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Angular Impact Mitigation system for bicycle helmets to reduce head acceleration and risk of traumatic brain injury

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ABSTRACT

Angular acceleration of the head is a known cause of traumatic brain injury (TBI), but contemporary bicycle helmets lack dedicated mechanisms to mitigate angular acceleration. A novel Angular Impact Mitigation (AIM) system for bicycle helmets has been developed that employs an elastically suspended aluminum honeycomb liner to absorb linear acceleration in normal impacts as well as angular acceleration in oblique impacts. This study tested bicycle helmets with and without AIM technology to comparatively assess impact mitigation. Normal impact tests were performed to measure linear head acceleration. Oblique impact tests were performed to measure angular head acceleration and neck loading. Furthermore, acceleration histories of oblique impacts were analyzed in a computational head model to predict the resulting risk of TBI in the form of concussion and diffuse axonal injury (DAI). Compared to standard helmets, AIM helmets resulted in a 14% reduction in peak linear acceleration ($p < 0.001$), a 34% reduction in peak angular acceleration ($p < 0.001$), and a 22–32% reduction in neck loading ($p < 0.001$). Computational results predicted that AIM helmets reduced the risk of concussion and DAI by 27% and 44%, respectively. In conclusion, these results demonstrated that AIM technology could effectively improve impact mitigation compared to a contemporary expanded polystyrene-based bicycle helmet, and may enhance prevention of bicycle-related TBI. Further research is required.

1. Introduction

Bicycle-related head injuries in the United States (US) resulted in an estimated 81,000 emergency room visits in 2011, and 77% of these patients were diagnosed with traumatic brain injury (TBI) (CPSC, 2011). Among children and teenagers, bicycling results in more cases of TBI than any other sport or recreational activity (Gilchrist et al., 2011). The US healthcare costs due to bicycle-related head injuries total over $2 billion annually (Schulman, 2002). The number of bicycle-related TBIs has increased steadily over the past fifteen years, in spite of increased rates of helmet use among cyclists (CPSC, 2011; Karkhaneh, 2006). Mandatory helmet test standards assess linear head acceleration but fail to capture angular head acceleration (BSI, 1997; CPSC, 1998), despite the fact that angular acceleration is also known to cause TBI (Goldsmith and Monson, 2005). Contemporary helmets are designed to meet these linear head acceleration standards, but lack specific mechanisms to mitigate angular head acceleration.

Conventional bicycle helmets consist of three layers: a plastic outer shell, an expanded polystyrene foam (EPS) liner, and an inner layer of comfort foam padding. This design is intended to mitigate skull fracture and focal brain injury. The current safety standards for bicycle helmets in the US and Europe establish limits for peak linear acceleration in response to an idealized normal impact test, in which a helmet is dropped vertically onto a horizontal surface and whereby the head surrogate is constrained to prevent angular acceleration (BSI, 1997; CPSC, 1998). These standards have been effective in driving the design of safer helmets: bicycle helmets have been shown to reduce the risk of head injury by an estimated 31–69% (Abu-Zidan et al., 2007; Amoros et al., 2012; Attewell et al., 2001; Cook and Sheikh, 2003; Thompson et al., 1996).

However, angular acceleration is also recognized as a cause of TBI. Primate studies conducted over thirty years ago demonstrated that angular acceleration can induce a range of traumatic brain injuries, including concussion, diffuse axonal injury (DAI), and acute subdural hematoma (SDH), even in the absence of a direct

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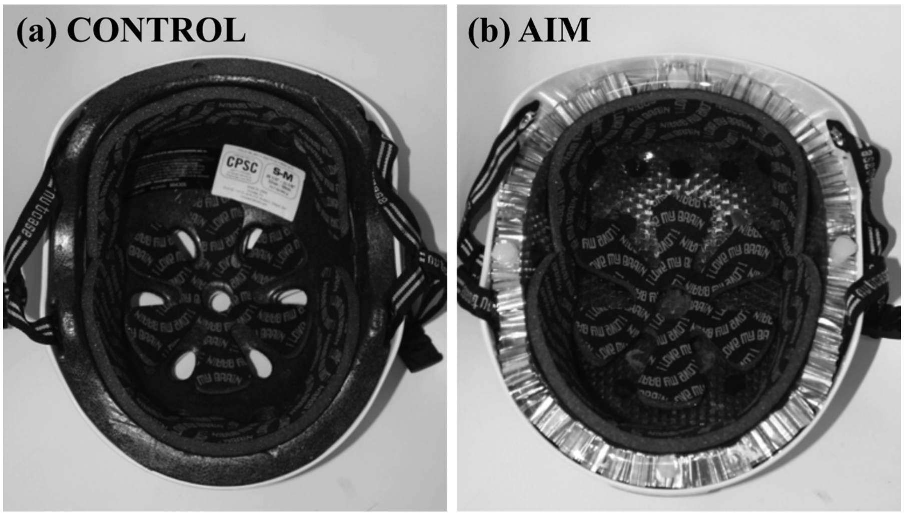

Fig. 1. (a) Commercially available CONTROL helmet, consisting of ABS outer shell, EPS energy-absorbing liner, and polyurethane comfort padding. (b) Prototype Angular Impact Mitigation (AIM) helmet, with EPS liner replaced by suspended aluminum honeycomb.

impact to the head (Gennarelli and Thibault, 1982; Gennarelli et al., 1982; Ommaya and Hirsch, 1971). The mechanism for these injuries has been further investigated through physical models (Bottlang et al., 2007; Margulies et al., 1990), cadaver studies (Hardy et al., 2007), and computational simulations (Deck et al., 2007; Takhounts et al., 2008; Weaver et al., 2012; Willinger et al., 1999; Zhang et al., 2004), which have demonstrated that the brain is highly susceptible to shear strain induced by angular head acceleration.

