Influence of Initial Head Position on Concussion Risk in Rear-End Vehicle Collisions

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Abstract Low- to moderate-speed rear-end motor vehicle collisions are a frequent cause of concussion, yet research has traditionally focused on whiplash-associated disorders and neck injury. Concussion mechanisms differ from whiplash, as injury risk may arise from torso and head interaction with the seatback and head restraint during body-first loading. This study investigated the influence of head position on head kinematics and brain strain using a 5th percentile Hybrid III headform representative of an adult female. Rear-impact reconstructions were performed using a rail-guided launcher at 5.5 m/s and 7.0 m/s. Two head-restraint positions were evaluated: direct contact, 0 cm, and offset, 7.6 cm. Head kinematics were recorded using a SLICE NANO array, and Maximum Principal Strain (MPS) was calculated using the University College Dublin Brain Trauma Model v2.0. Offset condition produced higher peak linear acceleration than direct contact at 5.5 m/s, 29.3 g versus 25.8 g, and 7.0 m/s, 40.8 g versus 38.1 g. However, offset reduced peak rotational acceleration at both velocities, 2237.9 versus 2489.0 rad/s^2 and 3021.9 versus 4413.1 rad/s^2. All conditions exceeded MPS values commonly reported as thresholds associated with a 50% probability of concussion, indicating that moderate-speed rear impacts pose a risk of concussion for braced female occupants.

Keywords Biomechanics, Concussion, Female occupant, Maximum Principal Strain, Rear-end vehicle collision.

I. INTRODUCTION

Motor vehicle collisions (MVCs) are the third leading cause of traumatic brain injury (TBI), with rear-end impacts accounting for a significant portion of these incidents [1]. While safety research has traditionally focused on whiplash-associated disorders (WAD) and neck injuries, clinical data reveal a significant incidence of concussion in low- to moderate-speed collisions [2]. Data from the Canadian Concussion Centre have identified that females are disproportionately affected, representing 69.1% of patients treated for concussions with persisting symptoms (C+PCS) following MVCs, with rear-end collisions being the most frequent injury mechanism [2].

A rear-end vehicle collision is defined as a type of impact occurring when a trailing car collides with the rear end of a leading or struck vehicle [3]. This study focuses on the driver of the struck vehicle, where the seatback is accelerated into the driver's torso before their head contacts the head restraint [3]. Previous research on body-first head impacts has implications for investigating head position in rear-end collisions [4]. A body-first head impact occurs when the body contacts a surface prior to head contact. In prior reconstructions of rear falls, it was found that accounting for a natural offset (where the upper back strikes the surface before the head) resulted in an increase in peak rotational acceleration and predicted concussion risk compared to simultaneous head-and-back impacts [4]. Real-world seating postures are highly variable among occupants, often creating an initial offset between the head and the headrest, therefore understanding how head position influences concussion risk is important [5]. This study evaluates how a realistic head-to-headrest offset contributes to tissue-level brain strain in moderate-speed rear-end impacts, providing the data necessary to inform the design of safer vehicle safety systems.

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II. METHODS

Experimental Design

Impact reconstructions were performed using a rail-guided system [6]. A production SUV seat was securely fastened to the end of the impactor to replicate real-world seatback and head-restraint interactions (Fig. 1). Impacts were conducted at two velocities: 5.5 m/s (20 km/h) and 7.0 m/s (25 km/h). These impact velocities were chosen to represent moderate-speed rear-end collisions [3]. A total of 11 valid trials were included in the analysis, with the number of valid trials for each velocity and head-position condition summarized in Table I. One 7.0 m/s direct trial was excluded due to a data acquisition system error, resulting in two valid trials for this condition.

