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Head Kinematics and Injury Analysis in Elite Bobsleigh Athletes Throughout a World Cup Tour

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Context: The neurocognitive health effects of repetitive head impacts have been examined in many sports. However, characterizations of head impacts for sliding-sport athletes are lacking.

Objective: To describe head impact kinematics and injury epidemiology in elite athletes during the 2021–2022 Bobsleigh World Cup season.

Design: Cross-sectional study.

Setting: On-track training and competitions during the Bobsleigh World Cup season.

Patients or Other Participants: Twelve elite bobsleigh athletes (3 pilots [1 female], 9 push athletes [5 females]; age = 30 ± 5 years; female height and weight = 173 ± 8 cm and 75 ± 5 kg, respectively; male height and weight = 183 ± 5 cm and 101 ± 5 kg, respectively).

Main Outcome Measure(s): Athletes wore an accelerometer-enabled mouthguard to quantify 6-degrees-of-freedom head impact kinematics. Isometric absolute and relative neck strength, number of head acceleration events (HAEs), workload (J), peak linear velocity ($m{\cdot}s^{-1}$), peak angular velocity ($rad{\cdot}s^{-1}$), peak linear acceleration (*g*), and peak angular acceleration ($rad{\cdot}s^{-2}$) were derived from mouthguard manufacturer algorithms. Linear mixed-effect models tested the effects of sex (male versus female), setting (training versus competition), and position (pilot versus push athlete) on the kinematic variables.

Results: A total of 1900 HAEs were recorded over 48 training and 53 competition days. No differences were found between the number of HAEs per run per athlete by sex (incidence rate ratio [IRR] = 0.82, $P = .741$), setting (IRR = 0.94, $P = .325$), or position (IRR = 1.64, $P = .463$). No sex differences were observed for workload (mean ± SD: males = 3.3 ± 2.2 J, females = 3.1 ± 1.9 J; $P = .646$), peak linear velocity (males = 1.1 ± 0.3 $m{\cdot}s^{-1}$, females = 1.1 ± 0.3 $m{\cdot}s^{-1}$; $P = .706$), peak angular velocity (males = 4.2 ± 2.1 $rad{\cdot}s^{-1}$, females = 4.7 ± 2.5 $rad{\cdot}s^{-1}$; $P = .220$), peak linear acceleration (male = 12.4 ± 3.9*g*, females = 11.9 ± 3.5*g*; $P = .772$), or peak angular acceleration (males = 610 ± 353 $rad{\cdot}s^{-2}$, females = 680 ± 423 $rad{\cdot}s^{-2}$; $P = .547$). Also, no effects of setting or position on any kinematic variables were seen. Male athletes had greater peak neck strength than female athletes for all neck movements, aside from right-side flexion ($P = .085$), but no sex differences were noted in relative neck strength.

Conclusions: We provide a foundational understanding of the repetitive HAEs that occur in bobsleigh athletes. Future authors should determine the effects of repetitive head impacts on neurocognitive function and mental health.

Key Words: concussion, sliding sport, instrumented mouthguard

Key Points

- The frequency and magnitude of head acceleration events during bobsleigh participation were similar between sexes (male versus female), positions (pilot versus push), and setting (training versus competition).
- Despite the high prevalence of head acceleration events (peak linear acceleration > 8*g*), only 1 sport-related concussion was observed throughout the World Cup Tour.
- Time-loss injury incidence rates during a World Cup Tour were 24.55 injuries per 1000 athlete-exposures, with males (sex) and pilots (position) sustaining more injuries than females and push athletes.

Despite the breadth of knowledge and awareness, prevention, and care of sport-related concussion (SRC) and the emerging evidence that suggests associations of repetitive, subconcussive head impacts with health and cognitive outcomes in contact sports such as American football,[1,2] boxing,[3–5] soccer,[6–8] and ice hockey,[9,10] our understanding of these occurrences in sliding sports (ie, bobsleigh, skeleton, and luge) is limited. Available evidence suggests that SRC accounts for 12% to 15% of all sliding-sports injuries.[11–14] Understanding concussive and subconcussive head impact kinematics in sliding sports may have significant long-term implications for health outcomes among these athletes. In previous literature, the potential for short- and long-term adverse physical and mental health effects from subconcussive and repetitive head impacts has been demonstrated.[15–26] Specifically, repetitive head impacts have been associated with cognitive decline, adverse blood biomarker changes, and brain structure and function alterations.[9,15,16,22,23,26] However, the occurrence and magnitude of head impacts during training and competition in sliding sports are unknown.

Stabilizing the neck via increased neck strength will alter head kinematics in response to a perturbation and has been proposed as a modifiable risk factor for concussions.[27] Previous authors determined that neck flexion and extension strength in soccer athletes was negatively correlated with head acceleration[28] and that neck strength explained 13.3% and 17.2% of peak linear acceleration (PLA) and peak rotational acceleration, respectively.[29] However, the association between neck strength and resulting kinematics is not a ubiquitous finding, as shown in a cohort of youth ice hockey players.[30] Moreover, earlier researchers identified differences in neck strength between sexes, with males showing greater absolute neck strength than females.[31,32] Reference data and whether neck strength is a critical factor for reducing head impacts and kinematics in bobsleigh athletes are not currently known.

