13.9	13.4	14.5	—
Penalties in minutes					
Least	2395 (30.8%)	13.6	13.1	14.0	0.014
Moderate	2538 (32.7%)	14.0	13.6	14.5	0.312
Greatest[2]	2837 (36.5%)	14.3	13.9	14.7	—
Total	7770	14.0	13.9	14.1	—

[1] *P* values reflect significant differences relative to reference category used by random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

CHAPTER V

GENERAL DISCUSSION

Most contemporary research in the field of mild TBI has taken place in laboratory settings. It is often difficult to research this injury in the general population since it occurs in many domains (e.g. motor vehicle accidents, workplace injuries, etc), and is not confined to the sports arena. With recent developments in real-time head impact tracking systems, researchers are now able to better elucidate the issue of mild TBI in the sports arena, with the overall potential of impacting health care change for all patients afflicted with this condition. Most of what clinicians understand about mild TBI is a direct result of what is observed in patients following a mild TBI. Increased symptomatology, decreased postural control, decreased cognitive function, and increased risk of subsequent injuries, are all characteristics of mild TBI researchers can only study once a patient has already sustained the injury. The study of biomechanics and the causes of mild TBI are likely the most important pieces of this clinical puzzle. Understanding these areas will directly improve equipment development, educational interventions, and rule changes, with the emphasis on improving safety in sports and other recreational activities.

To our knowledge, this dissertation was the first prospective study designed specifically to understand the multifaceted nature of descriptive, intrinsic, and extrinsic factors that clinicians have long believed to be associated with mild TBI, and to relate these factors to a sample of young, uninjured ice hockey players. On average, we observed head impacts in our sample of youth ice hockey players to be only slightly lower than what we have observed in Division I collegiate football players (Mihalik, et al., 2007). This may be of concern since the hockey players we have studied are shorter and lighter in stature than the football players we have studied, with the added risk of injury inherent to the developing brain. While we acknowledge little is known regarding the tolerance of the developing brain to head impacts, we postulate children may be less able to withstand repetitive subconcussive head impacts than their adult counterparts (Ommaya, Goldsmith, & Thibault, 2002).

Descriptive factors (Specific Aim 1)

This dissertation endeavored to evaluate how a number of descriptive factors associated with participation in youth ice hockey affect biomechanical measures of head impact severity including linear acceleration, rotational acceleration, and HITsp. These descriptive factors included playing position (defensemen vs. forwards), event type (game/scrimmage vs. practice), location of head impact (i.e. back, front, side, or top), and the striking nature of the subject involved in the body collision (striker vs. player struck).

Effects of player position

We followed 52 players over 2 complete seasons (Bantam = 31; Midget = 21) consisting of a total of 151 games and 137 practices, collecting in excess of 12,400 head impacts. This specific aim builds on our earlier work, where we reported a comparison of linear acceleration across playing position (Mihalik, Guskiewicz, et al., 2008). We found a similar finding in that linear acceleration did not differ between defensemen and forwards in our sample, adding also that no differences existed for measures of rotational acceleration and HITsp across the two playing positions. This agreed with our stated hypotheses. Due to the two-piece design of modern goaltender facial protection, we were unable to instrument a sufficient number of goaltenders in our sample to draw any meaningful comparisons within our positional data. Our study would appear to disagree with another report in which a single high school hockey player was instrumented and observed over a period of 6.52 player-hours (Naunheim, Standeven, Richter, & Lewis, 2000). Of the 161 impacts collected, Naunheim et al. reported a mean peak linear acceleration as high as 35.0 g, which contrasts greatly to the mean we reported in our sample of defensemen (18.3 g). Our data clearly do not seem to support a mean linear acceleration of this magnitude. A number of possibilities exist that may account for these differences. First, our project would appear to have accounted for a much higher sample (N = 19) of defensemen. Secondly, since head impact data are usually heavily skewed toward low-magnitude impacts, we employed the appropriate data transformations necessary to correctly analyze and report our measures. Next, Naunheim et al. did not use accelerometers coupled to the head. Previous research has demonstrated that head acceleration is significantly different than helmet acceleration in helmeted sports (Manoogian, et al., 2006). Previous work in the area of cervical spine injury management in ice hockey has suggested that a very high proportion of ice hockey players do not fit or wear their helmets properly (Mihalik, Beard, Petschauer, Prentice, & Guskiewicz, 2008), suggesting that helmet-coupled accelerometers do very little to measure actual head acceleration. Using head-mounted accelerometers directly in contact with the head provide a more accurate means of measuring in real-time the biomechanical characteristics of head impacts sustained in sports including youth ice hockey. While the mean magnitude reported by Naunheim et al. far exceeds what we report in this study, it should be noted that no impact sustained by the ice hockey player in their study resulted in an observable case of concussion, and no impacts as low as 35.0 g have resulted in a diagnosed concussion in the cases we have observed as part of our ongoing work in this area.

Effects of event type

While we observed statistically significant differences in head rotational acceleration and HITsp between impacts sustained in games and practices agreeing with our hypothesis, we were surprised to have not observed any differences in linear acceleration across event type. The latter opposed our earlier work in which we observed significantly greater linear acceleration in games than practices (Mihalik, Guskiewicz, et al., 2008). In our 2008 work, we studied fourteen Bantam-aged ice hockey players who were not included in the cohorts used for this dissertation. We observed three times more impacts in games and scrimmages (9,343) than in practice (3,115). Those who are opposed to introducing contact into ice hockey at young ages would likely appreciate the fact that our youth ice hockey players do not experience much in way of head impacts during practice. However, it can be argued the observed increase in frequency of head impacts during games and scrimmages, combined with the statistically significant increase in rotational acceleration and HITsp, should prompt coaches to emphasize the education of safe contact drills during practice in order to better prepare their athletes. These findings in event type differences are in contrast to data collected on Division I collegiate football players (Mihalik, et al., 2007), where as many as 77% of head impacts were sustained during practices and linear acceleration during practices were observed to be significantly higher than those sustained in games. Due to high ice rental costs, most youth ice hockey programs do not have an opportunity to practice daily. To date, we have not observed any concussions sustained during practice in the three years we have been studying youth ice hockey. The risk of mild TBI has been reported to be as high as 14.7

and 8.2 times higher during competition than during practice in men's and women's collegiate ice hockey, respectively (Agel, Dick, Nelson, Marshall, & Dompier, 2007; Agel, Dompier, Dick, & Marshall, 2007). Further, the game injury rate in Bantam players was reported to be as high as 10.9 per 1000 player-hours compared to 2.5 injuries per 1000 player-hours during practice (Stuart, et al., 1995). The finding of increased rotational acceleration and HITsp during games, coupled with the increased risk of injury during competition, seems justified by the initial work in this area dating back to Holbourn (1945). Holbourn showed that angular acceleration of the head propagated movements of the brain within the skull, generating shear strains most prominent at the surface of the brain. These shear strains are believed to result in the transient deficits clinicians observe following mild TBI, as opposed to deeper brainstem lesions resulting in more severe forms of TBI not as common during athletic participation. While animals tested in Holbourn's work were subjected to linear accelerations that typically did not result in loss of consciousness, many of them did sustain more serious cortical contusions and subdural hematomae. These results all connect to an important role for rotational movements eliciting an episode of mild TBI. Based on their definition of concussion, Ommaya and Gennarelli found that no observable injuries were produced when isolated linear impacts were imparted to twelve monkeys tested (1974). This was contrasted by thirteen monkeys who experienced a loss of consciousness for periods ranging from 2 to 12 minutes when impacted with their device while in the rotational mode. One of these thirteen monkeys never awoke and two others died within one hour of the impact. The question that still remains elusive to researchers and a matter of contention, for that matter, is "how do the relative contributions of angular and linear accelerations induce mild TBI?" Many factors play a role in the body's ability to mitigate head impact forces including individual differences in CSF levels and function, vulnerability to brain tissue injury, relative musculoskeletal strengths and weaknesses, and the anticipation of an oncoming impact or impulse. The potential effect of musculoskeletal strength is described in further detail later in this chapter, and the effect of anticipation is the focus of Manuscript 1 (see Appendix D). The ability to ever fully study these phenomena in real-time and in-vivo is a quandary that we sought to attain in this dissertation. Regardless of the type, attribute, or severity of a particular impact or impulse, the end result is as follows: the effective mass of the head has become too large for the body to overcome the acceleration or deceleration forces that have sent it in motion.

Effects of head impact location

One of the more interesting and unexpected findings in our analysis of descriptive factors is that of the location of head impact differences. First, we observed a smaller proportion of head impacts sustained to the top of the head (9.6%); we had previously reported as many as 14.2% of impacts occurred to the top of the head (Mihalik, Guskiewicz, et al., 2008). One reason is that we employed an elevation cutoff of 60° in our initial work in this area as opposed to the cutoff of 65° in elevation employed in this dissertation. Had the change not been made, we would have observed 12.7% of impacts occurring to the top of the head, which is more commensurate with our previously published result. Still, a 1.5% difference in our sample represents close to 200 fewer impacts to the top of the head. This decreased number could be the result of including an older subsample (Midget) in our analyses. These older players typically have the advantage of more playing experience and possess the skill required to "keep their head up" while playing ice hockey. Similar to our previous work, and in line with our stated hypothesis, the linear acceleration of head impacts sustained to the top of the head were significantly higher than those to other areas of the head (i.e. back, front, or side). These findings may have some implications to youth ice hockey and other collision sports. First, it reinforces our need to address the instruction of youth ice hockey players in keeping their heads up so that they can become more aware of their surroundings on the ice. Second, the finding also illustrates that an impact to the top of the head results in a greater resultant head linear acceleration than those impacts where players have their heads up. This is in agreement with previous work studying this effect in Division I collegiate American football players (Mihalik, et al., 2007). Given the propensity of using the crown of the head as a weapon in football, a direct comparison of this study and ours is difficult. We speculate that by seeing an impact approach, the player is able to better prepare themselves for the oncoming collision; thus, controlling head and neck movement in a more protective manner (see Appendix D). Contrasting our hypotheses, however, were the results pertaining to significantly lower rotational acceleration and HITsp to the top of the head compared to the other areas of the head. Biomechanically, this finding can likely be explained by the relative lack of an axis of rotation of the head for impacts directed through the top of the head. Impacts sustained to the back, front, or side of the head can more easily yield a rotational movement of the head. For example, an impact to the back of the head can more easily cause a rotation of the head in the anterior direction, and vice versa for impacts sustained to the front of the head. Impacts directly to the side of the head yield angular movements in the same movement direction as the impact, and more easily elicit rotation of the head about the neck. As posterior neck and upper back muscles tend to be much stronger than those of the anterior neck, it is not surprising we observed higher rotational accelerations following impacts to the front of the head compared to those of the back. Weak rotatory muscles of the neck are likely less able to overcome the effective mass of the head in impacts sustained to the sides.

Surprisingly, very little data are available to the effect of impact locations on mild TBI. Hodgson et al. (1983) studied reversible cerebral concussion in the context of head impact location. They imparted impacts to the front, side, back, and top aspects of rigid protective caps worn by six anaesthetized primates using an air-propelled striker. They reported an interesting finding in that impacts to the side produced periods of unconsciousness up to three times longer than loss of consciousness resulting from impacts imparted to the other areas of the head. It is difficult to compare more contemporary literature to our work and that of Hodgson et al. for two primary reasons: there was no follow-up work to this research question by the Hodgson group and modernized research in this area does not provide substantive data for which comparisons would be deemed meaningful. One such study is that of Guskiewicz et al. (2007). This paper employed a real-time helmet accelerometer data collection methodology—similar to the one employed in this dissertationin—in eighty-eight Division I collegiate football players across three playing seasons. The data in that study suggest a higher propensity of top-of-the-head impacts and the relative risk of concussion. In this regard, six of thirteen mild TBI occurred from impacts to the top of the head; this is in contrast to one concussion occurring following impact to the side. The latter injury, however, represented the greatest departures from preseason baseline in symptom score, postural stability, and neurocognitive function. It should be noted that this case represented the second injury sustained by the same athlete that season, and direct comparisons to our data are speculative at best. Guskiewicz et al. (Guskiewicz, et al., 2007) injury data support our findings in that top-of-helmet impacts typically result in relatively lower rotational acceleration values compared to injuries following impacts to the other areas of the head. This information brings into question the notion that rotational acceleration is the leading precursor to injury, and is suggestive that the type of acceleration, in combination with magnitude of impact and impact location, may be a better determinant for both onset and severity of injury.