The increased awareness of the connection between angular acceleration and TBI has sparked research to determine angular head accelerations in realistic impact scenarios. Physical tests and finite element models have been developed to measure the angular accelerations induced in oblique impacts that account for the tangential as well as normal forces that are typically present when a helmeted bicyclist contacts an impact surface (Aare and Halldin, 2003; Ivarsson et al., 2003; Mills and Gilchrist, 2008a,b; Pang et al., 2011). These angular accelerations have been shown to exceed the thresholds expected to cause TBI, even while linear accelerations remained below the limits established in helmet safety standards (BSI, 1997; CPSC, 1998; Pang et al., 2011). Moreover, a recent proposal was made to introduce tangential impact and improved brain injury criteria into future bicycle helmet test standards (Deck et al., 2012).

To reduce the risk of TBI among helmeted bicyclists, a novel bicycle helmet was developed with an Angular Impact Mitigation (AIM) system capable of reducing both linear and angular head acceleration. The AIM system is comprised of an aluminum honeycomb liner that is elastically suspended between an inner liner and outer shell. The aluminum honeycomb material provides a highly effective crumple zone, while the innovative suspension method mitigates angular acceleration by permitting elastic translation of the outer helmet shell relative to the head.

This study was designed to compare the impact mitigation performance between standard bicycle helmets with and without AIM technology, based on improved brain injury criteria. It was hypothesized that the AIM system would provide improved mitigation of linear and angular acceleration, and a reduction in TBI risk.

2. Methods

Bicycle helmets with and without AIM technology were subjected to impacts in a vertical drop test stand to compare the resulting head acceleration levels. First, linear head acceleration was measured in response to normal impact tests onto a horizontal surface. Second, angular head accelerations were captured in response to oblique impact tests onto a surface angled 30° from horizontal. Finally, acceleration histories of oblique impacts were implemented into a validated computational head model to predict the resulting TBI risk.

2.1. Helmets

For the CONTROL group, 10 identical commercially available bicycle helmets (Street Solid size S-M, Nutcase Helmets, Portland, OR) (Fig. 1a) were tested. These helmets consisted of a 3 mm thick acrylonitrile butadiene styrene (ABS) outer shell, a 17 mm thick, 85 kg/m^3 density expanded polystyrene (EPS) liner, and 8 mm thick polyurethane comfort padding. These hard shell bicycle helmets were chosen because they enabled the EPS liner to be replaced by an Angular Impact Mitigation (AIM) system, with no modification of the outer shell, retention system, or fit.

For the AIM group, 10 additional CONTROL helmets were modified by replacing their EPS liners with an AIM system, while retaining the outer shell, comfort padding, and retention straps (Fig. 1b). The AIM system consisted of a 17 mm thick aluminum honeycomb liner (5052/F40-0.0019 Flex-Core, Hexcel, Stamford, CT) of 50 kg/m^3 density that was elastically suspended between the outer shell and an inner liner. The unique cell structure of this particular honeycomb allowed forming the liner into a spherical shape inside the helmet shell while retaining a regular cell geometry. For mitigation of linear acceleration, this honeycomb served as a non-elastic crumple zone to absorb the normal component of the impact force that was directed perpendicular to the outer helmet shell. For mitigation of angular acceleration, the honeycomb was suspended between the outer ABS shell and an inner polymer liner, which was thermoformed from 0.8 mm thick polyethylene terephthalate (PETG). To enable elastic translation between the outer shell and inner liner, the honeycomb was attached at discrete fixation points to the crown of the outer shell and to the periphery of the inner liner by means of a permanent adhesive (Surebonder 707, FPC, Wauconda, IL) (Fig. 2a). Adhesive felt pads at the interior and exterior surfaces of the honeycomb facilitated sliding between the honeycomb and the adjacent layers. In this configuration, the honeycomb acted as an elastic spherical bearing between the outer shell and inner helmet liner to absorb the impact force component that acted tangential to the helmet shell, mitigating angular head acceleration

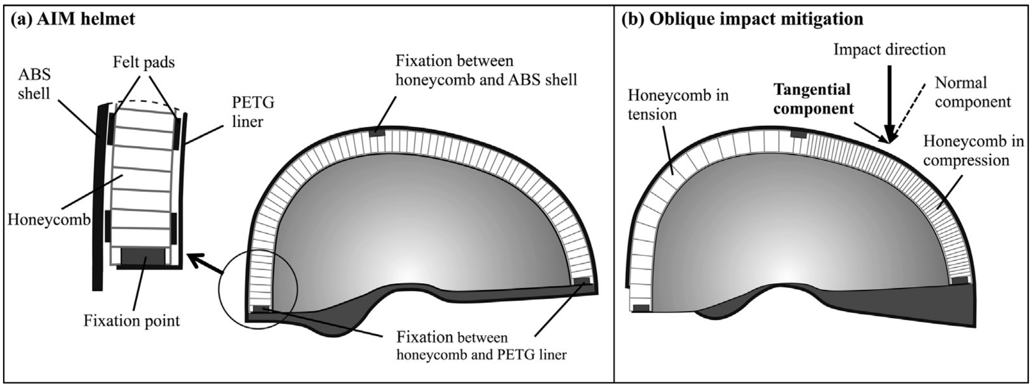

Fig. 2. (a) AIM helmet with elastically suspended honeycomb liner that is attached at the crown to the ABS shell, and at the periphery to the inner PETG liner. (b) The resulting spherical bearing absorbs the tangential impact component by allowing for relative displacement between the outer shell and inner liner, whereby the honeycomb liner undergoes compression in one segment and tension in the opposing segment.

