Journal of ASTM International, Vol. 6, No. 6
Paper ID JAI101848
Available online at www.astm.org

Joseph T. Gwin,[1,2] *Jeffery J. Chu,*[1] *Thomas A. McAllister,*[3,4] *and Richard M. Greenwald*[1,2]

In situ Measures of Head Impact Acceleration in NCAA Division I Men's Ice Hockey: Implications for ASTM F1045 and Other Ice Hockey Helmet Standards

ABSTRACT: A pilot study was performed to measure head impact accelerations in collegiate men's ice hockey during the 2005–2007 seasons using helmets instrumented with Head Impact Telemetry System technology to monitor and record linear head accelerations and impact locations in situ. The objectives of this study were (1) to quantify the relationship between resultant peak linear head acceleration and impact location for in situ head impacts in collegiate men's ice hockey, (2) to quantify the frequency and severity of impacts to the facemask, and (3) to determine if in situ impacts occurred such that the peak resultant linear head acceleration was higher than the peak resultant linear headform acceleration from a 40-in. linear drop (as in ASTM F1045–99) on the same helmet at a similar impact location. Voluntary participants (n=5 and 7 for years 1 and 2, respectively) wore instrumented helmets which monitored head impact accelerations sustained by each player during all games and practices. Head impact data were grouped by impact location into five bins representing top, back, side, forehead, and facemask. Forehead impacts represented impacts to the helmet shell as distinguished from facemask impacts. Additionally, a sample instrumented helmet was impacted in the laboratory at forehead, side, rear, and top impact locations (40-in. drop, three trials per location, test setup as specified in ASTM F1045-99). The mean peak resultant linear headform acceleration for each impact location was determined for analysis. Of the 4,393 recorded head impacts, 33.2 % were to the back of the helmet. This percentage increased to 59.2 % for impacts above 70 g. Facemask impacts accounted for 12.2 % of all impacts but only 2.4 % of impacts above 70 g. Over two seasons, five in situ impacts occurred such that the peak resultant linear head acceleration was greater than the mean peak resultant linear headform acceleration for a corresponding impact location in the laboratory. This study found that the most common impact location in ice hockey, particularly for impacts with higher peak linear accelerations, was the back of the head and demonstrated that facemask impacts were typically of a lower magnitude. The five impacts or ~0.4 per player/season that exceeded the peak linear acceleration associated with 40-in. laboratory drops suggested that the impact energy specified in ASTM F1045 may not reflect the highest energy impacts seen in situ.

KEYWORDS: hockey, impacts, helmets, collegiate sports, concussions

Introduction

Head impact is an inevitable part of the fast-paced game of hockey. Over the years the frequency and severity of severe or catastrophic focal head injuries resulting from these impacts have been reduced by the availability of standardized ice hockey helmets and the enforcement of helmet usage rules in leagues of all levels [1]. In 1961, the Swedish Ice Hockey Association (SIA) appointed a committee to investigate the ever-increasing number of head injuries and to develop a set of rules for ice hockey helmets. The committee established the first helmet standards for ice hockey, which were approved in January 1962. By October 1963, all players were required to wear certified helmets during SIA sanctioned games [2]. Since 1963, there have been no reported fatal head injuries from ice hockey in Sweden [1]. In North America, the Canadian Standards Association (CSA) and the American Standards for Testing and Materials (ASTM) soon followed suit, although most North American leagues did not mandate the use of helmets in ice hockey until the death of professional hockey player Bill Masterton in 1968. The National Hockey League did not mandate helmet use until 1979 and players who entered the league before this mandate continued to skate without headgear until the 1996/1997 season [3].

Manuscript received April 30, 2008; accepted for publication April 11, 2009; published online May 2009.
[1] Thayer School of Engineering, Dartmouth College, Hanover, New Hampshire 03755.
[2] Simbex, Lebanon, New Hampshire 03766.
[3] Dartmouth Medical School, Hanover, New Hampshire 03756.
[4] Dartmouth Hitchcock Medical Center, Lebanon, New Hempshire 03756.

The impact attenuation standards for ice hockey helmets that were developed in the 1960s remain largely unchanged to this day. They are based on measures of peak linear acceleration recorded by a helmeted and instrumented headform that is dropped onto a standardized surface from a preset height. The pass/fail criterion that was developed for these tests was based primarily on the threshold for skull fracture in the human cadaver [4]. In the 45 years since the development of the first SIA hockey helmet standard, much has been learned about traumatic brain injury (TBI), specifically about the mildest type of brain injury, mild traumatic brain injury (mTBI or concussion). In 1962, mTBI was thought to be a fully reversible phenomenon. It is now widely acknowledged that mTBI can have lasting effects and that these effects can be cumulative [5,6]. Concussion in sport makes up only 7 % of all mTBI cases presenting to emergency departments in the United States, but makes up 27 % of cases for the 5–14 age group [7]. In hockey, concussions affect 10 % of all athletes and make up 12–14 % of all injuries [8].

