

Comparison of the Hybrid III Head and Neck to a Detailed Head and Neck Finite Element Model with Active Musculature, in a Football Impact Scenario

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I. INTRODUCTION

Further improvements to the design of football helmets may be realised by experimental testing supported by advanced computational modelling. Helmet performance is assessed experimentally using the National Operating Committee on Standards for Athletic Equipment (NOCSAE) Drop Test, and the Linear Impact (LI) test [1]. Although the drop test is commonly used, it does not incorporate the compliance of the neck (the headform is rigidly mounted to a travelling carriage). In contrast, the LI test incorporates a Hybrid III (HIII) head and neck that is impacted with a linear ram. The ram has an end cap and foam pad to represent an impacting football helmet [1]. The HIII head and neck is mounted to a sliding carriage (Fig. 1), and is free to move in the x-direction. Football helmet design has recently started to address head rotation, which has been proven to be an important consideration in improved impact protection [2].

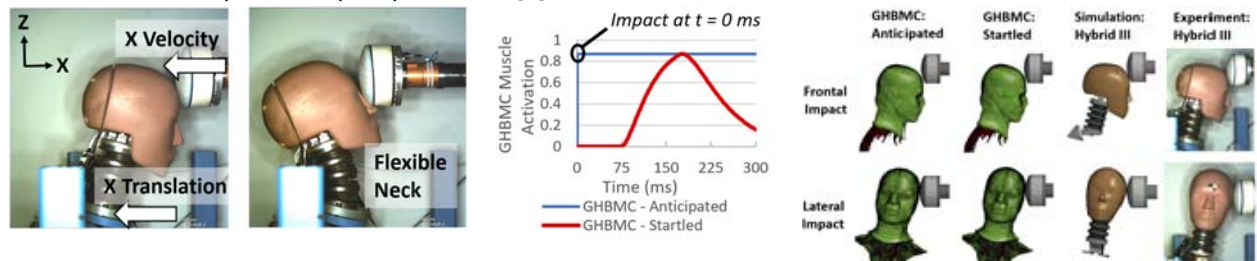


Fig. 1. Hybrid III Linear Impact (LI) Test, Front Configuration

Fig. 2. GHBMC Muscle Activation Schemes.

Fig. 3. Test Matrix showing all eight Impact configurations.

Helmet design may be enabled with computational models, and in that case, it may be desirable to use a detailed human model. For this purpose, it is beneficial to compare a detailed head and neck Finite Element (FE) model to a widely used experimental standard, beginning with bare head impacts to investigate the effect of active neck musculature on head response. This study compares the head kinematics of a LI experiment to the Global Human Body Models Consortium (GHBMC) head and neck model [3-4] in the same configuration.

II. METHODS

The HIII head and neck model was positioned to be consistent with the orientation in the experimental test. Similarly, the GHBMC model was repositioned to be consistent with the HIII experiment by rotating the neck 15° forward from the standard seated position, and rotating the head 4.5° in extension. The first thoracic vertebra was constrained to translate in the x-direction with a point mass corresponding to the mass of the carriage. The mass of the HIII and GHBMC head and neck models were within 0.05% of the experimental mass. A total carriage, head and neck mass of 23.1 kg and an impactor mass of 15.4 kg were used in all cases; these masses imitate a player's kinematic properties after an impact to the head [1]. The head centre of gravity was used as a reference point to align the impactor, to account for the differing head shapes of GHBMC and the HIII. All models were analysed using a commercial FE solver (LS-DYNA R7.1.2).

Active musculature in the GHBMC neck is achieved using Hill-type muscle elements in which the force is proportional to the muscle activation level. Two neck muscle activation schemes were applied to the GHBMC model (Fig 2): a delayed onset of muscle activation representing a startle response (activation starting 75 ms after impact [5]) and an anticipated impact where the muscles were fully activated at the time of impact, as peak muscle activation in human subjects occurs slightly before an anticipated impact to the head [6]. In this study, the computational models were compared to the LI experiment, as outlined in the test matrix (Fig. 3). The

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impact speed was held constant at 5.5 m/s; which is commonly used in helmet testing [7].