(Fig. 2b). After implementing this AIM system, the original comfort foam padding of the standard helmet was applied to the inner PETG liner. Compared to the CONTROL helmets, the resulting AIM prototype helmets had an impact liner of equivalent thickness and an identical ABS outer shell and comfort foam padding. The mass of AIM helmets (408 ± 4 g) and CONTROL helmets (408 ± 16 g) was also equivalent.

2.2. Impact testing

Normal impact tests were conducted to be consistent with the flat anvil impact test specified in the bicycle helmet safety standard §1203 of the US Consumer Product Safety Commission (CPSC, 1998). In these tests, helmets were fitted onto a magnesium alloy headform (ISO/DIN 622-1983 size J, Cadex, Montreal, Canada), and dropped in a guided free fall onto a horizontal, flat steel anvil. The helmet was attached to the headform with the original retention system for all tests. Headform motion was constrained to the vertical (z) direction by a fixed-angle connection between the headform and the monorail drop assembly (Fig. 3a). The mass of the entire drop assembly was 5.0 kg. The headform was fixed at a 45° angle to induce a frontal impact. The drop height was set to 2.15 m to target the impact velocity of 6.2 m/s specified in the CPSC standard. Linear acceleration (a_{cg}) during the impact was measured with a linear accelerometer (356B21, PCB, Depew, NY) mounted at the center of gravity (CG) of the headform, and oriented to capture acceleration along the z-axis. Impact velocity was measured with a laser trap (PC4200, Cadex) located 5 mm above the point of impact. To assess the amount of impact energy that was not depleted by the helmet but was projected into elastic recoil after impact, the helmet rebound velocity was also measured.

In addition, oblique impact tests were conducted in which the helmeted headform was also subjected to angular accelerations. This more realistic impact simulation employed the same vertical drop tower used for normal impact testing, but it implemented three modifications critical to assess a helmet's mitigation of angular head acceleration (Fig. 3b) (Dau et al., 2012). First, the horizontal anvil of the normal impact test was angled 30° from the horizontal to induce an oblique impact in response to a vertical drop. Second, a biofidelic neck surrogate (Hybrid III, 78051-336, FTSS Inc., Plymouth, MI) was used to connect the headform to the drop assembly, providing quasi-physiologic head restraints and enabling head rotation. The mass of the headform assembly was 4.5 kg, and the center of mass was located 219 mm superior from the inferior surface and along the centerline of the Hybrid III neck. Third, sensors were added to the headform and neck to measure angular head accelerations as well as neck loading. The headform was instrumented with two biaxial accelerometers (356B21, PCB), one located at the CG and a second located 78.5 mm anterior and 28.1 mm inferior of the CG. Both accelerometers were located in the mid-saggital plane and were used to calculate the angular acceleration (α_y) of the headform about the lateral (y) axis. Additionally, the base of the surrogate neck was instrumented with a 3-axis load cell (IF-203, FTSS Inc.) that measured neck shear (F_x), neck compression (F_z), and the neck flexion/extension moment (M_y). The mass of the entire drop assembly was 14.9 kg.

The performance of this oblique impact test system has previously been formally characterized for a reduced drop height of 1.2 m (Dau et al., 2012), resulting in an impact velocity of 4.8 m/s as specified in CPSC §1203 for impact tests onto curbstone and hemispherical anvils. The same drop height was used in the present study for the simulation of oblique impacts.

Ten AIM helmets and ten CONTROL helmets were tested under identical impact conditions. Five helmets of each group were tested in normal impacts, and five helmets of each group were tested in oblique impacts.

2.3. Data acquisition and analysis

Data were simultaneously captured at a sample rate of 20 kHz in one data acquisition system (PCI-6221, National Instruments, Austin, TX), and were post-processed using custom LabVIEW (National Instruments) software. Accelerations and forces were low-pass filtered at Channel Frequency Class (CFC) 1000, and moments were low-pass filtered at CFC 600, as specified by SAE J211 (2007).

For normal impacts, the linear acceleration history was used to calculate the Head Injury Criterion (HIC), which represents an established head injury metric that accounts for both the magnitude and duration of linear acceleration (Marjoux et al., 2008):

$$\mathrm{HIC} = (t_2 - t_1)\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2}(a_{cg}dt)\right]^{5/2}$$

where a_{cg} is the z-directional acceleration (g) at the CG of the headform, and whereby t_1 and t_2 are the initial and final times (s) over which HIC is maximized.

For oblique impacts, angular head acceleration about the lateral axis (α_y) was calculated from the two bi-directional accelerometer signals based on rigid body kinematics:

$$\alpha_y = \frac{r_x(a_{cg,z} - a_{front,z}) - r_z(a_{cg,x} - a_{front,x})}{r_x^2 + r_z^2}$$

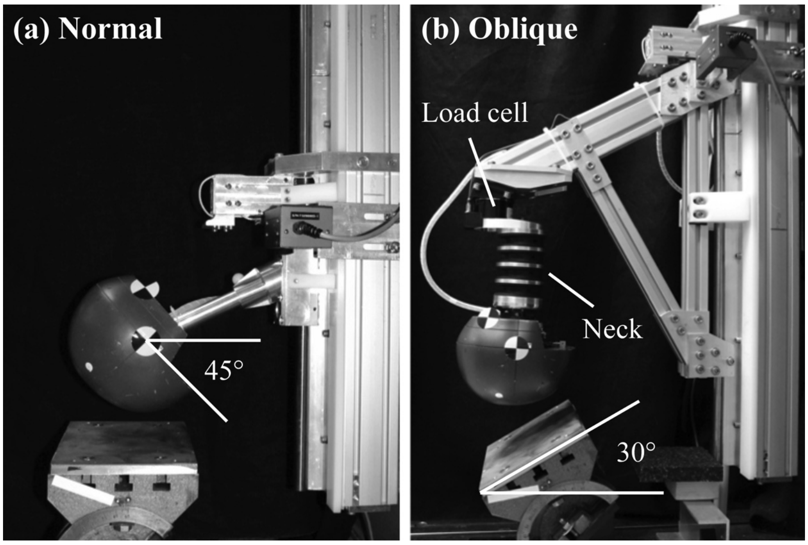

Fig. 3. Drop tester setups. (a) Normal impact setup, as specified by CPSC bicycle helmet testing standard. (b) Oblique impact setup, with angled anvil, Hybrid III biofidelic neck, and 3-axis load cell.

