Virginia Tech Helmet Lab
343 Kelly Hall
325 Stanger St
Blacksburg, Virginia 24061
P: (540) 231-8254
rowson@vt.edu

Equestrian STAR Protocol

Mark Begonia, Barry Miller, Steve Rowson and Stefan Duma

Laboratory Tests

A pendulum impactor was used to perform all impact tests (Figure 1) [1]. It was chosen for its increased repeatability and reproducibility compared to other impacting methods [2]. The pendulum arm is 190.5 cm long, has a total mass of 36.3 kg with a 16.3 kg impacting mass at the end, and has a moment of inertia of 72 kg·m^2. A flat, rigid nylon impactor face with a 12.7 cm diameter and 2.5 cm thickness was used to minimize any differences between helmets in comparative testing due to impactor compliance. The pendulum impacts a helmeted, medium NOCSAE head, which was modified to fit a 50th percentile male Hybrid III neck. The head and neck were mounted to a 5-degree-of-freedom Biokinetics slide table with a 16 kg sliding mass. This setup allowed for linear and rotational motion to be generated during an impact and representative of the head, neck and torso of a 50th percentile male. Test conditions included three impact locations (Table 1) and two impact speeds (4.0 and 6.3 m/s). Helmet position on the headform was set according to fitting guidelines provided by the manufacturer. Four samples of each helmet model were subjected to one impact per location and speed. Two trials were performed at each of the six test configurations, producing a total of 12 impacts per helmet model.

Figure 1: Pendulum impactor used for Equestrian STAR testing.

Figure 2: Impact locations from left to right: back, front, side.

Table 1: Relative NOCSAE headform translations and rotations on the linear slide table for each impact configuration.

Location	Y (cm)	Z (cm)	Ry (deg)	Rz (deg)
Back	0	+6.3	0°	180°
Front	0	+0.8	-30°	0°
Side	-1	+4.8	-10°	-100°

Notes: All measurements are made using the SAE J211 coordinate system in relation to a "zero" condition in which the headform is in a position of 0° Y and Z-axis rotation and the median (midsagittal) and basic (transverse) plane intersection of the headform is aligned with the center of the impactor. The x-position is set such that the helmet contacts the impactor face when the pendulum arm is in a neutral vertical position for each location.

The NOCSAE headform was instrumented with three linear accelerometers and a triaxial angular rate sensor (ARS) at its center of gravity to measure linear and rotational impact kinematics. Data were sampled at 20,000 Hz and filtered using a 4-pole Butterworth low pass filter with a cutoff frequency of 1650 Hz (CFC 1000) for accelerometer data and 256 Hz (CFC 155) for ARS data.

STAR Ratings

The STAR equation was originally developed to estimate the incidence of concussion that a college football player may experience while wearing a given helmet over the course of one season [1]. Common real-world impacts are simulated using laboratory testing, with the resulting concussion risk for each impact estimated and then weighted based on the relative frequency that a player might experience that impact scenario during a season of play (termed "exposure"). The STAR value is found by multiplying the predicted on-field exposure (E) at each impact location (L) and velocity (V) by the risk of concussion (R) for that impact. Concussion risk is modeled using the peak resultant linear acceleration (a) and rotational acceleration (α) from laboratory impacts (Equation 1 – Equestrian STAR).

$$STAR = \sum_{L=1}^{3}\sum_{V=1}^{2} E(L,V) * R(a,\alpha) \quad \text{(Eq. 1)}$$

Concussion risk was modeled using a multivariate logistic regression analysis of biomechanical data obtained from instrumented football players who sustained diagnosed concussions [3]. The associated risk function combines peak linear head acceleration (PLA) and peak rotational head acceleration (PRA), which are both known to be associated with brain injury, into a single metric (Equation 2) [4].

$$R(a, \alpha) = \frac{1}{1 + e^{-(-10.2 + 0.0433*a + 0.000873*\alpha - 0.00000092*a\alpha)}} \quad \text{(Eq. 2)}$$

Based on the raw absolute risk data, weighting factors (Table 2) were then assigned so that each location is more equally represented. The impact speeds were also considered with the lower energy impacts weighted higher based on real world injury scenarios [5]. The data was then normalized so that the performance of the average helmet from the initial testing batch of 40 helmets would have a STAR score of 1.0. A multiplier of 5 was then applied to spread the STAR scores to facilitate interpretation of the relative performance of different helmet models. The STAR score thus represents the average number of concussions a rider would likely get using that helmet model if exposed to 30 impacts (6 test configurations x 5) modeled in the lab.

Table 2: Exposure values used for each configuration, which were used as weightings to obtain a STAR Score for each helmet model.

Location	4.0 m/s	6.3 m/s
Back	18.0	0.48
Front	148.9	1.0
Side	4.1	0.42

The range of final STAR scores across helmets were then distributed into a discrete number of stars (1 to 5) for consumer interpretation (Table 3). Lower scores (i.e. lower concussion risk) are associated with a higher rating and representative of better performing helmets.

The Equestrian STAR rating thresholds were determined through the average STAR score across all tested helmets. A 5-star rating threshold was set to a 50% reduction in STAR score relative to the mean STAR score for equestrian helmets, and then each subsequent rating threshold was set in increments of 50% more risk from the 5-star threshold (Table 3). For example, the 4-star threshold is 1.5 times the 5-star threshold, and the 3-star threshold is two times the 5-star threshold.

Table 3: Thresholds to match STAR values to number of stars in a 5-star rating scale.

STAR Score	Number of Stars
< 2.50	5
< 3.75	4
< 5.00	3
< 6.25	2
< 7.50	1
> 7.50	0

References

[1] B. Rowson, S. Rowson, and S. M. Duma, "Hockey STAR: a methodology for assessing the biomechanical performance of hockey helmets," Annals of biomedical engineering, vol. 43, issue, 10, pp. 2429-2443, 2015.

[2] E. J. Pellman, D. C. Viano, C. Withnall, N. Shewchenko, C. A. Bir, and P. D. Halstead, "Concussion in professional football: helmet testing to assess impact performance--part 11," Neurosurgery, vol. 58, pp. 78-96; discussion 78-96, Jan 2006.

[3] S. Rowson and S. M. Duma, “Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration,” Annals of biomedical engineering, vol. 41, issue 5, pp. 873-882, 2013.

[4] A. K. Ommaya, "Biomechanics of Head Injuries: Experimental Aspects," in Biomechanics of Trauma, A. N. a. J. W. Melvin, Ed., ed Eat Norwalk, CT: Appleton-Century-Crofts, 1985.

[5] T. Connor, J. Clark, M. Stewart, A. Annaidh, Post-accident evidence basis for new equestrian standards: Relationship between helmet liner residual crush and accident parameters. Applications of Engineering Science vol. 6, 10004. 2021