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A comparison of the magnitude and duration of linear and rotational head accelerations generated during hand-, elbow- and shoulder-to-head checks delivered by hockey players

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ABSTRACT

Ice hockey has the highest rates for concussion among team sports in Canada. In elite play, the most common mechanism is impact to the head by an opposing player's upper limb, with shoulder-to-head impacts accounting for twice as many concussions as elbow- and hand-to-head impacts combined. Improved understanding of the biomechanics of head impacts in hockey may inform approaches to prevention. In this study, we measured the magnitude and duration of linear and rotational head accelerations when hockey players (n = 11; aged 21–25) delivered checks "as hard as comfortable" to the head of an instrumented dummy with their shoulder, elbow and hand. There were differences in both peak magnitude and duration of head accelerations across upper limb impact sites, based on repeated-measures ANOVA ($p < 0.005$). Peak linear head accelerations averaged 1.9-fold greater for hand and 1.3-fold greater for elbow than shoulder (mean values = 20.35, 14.23 and 10.55 g, respectively). Furthermore, peak rotational head accelerations averaged 2.1-fold greater for hand and 1.8-fold greater for elbow than shoulder (1097.9, 944.1 and 523.1 rad/s^2, respectively). However, times to peak linear head acceleration (a measure of the duration of the acceleration impulse) were 2.1-fold longer for shoulder than elbow, and 2.5-fold longer for shoulder than hand (12.26, 5.94 and 4.98 ms, respectively), and there were similar trends in the durations of rotational head acceleration. Our results show that, in body checks to the head delivered by varsity-level hockey players, shoulder-to-head impacts generated longer durations but lower magnitude of peak head acceleration than elbow- and hand-to-head impacts.

1. Introduction

Ice hockey is a popular sport, with approximately 1.7 million registered players worldwide (International Ice Hockey Federation (IIHF), 2017). While players often wear helmets and padding, the incidence of head impacts causing traumatic brain injury (TBI), including concussion, is greater in ice hockey than in other team sports (Cusimano et al., 2013; Hootman et al., 2007; Zuckerman et al., 2015).

The most common mechanism of concussion in ice hockey is player-to-player contact (Agel and Harvey, 2010; Cusimano et al., 2013; Hootman et al., 2007; Hutchison et al., 2015), most often involving impact to the head by the upper limb of the opposing player, and especially the shoulder. An analysis of video footage of head impacts surrounding 197 diagnosed concussions in the National Hockey League (NHL), which involved contact with an opponent but did not involve fighting, found that shoulder-to-head impacts accounted for 33% of concussions, elbow impacts caused 12%, and hand-to-head impacts caused 4% (Hutchison et al., 2015). An analysis of 25 concussions in men's university-level ice hockey found that shoulder impacts accounted for 32% of concussions, elbow impacts caused 28%, and none resulted from hand impacts (Delaney et al., 2014).

An improved understanding of the magnitude and duration of head accelerations generated during upper limb-to-head impacts in ice hockey may inform approaches to prevent TBI. During impact to the head, risk for TBI increases with increases in both the magnitude and duration of linear and rotational head acceleration, via the effect of these parameters on brain tissue stresses and strains (Fernandes and de Sousa, 2015; Gennarelli, 1983; Gennarelli and

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Thibault, 1982). The effects may be semi-independent, or relate to the high head velocities that can occur from low acceleration with long duration (Stemper et al., 2015). Our current understanding of the head accelerations accompanying impacts in hockey is based on two approaches, which each have their strengths and limitations: (a) mechanical test systems (Post et al., 2019; Rowson et al., 2015), which provide high resolution measures of head acceleration, but may not accurately simulate the complex dynamics of body-checks in hockey, and (b) acceleration data from helmet-mounted sensors during game play (Mihalik et al., 2012; Wilcox et al., 2014a), which have strong external validity, but limited accuracy based on current sensor technology (Allison et al., 2014; Allison et al., 2015).

In the current study, we sought to add new evidence by measuring the magnitude and duration of linear and rotational head accelerations, when hockey players delivered checks "as hard as comfortable" to the head of an instrumented dummy with their shoulder, elbow and hand. We hypothesized that the magnitude and duration of linear and rotational head accelerations would differ across upper limb contact sites. We expected, given the high prevalence of concussions in ice hockey from shoulder-to-head impacts (Delaney et al., 2014; Hutchison et al., 2015), that participants would deliver more severe head impacts with the shoulder than the elbow or hand.

2. Materials and methods

2.1. Participants

Eleven males participated, of mean age 22.6 years (SD = 1.4, range = 21–25), mean height 178.5 cm (SD = 6.5, range = 170.2–185.4) and mean weight of 83.2 kg (SD = 6.1, range = 72.7–93.2 kg). None had a documented muscle strain, joint sprain, bone fracture or TBI in the 6 months before participating. The experimental protocol was approved by the Research Ethics Committee of Simon Fraser University, and all participants provided written informed consent.

2.2. Body-checking dummy

A customized body-checking dummy (height = 172 cm; mass = 74 kg; Fig. 1A) simulated the checked player receiving head impacts. The dummy was similar to that described by Virani et al. (2017), except that a Hybrid III 50th-percentile male head and neck (Humanetics ATD, USA) replaced the foam core of the head and neck region of the dummy (Foster et al., 1977; Mertz, 1985). A CSA-certified caged helmet (Vector 08, CCM) was secured over the dummy's head. The foam torso and rubber skin of the dummy were taken from a commercially-available kickboxing dummy (BOB XL, Century Martial Arts, USA). The inner plastic tube of the kickboxing dummy was replaced with a sand-filled aluminum tube (20 cm diameter) onto which the Hybrid III head- and neck-form was bolted. The dummy was supported on a low-friction ball joint and by an overhead spring (resting length 63 cm, diameter 3.75 cm, 1157 N/m stiffness) that provided consistent positioning and rotational stiffness during impact. A cable was connected at one end to the spring, ran over a one-way climbing pulley (to prevent bounce-back from the dummy after delivery of the check), and connected at the other end to the mid-shoulder location of the dummy's torso (via a fall restraint harness secured rigidly to the dummy at the mid-shoulder location). Another cable connected the spring to a winch mounted on a force plate (FP4060-08, Bertec, USA), allowing monitoring and consistent baseline cable tension between trials.

