Towards a Representative Test Method for Rugby Knee Strikes

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I. INTRODUCTION

As of 2018, traumatic brain injury (TBI) was defined as the leading cause of death in young adults worldwide [1]. The risk of sustaining a TBI is heightened by participating in sports that expose individuals to deliberate contact events, such as Rugby Union. In Rugby Union, contact mechanisms account for 72% of all injuries [2], with tackling the leading cause of concussion [3]. Concussion is the most common injury in the sport [4], with a reported incidence rate of 21 per 1,000 match hours at the elite level, accounting for almost a quarter of all match injuries on average across a season [3-4]. Head-to-head, head-to-ground and knee-to-head impacts are the most frequent causes of concussion in the sport [4]. However, current World Rugby personal protective equipment testing only represents head-to-ground impacts and uses guided linear drop apparatus to generate impacts with energies of 13.8 J and resultant head accelerations of at least 200 g [5]. These impact tests are not representative of in-situ collisions, such as knee-to-head strikes, during a rugby match and do not allow for the influence of the neck to be represented. A previous study by Halkon *et al.* [6] developed a test methodology aimed at collecting data from laboratory-simulated knee-on-thigh and elbow-on-torso strikes in elite-level basketball, however the impactor had limited biofidelity. This work describes the initial development of a laboratory-based methodology for representing knee-to-head strikes in rugby with a similar but more biofidelic pendulum impactor and instrumentation to accurately predict impact location and magnitude.

II. METHODS

A bespoke impactor (knee surrogate) was designed and manufactured for this study. The impactor was constructed from ABS and silicone (Shore D hardness of 80-100 and Shore A hardness of 20) to represent bone and soft tissue (Shore D hardness of 80 [7] and Shore A hardness of 20-30 [8]), respectively. The impactor has a total mass of 2.8 kg, in line with the effective mass of a typical knee [9], and an end effector radius of 37.5 mm, representative of the patella at 90˚ flexion. A pendulum impact was preferred to a guided free fall as it enables controlled impact loading with the benefit of loading the headform whilst being constrained by a neck. The future inclusion of a more biofidelic neck will also allow the method to be used to investigate the effects of neck stiffness and loading direction with respect to the risk of TBI. The surrogate knee impactor was suspended from an A-frame using lightweight, inextensible strings; a schematic representation is shown in Fig. 1.

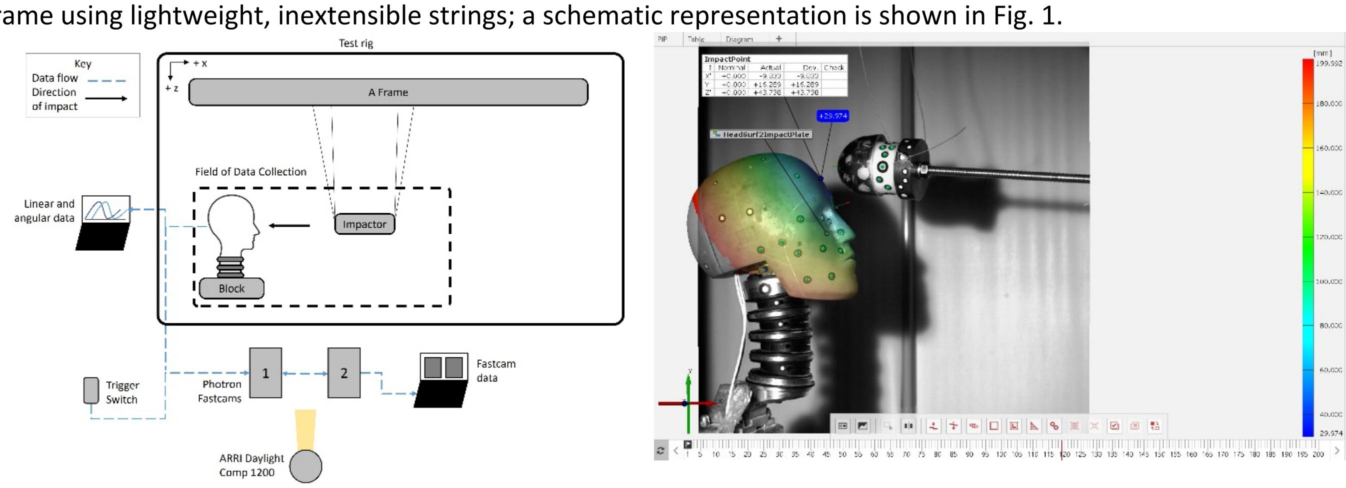

Fig. 1. Schematic of test setup (left) and impact location analysis (right).