Thus, in this pilot study, we aimed to describe head impact kinematics and head injury epidemiology throughout a Bobsleigh World Cup tour. Specifically, we sought to examine the frequency and magnitude of head acceleration events (HAEs) that occurred while training and competing in bobsleigh. We hypothesized that HAE kinematics would not differ between training and competition days, between male and female athletes, or between bobsleigh positions (pilots versus push athletes). We further hypothesized that absolute but not relative neck strength would be greater in male versus female athletes.

METHODS

The study was approved by the Institutional Review Board (IRB-FY22-116) at the University of North Carolina at Greensboro. Of the 18 athletes on the Team USA Bobsleigh World Cup roster eligible for participation in the study, 12 (67%) elite Team USA bobsleigh athletes (50% female; pilots, $n = 3$ [female pilots, $n = 1$]; push athletes, $n = 9$; age $= 30 \pm 5$ years; female height $= 173 \pm 8$ cm, female mass $= 75 \pm 5$ kg; male height $= 183 \pm 5$ cm, male mass $= 101 \pm 5$ kg) competing in the 2021–2022 International Bobsleigh and Skeleton Federation World Cup Tour were enrolled. *Pilots* were defined as the athletes responsible for steering the bobsleigh down the track, and *push athletes* were defined as the athletes responsible for assisting the pilot in propelling the bobsleigh down the track in 2-man or 2-woman, 4-man, and monobob bobsleigh (women only). Upon a review of the study's methods and procedures, athletes provided their written and informed consent to participate.

The 2021–2022 Bobsleigh World Cup season was an 8-race series held at 5 tracks in Europe. The 2021–2022 Bobsleigh World Cup season started on November 20–21, 2021, at the Olympiaworld-Eiskanal track in Innsbruck, Austria, and concluded on January 15–16, 2022, in St Moritz, Switzerland, at the Celerina Olympia Bobrun track. Details about the specific tracks—including length, vertical drop, and number of turns—used by the International Bobsleigh Federation and those that were used for the 2021–2022 World Cup season can be found in the Supplemental Table, available online at http://dx.doi.org/10.4085/1062-6050-0014.23.S1.

Before the Bobsleigh World Cup season started, isometric measures of absolute neck strength were obtained using a handheld dynamometer (MicroFET 2; Hoggan Scientific, LLC) and previously established methods.[33] We calculated relative neck strength by dividing peak strength values by the athlete's body mass. Throughout the 10-week World Cup season, athletes were instructed to wear an accelerometer-enabled boil-and-bite fitted mouthguard (Prevent Impact Monitor; Prevent Biometrics) to quantify HAE kinematics during all training sessions and competitions. As noted by Kuo et al,[34] mandible constraints can affect the accuracy of instrumented mouthguards. The validity of the mouthguard used in our study has been demonstrated in laboratory tests using a clamped-jaw model[35–37] that simulates a clenched jaw and in laboratory assessments with an articulating jaw without clamping.[37–39] Further, the mouthguard has been validated using video footage of collegiate rugby players, resulting in a positive predictive value of 94% to 96.4%.[35,36] The mouthguard has also been used in the field in conjunction with video confirmation of impacts during competitive rugby league matches[40] and soccer heading impacts[41] and in boxing and mixed-martial arts.[42] The mouthguard had an embedded triaxial accelerometer and gyroscope that both sampled at 3200 Hz. The accelerometer and gyrometer has a $\pm 200g$ and ± 35 rad/s full-scale sensor magnitude range, respectively. Additionally, the mouthguard uses an infrared sensor to determine the tightness of fit to the dentition, and HAEs considered off teeth were discarded. The manufacturer filtered the raw acceleration signals using a low-pass, second-order Butterworth filter and a zero-phase forward- and reverse-filtering process, used previously,[42] with cutoff frequencies established according to the level of noise in the HAE. The level of noise in the signal was determined by a manufacturer machine learning model, which classified each HAE into 1 of 3 classes: class 0 (minimal noise) HAEs were filtered at 200 Hz, class 1 (moderate noise) HAEs were filtered at 100 Hz, and class 2 (severe noise) HAEs were filtered at 50 Hz.

The number of HAEs, PLA (g), peak linear velocity (PLV, $m \cdot s^{-1}$), workload (J), peak angular acceleration (PAA, $rad \cdot s^{-2}$), and peak angular velocity (PAV, $rad \cdot s^{-1}$) were calculated from raw accelerometer waveforms per the manufacturer's algorithms. *Workload* was defined as the estimated kinetic energy transfer to the head for a registered event ($\text{workload} = 0.5I\omega^2 + 0.5mv^2$), and an *HAE* was further defined as an instant in time when PLA exceeded $8g$ (50-ms recording window; pretrigger duration $=$ 10 milliseconds, posttrigger duration $=$ 40 milliseconds, $\sim$160 samples). See Supplemental Figure, available online at http://dx.doi.org/10.4085/1062-6050-0014.23.S1, for an example time-series plot of an HAE. A research assistant was responsible for enforcing athlete adherence and ensuring that the mouthguard batteries were charged and the data were syncing. In addition to capturing HAEs, the team's medical provider recorded time-loss injuries. *Time-loss injuries* were defined as medical encounters that resulted in the athlete missing time from participation (either training or competition) due to the medical encounter. The mode of injury (acute or chronic/exacerbation of existing injury), type of injury, and anatomic location of injury were recorded.

