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Concussions in Sports

Laboratory Reconstructions of Real-world Bicycle Helmet Impacts

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Abstract—The best way to prevent severe head injury when cycling is to wear a bike helmet. To reduce the rate of head injury in cycling, knowing the nature of real-world head impacts is crucial. Reverse engineering real-world bike helmet impacts in a laboratory setting is an alternative to measuring head impacts directly. This study aims to quantify bike helmet damage using computed tomography (CT) and reconstruct real-world damage with a custom, oblique test rig to recreate real-world impacts. Damaged helmets were borrowed from a helmet manufacturer who runs a helmet warranty program. Each helmet was CT-scanned and the damage metrics were quantified. Helmets of the same model and size were used for in-lab reconstructions of the damaged helmets where normal velocity, tangential velocity, peak linear acceleration (PLA) and peak rotational velocity (PRV) could be measured. The damage metrics of the in-lab dropped helmets were quantified using the same CT scanning process. For each case, a multiple linear regression (MLR) equation was created to define a relationship between the quantified damage metrics of the in-lab tested helmets and the associated measured impact velocities and kinematics. These equations were used to predict the impact kinematics and velocities from the corresponding real-world damaged helmet based on the damage metrics from the original damaged helmet. Average normal velocity (3.5 m/s), tangential velocity (2.5 m/s), PLA (108.0 g), PRV (15.7 rad/s) were calculated based on a sample of 23 helmets. Within these head impact cases, five notes reported a concussion. The difference between the average PLA and PRV for concussive cases versus other impacts were not significantly different, although the average impact kinematics for the concussive cases (PLA = 111.4 g, PRV = 18.5 rad/s) were slightly higher than the remaining cases (PLA = 107.1 g, PRV = 15.0 rad/s). The concussive cases were not indicative of high magnitude impact kinematics.

Keywords—Biomechanics, Bicycle helmet, Concussion, Acceleration.

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INTRODUCTION

Cycling is a popular recreational sport, mode of transportation, and form of exercise. Bike crashes are rare, but they can result in mild to serious injuries or even death. Head injury is present in most bike crashes which result in death.[15,25] The best way to prevent severe head injury while cycling is to wear a bike helmet.[1,11,12,17,21,27] Helmets aim to decrease the energy delivered to the head upon impact. Historically, helmets have successfully reduced linear acceleration of the head associated with focal injuries to the brain, such as hematoma or contusion.[14]

Bike helmet standards were designed to ensure bike helmets manage the energy transferred to the head during impacts. The standard set by the Consumer Product and Safety Commission (CPSC) says that helmets must not exceed a peak linear acceleration (PLA) of 300 g (< 50% risk of skull fracture) during testing to be sold in the United States.[10] However, rotational motion which results in shear forces between the skull and brain interface and is associated with diffuse brain injuries, including concussions, is not considered in bike helmet standards.[18]

New helmet technologies have been introduced to the market in recent years, such as the Multi-directional Impact Protection System (MIPS® AB, Täby, Sweden) and WAVECEL™.[16] MIPS was designed to reduce rotational head acceleration resulting from helmet impacts and WAVECEL™ was designed to reduce both linear and rotational head kinematics during impact. Mitigating the magnitude of rotational motion of the head during an impact is crucial to lowering concussion incidence. Studies evaluating head impact kinematics in other sports have found that concussions occur over a wide linear and rotational kinematic range, with variance depending on many factors, such as age, sex, and previous concussion history.[22,28,30] However, equivalent studies regarding cycling related head impact kinematics are sparce due to the spontaneous and unpredictable nature of cycling accidents.

