Validation of a Helmet-Based System to Measure Head Impact Biomechanics in Ice Hockey

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ABSTRACT

ALLISON, M. A., Y. S. KANG, J. H. BOLTE, M. R. MALTESE, and K. B. ARBOGAST. Validation of a Helmet-Based System to Measure Head Impact Biomechanics in Ice Hockey. *Med. Sci. Sports Exerc.*, Vol. 46, No. 1, pp. 115–123, 2014. **Purpose**: This study aimed to quantify differences between head acceleration measured by a helmet-based accelerometer system for ice hockey and an anthropometric test device (ATD) to validate the system's use in measuring on-ice head impacts. **Methods**: A Hybrid III 50th percentile male ATD head and neck was fit with a helmet instrumented with the Head Impact Telemetry (HIT) System for hockey and impacted at various speeds and directions with different interfaces between the head and helmet. Error between the helmet-based and reference peak accelerations was quantified, and the influence of impact direction and helmet–head interface was evaluated. Regression equations were used to reduce error. System-reported impact direction was validated. **Results**: Nineteen percent of impacts were removed from the data set by the HIT System processing algorithm and were not eligible for analysis. Errors in peak acceleration between the system and ATD varied from 18% to 31% and from 35% to 64% for linear and rotational acceleration, respectively, but were reduced via regression equations. The relationship between HIT System and reference acceleration varied by direction ($P < 0.001$) and head–helmet interface ($P = 0.005$). Errors in impact azimuth were approximately 4%, 10%, and 31% for side, back, and oblique back impacts, respectively. **Conclusions**: This is the first comprehensive evaluation of peak head acceleration measured by the HIT System for hockey. The HIT System processing algorithm removed 19% of the impacts from the data set, the correlation between HIT System and reference peak resultant acceleration was strong and varied by head surface and impact direction, and the system error was larger than reported for the 6-degree-of-freedom HIT System for football but could be reduced via calibration factors. These findings must be considered when interpreting on-ice data. **Key Words:** mTBI, CONCUSSION, HEAD INJURY, HEAD ACCELERATION, KINEMATICS, SENSORS

With 1.7 million traumatic brain injuries (TBI) seen in emergency departments, the majority of those being considered "mild" (11), and an estimated 1.6–3.8 million sports- and recreation-related mild TBI (mTBI) (5) each year, the prevention of these injuries has been declared by the Centers for Disease Control and Prevention as a research priority (28). Concern regarding these injuries is heightened as recent data suggest that long-term neurological consequences can exist (17,31). Furthermore, research in a swine model has shown that, even in cases of mTBI, with the animal quickly recovering to normal-appearing behavior, permanent damage to the brain can still occur (4).

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To develop countermeasures to prevent these injuries from happening, it is important to understand the biomechanical inputs that lead to mTBI. Contact sports carry an increased risk of these injuries and provide an ideal means by which to study mTBI in humans as these are real-world scenarios in which the head regularly experiences accelerations that can lead to injury, and the athletes are a defined cohort on which to take baseline measures. Furthermore, in many of these sports, athletes already wear helmets, and instrumentation exists to measure head accelerations via the helmet (10). Thus, there is an implementable way to study head impact biomechanics during normal play in contact sports with helmeted athletes.

One example of helmet instrumentation that measures head accelerations during normal play is the Head Impact Telemetry (HIT) System for ice hockey (Simbex LLC, Lebanon, NH), which has been previously used with collegiate as well as youth hockey players during on-ice play (3,20–23,27). The HIT System for hockey consists of six single-axis linear accelerometers embedded in the padding of the helmet in a spring-loaded manner to encourage engagement of the sensors to the head (Fig. 1). Similar HIT System instrumentation has been

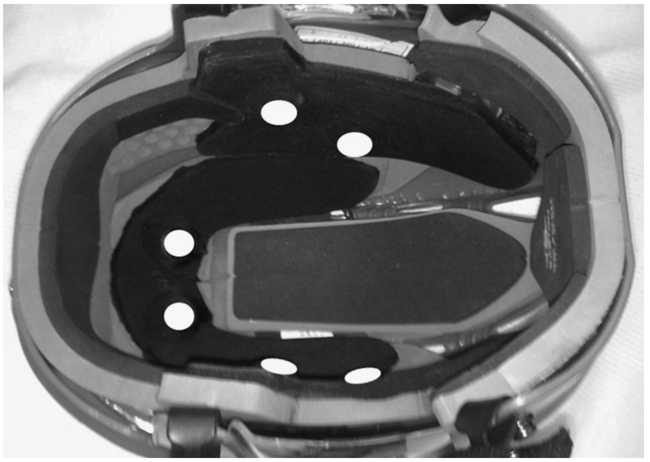

FIGURE 1—Accelerometer locations in the helmet indicated by white oval markings. The two accelerometer locations shown in the top of the photo are mirrored by the two accelerometer locations in the bottom of the photo.

developed for use in football helmets (1,18,29), boxing headgear (2), and soccer headgear (16). However, the football system differs from the system for ice hockey both in construction and in processing algorithm. Specifically, the orientation of the accelerometers in the HIT System for ice hockey is tangential to the head, whereas they are oriented normal to the head in the HIT System for football. The variation in accelerometer placement and orientation results in a different algorithm from which the three orthogonal linear and rotational accelerations of the head are calculated.