Summary data were calculated, including the number of days, the number of runs captured, the number of HAEs, and the mean values for each outcome metric (workload, PLV, PAV, PLA, and PAA). Furthermore, *athlete-exposures*, defined as the number of runs completed by each athlete, and time-loss injuries that occurred during training and competition were quantified. Injury incidence rates were reported as injuries per 1000 exposures (injury incidence $= \left(\frac{\text{No. of injuries}}{\text{athlete}-\text{exposures}}\right) \times 1000$) and were accompanied by 95% CIs. Generalized mixed-effects linear models with Poisson error distributions tested for differences in the number of recorded HAEs by sex (male versus female), setting (ie, training versus competition), and bobsleigh position (ie, pilot versus push athlete). Linear mixed-effects models were fit[43] to test differences in workload, PLV, PLA, PAV, and PAA between sexes, settings, and bobsleigh positions, covarying for the run number on each day. Models fit all independent variables as fixed effects and included random intercepts, nesting observations within each athlete. Model specifications were assessed for fit via χ^2 analysis. In addition, models were tested for residual normality and homoscedasticity assumptions and refit with alternative error distributions if required. Final models were fit using restricted maximum likelihood estimation. All analyses were completed using R statistical software (R Foundation for Statistical Computing),[44] and the α level for all fixed effects was set at $P < .05$.

RESULTS

A total of 101 separate days (training = 48 days, competition = 53 days) were recorded for the team, with a mean of 2 ± 1 (range = 1–3) runs per day per athlete. Throughout the World Cup season, a total of 1900 HAEs were recorded, with a mean of 11 ± 8 HAEs recorded per run per athlete (Table 1). No differences were found in the number of HAEs per run per athlete between males and females (incidence rate ratio [IRR] = 0.82, $P = .741$), training and competition (IRR = 0.94, $P = .325$), or pilot and push athletes (IRR = 1.64, $P = .463$). The number of HAEs per run and the summarized kinematic variables are presented in Table 1.

Visual inspection and formal tests demonstrated significant positive skewness and heteroscedasticity in general linear models. Therefore, we re-fit models as generalized linear mixed-effects models with γ distributions (log-link). The following coefficients are reported as the exponentiated log-odds coefficient. No effect of sex was evident on workload (Figure 1A, $\beta = 1.06$ [95% CI = 0. 83, 1.36], $P = .646$), PLV (Figure 2A, $\beta = 1.03$ [95% CI = 0.87, 1.23], $P = .706$), PAV (Figure 2D, $\beta = 0.86$ [95% CI = 0.67, 1.10], $P = .220$), PLA (Figure 3A, $\beta = 1.02$ [95% CI = 0.89, 1.16], $P = .772$), or PAA (Figure 3D, $\beta = 0.88$ [95% CI = 0.59, 1.32], $P = .547$).

Also, no effects of setting on workload (Figure 1B, $\beta = 0.98$ [95% CI = 0.93, 1.03], $P = .464$), PLV (Figure 2B, $\beta = 0.99$ [95% CI = 0.96, 1.01], $P = .317$), PAV (Figure 2E, $\beta = 1.05$ [95% CI = 1.00, 1.09], $P = .045$), PLA (Figure 3B, $\beta = 1.00$ [95% CI = 0.98, 1.03], $P = .697$), or PAA (Figure 3E, $\beta = 1.01$ [95% CI = 0.96, 1.06], $P = .706$) were found.

No effects of position were present on workload (Figure 1C, $\beta = 0.85$ [95% CI = 0.65, 1.10], $P = .211$), PLV (Figure 2C, $\beta = 0.94$ [95% CI = 0.74, 1.19], $P = .617$), PAV (Figure 2F, $\beta = 0.93$ [95% CI = 0.67, 1.28], $P = .641$), PLA (Figure 3C, $\beta = 1.10$ [95% CI = 0.96, 1.28], $P = .174$), or PAA (Figure 3F, $\beta = 1.38$ [95% CI = 0.88, 2.15], $P = .160$).

Male athletes had greater peak neck strength than female athletes for all absolute neck strength measures except right-side flexion (Table 2). No differences between males and females were observed for any relative neck strength measure.

Table 1. Group Head Kinematic Values. Mean ± SD (95% CI)[a]

Variable	Sex		Setting		Position	
	Female	Male	Competition	Training	Pilot	Push
Impacts, (No./athlete/run)	11 ± 9 (8, 13)	11 ± 8 (9, 12)	11 ± 8 (8, 13)	10 ± 8 (8, 12)	11 ± 6 (10, 12)	9 ± 11 (7, 12)
Workload, J	3.1 ± 1.9 (2.9, 3.2)	3.3 ± 2.2 (3.2, 3.4)	3.2 ± 2.1 (3.1, 3.4)	3.2 ± 2.1 (3.1, 3.3)	2.9 ± 2.0 (2.8, 3.1)	3.9 ± 2.2 (3.7, 4.0)
Peak linear velocity, $m \cdot s^{-1}$	1.1 ± 0.3 (1.0, 1.1)	1.1 ± 0.3 (1.1, 1.1)	1.1 ± 0.4 (1.1, 1.1)	1.1 ± 0.3 (1.1, 1.1)	1.0 ± 0.3 (1.0, 1.1)	1.2 ± 0.3 (1.2, 1.2)
Peak angular velocity, $rad \cdot s^{-1}$	4.7 ± 2.5 (4.5, 4.9)	4.2 ± 2.1 (4.1, 4.4)	4.3 ± 2.1 (4.1, 4.4)	4.5 ± 2.3 (4.4, 4.7)	4.4 ± 2.0 (4.3, 4.5)	4.4 ± 2.6 (4.1, 4.6)
Peak linear acceleration, *g*	11.9 ± 3.5 (11.6, 12.1)	12.4 ± 3.9 (12.2, 12.6)	12.2 ± 3.8 (11.9, 12.4)	12.3 ± 3.7 (12.0, 12.5)	12.5 ± 3.9 (12.3, 12.7)	11.5 ± 3.4 (11.2, 11.8)
Peak angular acceleration, $rad \cdot s^{-2}$	680 ± 423 (649, 711)	610 ± 353 (589, 630)	626 ± 355 (602, 650)	645 ± 403 (620, 669)	720 ± 363 (701, 740)	433.9 ± 349 (404, 462)