Effects of striking nature of collision

Our results did not support our underlying research hypothesis that the biomechanical measures of head impact severity would be lower for the striking player compared to the measures we observed when the player who was struck. A number of factors may contribute to these findings. First, regardless of who is striking or being struck, the notion that player anticipation of impending body collisions may play a role in mitigating head impact severity has long been believed, and the discussion of this particular topic can be found in Appendix D. In exploring this theory, we included the interaction effect between level of anticipation and striking nature of the collision in our analysis models and found linear accelerations were not influenced by the level of anticipation in our players (P = 0.170). Those who strike opponents could be considered by many to be more aggressive in nature. Our results pertaining to player aggression would suggest those who are more aggressive tend to experience more severe head impacts following body collision, and may lend support to our finding of higher linear accelerations in the striking player. When we controlled for BMI in our analyses, a surrogate for total cervical muscle strength, our findings did not change. Notwithstanding, the effects of cervical muscle strength, player anthropometrics, general aerobic fitness, and player aggression, are anecdotally believed to mitigate the forces associated with head impacts. These are all further discussed later in this chapter. We should also note that a difference of 1.3 g might not represent a clinical difference, although researchers are far from understanding the cumulative effect of subconcussive impacts on athletes, let alone adolescent athletes.

Extrinsic factor – Game-related exposure (Specific Aim 2)

We observed measures of linear acceleration to be greater in the third period compared to the second and first, which was in support of our research hypothesis. However, no differences existed between the first and second period, challenging our hypothesis. No differences existed between period of play for measures of rotational acceleration or HITsp. Given the evidence provided earlier suggesting rotational mechanisms result in more serious injuries than linear mechanisms, our period data suggest a non-increasing trend in injury risk over the course of the game. Our data suggest a similar mean number of playing shifts (5.6 vs. 5.4 vs. 5.4) per player across the three playing periods. While factors associated with fatigue could possibly explain an increase in the number of playing shifts in the third period (i.e. players are participating in more frequent, but shorter shifts), we did not observe this. We also included player BMI as a covariate, expecting that by controlling for a player's relative physique we would see changes in our original results. Player BMI did not affect our results. Overall, our analyses on cervical muscle strength (see Specific Aim 5 below) did not suggest an effect of muscle strength on mitigating the head impact severity following collisions in youth ice hockey. Since period of collision was collected with our retrospective sample, and cervical muscle strength with our prospective sample, we were unable to draw these into a single statistical model for direct comparisons. While we cannot neglect the effect of general fatigue that takes place over the course of the game, we speculate that as athletes tire, they are less likely to keep their head up and resort to lazy technical skills during play. This would align with our data described above in our discussion of significantly higher linear accelerations occurring to the top of the head.

To our knowledge, no one has published any work studying the short-term (i.e. one game) dose-response of game-related exposure on biomechanical measures of head impact severity. We sought to explore this in a number of different ways. First, we analyzed our measures of head impact severity by controlling for the absolute number of playing shifts the athlete participated in during the period the impact was sustained. We found that this did not affect our biomechanical measures. Next, we arbitrarily weighted the number of playing shifts such that those occurring in the period of the impact were weighted as “1” and those occurring in the preceding period were weight as “0.5.” For those occurring two periods preceding the period of collision (i.e. impacts in the third period), those shifts were weighted as “0.25.” While we felt this method would more adequately represent the cumulative nature of game-related exposure, this covariate did not identify any differences in our biomechanical measures of head impact severity. Next, we explored the potential for controlling for the average number of playing shifts for all periods up to and including the period in which the collision took place. We found that those who represented the highest average number of shifts played experienced higher rotational acceleration measures than those who represented the moderate and low average number of shifts played. It should be noted than approximately 55 rad/s^2 separates the lowest from the highest, and would be deemed clinically insignificant especially given our other non-significant findings pertaining to this specific aim. Lastly, the weighted average number of playing shifts (using the same weighting system, but dividing the sum by the total weights) did not yield any statistically significant findings. In addition, we performed a number of confirmatory interaction analyses to identify whether the effect of playing period was influenced by the game-related exposure. None of the interactions were significant, regardless of which of the four control variables we employed in our model, and indifferent across our three biomechanical measures of head impact severity. Our extensive testing of the data suggests that the number of playing shifts in a game does not lead to higher impact severity. Practically, those athletes who participate in an increasing number of playing shifts tend to be the better players on the team. They are usually attributed this role because they are more skilled and in a better state of fitness as a result. Injury researchers believe that the more a person is exposed to a potential event, the higher their risk of experiencing the event. Though this may be argued, perhaps more skilled players who are on the ice more often possess a skill set superior to their teammates who play less frequently. This skill set includes increased aerobic fitness, technical skill (i.e. skating with their head up), and possess an awareness of on-ice presence. These are all believed by coaches and players to reduce the risk of injury in ice hockey players, and may offset the increased exposure to a potential injury.

Intrinsic factors – strength, anthropometrics, fitness, and aggression (Specific Aim 5)

We sought to identify the effects of a number of intrinsic factors on biomechanical measures of head impact severity. The primary factors studied included body collision type (open-ice vs. along playing boards), anticipation of body collision, and relative body position. These are included in Manuscript 1 (Appendix D). Other intrinsic factors of interest included cervical muscle strength, anthropometrics, general aerobic fitness, and player aggression. These will be discussed further in the ensuing sections.

Cervical muscle strength

We evaluated the strength of cervical muscles using methods consistent with how these muscle groups are tested clinically (Hislop & Montgomery, 1986; Kendall, et al., 1993). Specifically, we evaluated anterior neck, anterolateral neck, posterolateral neck, cervical rotation, and upper trapezius muscle strength. To provide a comparative analysis, we categorized each strength measure into three ordinal categories: weak, moderate, and strong. Our results did not support our research hypothesis. That is, we did not find any differences between weak, moderate, and strong athletes in our sample for linear or rotational acceleration. Additionally, no differences across these tertile groups were observed for anterior neck, anterolateral neck, cervical rotation, and posterolateral neck strength. Surprisingly, subjects representing those with the weakest upper trapezius muscles experienced lower HITsp measures than those representing the strongest in our sample. The stronger players may be more likely to attempt open-ice collisions, which were shown in Appendix D to result in greater rotational acceleration and HITsp measures than impacts along the playing boards. The argument could be made that the players were likely in the "strong" group because they were bigger and heavier, and more likely to sustain a higher magnitude impact as a result. However, all strength measures—recorded in kilograms—were normalized to the athletes' body mass, adjusting these values for differences in player mass. Our findings are surprising, especially given the strong anecdotal support for cervical muscle strength as a factor in mitigating head impact severity. The basic tenet of the neck muscle theory is that an athlete who anticipates an oncoming collision will be better able to control head movement by contracting (i.e. tensing) their cervical musculature. When the cervical musculature is contracted, it is thought to significantly increase the effective mass of the head-neck-trunk segment, resulting in a lower acceleration of the head. When an impact is unanticipated, and the cervical musculature is not tensed and the athlete is unprepared for a collision, the effective mass is reduced to that of the head. Given an equal force from a body collision, the head would experience a substantially greater acceleration and, therefore, more likely to sustain an injury. While this seems rather intuitive, there is very little research to corroborate these beliefs. There is a total lack of this research in the area of ice hockey. Studies in this area have focused primarily on a soccer-heading task and do not serve as a particularly strong foundation for comparison (Mansell, et al., 2005; Tierney, et al., 2005).

Effect of player anthropometrics

Our hypothesis that taller and heavier players would sustain higher biomechanical measures of head impact severity was not supported by our data. No differences were observed for linear acceleration and HITsp, and the heaviest players sustained higher rotational accelerations, on average, than the lightest players. In retrospect, this is not surprising. The mass disparities between opponents—and within the same team—can range by as much as 48 kg (105.6 lbs). Given this, heavier players may perceive less of a threat from the other players on the ice, and be less likely to "tense up" during a collision as their lighter counterparts may be. The implication of player size has long been a concern, especially at the Bantam age level. While there tends to be minimal disparity in the height and mass of players within the Peewee (11 and 12 years of age) and Midget (15 and 16 years of age) playing levels, this trend is not shared among Bantam-aged players. Previous work has established anthropometric and biomechanical force profiles for each player in a sample of youth ice hockey players (Bernard, et al., 1993). When the authors simulated body checking between the smallest and largest players, they observed a 357% difference in the force of impact. However, these differences were not observed in this dissertation. The Canadian Academy of Sports Medicine notes that serious injury in ice hockey escalates in Bantam-aged players, further asserting body checking should not be allowed because of the differences in body size between players (Sullivan, 1992). In a study of children's ice hockey injuries presented earlier in this literature review (Brust, et al., 1992), more than half of the injuries occurred at the Bantam level (54%) compared to younger players in Peewee (27%) or Squirt (19%). They posit the reason for this trend to be a difference of as much as 53 kg (116.6 pounds) and 55 cm (21.65 inches) between players on Bantam teams in this study. In our study, we observed differences within Bantam-aged players to be as high as 30.5 kg (67.1 pounds) and 30 cm (11.8 inches).

While we did not hypothesize any differences across the other head and neck anthropometrics, there are some findings worthy of discussion. In agreement with the aforementioned explanations, all significant findings were such that measures representing the smallest players in our sample suggest linear and rotational acceleration, and HITsp are lower than those players representing the biggest in our sample. This was especially true of head anthropometrics, whereby we noted those with the smallest head circumference and the narrowest anterior-posterior and medial-lateral diameters sustained lower rotational accelerations than those representing the largest and widest of these anthropometric measures, respectively. Speculating those with smaller heads, as evidenced by shorter head circumferences and narrower diameters, have less head mass to overcome following a head impact, it is possible they are able to use their neck muscles to counter the effective mass that has been set in motion. As head and neck anthropometrics are genetic and we are unable to modify this factor, it is difficult to determine how to apply these findings clinically.

General aerobic fitness

Our hypothesis that increased general aerobic fitness would result in lower biomechanical measures of head impact severity was not supported. While our results suggest those with the least aerobic fitness experience lower linear and rotational accelerations than those who were deemed most fit, differences in linear acceleration did not exceed 1.0 g or 200 rad/s^2. These findings are in line with those reported for Specific Aim 2 (game-related exposure), where athletes with the lowest average number of shifts played experience lower rotational accelerations than those who played the highest average number of shifts. The importance of superior aerobic conditioning may not be as prevalent in a single event, but may play an important role in the last of five games held over a period of 2 days. Long-term prospective evaluations of general aerobic fitness in conjunction with the number of games played in a brief period of time (i.e. tournaments) are warranted to better understand the effect of general aerobic fitness.

Player aggression

Although we originally were going to evaluate trait aggressiveness with the Buss-Perry Aggression Questionnaire (Buss & Perry, 1992), we also opted to include penalties in minutes (PIM) as a measure of aggression commonly used in ice hockey to represent aggressive players. In support of our hypothesis, we observed more severe head impacts in players exhibiting a greater number of penalties in minutes than those who did not (i.e. more disciplined players). Our findings are in support of previous work that used the BPAQ to predict aggressive penalty minutes over the course of the season (Bushman & Wells, 1998). While the immediate purpose of our study was not to understand coaches' and players' aggressive behavior, it is difficult to ignore this aspect of ice hockey. The culture of ice hockey predicates a mentality among players to ignore injury, play recklessly, and encourages unsportsmanlike conduct such as fighting and illegal checking. In the United States, a study of Peewee-level players reported that fighting broke out in approximately 17 of 52 games observed. In that sample, players considered fighting a natural consequence of the game and experienced a certain resignation about fighting (Gerberich, et al., 1987). Another interesting finding reported by Brust et al. is that while 100% of coaches felt sportsmanship was "real important," only 59% of players shared this attitude (Brust, et al., 1992). Parents and coaches, in this sample, viewed the enforcement of rules as being the most important factor in reducing injuries.