Collecting impact data for real-world head impacts is challenging. Unlike football or hockey, it is impractical to put sensors inside of bike helmets due to the low occurrence rate of crashes at the individual level. Reverse engineering bike helmet impacts in a laboratory setting is an alternative to measuring head impacts directly. In the past, helmet damage reconstruction studies have been limited by unrealistic test rigs or simplified damaged measurement techniques.[26,29] CT scanning has been suggested to be an objective and precise method to measure damage characteristics of bike helmets.[19] Additionally, impacting helmets at an oblique anvil is a better representation of real-world bike helmet impacts than impacting on a flat surface.[4,6,7,20]. A novel study by Bland et al. conducted in-lab reconstructions of damaged bike helmets using CT scanning to quantify helmet damage and an oblique test rig to replicate real-world impacts.[3] This novel method measured helmet damage with a measurement error of .5 mm and recreated helmet damage in a laboratory setting to a high degree of accuracy compared to previous studies. This was accomplished using multiple impact surfaces and adjustable impact angles. In the study by Bland et al., helmets were collected from hospital systems after cyclists were admitted. This method produced realistic boundary conditions for real-world cycling related head impacts. Increasing the volume of data produced with this methodology can aid in the future development of injury risk functions, the understanding of cycling related concussion, and improve helmet design.[3] This study applied the methodology for reconstructing real-world bike helmet impacts created by Bland et al. to a sample of damaged bike helmets borrowed from a helmet manufacturer though their warranty program.

MATERIALS AND METHODS

Bike helmets damaged in real-world crashes were collected through a manufacturer warranty program. The damaged helmet and a note (if one were given by the returning customer) describing crash details were evaluated for each reconstruction. Helmets with no apparent damage, which was determined by searching for scraping on the shell or crush of the EPS liner, were not used in this study. Each helmet was CT scanned, along with an undamaged match in model and size. A 3D rendering of the helmets was created, and the damage metrics were quantified by subtracting the measurements of the damaged helmet from those of the undamaged helmet. This damage was replicated with a drop tower using undamaged helmets of the same make and model in the lab where the velocities and kinematics of each drop could be measured. Lastly, using linear regression and the relationships between the damage metrics and the impact velocities and kinematics, the impact velocities and kinematics of the original crash were estimated.

Quantifying Helmet Damage

Each helmet was CT scanned (Aquilion, Canon Medical Systems, Tustin, CA: 120 kV, 200 mA, 0.625 9 0.625 mm pixel spacing, 0.5 mm slice thickness) along with a matching undamaged helmet of the same model and size. The scans were segmented in MIMICS 23.0 software (Materialise, Leuven, Belgium) to isolate the EPS foam liner from the plastic helmet shell and surrounding environment. The foam liner corresponded to Hounsfield units (HU) of -950 to -850. Additional automatic and manual editing of the foam liner was applied when necessary. Liner masks were converted into a 3D surface mesh model ($>$ 600,000 triangular elements) and imported into 3-Matic 15.0 (x64) (Materialise, Leuven, Belgium). Each damaged helmet model was locally and globally aligned with the matching undamaged helmet model. The damage metrics were calculated by subtracting the damaged helmet measurements from those of the undamaged helmet. The metrics used to quantify helmet damage were crush depth, crush area, crush volume, scrape length, and centeredness of the point of maximum crush. Damage metrics were quantified using the methods described in Bland et al.; pictures of the damage quantification process can also be found in that publication.[3] Impact location of the damage was also measured. The impact location was determined by the point of maximum crush and was generalized into five regions of the helmet based on azimuth (front, front boss, side, rear boss, and rear) and three regions of the helmet based on elevation (rim, middle, and top).