The Hybrid III head was equipped with nine Endevco 7624C (Meggitt, USA) uniaxial accelerometers arranged in a 3–2–2–2 array for measuring the linear and rotational acceleration of the head centre-of-gravity (COG) during each trial (Padgaonkar et al., 1975). Each accelerometer sampled at 20 kHz with a SLICE Nano on-board DAQ system (DTS, USA). Appendix A details our accuracy evaluation of measured head COG acceleration.

2.3. Motion capture

We tracked participant upper limb kinematics with markers on padding over the acromion, elbow and back of the hand. Motion of the dummy head was tracked with markers on the helmet and left lateral cage. Kinematic data were collected at 640 Hz with an 8-camera passive motion capture system (Miqus M3, Qualisys, Sweden).

2.4. Experimental protocol

We designed our experiments to recreate impact scenarios that were common to ice hockey, and safe for the participants delivering the impacts. Accordingly, we instructed participants to deliver checks "as hard as comfortable" to the head of the dummy with the padded shoulder, elbow or hand, using checking styles that we observed to be common (Table 1; Fig. 1B and C) from an ongoing study of head impacts in regular season home games of the Simon Fraser University men's ice hockey team (Bruschetta et al., 2016). During shoulder checks, participants moved in a straight line to impact the head of the dummy with the lateral aspect of the shoulder. During elbow checks, participants moved their arm upward and laterally to impact the head of the dummy with the posterior aspect of the elbow. During hand impacts, participants delivered a jab punch to the head of the dummy while holding a hockey stick.

In each scenario, we included impacts to both the front and lateral aspect of the head, in order to test whether differences among upper limb contact sites in head impact severity depended on the location of head impact. In "frontal" impacts, the dummy faced the participant directly; in "lateral" impacts, the dummy was rotated 45° about its long axis. We instructed participants to contact a 4×4 cm coloured target secured to the anterior aspect (in frontal impacts) or anterior-lateral aspect (in lateral impacts) of the cage, at nose-height. In all trials, the dummy was inclined 20° from the vertical, toward the participant. For uniformity, participants delivered checks with their right upper limb. All trials were acquired on a vinyl floor with participants wearing running shoes. Participants wore shoulder pads (Supreme One.6, Bauer), elbow pads (Vapor APX2, Bauer), gloves (4R Pro, CCM) and shin pads (Supreme One.6, Bauer) that are designed for use in ice hockey, and a helmet (Bell Mirra Multi-Sport). In all trials, participants started by standing 1.4 m from the dummy, and (within this constraint) self-selected their speed at contact.

Each participant completed four repeated trials for each upper limb contact site and head contact site, in a blocked-randomized order. Participants rested approximately 3 min between trials, and practiced each type of check 1 to 3 times to familiarize themselves with the checking style and the dummy's physical attributes.

After testing, participants reported their: (a) comfort with shoulder, elbow and hand checks (on a 10-point Likert scale where 1 represented "extreme discomfort" and 10 represented "extreme comfort"); and (b) perceived similarity (realism) of the dummy and checking styles to those in an actual hockey game (using a 10-point Likert scale where 1 represented "extremely unrealistic" and 10 represented "extremely realistic").