Limitations

A discussion of this project would not be complete without a disclosure of some of the limitations. While no direct intervention was established on our part, the players knew they were wearing specially instrumented helmets. This may have altered their normal style of play knowing their impacts would be recorded. However, we recorded head impacts across every on-ice session over a period of two years and feel that whatever short-lived alterations in playing style would have been washed out by the high number of events recorded. Secondly, our sample was comprised of 13- to 16-year-old male ice hockey players only. As such, they may not be representative of all youth ice hockey players, and these results may not be related to collegiate and professional ice hockey players. Relating these findings to women's ice hockey, where bodychecking is not permissible, may be difficult. As we did not capture cervical muscle strength, player anthropometrics (aside from height and mass), general aerobic fitness, or aggression measures in our Year 1 sample, we were unable to include these variables into our explanatory statistical models in order to better explain our findings. We did find a moderate relationship ($r = -0.618$) between BMI and cervical muscle strength in our Year 2 cohort and, thus, used BMI as a covariate in the models associated with our Year 1 data analyses. We did not study the effect of any of these factors on actual injury incidence observed in our sample. During the course of the two-year period, we observed six separate instances of concussion, a number too few for any relevant analyses and comparisons. It will require significantly more injuries to substantiate the specific biomechanical factors associated with sports-related concussion in youth athletes.

Conclusions

Overall, the sport of youth ice hockey is a relatively safe sport, with only six injuries resulting from over 12,000 head impacts sustained by the hockey players in our sample. However, our findings suggest that collisions involving infractions still remain a problem in this sport. We observed as many as 17% of collisions resulting in some form of illegal infraction, primarily elbowing and head contact penalties. As these infractions resulted in the greatest measures of linear acceleration, rotational acceleration, and severity profile (HITsp), there should be concern from policy makers as to how we can best reduce the frequency of these collisions with the goal of reducing the incidence of head injuries related to these infractions. The notion that heightened player anticipation can mitigate the severity of head impacts seems supported by our data, especially as it pertains to measures of rotational acceleration sustained during collisions in the moderate range of intensity (50th to 75th percentile). Additionally, the results of this dissertation have implications to the current helmet testing standards, as our descriptive factors identified differences in linear and rotational accelerations depending on where the head impact occurred on the helmet. For example, while linear accelerations were greatest at the top of the head relative to the back, front, and sides, rotational accelerations were greatest at the sides. As each location and acceleration-type combination represent potential for injury, helmet standards must be modified to evaluate the efficacy of minimizing all types of acceleration at multiple head impact locations. In addressing these issues, testing standards can be modified to improve the efficacy of equipment to reduce injury, and may be the simplest manner in which widespread injury prevention can infiltrate athletes at the grass roots level. Prospective studies evaluating the effect of player education and technical training on reducing injury rates at the youth level should be undertaken. As no formal cervical strength training program was carried out by the athletes in our sample, it will be important to further understand the effects of cervical muscle strength on mitigating the severity associated with head impacts in youth ice hockey, and to do so in a controlled and prospective manner. Prior to implementing any long-term educational interventions in this young population, we must first evaluate the health behavior models associated with hockey players at the amateur level. In so doing, we will be better able to implement interventions with the end of goal of increasing athlete reporting of head injuries to their coaches and parents, which will reduce the risk of complications associated with secondary impacts. While this study did not investigate injuries, future work will better elucidate how the descriptive, intrinsic, and extrinsic factors studied in this dissertation affect outcome following mild traumatic brain injury.

Appendix A. Game Shift Log

Start time ________________ Stop time ________________

Game #: _____ Date (mm/dd/yyyy): ______________ Opponent: ________________ Location: _________________

Jersey #	1st period	1st Total	Jersey #	2nd period	2nd Total	Jersey #	3rd period	3rd Total
Forwards								
7			7			7		
9			9			9		
11			11			11		
17			17			17		
24			24			24		
33			33			33		
40			40			40		
87			87			87		
96			96			96		
Defensemen								
8			8			8		
15			15			15		
16			16			16		
22			22			22		
32			32			32		
55			55			55		
Goaltenders		**Length of period:**		**Length of period:**			**Length of period:**	
30			30			30		
62			62			62		

Technical difficulties with HIT System? YES / NO

If yes, please specify:

Appendix B. Carolina Hockey Evaluation of Children's Checking (CHECC) List

Draft

Carolina Hockey Evaluation of Children's Checking (CHECC) List

Subject ID ☐☐☐☐

Rater: ☐☐☐

Date of collision

☐☐☐☐ / ☐☐ / ☐☐

(year) (month) (day)

(example: 01 = January. . . 12 = December)

Time of collision

☐☐ : ☐☐ : ☐☐

(HH) (MM) (SS)

(example: 01=1:00AM; 15=3:00PM)

C1. Player looking ahead in direction of movement

☐ Yes (0) ☐ No (1)

C2. Player appears to be looking in direction of impending body collision

☐ Yes (0) ☐ No (1)

C3. Knee flexion angle at time of collision **>30 degrees**
Knees are flexed

☐ Yes (0) ☐ No (1)

C4. Trunk is flexed at time of collision

☐ Yes (0) ☐ No (1)

C5. Player drives into collision with shoulders

☐ Yes (0) ☐ No (1)

C6. Player uses elbow(s) in collision

☐ Yes (1) ☐ No (0)

C7. Player uses hands in collision

☐ Yes (1) ☐ No (0)

C8. Feet are shoulder width apart at time of collision

☐ Yes (0) ☐ No (1)

C9. Player uses stick during collision

☐ Yes (1) ☐ No (0)

C10. Player uses legs to drive into/through body collision

☐ Yes (0) ☐ No (1)

C11. Was the player receiving/delivering a pass at time of collision?

☐ Yes (1) ☐ No (0)

C12. **Overall impression of body collision**

☐ Anticipated, with good body position (0)

☐ Anticipated, with poor body position (1)

☐ Unanticipated (2)

Descriptive epidemiology:

C13. Player involvement in body collision

☐ Striking player (0) ☐ Player struck (1)

Extrinsic factors:

C14. Infraction type associated with collision

☐ Legal (clean) collision (0)

☐ Boarding/Charging(1)

☐ Check from behind (2)

☐ Elbowing/Head contact (3)

Intrinsic factors:

C15. Location of body collision

☐ Open-ice collision (0)

☐ Along playing boards (1)

Submit

Appendix C. The Buss-Perry Aggression Questionnaire

Please read the following 29 statements carefully. Circle the response you feel best represents your true feelings. Your options are *Extremely uncharacteristic of me*, *Somewhat uncharacteristic of me, Neither uncharacteristic nor characteristic of me, Somewhat characteristic of me,* or *Extremely characteristic of me*.

	The statement is…				
	Extremely uncharacteristic of me	*Somewhat uncharacteristic of me*	*Neither uncharacteristic nor characteristic of me*	*Somewhat characteristic of me*	*Extremely characteristic of me*
1. Some of my friends think I am a hothead.	1	2	3	4	5
2. If I have to resort to violence to protect my rights, I will.	1	2	3	4	5
3. When people are especially nice to me, I wonder what they want.	1	2	3	4	5
4. I tell my friends openly when I disagree with them	1	2	3	4	5
5. I have become so mad that I have broken things.	1	2	3	4	5
6. I can't help getting into arguments when people disagree with me.	1	2	3	4	5
7. I wonder why sometimes I feel so bitter about things.	1	2	3	4	5
8. Once in a while, I can't control the urge to strike another person.	1	2	3	4	5
9. I am an even-tempered person.	1	2	3	4	5
10. I am suspicious of overly friendly strangers.	1	2	3	4	5
11. I have threatened people I know.	1	2	3	4	5
12. I flare up quickly but get over it quickly.	1	2	3	4	5
13. Given enough provocation, I may hit another person.	1	2	3	4	5

	The statement is…				
	Extremely uncharacteristic of me	*Somewhat uncharacteristic of me*	*Neither uncharacteristic nor characteristic of me*	*Somewhat characteristic of me*	*Extremely characteristic of me*
14. *When people annoy me, I may tell them what I think of them.*	1	2	3	4	5
15. *I am sometimes eaten up with jealousy.*	1	2	3	4	5
16. *I can think of no good reason for ever hitting a person.*	1	2	3	4	5
17. *At times I feel I have gotten a raw deal out of life.*	1	2	3	4	5
18. *I have trouble controlling my temper.*	1	2	3	4	5
19. *When frustrated, I let my irritation show.*	1	2	3	4	5
20. *I sometimes feel that people are laughing at me behind my back.*	1	2	3	4	5
21. *I often find myself disagreeing with people.*	1	2	3	4	5
22. *If somebody hits me, I hit back.*	1	2	3	4	5
23. *I sometimes feel like a powder keg, ready to explode.*	1	2	3	4	5
24. *Other people always seem to get the breaks.*	1	2	3	4	5
25. *There are people who pushed me so far that we came to blows.*	1	2	3	4	5
26. *I know that "friends" talk about me behind my back.*	1	2	3	4	5
27. *My friends say that I am somewhat argumentative.*	1	2	3	4	5
28. *Sometimes I fly off the handle for no good reason.*	1	2	3	4	5
29. *I get into fights a little more than the average person.*	1	2	3	4	5

Appendix D. Manuscript 1

Title:

The effect of collision type and player anticipation on head impact severity in youth ice hockey players

Target journal:

Pediatrics

The effect of collision type and player anticipation on head impact severity in youth ice hockey players

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Short title: anticipation and head impacts in youth hockey

Abbreviations: Carolina Hockey Evaluation of Children's Checking (CHECC); Traumatic brain injury (TBI)

Keywords: child, concussion, injury, physical activity, sports, trauma

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Conflicts of interest: None declared

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ABSTRACT

Objective: The objective of the study was to determine the effects of body collision type and level of player anticipation on mitigating the severity of head impacts sustained by youth ice hockey players. In anticipated collisions, we also sought to identify an optimal body position for delivering and receiving body collisions. We hypothesized open-ice collisions would be more dangerous than collisions along the playing boards, and that anticipated collisions would be less severe than unanticipated collisions with respect to impact biomechanics.

Patients and Methods: The prospective study included sixteen male Bantam-aged ice hockey players (age = 14.0 ± 0.5 years) equipped with helmets instrumented with accelerometers to record biomechanical measures (i.e. linear acceleration, rotational acceleration, severity profile) associated with head impacts in real-time. Body collisions observed through video footage captured over a 54-game season were evaluated for collision type (open ice vs. along playing boards), level of anticipation (anticipation vs. unanticipation), and relative body positioning.

Results: The linear acceleration ($P = 0.036$) and rotational acceleration ($P = 0.003$) of collisions occurring in the open ice were significantly greater than those occurring in collisions along the playing boards. Though not statistically significant, our data are suggestive of a trend such that anticipated collisions resulted in less severe head impacts than unanticipated collisions.

Conclusions: Our data support the notion that anticipating collisions may play a role in minimizing head impact severity. We found that the differences in impacts occurring along the playing boards and those occurring in the open ice, in combination with our findings that unanticipated impacts result in higher impact forces, represent a continued need to educate our players with the necessary technical skills needed to heighten their awareness on the ice.

Clinically, coaches and athletes should incorporate bodychecking exercises in practices, and spend time educating young athletes on these proper checking techniques in order to minimize the risk of injury and increase the safety of ice hockey.