Impact Testing

Multiple undamaged helmets of the same model and size were purchased for each damaged helmet. Anywhere from three to eight new helmets were used for damage replication per damaged helmet. An oblique impact rig was used for the in-lab damage replications. The impact surface angle was adjustable in increments of 5° between 0° and 60° relative to the horizontal of the impact surface. National Operating Committee on

Standards for Athletic Equipment (NOCSAE) headforms in small (53.4 cm circumference, 3.54 kg mass), medium (57.6 cm, 4.42 kg), and large (61.4 cm, 5.35 kg) were used depending on the damaged helmet size and manufacturer fit recommendations. NOCSAE headforms are biofidelic and are commonly used in sports helmet testing.[9] Multiple impact surfaces were used to produce different types of helmet damage. If the helmet included a note describing the accident, the description was considered when deciding which surface to use. Additionally, visual inspection of the helmet damage profile along with the CT scan rendering of the damage was evaluated to determine the best impact surface. 80-grit sandpaper was used to simulate pavement and create long scraping profiles with generally even crush depth. A rough surface created from sand and rocks in epoxy was used to imitate gravel and create small, focal impacts. Curbstone and hemispheric anvils were also available, but not necessary to recreate damage for this sample of helmets.

The headform was positioned so that the impacts were body-driven, meaning the head was leading the body upon the impact, unless the note describing the accident indicated otherwise. Positioning of the helmet was measured using a dual-axis inclinometer (DMI600, Omni Instruments, Dundee, UK) for high repeatability between tests. The resultant impact velocity was measured using a photogate (BeeSpi V, NaRiKa Corp., Tokyo, Japan). Impact kinematics were collected at 20 kHz using three linear accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA) and a triaxis angular rate sensor (ARS3 PRO-18K, DTS, Seal Beach, CA) at the headform center of gravity (CG). The kinematic data from each helmet test were recorded using a TDAS SLICE PRO SIM data acquisition system (DTS, Seal Beach, CA).

The impact location was determined during damage quantification by finding the point of max crush. Scrape length was determined by the angle of the anvil. The angle of the anvil was adjusted using a hinge and series of holes which could be locked in place with a bolt in five-degree increments. For each test, the scrape direction and helmet location were matched to the original damage. The magnitude of the damage metrics (crush depth, crush area, crush volume, scrape length, and centeredness) were replicated at a lower and higher magnitude than the damage metrics from the original helmet. A combination of anvil angle and drop height were used to adjust the damage metric magnitudes.[2] The damage metrics of the in-lab tested helmets were quantified using the same process as the original damaged helmets which is described above. If the point of maximum crush was located more than half the distance of the crush area radius away from the original point of maximum crush, the trial was omitted.

Data Processing

Linear acceleration was processed using a channel frequency class (CFC) 1000 filter (SAE J211) and rotational velocities were processed using a CFC of 175. The drop data was processed using a custom MATLAB (R2020a, Mathworks Inc, Natick, MA) script to obtain the peak linear acceleration (PLA), peak rotational velocity (PRV), and duration of impact. The normal and tangential velocities were calculated based on the angle of the anvil and the resultant impact velocity.

Data Analysis

For each damaged helmet, correlation analyses using Pearson's correlation coefficient were computed between each damage metric (crush depth, crush area, crush volume, scrape length, and centeredness) and each impact velocity (normal velocity and tangential velocity) and kinematic (PLA and PRV). This resulted in twenty correlation analyses per helmet case. The two damage metrics with the highest correlation coefficient and their interactions were used to as explanatory variables in a multiple linear regression (MLR) model. If the process to quantify helmet damage resulted in a negative measurement, that damage metric was removed as an option for input into the MLR (e.g. the difference in crush volume between the damaged and undamaged helmets was small and the global registration error resulted in the damaged crush volume to be larger than the undamaged crush volume). Each equation defined an associated between the chosen damage metrics and one of the impact velocities (normal velocity and tangential velocity) or kinematics (PLA and PRV). The relationship between damage metrics and the resulting impact kinematics has been previously established by Bland et al.[2,3]

Various combinations of the chosen damage metrics and their interactions were used to create the model, until the model with the highest R^2 value was found. The significance of the parameters was monitored, removing the least significant terms until all terms in the model were significant, or until there was one term and an intercept remaining. The maximum number of parameters for each model was two less than the number of in-lab tests for that helmet. If models created from different combinations of parameters produced the same level of significance, the model with the greater amount of terms was used. The resulting equation would define a relationship between the chosen damage metrics and one of the impact velocities or kinematics. This equation was used to predict the corresponding impact velocity or kinematic for the original damaged helmet by plugging in the necessary original damage metric(s) as the new explanatory variable(s). During the in-lab testing, some trials were designated to exceed the damage metrics of the original helmet, and some were designated to fall inferior to the original damage metrics. The reason for this is when plugging the original damage metric into the MLR equation, the outcome would be interpolated, instead of extrapolated, improving the confidence of the results.

