A Novel Helmet Testing Apparatus for Assessing Rotational Head Kinematics and Its Applications in Military Helmet Design

Shuting Kou, Jaden Romero, Aravind Sundaramurthy, Ryan Doris, **Ron Szalkowski**

Team Wendy, LLC, 17000 St. Clair Avenue, Cleveland, OH 44110, USA
rszalkowski@teamwendy.com

Abstract. Traumatic brain injuries (TBIs) are a significant concern among military personnel, often leading to lasting cognitive and physical impairments. Traditionally, linear kinematics has been emphasized as a key predictor of head injuries. However, recent research increasingly points to rotational kinematics as a primary contributor. The current Department of Transportation (DOT) linear drop test, which utilizes a monorail drop tower with rigid neck and headforms, primarily captures linear kinematics, offering limited insights into rotational dynamics. To address this limitation, we developed the Device for Rotational Evaluation of Wearables (DREW), a novel testing platform designed to simulate realistic head-to-ground impacts and measure both linear and rotational kinematics. DREW incorporates a 50th percentile Hybrid-III head, neck, and torso, equipped with linear accelerometers, angular rate sensors, and neck load cells, enabling comprehensive helmet evaluations across various impact locations (front, rear, and sides) at velocities up to 6.1 m/s. Unlike monorail systems, DREW can also simulate the whiplash effects often experienced during backward falls. To determine if DREW can differentiate between the performance of different helmet designs and technologies, three bicycle helmets employing a variety of rotational mitigation technologies were evaluated. Furthermore, testing Team Wendy's ballistic helmet liners on DREW revealed that a high-density liner (optimized for impacts of 4.3–5.2 m/s), which outperformed a baseline liner (designed for 3.0 m/s impacts) under the current monorail drop test method, exhibited higher peak linear acceleration (20%) and peak angular acceleration (27%). These results suggest that the current monorail drop test methods may not accurately represent realistic impacts and could encourage the development of overly stiff helmet liners.

1. INTRODUCTION

Service members face frequent head impacts in both training and combat, with blunt force trauma posing a significant risk. Notably, Wojcik et al. documented a 15% hospitalization rate due to blunt head trauma during conflicts in Iraq and Afghanistan [1]. Although modern ballistic helmets have effectively reduced severe head injuries, such as penetrating trauma and skull fractures, they may not provide adequate protection against mild traumatic brain injuries (mTBIs). Indeed, between 2000 and the second quarter of 2024, 509,477 Service members sustained at least one traumatic brain injury (TBI), with 81.9% classified as mTBI (e.g., concussions) [2]. Recent studies have revealed that rotational kinematics, which generate substantial shear forces in brain tissue, are widely recognized as the predominant mechanism underlying concussive injuries [3, 4].

Rotational head kinematics, a common phenomenon during impact or impulsive head loading, plays a critical role in brain trauma. Due to the biomechanical properties of the brain, brain tissue deforms more readily in response to shear forces than to other forms of mechanical loading [5]. Rapid head rotations induce extensive shear forces throughout the brain, thereby elevating the risk of shear-induced tissue damage. Multiple studies across different laboratories have confirmed that shear deformation caused by rotational kinematics is the principal mechanism of concussion [4, 6–9]. This evidence underscores the critical need for next-generation helmet designs that specifically target the mitigation of rotational kinematics to reduce mTBI risk and enhance the overall protection.

Despite these insights, current ballistic helmet testing standards, such as AR/PD 10-02, predominantly focus on mitigating linear kinematics. These protocols typically utilize a Department of Transportation (DOT) headform dropped onto an anvil via a monorail drop system, a configuration that does not accurately replicate the complex rotational dynamics of the head and neck encountered in real-world impact scenarios. Alternative approaches, such as modified drop tests using a headform with flexible neck dropped onto an angled anvil to allow oblique impacts, offer improvements in simulating rotational kinematics during impacts [10]. However, these methods remain limited in replicating realistic pre-impact kinematics, such as the initial angular velocity generated by head whiplash following a fall. To better simulate a fall with accurate pre-impact kinematics, Niece et al. developed a backfall test platform that mimics the whiplash of the head during a backward fall [11]. However, this method is

https://doi.org/10.52202/081621-0046

restricted to rear impacts only, highlighting the need for a testing system capable of simulating impacts at various locations (front, sides, and rear) for a comprehensive evaluation of helmet performance.

In this study, we developed a novel test rig that simulates head-to-ground impacts while incorporating whiplash-induced head rotation, enabling a comprehensive evaluation of helmet performance. This system is capable of testing impacts at various locations, including the front, rear, and sides, thus providing a more realistic representation of the complex impact scenarios. Using this platform, we assessed the rotational kinematics mitigation capabilities of currently available bicycle helmets. Furthermore, we evaluated a high-density liner system, which had previously demonstrated improved linear acceleration mitigation in standard monorail drop tests at impact velocities of 4.3-5.2 m/s, by testing it on DREW at an impact velocity of 6.1 m/s and comparing its performance with that of a baseline liner.