INTRODUCTION

The Centers for Disease Control and Prevention (CDC) have acknowledged traumatic brain injury (TBI) to be a serious public health problem in the United States. Children less than 15 years of age represent as much as 40% of the 1.1 million TBIs resulting in emergency department visits each year.[1] Perhaps more alarming is the actual number of young TBI victims who do not seek out evaluation for their injuries and are not seen in emergency departments. In addition to representing one of the most difficult conditions to manage in sports medicine, TBIs were estimated in 2000 to account for over $60 billion in direct medical and indirect costs.[2]

Youth athletes participating in collision sports are at particular risk for sustaining concussion and other forms of mild TBI. First, technical development is often limited since parent volunteers often serve as coaches and lack the skill and training to adequately educate athletes on proper collision techniques. Medical supervision of these young athletes is also lacking in comparison to the medical personnel involved with collegiate and professional athletics. Second impact syndrome—a condition resulting in immediate brainstem swelling when an athlete sustains a head impact, often minor, when symptoms associated with an initial TBI have not fully resolved—has only been reported in the adolescent athlete and has typically resulted in cases where TBI has been mismanaged by parents, coaches, or medical personnel. Understanding the nature of head impacts sustained in youth athletics, with an emphasis on improving our comprehension of how best to mitigate the severity of head impacts, may lead to interventions directly related to reducing the incidence of TBI in our youth, minimizing the risks for second impact syndrome, and decreasing the financial costs incurred by our medical system.

Therefore, the purpose of this investigation was to study the effect of body collision type (i.e. open-ice vs. along playing board) and player anticipation (i.e. anticipated vs. unanticipated) on biomechanical measures of head impact severity including linear acceleration, rotational acceleration, and severity profile. A secondary purpose was to identify whether different relative body positions we observed resulted in lower biomechanical measures of head impact severity.

PATIENTS AND METHODS

Study Design and Participants

This study employed a prospective quantitative research design in order to evaluate the effect of body collision type and level of player anticipation on biomechanical measures of head impact severity in youth ice hockey players. We recruited sixteen male Bantam-level ice hockey players (age = 14.0 ± 0.5 years; height = 171.3 ± 4.5 cm; mass = 63.7 ± 6.6 kg) representing a convenient sample of participants from a AAA-level program. Our sample included nine forwards and six defensemen; one goaltender was removed from our analyses due to lack of body collisions. Data were collected during 54 games over the course of the season; data from 38 practices were not included since these events were not captured on video. Regardless of previous history of concussion or years of playing experience, none of the athletes were excluded from participating in the study. Parental permission and minor assent forms approved by the university's institutional review board were signed by each parent and player, respectively, prior to fitting an athlete with an instrumented ice hockey helmet (see Procedures section).

Instrumentation

Head Impact Telemetry (HIT) System. This study used commercially available Reebok RBK 6K and 8K helmets (Reebok-CCM Hockey, Inc.; St-Laurent, Quebec, Canada) modified to accept the Head Impact Telemetry (HIT) System technology (Simbex; Lebanon, NH). The helmet's foam liner was modified to accept six single-axis accelerometers, a battery pack, and the telemetry instrumentation (***Figure 1***). The custom helmets passed American Society for Testing and Materials (ASTM 1045-99) and Canadian Standards Association (CSA Z262.1-M90) helmet standards, and were approved by the Hockey Equipment

Certification Council (HECC) for use during competition. This instrumentation has been previously described in detail.[3] The head impact data were time-stamped, encoded, stored locally, and then transmitted in real time to the Sideline Response System (Riddell; Elyria, OH) via a radiofrequency telemetry link. The Sideline Response System was typically positioned along the playing surface sideboards or in the team's dressing room. The HIT System is capable of transmitting accelerometer data from as many as 100 players over a distance well in excess of the length of a standard international ice surface.

Carolina Hockey Evaluation of Children's Checking (CHECC) List. We developed the Carolina Hockey Evaluation of Children's Checking (CHECC) List to enable the use of a standardized evaluation rubric to be used when analyzing video footage of body collisions. The CHECC List is scored on 11 readily observable features (see **Table 1**) of human movement when involved in a body collision. In addition to these, whether the collision occurred in the open ice or along the playing boards, and whether the collision was anticipated or unanticipated were also variables of interest included on the CHECC List. We further distinguished the level of anticipation into the following overall impressions: anticipated in a good body position, anticipated in a poor body position, and unanticipated. Intrarater Kappa agreements ranged from 0.40 to 0.92 for the 15-item CHECC List. Interrater agreement suggested moderate to strong agreement between hockey coaches with no scientific experience. The range of agreement was 60% to 96% in which at least five of six coaches agreed on each individual criterion.

Video recording. Prior to each game, we synchronized the date and time on our video camera with the date and time on the Sideline Response System. This allowed us to accurately align body collisions observable on the video footage with the biomechanical measures of head impact severity recorded by our instrumented helmets. We recorded video footage for all 54 games over the course of the season. In an attempt to maximize player size in our footage and capture of collisions we could later analyze, we followed the movement of the puck carrier. As a result, some impacts occurred outside the view of the camera and were excluded from our analyses. In order to maximize length of video capture on our miniDV tapes, we began filming with the start of play, and paused recording when the official blew the whistle signaling a stop in the play. In some instances, late body collisions occurring shortly after the end of a play were sometimes not captured and were subsequently unable to be analyzed in our study. All raw videos were imported into a personal computer where the date and time-stamp information were inlayed onto the videos. The videos were then exported into a compressed format and later played back using QuickTime Player Pro (Version 7.5.5; Apple Inc.; Cupertino, CA) for evaluation via the CHECC List.

Procedures

Prior to the start of the season, players were measured for helmet and facemask size. They were properly fit with Reebok RBK 6K/8K helmets (and their personal facemasks) by a certified athletic trainer (ATC). The ATC instructed each participant to wet his hair to simulate sweating, and ensured that the facemask chinstrap was securely fastened to the helmet and fit tightly under the chin. Helmet and facemask fit was verified on a biweekly basis to ensure proper fit throughout the course of the season.

The raw head impact data collected over the course of the season were exported from the Sideline Response System into Matlab 7 (The Mathworks, Inc.; Natick, MA), where data were reduced to include only those impacts sustained during games and scrimmages. Impacts occurring outside of team-sanctioned events were omitted from our analyses. As impacts below 10 g of linear acceleration (measured in terms of gravity force, g) are considered negligible with respect to impact biomechanics and their relationship to head trauma, only impacts registering a linear acceleration greater than 10 g were included for the purposes of our analyses. Our biomechanical measures of head impact severity consisted of linear acceleration, rotational head acceleration (measured in rad/s^2), and Head Impact Technology severity profile (HITsp). The HITsp is a weighted composite score including aspects of linear and rotational acceleration, as well as impact location, and has previously been described in more detail.[4]

Statistical analyses

Head impact data were transformed using a natural logarithmic function to address the skewness of the data toward low-magnitude head impacts. All estimates were then back-transformed to their original scale for presentation and interpretation. Descriptive analyses (means and 95% confidence intervals) were calculated for resultant linear acceleration, resultant rotational acceleration, and the HITsp. In order to address our study purpose, separate random intercepts general mixed linear models were employed for each of these dependent variables. Body collision type and level of anticipation were included as separate independent variables (in addition to player) in the statistical models employed in this study. "Player" represented one level in each statistical model as a repeated factor. In addition to understanding the effects of body collision type and level of anticipation associated with head impact severity, we also sought to identify how relative body positioning affected head impact measures. As a result, we performed separate analyses that included each single relative body descriptor as an independent variable in separate random intercepts general mixed linear models. All random intercepts general mixed linear models (PROC MIXED) were performed in SAS/STAT (Version 9.1; SAS Institute, Inc.; Cary, NC). The level of significance was set at $P < .05$ *a priori*.

RESULTS

We observed a total of 666 body collisions for which we were able to complete a CHECC List and evaluate whether the collision took place along the boards or in the open ice, judge whether the collision was anticipated, and determine relative body positioning of these impacts. Of these collisions, 63.3% (421 of 666) took place along the playing boards, while the remaining 36.8% (245 of 666) occurred in the open ice. Anticipated collisions accounted for 84.7% (564 of 666) of the bodychecks we observed, while the remaining 15.3% (102 of 666) were deemed to be unanticipated collisions. We further categorized anticipation into an overall impression score assigning a body collision to one of the following three levels of anticipation: anticipated collision (with a good relative body position), anticipated collision (with a poor relative body position), and unanticipated collision. Of these collisions, 47.3% (315 of 666) were anticipated with a good relative body position, 37.4% (249 of 666) were anticipated with a poor relative body position, while the remaining 15.3% (102 of 666) were deemed to be unanticipated collisions.

Body collision type

We observed a statistically significant difference in head linear acceleration in impacts sustained along the playing boards compared to those sustained in the open ice ($F_{1,14} = 5.40$, $P = 0.036$). Linear accelerations sustained from open-ice collisions were significantly greater than those sustained from collisions along the playing boards (**Table 2**). However, the rotational acceleration measures for open-ice collisions were significantly greater than those we observed for collisions along the playing boards in our sample ($F_{1,14} = 12.75$; $P = 0.003$), and these data are reported in **Table 3**. With respect to the HITsp, a weighted composite score including aspects of linear and rotational acceleration, as well as impact location,[4] the data (**Table 4**) suggest a strong trend towards a significant difference between open-ice collisions and those occurring along the playing boards ($F_{1,14} = 4.38$; $P = 0.055$).

Level of anticipation

Though linear accelerations tended to be greater in unanticipated collisions compared to anticipated collisions, the differences we observed were not statistically significant ($F_{1,14} = 2.52$; $P = 0.135$). A similar trend was observed such that anticipated collisions resulted in lower rotational head accelerations than unanticipated, but these differences were not statistically significant ($F_{1,14} = 2.47$; $P = 0.138$). No significant differences in the HITsp between anticipated and unanticipated collisions were observed ($F_{1,14} = 0.10$; $P = 0.755$).

Linear accelerations were greatest under unanticipated conditions, followed by anticipated impacts with poor positioning, and anticipated collisions with good positioning, respectively, but these differences were not statistically significant ($F_{2,28} = 1.46$, $P = 0.249$) with a 2,28 degree of freedom mixed model. Since the data suggested the trend just described, we subsequently performed a 1 degree-of-freedom linear trend, observing suggestive evidence of a trend in our data ($F_{1,649} = 2.55$; $P = 0.111$). Likewise, a significant difference in rotational head accelerations across anticipation type was not observed ($F_{2,28} = 1.24$, $P = 0.304$). A linear trend in the mean data, however, is suggestive of lower rotational acceleration measures in anticipated collisions where players were in a good position to deliver or sustain the impact, compared to impacts that were anticipated in which the player was in a poor position and unanticipated collisions. No significant differences in the HITsp across the anticipation types were observed ($F_{2,28} = 0.70$, $P = 0.503$). After controlling for body mass index (BMI), we did not see any alterations in our original findings. As BMI is moderately related to measures of cervical muscle strength (r = -0.618) in our ongoing work (but not available for this current study), we felt this would represent a good surrogate for cervical muscle strength in our study.

Additionally, we explored a number of different impact ranges, particularly those between the 25th to 75th percentile of linear acceleration measures, as well as those between the 50th and 75th percentiles. For the former, we observed a statistically significant difference in linear acceleration ($F_{2,27} = 4.29$; $P = 0.024$), such that anticipated—good collisions (18.7 g; 95% CI: 18.0-19.4) were significantly lower than unanticipated (19.9 g; 95% CI: 19.1-20.7) body collisions ($P = 0.007$). We evaluated those collisions occurring between the 25th to 75th percentile of HITsp measures, as well as those between the 50th and 75th percentiles. For the latter, we observed a significant difference in rotational acceleration ($F_{2,19} = 6.83$; $P = 0.006$), such that impacts from anticipated—good (1215.11 rad/s^2; 95% CI: 1112.6-1327.1) and anticipated—poor (1218.9 rad/s^2; 95% CI: 1107.2-1341.9) collisions were significantly lower than unanticipated collisions (1465.7 rad/s^2; 95% CI: 1240.7-1731.4). We also observed a significant difference in HITsp ($F_{2,19} = 4.35$; $P = 0.028$), such that impacts from anticipated—good (15.2; 95% CI: 15.0-15.5) and anticipated—poor (15.3; 95% CI: 15.1-15.5) collisions were significantly lower than unanticipated collisions (15.6; 95% CI: 15.3-15.9). All other analyses did not yield any statistically significant findings ($P >$ 0.05).