This process was conducted to estimate the normal velocity, tangential velocity, PLA, and PRV for each damaged helmet. The standard errors for each predicted velocity and kinematic were normalized (NSE) by dividing it by the corresponding average value to account for the difference in magnitude between each velocity and kinematic. The percent difference between the velocities and kinematics were calculated to compare between the results in this study and the results found by Bland et al. All statistical analyses were conducted in RStudio (V 1.0.136, RStudio, Inc., Boston, MA).

RESULTS

Impact velocities and kinematics were estimated for 23 helmets damaged in real-world bike crashes. Of these helmets, 21 helmets included a note with some description of the incident, one helmet included a picture of the crash scene and damage to the bike without a note, and one helmet did not include any information. 16 cases reported some sort of injury, including 5 concussions. 60.9% of impacts were at the middle, 30.4% of impacts were at the rim, and 8.7% of impacts were on the top. 43.5% of impacts occurred on the side of the helmet, 34.8% of impacts occurred at the rear boss, 13.0% of impacts occurred at the front boss, and 8.7% of impacts occurred at the rear. The most common azimuth and elevation bin was the side middle (26.1%), followed by the rear boss rim (13.0%) and rear boss rim (13.0%).

Damage Characteristics

Five damage metrics were used to characterize helmet damage (Fig. 1). Crush depth had a median of 5.0 mm [IQR = 2.5–7.1 mm], crush area had a median of 25.73 cm^2 [IQR = 13.2–36.3 cm^2], crush volume had a median of 8.5 cm^3 [IQR = 3.5–13.9 cm^3], scrape length had a median of 5.3 cm [IQR = 4.2–6.9 cm], and centeredness had a median of 1.2 [IQR = .6–2.3]. The case which resulted in the maximum crush depth was crushed in the rear region of the helmet. The maximum values for crush area and crush volume were also observed from this case.

The two impact surfaces were almost evenly split between cases: 80-grit sandpaper (12 cases) and the rocky surface (11 cases). The majority of helmets fit a medium headform (13), followed by small (6) and large (4). An average of four undamaged helmets were used to replicate each case. The smallest number of replications for a case was three helmets to calculate the MLR. Within the replications for one case, the resulting damage metrics needed to range above and below the damage metrics from the original helmet. Some cases required additional tests to reach this goal. The most replications for one helmet case was eight, but two of those replications were eventually deemed unusable based on the large difference between points of maximum crush. The average impact angle was 37° with a minimum angle of 15° and a maximum of 60°.

Overall, crush area and crush depth were the damage metrics most used to predict impact velocities and kinematics (including interaction terms). Normal velocity was most commonly estimated by crush depth and crush area, closely followed by crush volume. Tangential velocity was most commonly estimated by crush depth, closely followed by crush area and scrape length. PLA was most commonly estimated by crush area. PRV was most commonly estimated using centeredness, closely followed by scrape length and crush depth.

High correlations between one damage metric and one impact characteristic were not observed for every case. Some cases used a combination of loosely correlated damage metrics to create an MLR with a moderate (0.4–0.7) R^2 value. PRV had the highest NSE (12%) of all the impact characteristics (Normal Velocity = 6%, Tangential Velocity = 8%, PLA = 7%).

Impact Characteristics

We were able to estimate normal and tangential velocity, PLA, and PRV for all 23 helmets (Fig. 2). The averages for each impact velocity and impact kinematic were calculated: normal velocity (3.5 m/s), tangential velocity (2.5 m/s), PLA (108.0 g), PRV (15.7 rad/s).

