ROCK CLIMBING HELMET IMPACT PERFORMANCE VARIES BY HELMET MODEL TYPE

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The purpose of this study was to compare the impact performance of a sample of rock climbing helmets. Each helmet was impacted at three locations (front, side, and back) and three impact speeds (2 m/s, 3.85 m/s, and 5 m/s) using a custom pendulum impactor. Peak resultant linear and rotational accelerations were compared by impact speed. The results of this study show that acceleration varies by helmet model and impact location. Differences in acceleration by helmet model support the need for relative performance ratings of rock climbing helmets. These tests can be used as the basis of methods to compare relative helmet performance. Relative performance evaluation would inform consumers of the safety of different helmet types, and would also inform manufacturers where improvements in helmet designs should be made.

KEYWORDS: concussion, skull fracture, head injury, risk, linear, rotational, acceleration.

INTRODUCTION: Rock climbing is a recreational sport that has seen a sharp rise in popularity in the U.S., with an increase of over 4.5 million Americans participating in indoor and outdoor climbing during 2007-2015 (The Outdoor Foundation, 2016). Despite the popularity of rock climbing, there is limited published research on the head protection used in this sport even though climbers may experience a wide range of injuries. Although injuries to extremities are more common, injuries to the head tend to be more severe or even fatal (Yamaguchi et al., 2014). Rock climbing helmets typically conform to two performance standards: UIAA 106 (International Climbing and Mountaineering Federation [UIAA], 2018) and EN 12492 (European Committee for Standardization [CEN], 2012). Both standards require a 5-kg striker to be dropped onto a rock climbing helmet that is mounted on a headform, which is attached to a load cell that measures neck force. The striker is dropped from a 2-m height onto the top of the helmet, or a 0.5-m height onto the front, side, and back of the helmet with the headform angled at 60° off-axis. Force must be below 8000 N for the UIAA standard and the 10,000 N for the EN standard for all impact locations. These standards were written with alpine climbing in mind where rock fall on the upper portion of the helmet is the biggest concern (MacNae & Taylor, 2000; Nelson & McKenzie, 2009; Rugg et al., 2020). However, falls are more likely to cause a head injury in other types of climbing like sport or trad climbing, which can result in an impact at any location on the head (Yamaguchi et al., 2014). Helmets generally have more padding at the top of the head in order to pass the standard, which could result in less protection at other locations impacted during a fall. All climbing helmets must pass current impact standards which protect the head from fatal injuries (Bowie et al., 1988; Schussman et al., 1990), but differences in impact performance have previously been identified for different helmet types (MacNae & Taylor, 2000; Yamaguchi et al., 2014). While these studies showed differences in impact performance, they had limited practical use for comparing helmet types. The purpose of this study is to investigate the relative impact performance of representative rock climbing helmet models using a repeatable methodology that measures linear and rotational head kinematics. It was hypothesized that head acceleration would differ between helmet models for impacts to the front, side, and back of the helmet where there is greater variability in helmet designs.

METHODS: Four different rock climbing helmet models with two samples per model were subjected to an identical impact protocol with a custom pendulum impactor (Figure 1) (Rowson et al., 2015). Helmet models tested were the Black Diamond Half Dome (BDHD), Beal Mercury (BM), Edelweiss Vertige (EV), and Petzl Boreo (PB). The pendulum arm consists of rectangular aluminum tubing with a 16.3 kg impacting mass at the end. The total length of the pendulum arm from the center of the pivot point to the center of the impacting mass is 190.5 cm, with a total mass of 36.3 kg. A flat nylon impactor face with a diameter of 12.7 cm was rigidly attached to the impacting mass. The impactor on the pendulum struck a helmeted medium NOCSAE (National Operating Committee on Standards for Athletic Equipment) headform instrumented with three linear accelerometers and three angular rate sensors to measure six-degree-of-freedom head kinematics (DTS, Seal Beach, CA). Each helmet was impacted at three locations (front, side, and back) and three speeds (2 m/s, 3.85 m/s, and 5 m/s). Pre-measured angles were marked in each plane of headform rotation, as well as the point of contact with the pendulum to ensure consistent impact locations. Impact speeds were selected to cover a wide range of head accelerations intended to test the lower and upper limits of helmet performance. With two samples of each helmet model, a total of 72 impact tests were performed.

Linear acceleration and angular rate data were sampled at a frequency of 20 kHz. Linear acceleration data were filtered at CFC 1000 and angular rate data were filtered at CFC 155. Rotational acceleration was calculated by differentiating angular rate data. To determine if impact performance differed between helmet models, two-factor ANOVAs were performed with helmet type and impact location as categorical factors. ANOVAs were run independently for each impact speed and for linear and rotational acceleration, for a total of six comparisons. Probabilities less than 0.05 were considered significant. Post hoc Tukey's HSD tests were performed to determine which helmets were significantly different at each speed.