Relative body positioning

Analyses of body positioning during collision revealed that athletes who drive into or through a body collision with their legs (20.5 g; 95% CI: 19.2-21.9) experience lower linear accelerations than athletes who do not use their legs (21.7 g; 95% CI: 20.1-23.5) during a collision ($F_{1,13}$ = 4.67; P = 0.049). No differences were noted for rotational acceleration ($F_{1,13}$ = 0.62; P = 0.446). A moderate trend observed for the HITsp ($F_{1,13}$ = 3.47; P = 0.085) suggests that athletes who use their legs to drive through a body collision (15.3; 95% CI: 14.6-16.1) experience lower severity profiles than those instances in which athletes do not use their legs to drive through a collision (16.0; 95% CI: 15.1-16.9). All other comparisons evaluating the effect of relative body positioning did not result in any statistically significant differences ($P > 0.05$).

DISCUSSION

To our knowledge, this is the first study focused on the biomechanical measures associated with head impact severity in youth ice hockey players, especially as it pertains to player anticipation, body collision type, and relative body positioning. The findings suggest head impact severity decreases with heightened player anticipation. It has long been believed by clinicians that athletes who are more aware of their surroundings and better at anticipating and preparing for impending body collisions would best be able to mitigate the forces to the head associated with those collisions. Surprisingly, there remains very little known about the types of forces that cause mild TBI and, perhaps alarmingly, very few suggested methods to reduce head impact forces. Cantu suggests there are five areas that can result in a reduction in the incidence of mild TBI: changes in rules and coaching technique, improvements in conditioning and equipment, and increasing medical supervision.[5] Of interest are our findings that differences in rotational acceleration between unanticipated and anticipated collisions in the mid-intensity range (defined as those impacts between the 50th and 75th percentile of HITsp) were greater than differences we observed in the high- and low-intensity ranges. This elucidates concern clinicians may have with impacts occurring within this intensity level. First, it highlights that severe impacts (those in the top 25th percentile) are perhaps equally dangerous regardless of anticipation. Second, in impacts that may not appear "severe" to coaches, parents, and other players, there is evidence to suggest that player anticipation may mitigate the severity of head impacts that may otherwise cause injury. Thus, anticipating collisions for these moderate-severity impacts may play a role in reducing risk of injury. While it has not yet been shown that the brain can be conditioned to accept repeated blows, it is believed the neck can be strengthened and the risk of mild TBI reduced. This is true only in cases where collisions are anticipated and the athlete is able to tense up their cervical musculature. Our data would support this plausible explanation, as unanticipated collisions yielded a trend towards higher linear and rotational accelerations compared to those collisions that were anticipated. Future research in this area by other independent researchers will be important in building on this area of research. Using a simple Newtonian approach, acceleration is the result of force divided by mass. When the cervical musculature is contracted, it is thought to significantly increase the effective mass of the head-neck-trunk segment, resulting in a lower acceleration of the head. When an impact is unanticipated, and the cervical musculature is not tensed and prepared for a collision, the effective mass is reduced to that of the head. Given an equal force from a body collision, the head would experience a substantially greater acceleration and, therefore, more likely to sustain an injury. Although this may seem rather intuitive, studies in the area of event anticipation have focused primarily on a soccer-heading task in collegiate athletes.[6, 7] Results have been ambiguous in this population so far, and extrapolations to a collision sport such as ice hockey and a younger population are very difficult. While we did not control for cervical muscle strength in our analyses, our data from ongoing work suggest a moderate relationship exists between BMI and total cervical muscles strength. As such, we felt that BMI would represent a comparable surrogate for cervical muscle strength in our statistical models. This did not affect any of our original findings and lends some support to the event anticipation work by Tierney et al.[6] described above. Notwithstanding some of the preliminary work in this area, there is still strong anecdotal support for the role neck musculature may play in reducing the risk of mild TBI that is worthy of investigation in a young, at-risk sample, and future research should investigate this as a potential intervention for preventing mild TBI.

Impacts occurring at different areas of the ice (open-ice vs. along boards) may present different concerns for the clinician ultimately responsible for the management of mild TBI sustained by young individuals. In real-world activities, there is usually some combination of both linear and angular accelerations associated with impacts and impulses. In isolation, however, TBI from linear accelerative forces are believed to result in more focal lesions while rotational mechanisms of injury result in diffuse cerebral injuries.[8-10] We offer that the increased linear forces in collisions we observed in the open ice are the result of allowing movement of the player's head since no contact with the rigid playing boards takes place following the body collision. In open-ice collisions, we observed significantly greater rotational accelerations than those collisions occurring along the playing boards. These differences were on the magnitude of approximately 200 rad/s^2, and previous work has measured the rotational acceleration causing a concussive injury to be as low as 163.35 rad/s^2.[11] When an athlete experiences a rotational mechanism, it is thought that rotation of the cerebrum about the brainstem produces shearing and tensile strains. Since activity in the midbrain and upper brainstem are responsible for alertness and responsiveness, it is perhaps not surprising that rotational mechanisms contributing to TBI are believed to more likely result in loss of consciousness than predominantly linear types of impacts or impulses. Anecdotally, this is reflected when one observes video highlights of "serious" injuries to players presented in the media. The question that still remains elusive to researchers and a matter of contention, for that matter, is "how do the relative contributions of rotational and linear accelerations induce mild TBI?" Many factors play a role in the body's ability to dissipate head impact forces including individual differences in cerebrospinal fluid levels and function, vulnerability to brain tissue injury, relative musculoskeletal strengths and weaknesses, and the anticipation of an oncoming impact or impulse. The ability to ever fully study these phenomena in real-time and in-vivo is a quandary that may never be attained. Regardless of the type, attribute, or severity of a particular impact or impulse, the end result is as follows: the effective mass of the head has become too large for the body to overcome the acceleration or deceleration forces that have sent it in motion. These whiplash-type mechanisms (i.e. hockey bodycheck, football tackle, automobile accident) are believed to contribute equally to mild TBI in addition to direct impacts to the head.[10, 12-15]

We also sought to identify whether a relative body position could be identified that would best mitigate impact forces associated with body collisions sustained in youth ice hockey. We were surprised in that only one of the relative body position descriptors yielded any significantly different results. Teaching players to "skate through the bodycheck" has been encouraged by USA Hockey and Hockey Canada, and has long been taught by coaches to young players in both countries and abroad. It is possible that due to the large number of low-magnitude non-injurious impacts we observed, we would not expect any differences in athletes who exhibited a positive relative body position descriptor compared to those who did not. As we continue in this line of research and record more injuries, it will be interesting to see the trends associated with relative body descriptors in better understanding the biomechanics associated with youth TBI and the subsequent clinical manifestations of those collisions. It should be important to note, in our opinion, that the "ready" position taught by USA Hockey and Hockey Canada should continue to be taught to our young hockey players until further research can suggest better interventions targeted at reducing mild TBI in youth athletes. Given the overall lack of significant differences, it further strengthens the notion that anticipation of a collision may be more important at minimizing head impact severity than the position an athlete may be in at the time of the collision.

Coaches can play a significant role with respect to teaching young hockey players to better anticipate collisions. In practices, game-related drills with full contact will pattern players to adapt to constantly changing scenarios during play. A relatively recent trend in coaching practices is the implementation of "small games" drills. These drills emphasize high speed, quick movements, and game-related tasks (i.e. passing, shooting, checking) in small and confined spaces (i.e. corner of the rink). These drills are excellent at forcing athletes to play with a heightened sense of awareness that allows them to anticipate better incoming body collisions. These drills are also excellent at promoting the quick movement of the puck, which limits the amount of time a player is susceptible to a potentially unanticipated collision. Officials are also mandated with the task of immediately and severely penalizing any player who takes advantage of opponents in susceptible and vulnerable position to deliver an unsuspecting collision. This act is not in the spirit of the rules, whereby body collisions are meant to simply separate a player from puck possession, and not meant to result in an attempt to injure. Player education is also important in promoting safe hockey practices. This may range from practice drills, coaching instruction during games related specifically to where a player's teammates were on the ice during a particular play (i.e. promoting player awareness of his or her surroundings), and interventions designed to evaluate the player's deemed importance of awareness during play.

Due to the costs associated with real-time biomechanical data collection, we were limited in sample size to a single team cohort of Bantam-aged hockey players. We chose this age due to the large discrepancies in player height and mass previously reported in this age group,[16, 17] and the resulting increases in injury risk. Since only the athletes enrolled in our study wore instrumented helmets, we only observed the quality of the body collision in our players and did not take into account the opponents. As studies in this area become more affordable, it will be interesting to study the effect of body collisions between two players both wearing instrumented helmets. This is partly possible in football since heavy hitting between teammates often takes place during practice, whereas this rarely occurs with the same frequency in ice hockey. We are also limited in that we were unable to report a sufficient number of injuries for which any meaningful conclusions could be drawn. We acknowledge that varying player heights and masses, differences in strength, and aggressive tendencies in young athletes may predispose some to more severe head impacts than others. This will be important to account for in future work, as our understanding of these intrinsic factors may have a direct implication to preventive interventions aimed at reducing TBI among our youth.

CONCLUSION

We employed an athletic research model with the goal of better understanding the nature of head impacts sustained by youth ice hockey players, and to describe how body collision type, level of anticipation, and relative body position, may mitigate the severity of these head impacts. Though we studied youth ice hockey players, the results of this study can easily be extended to other youth athletes participating in equipment-intensive collision sports including football and lacrosse. The notion that heightened player anticipation can mitigate the severity of head impacts is supported by our data, especially as it pertains to measures of rotational acceleration sustained during collisions in the moderate range of intensity (50th to 75th percentile). Our finding of increasing head impact severity with decreasing anticipation suggests that coaches target this aspect of ice hockey in their technical development of players during practices to promote the skills necessary to perform at high levels while keeping the safety of ice hockey at the forefront. Teaching players to be aware of their surroundings and, in particular, an opponent who may be striking them is a task not to be ignored by coaches. We recommend hockey coaches spend time during practices educating players on how to safely deliver and receive body collisions in all areas of the ice, including along the playing boards and in the open ice. There is also the responsibility of the player to develop this skill set as well. We strongly support that officials begin to implement more severe penalties for collisions delivered to unsuspecting players. Prospective studies evaluating the effect of player education and technical training on reducing injury rates at the youth level should be undertaken. As no formal cervical strength training program was carried out by the athletes in our sample, it will be important to further understand the effects of cervical muscle strength on mitigating the severity associated with head impacts in youth ice hockey, and to do so in a controlled and prospective manner. Though continued research in this area is necessary, we must use the information we know and implement interventions designed to increase the safety of youth ice hockey players in order to prevent the onset of TBI in young athletes.

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Table 1. The eleven descriptors of relative body position evaluated using the CHECC List.

Relative body position descriptor[1,2]
Was the player looking ahead in the direction of movement?
Did the player appear to be looking in the direction of impending body collision?
Were the player's knees flexed to greater than 30 degrees at the time of the collision?
Was the player's trunk flexed at the time of collision?
Did the player drive into the collision with their shoulders?
Did the player use their elbows, regardless of infraction or not, during the collision?
Did the player use their hands, regardless of infraction or not, during the collision?
Were the player's feet shoulder width apart at the time of the collision?
Did the player use their stick, regardless of infraction or not, during the collision?
Did the player use their legs to drive into or through the body collision?[3]
Was the player receiving or delivering a pass, or taking a shot, at the time of the collision?