In the last decade, mTBI has been recognized as a major public health issue and much emphasis has been put on proper treatment and prevention. The National Institutes of Health and the Centers for Disease Control and Prevention have declared that reducing the incidence, severity, and postinjury symptomology of mTBI should be a national research priority [9,10]. To this end, neuropsychologists are seeking ways to better identify symptomatic athletes including persistent refinement of computerized neuropsychological testing protocols [11], and athletic trainers are striving to implement the newest and best practices on the field as evident in the National Athletic Trainers' Association (NATA) Position Statement: Management of Sport-Related Concussion [12]. Moreover, leagues are seeking rule changes that preserve the integrity of the game and that attempt to reduce the incidence and severity of injuries to athletes. One such example is the International Ice Hockey Federation head checking rule, which was established in 2002 and which increased the penalty for any contact to the head [1,13]. Biomechanists and helmet designers are faced with understanding the dynamics of head impacts in situ with the end goal of establishing criteria for improved headgear with regard to reducing a specific injury such as mTBI. A tested method to promote design improvements is to implement test standards that mimic potentially injurious situations seen in actual play.

Previous research has suggested that advances in material engineering coupled with an increased understanding of the mechanisms and clinical sequelae of mTBI warrant reconsideration of ice hockey helmet standards [14]. Hoshizaki demonstrated that common hockey helmet liner materials do not attenuate energy uniformly across a range of impact velocities. Rather, engineered materials such as expanded polypropylene (EPP) can be tuned to perform optimally at specific energy levels such as those prescribed in a given helmet standard. For example, EPP outperformed vinyl nitrile, another commonly used liner material in ice hockey helmets, at 25 J but not at 5 J of impact energy [14]. Recently developed technology [Head Impact Telemetry (HIT) System, Simbex, Lebanon, NH; Sideline Response System, Riddell, Chicago, IL], has enabled in situ monitoring of head impact accelerations during helmeted activities. Studies of in-field head acceleration in American football, reported by the authors and others, have utilized this technology to demonstrate that a wide range of impact accelerations can result in clinically diagnosed concussion, including impacts that correspond to lower impact energies than those prescribed in current hockey helmet standards [15–20]. This suggests that energy attenuation properties of helmets should be evaluated across a range of impact velocities. Additionally, Halstead et al. [21] and others suggested that ice hockey helmet standards may not adequately reflect the highest energy impacts sustained during play.

The purpose of this study was to characterize the head impacts sustained in situ by Division I collegiate male ice hockey players during the 2005 and 2006 seasons. Specifically, the objectives of this study were (1) to quantify the relationship between resultant peak linear head acceleration and impact location for in situ head impacts, (2) to quantify the frequency and severity of impacts to the facemask, and (3) to determine if in situ impacts occurred such that the peak resultant linear head acceleration was higher than the peak resultant linear headform acceleration from a 40-in. linear drop (as in ASTM F1045–99, "Standard Performance Specification for Ice Hockey Helmets") [22] on the same helmet at a similar impact location. The hypothesis was that the majority of high magnitude impacts recorded during play would occur to the back of the head, in agreement with epidemiological studies dating back to 1962 when the first SIA helmet standard was developed [2].

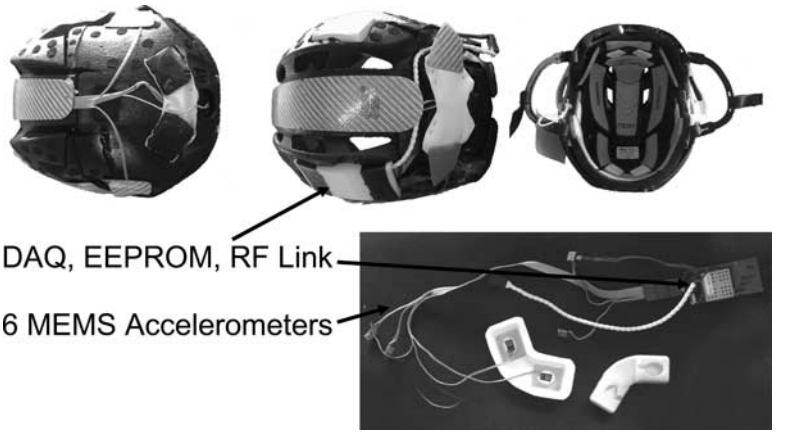

FIG. 1—*HIT System for hockey. The IHU for ice hockey consists of data acquisition and telemetry electronics as well as six MEMS accelerometers mounted within the liner of a commercially available ice hockey helmet.*

Methods

Commercially available ice hockey helmets (RBK 8, Reebok, Montreal, Canada) worn by men's ice hockey players ($n=5$ and 7 for years 1 and 2, respectively) at Dartmouth College (Hanover, NH) were instrumented with HIT System (Simbex, Lebanon, NH) in-helmet units (IHUs) to measure head impacts sustained during the 2005 and 2006 seasons. Players wore instrumented helmets during both practices and games. The participating players were selected by the medical staff from a pool of volunteers. First and second line players, who were expected to receive more ice time, were given preference over third and fourth line players. Two defensemen and three forwards participated in the study in Year 1, and three defensemen and four forwards participated in the study in Year 2. Informed consent was obtained for all participating athletes in accordance with the Dartmouth College Institutional Review Board.