[a] Aside from the No. of impacts (reported as the No. of impacts per athlete per run), all other data are presented as within-group summary statistics.

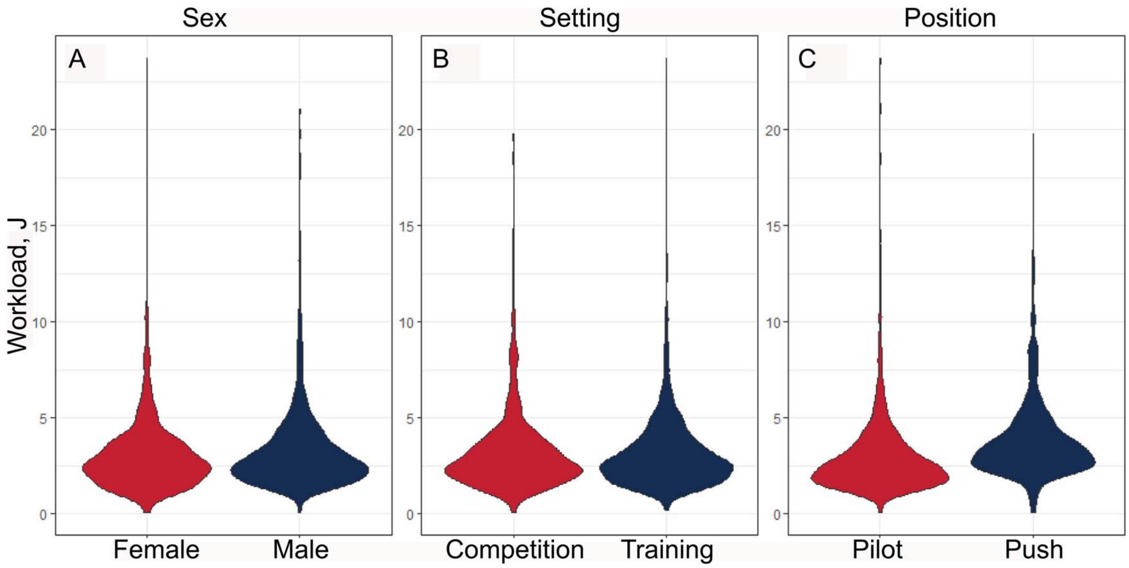

Figure 1. Workload (J) by, A, sex, B, setting (training versus competition), and C, position during the 2021–2022 World Cup season.

One crash event occurred, and the associated HAE was captured in this dataset, although in an attempt to retain de-identified data, that particular case is not available for publication. Time-loss injuries are reported in Table 3.

Despite many individual observations, the low sample size was an a priori concern. Because no differences were identified between any independent variables tested, we completed post hoc power analyses via Monte Carlo simulation.[45] We recognize and acknowledge the biased nature and other concerns with post hoc power analyses (eg, Hoenig and Heisey[46]). However, these data are provided for additional context, as opposed to justifying or substantiating any specific claims regarding our findings. Fitted models were tested for statistical power to detect a difference in the independent variable fixed effect at an α level of $P < .05$, using 1000 resamples. The mean power for all models was 28.7% (minimum = 3.9% [PAV $\sim$ session], maximum = 48.5% [PLV $\sim$ position]).

DISCUSSION

In agreement with our hypotheses, no differences in HAE kinematic measures were seen between sexes, settings, or positions among Team USA athletes throughout the Bobsleigh World Cup season. However, we observed an average of 11 HAEs per training and competition run. In this pilot work, we provide the first known descriptive data and a preliminary understanding of HAEs in bobsleigh athletes. Moreover, the model results revealed no differences in HAE kinematics between sexes, training and competition, or athlete positions within the bobsleigh.

The mean magnitude of HAE kinematic data was lower in the current study than pooled in American football (PLA mean = 25.4g; PAA = 1733 rad·s^{-2}).[47] In sports such as American football and ice hockey, previous literature characterized the mean number of head impacts as 10.6 to 24.1 per session.[47,48] Although our data showed that the number of HAEs was 50% of the minimum in other sports, we must acknowledge that we used a threshold of PLA $\geq$8g, whereas other authors considered a head impact using thresholds of 10g to 15g; these differences may overestimate the number of HAEs.