Within these head impact cases, five notes reported a concussion. Two of these cases reported being taken to the hospital and diagnosed with a concussion. One case reported a headache and sensitivity to light and noise, which lasted for two days, which we interpreted as a concussion. One case had extensive loss of memory extending from before the crash to after the hospital visit, which we also interpreted as a concussion. One case reported a concussion without further details.

The impact velocities and kinematics from the concussive cases were compared to the velocities and

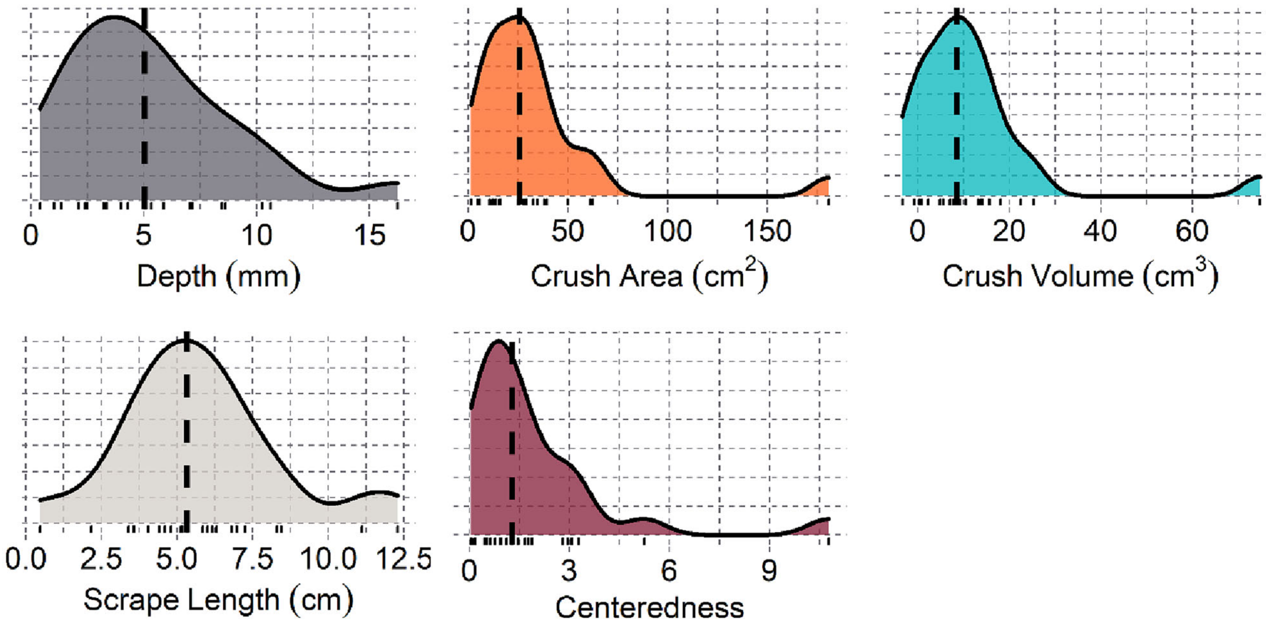

FIGURE 1. The distributions of damage metrics measured from a sample of helmets damaged in real-world bike accidents. The dashed line indicates the median value for each damage metric.