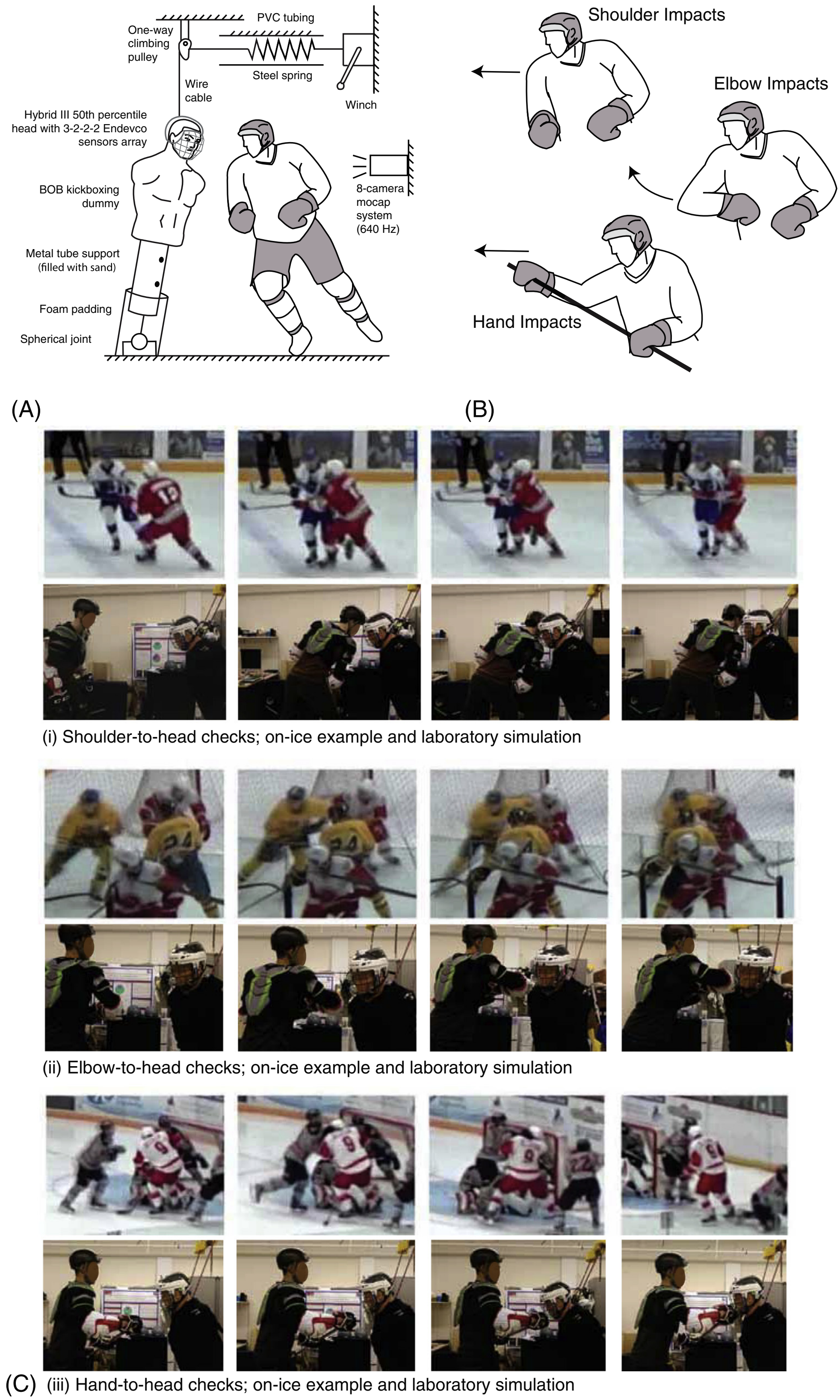

Fig. 1. Experimental setup with body-checking dummy. See text for additional details. **(A)** During the trials, participants delivered impacts to the head of customized dummy, containing a Hybrid III 50th percentile head and neck. **(B)** Checking styles during shoulder-, elbow- and hand-to-head impacts. **(C)** Snapshots of typical on-ice events and laboratory simulations of (i) shoulder-to-head impacts, (ii) elbow-to-head impacts and (iii) hand-to-head impacts.

Table 1
Instructions to participants for each impact scenario, and rationale for the instruction based on review of 40 real-life upper limb-to-head impacts in men's university ice hockey games (Bruschetta et al., 2016).

Impact scenario	Instructions to participants	Rationale for instruction based on real-life upper limb-to-head impacts from university games
Shoulder-to-head checks	Contact the dummy's head with the lateral aspect (side) of the shoulder	77% of cases (10/13) involved laterally-directed shoulder movement; in 62% of cases (9/13) shoulder movement was maintained in the horizontal plane In 75% of cases (10/13), skating speed of player delivering check was slow or at standstill when check was delivered*
Elbow-to-head checks	Contact the dummy's head with the lateral aspect of the elbow, using rapidly-increasing shoulder abduction while maintaining the elbow flexed at approximately 45°	82% of cases (9/11) involved laterally-directed elbow movement; 73% (8/11) of cases involved upward elbow movement In 91% of cases (10/11), skating speed of player delivering check was slow or at standstill when check was delivered*
Hand impacts	Impact the head while holding a hockey stick in both hands, using a "jab punch" motion similar to a cross-check. Contact the head with the glove and not the stick.	81% of cases (13/16) involved anteriorly-directed hand movement; in 63% of cases (10/16), hand movement was maintained in the horizontal plane In 94% of cases (15/16), skating speed of player delivering check was slow or at standstill when check was delivered*

* Justification for run-up distance of 1.4 m used in all trials.

2.5. Data processing

We filtered data from the nine Hybrid III head accelerometers with a CFC1000 filter (4th-order, recursive, low-pass Butterworth filter; 1650 Hz cut-off frequency), consistent with SAE J211-1 (Society of Automotive Engineers International, 1995). Linear head acceleration was calculated as the resultant from the three accelerometers mounted orthogonally at the head COG. Rotational head acceleration was calculated based on Padgaonkar's equations (Padgaonkar et al., 1975) for a 3–2–2–2 sensor array with the same sensor mounting dimensions as in our headform. The onset time for calculating times to peak acceleration was the instant that linear acceleration exceeded 3 g (National Highway Traffic Safety Administration, 2011).

We estimated upper limb contact velocity from the 3D resultant velocity of the corresponding motion capture marker at the instant of head contact (estimated from onset of steady movement of the dummy head markers). Velocities were calculated by applying a central finite difference algorithm to marker position data after filtering with a 4th-order, recursive, low-pass Butterworth filter having a 100 Hz cut-off frequency. All data processing was conducted with Matlab 2015b (The MathWorks Inc, USA).

2.6. Statistics

For each participant and impact scenario, we calculated average values (over the four repeated trials) of the magnitude and timing of peak linear head COG acceleration and peak rotational head acceleration from the Hybrid III sensors. We then used a full-factorial repeated measures ANOVA (JMP 12.2, SAS Institute Inc, USA) to test the main and interacting effects on these outcomes of upper limb contact site (hand, elbow or shoulder) and head contact site (front versus lateral). Each outcome variable was analyzed separately. When significant effects were observed, we used Tukey's Honest Significant Difference (HSD) tests to examine pair-wise comparisons. In all analyses, we used a significance level of $\alpha = 0.05$.

3. Results

3.1. Influence of upper limb contact site

Fig. 2(a) and (b) show linear and rotational head acceleration profiles from a typical participant for shoulder-, elbow- and hand-to-head impacts, while Fig. 2(c) and (d) show combinations from each participant of mean peak magnitude versus time-to-peak head acceleration.