2. MATERIALS AND METHODS

2.1 Device for rotational evaluation of wearables

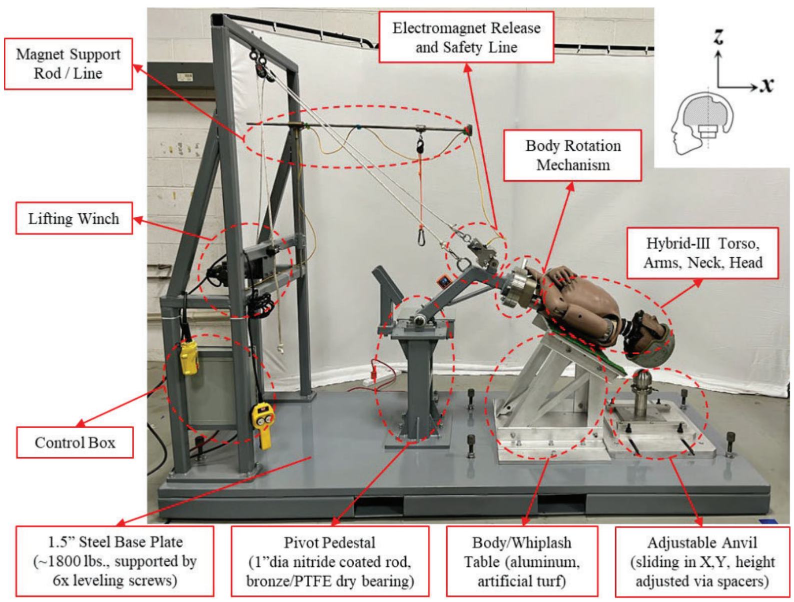

Figure 1. Experimental setup of Device for Rotational Evaluation of Wearables (DREW).

The Device for Rotational Evaluation of Wearables (DREW) is designed to simulate head-to-ground impacts while capturing both linear and rotational kinematics (Figure 1). It employs a 50th percentile Hybrid III (H3) Anthropomorphic Test Device (ATD) head, neck, and torso (56 kg; Humanetics, MI, USA) to mimic human upper-body dynamics. The ATD is mounted to a pivot leg via two adapters, one attached to the pivot leg and the other to the spine of the ATD. These adapters can rotate relative to one another and a locking mechanism secures them, permitting controlled rotation about the pivot leg's axis to simulate impacts at various helmet locations (e.g., front, rear, and side).

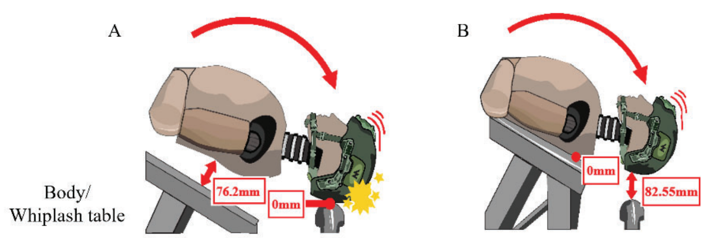

Figure 2. Description of two modes of impact: (A) headfirst and (B) whiplash test configurations.

https://doi.org/10.52202/081621-0046

The complete pivoting assembly weighs 63.5 kg, and the pivot leg is mounted on a pedestal via a shaft-and-bearing system, allowing the upper body to pivot about the y-axis (Figure 1). The pivot pedestal is rigidly fixed to an 820.0 kg steel base and supported by six leveling screws to ensure proper alignment of the test platform. An adjustable anvil system, capable of positioning along all three translational degrees of freedom, is secured to the base. In addition, a body/whiplash table is mounted on the steel base to induce two head impact modes on the ATD (Figure 2). In the first, the head contacts the anvil, followed by the torso striking the whiplash table to protect the instrumentation and ATD. In the second mode, the torso impacts the whiplash table first, inducing a whiplash motion on the head before it impacts the anvil. Finally, to perform the test, an electrically powered winch elevates the ATD to the desired height, and an electromagnetic release mechanism ensures consistent, repeatable drops by holding and then releasing the ATD into freefall. For safety, a pin through the magnet connection as well as a secondary safety line are attached until an impact test is run. The impact angle, used to characterize drop intensity, is defined as the angle between the lower pivot leg (connected to the bearing) and the horizontal plane. At the maximum impact angle of 104°, an impact velocity of 6.1 m/s is achieved.

To capture head kinematics during impact, the ATD head was instrumented with a Diversified Technical Systems (DTS, CA, USA) 6DX PRO sensor located at its center of mass. A Slice Nano modular data acquisition system, including a battery stack, Base+, and bridges, was employed to collect the sensor data. The DTS 6DX PRO sensor was selected for its ability to record head kinematics using three linear accelerometers (±2,000 g) and three angular velocity sensors (±18,000°/s), with all channels sampled at a data acquisition frequency of 30.3 kHz. Data for linear acceleration and angular velocity were recorded during impacts, and angular acceleration was subsequently estimated by differentiating the angular velocity time histories. In addition, high-speed video of the impacts was captured at 7,500 frames per second using an i-SPEED 509 camera (iX Cameras, MA, USA) to further examine the dynamics of head impact.

2.2 Testing

2.2.1 Direct vs. whiplash-induced impacts

To assess the differences in kinematics between direct head impact and whiplash-induced impact, experiments were conducted at an impact angle of 76°. For these tests, the ATD was fitted with an EXFIL ballistic helmet (Team Wendy, Cleveland, OH) and subjected to rear impacts, with five repetitions for each condition.

