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# A Novel Strategy for Mitigation of Oblique Impacts in Bicycle Helmets

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**Abstract**

A principal cause of traumatic brain injury is rotational head acceleration, which can induce brain injury even in absence of a direct impact to the head. A bicycle fall typically leads to an oblique impact of the head that induces rotational head acceleration. To mitigate this rotational head acceleration, a novel bicycle helmet concept has been developed that employs a collapsible cellular structure. This study quantified the efficacy of this technology in comparison to traditional bicycle helmets made of rigid Expanded Polystyrene (EPS). Prototype helmets with the Cellular structure (CELL) and standard EPS helmets (CONTROL) were subjected to oblique impacts in vertical drop tests onto angled anvils. Helmets were tested at impact speeds of 4.8 m/s and 6.2 m/s and at impact angles of 30°, 45°, and 60°. Linear and rotational headform acceleration and neck loads of an anthropometric head-neck surrogate were recorded and peak axonal strain was estimated from headform kinematics. CELL helmets significantly reduced rotational acceleration and associated axonal strain in all tests compared to CONTROL helmets, with reductions ranging from 34%-73% for rotational acceleration and 63%-85% for axonal strain. Results demonstrate the potential of the novel bicycle helmet strategy to reduce rotational head acceleration and axonal strain associated with brain injury risk.

**Keywords:** Bicycle helmet; Brain injury; Concussion; Oblique impact; Impact testing; Rotational acceleration

## Introduction

In the United States, the number of bicycle commuters increased by 61% between 2000 and 2012 [1] and an estimated 33 million children ride bicycles [2]. While bicycling provides clear health benefits, [3,4] it is also the leading cause of sport-related head injuries treated in U.S emergency rooms [5]. In 2013, bicycle crashes in the US caused over 600,000 emergency department visits, over 30,000 hospitalizations, and over 800 fatalities [2]. The associated direct medical treatment cost exceeded $2 billion, not including the far greater costs due to work loss and quality-of-life loss [6].

Injury to the head is of particular concern for bicyclists. Head injury from bicycle accidents caused 80,000 emergency department visits in 2015, with 13,000 of these visits including diagnosis of concussion and Traumatic Brain Injury (TBI) [7]. TBI is frequently referred to as the "silent epidemic", because deficiencies in thinking, sensation, language, or emotion typically manifest in a delayed manner [8]. Since medical interventions for treatment of concussions and TBI are limited, brain injury prevention becomes of utmost importance.

Bicycle helmets are the primary and most effective strategy to prevent TBI [9]. Traditional bicycle helmets employ a rigid shell of Expanded Polystyrene foam (EPS) that dampens the impact, reduces the impact force, and in turn reduces head accelerations. These helmets are highly effective in reducing the risk of skull fracture, penetrating injury, and severe brain injury [10-12]. Specifically, traditional helmets have been optimized to reduce linear acceleration of the head, as outlined in the in the mandatory impact test standard by the US Consumer Product Safety Commission (CPSC). This CPSC standard requires that a radial impact from a vertical drop of a helmeted headform onto a horizontal anvil results in less than 300 g linear acceleration of a test headform [13]. However, in contrast to radial CPSC impacts, real-world impacts typically occur at impact angles of 30°-60° degrees [14-16]. Such oblique impacts induce both radial and tangential forces to the head, leading to both linear and rotational head acceleration [12,17].

A large body of research has shown that rotational head acceleration can readily cause concussions by subjecting brain tissues to shear forces that induce diffuse axonal injury [18-25]. In fact, TBI can readily be induced by rotational head acceleration in absence of a direct impact to the head, such as in whip-lash injuries.

Since axonal shear strain caused by rotational head acceleration is a predominant mechanism of brain injury [26] advanced helmet designs should specifically target mitigation of rotational acceleration, and should be tested in real-world oblique impacts [27]. Recently, several bicycle helmet manufacturers have introduced novel bicycle helmet designs with dedicated mechanisms for mitigation of rotational head acceleration in order to provide improved protection from brain injury [11,28-30]. The most widely adopted strategy comprises a slip liner inside the helmet that seeks to reduce rotational acceleration of the head by permitting sliding between the helmet and head during impact. However, there is a general lack of research data on the performance of slip liners, and a recent study that tested two helmets with slip liners and eight traditional helmets did not find that slip liner technology provided superior mitigation of rotational head acceleration compared to standard EPS helmets [29]. As an alternative strategy for mitigation of rotational acceleration, the present research describes a rotational suspension system comprised of a collapsible cellular structure that is recessed within the helmet. This rotational suspension system represents an extension of earlier research by Hansen et al. on an angular impact mitigation system [30].

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This helmet impact study quantified impact mitigation provided by prototype helmets with the rotational suspension system in direct comparison to standard EPS bicycle helmets over a realistic range of oblique impact scenarios. Results were used to test the hypothesis that prototype helmets can significantly reduce rotational head acceleration and brain injury risk, without negatively affecting linear head acceleration or neck loading compared to standard EPS helmets.