Data collected using the HIT Systems for ice hockey and football have been used to compare head impact magnitudes based on playing position, sex, awareness of impending impact, cervical muscle strength, and sport specific scenarios (3,7,20–22,24,25,27,32). Furthermore, researchers have attempted to introduce new injury thresholds and criteria based on data collected with these instrumented helmets. A proposed injury criterion, the HIT Severity Profile, was developed using HIT System data via a principal component analysis of linear acceleration, rotational acceleration, Gadd Severity Index, and Head Injury Criteria (14). Concussion risk curves for football players based on peak resultant linear head acceleration have also been developed using data collected via the HIT System for football (12,13,30).

Critical to the success of these efforts to quantify head biomechanics is the ability of the helmet-mounted instrumentation to accurately estimate the accelerations (angular and linear) of the center of gravity (CG) of the head. Validation of the HIT System for football (1,18,29), boxing (2), and soccer (16) has been previously published. These studies have all used a similar methodology, fitting the instrumentation to an anthropometric test device (ATD) head and impacting the head in some manner to compare the helmet-based instrumentation measure to the reference head acceleration measured via sensors rigidly attached to the ATD head. However, data on the validation of the HIT System for ice hockey are extremely limited and do not include details on test methodology or data analysis (15). Because of the differences in the accelerometer orientation, processing algorithm, and helmet shape for ice hockey compared with these other systems, a comprehensive validation on the HIT System for ice hockey is needed. Therefore, the objective of this study was to compare the peak head acceleration measured by the HIT System for ice hockey with reference peak head acceleration. We accomplished this objective by subjecting a hockey-helmeted ATD head and neck complex to repeated head impacts of various intensities and directions and then by comparing the HIT System–reported peak head acceleration with the peak acceleration measured at the CG of the ATD head.

METHODOLOGY

A Hybrid III (HIII) 50th percentile male ATD head and neck with the 3-2-2-2 accelerometer array (26) was rigidly mounted at T1. Rigid mounting of the ATD head and neck was chosen to reach the desired range of resultant accelerations within the limits of impact velocities able to be generated by the linear impactor and to simulate a large effective mass of the torso. Resultant accelerations from this approach were similar to those test setups that used a sliding mount for T1 (29); however, they were achieved with lower impact velocities. A large Easton S9 hockey helmet (Easton-Bell Sports Inc., Van Nuys, CA) instrumented with the HIT System for ice hockey was fit to the ATD head. USA Hockey guidelines were adapted by marking the HIII head with approximate eyebrow locations and then centering the helmet on the head with the "rim" one finger width above the eyebrows (33). This was the neutral position, to which the helmet was aligned before each impact. The helmet had an Easton S9 facemask attached, as hockey players younger than 18 years are required by USA Hockey rules to wear such a facemask.

The HIT System for ice hockey consists of six linear single-axis accelerometers, oriented tangentially to the head. A spring between the helmet shell and each accelerometer's housing is designed to enhance contact between the accelerometer and the head (18). When one of the accelerometers detects an acceleration of at least 10*g*, the system is triggered to collect 40 ms of data, 8 ms before the threshold is reached and 32 ms after at 1000 Hz (22). Data from the six accelerometers are passed through a 0.5-Hz AC hardware filter and a 400-Hz low-pass filter and then automatically and wirelessly transferred to a sideline data storage system. The data are then uploaded to a Simbex server, and based on the individual accelerometer measurements, an algorithm proprietary to Simbex, which is not available to end users (including the authors of the current study), is used to calculate linear and rotational acceleration at an estimated center of gravity of the head based on rigid body dynamics and iterative optimization (6). The processed impact data are then sent back to the end users. The general theory and equations used to calculate linear acceleration based on data from six single-axis accelerometers oriented normal to the head (as used in the football application of this system) on a hemispherical object were published in 2004 (8). The theoretical approach was subsequently updated to calculate linear and rotational acceleration based on data from six accelerometers oriented tangential to the head, as is the case for the HIT System for ice hockey, and published in abstract form (6). However, the raw accelerometer data and the processing algorithm itself are unavailable to end users of the HIT System for ice hockey.