[1] All relative body descriptors were evaluated using a dichotomous outcome of "yes" or "no"

[2] All comparisons between "yes" and "no" responses for linear acceleration, rotational acceleration, and HITsp were not statistically significant (with the exception noted in [3] below)

[3] Using legs resulted in lower linear acceleration compared to collisions where the athlete did not use legs. A moderate trend in the same direction for HITsp was observed. No differences in rotational acceleration were noted.

Table 2. Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts sustained by body collision type, anticipation, and overall impression. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts[1]	Linear acceleration (g)	95% CI Lower	95% CI Upper	*P* value[2]
Body collision type					
Along playing boards	421 (63.3%)	20.7	19.4	22.2	0.036
Open-ice[3]	245 (36.8%)	22.4	20.6	24.3	—
Anticipation					
Anticipated	564 (84.7%)	21.1	19.5	22.8	0.135
Unanticipated[3]	102 (15.3%)	22.6	20.9	24.5	—
Overall impression					
Anticipated—good	315 (47.3%)	20.7	19.1	22.5	0.098
Anticipated—poor	249 (37.4%)	21.4	19.6	23.4	0.279
Unanticipated[3]	102 (15.3%)	22.6	20.9	24.5	—
Total	666		—	—	—

[1] Percentages may add up to 100.01% due to rounding

[2] *P* values reflect significant differences relative to the reference category employed by the random intercepts general mixed linear model analyses

[3] Denotes the reference category used in mixed linear models

Table 3. Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts sustained by body collision type, anticipation, and overall impression. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts[1]	Rotational acceleration (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[2]
Body collision type					
Along playing boards	421 (63.3%)	1367.7	1295.6	1443.9	0.003
Open-ice[3]	245 (36.8%)	1564.7	1440.3	1699.9	—
Anticipation					
Anticipated	564 (84.7%)	1414.3	1330.6	1503.3	0.138
Unanticipated[3]	102 (15.3%)	1550.0	1377.4	1744.2	—
Overall impression					
Anticipated—good	315 (47.3%)	1409.4	1303.0	1524.4	0.145
Anticipated—poor	249 (37.4%)	1420.4	1312.4	1537.3	0.184
Unanticipated[3]	102 (15.3%)	1549.9	1377.3	1744.2	—
Total	666		—	—	—

[1] Percentages may add up to 100.01% due to rounding

[2] *P* values reflect significant differences relative to the reference category employed by the random intercepts general mixed linear model analyses

[3] Denotes the reference category used in mixed linear models

Table 4. Frequency (percentage) of recorded impacts, mean HITsp of head impacts sustained by body collision type, anticipation, and overall impression. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts[1]	HITsp	95% CI Lower	95% CI Upper	*P* value[2]
Body collision type					
Along playing boards	421 (63.3%)	15.5	14.74	16.3	0.055
Open-ice[3]	245 (36.8%)	16.3	15.3	17.3	—
Anticipation					
Anticipated	564 (84.7%)	15.8	15.0	16.6	0.755
Unanticipated[3]	102 (15.3%)	15.5	14.2	17.1	—
Overall impression					
Anticipated—good	315 (47.3%)	15.6	14.6	16.5	0.990
Anticipated—poor	249 (37.4%)	16.1	15.2	17.0	0.491
Unanticipated[3]	102 (15.3%)	15.5	14.2	17.1	—
Total	666		—	—	—

[1] Percentages may add up to 100.01% due to rounding

[2] *P* values reflect significant differences relative to the reference category employed by the random intercepts general mixed linear model analyses

[3] Denotes the reference category used in mixed linear models

Appendix E. Manuscript 2

Title:

The effect of infraction type on head impact severity in youth ice hockey

Target journal:

Medicine & Science in Sports & Exercise

The effect of infraction type on head impact severity in youth ice hockey

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Running title: Infractions and head impacts in youth hockey

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ABSTRACT

Purpose: Considering the issues regarding the management of sports-related concussion, especially in youth athletes, we aimed to identify the effects of infractions sustained during participation in youth ice hockey on biomechanical measures of head impact severity.

Methods: Sixteen young male Bantam-aged ice hockey players (age = 14.0±0.5 years; height = 171.3±4.5 cm; mass = 63.7±6.6 kg) were recruited and subsequently equipped with instrumented helmets. The helmets housed six single-axis accelerometers capable of measuring linear acceleration, rotational acceleration, and Head Impact Technology severity profile (HITsp) associated with collisions sustained while participating in ice hockey. Video footage from 54 games was synchronized with the helmet data and all viewable collisions (N = 665) were evaluated as resulting from a legal collision or an infraction. Infractions were further categorized into boarding or charging, checking from behind, and elbowing or intentional head contact. Statistical analyses included random intercepts general linear mixed models.

Results: As many as 17.3% (115 of 665) of all body collisions were the result of observable infractions. Illegal collisions resulted in significantly higher linear accelerations (P=0.012) and HITsp (P=0.021) than legal collisions. Specifically, elbowing, head contact, and high sticking infractions resulted in greater linear (P=0.012) and rotational accelerations (P=0.028) than legal collisions. A strong trend for HITsp was also present for this infraction type (P=0.059).

Conclusion: Based on our data, infractions occur at a relatively high frequency and typically result in higher measures of head impact severity than legal collisions. This information provides objective data to strengthen the ongoing need to educate athletes and coaches to

educate conformity to playing rules, and for establishing a baseline for officiating interventions at the youth ice hockey level.

INTRODUCTION

It has been estimated that between 1.6 and 3.8 million traumatic brain injuries (TBI) result from sports each year in the United States (14). These injuries cost the American health care system approximately $56.3 billion annually in direct and indirect costs (13), and make TBI among the most expensive conditions to treat in children (26). While teams of medical professionals surround collegiate and professional athletes, it is often adolescent athletes who are cared for by parent volunteer coaches with very little medical knowledge. As a result, there is a need to study the factors that may contribute to mild TBI in order to minimize the risk of injury in our young athletes. In the midst of the popularity of professional and collegiate athletics, the young athlete has often been ignored in research. In the area of TBI-related research, it has been found that children with moderate to severe TBI suffer more serious adverse physiological effects after injury than adults with equal severity of brain injury (2). Other studies have also elucidated the increased severity and prolonged recovery of TBI in younger athletes including complications resulting from Second Impact Syndrome (5, 8, 12, 24).

While much has been published in the area of sports-related concussion and mild TBI research in the last 10 years, few studies have sought to understand the biomechanics associated with concussive injuries (10, 11, 17, 18, 22, 25). To date, only one study to our knowledge has disseminated descriptive data on biomechanical measures of head impacts in youth ice hockey players (19). The latter study reported that impacts occurring during games and scrimmages resulted in significantly greater linear accelerations than those sustained during practices. This study, however, failed to report rotational acceleration, a measure that previous animal research has elucidated to be an important contributor to injury (20). Ice hockey injury rates are among the highest of collision sports. Injury rates as high as 75 injuries per 100 ice hockey players have been reported, with 22% of these injuries sustained to the head and neck (9). The injury rate in Bantam players was reported to be 4.3 injuries per 1000 player-hours, with a game injury rate as high as 10.9 per 1000 player-hours (27). Ice hockey presents a number of factors that predispose the participants to higher risk of injury compared to other collision sports. First, the playing surface is made of solid ice and uses rigid boards that contain the playing area. Second, players use a stick to manipulate a rigid frozen projectile (the playing puck) that can sometimes exceed 80 mph. Compounding these two factors, twelve ice hockey players wearing skates with sharp blades taking up position on the ice travel at high speeds, and are encouraged to purposefully collide with any opponent in possession of the puck.

Allowing body collisions at the youth amateur level is a topic not without controversy. Proponents for allowing body collision at young ages offer it is an opportunity to teach proper technique while athletes are not big and strong enough to cause undue injury. Opponents to body collision at young ages argue that the size of the players, especially at the Bantam level (13- and 14-year-olds), can vary significantly and predispose smaller players to injury. Previous work has established anthropometric and biomechanical force profiles for each player in a sample of youth ice hockey players (3). The height and mass of Bantam players differed by as much as 41 cm (16.1 inches) and 48 kg (105.6 pounds), respectively.

The culture of ice hockey often predicates a mentality among players to ignore injury, play recklessly, and encourage unsportsmanlike conduct such as fighting and illegal checking. In a sample of 12 and 13 year old players, fighting was considered a natural consequence of the game (9). Further, while 100% of coaches reported that sportsmanship was “real important,” only 59% of players shared this attitude (4). Parents and coaches in this sample viewed the enforcement of rules as being the most important factor in reducing injuries.

The sports arena provides us with an optimal laboratory in which we can better study the effects of head impacts in young ice hockey players. To our knowledge, this represents the first study to objectively evaluate biomechanical measures of head impact severity in the context of understanding the effects of player infractions on youth ice hockey players. Therefore, the primary purpose of this study was to determine the effects of infractions and infraction types on the biomechanical measures of head impact severity in youth ice hockey players. A secondary purpose was to evaluate whether an interaction existed between infraction type and whether a player caused the offending penalty or was struck illegally by an opponent.

METHODS

Study Design and Participants

This study employed a prospective quantitative research design to evaluate the effect of infractions on head impact severity in youth ice hockey players. Data collection occurred at a number of local, state, national, and international venues, as the ice hockey team traveled for many of their competitions. The study included sixteen male Bantam-level ice hockey players (age = 14.0 ± 0.5 years; height = 171.3 ± 4.5 cm; mass = 63.7 ± 6.6 kg) representing a convenient sample of participants from an AAA-level program. One of the players was a goaltender not actively involved in body collisions and did not have any penalty minutes served to him during the season. Therefore, he was removed from the analyses included in this study. Our sample represented 9 forwards and 6 defensemen. Data were collected during 54 games and 38 practices.

Athletes in our sample participated in at least three sessions (practice or game) each week over the course of the playing season. A detailed explanation of the study was provided for all the athletes, coaches, and parents, prior to the start of the season. While data pertaining to previous history of concussion and years of playing experience were collected, they did not serve as exclusion criteria. Parental permission and minor assent forms approved by the university's institutional review board were signed by each parent and player, respectively, prior to fitting an athlete with an instrumented ice hockey helmet (see Procedures section).

Apparatus

Head Impact Telemetry (HIT) System

This study used commercially available Reebok RBK 6K and 8K helmets (Reebok-CCM Hockey, Inc.; St-Laurent, Quebec, Canada) modified to accept the Head Impact Telemetry (HIT) System technology (Simbex; Lebanon, NH). The helmet's foam liner was custom cut to accept six single-axis accelerometers, a battery pack, and the telemetry instrumentation (***Figure 1***). The custom helmets passed American Society for Testing and Materials (ASTM 1045-99) and Canadian Standards Association (CSA Z262.1-M90) helmet standards, and were approved by the Hockey Equipment Certification Council (HECC) for use during competition. The HIT System utilized spring-loaded accelerometer holders to maintain contact with the head during an impact event. This method has been shown to successfully decouple accelerometers from the head allowing for measurement of head—not helmet—acceleration (15). These accelerometers were positioned tangentially to the head. Linear acceleration of the center of gravity of the head was computed using a least-squares regression algorithm (6, 7). Data from the six accelerometers were collected at 1 kHz for a period of 40 ms (8 ms pre-trigger and 32 ms post-trigger) following the acceleration of any individual accelerometer exceeding 10 g. The data were time-stamped, encoded, stored locally, and then transmitted in real time to a sideline controller (antenna) incorporated within the Sideline Response System (Riddell; Elyria, OH) via a radiofrequency telemetry link. The sideline controller was typically positioned along the playing surface sideboards or in the team’s dressing room. Biomechanical measures of head impact severity were computed and stored. The HIT System is capable of transmitting accelerometer data from as many as 100 players over a distance well in excess of the length of a standard international ice surface. In some instances when the real-time transmission of head impact data was unavailable (i.e. signal interruptions, sideline system not set up, etc), information from 100 separate head impacts were capable of being stored in non-volatile memory built into the acceleration monitoring system.