## Methods

Standard EPS helmets (CONTROL group) and helmets with a collapsible cellular structure (CELL group) were subjected to oblique impacts in guided vertical drop tests onto an angled anvil. Helmet performance was evaluated at 4.8 m/s impact velocity at impact angles of 30°, 45°, and 60° from the horizontal plane. In addition, helmet performance was tested at a higher velocity of 6.2 m/s at a 45° impact angle to evaluate the effects of low and high impact velocities at a given impact angle. Five helmets from each group were tested at each drop condition, requiring a total of 40 helmets. Headform kinematics and neck loads were acquired for each impact using an instrumented head and neck surrogate. The risk of brain injury for each drop condition and helmet type was estimated by explicit calculation of injury risk criteria based on headform kinematic histories. Furthermore, headform kinematic data were implemented into a validated finite element model of the head and brain to determine the probability of brain injury based on axonal strain.

### Helmets

For the Control group, 20 standard bicycle helmets (Scott ARX, www.scott-sports.com) were tested. These midrange helmets had an in-molded polycarbonate micro-shell and a standard Expanded Polystyrene (EPS) liner (Figure 1A). For the CELL group, 20 additional Scott ARX helmets were obtained and modified to implement a collapsible cellular structure (WAVECEL™, Milwaukee, OR) without affecting the overall thickness of the helmet (Figure 1B). A 15 mm thick portion of the EPS material was removed by a programmable milling machine from the inside of the helmet, while leaving intact approximately 10 mm of the original outer EPS shell. The 15 mm thick cellular structure was placed inside the machined recess to restore the original helmet thickness. This cellular liner has a specifically designed cell structure to provide distinct mechanisms for absorption of radial and tangential impact forces (Figure 1C). For radial impact forces, each cell has a transverse crease to initiate organized cell buckling. For oblique impact forces, cells can fold in shear direction and the structure can elastically deform in-plane to serve as a rotational suspension between the head and the