Data Collection

Instrumented helmets contained six single axis micro electric-mechanical systems (MEMS) accelerometers (Analog Devices, Inc., Cambridge, MA), data acquisition electronics, 128 kbyte of memory (capable of storing data for up to 100 impacts), and a rf transceiver. These components were built into the liners of commercially available EPP ice hockey helmets (Fig. 1) and were collectively referred to as an IHU.

The IHU recorded 40 ms of data at 1,000 Hz (8 ms pretrigger and 32 ms post-trigger) when a preset acceleration threshold (10 g in this study) was exceeded by any one single accelerometer. These data were also associated with an impact time and a unique IHU identifier, to distinguish impacts downloaded from multiple IHUs. The bandwidth of the acceleration data was limited by a 0.5 Hz ac filter, used to remove any dc offsets, and a 400 Hz low pass filter that was on the accelerometer. Each IHU transmitted data in real time to the HIT sideline controller (SC), which consisted of a rf telemetry link and a laptop computer for data processing and storage. Within the SC, the data from the six nonorthogonal accelerometers in the helmet were used to compute the head center of gravity (CG), resultant linear acceleration time series, and impact location [23,24]. Resultant linear acceleration refers to the magnitude of the linear acceleration vector. The maximum value of this resultant time series is herein referred to as peak linear acceleration. Instrumented helmets have passed relevant ASTM and CSA standards tests at several independent testing facilities. The measurement accuracy of HIT System IHUs in ice hockey helmets was previously validated [25].

To facilitate comparison of the magnitude of linear acceleration of the head measured from in situ head impacts with the linear acceleration of the headform measured during laboratory testing, an instrumented helmet of the same model worn by the players in this study was impacted at forehead, rear, side, and top impact locations for a 40-in. drop according to ASTM F1045–99. Three trials were performed per impact location. The peak linear headform acceleration was computed for each impact and mean peak linear headform acceleration for each impact location was determined. The ASTM ice hockey helmet test standard (ASTM F1045–99) involves dropping a helmeted metallic headform, which is instrumented with a

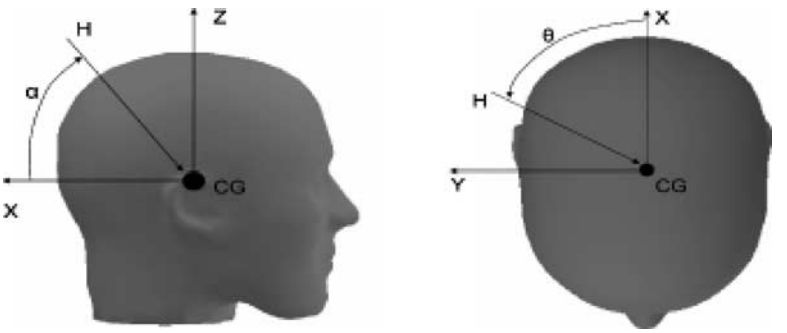

FIG. 2—*Impact location was defined by azimuth (α) and elevation (θ). The azimuth (θ) was defined from −180° to 180° with 0° at the positive x axis and positive θ to the right side of the head. The elevation (α) was defined from 0° (horizontal plane passing through the head CG) to 90° (crown of the head at the positive z axis). The xz plane represented the midsagittal plane with positive x corresponding to the caudal direction. The xy plane represented the coronal plane with positive y to the right side of the head.*

triaxial linear accelerometer at the headform CG, onto a rubber drop surface from various heights and at prescribed impact locations. The headform is rigidly attached to a drop bar which limits any rotation of the headform during impact. The pass/fail criterion is that the average peak linear acceleration for three impact tests to a prescribed impact location may not exceed 275 g and the peak linear acceleration for a single trial may not exceed 300 g.

Data Analysis

Head impact data were grouped by impact location into five bins representing top, back, side, forehead, and facemask. Top impacts were defined as impacts >65° elevation where 0° was the horizontal plane passing through the head CG and 90° was the ray through the crown of the head (Fig. 2). The remaining impacts were classified into equally spaced bins where front and back were centered on the sagittal plane. Forehead impacts represented impacts to the helmet shell as distinguished from facemask impacts. Only impacts with peak linear acceleration >10 g were considered for analysis, as in Mihalik et al. [19].[5] For each impact location bin, the frequency of in situ impacts such that the peak linear head acceleration was higher than the mean peak linear headform acceleration for the corresponding laboratory test impact location was determined. Impacts to the facemask were excluded from comparison with laboratory impacts because there is no corresponding facemask test location prescribed in the ASTM certification standard, but were included in analyses of the distribution of in vivo head impacts.

Analysis of variance (ANOVA, MATLAB, The MathWorks Inc., Natick, MA) was used to compare the mean of the top 1 %, 2 %, and 5 % of peak linear accelerations across location groups, as in the authors' analysis of head impact accelerations in football [17]. Additionally, chi squared independence tests were used to evaluate differences in the distribution of impact locations as a function of peak linear acceleration. Head impact data were grouped by peak linear acceleration into 20 g bins (10–29.9, 30–49.9, 50–69.9, 70–89.9, and 90 g and above). The distribution of impact locations for each linear acceleration group was compared to the distribution for all data. The significance level was set at $\alpha = 0.05$ for all analyses. To visually represent the data, heat maps, which projected the concentration of impacts onto a three-dimensional headform, were generated for each peak linear acceleration bin (MATLAB, The MathWorks Inc., Natick, MA).