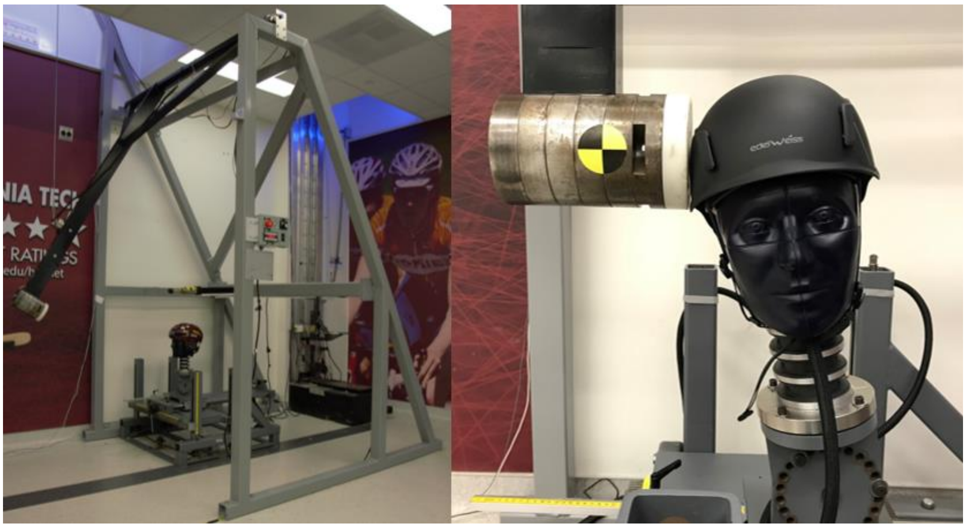

Figure 1: Photos of pendulum impactor (left) and side impact configuration with a representative rock climbing helmet model (right).

RESULTS: Both helmet model and impact location had significant main effects for all impact speeds for both linear and rotational accelerations. There was also a significant interaction between helmet model and impact location for all comparisons except rotational speed at 5 m/s. For linear acceleration resulting from 2-m/s impacts, EV (26 *g* mean acceleration) was significantly lower than BDHD (41 *g*) and PB (37 *g*). At 3.85 m/s, all helmet models had significantly different linear acceleration, with BM (201 *g*) at the highest and PB (84 *g*) at the lowest. At 5 m/s, BDHD (225 *g*) and PB (162 *g*) both had lower linear accelerations than BM (346 *g*) and EV (334 *g*). For impacts at 2 m/s, rotational acceleration of BDHD (2900 rad/s^2) was significantly higher than all other helmet models (1717-2093 rad/s^2). At 3.85 m/s, BM, BDHD, and PB all had significantly different rotational accelerations, with BM (8145 rad/s^2) at the highest and PB (4060 rad/s^2) at the lowest. At 5 m/s, PB (7587 rad/s^2) had significantly lower rotational acceleration than all other helmet models (12802-14208 rad/s^2). For linear acceleration, helmet model generally accounted for most of the variation (Figure 2, left). For rotational acceleration, impact location accounted for more variance, with significantly higher acceleration at the side for all impact speeds (Figure 2, right). Rotational acceleration is more variable by impact location because the distance between the point of impact and head center of gravity changes by location.

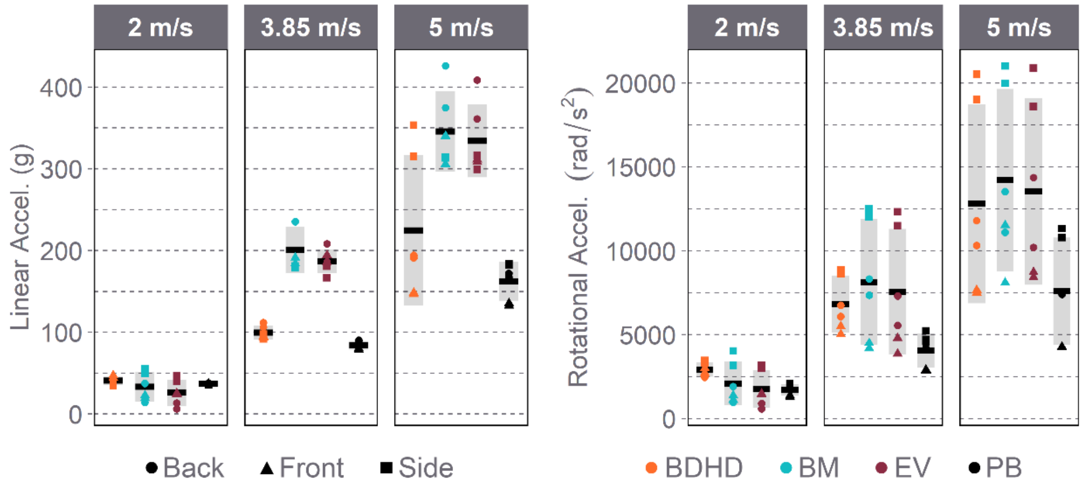

Figure 2: Peak linear acceleration (left), and peak rotational acceleration (right) by helmet type and impact location for each impact speed. Each point represents a single impact test. There are six tests (three locations and two helmet samples) per helmet model at each speed. Means (horizontal lines) and 95% confidence intervals (shaded areas) are shown for each helmet model at each speed.