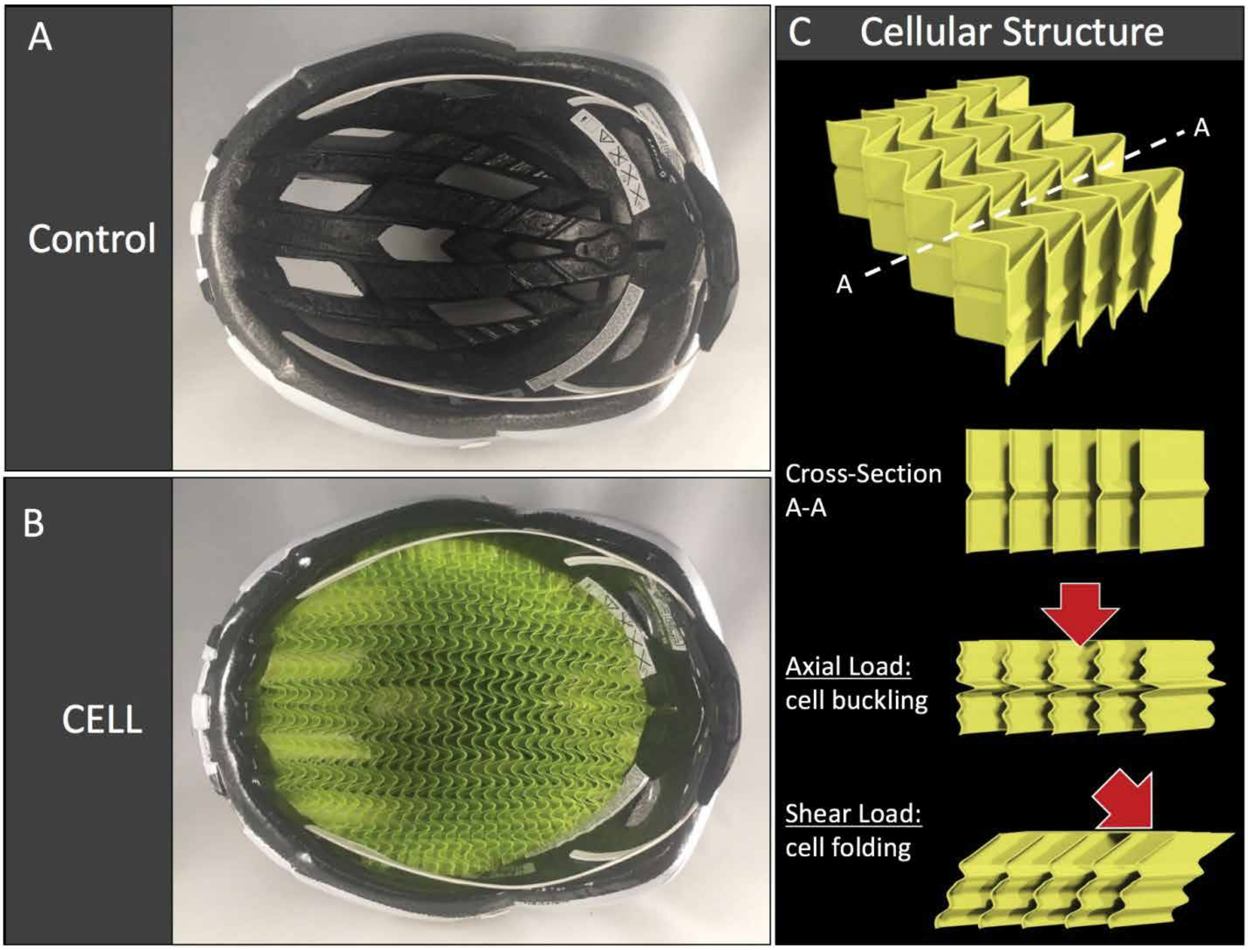


**Figure 1:** Two helmet types with identical outer shell and liner thickness were tested: A) Standard EPS helmets (Control); B) Helmets with a cellular structure for mitigation of linear and rotational acceleration (Cell); and C) The cellular structure has specifically designed cell geometry to provide distinct mechanisms for absorption of axial and shear loading.

outer helmet shell. All helmets had the same retention system, outer shell, and overall liner thickness.

## Test setup

Helmet testing was conducted at the Helmet Impact Testing (HIT) facility of the Portland Biomechanics Laboratory (Figure 2A). In absence of an accepted standard for oblique impact testing of bicycle helmets, the HIT facility was designed to follow recommendations of a recent consensus paper on advanced methods for oblique impact testing [17] and closely corresponds to several published methods of vertical drops onto oblique anvils [29-31]. Specifically, the HIT facility employed a Hybrid III 50$^{th}$ percentile male anthropomorphic head [32] and neck surrogate [17,33] (78051-336, Humanetic Innovative Solutions, Plymouth, MI) that was connected to a vertical drop tower rail (Figure 2B). A flat anvil adjustable from 30° to 60° [14-16] was used to induce oblique impacts in response to vertical drops (Figure 2B). Linear head acceleration was captured with a three-axis linear accelerometer (356B21 ICP Triaxial, PCB Piezotronics, Depew, NY) mounted at the center of gravity of the Hybrid III head. The resultant linear acceleration $a_r$ was calculated from the three linear acceleration components. Rotational acceleration $\alpha_y$ and rotational velocity $\omega_y$ of the headform around the transverse y-axis were measured with a rotational accelerometer (#8838, Kistler Instruments Corp., Amherst, NY). Assessment of headform rotation was limited to rotation around the transverse y-axis, since all impacts were centered onto the sagittal midline of the helmet, and since the anvil surface was aligned parallel to the headform transverse axis [30]. Neck loading was measured with a 3-axis load cell (IF-203, Humanetic Innovative Solutions, Plymouth, MI) at the base of the Hybrid III neck to capture neck compression ($F_C$), neck shear ($F_S$), and the neck flexion moment ($M_{Flex}$). Impact velocity was measured with a time gate (#5012 Velocimeter, Cadex Inc., Quebec, CA).

Five helmets of each group were tested at 4.8 m/s impact speed on 30°, 45°, and 60° inclined anvils, and additionally at 6.2 m/s on the 45° anvil. The impact speeds represent those specified in the bicycle helmet safety standard §1203 of the US Consumer Product Safety Commission (CPSC) [13]. Helmets were properly fitted to the headform with their original fit system. Consistent with prior studies that utilized the Hybrid III headform in helmeted drop tests, a nylon stocking was fitted over the silicone skin surrogate of the Hybrid III headform to better represent the surface friction of the human head [8,34]. All drop tests were performed on a frontal impact location in the mid-sagittal plane to induce headform rotational acceleration around a transverse axis. Before each test, new 80 grit sandpaper was applied to the anvil surface [35].

## Data acquisition and analysis

Accelerometer and neck loading data were simultaneously captured at a sample rate of 20 kHz in a data acquisition system (PCI-6221, National Instruments, Austin, TX). Accelerations and forces were low-pass filtered at Channel Frequency Class (CFC) 1000, and neck moments were low-pass filtered at CFC 600, as specified by SAE J211 [36]. Rotational velocity $\omega_y$ was calculated in LabVIEW software using trapezoidal integration of rotational acceleration data.