Results

A total of 4,393 head impacts were recorded for the twelve players over the two seasons. This amounts to approximately five impacts per player per day of data collection. The mean of the top 1 %, 2 %, and 5 % of all peak linear accelerations were 104.9 g, 90.0 g, and 62.4 g, respectively. 33.2 % of all impacts were

[5]Data collection was triggered if a single accelerometer within the IHU exceeded 10 g. It is possible for a single accelerometer to exceed 10 g while the resultant linear head acceleration is <10 g. These recorded impacts of <10 g peak resultant linear acceleration were removed prior to analysis.

TABLE 1—*Differences in the distribution of peak linear acceleration as a function of impact location*

	Back			Side			Front[a]			Top			
	n	Cutoff	Mean (g)	*n*	Cutoff	Mean (g)	*n*	Cutoff	Mean (g)	*N*	Cutoff	Mean (g)	Significance
Top 1 %	15	117.2	151.9	14	90.7	119.2	12	68.0	90.9	3	141.0	185.7	$p>0.18$
Top 2 %	29	97.9	130.6	29	72.0	99.1	23	53.1	75.3	7	121.2	154.5	$p<0.05$
Top 5 %	73	81.3	106.1	72	52.1	75.9	58	42.4	58.0	17	94.7	125.6	$p<0.02$

[a]Front includes forehead and facemeask impact location groups.

to the back of the head, 32.6 % were to the side, 14.3 % were to the forehead, 12.2 % were to the facemask, and 7.8 % were to the top of the head.

There were five head impacts that resulted in a peak linear head acceleration that was greater than the corresponding mean laboratory peak linear headform acceleration for a 40-in. linear drop (as in ASTM F1045–99) to a similar location. These five impacts were 268.1 g, 91.9°, and 45.0° (peak linear head acceleration, impact location azimuth, and impact location elevation, respectively); 197.0 g, −9.8°, and 31.5°; 239.7 g, −4.1°, and 38.9°; 154.9 g, −59.6°, and 72.6°; 255.0 g, −0.06°, and 86.6°. The highest magnitude linear acceleration recorded at any location in the laboratory testing was 183 g.

Differences in the mean peak linear accelerations across impact location groups were statistically significant for the top 2 % and 5 % of all data ($p<0.05$) but not for the top 1 % of all data ($p>0.18$). Specifically, the top 2 % and 5 % of all impacts to the top of the head had significantly higher peak linear acceleration than impacts to the back of the head, which had higher peak linear accelerations than the side and front (Table 1). There were no significant differences between forehead and facemask impacts for the top 1 %, 2 %, or 5 % levels ($p=0.92$, 0.29, and 0.11, respectively). As such, forehead and facemask impacts were grouped together as front impacts (Table 1).

In addition to grouping by impact location, impact data were grouped by peak linear acceleration. Chi squared tests for independence demonstrated that the distribution of impact locations was significantly different for each peak linear acceleration group compared to the distribution for all data. Specifically, the highest magnitude impacts (>70 g) tended to be to the back of the head while the lowest magnitude impacts (<50 g) were relatively evenly distributed (Table 2).

Heat maps were generated to visually represent the data. Each heat map represented a peak linear acceleration group and projected the relative concentration of head impacts onto a three-dimensional headform. For the lowest peak linear acceleration grouping (10–30 g), the frequency of impacts to the back, side, and front (including forehead and facemask) were similar (Table 2). However, the side location bin covers approximately twice as much head surface area as the back and the facemask location bin covers approximately twice as much head surface area as the forehead. When normalized by the surface area, the highest concentration of low magnitude impacts (10–30 g) occurred to the forehead and the highest concentration of high magnitude impacts (>70 g) occurred to the back of the head (Fig. 3).

Discussion

Collegiate hockey players at Dartmouth College wore instrumented helmets to monitor head impact accelerations sustained during all games and practices during the 2005 and 2006 seasons. The 4,393 head

TABLE 2—*Chi squared independence tests for differences in impact location distribution by peak linear acceleration subgroups*

Peak linear acceleration	Impact location						Chi squared independence test	
	Back	Side	Forehead	Facemask	Top	Total	$x^2(df=4)$	Significance
10–30 g	1,076	1,200	540	466	236	3,518	21.04	$P<0.001$
30–50 g	202	152	70	59	52	535	9.89	$P<0.001$
50–70 g	80	48	12	8	23	171	31.87	$P<0.001$
70–90 g	48	16	4	0	13	81	42.60	$P<0.001$
90 g+	52	14	0	4	18	88	60.20	$P<0.001$
All data	1,458	1,430	626	537	342	4,393		