where r_x and r_z are the *x*- and *z*-directional distances between the two accelerometers, and *a* denotes linear acceleration, with subscripts indicating sensor location (*cg* or *front*) and direction (*x* or *z*). An additional CFC 180 filter was applied to the angular acceleration signal in post-processing to reduce high-frequency noise (Newman et al., 2005).

Statistical analyses were conducted using SPSS software (IBM, Armonk, NY). Acceleration, HIC, neck loading, rebound velocity, and rebound energy results were compared between the AIM and CONTROL groups using independent, two-sided, Student's *t*-tests. A value of $\alpha = 0.05$ was used for the evaluation of statistical significance.

2.4. Computational modeling

For correlation of acceleration signals to TBI risk, head acceleration histories of oblique impact tests were analyzed in the validated Strasbourg University Finite Element Head Model (SUFEHM) (Deck and Willinger, 2008; Willinger et al., 1999). This computational model of an adult head consists of 13,208 elements that represent the skull, face, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum, and brain stem, and it accounts for cerebrospinal fluid and the viscoelastic constitutive properties of brain tissue. Injury tolerance curves for this model have been established by reconstructing 68 real-world head impacts to retrieve head acceleration histories as input for the SUFEHM, and by correlating the resulting intracerebral stress distributions with the TBI diagnoses of those 68 patients (Deck and Willinger, 2008). This SUFEHM validation established risk functions for concussion, defined by an Abbreviated Injury Scale (AIS) level of 2 or 3, and DAI, defined by an AIS level of 4 or greater.

For the present study, the SUFEHM was used to calculate the transient intracerebral stresses in response to prescribed head acceleration histories. For this purpose, an average acceleration history for the CONTROL helmet and for the AIM helmet was calculated from the five oblique impact tests conducted for each helmet group. These average acceleration histories were implemented into the SUFEHM to determine intracerebral loading in terms of von Mises stress and to predict the risk of concussion and DAI.

3. Results

3.1. Normal impact test

The average impact velocity of AIM helmets (6.2 ± 0.1 m/s) was comparable to that of CONTROL helmets (6.1 ± 0.1 m/s). The non-elastic crumple zone of AIM helmets yielded a 24% lower rebound velocity ($p < 0.001$), and a 43% reduction in rebound energy ($p < 0.001$) compared to CONTROL helmets (Table 1). The maximum linear acceleration (a_{cg}) of the headform during impact was 14% lower with AIM helmets than with CONTROL helmets ($p < 0.001$). The corresponding HIC values were 15% lower in AIM helmets

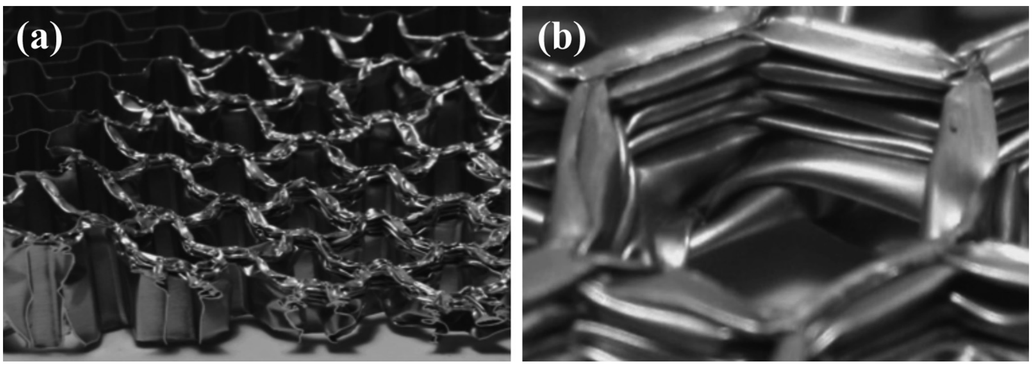

Fig. 4. (a) Representative cross-sectional cut of aluminum honeycomb after a normal impact, demonstrating (b) highly organized buckling.

Table 1
Summary of results from the normal impacts.

Helmet	Rebound velocity (m/s)	Rebound energy (J)	Max linear acceleration (g)	HIC
CONTROL	2.23 (0.08)	1.01 (0.11)	281 (9)	1938 (47)
AIM	1.69 (0.22)	0.58 (0.14)	242 (13)	1640 (112)
Difference	−24%*	−43%*	−14%*	−15%*

Standard deviations are in parentheses.
* Statistical significance.

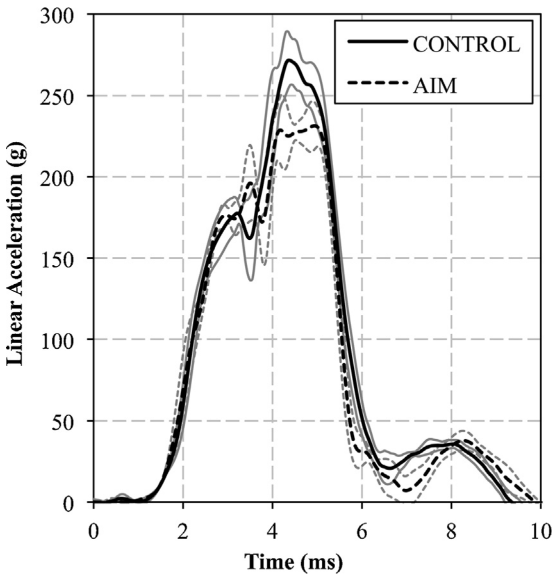

Fig. 5. Averaged linear acceleration signals from normal impact tests. Gray lines represent the one standard deviation corridor.