To estimate the probability of brain injury, injury probability was predicted by implementing headform kinematic histories into the Strasbourg University Finite Element Head Model (SUFEHM) [17,37,38]. SUFEHM is a validated computational model that has been

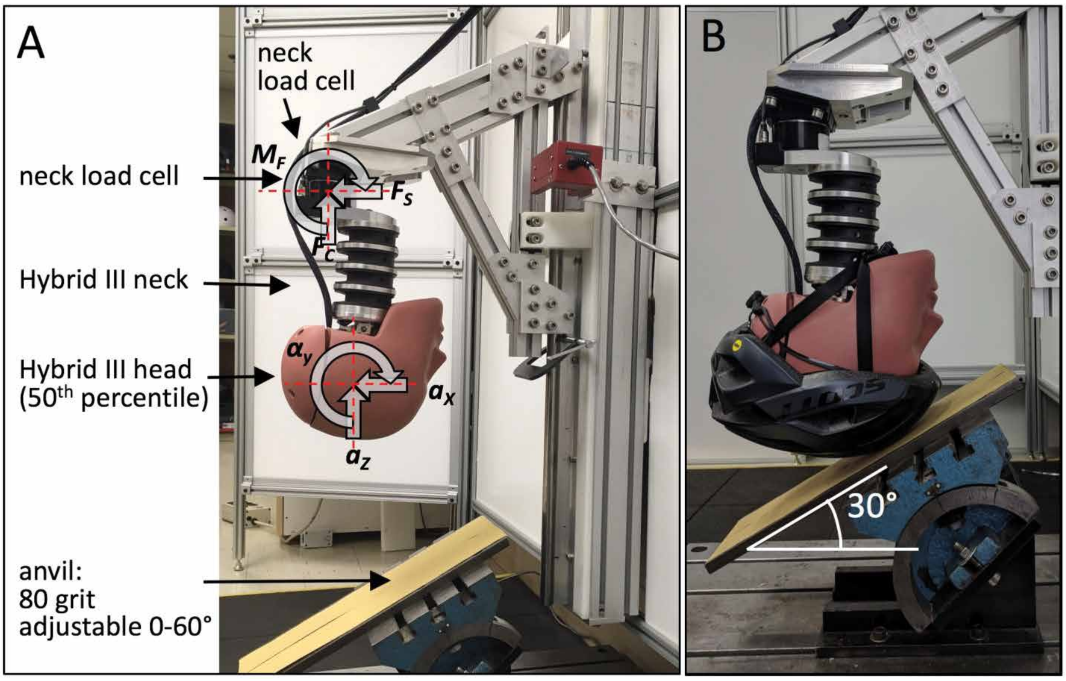


**Figure 2:** A) The Helmet Impact Testing (HIT) facility for vertical drop of a Hybrid III head and neck assembly onto a 0°-60° adjustable anvil to simulate oblique impacts. The drop assembly with integrated neck load cell and linear and rotational headform accelerometers captures neck loading (FC=compression, FS=shear, MF=flexion moment) and headform kinematics in terms of linear acceleration: (a) and rotational acceleration (α); B) Drop test shown for impact on the 30° anvil, corresponding to impact angles between the head trajectory and impact surface of 60°.

previously used in similar studies to quantify the damaging effects of impacts to the brain [30,39,40]. The model represents a 50th percentile adult human head, including the skull, brain, cerebrospinal fluid, falx, tentorium, and the main axonal bundles. It has been extensively validated by Sahoo et al. [38] against local brain motion data [41] and intercranial pressure data [42]. An injury risk curve for SUFEHM has been established by reconstruction of 109 real-world head trauma cases, and by correlating axonal strain results with the risk of sustaining moderate Diffuse Axonal Injury (mDAI) [25]. For the present study, average headform acceleration histories of the five tests conducted for each helmet type and impact condition were calculated. These average acceleration histories were implemented into SUFHEM to determine the peak axonal stain and the resulting injury probability for moderate DAI, as defined by the SUFHEM model [37,38].

For statistical analysis, headform kinematics ($a_r$, $\alpha_y$, $\omega_y$), neck loading ($F_C$, $F_S$, $M_F$) results of the CELL group were individually compared to the CONTROL group results using two-sided Student's t-tests. To test the stated hypothesis, a value of $\alpha=0.05$ was used for the evaluation of statistical significance.

## Results

The average impact speed for slow impacts did not differ significantly between the 20 CONTROL helmet impacts (4.80 ± 0.02 m/s) and the 20 CELL helmets impacts (4.80 ± 0.03 m/s, p=0.77). Similarly, the average speeds for fast impacts with CONTROL helmets (6.20 ± 0.02 m/s) was not significantly different from that of CELL helmets (6.17 ± 0.04, p=0.13).

The absolute magnitudes of all outcome parameters for each impact condition are summarized in Table 1. The following section describes relative differences in performance between CONTROL and CELL helmets by means of statistical comparisons.

### Headform kinematics

Peak linear acceleration $a_r$ was significantly lower in CELL helmets compared to CONTROL helmets for all slow impacts (4.8 m/s), with reductions ranging from 16% on the 60° anvil to 26% on the 30° anvil (Figure 3A). For fast impacts (6.2 m/s), there was no significant difference in linear acceleration between CELL and CONTROL helmets.