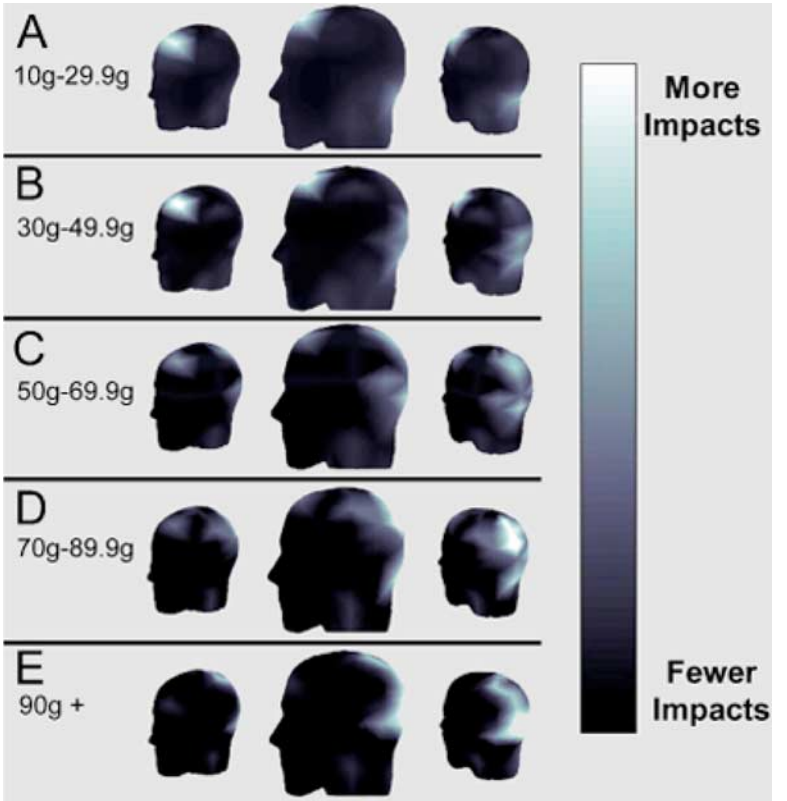

FIG. 3—*Heat maps represent the concentration of impacts per unit of head surface area. Panel (a) contains data with peak linear acceleration between 10 and 30 g, panel (b) 30–50 g, panel (c) 50–70 g, panel (d) 70–90 g, and panel (e) 90 g and above.*

impacts quantified, for the first time, the distribution of head impact accelerations that occurred in men's ice hockey at the collegiate level. This researched focused on the implications of the in situ data on ASTMF1045 and other ice hockey helmet test standards. Three independent findings are reported.

The first finding was that there were significant differences in the location of head impacts as a function of peak linear acceleration. Specifically, high magnitude linear head accelerations were most common for impacts to the back of the head. This finding was consistent with epidemiological statistics dating back to the 1960s. The committee of the SIA that developed the first ice hockey helmet standard in 1962 agreed, based on these epidemiological statistics, that the most critical area for impact protection in ice hockey was the back of the head, an area at risk during slips when the player falls backward onto the ice [2]. It is interesting to note that the highest magnitude impacts to the top of the head (top 1 %, 2 %, and 5 % by peak linear acceleration) were of a higher magnitude than impacts to the back of the head. However, there were substantially more impacts to the back (33.2 % of total) than to the top (7.8 % of total).

The second finding of this study was that in situ head accelerations resulting from impacts to the forehead and facemask tended to be of a lower magnitude than other impact locations. High magnitude (>70-g peak linear acceleration) impacts to the forehead or facemask were uncommon ($n=8$ impacts, 0.2 % of total). This finding is relevant given that the National Operating Committee for Standards in Athletic Equipment (NOCSAE), the organization whose helmet test standard is utilized by most American football leagues, is considering adopting a headgear impact attenuation standard which incorporates a facemask impact location [26]. This proposed standard is not explicitly designated for football helmets. Currently, ice hockey facemasks are tested for protection from projectiles (pucks) (ASTM F513–00, "Standard Safety Specification for Eye and Face Protective Equipment for Hockey Players") but not for direct impacts to rigid surfaces [27]. The in situ data could be useful in determining an appropriate impact energy for an ice hockey facemask impact attenuation test standard. However, head acceleration data from impacts to the facemask presented in this study were lower than at other impact locations.

The third finding was that in situ head impacts occurred such that the peak resultant linear head

acceleration was higher than the peak resultant linear headform acceleration from a 40-in. linear drop (as in ASTM F1045–99) on the same helmet at a similar impact location. The incidence of these impacts was uncommon ($n=5$ impacts, 0.1 % of all). However, the occurrence of any such impacts suggested that the ASTM ice hockey helmet test standard may not be reflective of the highest energy impacts sustained in situ. It is difficult to practically measure impact energy on the ice (e.g., the kinematics of the player relative to the impact surface). Previous studies have utilized high-speed video to determine these kinematic parameters for a select group of open field impacts in football [28]. However, this determination is costly and time consuming, and is not feasible for a large-scale analysis of all impacts occurring during play. As such, output measures (peak linear acceleration in this study) were used to compare laboratory data to in situ data. It is telling that these high magnitude peak linear head accelerations occurred in situ. This finding supports the previous hypothesis that ASTM F1045 may not be representative of the highest magnitude impacts that occur during play [21].