2.2.2 Assessment of rotational mitigation

To demonstrate the capabilities of the DREW system, we examined the performance of rotational kinematics mitigation technologies in bicycle helmets. Three helmet designs were evaluated: a baseline helmet featuring a traditional EPS (expanded polystyrene) liner (G1), a variant of the baseline helmet equipped with a Multi-Directional Impact Protection System (MIPS [12]; G2), and a helmet incorporating WaveCel [10] technology (G3). The ATD head was fitted with each helmet and subjected to frontal direct impacts on a hemispherical anvil, with each test condition repeated three times. Impacts were performed at an angle of 104°, and both linear and angular kinematics were measured and compared between the baseline helmet and those incorporating mitigation technologies.

2.2.3 Evaluation of ballistics helmet liner

Finally, we evaluated a high-density liner that was previously developed and optimized for high velocity impacts using a monorail drop test with a DOT headform. The goal for this liner was to maintain linear acceleration below 150 g during the first impact at impact velocities of 4.3 and 5.2 m/s. To assess the performance of the new liner under more realistic impact conditions of DREW, tests were conducted at an impact velocity of 6.1 m/s, and its kinematics were compared with those of a baseline liner that did not meet the 150 g criterion at 4.3 m/s. Each condition was repeated three times, and the impact angle for these tests was maintained at 104°.

2.3 Injury metrics

Universal Brain Injury Criteria (UBrIC) [13] was used to estimate the 95th percentile maximum principal strain (MPS) in the brain. UBrIC, as defined in Equation 1, computes MPS based on measurements of angular velocity and angular acceleration. It is formulated to be broadly applicable across a range of impact conditions, including automotive accidents and sports-related impacts.

$$UBrIC = \left\{\sum_i \left[\omega_i^* + (\alpha_i^* - \omega_i^*)e^{-\frac{\alpha_i^*}{\omega_i^*}}\right]^r\right\}^{\frac{1}{r}} \quad where\ i = x, y, and\ z \qquad (1)$$

Here, $\omega_i{}^*$ represents the difference between the maximum and minimum angular velocities (i.e., positive and negative peaks) normalized by a critical value ($\omega_i{}^* = \omega_i/\omega_{icr}$) along the x, y, and z axes. Similarly, $\alpha_i{}^*$ represents the peak angular acceleration normalized by a critical value ($\alpha_i{}^* = \alpha_i/\alpha_{icr}$) along the x, y, and z axes. The exponent r, used in the evaluation of UBrIC, was set to 2, as recommended by the authors [13]. Although UBrIC provides MPS for various rotational kinematics parameters, it does not yield a direct injury probability value. Instead, it informs helmet and countermeasure design by capturing a broader spectrum of brain injury mechanisms associated with high brain tissue strains.

2.4 Data processing and statistical analyses

Raw experimental signals are processed following the Society of Automotive Engineers (SAE) J211 standards using a fourth-order Butterworth low-pass filter. Linear acceleration signals were filtered using a channel frequency class (CFC) of 1000 (cut-off frequency: 1650 Hz), while angular velocity signals were filtered using CFC 180 (cut-off frequency: 300 Hz). Angular velocity data, recorded in degrees per second (°/s), were converted to radians per second (rad/s) before filtering and then differentiated to compute angular acceleration. Data processing was performed using a MATLAB script (MathWorks, MA, USA).

Statistical analyses were performed using Prism (GraphPad Software, MA, USA). Differences in parameters among the three bicycle helmets were evaluated using one-way analysis of variance (ANOVA) with Tukey's post-hoc test for pairwise comparisons. An unpaired Student's t-test was used to assess the mean differences in kinematic behavior between the high-density liner and the baseline liner. Statistical significance was set at $p < 0.05$, and all peak kinematic results are reported as mean ± one standard deviation.

3. REPEATABILITY ASSESSMENT

Table 1. Percentage differences in peak linear acceleration (PLA) and peak angular velocity (PAV) across four repeatability test setups. Results are presented as mean [standard deviation (SD)], with the percentage difference (% diff) calculated as the difference between the maximum and minimum values across five tests for each condition.

Impact angle (deg)	*PLA (g) mean (SD)*	*PLA (min/max % diff)*	*PAV (rad/s) mean (SD)*	*PAV (min/max % diff)*
51° (Front)	101 (1.5)	3.5%	26.0 (0.3)	2.6%
78° (Front)	223 (1.2)	1.5%	35.8 (0.1)	0.6%
40° (Rear)	86 (1.2)	3.6%	17.2 (0.2)	3.0%
61° (Rear)	212 (1.3)	1.5%	26.7 (3.8)	1.8%

The repeatability of the DREW system was evaluated through controlled drop tests using a bare Hybrid III headform (without a helmet) onto a CADEX modular elastomer programmer (MEP) pad (hardness: 60 ± 2 Shore A; thickness: 0.025 m; diameter: 0.152 m). Four impact conditions were tested: two front-facing impacts at impact angles of 51° (low) and 78° (high), and two rear-facing impacts at 40° (low) and 61° (high), as outlined in Table 1. Each condition was repeated five times. For front impacts, the 51° impact angle produced a mean linear acceleration of 101 g [standard deviation (SD) = 1.5] and a mean angular velocity of 26.0 rad/s (SD = 0.3), with 3.5% and 2.6% variability, respectively. At the higher 78° angle, linear acceleration increased to 223 g (SD = 1.2) with a 1.5% variation, and angular velocity reached 35.8 rad/s (SD = 0.1) with only 0.6% variation.