Peak rotational acceleration $\alpha_y$ was significantly lower in CELL helmets compared to CONTROL helmets for all impacts. Reductions in $\alpha_y$ for CELL helmets compared to CONTROL helmets ranged from 34% for slow impacts on the 60° anvil to 73% for fast impacts on the 45° anvil (Figure 3B). Similarly, peak rotational velocity $\omega_y$ was significantly lower in CELL helmets compared to CONTROL helmets for all impacts, with reductions ranging from 50% for slow impacts on the 30° anvil to 84% for fast impacts on the 45° anvil.

### Brain injury risk prediction

Axonal strain predicted by SUFHEM computational modeling was significantly lower in CELL helmets compared to CONTROL helmets for all slow impacts, with reductions ranging from 63% for slow impacts on the 60° anvil to 85% for fast impacts on the 45° anvil (Figure 4). Peak axonal strain in all CELL helmet tested ranged from 4%-11%, and therefore remained below the 15% strain threshold indicative for a 50% risk of sustaining a moderate DAI [37,38]. Accordingly, the probability p (mDAI) of sustaining moderate DAI ranged from 0% to 3.3% for all impacts with CELL helmets. Conversely, only slow impacts on the 60° anvil yielded a low p (mDAI) value of 3.9% within the foot region of the

| Outcome Category | Result Parameter | Helmet Type | 30° anvil, slow | | | 45° anvil, slow | | | 60° anvil, slow | | | 45° anvil, fast | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | | AVG | STDEV | p-value | AVG | STDEV | p-value | AVG | STDEV | p-value | AVG | STDEV | p-value |
| Impact Conditions | Impact speed [m/s] | CONTROL | 4.80 | 0.02 | - | 4.81 | 0.01 | - | 4.78 | 0.02 | - | 6.20 | 0.02 | - |
| | | CELL | 4.79 | 0.0 | 0.559 | 4.83 | 0.02 | 0.551 | 4.78 | 0.02 | 0.811 | 6.17 | 0.04 | 0.259 |
| | Impact Energy [J] | CONTROL | 163.8 | 1.4 | - | 164.2 | 1.8 | - | 162.6 | 1.1 | - | 272.9 | 1.5 | - |
| | | CELL | 162.8 | 1.3 | 0.558 | 165.3 | 1.0 | 0.553 | 162.0 | 1.1 | 0.813 | 269.9 | 3.6 | 0.259 |
| Head Kinematics | lin. acceleration $a_r$ [g] | CONTROL | 87 | 1.1 | - | 65 | 0.7 | - | 45 | 2.3 | - | 81 | 7.7 | - |
| | | CELL | 64 | 1.0 | <0.001 | 53 | 2.7 | <0.001 | 38 | 1.4 | 0.001 | 80 | 4.2 | 0.808 |
| | rot. acceleration $\alpha_y$ [rad/s$^2$] | CONTROL | 6821 | 219 | - | 6237 | 255 | - | 2743 | 176 | - | 7243 | 574 | - |
| | | CELL | 3262 | 63 | <0.001 | 1702 | 98 | <0.001 | 1802 | 98 | <0.001 | 1962 | 644 | <0.001 |
| | rot. velocity $\omega_y$ [rad/s] | CONTROL | 26 | 0.3 | - | 26 | 0.5 | - | 12 | 1.2 | - | 31 | 2.5 | - |
| | | CELL | 13 | 0.5 | <0.001 | 7 | 1.0 | <0.001 | 3 | 1.9 | <0.001 | 5 | 3.5 | <0.001 |
| Brain Injury Risk | axonal strain SUFEHM [%] | CONTROL | 30 | - | - | 32 | - | - | 12 | - | - | 53 | - | - |
| | | CELL | 11 | - | - | 8 | - | - | 4 | - | - | 8 | - | - |
| | P(mDAI) SUFEHM [%] | CONTROL | 100 | - | - | 100 | - | - | 3.9 | - | - | 100 | - | - |
| | | CELL | 3.3 | - | - | 0.1 | - | - | 0 | - | - | 0.1 | - | - |
| Neck Loading | compression $F_C$ [N] | CONTROL | 4542 | 18 | - | 2817 | 32 | - | 1347 | 53 | - | 3638 | 210 | - |
| | | CELL | 3676 | 44 | <0.001 | 2092 | 57 | <0.001 | 1115 | 109 | 0.005 | 3251 | 234 | 0.049 |
| | shear $F_S$ [N] | CONTROL | 1656 | 21.3 | - | 2062 | 35 | - | 1330 | 88 | - | 2786 | 300 | - |
| | | CELL | 1358 | 17.4 | <0.001 | 1299 | 11 | <0.001 | 950 | 68 | <0.001 | 1639 | 177 | <0.001 |
| | flexion $M_F$ [Nm] | CONTROL | 94 | 0.8 | - | 51 | 1.9 | - | 39 | 1 | - | 64 | 2.1 | - |
| | | CELL | 77 | 0.9 | <0.001 | 40 | 1.6 | <0.001 | 37 | 1.5 | 0.071 | 63 | 6.2 | 0.880 |

**Table 1:** Summary of results from all impact tests in terms of the average outcome and standard deviations (S.D). P-value denotes the significance of differences compared to the CONTROL group.