A pneumatic linear impactor, weighing 23.9 kg, was used to contact the helmet at various speeds and in different directions (Fig. 2). Reported impacting speeds were measured immediately before impact with the impactor in free flight. One of two ultrahigh molecular weight polyethylene (UHMWPE) impacting surfaces were used in each test, both of which were cylindrical with the flat end of the cylinder contacting the helmet. Both cylinders were 2 inches thick from the impacting surface to the site of attachment on the impactor. One impacting surface had a 3.25-inch diameter and weighed 0.4 kg, and the other was 4 inches in diameter and weighed 0.6 kg. For both impacting surfaces, the flat surface of the cylinder had rounded edges. The specific impacting surface used for each impact direction was chosen based on helmet geometry at the site of impact to avoid impacts centered on geometrical irregularities of the helmet. In previous validations of the HIT System for football (1,29), a layer of foam was interposed between the UHMWPE impacting surface and the impactor because helmet-to-helmet impacts are common in football. Helmet-to-helmet impacts have not been shown to be a primary cause of concussion in ice hockey (9). Instead, the UHMWPE impacting surface without foam backing is more in line with head contact to the boards or ice.

In the first phase of testing, impacts were conducted at 2, 3.5, and 5 $m \cdot s^{-1}$ in the front, back and side impact directions. The impacting speeds were chosen to produce accelerations across the range of measures observed during on-ice play (20). To analyze the effect of the interface between the ATD head and the helmet, three different head surfaces were tested: a nylon skull cap to mimic previous validation efforts on the football HIT System (1,29), a dry human hair wig adhered to the ATD head using a strong double-sided tape that kept the wig from displacing relative to the ATD head, and the same wig sprayed with water to simulate perspiration. Three to five impacts were performed per speed–direction–head surface combination.

In the second phase of testing, an expanded test matrix was performed. We observed youth hockey teams in play and practice and found that player hair is wet during play. Thus, we tested only the wet wig condition in phase 2, with five repeat tests conducted at each of five impact directions (front, back, side, oblique back-side, or oblique front-side) and each of four speeds (1.5, 2.5, 3.75, or 5 $m \cdot s^{-1}$). In oblique back impacts, the impact vector was in the axial plane and angled 30° from the sagittal plane. Similarly, oblique front impacts were angled 30° from the sagittal plane (Fig. 2). The same test matrix was performed on two identical sets of HIT System instrumentation to assess inter-HIT System variability.

ATD-collected acceleration time histories were processed with a CFC 1000 filter, and rotational accelerations were calculated from the nine accelerometer array via the process outlined by Padgaonkar et al. (26). The peak values of the ATD and HIT System resultant linear and angular head acceleration for the same impact were compared. The correlation between the HIT System–measured peak acceleration and the ATD peak head acceleration was quantified using two regression techniques: a linear fit and a power fit for each impact direction. The quality of the regression was assessed using coefficient of determination (R^2).

All statistical analysis was performed using ANOVA for unbalanced data carried out in SAS 9.3 (SAS Institute Inc., Cary, NC). The relationship between the primary outcome measure (ATD peak resultant acceleration) and the HIT System peak resultant acceleration was assessed via ANOVA with impact direction, HIT System number (one or two),

FIGURE 2—Test setup for the side (*left*), oblique back (*middle*), and back (*right*) impact directions.

and head–helmet interface (for the first phase of testing) included in the model as categorical variables. The effect of these categorical variables on the interaction between peak HIT System–calculated and ATD-measured peak resultant acceleration was assessed.

Two types of error were calculated to compare the HIT System peak resultant acceleration to the ATD peak resultant acceleration. The first was total percent error (equation 1), that is, the absolute difference between the two measures, expressed as a percentage of the Hybrid III measure. The second was the percent error of the calibrated data (equations 2a and 2b); in other words, the error was recalculated after the linear (equation 2a) and the power (equation 2b) regression equations were applied to the HIT System measures. The average and standard deviations of these errors were calculated, stratified by impact direction

$$\text{percent error} = \frac{|\mathrm{HIT_{peak}} - \mathrm{HIII_{peak}}|}{\mathrm{HIII_{peak}}} \times 100 \quad [1]$$

$$\text{percent error of linear relationship calibrated data} = \frac{|(a \times \mathrm{HIT_{peak}} + b) - \mathrm{HIII_{peak}}|}{\mathrm{HIII_{peak}}} \times 100 \quad [2a]$$

where a and b are coefficients of the linear regression equation

$$\text{percent error of power relationship calibrated data} = \frac{|(a \times \mathrm{HIT^{b}}_{\mathrm{peak}}) - \mathrm{HIII_{peak}}|}{\mathrm{HIII_{peak}}} \times 100 \quad [2b]$$

where a and b are coefficients of the power regression equation.