Further, a critical difference was the sampling period between the current dataset and other sports. Specifically, the sampling period in bobsleigh athletes was approximately 2 minutes (timing of each run) compared with 60 to 90 minutes for other sports. This timing discrepancy may be clinically significant and warrants continued investigation to better understand the implications of the number of HAEs during a bobsleigh run and the frequency and time course in which they occur.

The most comparable data are from pilot data in dirt-track car racing athletes.[49] Interestingly, our data (median PLA = 11.2g, PAA = 573 rad·s^{-2}, and PAV = 4 rad·s^{-1}) suggested higher HAE kinematics than those recorded during racing laps (median PLA = 5.33g, PAA = 179 rad·s^{-2}, and PAV = 2.89 rad·s^{-1}), and in fact, our data are more consistent with kinematics recorded during the car racing crash events (median PLA = 13.4g, PAA = 630 rad·s^{-2}, and PAV = 9.67 rad·s^{-1}).[49] It is possible that the additional equipment restraints (eg, seatbelts, neck restraints, car frame, seat design), none of which are available to bobsleigh athletes, help to limit head forces in the racing car drivers. In its simplest form, a bobsleigh consists of an aerodynamic shell, front and back metal runners, and a front bumper; however, the ability to modify the bobsleigh design is limited.[50] This highlights the need for research focused on sliding-sport ergonomic and protective equipment.

Sex differences in the number of head impacts have been observed in various sports, including combat sports,[42] soccer,[51] ice hockey,[52] and Australian rules football.[31] These sports are generally considered contact sports, and the mechanism for head impacts is different than during a bobsleigh run, in which the HAEs are primarily dictated by the sporting environment (ie, the track) rather than opponents. Some sports (eg, lacrosse) limit high-intensity body-to-body contact in the women's game, thereby providing a clear explanation for the greater incidence of head injury in male athletes. However, this explanation does not hold for sports with sex-agnostic rules, such as rugby and soccer or, indeed, bobsleigh. An explanation for the higher incidence of head impacts in team contact sports remains speculative, but it may result from more aggressive and dangerous play by male participants. Yet in bobsleigh, this

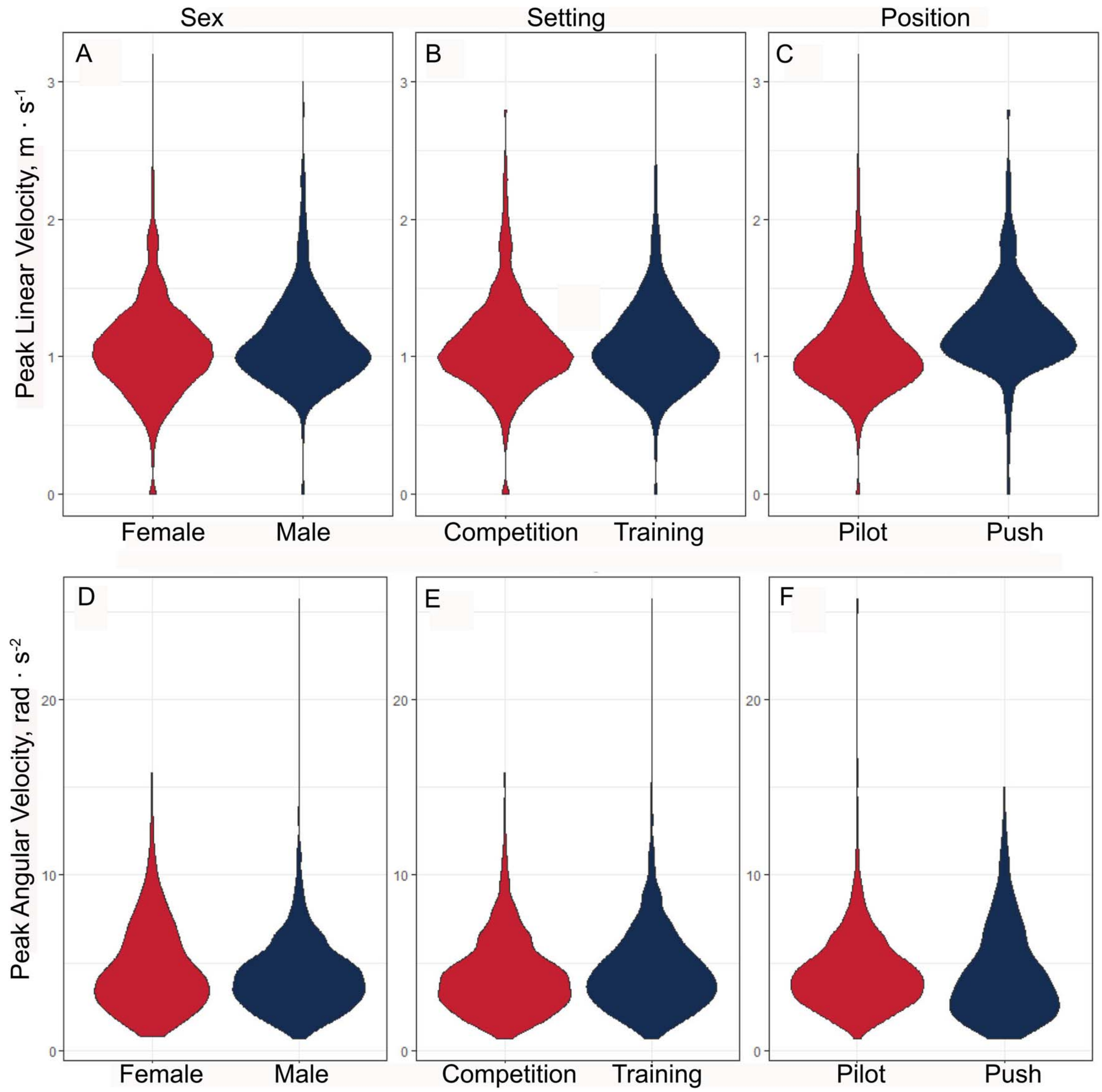

Figure 2. Peak linear velocity ($m \cdot s^{-1}$) by, A, sex, B, setting (training versus competition), and C, position; and peak angular velocity ($rad \cdot s^{-1}$) by, D, sex, E, setting, and F, position during the 2021–2022 World Cup season.