Upper limb contact site significantly affected the magnitudes of and times to peak linear and rotational head acceleration ($p < 0.005$; Table 2, Figs. 2 and 3). Peak linear head accelerations from hand impacts (mean ± 1 SD: 20.35 ± 6.65 g) exceeded elbow (14.23 ± 4.35 g) and shoulder (10.55 ± 2.60 g) by 1.4-fold and 1.9-fold respectively ($p \leq 0.002$). Differences between elbow and shoulder were borderline-significant ($p = 0.05$). Peak rotational head acceleration was higher for hand (1097.9 ± 328.9 rad/s^2) and elbow (944.1 ± 375.5 rad/s^2) than shoulder impacts (523.1 ± 158.0 rad/s^2), by 2.1-fold and 1.8-fold respectively ($p < 0.001$). Peak rotational head accelerations did not differ between hand and elbow impacts ($p = 0.3$).

Time-to-peak linear head acceleration (after contact) was greater for the shoulder (12.26 ± 4.28 ms) than both the hand (4.98 ± 1.16 ms) and elbow (5.94 ± 4.01 ms), by 2.5-fold and 2.1-fold respectively ($p < 0.0001$). Time-to-peak rotational head acceleration was 1.3-fold greater for the shoulder (11.71 ± 4.84 ms) than for the hand (9.11 ± 4.55 ms; $p = 0.01$), and there was a borderline-significant difference of 1.2-fold between shoulder and elbow (9.83 ± 4.56 ms; $p = 0.07$). For impacts delivered by the hand and elbow, the time-to-peak linear acceleration was shorter than the time-to-peak rotational acceleration (4.98 vs 9.11 ms for the hand; 5.94 vs 9.83 ms for the elbow; $p < 0.0001$ by paired t-tests). For the shoulder, the time-to-peak acceleration was similar for linear and rotational head acceleration (11.71 ms for linear and 12.26 ms for rotational; $p = 0.4$).

Upper limb contact velocity differed significantly between each limb contact site ($p < 0.0004$), and was 2.0-fold greater for the elbow (5.15 ± 0.66 m/s) than the shoulder (2.53 ± 0.47 m/s), and 1.7-fold greater for the hand (4.39 ± 0.47 m/s) than the shoulder.

3.2. Influence of head impact site

Site of head impact (frontal versus lateral) had little effect on the associations between head kinematics and upper limb contact site. There were no significant interactions between upper limb and head contact site for peak rotational head acceleration, time-to-peak linear and rotational acceleration, and contact velocity ($p \geq 0.08$). Upper limb contact site and head impact site interacted significantly for peak linear head acceleration ($p = 0.03$), reflecting similar trends but greater differences between upper limb contact sites in lateral than frontal impacts (Fig. 3(a)).

There was no significant main effect of head impact site for peak linear acceleration, time-to-peak linear and rotational acceleration, or contact velocity ($p \geq 0.09$). Peak rotational head accelerations were significantly greater ($p = 0.02$; Table 2; Fig. 3(b)) in lateral impacts (936.4 ± 408.7 rad/s^2) than frontal impacts