The impact direction calculated by the HIT System, which includes a categorical description of direction (side, front, top, and back) and the estimated azimuth and elevation angles of each impact, was compared with the actual impact direction for each test. As the attachment of T1 was fixed relative to the impactor and the line of action of the impactor was known, the actual impact direction was initially measured on setup and reconfirmed before each test.

In addition to calculating the head acceleration and impact direction, the HIT System for ice hockey algorithm determines whether an impact is considered "valid." There are two reasons that an impact may not be considered valid: 1) the resultant linear acceleration is less than 10*g* or 2) based on rigid body dynamics, the acceleration pulse does not have characteristics of an impact to a helmeted head. The purpose of the latter is to remove data resulting from occurrences such as a player throwing his or her helmet down on the bench. Impacts that, based on the algorithm, fall into one of these categories are removed from the processed data set and are not normally sent back to the end user. In this test series, we recorded the time of each impact performed and were able to confirm that raw data were collected for all impacts by viewing the data before it was wirelessly uploaded to the server for processing. Through cooperation with the HIT System manufacturer, we were able to obtain those data that were removed from processing per the reasons outlined above and compare them to data from similar impacts (having the same speed and direction) that were not removed from the data set.

RESULTS

In phase 1 testing, statistical analysis confirmed a significant effect of head–helmet interface on the relationship between the reference and the HIT System–reported peak accelerations ($P = 0.005$). This led to the decision to use the most realistic head–helmet interface; hence, the wet human hair wig was used during phase 2 of testing.

In phase 2 testing, 48 of the 218 impacts were removed from the data set by the HIT System processing algorithm. ATD-measured peak resultant acceleration showed that 8 of these 48 impacts were lower than the 10*g* threshold and therefore accurately removed according to the lower threshold defined in the algorithm. Of the 40 impacts remaining, 28 (70%) of these were front and oblique front impacts, representing 53% of the data for these impact directions. Thus, with more than half of data in the frontal and frontal oblique impact directions removed by the HIT algorithm, the remainder of the results section will focus exclusively on the 139 impacts in the side, back, and oblique back directions.

The relationship between HIT System–measured and reference peak head acceleration did not vary between the two sets of HIT System instrumentation ($P = 0.49$); therefore, the data presented is from both sets of instrumentation combined. The relationship between peak HIT System–measured and reference head acceleration varied by impact direction (Fig. 3, Table 1). Specifically, the relationship between peak reference acceleration and HIT System–measured acceleration significantly differed for back and oblique impacts ($P < 0.001$) and side and oblique back impacts ($P < 0.001$). The relationship between peak reference acceleration and HIT System–measured acceleration was not statistically different for side and back impacts ($P = 0.08$).

Of note, the intercepts associated with the linear fits were not equal to zero, and as a result, a power fit was also explored. The power fit through zero improved the correlation between the reference acceleration and the HIT System–measured acceleration for all impact directions for peak linear acceleration and for all impact directions except side for peak rotational acceleration (Table 1).

For peak resultant linear acceleration, the average error between the HIT System for ice hockey and the reference acceleration for various impact directions ranged from 18% to 31% for the raw data. Applying the linear regression equations (Table 1) to the HIT System measures reduced these errors to 7%–27%. The smallest average percent errors were for the side and back impact directions (Table 2). For peak resultant rotational acceleration, average errors ranged from 35% to 64% and from 13% to 38% for the

