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Publication date	2016-07-22
Publication information	Clark, J. Michio, Andrew Post, Thomas A. Connor, Thomas Blaine Hoshizaki, and M. D. Gilchrist. “Effect of Impact Surface in Equestrian Falls.” International Society of Biomechanics in Sports (ISBS), July 22, 2016.
Conference details	34th International Conference on Biomechanics in Sports, Tsukuba, Japan, 18-22 July 2016
Publisher	International Society of Biomechanics in Sports (ISBS)
Item record/more information	http://hdl.handle.net/10197/8375

EFFECT OF IMPACT SURFACE ON EQUESTRIAN FALLS

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This study examines the effect of impact surface on head kinematic response and maximum principal strain (MPS) for equestrian falls. A helmeted Hybrid III headform was dropped unrestrained onto three impact surfaces of different stiffness (steel, turf and sand) and three locations. Peak resultant linear acceleration, rotational acceleration and duration of the impact events were measured. A finite element brain model was used to calculate MPS. The results revealed that drops onto steel produced higher peak linear acceleration, rotational acceleration and MPS but lower impact durations than drops to turf and sand. However, despite lower MPS values, turf and sand impacts compared to steel impacts still represented a risk of concussion. This suggests that certification standards for equestrian helmets do not properly account for the loading conditions experienced in equestrian accidents.

KEYWORDS: Helmet; Concussion; Injury biomechanics; Head kinematics; Brain strain

INTRODUCTION: Equestrian helmets are designed to pass standards which involve a drop test to a rigid steel anvil (EN 1384:1997). The introduction of these standards and use of these helmets has considerably reduced the incidence of traumatic brain injuries (TBI) (Harrison, Mills & Turner, 1996; Northey, 2003). However, concussive injuries continue to occur even when equestrian helmets are worn by jockeys. One possible explanation for the continued high incidence of concussion could be that the rigid steel anvil used in equestrian helmet standards may not reflect the type of impact surfaces that are commonly associated with concussion. Concussions in equestrian accidents are typically a result of falls to soft surfaces such as grass, turf and sand (Mills & Whitlock, 1989). These soft surfaces absorb more impact energy than harder surfaces (Hunt & Mills, 1989) and can affect the loading conditions of the impact. Different impact events are characterized by different loading conditions that can affect the performance of a helmet (Hoshizaki, Post, Oeur & Brien, 2014). The manner in which the head and brain are loaded during an impact is an important consideration for helmet designs. Currently, the loading conditions simulated in equestrian standards may not reflect those of real world accidents. A better understanding of how different impact surfaces influence the response of the head and brain may provide a more effective strategy for developing a safer riding environment through improved helmet design. The purpose of this study is to examine the effect of impact surface on head kinematic response and maximum principal strain for equestrian falls.

METHODS: A 50th percentile Hybrid III headform was used for all impact conditions. The headform used Endevco7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA) in a 3-2-2-2 accelerometer array (Padgaonkar, Krieger & King, 1975). Accelerometer signals were collected at 20 kHz by a TDAS Pro Lab system (DTS, Seal Beach CA) and filtered with a CFC 1000 filter. The headform was dropped unrestrained with the use of a halo which was attached to the drop carriage of a monorail drop rig (Figure 1). The drop carriage ran along a 4.7 m long rail on ball bushings to reduce the effects of friction on the inbound velocity of the headform. The monorail drop rig was connected to a computer equipped with Cadex Software (Cadex Inc., St-Jean-sur-Richelieu, QC), which was used to control the velocity and release mechanism for the impact. The headform was released by a pneumatic piston and the inbound velocity was measured using a photoelectric time gate. The headform was dropped at 5.4 m/s in accordance with equestrian helmet standards (EN 1384:1997) onto steel, turf and sand anvils. The impact locations were the front, side and rear of the head. Three trials were conducted for each condition and peak resultant linear and rotational accelerations and impact duration of the headform were obtained. The resulting linear and rotational accelerations served as input to a finite element model which calculated the magnitude of peak maximum principal strain (MPS) in the cerebrum.

Figure 1: Monorail drop system used for unrestrained drops

The model used in this study was the University College Dublin Brain Trauma Model (UCDBTM) and consisted of 26,000 hexahedral elements representing the scalp, skull, pia, falx, tentorium, cerebral spinal fluid (CSF), grey and white matter, cerebellum and brain stem (Horgan & Gilchrist, 2003; 2004). The head geometry of the UCDBTM was extract from computed tomography (CT) and magnetic resonance imaging scans (MRI) of a male human cadaver (Horgan & Gilchrist, 2004). The material properties of the model were based on cadaveric anatomical research (Horgan & Gilchrist 2003). The brain tissue was modelled as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus (Horgan & Gilchrist, 2003). The compressive behaviour of the brain was considered elastic. The viscoelastic behaviour representing the shear characteristics was defined using the following equation:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{\beta t}$$

where G_∞ represents the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor (Horgan & Gilchrist, 2003). To simulate a sliding boundary condition CSF was modelled as solid elements with a low shear modulus and a high bulk modulus. There was no separation for the contact interaction and a friction coefficient of 0.2 was used (Miller et al., 1998). The UCDBTM was validated through comparisons of cadaveric pressure responses conducted by Nahum, Smith, & Ward (1977) and brain motion research conducted by Hardy et al. (2001). Further validations were done by comparing reconstructions of real world traumatic brain injury (Post et al., 2015).