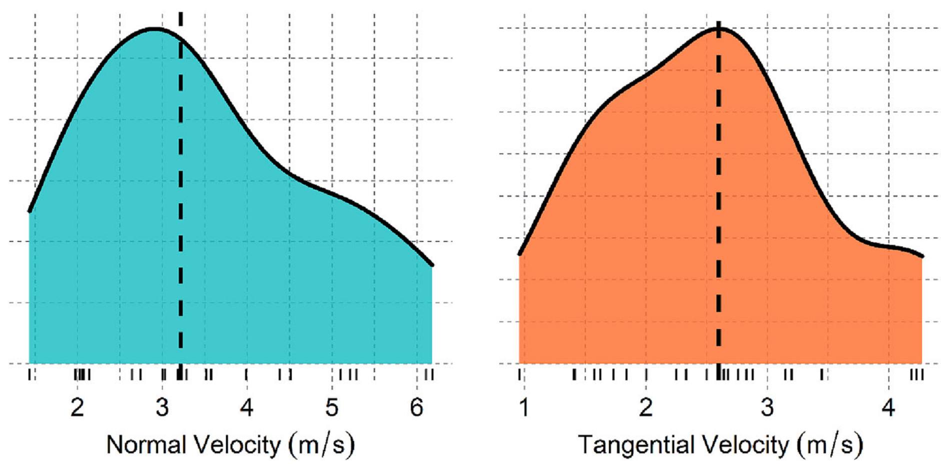

FIGURE 2. The distribution of normal velocity, tangential velocity, peak linear acceleration (PLA), and peak rotational velocity (PRV) estimated for 23 real-world bike helmet impacts. The dashed lines represent the median value of each metric.

kinematics from the remainder of the sample. Within the concussive sample, the average normal velocity was 3.7 m/s ± 0.2 and the average tangential velocity was 2.1 m/s ± 0.3. The average impact kinematic values for the concussive cases (PLA = 111.4 ± 34.5 g, PRV = 18.5 ± 7.0 rad/s) were slightly higher than the remaining cases (PLA = 107.1 ± 44.0 g, PRV = 15.0 + 10.0 rad/s). The differences between PLA and PRV between the concussive cases and the remainder of the sample were not significantly different. The impact kinematics for the concussive cases fell within the range of the kinematics for the impact cases (Fig. 3).

DISCUSSION

This study's objective was to estimate the impact velocities (normal and tangential velocity) and impact kinematics (PLA and PRV) of real-world bike helmet impacts using the relationship between damage metrics and impact boundary conditions. This study replicated the methods presented in Bland et al. to recreate real-world bike helmet impacts with a high degree of accuracy and objectivity.[3] The impact velocities estimated in this study align with those estimated using computational simulations[7] and other in-lab helmet

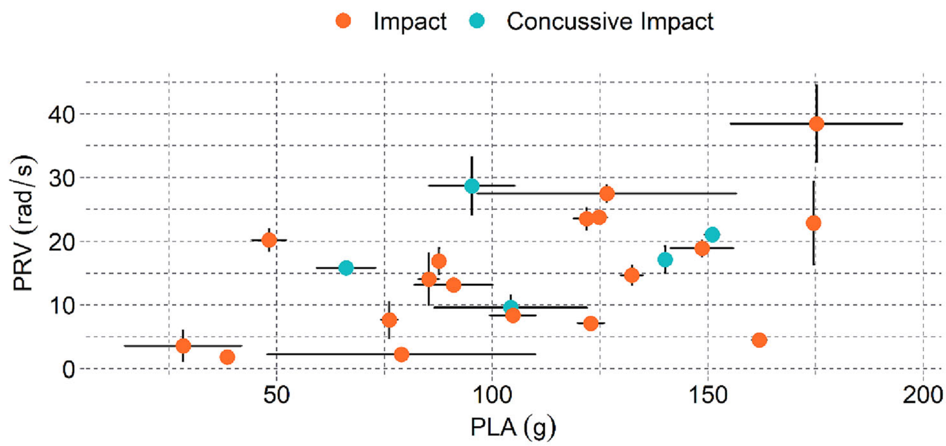

FIGURE 3. The relationship between peak linear acceleration and peak rotational velocity for bike helmet impacts, including concussive cases. The lines emitting from each point represent the standard error of each prediction.

reconstructions.[3,26] During standards testing, helmets are dropped at normal velocities of 4.8 m/s and 6.2 m/s. The average normal velocity from this sample was 3.5 m/s, and the maximum value was 6.2 m/s. The velocities used in standard testing are representative of the range of the velocities in real-world impacts.