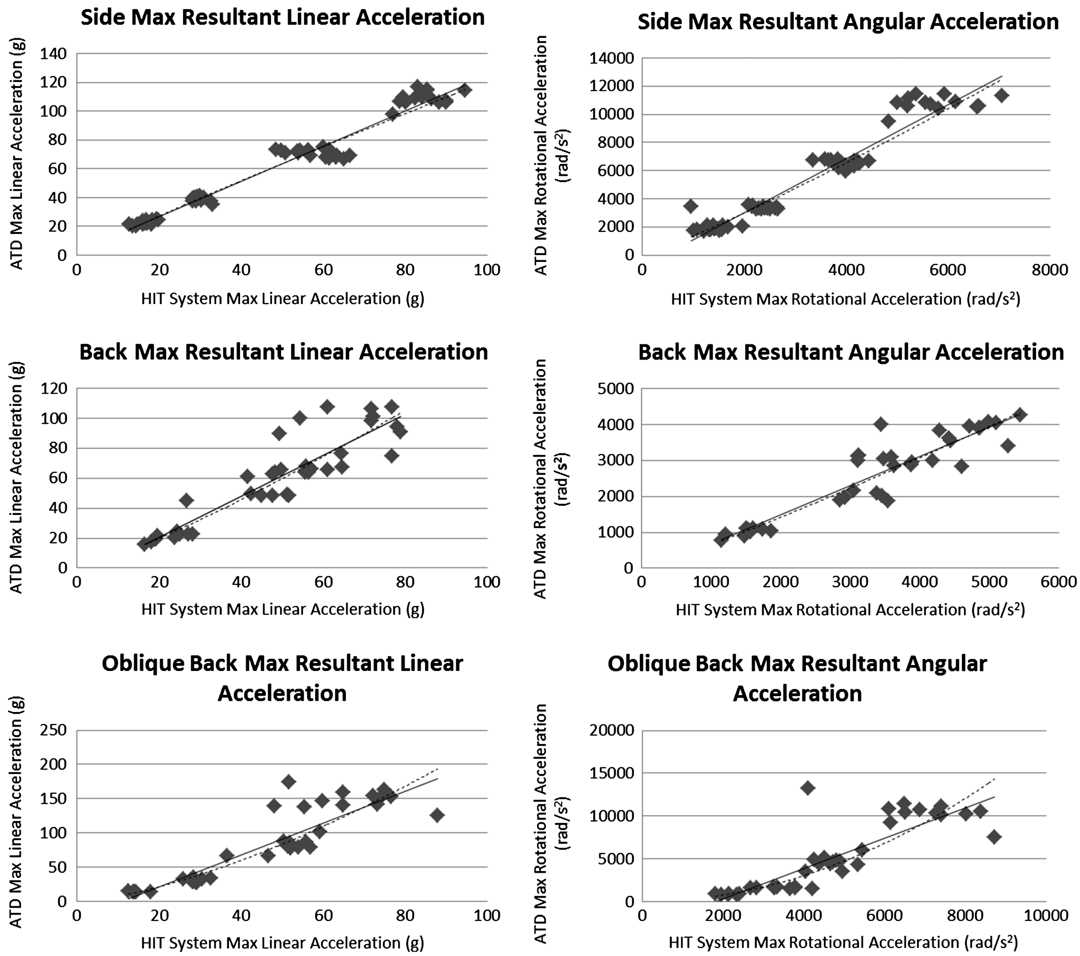

FIGURE 3—Comparison between peak resultant linear (*left*) and rotational (*right*) acceleration as measured by the ATD and by the HIT System for side (*top*), back (*middle*), and oblique back (*bottom*) impacts. Each data point represents a single impact, with HIT System measure on the abscissa and ATD measure on the ordinate. The solid line corresponds to the linear regression relationship, and the dashed line corresponds to the power regression.

raw and linearly calibrated data, respectively (Table 2). The average errors for data calibrated with the power regression equations were slightly smaller, ranging from 7% to 18% for peak linear acceleration and from 12% to 27% for peak rotational acceleration (Table 2).

The HIT System accurately determined the general categorical impact direction (front, back, or side) for 100% of side and back impacts and for 79% of oblique back impacts. Comparing the HIT System–reported impact azimuth with the actual impact direction, the mean ± SD errors were 4.0% ± 3.3%, 9.6% ± 5.1%, and 30.5% ± 15.2% for side, back, and oblique back impacts, respectively. For oblique back impacts, the HIT System–measured azimuth had systematic error with reported azimuth biased toward the back impact direction (Fig. 4).

DISCUSSION

This is the first comprehensive evaluation of the correlation between peak head acceleration measured by the HIT System for ice hockey and reference head acceleration as measured by an ATD. It is also the first to examine the effect of the interface between the head and the helmet surface on HIT System data. Several key findings were noted:

TABLE 1. Linear and power regression fit equations and their associated R^2 values.

	Direction	Regression Equation: Linear Fit	Linear Fit R^2	Regression Equation: Power Fit	Power Fit R^2
Linear acceleration	Side (n = 66)	$y = 1.21x + 3.14$	0.97	$y = 1.80x^{0.91}$	0.98
	Back (n = 36)	$y = 1.37x - 6.70$	0.81	$y = 0.53x^{1.21}$	0.89
	Oblique back (n = 38)	$y = 2.33x - 25.19$	0.81	$y = 0.25x^{1.49}$	0.92
	All combined (n = 140)	$y = 1.47x - 3.71$	0.73	$y = 0.88x^{1.11}$	0.88
Rotational acceleration	Side (n = 66)	$y = 1.92x - 860.57$	0.94	$y = 0.53x^{1.14}$	0.92
	Back (n = 36)	$y = 0.81x - 141.02$	0.85	$y = 0.30x^{1.11}$	0.90
	Oblique back (n = 38)	$y = 1.76x - 3123.39$	0.71	$y = 0.0002x^{1.99}$	0.84
	All combined (n = 140)	$y = 1.51x - 971.57$	0.64	$y = 0.40x^{1.12}$	0.60

TABLE 2. Absolute error ± SD between the HIT System–measured and reference peak resultant accelerations as calculated by equations 1, 2a, and 2b.