DISCUSSION: This study showed that different rock climbing helmet models have different protective capabilities for identical impact conditions. While previous studies have shown differences in impact performance (MacNae & Taylor, 2000; Yamaguchi et al., 2014), this study adds to these findings by reporting differences in linear and rotational head kinematics and presenting a highly repeatable and reproducible method to compare helmet performance. Linear and rotational head acceleration are correlated with different head injury mechanisms, so injury risk can be estimated from impact tests and compared between helmet types (Mertz et al., 1996; Rowson & Duma, 2013). Repeatable and reproducible test methods are especially important for comparing relative impact performance of helmets so that true differences can be identified.

The authors chose to compare impact performance at the front, side, and back of the helmet where design is generally not as robust as the top, and may have more differences between helmet models. Differences were apparent across all impact speeds, but it is important to account for effect size. For linear acceleration, the mean difference between the best and worst performing helmets was 15 *g* for a 2-m/s impact, 117 *g* for a 3.85-m/s impact, and 183 *g* for a 5-m/s impact. While significant differences between helmet model types were identified at all impact speeds, the 5 m/s difference is more meaningful in terms of head injury risk. To put these numbers in perspective, all helmets have a 0% risk of skull fracture at 2 m/s, range from 0% to 10% risk at 3.85 m/s, and range from 2% to 96% at 5 m/s (Figure 3, left) (Mertz et al., 2003; Mertz et al., 1996). In terms of concussion risk, all helmets have a 0% risk at 2 m/s, range from 3% to 98% at 3.85 m/s, and range from 91% to 100% at 5 m/s (Figure 3, right) (Rowson & Duma, 2013). The significant interactions between helmet type and impact location indicate that helmet designs vary in protection depending on the impact location.

This study had several limitations. First, a limited number of helmet samples were available for use, so each helmet sample was tested at three impact speeds. Although visible damage was not apparent until after the 5-m/s impact speed, compression of the foam from previous impacts could influence head accelerations at higher impact speeds. Second, data on impact locations and speeds most commonly experienced by rock climbers are not currently available, so the impact speeds were selected to test the limits of helmet performance rather than real-world conditions.

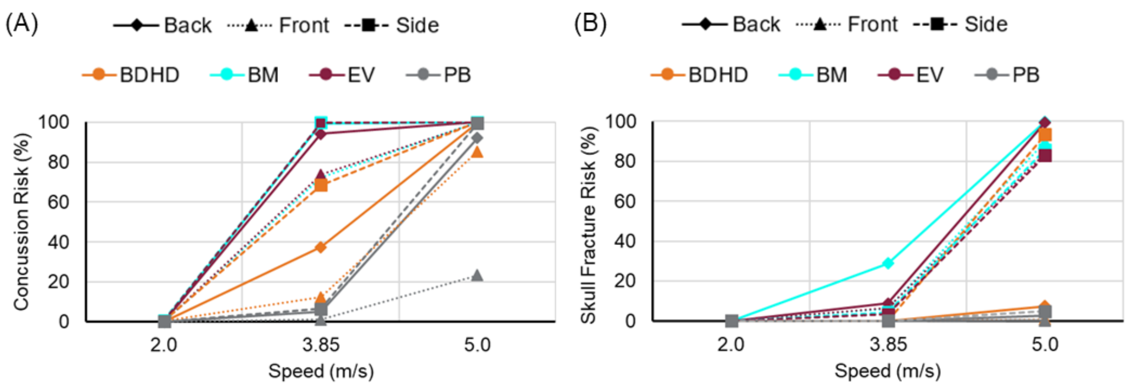

Figure 3: Comparison of concussion risk (Rowson & Duma, 2013) (left) and skull fracture risk (Mertz et al., 2003; Mertz et al., 1996) (right) by helmet model and impact speed. The most notable differences are seen at 3.85 m/s for concussion risk, and 5 m/s for skull fracture risk.

CONCLUSION: The differences in head accelerations and injury risks found in this study represent opportunities for improvements in helmet design to better protect climbers in the event of a fall or other collision. These preliminary tests can be used to inform development of methods for relative impact performance evaluation of rock climbing helmets. Quantifying helmet performance would provide consumers with information on the relative protection provided by different helmet models, and provide manufacturers with data to improve helmet designs.

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