Impact Performance of Equestrian Helmets in Oblique Impacts.

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I. INTRODUCTION

Many equestrian falls can result in oblique head impacts that have both vertical and horizontal velocity components. Current certification standards for equestrian helmets, however, do not evaluate helmet performance in oblique impacts. The Fédération Equestre Internationale (FEI) has proposed that equestrian helmets worn during competitions be tested in oblique impacts with a 2.2-m drop height onto a 45° anvil using the latest headform [1]. They have also proposed acceptance criteria for peak linear acceleration (PLA) of ≤150 g and peak angular acceleration (PAA) of ≤5500 rad/s^2 [1]. Concussion risk functions have been developed using PLA and PAA data collected from American football players wearing the Head Impact Telemetry System (HITS) [2] and separately using maximum principal strain (MPS) in a finite element model of the brain calculated from reconstructions of American football impacts [3]. These risk functions have been used in prior studies to evaluate equestrian helmets [4–5]. Our goal here was to evaluate common equestrian helmets using the proposed FEI test specifications and to compute the risk of concussion from the test results using both risk functions.

II. METHODS

Seven make/models of equestrian helmets (sizes 57-60 cm) that are currently on the market were tested on a EN 17950 headform (size 570 mm, Humanetics, Farmington Hills, MI, USA). The helmets had either an acrylonitrile butadiene styrene or polycarbonate shell and an expanded polystyrene liner. The brow of each helmet was positioned 16 mm above the headform's reference plane. When present, the ring-fit system was tightened until slight resistance was met, and when necessary, the helmet's visor was trimmed to prevent contact with the test assembly during impacts. The chin strap was adjusted so that 1-2 fingers fit comfortably between the buckle and headform. The headform and helmet were placed on a U-shaped freefalling trolley that fell past a 45° anvil covered with 80-grit sandpaper. The headform was oriented to produce rotations about the X, Y, or Z-axis at 6.6 m/s. Each helmet experienced at most three impacts (one about each rotation axis) separated by at least 135 mm. For X-axis rotations (X-rot), the headform was inverted (X- and Y-axes ±1° of horizontal); for Y-axis rotations (Y-rot), the headform was rotated 45° (+Y rotation); and for Z-axis rotations (Z-rot), the headform was rotated 65° (-Y rotation) (Fig. 1a). A cantilevered arm holding the headform during freefall released prior to impact and a slack tether attached to the base of the headform prevented secondary head impacts. Six degree-of-freedom headform kinematics were captured by three 500g linear accelerometers, three 8k deg/s angular rate sensors (DTS, Seal Beach, CA, USA), and an on-board data acquisition system (mg-sensor, Farmington Hills, MI, USA). PLA and peak angular velocity (PAV) were extracted directly from the filtered data (acceleration: CFC 1000; angular rate: CFC 180). PAA was calculated from PAV on a component basis. All channels were sampled at 20 kHz and high-speed video was recorded at 1 kHz (Mikrotron, MotionBLITZ EoSens mini2, Gilching, Germany). For each impact, concussion risk was computed from Virginia Tech's combined risk function [2]. In addition, MPS 100th-percentile nodal average element values were calculated with the KTH Royal Institute of Technology head model [6] and concussion risk was determined from Hybrid III reconstructions of American football impacts [3]. For each helmet make/model, concussion risks were averaged for each impact location and then averaged across impact locations.

III. INITIAL FINDINGS

Across all 42 impacts, the impact speeds were 6.54 ± 0.04 m/s, PLAs were 154.7 ± 16.1g, PAAs were 4.017 ± 1.289 krad/s^2, and PAVs were 19.5 ± 3.8 rad/s (Fig.1 c and d). All helmets exceeded the 150g PLA threshold at one or more impact locations. Two helmets exceeded the 5500 rad/s^2 threshold at two impact locations. Averaged across X, Y, and Z rotations, concussion risk varied from 25.2% to 85.2% using the Virginia Tech method and from 12.1% and 40.4% using the KTH method (Figure 1b).