	Direction	Average Error	Calibrated Error: Linear Fit	Calibrated Error: Power Fit
Linear acceleration	Side (*n* = 66)	23% ± 8%	7% ± 5%	7% ± 6%
	Back (*n* = 36)	18% ± 13%	16% ± 11%	15% ± 11%
	Oblique back (*n* = 38)	31% ± 22%	27% ± 18%	18% ± 14%
	All combined (*n* = 140)	24% ± 15%	19% ± 15%	18% ± 14%
Rotational acceleration	Side (*n* = 66)	35% ± 12%	13% ± 11%	12% ± 10%
	Back (*n* = 36)	37% ± 22%	13% ± 11%	12% ± 11%
	Oblique back (*n* = 38)	64% ± 58%	38% ± 41%	27% ± 25%
	All combined (*n* = 140)	43% ± 35%	50% ± 48%	45% ± 38%

1) The HIT System for ice hockey's processing algorithm removed 19% of the impacts from the data set as they were identified as being perturbations not associated with an impact to a helmeted head, with a higher percentage of impacts to the facemask being removed; 2) the HIT System and the reference peak resultant linear and rotational accelerations are strongly correlated (all power fit R^2 values ≥0.84 for individual directions, Table 1) but not equivalent, and this correlation varies by head surface and impact direction; and 3) the magnitude of error associated with HIT System for ice hockey is larger than previously reported for the 6-degree-of-freedom HIT System for football (29) but has the potential to be reduced via impact direction-specific calibration factors. These results provide insight into how this technology can be practically used to quantify head biomechanics during normal on-ice play.

As stated in the methods, the HIT System's processing algorithm removes impacts from the data set if 1) the resultant linear acceleration is less than 10*g* or 2) based on rigid body dynamics, the acceleration pulse does not have characteristics of an impact to a helmeted head. The purpose of the latter is to remove data resulting from occurrences such as a player throwing his or her helmet on the bench. In this analysis, 19% of the impacts were removed from the data set. Through cooperation with the HIT System's manufacturer, we were able to obtain those removed data via matching the time stamp of impact. Many of these removed impacts were to the facemask, likely due to the irregular shape of the facemask as well as its nonrigid attachment to the helmet itself, causing the acceleration pulses recorded by the helmet instrumentation system to be atypical compared with impacts to the shell of the helmet. An inspection of the acceleration time histories of the removed impacts from impact directions other than the front and a comparison of impacts performed at the same speed and direction that were not removed from the data set did not identify any key characteristics of the pulse that were different: the magnitude, shape, and duration were similar. The details of the algorithm used to filter the data as well as the theory behind it are proprietary and unavailable to the authors making it difficult to identify the specific reason these impacts may have been removed. However, in a real-world situation, unless each impact is being uploaded, tracked, and verified as an on-ice occurrence while it is happening, if an impact was removed from the data set, the researchers would be unaware that it occurred. Given that this study evaluated a single-impact scenario across a range of speeds and directions, an on-ice analysis is necessary to understand how often impacts occur during play but are removed from the data set during processing. This will aid in understanding whether this finding affects the interpretation of on-ice data.

A comparison of peak linear and angular acceleration as calculated by the HIT System for ice hockey and the ATD highlighted that the error between the two measures reported