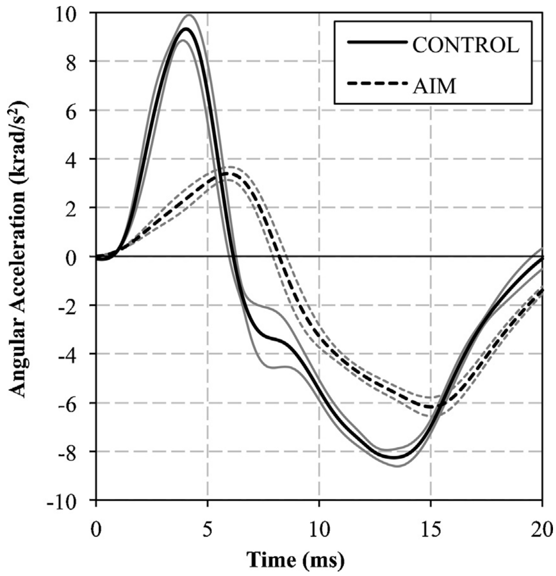

Fig. 6. Averaged angular acceleration signals from oblique impact tests. Gray lines represent the one standard deviation corridor. Positive angular acceleration is in the direction of neck flexion.

than in CONTROL helmets ($p < 0.001$). Post-impact inspection demonstrated a highly ordered crumpling of the honeycomb liner in AIM helmets, with a maximal liner compression of 67% (Fig. 4). The maximal compression of the EPS liner in CONTROL helmets was 31%.

3.2. Oblique impact test

The average impact velocity of AIM helmets (4.8 ± 0.0 m/s) and CONTROL helmets (4.8 ± 0.1 m/s) was equivalent. Unlike linear acceleration histories in normal impacts (Fig. 5), angular acceleration histories in oblique impacts exhibited a reversal of the direction of headform acceleration (Fig. 6). This reversal was evident on high-speed imaging of impacts, which demonstrated headform rotation in the direction of neck flexion in the early impact phase, followed by headform rotation in the direction of neck extension toward the rebound phase (Fig. 7). The peak magnitude of angular acceleration (α_y) during impact was 34% lower in AIM helmets than in CONTROL helmets ($p < 0.001$) (Table 2). The peak-to-peak value of α_y was 46% lower in AIM helmets than in CONTROL helmets ($p < 0.001$). AIM helmets reduced neck shear (F_x) by 32% ($p < 0.001$), neck compression (F_z) by 25% ($p < 0.001$), and the neck moment (M_y) by 22% ($p < 0.001$) compared to CONTROL helmets.

Calculation of intracerebral stress with the SUFEHM in response to the average acceleration histories demonstrated that AIM helmets reduced peak von Mises stress in the brain by 39% compared to CONTROL helmets (Fig. 8). Based on the established TBI risk functions of the SUFEHM, the risk of sustaining concussion from the oblique impact was 99% for the CONTROL helmet and 72% for the AIM helmet (Fig. 9). The risk of DAI was 54% for the CONTROL helmet and 10% for the AIM helmet.

4. Discussion

The results of this study confirmed the hypothesis that, compared to a standard helmet with an EPS liner, an AIM helmet can provide improved mitigation of acceleration in both normal and oblique impacts, and can reduce the risk of TBI in oblique impacts.

In normal impacts, AIM helmets delivered a 14% reduction of linear acceleration and a 15% improvement in HIC value, both of which may largely be attributed to the superior impact absorption properties of aluminum honeycomb. In vehicle crash testing, aluminum honeycomb represents the gold standard for impact energy absorbers, where it is valued for its uniform buckling behavior and nearly constant crush response (Zarei and Kröger, 2008). In both static and dynamic crush tests, aluminum honeycomb has been shown to absorb more than twice the energy per unit volume as EPS (Caserta et al., 2010). Additionally, compression of the aluminum honeycomb liner is primarily plastic, while EPS undergoes more elastic deformation that causes an increased storage of impact energy. The effect of this elastic liner deformation was evident in the rebound energy, which is 43% lower in AIM helmets than in CONTROL helmets. This effect can also be seen in the linear acceleration time histories of the two helmets. Because the area under an acceleration time history curve represents a change in velocity, one would typically expect a lower peak acceleration to correspond to a longer impact duration for the same impact velocity. The acceleration pulse durations are virtually identical between helmets, however, even though peak acceleration is reduced by 14% for the AIM helmets. This disparity is most likely due to the higher rebound velocity for the CONTROL helmets. In addition to improved impact absorption properties, aluminum honeycomb combines several properties that are desirable for bicycle helmets.