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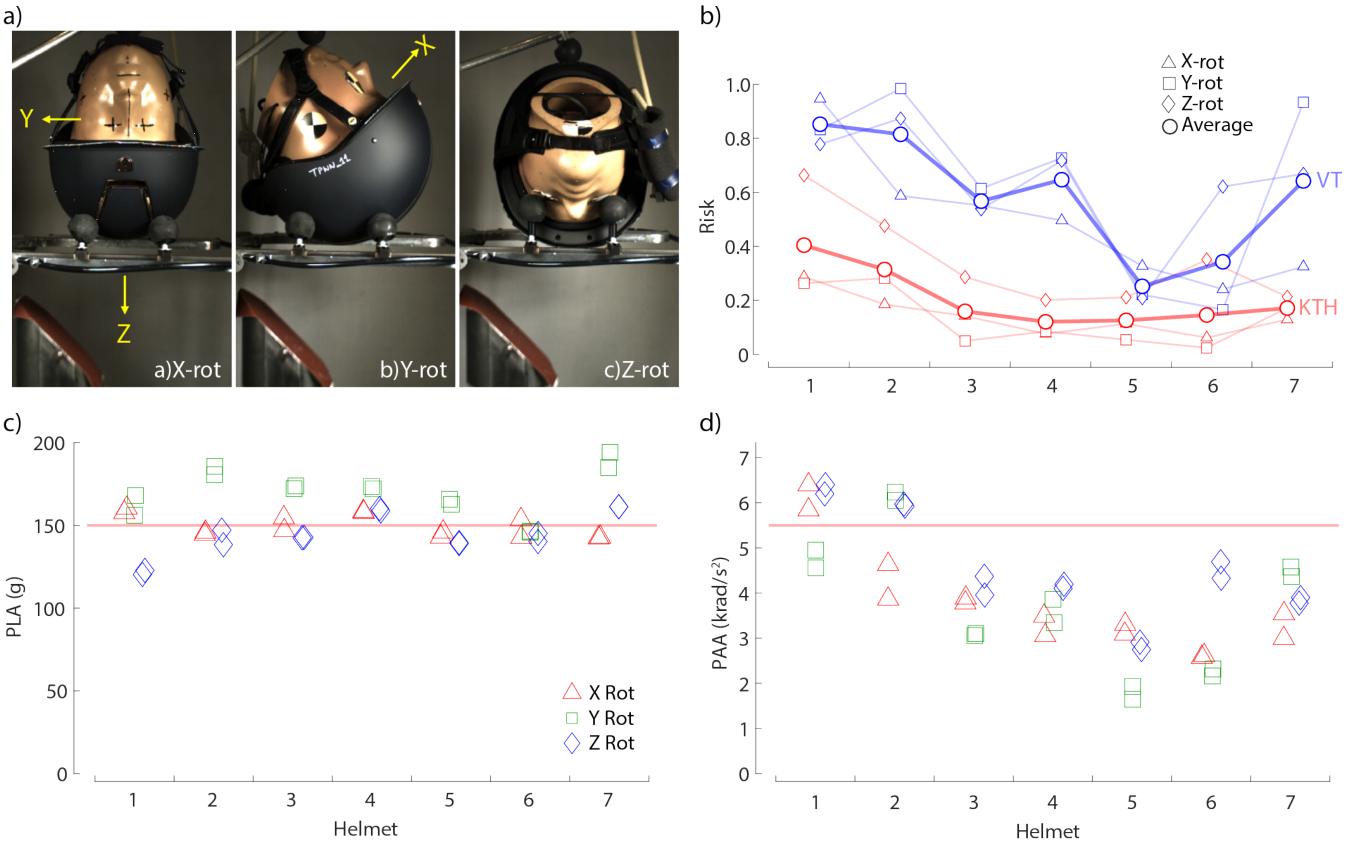

Fig. 1. (a) Headform orientation relative to the oblique anvil; (b) VT (blue) and KTH (red) risks averaged for each impact location (thin lines) and averaged across all impact locations (thick lines); (c) PLA and (d) PAA for each impact location. The horizontal red lines represent the proposed FEI thresholds (i.e., 150g and 5.5 krad/s^2).

IV. DISCUSSION

Our goal was to evaluate common equestrian helmets against the proposed FEI specifications and to compute the concussion risk metrics using two available methods. Overall, we found that the FEI's proposed PLA threshold was exceeded for all the helmet make/models we tested, whereas the FEI's proposed PAA threshold was exceeded for only two of the helmets. Concussion risks computed using the VT and KTH methods were markedly different for most of the helmets (Fig. 1b). The VT risk calculation relies on peak kinematics, which likely explains the similar patterns present in the PAA data and VT risk data in Figs 1d and 1b, respectively. The KTH risk calculation relies on brain strain, which is highly correlated (r^2=0.92) to angular velocity (not shown here). The VT risk function is based on data collected with HITS, which has known errors in measuring impact direction and magnitude [7], and the KTH risk function is based on Hybrid III reconstructions of American football impacts. The differences in risk predicted by these two methods suggest more work is needed to understand how these two methods relate to one another and which method correlates best with real-world concussive injuries.

The FEI's proposed test criteria aim to reduce concussion risk in riders by encouraging improved helmets that better attenuate peak head kinematics during an impact [1]. Based on our findings, these proposed criteria may encourage helmet manufactures to focus on reducing PLA rather than PAA, although reducing PLA will likely act to also reduce PAA. Further improvements may be possible by including direction-dependent thresholds.

The study is limited to seven different helmet designs across a broad price range. Helmets in this study struck a rigid surface, whereas the most common impact surface in equestrian-related falls is a softer compliant surface [8]. Future work should evaluate equestrian helmet performance during oblique impacts against surfaces encountered during equestrian-related falls.

V. REFERENCES

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