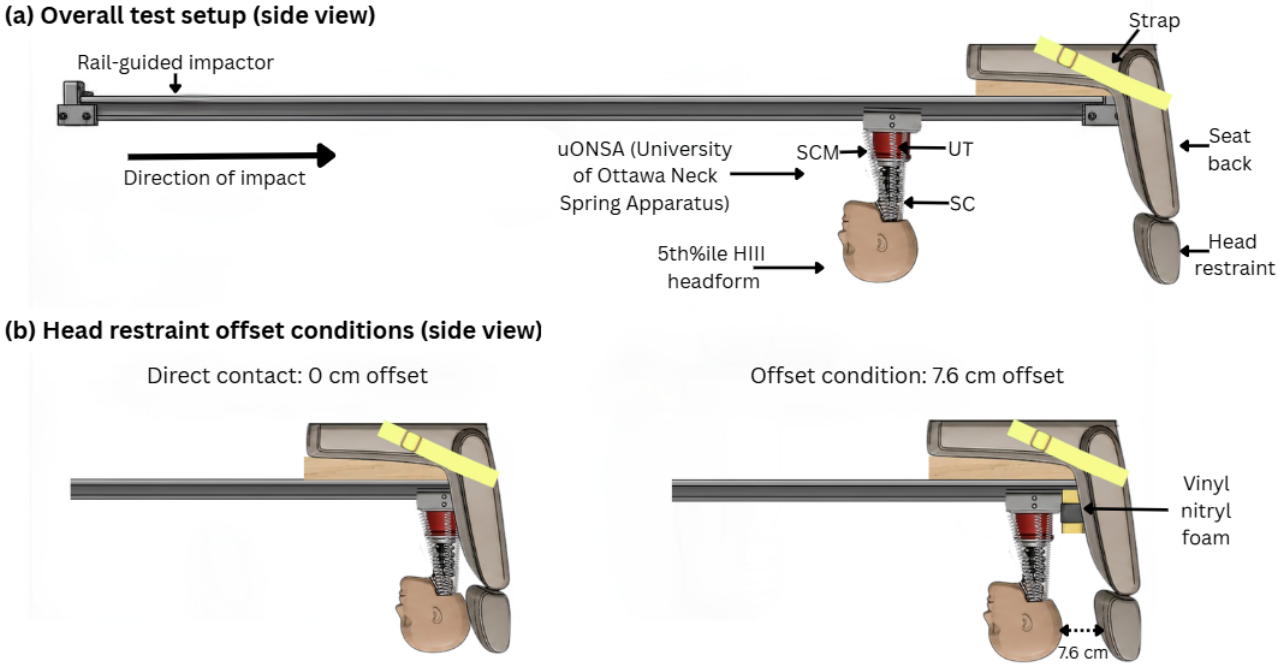

Fig. 1. Experimental setup: (a) Side-view schematic of the rail-guided impactor, inverted SUV seat, head restraint, 5th percentile Hybrid III headform, and University of Ottawa Neck Spring Apparatus. The arrow indicates the direction of impactor motion. (b) Initial head-restraint positions tested: direct contact, 0 cm offset, and 7.6 cm offset. Offset was measured horizontally from the posterior surface of the headform to the anterior surface of the head restraint at the approximate level of initial head contact.

TABLE I
TEST MATRIX

Velocity	*Head position*	*Number of trials*
5.5 m/s	Direct (0 cm)	3
5.5 m/s	Offset (7.6 cm)	3
7 m/s	Direct (0 cm)	2
7 m/s	Offset (7.6 cm)	3