Evaluation of body collisions

A tool was designed to evaluate body collisions observed during video footage playback. The Carolina Hockey Evaluation of Children's Checking (CHECC) List was scored on 11 readily observable features of human movement when involved in a body collision (**Table 1**). Additional variables of interest on the CHECC List included: whether the player was striking an opponent or was the player struck, and whether an infraction took place during the collision. We further distinguished the illegal infractions into the following types: boarding or charging, checking an opponent from behind, and elbowing an opponent or deliberately making head contact (with their body or playing stick). Intrarater Kappa agreements ranged from 0.40 to 0.92 for the 15-item CHECC List. Interrater agreement suggested moderate to strong agreement between hockey coaches with no scientific experience. The range of agreement was 60% to 96% in which at least five of six coaches agreed on each individual criterion.

Video recording

This study employed the use of a standard digital video camera (Model: PV-GS35; Panasonic Corp.; Secaucus, NJ) to record live game footage onto 60-minute miniDV tapes (Model: M-DV60ME; JVC Americas Corp.; Wayne, NJ). The video camera was capable of recording video footage at 120 Hz, and had a built-in sports exposure mode that allowed for clear video recording of quick action. It was equipped with a 30X optical zoom and 1000X digital zoom, allowing for close-up contained images regardless of where the play was occurring on the ice relative to the research assistant filming the game. Video footage was recorded for every game and scrimmage during the season. Prior to each game, the video camera date- and time-stamping features were synchronized to the Sideline Response System date and time. In order to maximize video image size, the camera followed movement of the puck in an attempt to isolate body collisions during play and to maximize the capture of these events by the camera. Impacts occurring outside the view of the camera were excluded from our analyses, as there was no way to analyze these collisions using the CHECC List. In some instances, late body collisions occurring shortly after the officials blew the whistle were not captured and, therefore, were excluded from our analyses. All raw videos were imported from the video camera connected to a personal computer using a standard universal serial bus (USB) cable. Date and time-stamp information were inlayed onto the videos, and the videos were then exported to a DivX-encoded audio video interleave (AVI) file for storage and future playback. Video playback during CHECC List evaluations was performed on a personal laptop (MacBook; Apple, Inc.; Cupertino, CA) using QuickTime Player Pro (Version 7.5.5; Apple Inc.; Cupertino, CA).

Procedures

Helmet fitting

Prior to the start of the season, players were measured for helmet and facemask size. They were properly fit with Reebok RBK 6K/8K helmets by a certified athletic trainer (ATC). The ATC instructed each participant to wet his hair to simulate sweating. Facemasks owned and used by the players were secured to the new helmet if they were deemed to be in good condition. Otherwise, players were asked to purchase a new facemask or were allocated one by the research team. They were then fitted with the helmet such that the brim of the helmet rested 2.5-3.0 cm (two finger-widths) above the participant's eyebrows. The facemask chinstrap fit tightly under the chin and was securely fastened to the helmet. As a quick test, participants were instructed to hold their head still while the principal investigator attempted to move the helmet. If the investigator was able to move the helmet with no movement of the head, the fitting procedure was repeated. Helmet and facemask fit was verified on a biweekly basis to ensure proper fit throughout the course of the season.

Biomechanical measures of head impact severity

The raw head impact data collected over the course of the season were exported from the Sideline Response System into Matlab 7 (The Mathworks, Inc.; Natick, MA), where data were reduced to include only those impacts sustained during practices, scrimmages, and games. Impacts occurring outside of team-sanctioned events (i.e. pick-up hockey, impacts imparted to the helmet during handling of equipment or travel, etc) were thus omitted from our analyses. Only impacts registering a linear acceleration greater than 10 g were included for the purposes of our analyses as impacts below this threshold are considered negligible with respect to impact biomechanics and their relationship to head trauma. As each impact was linked to a player enrolled in our study by unique identifiers, we were able to easily identify impacts that belonged to a particular player. Resultant linear head acceleration (measured in terms of gravity force, g), resultant rotational head acceleration (measured in rad/s^2), and Head Impact Technology severity profile (HITsp), were the outcome measures of interest and retained for further analysis. The HITsp is a weighted composite score including aspects of linear and rotational acceleration, as well as impact location, and has previously been described in more detail (10). These variables were computed by the HIT System.

Statistical analyses

Since head impact data were highly skewed in favor of low-magnitude impacts, data were transformed using a natural logarithmic function in order to meet the assumptions of normality for the analyses described below. All estimates were then back-transformed to their original scale for purposes of presentation. Descriptive analyses (means and 95% confidence intervals) were calculated for the three biomechanical measures of head impact severity (dependent variables): resultant linear acceleration, resultant rotational acceleration, and the HITsp. In order to address our study purpose, separate random intercepts general mixed linear models were employed for each of our dependent variables. Presence of infraction and infraction type were included as separate independent variables (in addition to player) in the statistical model employed in this study. “Player” represented one level in each statistical model as a repeated factor. We also included the independent variable of “strike,” coded to distinguish between collisions in which the player struck an opponent or in which the player was struck by an opponent, in our statistical model. We did so in order to identify how infraction types affect those players who are struck by the offending players. All random intercepts general mixed linear models (PROC MIXED) were performed in SAS/STAT (Version 9.1; SAS Institute, Inc.; Cary, NC). The level of significance was set at $P < .05$ *a priori*.

RESULTS

Legal vs. illegal collisions

We recorded 4608 head impacts during the 2007-2008 season. We observed a total of 665 body collisions for which we were able to complete a CHECC List and assign a level of infraction. Of these collisions, 82.7% (550 of 665) were deemed to be legal body collisions, while the remaining 17.3% (115 of 665) were deemed to be illegal in nature. Linear accelerations measured during collisions involving illegal infractions (23.0 g; 95% CI: 21.4-24.8) were significantly greater than those sustained during legal collisions (20.9 g; 95% CI: 19.5-22.5) in our sample ($F_{1,13} = 8.46$, $P = 0.012$). The HITsp measures for illegal infractions (16.8; 95% CI: 15.8-17.9) were significantly greater than those we observed for legal collisions (15.5; 95% CI: 14.7-16.4) in our sample ($F_{1,13} = 6.86$; $P = 0.021$). No significant differences between illegal infractions and legal collisions were observed for measures of rotational acceleration ($F_{1,13} = 2.45$; $P = 0.142$).

Infraction types

Of all impacts evaluated using the CHECC List, 82.7% (550 of 665) were legal body collisions, 3.0% (20 of 665) were boarding or charging infractions, 2.9% (19 of 665) were a result of a check from behind, and 11.4% (76 of 665) were a result of elbowing, intentional head contact, or high sticking to the head. Linear head accelerations due to elbowing, intentional head contact, or high sticking to the head (24.0 g; 95% CI: 21.9-26.2) were significantly greater than those observed in legal collisions (20.9 g; 95% CI: 19.5-22.5). There were no differences between legal collisions, those sustained from boarding or charging, and checking from behind. Rotational head accelerations differed across legal collisions and infraction types ($F_{3,28}$ = 3.53, P = 0.028). Impacts involving elbowing, head contact, or high sticking infractions (1614.3 rad/s^2; 95% CI: 1419.6-1835.7) were significantly greater than those observed for legal collisions (1418.4 rad/s^2; 95% CI: 1335.4-1506.5). Though not statistically significant, we observed a trend in the data to suggest differences between boarding or charging infractions (1575.9 rad/s^2; 95% CI: 1419.2-1749.9), and legal collisions (P = 0.103). Checking from behind did not result in any significant differences in head rotational acceleration compared to legal collisions. With respect to the HITsp, the data are suggestive of a trend towards a significant difference between legal collisions and the different infraction types ($F_{3,28}$ = 2.78; P = 0.059). Further exploring this finding, we observed impacts resulting from elbowing, head contact, or high sticking infractions (17.6; 95% CI: 16.0-19.2) to exhibit higher severity profiles than legal collisions (15.5; 95% CI: 14.7-16.4). No significant differences were observed between legal collisions and those sustained as a resulting from boarding or charging infractions and those as a result of checking from behind. When we controlled for body mass index (BMI) in our analyses, we did not observe any changes suggesting body size does not appear to have mitigated any of our previously reported findings.

Interactions between infraction types on striking and struck players

We observed a significant interaction between infraction type and whether a player was striking an opponent or was struck by an opponent on measures of rotational acceleration ($F_{3,15}$ = 4.81; P = 0.015). Players who were checked from behind sustained lower rotational head accelerations (1151.6 rad/s^2; 95% CI: 910.0-1457.3) than those who struck opponents from behind (1395.9 rad/s^2; 95% CI: 1200.5-1623.2). A moderate trend was observed suggesting players who were struck as a result of a boarding or charging infraction sustained greater rotational accelerations than those players who boarded or charged opponents (*P* = 0.083). There were no significant effects of elbowing, head contact, or high sticking infractions between players who were struck and those who delivered the collisions (*P* = 0.116). We did not observe any interaction effects between infraction type and whether a player delivered or received an illegal infraction at the time of the body collision for linear acceleration ($F_{3,15}$ = 0.67; *P* = 0.583) or the HITsp ($F_{3,15}$ = 1.07; *P* = 0.391). All means, 95% confidence intervals, and associated *P* values for significant and non-significant comparisons are presented in ***Tables 2 to 4***.

DISCUSSION

Infractions that occur while participating in youth ice hockey result in more pronounced biomechanical measures of head impact severity compared to legal contact. Our primary finding was that illegal collisions, particularly those involving elbowing, intentional head contact, or high sticking to the head, result in higher measures of linear and rotational acceleration, and HITsp. Unfortunately, based on our analyses, more than 17% of body collisions were deemed to result in an infraction and as many as two-thirds of those infractions were the result of intentional head contact with the elbow, playing stick, or player body. To our knowledge, this study is the first to objectively evaluate the effect infractions may have on brain trauma in the young athlete. It is the first to employ a novel real-time data collection system for the purposes presented in this study. Due to the increasing popularity of ice hockey in the United States, and its continued popularity and growth worldwide including Canada and many countries in Europe, this young population provides us an excellent opportunity to study body collisions in youth sport, and how we may begin to implement skill changes in order to mitigate the forces associated with head impacts sustained during participation.

Our results did not entirely agree with our hypotheses. We anticipated that all types of illegal infractions would result in higher biomechanical measures of head impact severity. Boarding and charging infractions resulted in similar linear and rotational accelerations compared to legal collisions sustained while participating. A boarding penalty is imposed to a player at the discretion of the official based upon the degree of violence of the impact causing an opponent to be thrown violently into the playing boards. Charging is assessed to a player who takes more than 2 steps or strides (i.e. takes a run), or jumps into or charges an opponent. Given the overly aggressive nature of these infractions, it was surprising to not have found any differences in head impact measures compared to legal collisions. There are some possible explanations for this finding. First, our initial analyses did not differentiate between the striking player and the player struck. Secondly, though players may have been boarded or charged by opponents, their level of anticipation or overall preparedness of these impending collisions may have led them to better absorb the forces associated with these collisions. Checking from behind, a very dangerous infraction imposed to a player who body checks or pushes an opponent from behind. This infraction is usually delivered to an unsuspecting opponent who is often unable to protect him or herself. It also represents the leading mechanism of injury for catastrophic cervical spine injuries in ice hockey at all levels (29). Given this, the result that biomechanical measures of head impact severity did not differ between collisions involving checks from behind and legal collisions was unexpected. We hypothesized that this effect could best be described by the relative body position an athlete was in at the time of the collision. When we included the interaction term of infraction type and the total CHECC List score for the eleven body descriptors (higher score results in better relative body position), we did not find any interaction effects suggestive of the role of relative body position on mitigating the effects of infraction type.