III. INITIAL FINDINGS

Considering a bare head impact, the GHBM head and neck model exhibited a similar acceleration-time response to the HIII model and experiment in a Frontal impact (Fig. 4) and a Lateral impact (Fig. 5). Linear and angular acceleration peaks for the FE models were compared to the experiment (Table 1).

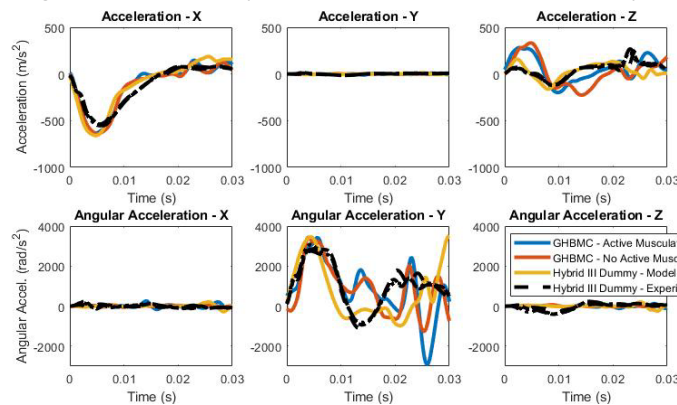


Fig. 4. Frontal Impact, Linear & Angular Acceleration

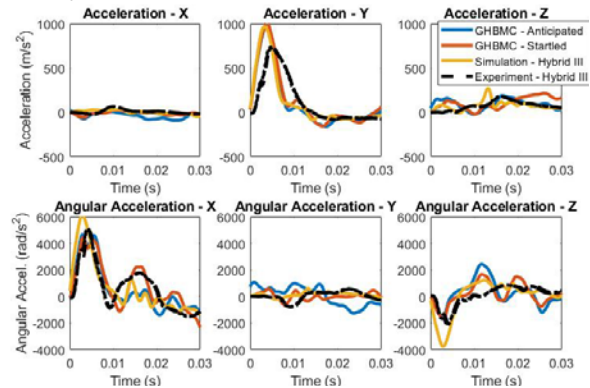


Fig. 5. Lateral Case, Linear & Angular Acceleration

TABLE 1 - PEAK RATIOS FOR PRIMARY ACCELERATION PEAKS

Peak Ratio, Acceleration	Simulation: Hybrid III	Frontal			Lateral	
		GHBM: Anticipated	GHBM: Startled	Simulation: Hybrid III	GHBM: Anticipated	GHBM: Startled
<i>X Linear</i>	1.2	1.17	1.15	1.32	1.38	1.38
<i>Y Angular</i>	1.19	1.19	1.18	1.2	0.87	0.92

IV. DISCUSSION

For all models, the peak values, timing and shape of the acceleration traces were in good agreement with the experiments. The timing of the model acceleration peaks was within 1 ms of the measured experimental peaks. Importantly, the primary acceleration peaks occurred very early in the impact scenario, at approximately 4 ms following impactor contact with the bare head, and thus the translational (~4mm) and rotational (~2°) displacements were small at that time. The very short timeframe and small displacements suggest that the primary acceleration peaks were governed by the inertial properties of the head and the stiffness properties of the impactor, and that the bending stiffness of the neck played a lesser role in the early kinematics. However, the neck musculature did have an effect on linear and angular displacements later in time. The later time responses for the HIII and GHBM model with early muscle activation were similar, while the startled response model (75 ms activation time) exhibited larger displacements, especially at times great than 50 ms after the onset of the impact. Future research will investigate the importance of muscle activation for helmeted impacts, where the maximum accelerations will be smaller and occur later in time such that active musculature is expected to play a stronger role in head response.

The results of this study demonstrate similar kinematics between models and experiments, which is a positive outcome in terms of using detailed human models to investigate head impact, and more importantly is a first step towards the assessment and optimisation of head protection using detailed helmet FE models.

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