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Effects of external helmet accessories on biomechanical measures of head injury risk: An ATD study using the HYBRIDIII headform

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ABSTRACT

Competitive cycling is a popular activity in North America for which injuries to the head account for the majority of hospitalizations and fatalities. In cycling, use of helmet accessories (e.g. cameras) has become widespread. As a consequence, standards organizations and the popular media are discussing the role these accessories could play in altering helmet efficacy and head injury risk. We conducted impacts to a helmeted anthropomorphic headform, with and without camera accessories, at speeds of 4 m/s and 6 m/s, and measured head accelerations, forces on the head-form skull, and used the Simulated Injury Monitor to estimate brain tissue strain. The presence of the camera reduced peak linear head acceleration (51% – 4 m/s impacts, 61% – 6 m/s, $p<0.05$). Skull fracture risk based on kinematics was always less than 1%. For 4 m/s impacts, peak angular accelerations were lower (47%, $p<0.05$), as were peak angular velocities (14%) with the velocity effect approaching significance ($p=0.06$), with the camera accessory. For 6 m/s impacts, accelerations were on average higher (5%, $p>0.05$) as were velocities (77%, $p<0.05$). Skull forces were never greater than 443.2 N, well below forces associated with fracture. Brain tissue strain, the cumulative strain damage measure at 25% (CSDM-25), was lower (56%, $p<0.05$) in 4 m/s but higher (125%, $p>0.05$) in 6 m/s impacts with the camera accessory. Based on CSDM-25 for 4 m/s tests, the risk of severe concussion was reduced ($p<0.05$) from 25% (no camera) to 7% (camera). For 6 m/s tests, risks were on average increased ($p>0.05$) from 18% (no camera) to 58% (camera).

1. Introduction

US statistics show bicycle-related injury as the second most common related to sport and recreation (NEISS Data Highlights, 2012) with 44,000 cyclists injured (726 fatally) in the US in 2012 (NHTSA Traffic Safety Facts Data, 2014). In 2010, 4324 Canadians were injured and hospitalized as a result of a cycling accident (Trauma Registries: Cycling Injury Hospitalizations in Canada, 2009-2010, 2011). Among cycling injuries, head injuries, despite widespread helmet use, account for the majority of hospital admissions and fatalities (Thompson et al., 1999).

One of the primary functions of a helmet is to attenuate forces on the head by redistributing focal impact forces to a larger area of the skull, attenuate peak head acceleration and, thereby, shield the underlying head (Newman, 2002). Helmet performance requirements are dictated by standards committees such as the National Operating Committee on Standards for Athletic Equipment (NOCSAE), The American Society for Testing and Materials (ASTM) and the Consumer Product Safety Commission (CPSC). ASTM and CPSC set minimum criteria on head protection by limiting peak linear head acceleration to be less than 300 g (275 g in some cases) in a drop-impact test. NOCSAE standards set limits on an acceleration–time functional (the Severity Index). Helmets meeting these limits have been credited with eliminating fatal head injuries in many sports. Research assessing ability to prevent the spectrum of relatively milder brain injuries (including concussions) suggests simply minimizing linear acceleration is insufficient.

Epidemiology suggests rates of mild brain injury remain high despite helmet adoption in many sports (Benson et al., 2009; Hootman et al., 2007; McCrory et al., 2009). Biomechanical literature suggests that parameters including linear and angular acceleration (Greenwald et al., 2008), velocity (Takhounts et al., 2013), and directional kinematics (Zhang et al., 2001; Newman et al., 2000) are needed to accurately assess injury risk and protective efficacy. Finite element analysis (FEA) with skull-brain models suggest that transient focal impact forces on the skull lead to cranial deformation and stress propagations, which initially

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form at the impact force site and subsequently migrate through the brain as time progresses (Zhang et al., 2001). Given that these force-induced deformations, tissue stresses and strains are hypothesized to be predictive of injury (Zhang et al., 2001; Hardy et al., 2007), knowledge of head-helmet interaction forces could augment our understanding of brain protection. This logic has motivated recent work assessing force distributions on certification-type magnesium headforms using small flexible force transducers (Ouckama and Pearsall, 2012, 2011), and how these force distributions relate to the classical kinematic measures of helmet performance. In the context of automotive injury research using data from the Crash Injury Research Engineering Network, it has recently been suggested that head contact loading measures should augment kinematics because focal head contact (for non-helmeted vehicle occupants) was evident in $>95\%$ of cases of diagnosed diffuse brain injury (Yoganandan et al., 2009).