effect is limited; thus, in the absence of direct contact with other athletes, rule differences, or a limited ability to compete less safely, the number of HAEs in male and female bobsleigh athletes appears to be similar.

The intricate interrelation between neck strength and head kinematics has been the focus of extensive research in the field of traumatic brain injury. Although some researchers have suggested that a strong neck musculature may confer enhanced head stability[29,53] and decreased likelihood of concussive events,[27] others have proffered evidence that cervical muscle force does not exert a discernible influence on head kinematics.[30,54–56] Consequently, the precise role of neck strength and stabilization in reducing head trauma remains a subject of considerable debate, and the effectiveness of neck-strengthening programs in reducing the concussion risk is not yet definitively established. The authors of 1 literature review proposed that targeted neck-strengthening protocols may be effective in reducing the incidence of concussions,[57] whereas the authors of another review have posited more limited evidence for such interventions.[58] Further exploration is thus warranted to more fully elucidate the complex mechanisms that underpin the interplay between neck strength, head motion, and concussion risk to devise more effective concussion-prevention strategies across a range of athletic contexts, including bobsleigh and other sports.

Earlier investigators demonstrated greater neck strength in male athletes,[32,59] which was confirmed by our data. However, given the lack of differences between males and females for any kinematic variables we measured, our data suggest that, despite the recording of an HAE, multidirectional absolute neck strength in bobsleigh athletes does not appear to be strongly predictive of the resultant kinematics. This result partly agrees with observations from rugby union[32] and boxing,[42] which also indicated no sex differences in the kinematic variables explored (ie, PLA and PAA). Nonetheless, importantly, we also assessed neck strength relative to body mass. Assuming head mass is at least partially proportional to body mass,[60] based on our data, the lack of sex differences in kinematics may be driven by relative neck strength.

The influence of setting (ie, training versus competition) has been identified as a risk factor for head impacts. For example, in female soccer athletes, higher PLV, PAV, PLA, and PAA