Occupant Surrogate and Neck Stiffness

To represent adult females, a 5th percentile Hybrid III (HIII) headform with a mass of 3.73 kg was used (Fig. 1). The headform was mounted on the University of Ottawa Neck Spring Apparatus (uONSA), which consists of an unbiased neckform augmented with external spring elements [7]. The uONSA was selected instead of the standard Hybrid III neck because this study focused on concussion-related head kinematics and finite element-derived brain strain, rather than whiplash-associated neck injury. The Hybrid III neck is commonly used in automotive testing and is primarily characterized for sagittal-plane flexion-extension responses relevant to neck injury [7]. However, concussion risk depends on resultant head motion and brain deformation, which may be influenced by linear and rotational kinematics across multiple anatomical planes [7]. The unbiased neckform was used to reduce directional bias and allow the headform to respond more freely during dynamic loading. The apparatus incorporated three sets of springs representing the primary muscles responsible for head stabilisation: the sternocleidomastoid (SCM); upper trapezius (UT); and splenius capitis (SC) [8]. These muscle groups were selected due to their dominant roles in head motion and stabilisation during impact [8]. The SCM contributes primarily to neck flexion, the SC to neck extension, and the UT to lateral flexion and stabilisation [8]. Spring forces were set to 100% maximum voluntary contraction to represent a braced/aware occupant condition. This force level was determined from peak isometric neck-muscle force measurements collected from female participants, with electromyography (EMG) used to confirm muscle activation during the contractions (unpublished data in preparation). Therefore, the results should be interpreted as representing rear-impact reconstructions with an active/braced neck configuration, rather than a passive or unaware occupant response. A prospective cohort study found that approximately one-quarter of occupants reported being aware of the impending collision, suggesting that this condition is not the dominant scenario but is a real-world exposure [9].

Two initial head positions were tested: a direct condition (0 cm offset), representing a driving position where the head is in contact with the head restraint; and an offset condition (7.6 cm), representing a driving posture in which the head is positioned slightly forward of the restraint [10]. The 7.6 cm offset was measured as the horizontal distance between the posterior surface of the headform and the anterior surface of the head restraint at the approximate height of the headform center of gravity. The offset of 7.6 cm aligns with the guidelines from the Canadian Centre for Occupational Health and Safety, which recommends that the head be positioned as close as possible to the head restraint, with an acceptable backset of approximately 7 cm and not exceeding 10 cm [10]. The offset in reconstructions was created using a piece of vinyl nitrile foam that represented the compliance of the upper back based on previous research [11], [12].

Data Acquisition and Brain Modelling

High-speed video was recorded for each impact and was used to determine impact velocity. Head kinematics were recorded at 20 kHz using a DTS SLICE NANO data acquisition system, consisting of a triaxial accelerometer and three angular rate sensors mounted at the headform centre of gravity [6]. The coordinate system followed the headform-mounted sensor axes. Positive x corresponded to the anterior direction of the headform, positive y corresponded to the left lateral direction, and positive z corresponded to the superior direction. Rotational velocities and accelerations followed the right-hand rule about each corresponding axis. Linear acceleration and angular velocity time histories were recorded about the x-, y-, and z-axes of the headform. All kinematic signals were filtered using a CFC 180 filter. Angular acceleration was calculated from the filtered angular velocity time histories. Peak resultant linear acceleration, angular velocity, and angular acceleration were calculated as the vector magnitudes of their respective x-, y-, and z-components. The filtered linear acceleration and angular acceleration time histories were then directly applied to the head centre of gravity of the University College Dublin Brain Trauma Model v2.0 (UCDBTM v2.0). Each simulation was run for the duration of the experimental acceleration pulse, with the model output used to calculate peak global Maximum Principal Strain (MPS).

The UCDBTM v2.0 is a three-dimensional finite element model of the human head developed to estimate tissue-level deformation during head impact [12], [13]. The model is an updated version of the original UCDBTM and contains 184,261 hexahedral elements, representing the scalp, skull, cerebrospinal fluid, pia, dura, falx, tentorium, cerebral grey and white matter, cerebellum, brainstem, and ventricles [12], [13]. The UCDBTM v2.0 has been validated against two sets of cadaveric impact data, including cadaveric drop-test acceleration responses and neutral density target brain displacement measurements, supporting its use for estimating impact-induced brain deformation [12], [13].

To approximate the 5th percentile female headform used, the UCDBTM v2.0 geometry was uniformly scaled to 95% of its original dimensions. A single global scale factor was applied to all model components, rather than independently scaling individual anatomical regions. The 95% scale factor was selected to approximate the external head dimensions of the 5th percentile female surrogate. Material properties were not scaled or modified for sex. The original UCDBTM v2.0 material definitions were retained because subject-specific female material properties were not available, and modifying material properties without validation would introduce additional uncertainty. Therefore, the scaled model should be interpreted as a geometrically scaled representation of an adult female head, not a sex-specific finite element model.