To assess the influence of impact surfaces two-way ANOVAs were conducted for peak linear and rotational acceleration, MPS and impact duration. When significant main effects were found post hoc Tukey tests were performed. For all comparisons $\alpha=0.05$. Data analysis was performed using SPSS 20.0 for Windows.

RESULTS: The effect of different impact surfaces on linear and rotational acceleration, MPS and impact duration in equestrian falls are shown in Tables 1 and 2. Significant main effects of impact surfaces were found across all dependant variables ($p < 0.01$). Tukey post hoc tests found impacts to a steel anvil produced significantly higher linear and rotational accelerations and MPS compared to turf and sand surfaces ($p < 0.05$). Impact duration for steel impacts was found to be significantly lower than all other impact surfaces ($p < 0.05$). Impacts to the turf and sand surfaces were not significantly different from one another across all dependant variables measured ($p > 0.05$).

Table 1
Mean kinematic response and maximum principal strain (±1 standard deviation) as measured from a hybrid III headform and finite element analysis.

Location	Surface	Linear Acceleration (g)	Rotational Acceleration (rad/s^2)	Maximum Principal Strain (mm/mm)	Impact Duration (ms)
Front	Steel	180.6 (8.8)	4192 (453)	0.304 (0.021)	8.3 (0.9)
	Turf	37.0 (2.8)	2031 (199)	0.192 (0.017)	27 (3.8)
	Sand	45.4 (1.9)	1753 (95)	0.190 (0.033)	24.9 (0.5)
Side	Steel	168.4 (6.2)	6472 (738)	0.347 (0.091)	8.9 (0.5)
	Turf	36.4 (6.1)	1363 (405)	0.125 (0.014)	25.6 (2.3)
	Sand	38.4 (2.5)	1385 (145)	0.135 (0.022)	23.8 (1.8)
Rear	Steel	164.6 (1.7)	2840 (619)	0.310 (0.038)	9.1 (0.2)
	Turf	39.4 (2.6)	1216 (166)	0.120 (0.014)	24.5 (3.6)
	Sand	43.3 (2.4)	910 (99)	0.145 (0.019)	26.1 (1.0)

DISCUSSION: This study examined the effect of impact surface on head kinematic response and MPS for equestrian falls. The results revealed that impacts to a steel anvil were significantly different from all other impact surfaces for all measured variables and that this was due to differences in surface compliance. Soft surfaces absorb more impact energy than harder surfaces (Hunt & Mills, 1989). This results in a change in the head's response to an impact. Falls to rigid impact surfaces such as steel result in a large amount of energy being transferred to the head due to the rigid surface offering little compliance, resulting in high magnitude and short duration linear and rotational accelerations (Post et al., 2015). More compliant impact surfaces result in lower magnitude and longer duration linear and rotational accelerations (Rousseau, 2014). These differences are reflected in impacts to the steel anvil compared to the impacts to the more compliant anvils made of turf and sand. These differences in compliance can further be seen in brain strain levels as impacts to a steel anvil resulted in higher MPS values than impacts to turf and sand anvils. Clearly, the conditions simulated in current equestrian standards (EN 1384:1997) do not reflect those of surfaces which are commonly impacted in equestrian accidents.
Although the more compliant turf and sand surfaces absorb more impact energy, the load which is transferred to the brain can still result in injury. Despite the associated MPS values being lower for impacts to turf and sand compared to impacts to steel, all impact surfaces produced MPS values within the range of concussion (Galbraith, Thibault & Matteson, 1993; Zhang, Yang & King, 2004; Rousseau, 2014). Long duration impacts have been suggested to cause high brain stress and strain (Willinger, Taleb & Kopp, 1992; Gilchrist, 2003). Impacts of this nature have been found to cause diffuse axonal injury and concussions (Rousseau, 2014). Research in ice hockey has found that helmets designed to pass standards involving a drop to a rigid surface are not effective at decreasing MPS values below the level for concussion because the impact surface will absorb most of the energy and the helmet will absorb very little energy (Clark, Post, Hoshizaki & Gilchrist, 2015). As different loading conditions can affect the protective capabilities of a helmet, it is important that the method used to test helmets should be representative of the environment in which they are used (Hoshizaki, et al., 2014). Impact test protocols which better reflect the impact conditions experienced in real world accidents should be developed.

CONCLUSION: This study examined the effect of impact surface on kinematic head response and maximum principal strain for equestrian falls. When comparing impact surfaces, it was found that impacts to a steel anvil produced higher peak linear acceleration, rotational acceleration and MPS values but lower impact durations than the turf and sand surfaces. This suggests that current equestrian helmet standards may not properly account for the loading conditions experienced in equestrian accidents. As MPS values for turf and sand impacts are still within the concussive range, test protocols which better reflect the impact conditions experienced in real world equestrian accidents should be developed.

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Acknowledgements

Equestrian helmets for this research were supplied by Charles Owen. Funding for this research was received from the European Union's Horizon 2020 research and innovation programme under the Marie Skiodowska-Curie grant agreement No. 642662.