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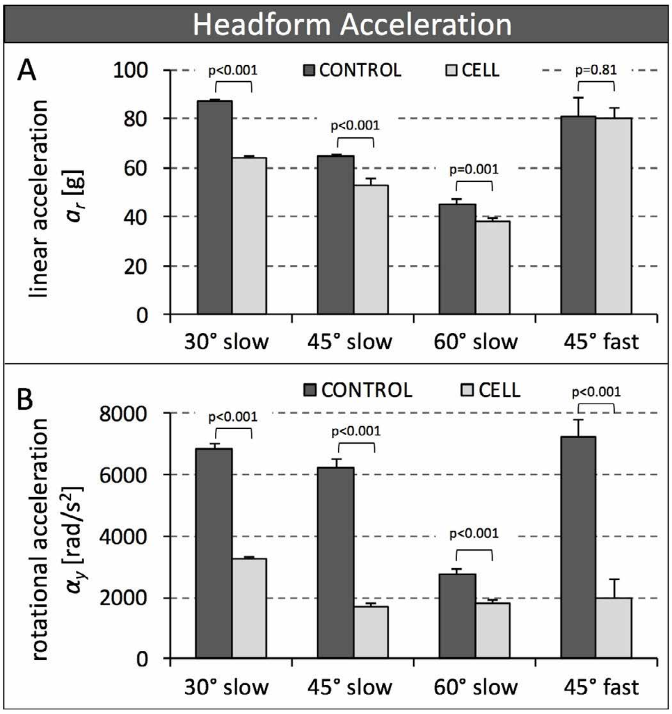


**Figure 3:** Headform kinematic results from impacts onto the three anvil angles at 4.8 m/s (slow), and for the 45° anvil angle at 6.2 m/s (fast): A) Resultant linear headform acceleration; and B) Headform rotational acceleration.

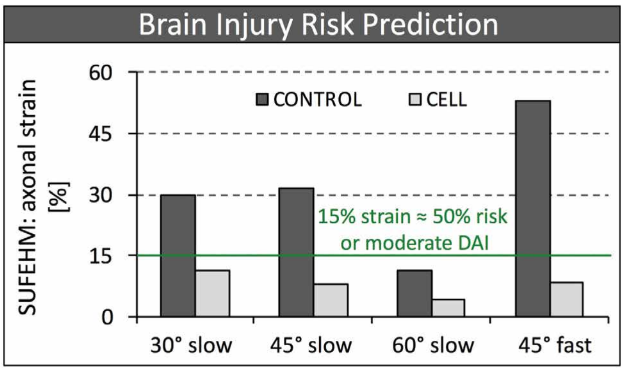


**Figure 4:** Axonal strain from impacts onto the three anvil angles at 4.8 m/s (slow), and for the 45° anvil angle at 6.2 m/s (fast), predicted by the SUFEHM finite element model that accounted for the entire head kinematic history during impact.

injury probability function, while all other impacts with CONTROL helmets caused p (mDAI) values to fall within the saturated region of the injury probability function (Figure 5).

## Neck loading

Peak neck compression $F_C$ was significantly lower in CELL helmets compared to CONTROL helmets for all impacts, with reductions ranging from 11% for fast impacts on the 45° anvil to 26% for slow impacts on the 45° anvil (Figure 6A). Similarly, peak neck shear $F_S$ was significantly lower in CELL helmets compared to CONTROL helmets for all impacts, with reductions ranging from 18% for slow impacts on the 30° anvil to 41% for fast impacts on the 45° anvil (Figure 6B).

The neck flexion moment $M_F$ was significantly lower in CELL helmets compared to CONTROL helmets for slow impacts on the 30° and 45° anvils. However, there was no significant difference in $M_F$ between CELL and CONTROL helmets for slow impacts onto the 60° anvil and for fast impacts onto the 45° anvil (Figure 6C).

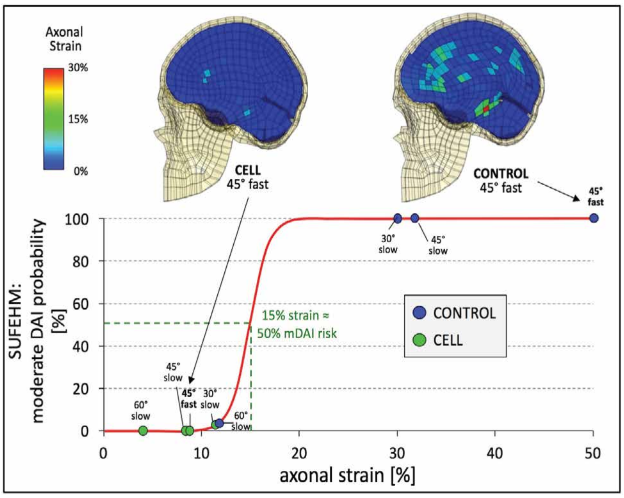


**Figure 5:** Probability of moderate DAI, estimated by the SUFEHM model. Results for all test scenarios are show on the injury risk function (red), which correlates 15% axonal strain with a 50% risk of moderate DAI [25]. Color maps of axonal strain distributions are shown for 6.2 m/s impacts on the 45° anvil.