Statistical Analysis

Statistical analyses were performed using an exploratory two-way factorial approach to evaluate the effects of impact velocity, initial head position, and their interaction on each outcome measure. Separate models were conducted for peak resultant linear velocity, peak resultant linear acceleration, peak resultant rotational acceleration, and maximum principal strain. The independent variables were impact velocity and head position, with an interaction term included to assess whether the effect of head position differed by impact velocity. Due to the small and unequal number of repeated trials per condition, statistical analyses were considered exploratory. Pairwise comparisons between direct and offset head positions were also performed within each velocity using Welch's independent-samples t-tests, in which statistical significance was set at $\alpha = 0.05$.

III. RESULTS

Kinematic Responses

Average peak kinematic responses for each condition are summarised in Table II. At an impact velocity of 5.5 m/s, the offset condition produced a lower response in peak rotational acceleration compared to direct head restraint contact (2237.9 rad/s² vs 2489.0 rad/s²). Peak linear acceleration increased slightly in the offset condition (29.3 g vs 25.8 g), while MPS values were similar between conditions (27.0% offset vs 26.1% direct). At an impact velocity of 7.0 m/s, a larger reduction in peak rotational acceleration was observed with head restraint offset (3021.9 rad/s² vs 4413.1 rad/s²). Peak linear acceleration was slightly higher in the offset condition (40.8 g vs 38.1 g), as well as MPS values (34.7% vs 33.7%). Exploratory two-way analyses indicated that impact velocity had a significant effect on all measured outcomes, including peak linear velocity, peak linear acceleration, peak rotational acceleration, and MPS (all $p \leq 0.039$). Head position had a significant main effect only for peak linear acceleration ($p = 0.013$), while a significant velocity-by-position interaction was observed for peak rotational acceleration ($p = 0.045$). Follow-up Welch's comparisons indicated that the only significant direct-versus-offset pairwise difference occurred for peak linear acceleration at 5.5 m/s ($p = 0.006$). No other direct-versus-offset comparisons reached statistical significance. Given the small and unequal sample sizes, these statistical findings were interpreted as exploratory and used to support descriptive trends rather than definitive causal conclusions.

The duration of the direct impacts was approximately 60 ms, while the offset impacts were of shorter duration, approximately 50 ms (Figs 2–5). Kinematics were captured for the full duration of the impacts. The linear acceleration curves in Figs 2–5 begin at approximately 10 g because the data acquisition system was triggered using a 10 g acceleration threshold. Across both velocities, the offset condition consistently produced lower peak resultant rotational accelerations with higher peak linear accelerations compared to the direct condition.

Neural Tissue Strain (MPS)

Finite element modelling using UCDBTM v2.0 showed that all peak global MPS averages ranged from 26.1% to 34.7% (Table II). The highest strain (34.7%) occurred at 7.0 m/s with a 7.6 cm offset.

TABLE II
AVERAGE PEAK HEAD KINEMATICS AND NEURAL STRAIN (5TH PERCENTILE HIII)

Velocity & Head Position	*Peak Resultant Linear Velocity (m/s) ± SD*	*Peak Resultant Linear Acceleration (g) ± SD*	*Peak Resultant Rotational Acceleration (rad/s^2) ± SD*	*MPS (%) ± SD*
5.5 m/s direct	7.6 (±0.1)	25.8 (±0.6)	2489.0 (±66.4)	26.1 (±0.8)
5.5 m/s offset	7.6 (±0.1)	29.3 (±0.9)	2237.9 (±152.7)	27.0 (±1.6)
7 m/s direct	9.4 (±0.4)	38.1 (±2.0)	4413.1 (±799.5)	32.9 (±0.6)

7 m/s offset	9.3 (±0.1)	40.8 (±1.8)	3021.9 (±405.3)	34.7 (±0.9)