For rear-facing impacts, the 40° drop resulted in a mean linear acceleration of 86 g (SD = 1.2) and an angular velocity of 17.2 rad/s (SD = 0.2), with respective variability of 3.6% and 3.0%. At the higher 61° angle, linear acceleration rose to 212 g (SD = 1.3) with a 1.5% difference between maximum and minimum values, while angular velocity increased to 26.7 rad/s (SD = 3.8) with a 1.8% difference. These

https://doi.org/10.52202/081621-0046

results demonstrate the DREW system's good repeatability across a range of impact angles and directions, with less than 3.6% variation in linear acceleration and less than 3.0% in angular velocity.

4. RESULTS

4.1 Direct head impact vs. whiplash-induced impact

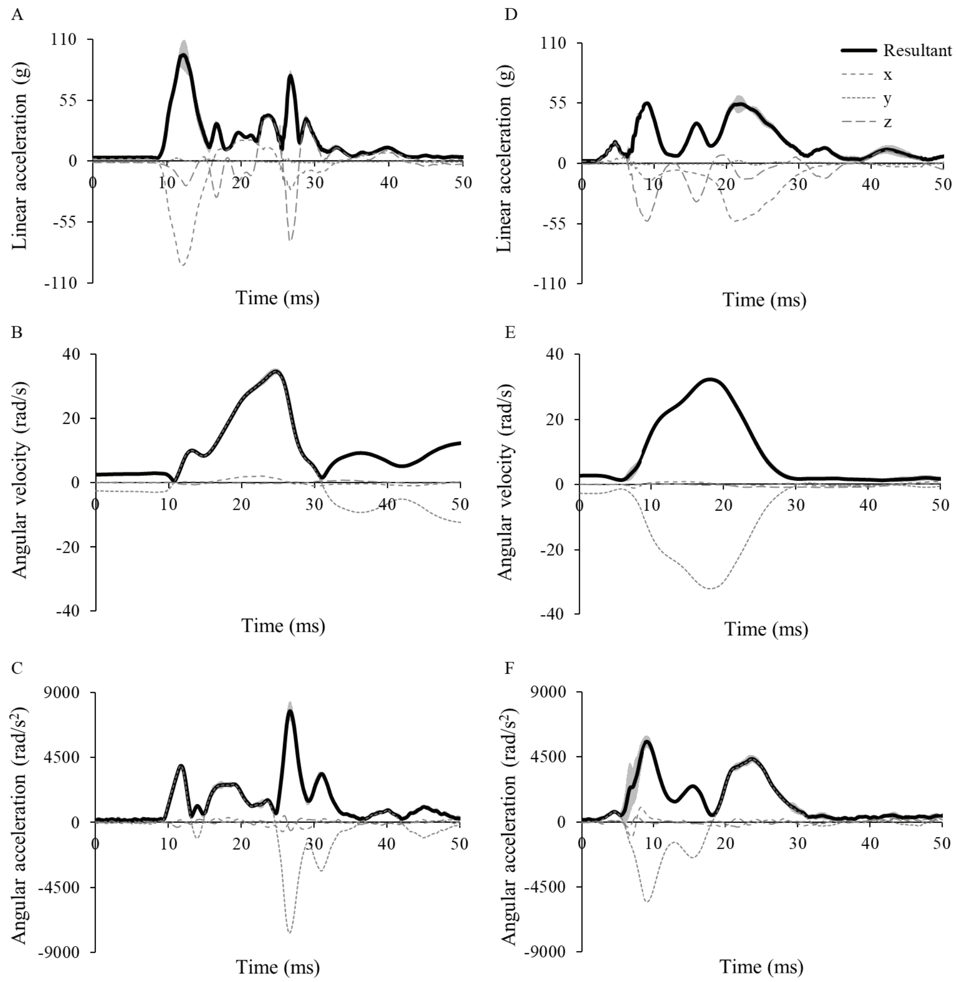

Figure 3. Mean traces of linear acceleration, angular velocity, and angular acceleration for head-first impact (Panels A, B, and C) and whiplash-induced impact (Panels D, E, and F). The solid line represents the mean resultant (n = 5), with the shaded area indicating the standard deviation. Mean component traces (along x, y, and z) are shown as dotted lines.

During direct impact, the linear acceleration peaked after the headform contacted the anvil, reaching a peak resultant value of 96 ± 13 g, with a secondary peak observed after 25 ms. Analysis of the component accelerations indicated that the first peak was due to acceleration along the x-axis, while the second peak arose from acceleration along the z-axis toward the pivot point. The angular velocity did not increase monotonically; instead, it exhibited a transient dip at approximately 12 ms, followed by a rise to a peak value of 34.6 ± 1.6 rad/s. Similar to linear acceleration, the angular acceleration displayed two distinct peaks, with the second peak (7727 ± 1079 rad/s²) exceeding the first.