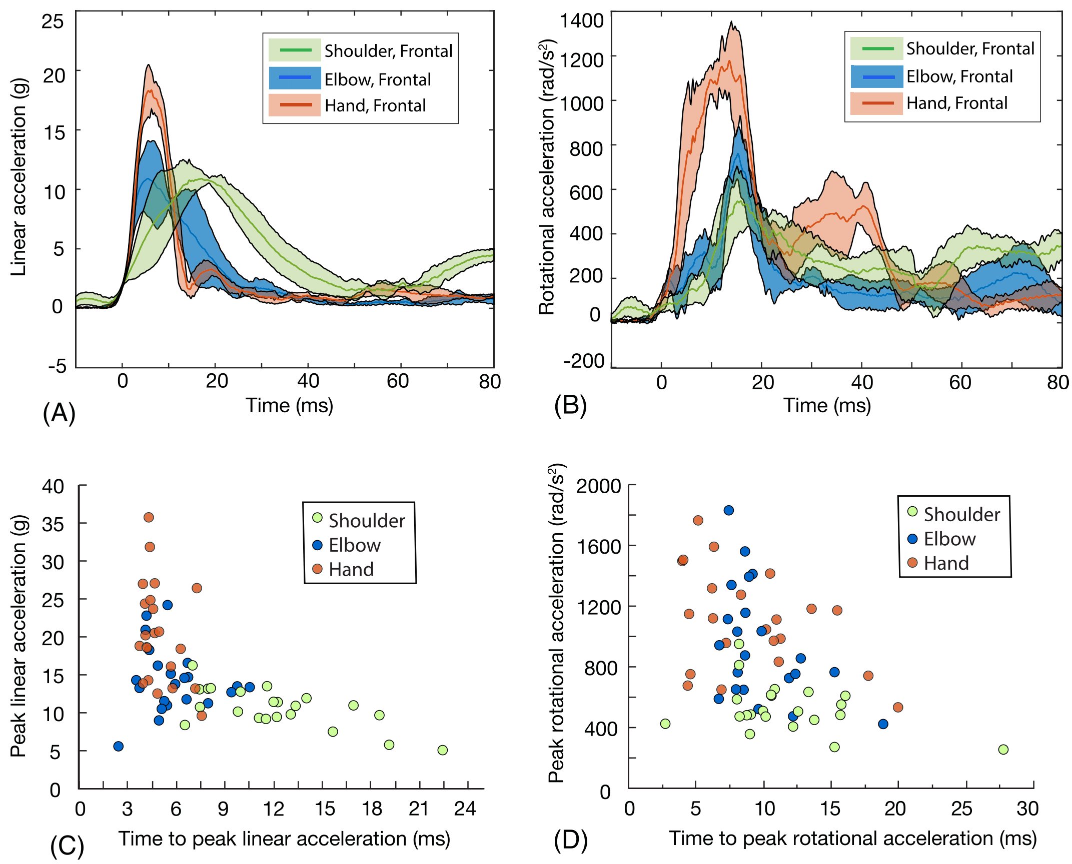

Fig. 2. (A) Temporal variations in linear head acceleration and (B) head rotational acceleration for a typical participant. Data show mean values over the four repeated trials, with clouds that represent ± one standard deviation. Onset of impact is at t = 0 ms. See legend to match colours to upper limb contact site and head contact site. (C) Combinations of magnitude versus time-to-peak linear acceleration and (D) peak rotational acceleration.

Table 2
Summary of the magnitudes and times to peak linear and peak rotational head acceleration, and upper limb contact velocity.

Variable	Checks to FRONTAL Aspect of Head			Checks to LATERAL Aspect of Head		
	Hand	Elbow	Shoulder	Hand	Elbow	Shoulder
Peak linear head acceleration						
Mean (SD) in g's *	19.13 (6.87)	15.20 (4.78)	11.18 (2.39)	21.57 (6.52)	13.26 (3.85)	9.91 (2.77)
Range in g's	13.12–35.66	8.91–20.84	5.00–13.42	9.52–31.76	5.50–20.84	5.70–16.16
Time-to-peak (SD) in ms*	4.92 (1.13)	5.88 (2.04)	12.31 (4.61)	5.04 (1.23)	6.00 (2.19)	12.20 (4.14)
Peak rotational head acceleration						
Mean (SD) in rad/s^2 * †	1026.5 (334.0)	783.0 (338.4)	511.4 (123.0)	1169.2 (323.2)	1105.2 (352.9)	534.8 (192.4)
Range in rad/s^2	645.9–1586.7	419.2–1407.2	251.3–649.0	529.2–1760.4	647.5–1826.6	267.6–946.7
Time-to-peak (SD) in ms*	8.86 (4.43)	10.58 (3.75)	12.88 (6.17)	9.35 (4.88)	9.07 (1.89)	10.53 (2.85)
Limb contact velocity						
Mean (SD) in m/s *	4.29 (0.31)	5.19 (0.46)	2.49 (0.66)	4.49 (0.29)	5.11 (0.48)	2.58 (0.99)
Range in m/s	3.60–5.38	3.73–6.04	1.63–3.37	3.58–5.16	4.04–5.96	1.94–3.09

* Indicates significant differences between upper limb contact sites (Hand/Elbow/Shoulder), $p < 0.05$.
† Indicates significant differences between head impact sites (Frontal/Lateral), $p < 0.05$.

(773.6 ± 347.9 rad/s^2). Post-hoc comparisons revealed significant differences in peak rotational head acceleration between lateral and frontal impacts for the elbow (p = 0.03) but not for the shoulder or hand ($p \geq 0.2$). There were no differences in any of our head impact severity measures between lateral and frontal impacts in the case of the shoulder and hand ($p > 0.2$).

3.3. Similarity to real-life and comfort in delivering checks

Participants rated the similarity of their checks to those in ice hockey with a mean score of 6.9/10.0, and their comfort in checking the dummy with mean scores of 9.0/10.0, 7.5/10.0 and 7.0/10.0 for shoulder, elbow and hand checks respectively (Table 3).

4. Discussion

In this study, we used a novel method to measure and compare head kinematics when players delivered checks "as hard as comfortable" to the head of an instrumented dummy with their padded shoulder, elbow and hand. We expected, based on the documented circumstances of concussions in ice hockey (Delaney et al., 2014; Hutchison et al., 2015) that participants would deliver more severe impacts, involving larger magnitudes and/or durations of head acceleration, with the shoulder than with the elbow or hand.

We found that shoulder-to-head impacts produced head linear and rotational accelerations that were longer in duration (based on time-to-peak), but lower in peak magnitude than head