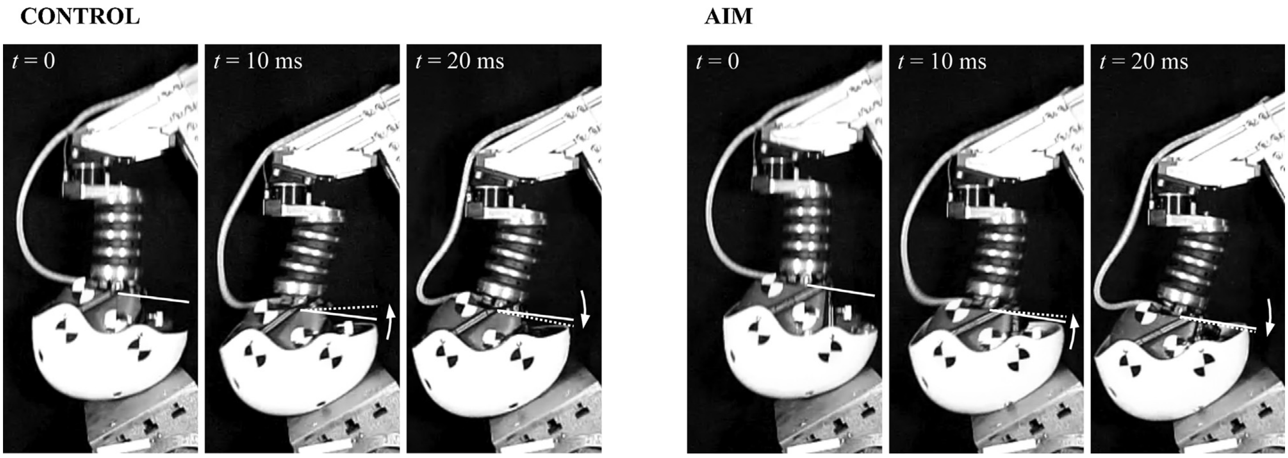

Fig. 7. High-speed capture of oblique impacts with CONTROL and AIM helmets. From $t = 0$ to $t = 10$ ms, the headform rotates in the direction of neck flexion. From $t = 10$ ms to $t = 20$ ms, the headform rotates in the direction of neck extension.

Table 2
Summary of results from the oblique impacts.

Helmet	Max angular acceleration (krad/s^2)	Peak-to-peak angular acceleration (krad/s^2)	Max neck shear (N)	Max neck compression (N)	Max neck moment (N m)
CONTROL	9.39 (0.51)	17.69 (0.80)	1930 (45)	5550 (88)	102.7 (2.6)
AIM	6.20 (0.39)	9.64 (0.63)	1318 (47)	4160 (41)	79.7 (1.3)
Difference	−34%*	−46%*	−32%*	−25%*	−22%*

Standard deviations are in parentheses.
* Statistical significance.

It is lightweight, allows for heat transfer and airflow, and is resistant to moisture and temperature changes. Aluminum honeycomb has previously been explored in a motorcycle helmet design that incorporated a hybrid honeycomb/EPS liner. Motorcycle helmets were tested in normal impacts at 7.6 m/s onto both flat and curbstone anvils (Caserta et al., 2011). The hybrid liner reduced peak linear acceleration by up 27%. However, the hybrid liner was only tested in motorcycle helmets, and was neither designed nor tested in regard to mitigation of angular head acceleration in oblique impacts.

In oblique impacts, AIM helmets delivered a 34% reduction in peak angular head acceleration. This improved impact mitigation may be attributed to the combined effects of the improved energy absorption of aluminum honeycomb, and the elastic suspension of the honeycomb liner. This suspension system effectively decoupled the head from the outer shell, and enabled dissipation of impact energy through in-plane deformation of the honeycomb. This functionality of the honeycomb suspension was confirmed by the relative displacement between the inner liner and outer shell (Fig. 10), which was visualized by a permanent marker tip embedded in the honeycomb to trace relative liner displacement during impact. Marker tracings demonstrated a peak elastic displacement of 4.5 cm and a residual post-impact displacement of 2.1 cm. High-speed imaging revealed that this decoupling effect was particularly pronounced during the first phase of the impact (0–10 ms), during which the helmet shell remained compressed onto the anvil without sliding, and in which the majority of displacement between the inner liner and the outer shell occurred. During this first impact phase, AIM helmets reduced the peak angular acceleration by 67% compared to CONTROL helmets. The second phase of the impact (10–20 ms) was dominated by helmet rebound from the anvil, whereby the helmet separated from the anvil and the headform accelerated from flexion into extension.

In oblique impacts, AIM helmets furthermore reduced neck forces and moments by 22–32% compared to CONTROL helmets.

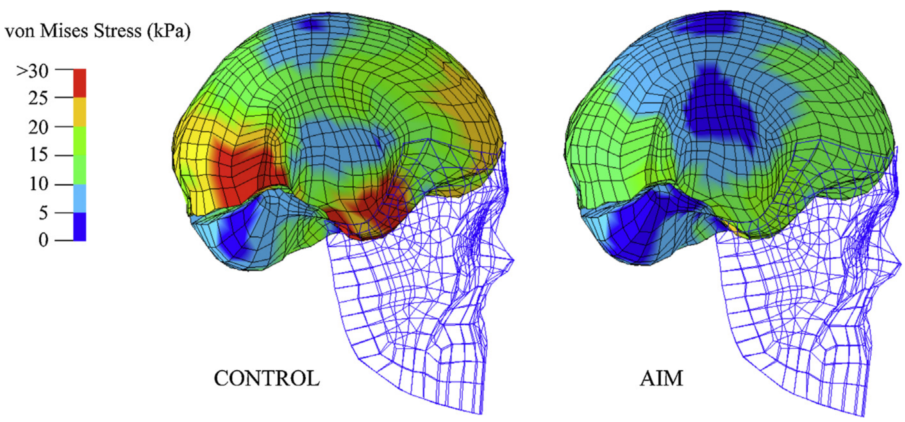

Fig. 8. Contour plot of von Mises stress in the oblique impact tests with CONTROL and AIM helmets, calculated by the Strasbourg University Finite Element Head Model (SUFEHM) based on measured acceleration reports.