There were several limitations of this study. Head acceleration data were collected at one organization from twelve players wearing one helmet type. Future work should include an analysis of head impact accelerations in both men's and women's ice hockey at several organizations. Multiple helmet models, including models with vinyl nitrile liners, should be evaluated. Furthermore, this study addressed the biofidelity of impact energies prescribed in ASTM F1045 but did not address the biofidelity of the temporal properties of the headform impact response. The ability to rapidly and cost-effectively collect temporal data from in situ head impacts facilitates study of the frequency response of the head and neck to impact under various conditions. Temporal properties of laboratory headform acceleration (for various headforms and impact protocols) should be compared to in situ data to further guide the development of a more biofidelic headforms and helmet test standards.

In situ head impact data can help guide the development of helmet test standards. However, it does not seem appropriate to use in situ and epidemiologic data alone to define a pass/fail criterion with respect to a specific injury. When the pass/fail criterion for ASTM F1045 (275 g average peak linear acceleration across three trials for a given location and 300 g maximum for any single trial) was adopted, there was extensive data which suggested that this threshold would reduce the occurrence of skull fracture and subdural hematoma [4].

At this time, there are not sufficient data to facilitate association of concussive injury with specific biomechanical characteristics of head impact. Such associations have been made in the literature [28–33] but none have been verifiable based on in situ measures of head impact acceleration. Ongoing and future work at various institutions is aimed at quantifying the biomechanical properties of head impact acceleration that are associated with clinically diagnosed concussive injury. Based on this work, a recommendation could be made for an appropriate pass/fail criterion that would be related to the incidence or risk of mTBI for a given set of head impact characteristics. These characteristics may include peak linear acceleration, peak rotational acceleration, impact duration, impact location, or a combination of these, and possibly other variables, such as a weighed principle component score previously proposed by the authors [17].

In an effort to reduce the incidence of mTBI in ice hockey, lowering the pass/fail criterion for the ASTM ice hockey helmet test standard has been discussed. Given that the mechanisms of mTBI are not fully understood and that the impact characteristic tested in ASTM F1045 (peak linear acceleration) may not be the sole or even primary contributing factor to mTBI [34–40], lowering the threshold does not seem appropriate with respect to reducing the incidence of mTBI. Additionally, a lower pass/fail criterion (or a higher impact energy) would likely lead to a thicker helmet and thus might increase the rotational accelerations of the head by increasing the length of the lever arm between the CG of the head and the outer surface of the helmet shell. Bishop [41] demonstrated the inadequacy of hockey helmets in preventing rotational accelerations and considering the rotational kinematics of the headform when setting helmet test standards has been suggested [14]. However, to the authors' knowledge, no repeatable method of simulating rotational accelerations in the laboratory for the purpose of standards testing has been validated. The NOCSAE linear impactor test method [26] simulates rotational accelerations but the rotations are dependent on the stiffness of the anthropomorphic neck and geometric orientation of the headform with respect to the impactor, which can be difficult to standardize across test facilities.

The data presented herein demonstrate that current ice hockey test standards such as ASTM F1045–99 may not be representative of the highest magnitude linear accelerations sustained in situ. However, current standards led to the manufacture of high quality helmets that have been proven to limit localized loading

on the head, which is associated with skull fracture and subdural hematoma [42]. These helmets have undoubtedly protected many players from what would have been a catastrophic head injury had they not been wearing a helmet [1]. Ongoing and future research regarding human head impact tolerance will lead to improved helmet designs and testing protocols in an effort to further reduce the incidence and severity of sports related head and brain injury.

Acknowledgments

This work was supported in part by an award from the National Center for Medical Rehabilitation Research at the National Institute of Child Health and Human Development at the National Institutes of Health (Grant No. R44HD40473) and research and development support from Riddell, Inc. (Chicago, IL). Analog Devices (Cambridge, MA) provided sensors and expertise in sensor integration. The writers acknowledge Reebok for providing the helmets used in both the on ice and laboratory testing. The writers also appreciate and acknowledge their colleagues David Halstead and Mel Cook, Southern Impact Research Center, Rockford, TN, for performing the laboratory testing, Jeff Frechette ATC, Dartmouth College Sports Medicine, Hanover, NH, for invaluable insight and daily monitoring of the in vivo data acquisition system, Siena Calabro, Simbex, Lebanon, NH, for her role in manufacturing the instrumented helmets, and Sol Diamond Ph.D., Thayer School of Engineering at Dartmouth College, for his assistance with data analysis and presentation. Additionally, the writers would like to thank the players, athletic training staff, equipment staff, and coaches from the Dartmouth College men's hockey team for their ongoing support and cooperation.

Disclosure

Authors Richard M. Greenwald and Jeffrey J. Chu have a financial interest in the Head Impact Telemetry (HIT) System used in this study to collect in-vivo head impact data. The technology is currently sold commercially by Riddell, Chicago IL, USA under the name Sideline Response System.

References

[1] Biasca, N., Wirth, S., and Tegner, Y., "The Avoidability of Head and Neck Injuries in Ice Hockey: A Historical Review," *Br. J. Sports Med.*, Vol. 36, 2002, pp. 410–427.

[2] Odelgard, B., "The Development of Head, Neck, and Face Protectors for Ice Hockey Players," *Safety in Ice Hockey, ASTM STP 1050*, Castaldi, C. R., and Hoerner, E. F., Eds., ASTM International, West Conshohocken, PA, 1989, pp. 164–183.