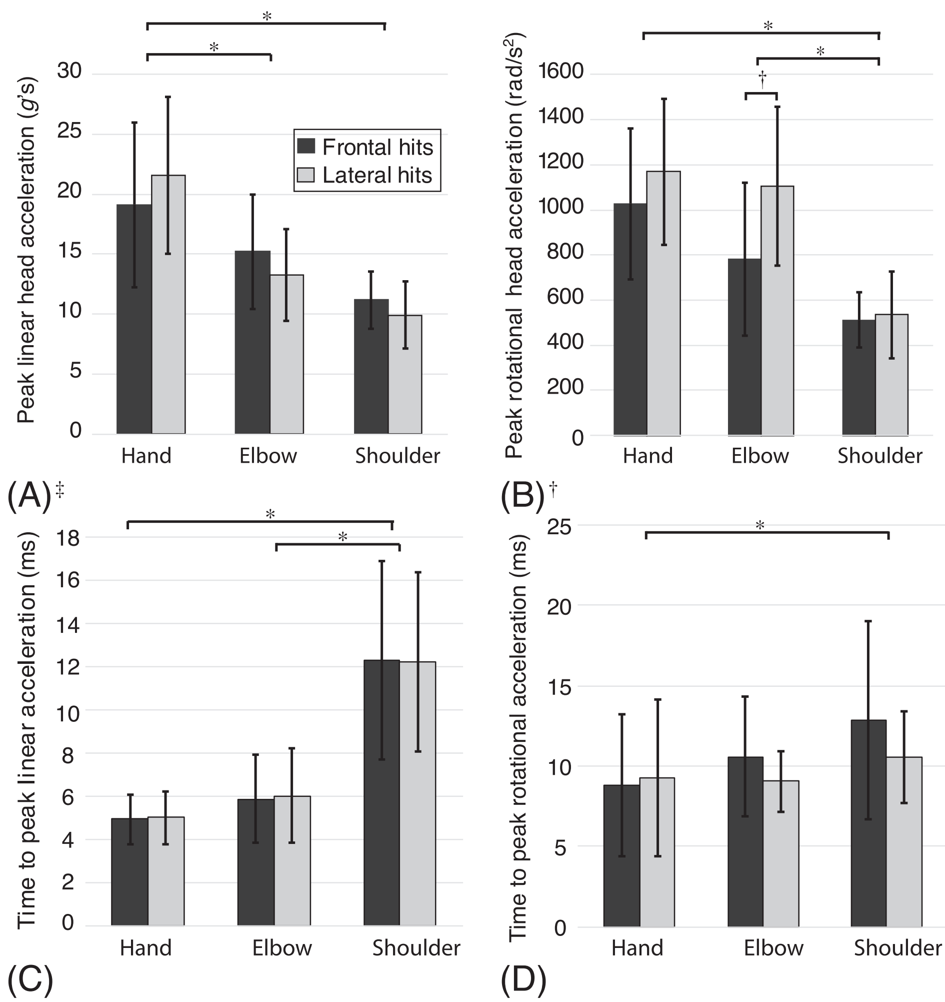

Fig. 3. Mean values and results from ANOVA analysis on (A) peak linear head acceleration (*g*); (B) peak angular head acceleration (rad/s^2); (C) time-to-peak linear head acceleration (ms); and (D) time-to-peak rotational head acceleration (ms). Error bars show ± one standard deviation. * Indicates significant differences ($p < 0.05$) across upper limb contact sites from Tukey's HSD comparisons. † Indicates significant differences ($p < 0.05$) between pairs across head contact sites from Tukey's HSD comparisons. § For peak linear acceleration, there was a significant interaction between upper limb contact site and head contact site ($p = 0.03$). ¶ For peak rotational head acceleration, there was a significant effect of head contact site ($p = 0.02$).

Table 3
Participant ratings on a 10-point Likert scale for similarity to on-ice checks and comfort in delivering checks.

Participant number	Similarity to On-Ice Check[a]	Comfort in Delivering Check[b]		
		Hand	Elbow	Shoulder
1	7	2	7	10
2	8	7	9	10
3	6.5	8	8	9
4	7	7	6	9
5	7	7	7	9
6	6	7	5	5
7	7	6	8	10
8	8	10	10	10
9	4	6	6	8
10	8	7	7	9
11	...	10	9	10
Mean (SD)	6.9 (1.2)	7.0 (2.1)	7.5 (1.5)	9.0 (1.5)
Median	7.0	7.0	7.0	9.0

"..." denotes missing data.
[a] For similarity, a rating of 1 corresponded to "extremely different", while a rating of 10 corresponded to "extremely similar".
[b] For comfort, a rating of 1 corresponded to "extreme discomfort", while a rating of 10 corresponded to "extreme comfort".

accelerations produced by hand- and elbow-to-head impacts. These trends existed for both linear and rotational head acceleration, and for impacts delivered to both the frontal and lateral aspects of the head. The differences were explained in part by lower contact velocities for shoulder impacts. However, they may also relate to differences in biomechanical parameters beyond the scope of our analysis, such as participants' effective (end-point) mass, stiffness and damping when delivering impacts with the shoulder, elbow and hand (Rousseau and Hoshizaki, 2015).

We observed greater differences across upper limb contact sites in peak linear head acceleration during lateral than frontal impacts. However, the trends were similar in terms of accelerations being highest for hand impacts, and lowest for shoulder impacts. We also found that peak rotational head accelerations were larger in lateral than frontal impacts, although post-hoc comparisons revealed significant differences only for the elbow. For both the shoulder and hand, there were no differences between frontal and lateral impacts in any of our measures of head kinematics. This suggests that the effect of head impact site on peak rotational acceleration for elbow checks was probably due to differences in the manner in which elbow checks were delivered in lateral and frontal impacts, as opposed to systematic differences in dummy characteristics.

The opposite trends we observed in the magnitude and duration of head acceleration across upper limb contact sites make it difficult for us to rank head impact severity, or identify a clear "worst case", among the scenarios we examined. The effects of both magnitude and duration of head acceleration on brain injury risk are well-established, with longer durations of head acceleration increasing brain injury risk for a given magnitude of head acceleration (Fernandes and de Sousa, 2015; Gennarelli, 1983; Gennarelli and Thibault, 1982; Hoshizaki et al., 2017; Post et al., 2017).

Our approach (of having players deliver checks to the head of an instrumented dummy) provides new knowledge by overcoming some of the challenges of accurately measuring head accelerations from helmet-mounted sensors during game play (Mihalik et al., 2012; Wilcox et al., 2014a; Wilcox et al., 2014b), and of accurately re-creating the dynamics of head impacts in hockey with mechanical test systems (Clark et al., 2016; Post et al., 2019; Rowson et al., 2015). At the same time, it is instructive to compare our results to studies that have used these approaches.