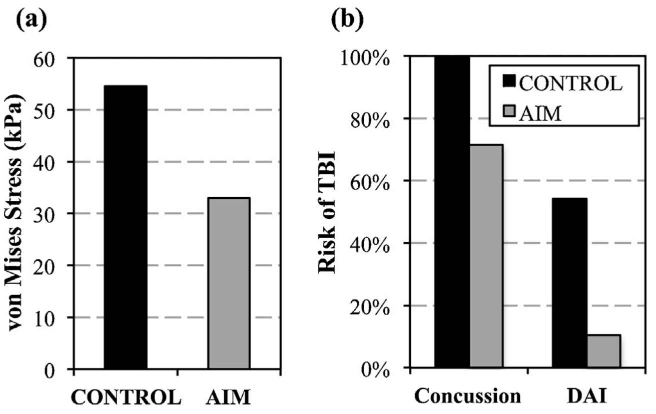

Fig. 9. Oblique impact results from SUFEHM. (a) Maximum intracerebral von Mises stress. (b) Risk of concussion and DAI.

Multiple factors may have contributed to the observed reduction in neck loading. First, the honeycomb suspension decoupled the outer shell and inner liner and thereby mitigated transmission of rotational forces to the neck. Second, the superior impact absorption of aluminum honeycomb compared to an EPS liner reduced the peak impact load onto the neck. Third, the reduced recoil energy observed for the plastically deforming aluminum honeycomb decreased the neck extension forces observed in the rebound phase of oblique impacts. Such reduction of neck loading is likely a critical component for injury prevention. Williams (1991) reported a series of 64 helmeted bicyclists with head injuries, 6 of which (9%) also had neck injuries, including fractures of the cervical spine. Furthermore, several studies on bicycle helmet efficacy have found that bicycle helmets reduce the risk of fatalities and TBI, but may in fact increase the risk of neck injury (McDermott et al., 1993; Wasserman and Buccini, 1990).

Computation of the intracerebral stress response to headform acceleration in the SUFEHM furthermore demonstrated that AIM helmets decreased the risk of concussion and DAI by 27% and 44%, respectively, compared to CONTROL helmets. The SUFEHM was selected because it takes into account the entire acceleration history during impact, and because it provides thoroughly validated injury tolerance curves that correlate an intracerebral stress response to the risk of TBI. Alternative injury risk functions (IRFs) have been introduced that correlate peak angular acceleration to specific brain injury criteria. Rowson et al. (2011) developed a concussion IRF based on field event data from college football players wearing instrumented helmets. Zhang et al. (2004) also correlated angular acceleration to the risk of concussion, using laboratory recreations of National Football League impacts. Takhounts et al. (2008) developed a brain injury FE model that could be used to correlate peak angular acceleration to maximum principal strain in the brain tissue, and in turn to the risk of DAI. For a comparison with the SUFEHM results, the peak angular accelerations from the present study were evaluated in these alternative IRFs (Table 3). The SUFEHM predicted a 99% risk of concussion for CONTROL helmets, which closely correlated to the 97% and 99% risks predicted by the Rowson and Zhang IRFs, respectively. For AIM helmets, the SUFEHM predicted a 27% reduction in risk of concussion, which was less than the reduction predicted by the Rowson IRF (32%) and Zhang IRF (52%). Furthermore, the SUFEHM predicted a 54% risk of DAI for CONTROL helmets, which closely correlated to the 58% risk predicted by the Takhounts IRF. For AIM helmets, the SUFEHM predicted a 44% reduction in the risk of DAI, which was greater than the 28% reduction predicted by the Takhounts IRF. Despite the inherent difficulty in any injury prediction, all of the IRFs and the SUFEHM agreed in their prediction that the AIM helmet reduces the risk of TBI compared to the CONTROL helmet. There are two additional limitations to these results, however. First, the results are limited to the one specific set of impact conditions used in this study. Second, an averaged acceleration time history, rather than each individual impact, was analyzed for each group. Although the impacts were relatively consistent within each group, this averaging may have affected the results.

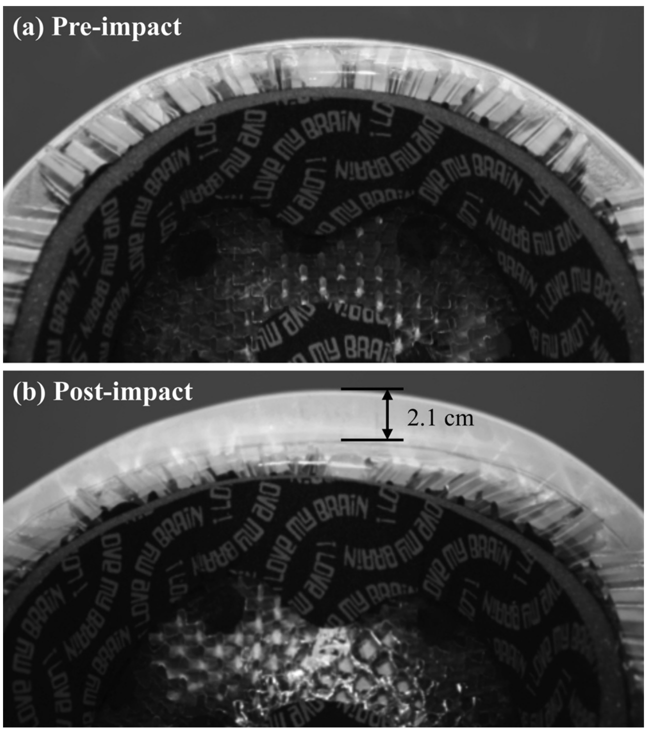

Fig. 10. Front edge of AIM helmet (a) before and (b) after an oblique impact. Note the relative displacement between the inner liner and outer shell.