During whiplash-induced impacts, the head's linear acceleration exhibited multiple peaks: the first occurred before 10 ms (55 ± 1 g), followed by a slightly reduced peak after 15 ms, and finally a third peak, with a magnitude comparable to the first (54 ± 6 g), appeared beyond 20 ms. Component analysis revealed that the first two peaks were caused by acceleration along the z-axis toward the pivot point,

https://doi.org/10.52202/081621-0046

while the third peak resulted from acceleration along the x-axis. Unlike direct impact, the angular velocity for whiplash-induced impacts increased monotonically, reaching a peak value of 32.1 ± 0.6 rad/s. Additionally, analysis of the acceleration traces revealed that the first peak (5566 ± 193 rad/s²) was higher than the second.

4.2 Bicycle helmet

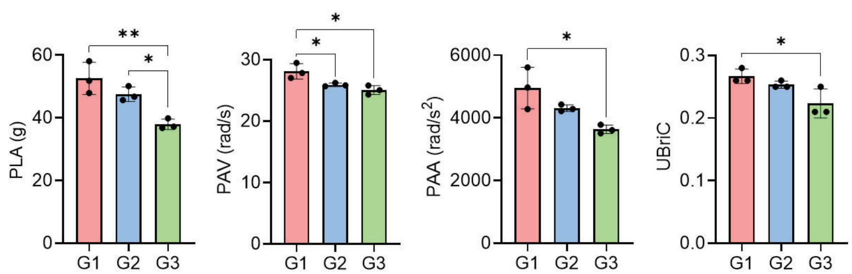

Figure 4. Kinematic evaluation of bicycle helmets under frontal head-first impacts: comparison of peak linear acceleration (PLA), peak angular velocity (PAV), peak angular acceleration (PAA), and Universal Brain Injury Criteria (UBrIC) in three helmet designs (G1: Baseline, G2: Multi-Directional Impact Protection System, G3: WaveCel). Data are presented as mean ± standard deviation (n = 3); *p ≤ 0.05, **p ≤ 0.01.

Using frontal impacts without the whiplash mode, we assessed the efficacy of rotational kinematics mitigation systems available in bicycle helmets. Three helmet conditions were tested: a baseline helmet (G1), a MIPs-equipped helmet (G2), and a WaveCel-equipped helmet (G3). The baseline helmet produced a mean peak linear acceleration (PLA) of 53 g (SD = 5), a mean peak angular velocity (PAV) of 28.1 rad/s (SD = 1.2), and a mean peak angular acceleration (PAA) of 4954 rad/s² (SD = 664). Compared to the baseline helmet, the WaveCel helmet demonstrated significant reductions across multiple kinematic metrics.

For PLA, the WaveCel helmet achieved a mean of 38 g (SD = 2), representing a significant 28% reduction relative to the baseline. Similarly, for PAV, the WaveCel helmet yielded a mean of 25.1 rad/s (SD = 0.7), corresponding to a 11% decrease from the baseline. For PAA, the WaveCel helmet recorded a mean of 3634 rad/s² (SD = 138), a 27% reduction relative to the baseline. Additionally, the UBrIC, which estimates brain's 95th percentile MPS, was significantly reduced by 16% compared to the baseline. In contrast, the MIPS-equipped helmet only demonstrated a significant reduction in PAV, with a mean of 25.9 rad/s (SD = 0.3), representing an 8% decrease relative to baseline. Notably, the WaveCel helmet outperformed the MIPs helmet in mitigating linear acceleration, however, there was no significant difference between MIPs and WaveCel among angular kinematics and UBrIC estimate.

4.3 Ballistic helmet liners

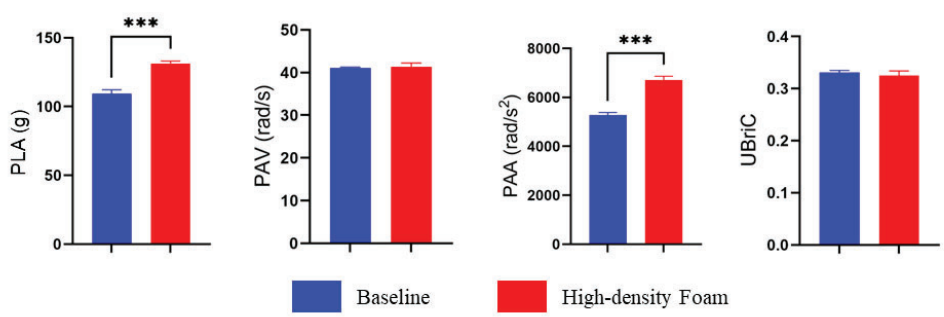

Figure 5. Kinematic evaluation of ballistic helmet liners under rear whiplash-induced impacts: comparison of peak linear acceleration (PLA), peak angular velocity (PAV), peak angular acceleration (PAA), and Universal Brain Injury Criteria (UBrIC) for two liners, Baseline (blue) and high-density liner (red). Data are presented as mean ± standard deviation (n = 3); ***p ≤ 0.001.