With regard to accelerations measured from helmet-mounted sensors during game play, Wilcox et al. (2014a) used the HIT system to record accelerations from 19,880 head impacts in men's university hockey. Median values of peak acceleration were 15.7 g and 1630 rad/s^2. Mihalik et al. (2012) used the HIT system to record accelerations from 12,253 head impacts in male youth (13–16 year old) hockey, which averaged 18.4 g and 1464 rad/s^2. In comparison, our mean accelerations were 15.0 g and 855 rad/s^2. The differences may stem from the fact that we examined only upper limb-to-head impact scenarios, while these previous on-ice studies included all types of head impacts (including those to the boards, glass and ice). The differences may also relate to the documented inaccuracies and trend for helmet-mounted sensors to overestimate head accelerations when compared to Hybrid III accelerometer arrays (Allison et al., 2014; Allison et al., 2015).

With regard to accelerations measured from purely mechanical testing systems, Clark et al. (2016) simulated padded elbow- and shoulder-to-head impacts at 3, 5 and 7 m/s. They reported acceleration durations that averaged 1.7-fold longer for shoulder than elbow impacts, and peak linear and rotational head accelerations (estimated from their bar graphs) that were approximately two-fold higher for elbow- than shoulder-to-head impacts. We observed similar trends, which supports the biofidelity of their test systems. Rousseau (2014) recreated 14 concussive cases of shoulder- and elbow-to-head impacts in professional hockey. They reported mean values of peak head acceleration of 27 g and 3.0 krad/s^2, which are considerably higher than our values (which averaged 12.4 g and 733 rad/s^2 when combining shoulder and elbow impacts). The differences may not be surprising, given that only a small portion of head impacts in hockey result in concussion (Hutchison et al., 2015; Wilcox et al., 2014b), and, as discussed in greater detail below, our experiment was designed to measure head accelerations during re-enactments of common checks that university-level hockey players delivered "as hard as comfortable."

Our study had strengths and limitations. Our participants re-enacted shoulder-, elbow- and hand-to-head checks that we observed to be common in video footage of men's university ice hockey games. However, for safety, we limited the run-up distance to 1.4 m, and instructed participants to check the head of a dummy "as hard as comfortable" while wearing pads. Accordingly, we cannot assume that our experiments measured the hardest hits that participants were capable of delivering. Furthermore, real-life head impacts in hockey involve highly varied contact speeds, body configurations, states of muscle activation, and types of protective gear. Our results are specific to the impact conditions tested in our experiments, and a more comprehensive comparison is warranted for a variety of checking styles and scenarios.

Participants rated the "real life" similarity of their checks to the dummy with a median score of 7.0/10. While encouraging, this is not a guarantee that the motions of the dummy were representative of those occurring during head impacts in hockey. We did not examine how head accelerations depend on dummy characteristics (e.g., neck bending stiffness), or on the nature of protective equipment (helmet, padding) worn by participants or by the dummy. The dummy was equipped with a standard hockey helmet and cage, but not with shoulder pads. It is conceivable that, especially for lateral impacts, the presence and geometry of shoulder pads might affect the initial point of contact, or the manner in which checks are delivered. The experiment was conducted on a vinyl floor with participants wearing running shoes, and there may be differences in the dynamics of upper limb-to-head checks while wearing skates on ice. Additional work is warranted to examine the effect of these variables on head kinematics during upper limb impacts, and to identify and simulate a wider range of impact scenarios that are common in hockey.

In summary, we measured the magnitudes and times to peak head acceleration during checks that university-level hockey players delivered with the padded shoulder, elbow and hand to the head of an instrumented dummy. Shoulder-to-head impacts resulted in the longest times-to-peak acceleration, while hand impacts resulted in the highest magnitudes of acceleration. Increases in peak acceleration magnitude were generally associated with reduced time-to-peak acceleration. The trends we observed prevent us from identifying a worst case, among the padded shoulder, elbow and hand impacts we examined, in terms of the magnitude and duration of head acceleration produced by these impacts. Our results contribute to our evolving understanding of brain health in ice hockey, by providing new evidence from experiments with varsity players on the magnitudes and durations of head accelerations generated during common upper-limb-to-head impacts in hockey.

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Conflict of interest statement

None of the authors have any conflicts to declare related to this research.

Appendix A

Evaluation of the accuracy of the head COG acceleration measures

The accuracy of the head COG acceleration measures was evaluated through a series of drop tests (n = 30) of the isolated head and neck onto a force plate (model 4060H, Bertec) from various heights, which produced peak linear accelerations between 5.0 g and 82.1 g. The measured mass of the head neck system (4.71 kg) was estimated within 0.6% error (at 4.68 kg), based on the best-fit line ($F_{max} = 4.68 * a_{max}$; $R^2 = 0.99$) between peak resultant force from the force plate (F_{max}) and peak resultant acceleration from the Hybrid III sensors (a_{max}).

References

Agel, J., Harvey, E.J., 2010. A 7-year review of men's and women's ice hockey injuries in the NCAA. Can. J. Surg. 53, 319–323.

Allison, M.A., Kang, Y.S., Bolte, J.H.T., Maltese, M.R., Arbogast, K.B., 2014. Validation of a helmet-based system to measure head impact biomechanics in ice hockey. Med. Sci. Sports Exer. 46, 115–123.

Allison, M.A., Kang, Y.S., Maltese, M.R., Bolte IV, J.H., Arbogast, K.B., 2015. Measurement of hybrid III head impact kinematics using an accelerometer and gyroscope system in ice hockey helmets. Ann. Biomed. Eng. 43, 1896–1906.

Bruschetta, M., Virani, S., Hua, K.N., Robinovitch, S.N., 2016. Development and validation of a tool for analysis of hockey-related head impacts captured on video. In: Accepted Abstracts from the International Brain Injury Association's Eleventh World Congress on Brain Injury. Brain Injury, pp. 481–817.

Clark, J.M., Post, A., Hoshizaki, T.B., Gilchrist, M.D., 2016. Protective capacity of ice hockey helmets against different impact events. Ann. Biomed. Eng. 44, 3693–3704.