In agreement with our hypotheses was that elbowing, head contact, and high sticking to the head infractions resulted in more pronounced biomechanical measures of head impact severity. An elbowing infraction is imposed to any player who uses his or her elbow in such a way as to attempt to foul an opponent. A head contact is imposed on any player who intentionally or recklessly contacts a player in the head, including with the stick or with an illegal body check. Since contact for this group of infractions is sustained directly to the head, with the intent to strike the head with excessive force, our findings of increased linear and rotational acceleration compared to legal collisions was expected. In cases where an athlete would anticipate the impending collision, it could be argued the forces generated by the cervical musculature would be unable to overcome the inertial forces of an opponent traveling at high speeds and directing their body mass through the opponent's head during a collision. Controlling for cervical muscle strength in these types of analyses will be an important step for future work in this area. In our own ongoing work, we have observed a good relationship between total cervical muscle strength and BMI ($r = -0.618$). This would suggest that as BMI increases (i.e. athletes become more overweight and, thus, less fit) athletes have weaker cervical muscle strength. Given this relationship, we included BMI as a surrogate covariate to cervical muscle strength in our analyses. We felt that including cervical muscle strength (represented by BMI in these analyses) would provide us with an outlet to better understand and explore the effects these covariates may play on the severity of head impact measures following illegal collisions. In other words, would players with strong cervical muscles better able to mitigate the magnitude of head impact forces following an illegal collision? After controlling for BMI, the effects of infraction type on biomechanical measures of head impact severity do not appear to be affected. Elbowing and head contact infractions, in our opinion, provide the greatest likelihood of incurring brain injury to ice hockey players. Given their nature, it is our opinion that these findings may be extended to other collision sports including football and lacrosse. Though some of these infractions may not be relevant in those sports, the danger of deliberate head contacts, especially in unsuspecting athletes, remains a concern to sports medicine professionals tasked with the care of these athletes. This presents a number of challenges to maintaining the safety of participants involved with ice hockey. Coaches should be tasked to promote a fair method of play among their players. Taking the time during practices to promote safe body checking techniques, instructing players not to "take runs" at unsuspecting opponents, and promoting body collisions in the spirit of their original intent which was to separate the opponent from puck possession, and not to deliberately injure an opponent. Officials should be responsible for maintaining a safe playing environment, which often can be as simple as keeping control of the game and not allowing it to get out of hand from a hostility standpoint. Players should recognize that their actions may directly injure an opponent, and should be taught to play fairly and within the confines of the rules set forth by their respective governing bodies. Lastly, parents should recognize the value of body collisions for their purpose, and to foster an environment for their children such that safe and fair play are rewarded and illegal actions are dealt with swiftly.

While the immediate purpose of this study was not to understand coaches' and players' aggressive behaviors and why they may behave in this manner, it is difficult to ignore this aspect of ice hockey, and a brief discussion of this is warranted in the context of the results presented in this study. Although USA Hockey and Hockey Canada strongly encourage sportsmanlike behavior, the culture of ice hockey predicates a mentality among players to ignore injury, play recklessly, and encourages unsportsmanlike conduct such as fighting and illegal checking. In the United States, a study of Peewee-level players reported that fighting broke out in approximately 17 of 52 games observed. In this sample, players considered fighting a natural consequence of the game and experienced a certain resignation about fighting (9). Another interesting finding is that while 100% of coaches felt sportsmanship was "real important," only 59% of players shared this attitude (4). Parents and coaches in the latter sample viewed the enforcement of rules as being the most important factor in reducing injuries. Tator's work in the area of catastrophic cervical spine injury agrees with this statement and emphasizes a need for strict enforcement of the hit-from-behind rule and the necessity of continued education for coaches and players regarding the risk of head and neck injuries in ice hockey (29). Notwithstanding, our results suggest that as many as 17.3% of all collisions we observed involved some form of illegal infraction. While a low number of checks from behind (2.9%) were still observed, they occurred at a much lower frequency than elbowing and head contact infractions. This is suggestive that interventions implemented by USA Hockey and Hockey Canada to reduce the risk of catastrophic neck injuries resulting from checks from behind appear, at quick glance, to be effective.

A number of interventions have sought to mitigate the frequency of penalties and aggressive play in order to reduce injury rates at the youth ice hockey level. A Canadian program labeled *Fair Play* was introduced to youth ice hockey in the province of Quebec. *Fair Play* was designed to penalize unnecessary roughness by awarding a fair play point to teams that reduce the number and severity of their penalties. In one study, penalties issued to teams playing under the *Fair Play* program were compared to those teams not using the system (16). The authors reported 30% less major penalties and 25% less game suspensions were issued to the Bantam-level *Fair Play* teams compared to their non-program counterparts. While our study did not compare a pre- and post-intervention, our data would suggest that elbowing and head contact infractions still occur at a fairly high rate in youth ice hockey. At the Peewee level, *Fair Play* teams averaged 1.3 major penalties per season compared to 6.3 major penalties for non-program teams. Further, among teams using the *Fair*

Play system, 71% of them did not receive a single game suspension. The *Fair Play* study highlights a number of key points as they relate to the current study. First, interventions specifically designed to reward teams' proper behavior appear to result in decreases in illegal conduct and, specifically, severe misconduct more likely to result in injuring an opponent by possibly mitigating the head impact severity associated with those collisions. Secondly, according to this study, it appears as though interventions such as *Fair Play* have a greater impact on younger players. This suggests that targeting younger players, such as those included in our sample or younger, may be best suited by these interventions. Since our data suggest that almost 1 in 5 collisions involve some form of infraction, it will be important to educate officials on the dangers of unsafe actions during play. Future work in this area should draw on these findings and study the effects on injury rates as a result of targeted officiating designed specifically to reduce head trauma and its severity in youth ice hockey.

Allowing body checking at young ages in ice hockey is not without controversy. Many believe body checking leads to a laissez-faire attitude toward body collisions and an increase in rule infractions (21). Cerebral concussion (i.e. mild TBI) occurred in each of four tournaments that allowed body checking (23). The rate of mild TBI ranged from 10.7 to 23.1 per 1000 player-hours in the tournaments observed by Roberts et al. This rate is markedly higher than regular season rates previously reported. Sutherland et al., for example, recorded 0.09 concussions per 1000 player-hours (28), while Brust et al. and Stuart et al. estimated 0.75 concussions per 1000 player-hours and no concussions during a season, respectively (4, 27). In contrast, as many as 10% of high school ice hockey players sustained a concussion during the regular season (9). Given the reported literature in this area, it is surprising that the national governing bodies for amateur hockey in the United States and Canada have not done more to educate players, parents, officials, and coaches on this topic.

Our study is not without a number of limitations worthy of discussion. First, we were limited to only 665 body collisions in which we were able to observe video footage in order to complete the CHECC List, and an even smaller number (N = 115) consisted of observable infractions. Second, our athletes represented a convenient sample of Bantam-aged ice hockey players from a single elite hockey team. As a result, the results of this study may be difficult to extend to non-elite levels of youth ice hockey including local travel, recreational, and house leagues. Since every participant knew he was wearing an instrumented helmet, there may have been a level of competition among the teammates (unknown to the researchers) that may have predisposed an athlete to want to hit an opponent harder than their fellow teammates. Finally, infractions were assessed based on careful review of game footage. Videos were played back in regular time, slow motion, and freeze frame forwarding, a number of times before an infraction type was assigned to a given body collision. We are aware that on-ice officials are tasked with making immediate calls and are often unable to identify potentially injurious collisions that may occur beyond their field of vision. It should be noted that not every infraction we observed was called as such by the on-ice officials.

As we begin to better understand the causes and effects of cerebral concussion in pediatric athletes, it is believed we will become better prepared to introduce prevention programs and improved emergency care and injury management programs. The outcomes from this study and others in this area have the potential to help create a more positive and safe environment for youth hockey players. The results provide valuable information regarding situations in which youth hockey players are at the greatest risk for sustaining head impacts of higher, and potentially dangerous, magnitudes. Since the tempo and aggressiveness of ice hockey is a direct reflection on a game's officiating, this study serves to better elucidate the effects of player infractions on measures of head impact severity. This information, in addition to educational interventions, can have a targeted purpose of improving the awareness of these infractions among players, coaches, and officials, and emphasize the importance of enforcing player infractions likely to result in more severe head impacts.

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CONFLICT OF INTEREST

Dr. Greenwald has a financial interest in the HIT System technology used to collect data for this study.

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Table 1. The eleven descriptors of relative body position evaluated using the CHECC List.

Relative body position descriptor[1]
Was the player looking ahead in the direction of movement?
Did the player appear to be looking in the direction of impending body collision?
Were the player's knees flexed to greater than 30 degrees at the time of the collision?
Was the player's trunk flexed at the time of collision?
Did the player drive into the collision with their shoulders?
Did the player use their elbows, regardless of infraction or not, during the collision?
Did the player use their hands, regardless of infraction or not, during the collision?
Were the player's feet shoulder width apart at the time of the collision?
Did the player use their stick, regardless of infraction or not, during the collision?
Did the player use their legs to drive into or through the body collision?
Was the player receiving or delivering a pass, or taking a shot, at the time of the collision?

[1] All relative body descriptors were evaluated using a dichotomous outcome of "yes" or "no"

Table 2. Frequency (percentage) of recorded impacts, mean resultant linear acceleration of head impacts sustained by legality of collision, and infraction type. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts	Linear acceleration (g)	95% CI		*P* value[1]
			Lower	Upper	
Legality of body collision					
Illegal collision	115 (17.3%)	23.0	21.4	24.8	0.012
Legal collision[2]	550 (82.7%)	21.0	19.5	22.5	—
Type of infraction					
Boarding/charging	20 (3.0%)	21.2	18.7	24.0	0.868
Checking from behind	19 (2.9%)	21.4	18.8	24.3	0.722
Elbowing/head contact	76 (11.4%)	24.0	21.9	26.2	0.005
Legal collision[2]	550 (82.7%)	21.0	19.5	22.5	—
Total	665		—	—	—

[1] *P* values reflect significant differences relative to the reference category employed by the random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 2. Frequency (percentage) of recorded impacts, mean resultant rotational acceleration of head impacts sustained by legality of collision, and infraction type. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts	Rotational acceleration (rad/s^2)	95% CI Lower	95% CI Upper	*P* value[1]
Legality of body collision					
Illegal collision	115 (17.3%)	1529.9	1388.5	1685.8	0.142
Legal collision[2]	550 (82.7%)	1417.5	1334.8	1505.3	—
Type of infraction					
Boarding/charging	20 (3.0%)	1575.9	1419.2	1749.9	0.103
Checking from behind	19 (2.9%)	1197.7	953.3	1504.9	0.144
Elbowing/head contact	76 (11.4%)	1614.3	1419.6	1835.7	0.059
Legal collision[2]	550 (82.7%)	1417.5	1334.8	1505.3	—
Total	665		—	—	—

[1] *P* values reflect significant differences relative to the reference category employed by the random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

Table 3. Frequency (percentage) of recorded impacts, mean HITsp of head impacts sustained by legality of collision, and infraction type. The associated 95% confidence intervals and p-values are provided.

	Frequency of impacts	HITsp	95% CI Lower	95% CI Upper	*P* value[1]
Legality of body collision					
Illegal collision	115 (17.3%)	16.8	15.8	17.9	0.021
Legal collision[2]	550 (82.7%)	15.5	14.7	16.4	—
Type of infraction					
Boarding/charging	20 (3.0%)	16.8	14.6	19.3	0.364
Checking from behind	19 (2.9%)	14.1	12.0	16.7	0.199
Elbowing/head contact	76 (11.4%)	17.6	16.0	19.2	0.010
Legal collision[2]	550 (82.7%)	15.5	14.7	16.4	—
Total	665		—	—	—

[1] *P* values reflect significant differences relative to the reference category employed by the random intercepts general mixed linear model analyses

[2] Denotes the reference category used in mixed linear models

FIGURE 1. The protective foam of the ice hockey helmets were removed from the helmet shell. Following this, six single-axis accelerometers were fitted into custom holes cut into the foam. The figure depicts the location of the helmet accelerometers in the protective foam (hard shell removed) in both front (A) and rear (B) views. The arrows identify the location of the six accelerometers as viewed from the inside of a fully assembled playing ice hockey helmet (C).

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