Whiplash-induced rear impact tests were conducted at a calculated impact velocity of 6.1 m/s to compare the kinematic performance of a high-density liner with a baseline liner. In contradiction to the standard

https://doi.org/10.52202/081621-0046

monorail test, the high-density liner produced a significantly higher PLA, with a mean value of 131 g (SD = 2) compared to 109 g (3 g) for the baseline (Figure 5). Similarly, PAA was significantly elevated with the high-density liner (6708 rad/s², SD = 153) relative to the baseline (5286 rad/s², SD = 95). In contrast, there was no significant difference in PAV and UBrIC between the high-density liner (41.4 rad/s, SD = 0.8) and baseline (41.04 rad/s, SD = 0.2). These results indicate that the high-density liner did not perform as intended when tested in a more realistic fall simulation.

5. DISCUSSIONS

Table 3. Comparison of impact testing capabilities: monorail drop tests (department of transportation [DOT] and hybrid III [H3] headforms), backfall system, and the device for rotational evaluation of wearables (DREW) Platform.

Capabilities	*Monorail with DOT head*	*Monorail with H3 head*	*Backfall*	*DREW*
Linear motion	O	O	O	O
Rotational motion	X	O	O	O
Various impact locations	O	O	X	O
Various impact speed	O	O	X	O
Whipping motion	X	X	O	O
Biofidelic dummy	X	X	O	O

We developed a novel test rig, DREW, to simulate head-to-ground impacts for helmet evaluation. DREW is capable of generating two distinct impact scenarios: (1) head-first impacts, where the head directly strikes an anvil, and (2) whiplash-induced impacts, where the torso of the ATD first contacts a table, inducing a whiplash motion in the head before it impacts the anvil. To assess the rig's capability in characterizing rotational mitigation technologies, we tested bicycle helmets equipped with MIPs and WaveCel systems and compared their performance to that of a baseline helmet. Finally, we evaluated a ballistic helmet liner, previously optimized for high-velocity performance at impact velocities of 4.3–5.2 m/s, against a baseline liner to determine whether its enhanced performance translates to improved protection under more realistic impact conditions simulated through DREW.

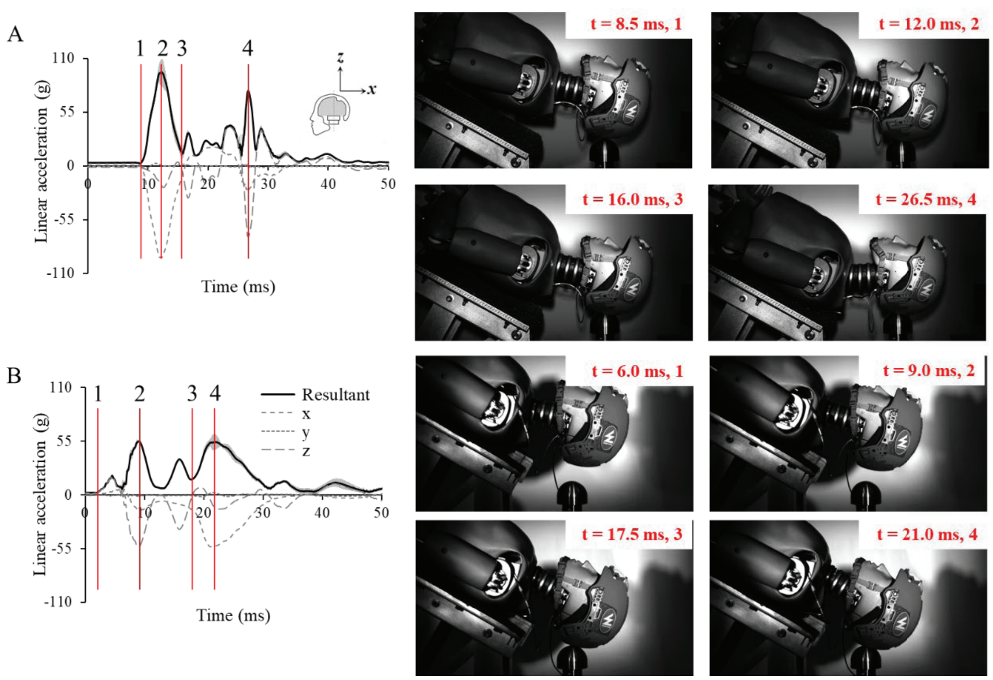

Figure 6. High-speed video frames captured at four key time points of drop and impact for head-first (A) and whiplash-induced (B) impacts are presented. The corresponding time points are numbered and marked (red line) on the linear acceleration trace to elucidate the impact dynamics.

Table 3 provides a comparison of the capabilities of conventional helmet evaluation techniques and the DREW system. Traditional methods, such as the monorail drop test with a DOT headform, reliably simulate linear motion and can accommodate various impact locations and speeds; however, they lack the ability to simulate rotational motion, whipping motion, and do not utilize a biofidelic dummy. In response, researchers have modified these methods by employing a H3 ATD head, which permits neck bending and better captures rotational dynamics, though this approach still falls short in replicating whipping motion, i.e., pre-impact kinematics. Backfall rigs that induce whiplash impacts provide an improvement by simulating the head's whipping motion; nevertheless, they are limited in that they cannot simulate impacts at multiple locations or various impact speeds. In contrast, the DREW system integrates all these critical capabilities, i.e., it can simulate and capture both the linear and rotational kinematics associated with head impacts, accommodates impacts from multiple locations at various speeds, simulates whipping motion, and employs a biofidelic dummy. This comprehensive functionality makes the DREW system a robust platform for evaluating helmet performance, particularly in the context of mitigating the rotational kinematics experienced in real-world impact scenarios.