[3] Hoshizaki, T. B., and Brien, S. E., "The Science and Design of Head Protection in Sport," *Neurosurgery*, Vol. 55, 2004, pp. 956–967.

[4] Gurdjian, E. S., Roberts, V. L., and Thomas, L. M., "Tolerance Curves of Acceleration and Intracranial Pressure and Protective Index in Experimental Head Injury," *J. Trauma*, Vol. 6, 1966, pp. 600–604.

[5] Guskiewicz, K. M., Marshall, S. W., Bailes, J., McCrea, M., Cantu, R. C., Randolph, C., and Jordan, B. D., "Association Between Recurrent Concussion and Late-life Cognitive Impairment in Retired Professional Football Players," *Neurosurgery*, Vol. 57, 2005, pp. 719–726.

[6] Guskiewicz, K. M., Marshall, S. W., Bailes, J., McCrea, M., Harding, H. P., Jr., Matthews, A., Mihalik, J. R., and Cantu, R. C., "Recurrent Concussion and Risk of Depression in Retired Professional Football Players," *Med. Sci. Sports Exercise*, Vol. 39, 2007, pp. 903–909.

[7] Bazarian, J. J., McClung, J., Shah, M. N., Cheng, Y. T., Flesher, W., and Kraus, J., "Mild Traumatic Brain Injury in the United States, 1998–2000," *Brain Inj*, Vol. 19, 2005, pp. 85–91.

[8] Jorgensen, U., and Schmidt-Olsen, S., "The Epidemiology of Ice Hockey Injuries," *Br. J. Sports Med.*, Vol. 20, 1986, pp. 7–9.

[9] NIH Consensus Development Panel on Rehabilitation of Persons with Traumatic Brain Injury, "Rehabilitation of Persons with Traumatic Brain Injury," *J. Am. Med. Assoc.*, Vol. 282, 1999, pp. 974–983.

[10] National Center for Injury Prevention and Control, *Report to Congress on Mild Traumatic Brain Injury in the United States: Steps to Prevent a Serious Public Health Problem*, Centers for Disease

Control and Prevention, Atlanta, GA, 2003.

[11] Bender, S. D., Barth, J. T., and Irby, J., "Historical Perspectives," in *Traumatic Brain Injury in Sports: An International Neuropsycological Perspective*, Lovell, M. R., Echemendia, R. J., Barth, J. T., and Collins, M. W., Eds., Swets & Zeitlinger, Lisse, 2004, pp. 3–21.

[12] Guskiewicz, K. M., Bruce, S. L., Cantu, R. C., Ferrara, M. S., Kelly, J. P., McCrea, M., Putukian, M., and Valovich McLeod, T.C., "National Athletic Trainer's Association Position Statement: Management of Sport-Related Concussion," *J. Athl. Train.*, Vol. 39, 2004, pp. 280–297.

[13] International Ice Hockey Federation (IIHF), *Official Rule Book 2006–2010*, IIHF, Zurich, 2006.

[14] Hoshizaki, T. B., "The Relationship between Helmet Standards and Head Protectors Safety in Ice Hockey," in *IIHF International Symposium on Medicine and Science in Hockey*, Biasca, N., Montag, W.-D., and Gerber, C., Eds., University of Zürich, Zürich, 1995, pp. 90–95.

[15] Brolinson, P. G., Manoogian, S., McNeely, D., Goforth, M., Greenwald, R., and Duma, S., "Analysis of Linear Head Accelerations from Collegiate Football Impacts," *Curr. Sports Med. Rep.*, Vol. 5, 2006, pp. 23–28.

[16] Duma, S. M., Manoogian, S. J., Bussone, W. R., Brolinson, P. G., Goforth, M. W., Donnenwerth, J. J., Greenwald, R. M., Chu, J. J., and Crisco, J. J., "Analysis of Real-time Head Accelerations in Collegiate Football Players," *Clin. J. Sport Med.*, Vol. 15, 2005, pp. 3–8.

[17] Greenwald, R., Gwin, J., Chu, J., and Crisco, J., "Head Impact: Severity Measures for Evaluating Mild Traumatic Brain Injury Risk Exposure," *Neurosurgery* Vol. 62, No. 4, 2008, pp. 789–798.

[18] Schnebel, B., Gwin, J. T., Anderson, S., and Gatlin, R., "In Vivo Study of Head Impacts in Football: A Comparison of National Collegiate Athletic Association Divison I vs. High School Impacts," *Neurosurgery*, Vol. 60, 2007, pp. 490–495.

[19] Mihalik, J. P., Bell, D. R., Marshall, S. W., and Guskiewicz, K. M., "Measurement of Head Impacts in Collegiate Football Players: An Investigation of Positional and Event-type Differences," *Neurosurgery*, Vol. 61, 2007, pp. 1229–1235.

[20] Funk, J. R., Duma, S. M., Manoogian, S. J., and Rowson, S., "Biomechanical Risk Estimates for Mild Traumatic Brain Injury," *Annu. Proc. Assoc. Adv. Automot. Med.*, Vol. 51, 2007, pp. 343–361.