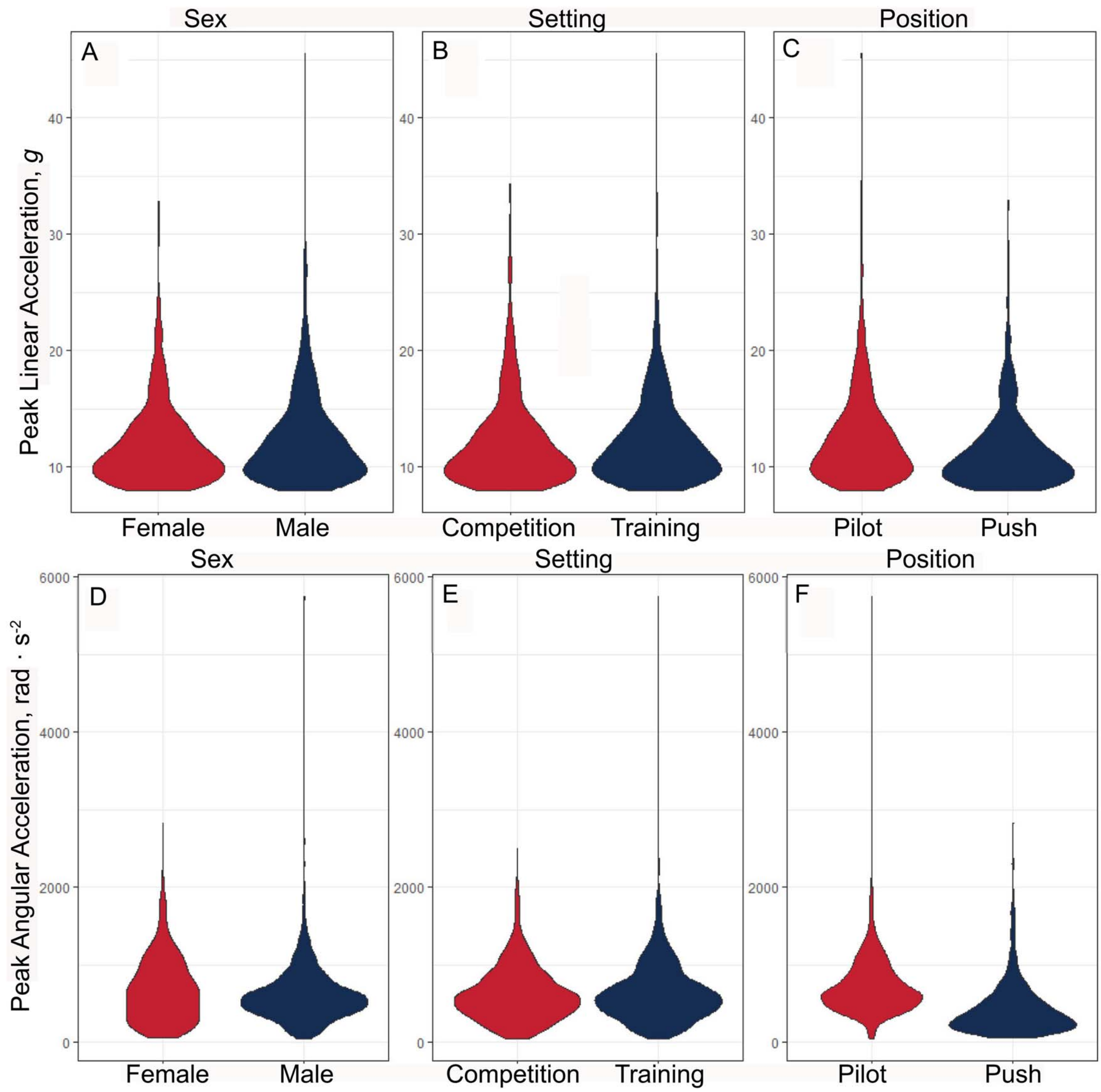

Figure 3. Peak linear acceleration (g) by, A, sex, B, setting (training versus competition), and C, position; and peak angular acceleration (rad·s^{-2}) by, D, sex, E, setting, and F, position during the 2021–2022 World Cup season.

were noted during competition.[61] Our analysis suggests that bobsleigh athletes are exposed to the same kinematic factors in training and competition, even when the number of runs completed is controlled in both scenarios. This highlights the similarity and specificity of bobsleigh training relative to competition.

We also tested for differences in bobsleigh position because of speculation that, due to the visual cues of the track available to the pilots but not push athletes (in both 2- and 4-person bobsleigh), the pilots may have greater visual feedback to brace themselves before cornering events and thereby reduce kinematic forces imposed on the head. Our results suggest, however, that this likely does not play a prominent role in the forces felt by bobsleigh athletes. The push athletes rely on both reaction and memory of the track to make the appropriate biomechanical adjustments, and these data may indicate that they can do so as well as the pilots, at least as far as this is reflected in head kinematics.

Importantly, though, the current data and analysis included only 3 pilot athletes, thus limiting position comparisons. Pilots not only have potential risk profile differences due to their first position in the bobsleigh, but they are also required to complete a minimum number of runs on each track in the week before competition, which necessitates more exposure risk than for push athletes who do not have a minimum practice run requirement. Therefore, differences between bobsleigh position in HAEs, head impacts, and concussion risk should continue to be explored in larger datasets. It should be noted that we recorded only 1 crash event during this study, but the magnitudes of head kinematic metrics were 2 to 6 times greater than the HAEs captured during noncrash runs. Extrapolating from this single event, crash events would appear to pose a greater risk of head injuries for the athlete. Nonetheless, because crashes are rare events, it is also critical to evaluate health risks to athletes during otherwise clean runs, as the cumulative HAEs from both clean and crash runs may contribute to long-term health concerns.

It is important to understand the risk profile associated with participation in bobsleigh. Throughout the 2021–2022 World Cup season, 11 time-loss injuries (incidence rate = 24.55 [95% CI = 10.04, 39.06]) were incurred among Team USA bobsleigh athletes. We also found that male athletes

Table 2. Neck Strength Measures Between Male and Female Athletes

Variable	Males Absolute N	Males Relative N·kg^{-1}	Females Absolute N	Females Relative N·kg^{-1}	Model Results Absolute N	Model Results Relative N·kg^{-1}
Extension	32.4 ± 8.73	0.32 ± 0.08	17.90 ± 6.73	0.24 ± 0.09	14.48 (4.46, 24.51)[a]	0.08 (−0.03. 0.19)
Flexion	26.80 ± 10.6	0.27 ± 0.10	14.30 ± 5.86	0.19 ± 0.08	12.45 (1.47, 23.43)[a]	0.08 (−0.04, 0.19)
Rotation						
Left	18.2 ± 6.26	0.18 ± 0.07	10.10 ± 4.81	0.13 ± 0.06	8.02 (0.84, 15.20)[a]	0.05 (−0.03, 0.13)
Right	17.2 ± 6.42	0.17 ± 0.07	9.64 ± 3.98	0.13 ± 0.05	7.60 (0.72, 14.48)[a]	0.04 (−0.03, 0.12)
Side flexion						
Left	20.2 ± 5.89	0.20 ± 0.06	12.10 ± 4.86	0.16 ± 0.06	8.15 (1.54, 14.76)[a]	0.04 (−0.03, 0.12)
Right	19.0 ± 7.04	0.19 ± 0.07	12.30 ± 4.34	0.16 ± 0.05	6.69 (−1.12, 14.49)	0.02 (−0.06, 0.11)
Side flexion with rotation						
Left	18.4 ± 4.95	0.18 ± 0.05	10.00 ± 3.14	0.13 ± 0.04	8.39 (3.06, 13.73)[a]	0.05 (−0.01, 0.11)
Right	17.6 ± 4.57	0.18 ± 0.05	9.83 ± 3.64	0.13 ± 0.05	7.78 (2.46, 13.09)[a]	0.05 (−0.02, 0.11)

[a] Indicates a difference between males and females. Data are reported as mean ± SD for males and females and as the magnitude of differences (95% CI) for model results.