High-speed video recordings were used to interpret the mechanics of both head-first and whiplash-induced impacts. Snapshots from the footage were synchronized with four key time points on the linear acceleration trace. For head-first impacts (Figure 6A), the video shows that the headform contacts the anvil at t = 8.5 ms, initiating an increase in acceleration that peaks at 12.0 ms. At t = 16.0 ms, the ATD torso impacts the table, resulting in a marked reduction in the head's acceleration. Component analysis indicates that the initial acceleration peak is primarily due to motion in the x-direction, corresponding to the head's impact. A secondary acceleration peak is observed at t = 26.5 ms, predominantly arising from acceleration in the z-direction. This z-direction acceleration is attributed to the centripetal acceleration generated as the torso rebounds toward the table, causing the head, while in contact with the anvil, to undergo angular motion, as evidenced by neck flexion at t = 26.5 ms. For whiplash-induced impacts (Figure 6B), the torso first contacts the table at t = 6.0 ms, inducing a whiplash motion in the head that reaches its peak acceleration at 9.0 ms. This whiplash effect generates significant angular motion, resulting in centripetal acceleration along the z-axis. Notably, oscillations in the z-axis acceleration are observed, likely reflecting dynamic changes in head motion as the torso rebounds off and back onto the table. Subsequently, the head contacts the anvil around t = 17.5 ms, with the x-direction acceleration peaking at t = 21.0 ms.

Analysis of the angular kinematics traces revealed distinct behaviors between head-first and whiplash-induced impacts. In head-first impacts, angular velocity increases non-monotonically, with a slight dip observed around t = 12.0 ms, likely due to the torso impacting the table, before rising to a peak and then decreasing. Notably, the second peak in angular acceleration exceeds the first, indicating that the rate of velocity reduction is greater than the initial rate of increase. During whiplash-induced impacts, the headform exhibits a pre-impact angular velocity, with the peak occurring before the head contacts the anvil (approximately 18 ms). Similar observations have been reported in mouthguard data from an American football player [11]. Moreover, the angular acceleration during the whiplash phase is higher than that observed at impact, likely due to helmet offering protection during impact. These findings underscore the importance of simulating pre-impact, rather than focusing solely on post-impact kinematics. Because standard monorail drop tests fail to capture these dynamics, alternative testing methods such as the DREW system may provide a more comprehensive assessment, thereby informing strategies to better prevent mTBI.

Assessment of rotational kinematics mitigation technology among bicycle helmets in non-whiplash events revealed differences in linear and rotational kinematics among different helmet designs, with the WaveCel helmet consistently exhibiting the lowest values across all metrics. Statistical analyses revealed that the PLA was significantly lower for the WaveCel helmet compared to both the baseline and MIPs helmets. For PAV, significant differences were observed between the baseline and both the MIPs and WaveCel conditions. In terms of PAA and UBriC, a statistically significant reduction was found only between the baseline and the WaveCel helmet. These findings are consistent with previous research by Bliven et al. and Abayazid et al., which reported substantial decreases in both rotational acceleration and velocity, as well as improvements in injury metrics such as the brain injury criterion [10,14]. Overall, these results underscore the potential benefits of optimizing helmets to improve protection against the rotational kinematics associated with traumatic brain injury.

Finally, the high-density liner, optimized via conventional monorail drop tests using a DOT headform, previously demonstrated promising performance by reducing PLA by 15% and 57% at impact velocities of 4.3 and 5.2 m/s, respectively, compared to the baseline liner. However, when evaluated using the DREW system with an H3 ATD, although both liners maintained the PLA below 150 g, the high-density liner produced a statistically significant increase in PLA of up to 20% relative to the baseline. In addition, while standard testing protocols do not account for rotational kinematics, emerging

https://doi.org/10.52202/081621-0046

research highlight their importance in mTBI. In our study, the high-density liner increased PAA by 27% compared to the baseline, with this difference reaching statistical significance. No significant difference was observed in PAV and UBrIC, likely because whiplash-induced impacts generate peak PAV during the whiplash phase before the liner is engaged. These results suggest that the current DOT-based monorail drop tests, with their reliance on rigid neck configurations, may favor stiffer foam designs that do not necessarily translate to improved performance under more realistic impact conditions.

One limitation of our study is that the H3 ATD system is not specifically designed for simulating backward falls. While it is traditionally used in automotive impact testing, backward falls may involve more complex kinematic patterns than those for which the surrogate was originally intended. Nonetheless, our primary objective was to assess the DREW system's ability to differentiate helmet performance in mitigating rotational kinematics. The Hybrid III provided a repeatable and standardized platform for this comparative evaluation.