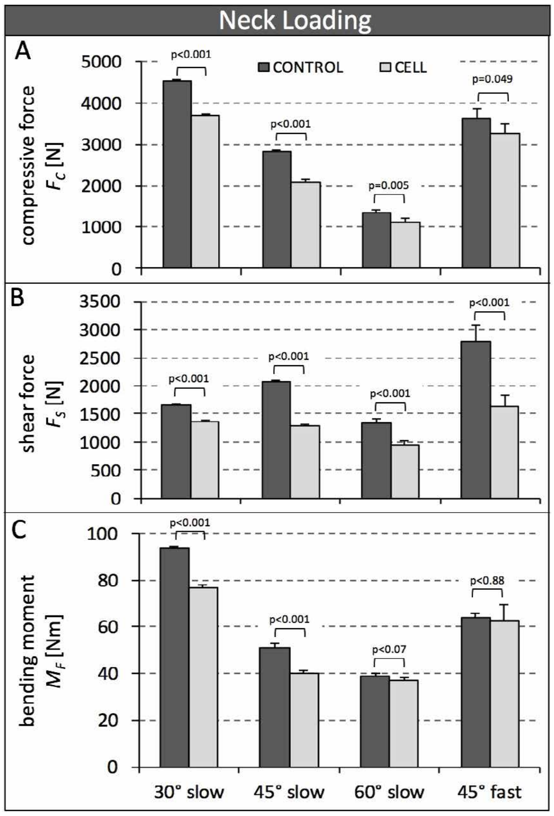


**Figure 6:** Neck loading for impacts onto the three anvil angles at 4.8 m/s (slow), and for the 45° anvil angle at 6.2 m/s (fast). A) Neck compression; B) Antero-posterior neck shear force; and C) Neck bending moment in flexion.

## Discussion

Results demonstrated that CELL technology may provide better protection from rotational head acceleration and associated brain injury, and may reduce neck loading compared to standard EPS helmets.

CONTROL group helmets demonstrated that linear acceleration was effectively suppressed to 87 g or less in all impacts. This linear acceleration is far below the 300 g linear acceleration threshold mandated by the CPSC safety standard, which was developed to prevent skull fractures [13]. Moreover, this linear acceleration correlates with an average linear acceleration of 89 g reported by Bland et al. for oblique impact tests of 10 different bicycle helmet models onto a 30° anvil at 5.1 m/s [29]. While they employed the same Hybrid III neck than the present study, they used a National Operating Committee of Standard for Athletic Equipment (NOCASE) headform. In contrast to the low linear accelerations, they reported average rotational accelerations as high as 4.9 krad/$s^2$ and 6.4 krad/$s^2$ for impact speeds of 5.1 m/s and 6.6 m/s, respectively, whereby the associated concussion risk was nearly saturated at 100% for two of the 10 helmets. While the present study employed a different headform type and orientation, it similarly found high rotational accelerations indicative of brain injury, with impacts onto the 30° anvil inducing on average rotational acceleration of 6.8 krad/$s^2$. A rotational head acceleration of 5.9 krad/$s^2$ has been correlated to a 50% probability of sustaining a concussion [43]. Accordingly, injury risk analysis by SUFEHM predicted a 100% probability of sustaining a moderate DAI for conventional helmets tested on the 30° and 45° anvils. These results confirm the growing recognition that contemporary bicycle helmets can effectively prevent skull fractures, but may not be optimized to mitigate brain injury [27].

CELL group helmets demonstrated a significant reduction in linear acceleration by up to 26% compared to CONTROL group helmets. This finding suggests that controlled buckling of an organized cellular structure may attenuate radial impacts better than compression of traditional EPS foam [29]. Cellular honeycomb structures for protective helmets have been previously explored, since they can deliver controlled energy absorption in a light-weight structure that also permits heat transfer and airflow [30,44,45]. In the most recent helmet comparison study by Bland et al. the highest-ranked bicycle helmet of the 10 helmet models tested was also the only helmet that incorporated a honeycomb structure [29]. More importantly, CELL helmets reduced rotational acceleration to well below 4 krad/$s^2$ in all tests. As a result, axonal strain remained below the 15% axonal strain threshold [25] and the probability of moderate DAI did not exceed 3%, regardless of the test condition. An earlier attempt of employing a cellular structure as a rotational suspension system in bicycle helmets has been introduced by Hansen et al. in the form of an Angular Impact Mitigation (AIM) system, comprised of an elastically suspended aluminium honeycomb liner [30]. In vertical drop tests at 4.8 m/s onto a 30° anvil, their cellular structure reduced linear acceleration by 14%, rotational acceleration by 34%, and neck loading by up to 32% compared to a traditional EPS bicycle helmet.