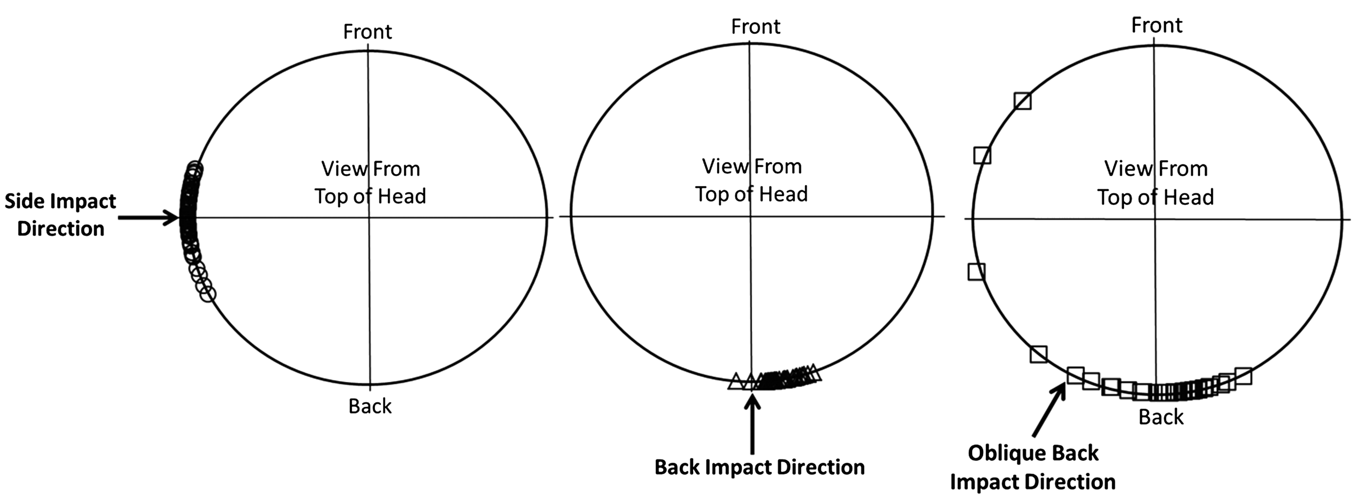

FIGURE 4—Comparison of HIT System–reported and actual impact azimuth for side (*left*), back (*middle*), and oblique back (*right*) impacts. Actual impact azimuth is indicated by the arrow, and system-reported azimuths for each impact are indicated by the markers.

herein is greater than that previously reported for the specialized version of the HIT System for football that includes twelve single-axis accelerometers organized into six orthogonal pairs (6-degree-of-freedom system). This can partially be attributed to differences in how error was calculated. In a validation study of the 6-degree-of-freedom system for football, Rowson et al. (29) included the sign of the difference between the HIT System and reference acceleration measures in their average error calculations, thus averaging values that were both positive and negative, resulting in a small overall error. Given that the error associated with this type of system appears to be random, it is not surprising that this approach led to an average error close to zero. In contrast, in the current study, the absolute value of the difference between the measures was included in the error calculations, calculating absolute error rather than relative error. As a result, the errors reported herein for the ice hockey HIT System are larger than those reported for the 6-degree-of-freedom football system, but the method of error calculation is more appropriate, particularly when assessing the possible error associated with a single on-field measurement from the HIT System. Other validation studies on HIT System instrumentation for football, boxing, and soccer did not report the average error between the instrumentation-measured acceleration and reference head acceleration as measured via the ATD (1,2,16,18). These include validations of the most widely used system, which is the HIT System for football that consists of six single-axis accelerometers (5-degree-of-free dom system). However, coefficients of determination reported herein are comparable with those reported, by impact location, in a validation study of the 5-degree-of-freedom system for football (1). In the future, the errors associated with this helmet instrumentation system for ice hockey must be accounted for in the analysis of real-world data, particularly when working with a small sample size or analyzing individual impacts.

Another key aspect of the HIT System for ice hockey performance highlighted in this study was that although the HIT System data strongly correlates with the ATD measures, that correlation is not one to one and varies by impact direction and head–helmet interface. One method to account for this variability, introduced herein, is to calculate regression equations to be used as calibration factors for the on-ice data. We have shown that this method has the potential to appreciably reduce measurement errors for the HIT System for ice hockey when applied on an impact direction-specific basis. The smallest errors were measured in side and back impacts that, of note, have been reported to account for close to 60% of all on-ice impacts in youth hockey players (20). However, this method requires researchers to accurately determine the impact direction, which adds a level of complexity to the data analysis. Many researchers who use this instrumentation also collect game film to confirm that the data are associated with true impacts and to gather information about the circumstances surrounding each impact (i.e., player awareness of impending impact) (7,22,25). This film may help confirm the impact direction, providing more information to help determine which calibration factor should be applied to the data. Further development of robust calibration equations should be subject of future study.

As mentioned previously, the HIT System for football is different from that for ice hockey, so this study's findings are not directly transferrable to the football instrumentation. However, given the growing use of the football system for research purposes, a validation study on the football instrumentation system calculating the absolute error as outlined herein and quantifying any differences in system performance with impact direction is critically important for researchers using this system, particularly for those studies which analyze data from small sample sizes, individual impacts, or a small number of injuries.

There are several limitations of this study which must be considered. First, rigidly mounting the HIII head and neck at the level of T1 is unlike on-ice conditions in which the torso can move. The rigid mounting simulates a torso with an infinite effective mass. Given the relatively large mass of the torso compared with the head, and the coupling of the two via the neck, the real-world scenario is likely more similar to an infinite effective mass, particularly in the initial milliseconds of the impact when the acceleration reaches its peak. In addition, ice hockey impacts frequently occur against the boards, and this coupling of the torso to the boards would further increase the effective mass of the torso. Furthermore, the peak acceleration occurs early in the time history. The high-speed video of the impact at the time of peak acceleration shows that the neck has flexed very little, if at all, and is not near the end of its range of motion. Thus, the contribution of torso inertia on the peak head acceleration is likely negligible.