Third-party helmet testing systems use oblique impact rigs to test bike helmets. This testing is not necessary to sell helmets on the market, but it gives consumers additional insight into helmet performance past standards testing. These test systems drop helmets onto a 45° anvil to evaluate helmets' ability to mitigate rotational motion of the head. The resultant impact velocity of these tests ranges from 6.3 to 8 m/s.[5,8,13] These velocities are associated with normal and tangential components equaling 4.5 and 5.7 m/s on a 45-degree anvil. These velocities are much higher than the average normal (3.5 m/s) and tangential (2.5 m/s) velocities found in this study. However, helmet tests are designed to evaluate helmets at high velocities to evaluate performance during high-risk impacts.

When we compared the data between this study and the data in a previous study by Bland et al., we saw the maximum velocities and kinematics tended to be lower in this study in all cases except for normal velocity, when both studies produced the same maximum value.[3] The linear metrics (normal velocity and linear acceleration) showed very similar medians between the two sample groups (Fig. 4). The median normal velocity in this study was 5.5% lower than the median normal velocity found in Bland et al.,[3] and the median linear acceleration was 4.2 % higher. However, the rotational metrics (tangential velocity and rotational velocity) were much lower in the current sample. The median tangential velocity was 45.8% lower in this study than in the study by Bland et al.,[3] and the rotational velocity was 40.4% lower. The difference between the estimated velocities and kinematics between the two samples is most extreme in the median tangential velocity and rotational velocities (Fig. 4).

The helmets in this study were borrowed from a helmet manufacturer warranty program, whereas the helmets in Bland et al. were collected from crashes which resulted in hospitalizations.[3] The sample in Bland et al, most likely included helmets from riders who, on average, were involved in crashes of a higher severity, due to the fact that they were hospitalized. This would contribute to the higher rotational forces upon impact. Collecting helmets from a manufacturer warranty program may have resulted in a lower threshold of damage for the helmet to be returned, due to the fact that manufacturers recommend helmets be replaced after an impact even if there is no visible damage.

In both Bland et al. and the current study, the average PLA and PRV for the concussive cases were higher than cases which did not report a concussion.[3] The PLA for concussive cases (111.3 ± 34.5 g) in this sample was similar to those estimated from Bland et al (109.8 ± 55.0 g). The PRV for concussive cases was slightly higher in Bland et al. (19.3 ± 7.4 rad/s) than the concussive cases in this study (18.5 ± 8.9 rad/s).[3] The average PLA for concussive impacts in this sample (111.4 ± 34.5 g) aligns with the average PLA for concussive hits in American football (104 ± 30 g).[23] Similarly, the average PRV for concussive impacts in this sample (18.5 ± 7.0 rad/s) aligns with the average PRV found in concussion in American football (22.3 rad/s).[24] It's important to note that we did not receive medical records from any of these cases, so we cannot comfirm that the cases which did not report a concussion in their description of the accident did not experience a concussion.