In recent years, aftermarket helmet accessories (cameras) made of hard and impact resistant polymers have been marketed to the consumer and are being increasingly used in many sports including road and off-road cycling. These accessories, like external bosses and retaining clips, could focus impact forces and cause injuries, as specifically mentioned in the CPSC Bicycle Helmet Standard (Safety Standard for Bicycle Helmets, Final Rule, 1998). In the popular media, there is speculation that helmet mounted accessories have contributed to serious injury in high profile athletes (Williams, 2014). Large projections, such as external helmet accessories, could pose a problem to helmet wearers because they could amplify and concentrate impact forces on the helmet, penetrate through the helmet liner and make direct contact with the skull causing fracture. It is also possible that the external accessory could be considered a snagging hazard which, in a crash situation, could cause the helmet to be dislodged from its proper position on the wearer's head. Worldwide, the helmet standards communities have yet to adopt specific language for their integration and application with consumer helmets, possibly because the effect these accessories have on head injury risk is unclear. In the case of bicycle helmets, CPSC states that external projections be 7 mm or less, unless the projections collapse or break away during impact. A majority of contemporary helmet-mounted accessories (including the camera mount and camera itself) are larger than the CPSC guideline. Typical camera mounts (not including the camera) project more than 7 mm, and are held onto the helmet with either adhesive tapes or tension straps. It is not clear whether these mounts are designed to collapse or break away on impact. These facts could be viewed as worrisome if it is true that helmeted impact to an external accessory amplifies forces or head kinematics. To our knowledge, there is no biomechanical data on transient head forces or the common kinematic measures used to assess head injury risk during helmeted impact with helmet-mounted accessories.

Focusing on cycling, one of North America's most popular sports, we investigated the hypothesis that camera accessories can cause significant alterations in skull and brain injury risk and head-helmet interaction forces in direct impact. We measured head kinematics, head-helmet interaction forces adjacent to the camera accessory, and computed FE predicted measures of brain tissue strain for impacts, both with and without helmet mounted camera hardware, to determine whether the presence of an external camera mount increases or decreases the risk of skull fracture and brain injury.

2. Methods

We simulated helmeted impact using a purpose-built impact tower comprising an adjustable drop gimbal, an anthropomorphic test device (ATD) head and neck (HybridIII 50th Percentile, total mass of gimbal and head-neck of 10 kg, head circumference 575 mm), and a stationary steel impact anvil (Fig. 1a). In total, 24 CPSC certified helmets (CCM Nexus size medium (circumference range = 540–580 mm), mass: 283 ± 3 g (average $\pm$ SD)) were impacted, 12 with camera accessories (GoPro vented helmet strap camera mount, GoPro Inc., San Mateo CA, mass: 46 ± 1 g) and 12 without. A surrogate camera model (comprising impact resistant polyvinyl chloride) matching the mass (150 g) and dimensions (70 mm $\times$ 60 mm $\times$ 30 mm) of contemporary cameras was used (Fig. 1d). All impacts were frontal, one common region of head impact in cycling (Depreitere et al., 2004), approximately halfway between the head vertex and anterior-most brim of the bicycle helmet along the sagittal plane, to simulate a direct impact to the camera accessory (Fig. 1a). Two realistic impact speeds for bicycling (Fahlstedt et al., 2012) were simulated: nominally 4 m/s (12 impacts) and 6.0 m/s (12 impacts). Drop heights were kept the same for helmeted impacts with and without an external camera accessory so that the helmeted impacts with a camera would have a lower impact speed (as would occur with actual cyclists wearing cameras and falling from the same height as a cyclist without a camera). The difference in impact velocities with and without a camera accessory was not significantly different (low: p-value = 0.85, high: p-value = 0.75).

The HybridIII headform was instrumented with a 9 uniaxial accelerometer (Measurement Specialties Inc. Hampton VA, model 64C-2000-360) array, allowing both linear and angular head accelerations at the head center of mass (Padgaonkar et al., 1975) to be determined. Impact velocities were measured using a purpose-built velocity gate that captures impact speed at a height 30 mm before impact. Between the aluminum skull and vinyl skin of the HybridIII headform, at a location directly beneath the camera mount, we added 4 purpose-built impact force transducers (Fig. 1b, 12 mm diam. each) developed for transient impact force measurements in helmeted impact (Butz and Dennison, 2015) and another 4 impact force transducers remote from the camera mount. The 4 transducers directly beneath the camera mount were covered by impact liner (as opposed to air vents), and captured transient force magnitudes on the region of the head beneath the accessory (circular active sensing region of 36 mm diam.). The 4 remote transducers (anterior–posterior transducers were covered by impact liner, lateral transducers positioned under air vent) capture forces distant (nominally 40 mm) from the accessory. The total mass of all force transducers was 12 g. Paired impacts were performed (with force transducers and without) and a paired t-test was performed to confirm the HybridIII head kinematic response is unaltered as a result of addition of the force sensors: linear head accelerations of the paired impacts were not significantly different (with: 104.9 ± 7.0 g, without: 104.1 ± 9.0 g, $n = 3$, $p = 0.93$).

National instruments hardware and software (PXI 6251 and LabVIEW v.8.5, Austin TX) were used to collect impact telemetry: accelerations at 100 kHz on all 9 channels and impact forces at 10 kHz on 8 channels. Analog voltages were anti-alias filtered using hardware, and subsequently low-pass filtered per CFC 1000 in post processing (SAE J211 Instrumentation for Impact Test - Part 1: Electronic Instrumentation, 2007).

Linear acceleration and rotational acceleration data were processed to obtain peak resultant linear acceleration, peak resultant angular acceleration, linear velocity, and angular velocity.

Following processing of kinematics, three dimensional angular velocities and linear accelerations were input to the improved Simulated Injury Monitor (SIMon) (Takhounts et al., 2008): an FE model approximating the geometry, anatomy and mechanical properties of the average male skull, cerebrospinal fluid layers, bridging veins, and brain. The SIMon model was used to estimate brain strain