At rest, the impactor was in contact with the Hybrid-III headform in the target impact locations, determined from epidemiology studies [10]. In total, 45 trials (three orientations at three different velocities, with five repeats of each impact condition) were conducted under ambient conditions. Impact location and velocity of the impactor were tracked throughout the duration of the impact using a combination of high-speed video capture (4000 fps,

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1/10000 s) and a 3D scanning system (GOM Atos Core structured light 3D scanner [11]). High-speed video images were imported into a bespoke GOM template (Fig. 1 (right)) to track the location of the pendulum impactor relative to the headform and to extract the exact coordinates of the impact site for all trials. The headform was instrumented with a tri-axial linear accelerometer and angular rate sensor mounted at the centre of gravity (6DXPRO 2K-18K, ±2000 g, ±18000 deg/s, 10 kHz sampling rate) to capture the skull kinematics. Data were acquired with DTS SLICEWare and filtered using a standard CFC 1000 filter [12]. The headform was attached to a Hybrid-III surrogate neck and rigidly mounted to an adjustable rig that allowed for the position of the head-neck complex to be adjusted in the x, y and z axes to achieve an accurate and repeatable impact location.

III. INITIAL FINDINGS

A visual representation of the target impact locations, the mean actual locations and their standard deviations is shown in Fig. 2.

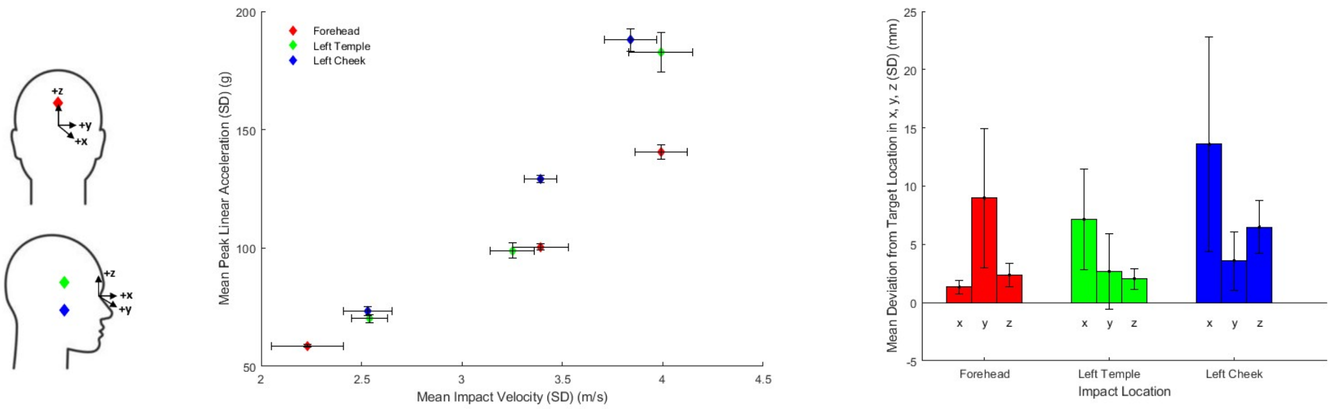

Fig. 2. Target impact locations (left), mean peak linear acceleration against mean impact velocity (middle) and mean deviation from target location in x, y, z (right).

IV. DISCUSSION

The research has introduced a test method that can be used to investigate the risk of TBI (e.g. concussion) when an athlete is subjected to knee-to-head strikes, commonly seen in rugby and other sports. The maximum deviations from the target location in the x, y and z axes were 28.09* mm, 17.15 mm and 8.6* mm, respectively (*maximum recorded at highest velocity). The maximum deviations for peak linear acceleration and impact velocity were 8.31 g and 0.18 m/s, respectively. The largest deviations in impact location were consistently left and right of the target location, i.e. in the x axis for the left temple and cheek and in the y axis for the forehead location. The manual release of the impactor is believed to be the main source of error for the deviation in impact location. There was difficulty with raising the impactor to the height necessary for the highest impact velocity to be achieved due to the spatial constraints of the A-frame and the requirement of the test operator to maintain this position. This is likely to have induced a parallax error, resulting in the impactor being released at an angle as opposed to horizontally, as intended. Future improvements to the method will include an automated release mechanism to improve both the accuracy of the impact location and impact velocity.

Based on other literature, the threshold for a concussion is dependent on both impact duration and head rotation, in addition to PLA. This was beyond the scope of this study and as a result, more data needs to be conducted in order to define concussion and skull fracture probabilities of these impacts.

The method has useful utility in validating the predictive nature of current and future sensor technologies that aim to predict the magnitude and location of impacts. Further, the introduction of a neck with improved biofidelity and tunable properties will enable the investigation of neck stiffness effects on the head response during such an impact.

V. REFERENCES

[1] Trotta, *et al*., *J Biomech*, 2018.
[2] Brooks, *et al*., *Br J Sports Med*, 2005.
[3] Gregory, *et al*., *J Sci Med Sport*, 2018.
[4] Cross, *et al*., *Br J Sports Med*, 2017.
[5] World Rugby, Handbook, 2015.
[6] Halkon, *et al*., *Procedia Eng*, 2014.
[7] Kannan, *et al., JETIR,* 2020.
[8] Falanga, *et al., JAAD,* 1993.
[9] Osterhoff, *et al*., *Knee*, 2011.
[10] McIntosh, *et al., Med Sci Sports Exerc,* 2000.
[11] https://www.gom.com/en/products/3d-scanning
[12] SAE, J211, 1995.