Cusimano, M.D., Cho, N., Amin, K., Shirazi, M., McFaull, S.R., Do, M.T., Wong, M.C., Russell, K., 2013. Mechanisms of team-sport-related brain injuries in children 5 to 19 years old: opportunities for prevention. PLoS One 8, e58868.

Delaney, J.S., Al-Kashmiri, A., Correa, J.A., 2014. Mechanisms of injury for concussions in university football, ice hockey, and soccer. Clin. J. Sport Med. 24, 233–237.

Fernandes, F.A., de Sousa, R.J., 2015. Head injury predictors in sports trauma–a state-of-the-art review. Proc. Inst. Mech. Eng. [H] 229, 592–608.

Foster, J.K., Kortge, J.O., Wolanin, M.J., 1977. Hybrid III-A biomechanically-based crash test dummy. SAE Tech. Paper, 770938.

Gennarelli, T.A., 1983. Head injury in man and experimental animals: clinical aspects. Acta Neurochir. Suppl. 32, 1–13.

Gennarelli, T.A., Thibault, L.E., 1982. Biomechanics of acute subdural hematoma. J. Trauma 22, 680–686.

Hootman, J.M., Dick, R., Agel, J., 2007. Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives. J. Athletic Train. 42, 311–319.

Hoshizaki, T.B., Post, A., Kendall, M., Cournoyer, J., Rousseau, P., Gilchrist, M.D., Brien, S., Cusimano, M., Marshall, S., 2017. The development of a threshold curve for the understanding of concussion in sport. Trauma 19, 196–206.

Hutchison, M.G., Comper, P., Meeuwisse, W.H., Echemendia, R.J., 2015. A systematic video analysis of national hockey league (NHL) concussions, part II: how concussions occur in the NHL. Br. J. Sports Med. 49, 552–555.

International Ice Hockey Federation (IIHF). Survey of Players. International Ice Hockey Federation, Zurich, Switzerland; 2017.

Mertz, H.J., 1985. Biofidelity of the hybrid III head. SAE Trans. 94, 97–105.

Mihalik, J.P., Guskiewicz, K.M., Marshall, S.W., Blackburn, J.T., Cantu, R.C., Greenwald, R.M., 2012. Head impact biomechanics in youth hockey: comparisons across playing position, event types, and impact locations. Ann. Biomed. Eng. 40, 141–149.

National Highway Traffic Safety Administration. Anthropometric Test Devices, 49 CFR Part 572, Washington DC; 2011.

Padgaonkar, A.J., Krieger, K.W., King, A.I., 1975. Measurement of angular acceleration of a rigid body using linear accelerometers. J. Appl. Mech. 42, 552–556.

Post, A., Hoshizaki, T.B., Gilchrist, M.D., Cusimano, M.D., 2017. Peak linear and rotational acceleration magnitude and duration effects on maximum principal strain in the corpus callosum for sport impacts. J. Biomech. 61, 183–192.

Post, A., Hoshizaki, T.B., Karton, C., Clark, J.M., Dawson, L., Cournoyer, J., Taylor, K., Oeur, R.A., Gilchrist, M.D., Cusimano, M.D., 2019. The biomechanics of concussion for ice hockey head impact events. Comput. Methods Biomech. Biomed. Eng. https://doi.org/10.1080/10255842.2019.1577827, pp. 1-13.

Rousseau, P., 2014. An Analysis of Concussion Metrics Associated with Elite Ice Hockey Elbow-to-Head and Shoulder- to-Head Collisions PhD thesis. University of Ottawa, Canada.

Rousseau, P., Hoshizaki, T.B., 2015. Defining the effective impact mass of elbow and shoulder strikes in ice hockey. Sports Biomech. 14, 57–67.

Rowson, B., Rowson, S., Duma, S.M., 2015. Hockey star: a methodology for assessing the biomechanical performance of hockey helmets. Ann. Biomed. Eng. 43, 2429–2443.

Society of Automotive Engineers International. SAE J211-1: Instrumentation for Impact Test - Part 1 – Electronic Instrumentation. Society of Automotive Engineers International Inc., Warrendale, PA; 1995.

Stemper, B.D., Shah, A.S., Pintar, F.A., McCrea, M., Kurpad, S.N., Glavaski-Joksimovic, A., Olsen, C., Budde, M.D., 2015. Head rotational acceleration characteristics influence behavioral and diffusion tensor imaging outcomes following concussion. Ann. Biomed. Eng. 43, 1071–1088.

Virani, S., Russel, C.N., Bruschetta, M.L., Hua, K.N., Potvin, B.M., Cox, D.N., Robinovitch, S.N., 2017. The effect of shoulder pad design on head impact severity during checking. Med. Sci. Sports Exerc.

Wilcox, B.J., Machan, J.T., Beckwith, J.G., Greenwald, R.M., Burmeister, E., Crisco, J.J., 2014b. Head-impact mechanisms in men's and women's collegiate ice hockey. J. Athletic Train. 49, 514–520.

Wilcox, B.J., Beckwith, J.G., Greenwald, R.M., Chu, J.J., McAllister, T.W., Flashman, L. A., Maerlender, A.C., Duhaime, A.C., Crisco, J.J., 2014a. Head impact exposure in male and female collegiate ice hockey players. J. Biomech. 47, 109–114.

Zuckerman, S.L., Kerr, Z.Y., Yengo-Kahn, A., Wasserman, E., Covassin, T., Solomon, G. S., 2015. Epidemiology of sports-related concussion in NCAA athletes from 2009–2010 to 2013–2014: incidence, recurrence, and mechanisms. Am. J. Sports Med. 43, 2654–2662.