The relevance of neck loading results may best be interpreted in the context of neck loading during sports activities that present a minimal risk of injury, and those neck loads that cause neck injury. Funk et al. determined neck loading in response to heading a soccer ball at 11.5 m/s for 20 human volunteers [46]. They reported average neck compression (414 N), shear force (144 N) and flexion moments (9 Nm), which are approximately one order of magnitude smaller than neck loading reported in the present study for CONTROL helmets on the 30° anvil. For injurious neck loading, cervical quadriplegia from real-life head-first impacts in athletes was associated with neck compression forces in the range of 3.6-8.1 kN [47]. Biomechanical studies induced compression fractures of cadaveric neck specimens in response to 7.5 kN compressive impact loading [48]. In human cadaveric head and neck specimens, bony and soft tissue injuries resulted from compressive impact forces ranging from 1.6-6.2 kN [49] In the present study, neck compression was as high as 4.5 kN for CONTROL helmets, and 3.6 kN for CELL helmets. Since the magnitude of neck loading in the present study approaches the injurious neck loading range, mitigation of neck loading should be considered for optimization of helmet designs.

Results of this study described the performance of a novel CELL technology for mitigation of rotational acceleration in direct comparison to a traditional EPS helmet design, tested at three impact angles and two impact speeds in the same helmet design. Results of this study are therefore limited to these specific study parameters and may not be extrapolated outside the tested parameter range. The test setup and parameters were selected to align as much as possible with established test standards, precedence from similar studies, and to facilitate reproduction of the test setup in other test facilities. Specifically, impact testing by guided free-fall onto an angled anvil [29-32,50] was chosen over vertical drops onto a laterally translating impact surface [12,28,51] or pendulum impact tests [33,52] for its greater simplicity and high reproducibility [28]. The Hybrid III 50th percentile male anthropomorphic head was chosen, since it is the most widely used human head surrogate employed for impact testing [33]. It readily allows for sensor integration and Hybrid III neck attachment, provides an elastic skin envelope, and its inertial properties are considerably more biofidelic than that of ISO headforms specified in the CPSC safety standard [17]. Quasi-physiologic head constraints have been simulated with a Hybrid III neck, which is the most commonly used human neck surrogate [29]. While there is also precedence for impact testing using an unconstrained headform without a neck surrogate [31,32,50,51] a neck was required in the present study to capture neck loading. Although the Hybrid III neck has been shown to be overly stiff in lateral bending, it was specifically developed and validated for flexion and extension [53] which corresponds to the principal neck motion in response to mid-sagittal frontal impacts in the present study. The Hybrid III head and neck combination has been used in a wide range of impact studies [12,24,29,30,33] and has been proposed for advanced testing of bicycle helmets [17]. The experimental design was limited to an impact location at the helmet front. The helmet front is a commonly impacted region [54] which is involved in over 50% of bicycle helmet impacts [14]. A mid-sagittal impact location was chosen to simplify the impact kinematics, and to match the impact scenarios in previously published studies [21,28,30,31,51]. Limiting the experimental design to one impact location was required to enable exploration of three impact angles and two velocities without exceeding available test resources. Impact angles were chosen to represent the 30°- 60° range determined from reconstruction of real-world bicycle accidents [14,15,28].

In addition to limitations due to simplified simulation of real-world impacts under reproducible laboratory conditions, further limitations must be considered when predicting brain injury risk from impact kinematics data. Headform kinematics was used to compute axonal strain in the SUFEHM finite element model. However, prediction of brain injury risk from axonal strain depends on the accuracy of injury risk functions that have been reconstructed from a limited number of real-world injury data to estimate brain tolerance limits. Moreover, these injury risk functions are highly non-linear, for which reason a relatively small difference in peak rotational velocity or axonal strain can translate into a large difference in injury probability [29]. The uncertainty in defining brain tolerance limits combined with the non-linear nature of injury risk curves necessarily limits the accuracy in predicting an absolute probability of brain injury. However, relative differences in brain injury probability between helmet technologies should provide a meaningful comparison, since the two helmet groups were tested in the same helmet model under reproducible impact conditions. Nevertheless, future studies will be required to expand the parameter range of impact conditions.

## Conclusion

Low linear acceleration results suggest that traditional EPS bicycle helmets are highly effective in preventing skull fractures. Conversely, high rotational acceleration results similarly suggest that these helmets have not been optimized to reduce the risk of concussion and brain injury. Since axonal shear strain caused by rotational acceleration is a predominant mechanism of injury in concussions strategies for improved helmet designs should therefore target mitigation of rotational acceleration. CELL helmets not only reduced rotational acceleration and associated brain injury risk compared to standard EPS helmets, but they also reduced linear acceleration and neck loading.

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### Disclosure

Some of the authors (MB, SMM) are co-inventors of technology described in this manuscript, have filed patents, and have a financial interest in the company that owns this technology.

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