6. CONCLUSION

In conclusion, the DREW test rig provides a robust and comprehensive platform for evaluating helmet performance under realistic impact conditions. DREW simulates both head-first and whiplash-induced impacts at various locations on the helmet, thereby capturing both translational and rotational dynamics. The system exhibited high repeatability, with variability in kinematic measurements remaining below 4%. Using the DREW system, our analysis of a selection of bicycle helmets revealed that those incorporating rotational mitigation technology offer improved performance in reducing rotational kinematics compared to the baseline. Furthermore, we assessed a high-density liner, developed and optimized through DOT-based monorail drop tests, against a baseline liner. In monorail tests, the high-density liner demonstrated superior performance relative to the baseline at impact velocities of 4.3 and 5.2 m/s, achieving the required standard of maintaining PLA below 150 g during. However, when tested using the DREW system at a higher impact velocity, the baseline liner achieved a significantly lower peak linear acceleration than the high-density liner. Additionally, our evaluation of rotational kinematics revealed that the baseline liner exhibited considerably lower peak angular acceleration compared to the high-density liner.

These findings indicate that a realistic simulation of impact dynamics is essential for helmet evaluation. Achieving optimal linear performance alone may not be sufficient to reduce the risk of brain injury, as rotational kinematics also play a crucial role. Therefore, incorporating the evaluation of rotational kinematics into helmet testing protocols is critical. Our results underscore that assessment methods accounting for both linear and rotational dynamics offer a comprehensive evaluation of helmet performance and are vital for developing next-generation helmets that deliver superior head protection in real-world scenarios.

Acknowledgments

This research was supported by the U.S. Office of Naval Research under PANTHER award number N00014-21-1-2916 through Dr. Timothy Bentley.

Disclaimer

References to trade names are provided solely for clarity and do not constitute an endorsement of any commercial products. It is important to note that the bicycle helmet testing described herein was conducted under specific conditions and impact velocities optimized for evaluating ballistic helmet performance; these results are not intended to serve as a direct comparison or commentary on the performance of different helmet technologies.

References

[1] Wojcik B. E., Stein C. R., Bagg K., Humphrey R. J., and Orosco J., "Traumatic brain injury hospitalizations of US army soldiers deployed to Afghanistan and Iraq." *Am J Prev Med*, 38, no. 1 (2010): S108-S116.

[2] U.S. Military Health System. DOD Numbers for Traumatic Brain Injury Worldwide 2000 – 2024 Q2. (2024). https://health.mil/Reference-Center/Reports/2024/12/30/2000-2024-Q2-DOD-Worldwide-Numbers-for-TBI. Accessed 13 March 2025.

[3] Tierney G., "Concussion biomechanics, head acceleration exposure and brain injury criteria in sport: a review." *Sports Biomech*, 23, no. 11 (2024): 1888-1916.
[4] Bian K., and Mao H., "Mechanisms and variances of rotation-induced brain injury: a parametric investigation between head kinematics and brain strain." *Biomech Model Mechanobiol*, 19, no. 6 (2020): 2323-2341.
[5] Holbourn, A. H. S., "Mechanics of head injuries." *The Lancet* 242, no. 6267 (1943): 438-441.
[6] Kimpara H., and Iwamoto M., "Mild traumatic brain injury predictors based on angular accelerations during impacts." *Ann Biomed Eng*, 40 (2012): 114-126.
[7] Rowson S., Duma S.M., Beckwith J.G., Chu J.J., Greenwald R.M., Crisco J.J., Brolinson P.G., Duhaime A.C., McAllister T.W. and Maerlender A.C., "Rotational head kinematics in football impacts: an injury risk function for concussion." *Ann Biomed Eng*, 40 (2012): 1-13.
[8] Kleiven S., "Why most traumatic brain injuries are not caused by linear acceleration but skull fractures are." *Front bioeng biotechnol*, 1 (2013): 15.
[9] Fernandes, F.A. and De Sousa, R.A., "Finite element analysis of helmeted oblique impacts and head injury evaluation with a commercial road helmet." *Struct Eng Mech*, 48, no. 5 (2013): 661-679.
[10] Bliven E., Rouhier A., Tsai S., Willinger R., Bourdet N., Deck C., Madey S.M. and Bottlang M., "Evaluation of a novel bicycle helmet concept in oblique impact testing." *Accid Anal Prev*, 124 (2019): 58-65.
[11] Neice R.J., Lurski A.J., Bartsch A.J., Plaisted T.A., Lowry D.S. and Wetzel E.D., "An experimental platform generating simulated blunt impacts to the head due to rearward falls." *Ann Biomed Eng,* 49 (2021): 2886-2900.
[12] Halldin P., Gilchrist A., Mills N. J., "A new oblique impact test for motorcycle helmets." *Int J Crashworthiness* 6, no. 1 (2001): 53-64.
[13] Gabler L.F., Crandall J.R. and Panzer M.B., "Development of a metric for predicting brain strain responses using head kinematics". *Ann Biomed Eng*, 46 (2018), 972-985
[14] Abayazid F., Ding, K., Zimmerman K., Stigson H., & Ghajari M. (2021). "A new assessment of bicycle helmets: the brain injury mitigation effects of new technologies in oblique impacts." *Ann Biomed Eng*, 49, no. 10 (2021): 2716-2733.

 https://doi.org/10.52202/081621-0046