(compared with females) and push athletes (compared with pilots) were at greater risk of sustaining a time-loss injury. Further, the athletes in our investigation were at greater risk of sustaining an acute injury than a chronic injury or exacerbation of a previous injury. Of all injuries, only 1 (9.1%) time-loss injury resulted from an SRC, in contrast to the prevalence of SRC previously reported among other sliding-sport athletes (injury proportion = 12%–15%).[11–14] This low number of observed SRC occurred despite the high prevalence and magnitude of the recorded HAEs. Yet we did not assess how these HAEs may have affected acute changes in mental health nor long-term effects on cognitive, brain, and mental health function, which subconcussive HAEs could theoretically influence.[15–26] Future authors should evaluate these possible effects on other aspects of athlete health.

Table 3. Incidence of Injury and Illness During the 2021–2022 Bobsleigh World Cup

	Frequency	Incidence Rate per 1000 Athlete Runs (95% CI)
Total	11	24.55 (10.04, 39.06)
Female	2	4.46 (0, 10.65)
Male	9	20.09 (6.96, 33.21)
Position		
Driver	3	6.7 (0, 14.27)
Push athlete	8	17.86 (5.48, 30.23)
Acute or chronic/exacerbation		
Acute	8	17.86 (5.48, 30.23)
Chronic/exacerbation	3	6.7 (0, 14.27)
Type		
Cartilage	1	2.23 (0[a], 6.61)
Concussion	1	2.23 (0, 6.61)
Infection	1	2.23 (0, 6.61)
Laceration	1	2.23 (0, 6.61)
Pain	2	4.46 (0, 10.65)
Soft tissue	1	2.23 (0, 6.61)
Sprain	2	4.46 (0, 10.65)
Strain	2	4.46 (0, 10.65)
Anatomical location		
Head	2	4.46 (0, 10.65)
Shoulder	1	2.23 (0, 6.61)
Knee	1	2.23 (0[a], 6.61)
Leg	2	4.46 (0, 10.65)
Ankle	1	2.23 (0, 6.61)
Back or pelvis	3	6.7 (0, 14.27)
Skin	1	2.23 (0, 6.61)

[a] The lower bound of the 95% CI was deliberately set to 0, considering only plausible values.

Our work had numerous strengths, including the number of observations, the elite and novel population tested, and our technological approach to assessing HAEs. However, several limitations should be noted. First, only 12 athletes from the same country were included. This limited our ability to detect small, potentially clinically meaningful differences between sexes, settings, and positions. Moreover, the sample findings may not extrapolate to non-Olympic-level athletes or other Olympic-level athletes from countries with different levels of sport participation, resources, and training histories. Relatedly, athletes were permitted to wear their own helmets of choice. Therefore, slight differences in helmet structure and materials may have contributed to between-athletes differences in HAE kinematics. Furthermore, although a mouthguard-based monitoring system confers many benefits, athletes may also bite and chew the mouthguards throughout the season, perhaps altering the fit over time. Athletes wore the same mouthguard throughout the study, which may have affected the results, although to what extent is unknown. Second, our positional analysis compared only 2 levels (pilot versus push). In the 4-man bobsleigh, push athletes can be in the second, third, or fourth position in the bobsleigh, which may theoretically influence the HAEs. Our athletes did not have a consistent position in the bobsleigh among runs, and unfortunately, we were unable to account for actual position in the bobsleigh. Future researchers should determine whether the push positions within the bobsleigh do indeed influence the HAEs. Third, the present analysis lacked a degree of contextual data that could have refined it. For example, video footage of all training and competition runs was unavailable, and thus, the proportion of recorded HAEs due to helmet-to-sled contact compared with whiplash events was unknown. Further, because video footage was not available, we were unable to distinguish false-positive or false-negative impact rates (CHAMPS 7d) or conduct a blinded impact review (CHAMPS 5a). These rates and events may differentially manifest by sex, setting, and position and deserve follow up. Lastly, we used a cutoff of $>8g$ for determining an HAE, which was decidedly lower than previous thresholds (eg, 10–20g). This decision was based partly on the observation[62] that mouthguard-based accelerometers can miss many head impacts. As a result, we wanted to take a more conservative approach in defining when a possible head impact occurred. A less conservative approach (ie, a higher threshold for a head impact) may reveal differences that were not observed using the current approach, as our procedure may also have introduced more spurious events. Governing bodies for sliding sports should establish sport-specific thresholds of mouthguard kinematic data that maximize sensitivity and specificity for detecting head impacts and concussive events.

CONCLUSIONS AND FUTURE DIRECTIONS

We provided a foundational understanding of head impact kinematics that elite bobsleigh athletes experience during training and competition. Despite the 1900 HAEs (each $\geq 8g$) recorded in bobsleigh athletes, only 1 SRC was noted, indicating that almost all HAEs were subconcussive. Future work is required to determine the clinical significance of repetitive subconcussive HAEs on short- and long-term physical and mental health in sliding-sport athletes.

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SUPPLEMENTAL MATERIAL

Supplemental Table. International Bobsleigh Federation Bobsleigh Track Details.

Supplemental Figure. An example of a head acceleration event acceleration time series.

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