[21] Halstead, P. D., Alexander, C. F., Cook, E. M., and Drew, R. C. "Hockey Headgear and the Adequacy of Current Design Standards," *Safety in Ice Hockey, ASTM STP 1341*, Ashare, A. B., Ed., ASTM International, West Conshohocken, PA, 2000, pp. 93–100.

[22] ASTM Standard F1045–99, "Standard Performance Specification for Ice Hockey Helmets," *Annual Book of ASTM Standards*, ASTM International, West Conshohocken, PA, 1999.

[23] Crisco, J. J., Chu, J. J., and Greenwald, R. M., "An Algorithm for Estimating Acceleration Magnitude and Impact Location Using Multiple Non-Orthogonal Single-Axis Accelerometers," *J. Biomech. Eng.*, Vol. 126, 2004, pp. 849–854.

[24] Chu, J. J., Beckwith, J. G., Crisco, J. J., and Greenwald, R. M., "A Novel Algorithm to Measure Linear and Rotational Head Acceleration Using Single-Axis Accelerometers," *J. Biomech.*, Vol. 39, 2006, pp. S153–6514.

[25] Gwin, J. T., Chu, J. J., and Greenwald, R. M., "Head Impact Telemetry System for Measurement of Head Acceleration in Ice Hockey," *J. Biomech.*, Vol. 39, 2006, pp. S153–6512.

[26] NOCSAE Standard Linear Impactor Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear and Faceguards. National Committee on Standards for Athletic Equipment. Document #081–04m04 (*Proposed Status*). January 2006.

[27] ASTM Standard F513–00, "Standard Safety Specification for Eye and Face Protective Equipment for Hockey Players," *Annual Book of ASTM Standards*, ASTM International, West Conshohocken, PA, 2007.

[28] Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., and Waeckerle, J. F., "Concussion in Professional Football: Reconstruction of Game Impacts and Injuries," *Neurosurgery*, Vol. 53, 2003, pp. 799–814.

[29] Newman, J. A., "A Generalized Acceleration Model for Brain Injury Threshold (GAMBIT)," *Proceedings of the IRCOBI Conference on Biomechanics of Impacts*, Zurich, 1986.

[30] Newman, J. A., Shewchenko, N., and Welbourne, E., "A Proposed New Biomechanical Head Injury Assessment Function–The Maximum Power Index," *Proceedings of the 44th STAPP Car Crash Conference*, Society of Automotive Engineers, Warrendale, 2000.

[31] Mertz, H. J., and Prasad, P., "Injury Risk Curves for Children and Adults in Frontal and Rear Collisions," SAE Technical Paper No. 973318, Warrendale, 1997.

[32] Margulies, S. S., and Thibault, L. E., "A Proposed Tolerance Criterion for Diffuse Axonal Injury in Man," *J. Biomech.*, Vol. 25, 1992, pp. 917–923.

[33] Zhang, L., Yang, K. H., and King, A. I., "A Proposed Injury Threshold for Mild Traumatic Brain Injury," *J. Biomech. Eng.*, Vol. 126, 226–236, 2004.

[34] Ommaya, A. K., and Gennarelli, T. A., "Cerebral Concussion and Traumatic Unconsciousness," *Brain*, Vol. 97, 1974, pp. 633–654.

[35] Ommaya, A. K., Hirsch, A. E., and Martinez, J. L., "The Role of Whiplash in Cerebral Concussion," *Proceedings of the 33rd Stapp Car Crash Conference*. Society of Automotive Engineers, Warrendale, 1966.

[36] DiMasi, F., Marcus, J., and Eppinger, R. H., "3-D Anatomic Brain Model of Relating Cortical Strains to Automotive Crash Loadings," *Proceedings of the 13th International Technical Conference on Experimental Safety Vehicles*, Paris, France, 1993, Paper No. 91–S8-0–11.

[37] Hodgson, V. R., Thomas, L. M., and Khalil, T. B., "The Role of Impact Location in Reversible Cerebral Concussion," *Proceedings of the 27th Stapp Car Crash Conference*, 1983, SAE Paper No. 831618.

[38] Zhang, L., Yang, K. H., and King, A. I., "Comparison of Brain Responses Between Frontal and Lateral Impacts by Finite-Element Modeling," *J. Neurotrauma*, Vol. 18, 2001, pp. 21–30.

[39] Kleiven, S., "Influence of Impact Direction on the Human Head in Prediction of Subdural Hematoma," *J. Neurotrauma*, Vol. 20, 2003, pp. 365–379.

[40] Scott Delaney, J., Puni, V., and Rouah, F., "Mechanisms of Injury for Concussions in University Football, Ice Hockey, and Soccer: A Pilot Study," *Clin. J. Sport Med.*, Vol. 16, 2006, pp. 162–165.

[41] Bishop, P. J., "Head Protection in Sport with Particular Application to Ice Hockey," *Ergonomics*, Vol. 19, 1976, pp. 451–464.

[42] Bishop, P. J. "The Effectiveness of Hockey Helmets in Limiting Localized Loading on the Head," *Safety in Ice Hockey, ASTM STP 1212*, Castaldi, C. R., Bishop, P. J., and Hoerner, E. F., Eds., ASTM International, West Conshohocken, PA, 1993. Vol. 2, 175–182.