Several previous oblique impact studies have utilized a laterally translating impact surface (Aare and Halldin, 2003; Mills and Gilchrist, 2008a; Pang et al., 2011), while others have used an angled surface similar to the one in this study (Finan et al., 2008; Ivarsson et al., 2003). While this difference could potentially change the impact kinematics, both impacts are representative of potential real-world impact scenarios. The angled anvil impact was chosen for this study because of its greater simplicity and repeatability.

The Hybrid III neck was originally validated for whiplash-type impacts, and is overly stiff and less biofidelic in compression (Sances et al., 2002). This increased stiffness may decrease angular acceleration levels compared to a more biofidelic neck (Rousseau et al., 2010). Despite its limitations, the Hybrid III is the most widely accepted human neck surrogate, and there is precedence for its use in oblique impact helmet testing (Pang et al., 2011). The Hybrid III neck was chosen for this study to provide a simple, reproducible means of simulating an oblique impact, and to compare relative differences in oblique impact mitigation between helmet designs.

AIM helmets were tested against a single type of CONTROL helmet. This hard shell helmet was chosen because it easily allowed for a direct replacement of the EPS liner with an AIM construct. Preliminary tests in this laboratory have shown that angular acceleration levels for microshell helmets may be up to 20% lower than those for hard shell helmets. McIntosh and Patton (2012) reported an average linear acceleration of 207 g for microshell helmets dropped from 2 m, which is 26% lower than the CONTROL helmets in this study. Changes to the density and thickness of an EPS liner can also lead to improvements in impact performance. The results of this study are limited to a comparison of the AIM system to one specific CONTROL helmet, and future work should investigate whether the

Table 3
Results of oblique impacts correlated to published injury risk functions.

Helmet	Risk of concussion (Rowson et al., 2011)	Risk of concussion (Zhang et al., 2004)	Risk of DAI (Takhounts et al., 2008)
CONTROL	97%	99%	58%
AIM	65%	47%	30%

AIM system will provide improved impact mitigation over other helmets.

Oblique impact test results were limited to a single impact scenario which was designed to simulate a potentially injurious impact that is representative of a typical bicycle crash, and that has precedence in prior studies. The front of the helmet was chosen as the impact location because it is the most commonly impacted region in bicycle accidents (Ching et al., 1997). A mid-saggital plane impact was chosen to simplify the impact kinematics, and to match the impact scenarios in previously published studies (Aare and Halldin, 2003; Finan et al., 2008; Ivarsson et al., 2003; Mills and Gilchrist, 2008a; Pang et al., 2011). The impact angle was set to 30° to match previously published studies of angular acceleration in oblique head impacts (Finan et al., 2008; Fredriksson et al., 2007; Ivarsson et al., 2003). The impact velocity of 4.8 m/s was chosen to represent a severe impact corresponding to a high probability of TBI when wearing a conventional EPS helmet, which is confirmed by the finite element and IRF analyses showing a risk of TBI greater than 97%. In addition, the impact velocity falls within the range representing a moderate bicycling speed (2.2–6.7 m/s), where most crashes occur (Ching et al., 1997). Furthermore, the 4.8 m/s impact velocity has precedence in the CPSC bicycle helmet standard for impact tests on curbstone and spherical anvils. Prior studies have shown that the friction between the helmet and anvil (Camacho et al., 1999; Finan et al., 2008), and between the headform and helmet (Aare and Halldin, 2003), can affect the kinematics of an oblique impact. Previous head impact studies have included high-friction and low-friction surfaces on the impact anvil (Finan et al., 2008; Mills and Gilchrist, 2008a; Pang et al., 2011), and headform surface modifications such as an artificial scalp and wig (Aare and Halldin, 2003; Mills and Gilchrist, 2008a). The present study used a smooth anvil with a mill-finished surface, and no modification of the interaction between headform and helmet. These interfaces were chosen to provide a simple, low-variance method of comparison between different helmet designs. Since a rough anvil surface would likely have increased angular head acceleration, the present results obtained with a smooth anvil likely represent a conservative “best-case” scenario for CONTROL helmets. Additionally, helmet testing was limited to a single impact for each helmet. As aluminum honeycomb undergoes more permanent plastic deformation than EPS, it is possible that the performance of AIM helmets would decrease if subjected to multiple impacts. While testing was necessarily limited to simplifications of a particular impact scenario that may affect the absolute magnitude of outcome measures, the validity of relative differences between outcome measures was preserved by direct comparison between AIM and EPS liners in the same helmet shell and in well-defined and reproducible impact simulations. Furthermore, while results suggest that the AIM concept improves impact mitigation in linear and angular impacts, future work should further verify AIM performance over a range of impact locations, impact angles, impact velocities, frictional interfaces, and multi-impact scenarios.

5. Conclusion

Contemporary bicycle helmets with standard EPS liners may not be optimized to mitigate head angular acceleration caused by oblique impacts representative of a realistic bicycle crash. AIM helmets with a suspended aluminum honeycomb liner that decouples the outer helmet shell from the inner helmet liner can significantly reduce angular head acceleration and neck loading in an oblique impact compared to EPS-based bicycle helmets. AIM helmets can furthermore reduce linear head acceleration in response to a normal impact. A computational model predicted that the improved mitigation of linear and angular head acceleration by AIM helmets would reduce the risk of TBI. In summary, these results demonstrated the feasibility and efficacy of a bicycle helmet design that is directed toward mitigation of angular acceleration to provide better protection from brain injury than contemporary bicycle helmets with EPS liners.

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Conflict of interest disclosure

Some of the authors of this manuscript (Bottlang, Dau, Hansen, Madey) are listed as inventors on a provisional patent application for a novel bicycle helmet design (Apex Biomedical LLC). Some of the authors are part-time employees of Apex Biomedical LLC (Bottlang, Dau), and members of Apex Biomedical LLC (Bottlang, Madey).

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