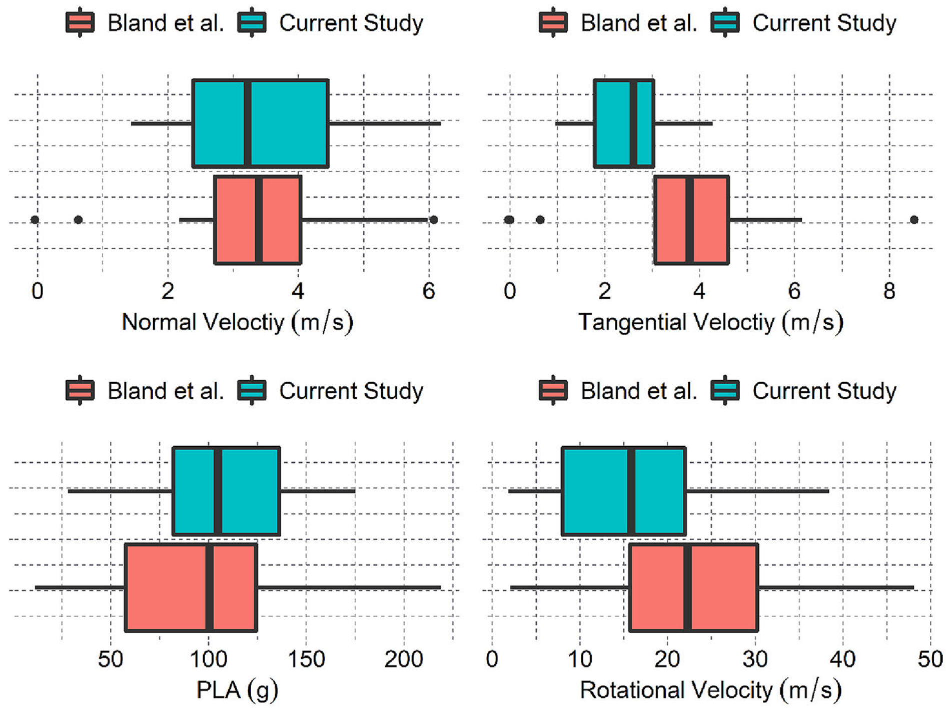

FIGURE 4. The estimated impact velocities (normal and tangential velocity) and kinematics (PLA and PRV) of real-world bike accidents in this study compared to the study by Bland et al.[3] The velocities and kinematics from Bland et al. were found to span a much wider range than this study. The tangential velocity and peak rotational velocity were much greater in Bland et al.

The limitations of this study affect our ability to estimate impact boundary conditions with minimal error. The CT scans used in this study have a resolution of .5mm. This is compounded with the global registration error in 3-Matic (ranging from .9 mm to .4 mm) when aligning the damaged and undamaged helmets. These measurement errors can affect final estimates from the regression equations. In some cases, measuring the damage metrics resulted in a negative number. When this occurred, we removed that damage metric as an option in the creation of the MLR. This could have resulted in an MLR which was less robust due to one less damage metric available for regression. Additionally, the sample size of this study was affected by the availability of new helmets of the same model and size as the damaged helmets in our study. We were not able to use multiple damaged helmets due to a lack of new matches. This unavailability was due partly to the age of some helmet models that are no longer being produced and party due to the increase in demand and low supply of bike helmets during the COVID-19 pandemic. Additionally, one helmet model had metal artifacts within the EPS foam which were too embedded to remove prior to scanning and too intrusive to remove in post-processing, so we could not use the helmet in the study.

The test rig used in this study can accommodate multiple variables in helmet impacts, including impact surface, angle of impact, and impact location. However, our test method and set-up only allow for one impact per trial. It is common for the helmet to experience more than one impact during a crash. Our study design only recreated the most severely damaged region, which was assumed to be the initial impact. This would result in our impact velocity and kinematic predictions to only account for a portion of the impact experienced by the cyclist during the crash. In some cases, damage metrics were left out of the analyses because the test rig could not achieve a magnitude of damage higher than the original helmet. On the other hand, some damage profiles were extremely small. There is a minimum height on the drop tower to ensure the helmet is released from the carriage before impacting the anvil, and in some cases this minimum height was too high to recreate the small damage magnitudes.

The data in this study can inform test methods for improved helmet testing and design. As more test methods are designed to evaluate rotational velocities, real-world impact characteristics are of increasing importance. As helmet design is influenced by both standards testing and supplemental third-party testing, creating test methods that accurately reflect real-world impacts is critical to helmet manufacturers. Increasing the amount of available bike helmet reconstruction data, including cases that resulted in head injury, is key to improving helmet testing and design and decreasing the risk of brain injury while cycling.

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CONFLICT OF INTEREST

The authors declare that they have no conflict of interest.

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