Second, helmet fit varies among individuals as athletes may or may not wear their helmets with an ideal fit; some may find the helmet uncomfortable if it is too tight. The design of the HIT System is based on spring-loaded accelerometers, which are intended to maintain better contact between the head and the helmet so that head acceleration is measured rather than helmet acceleration (18). However, this is most likely not always the case so the more tightly the helmet is coupled to the head upon impact the better. In this analysis, the HIT System was evaluated under one fit condition, that is, a wet human hair wig with a Hybrid III 50th percentile male head and an Easton S9 size large helmet. Through pilot observations of how adolescent hockey players wear their helmets, we felt this scenario most closely mimicked the on-ice situation. In this study, we opted to use a helmet fit that, although realistic, most likely represents a worst-case scenario, using a large helmet that allowed some movement between the HIII head and the helmet rather than using the medium-sized helmet, which was extremely tight on the ATD head. USA Hockey guidelines specify that the helmet should be snug enough to prevent rotation (33), and our test setup ensured this criterion was met. As noted in the methods, the helmet was aligned to the neutral position before each impact; however, little displacement of the helmet relative to the ATD head occurred, particularly for side, back, and oblique back impacts.

Two aspects of helmet fit around the chin area in our tests must be noted. First, due to a lack of structure on the underside of the ATD chin, the chin strap could not be tightened against an anatomical structure to further couple the helmet to the head. Second, the chin pad on the facemask did not come in contact with the ATD chin (approximately a finger width apart) when the helmet was in the neutral position (as defined in the methods). In sum, the fit of the helmet on the cranium of the ATD appeared to mimic resistance to rotation prescribed by guidelines and observed in actual players; however, the lack of coupling of the helmet to the ATD in the chin region may have led to an increased ability of the helmet to translate upward if force was applied in that direction. Impacts to the facemask were likely those impacts that were most affected by this chin coupling, and these impacts were removed from the analysis.

Also, the outer shape of the hockey helmet has more protuberances than a football helmet and can deform upon impact. These characteristics likely influence how well the accelerometer is coupled to the head during an impact. The ability of the HIT System for ice hockey to accurately measure head acceleration may vary by these geometric and deformation parameters, but the evaluation of this variability was not the focus of the current analysis. Future studies should develop methods to rigorously quantify helmet fit and evaluate its effect on the HIT System performance.

Third, we chose a flat impacting surface (with rounded edges) to mimic surfaces such as the boards or the ice which are common surfaces of head impact during ice hockey. We note that the size of the impacting surface is infinitely smaller than the flat surface of the boards or ice. Although in our testing we did not observe the helmet-to-impactor contact area extending beyond the edges of the impactor surface, and thus our impact conditions appear to be applicable to impacts to large flat surfaces, a larger impacting surface would more accurately mimic this scenario. In hockey, players may also contact other objects such as another player's elbow or hockey stick. The impacting surface was not varied in this validation study. Other impact types may have distinguishing characteristics in their acceleration profiles that were not evaluated in this study.

Lastly, it is important to note that we only evaluated the peak values of linear and angular acceleration in the current study. Linear acceleration brain injury metrics such as HIC consider the acceleration time history, not just the peak value (34). Similarly, rotational kinematic metrics for injury also have been shown to be dependent on the shape of the acceleration pulse, not just the peak value (19,35). Thus, further analysis of the HIT System acceleration time history, relative to a suitable reference system, is required.

CONCLUSIONS

This study represents the first comprehensive evaluation of the correlation between peak head acceleration measured by the HIT System for ice hockey and reference head acceleration. Several key findings were noted. The processing algorithm for the HIT System removed 19% of the impacts from the data set due to the appearance of differences from impacts to a helmeted head. There is a correlation between HIT System peak resultant acceleration and reference acceleration; however, this relationship is not one to one, and it varies by impact direction and the interface between the ATD head and the helmet. The error associated with the HIT System for ice hockey was larger than those previously reported for an advanced HIT System for football (6-degree-of-freedom system), but these errors could potentially be reduced via impact direction-specific calibration factors. Obtaining head impact biomechanics via helmet-based sensors has the potential to contribute valuable real-world data to the biomechanics field as such approaches can relate head impact metrics with clinical outcomes of concussion, which are difficult to measure in other human surrogates such as cadavers and animals. It is essential, however, that the measurement error associated with such systems outlined herein be incorporated into analyses of such kinematic data obtained during normal on-ice play. It is important to note that this study evaluated a single-impact scenario across a range of speeds and directions. On-ice play is characterized by a diverse set of impact conditions with a variety of surfaces, some of which may be characteristically different from that evaluated herein. Future work should expand this work and characterize the influence of parameters such as impact scenario, helmet fit, and impacting surface on the magnitude and nature of differences between HIT System for ice hockey-measured acceleration and reference head acceleration.

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