Reconstruction of Real World Concussive and Non-Concussive Accidents in Equestrian Sports

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I. INTRODUCTION

Equestrian sports are popular worldwide, however, racing jockeys have a higher rate of concussion than American Football players and boxers [1]. Currently, equestrian helmets are designed to pass certification standards which involve a linear drop test onto a rigid steel surface. Concussions in equestrian sports, however, are most commonly a result of being thrown off the horse and impacting surfaces such as turf or sand [2]. This results in an oblique impact to a compliant surface, the mechanics of which has not been widely studied. Understanding the event characteristics and mechanics which lead to concussions in equestrian sports will have significant implications for equestrian helmet standards and helmet designs. The purpose of this study was the reconstruction of concussive and non-concussive falls in equestrian sports to determine the biomechanics and thresholds of concussive injury in equestrian sports.

II. METHODS

Reconstructions

Concussive and non-concussive falls in equestrian sports were reconstructed using computational and physical methods. For all computational (multiple body dynamics and finite element) and physical models the 50th percentile was used for cases involving males and the 5th percentile for females. Video analysis was conducted using Kinovea 0.8.20 (open source, kinovea.org) to determine impact location, surface type and initial velocity. Report forms were used to confirm ground conditions, .i.e, Going Rating, on the day of the accident and one of three anvils representing Soft, Good or Hard ratings was used to reflect the compliance of the surface for the reconstructions. Mathematical Dynamic Models (MADYMO) simulations were conducted based on body position, impact location and velocity determined from video analysis. Impact velocity and trajectory angle at contact were obtained from these simulations. Impacts with a Hybrid III headform and rail guided launcher (RGL) were performed according to the obtained impact parameters (location, compliance, velocity, angle). Fig. 1 illustrates a schematic example of the experimental reconstructions. Resulting head kinematics (linear acceleration, rotational velocity and rotational acceleration) were obtained from the helmeted headform. Linear and rotational acceleration histories served as input into a University College Dublin Brain Trauma Model (UCDBTM) which was used to determined maximal principal strain (MPS).

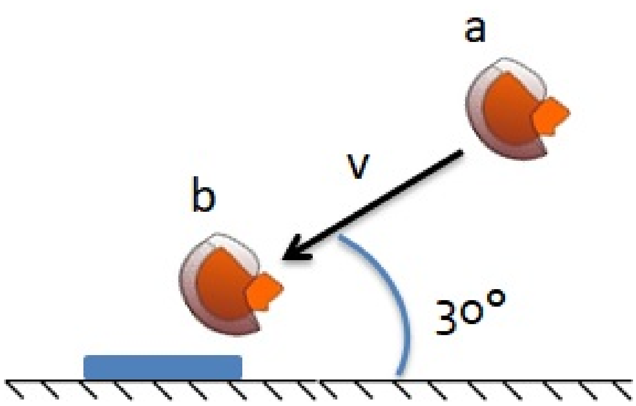

Fig. 1. Schematic example of experimental reconstruction illustrating the trajectory of Hybrid III headform from rail guided launcher to anvil.

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Case Selection

A total of 466 videos of jockey/rider accidents in which they fall off their horses in racing and eventing have been collected to date. The inclusion criteria for analysis required the video 1) to have a clear view of the impact and 2) be able to produce a reliable calibration (camera angle 30-90°, calibration parallel to line of action). If these criteria were not met the case was excluded from the research dataset. Additionally, cases in which multiple impacts occurred to the jockey/riders head (e.g. fall and a kick) were also excluded from this research. Ten concussive and 10 non-concussive cases from horse racing and eventing have been reconstructed to date.

I. NITIAL FINDINGS

The results of an independent samples t-test showed the concussive cases had significantly higher impact velocity, linear accelerations, rotational accelerations, rotational velocity and MPS than the non-concussive cases ($p < 0.05$). No significant difference was found for trajectory angle and impact duration between concussive and non-concussive cases ($p > 0.05$). A binary logistic regression analysis found impact velocity, linear acceleration, rotational acceleration, rotational velocity and MPS were significant predictors of injury ($p < 0.05$). The head kinematic response and MPS results for concussive and non-concussive cases along with the percentile risk classification are presented in Table 1.

TABLE I

THE HEAD KINEMATIC RESPONSE AND MAXIM PRINCIPAL STRAIN RESULTS FOR CONCUSSIVE AND NON-CONCUSSIVE CASES

Variable	Injury Outcome		Risk Classification		
	Non-Concussive	Concussive	25 %	50 %	80 %
Impact Velocity (m/s)	6.6 (2.1)	9.3 (2.3)	7.4	8.0	8.9
Trajectory Angle (°)	27 (10)	33 (18)	NA	NA	NA
Linear Acceleration (g)	49.8 (23.0)	70.0 (19.8)	49	60	74
Rotational Acceleration (rad/s^2)	2072 (1271)	3134 (1238)	1800	2600	3500
Rotational Velocity (rad/s)	19.7 (8.4)	32.0 (15.1)	19	25	33
Impact Duration (ms)	27.0 (2.8)	26.8 (2.8)	NA	NA	NA
Maximum Principal Strain (%)	13.1	29.4	16	23	32

IV. DISCUSSION

The kinematics and MPS during the reconstructed head impacts are similar to those previously reported in the literature [3-5]. Despite similarities with reported values in the literature, these results represented a unique combination of head kinematics compared to other sports. The impacts were of long duration (> 22ms) with low peak rotational acceleration similar to collisions in ice hockey [5] but relatively high peak linear accelerations for the given impact duration with magnitudes more comparable to American football and Australian rules football [3,4]. The low magnitude rotational accelerations and long durations are a reflection of the compliant turf surface, whereas the relatively high magnitude linear accelerations were a reflection of the large amount of energy transferred to the head during falls. These results indicate that when determining the likelihood of injury from peak linear and rotational acceleration the sport in which the injury occurs must be considered. Additionally, the oblique impact to a compliant surface representing concussive conditions in this study resulted in considerably lower and longer accelerations than the commonly used 250 g threshold in current equestrian standards involving a linear drop to a steel surface [6] which result in durations less than 15ms. It is apparent that the current equestrian helmet standards and design do not necessarily account for the loading conditions associated with concussion. An opportunity exists to improve equestrian helmet standard and designs by accounting for impacts that result in acceleration duration of 20 to 30 ms as well as shorter duration traumatic brain injury type events in order to reduce head and brain injury in equestrian sports.

V. REFERENCES

[1] Press, J.M., et al., *Clin J Sport Med*, 1995.

[2] Mills, N., Whitlock, M., *Injury*, 1989.

[3] McIntosh, A.S., et al., *BMJ Open*, 2014.

[4] Zhang, L., et al., J Biomech Eng, 2004.

[5] Karton, C., et al., *Mayo Clinic*, 2017.

[6] BS EN 1384:2012.