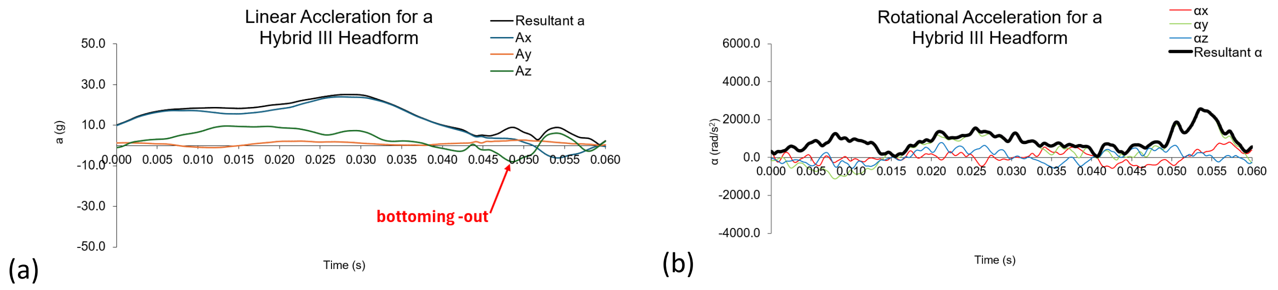

Fig. 2. Kinematic curves for 5.5 m/s, direct (0 cm) impact condition showing a 60 ms duration of (a) linear acceleration and (b) rotational acceleration.

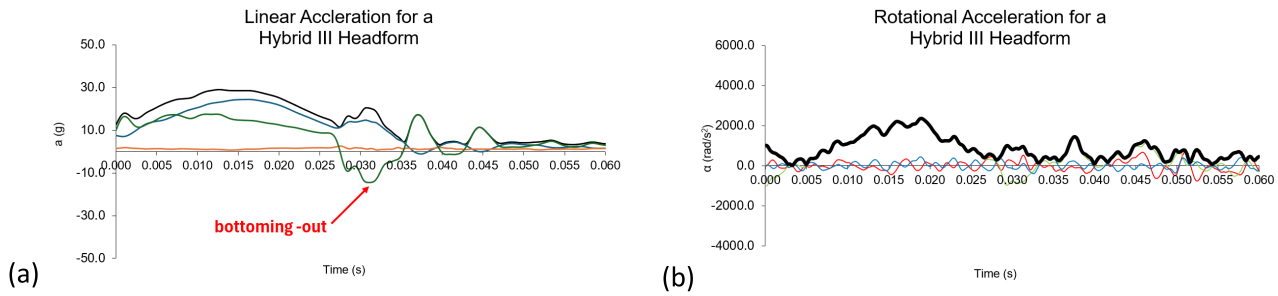

Fig. 3. Kinematic curves for 5.5 m/s, offset (7.6 cm) impact condition showing a 50 ms duration of (a) linear acceleration and (b) rotational acceleration.

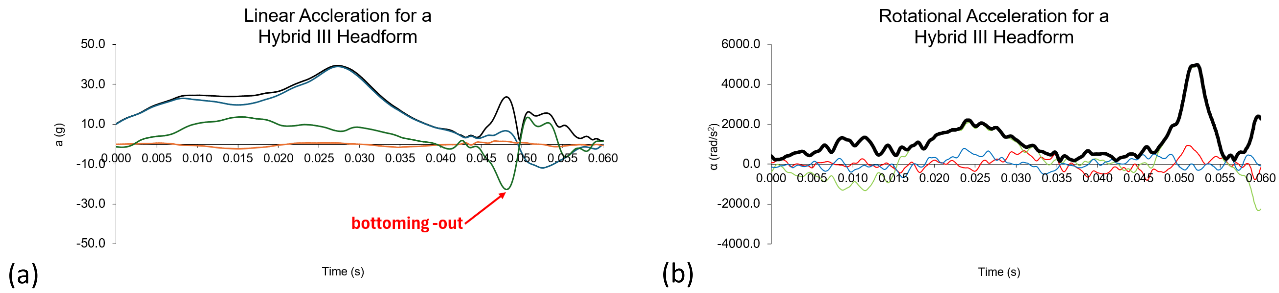

Fig. 4. Kinematic curves for 7.0 m/s, direct (0 cm) impact condition showing a 60 ms duration of (a) linear acceleration and (b) rotational acceleration.

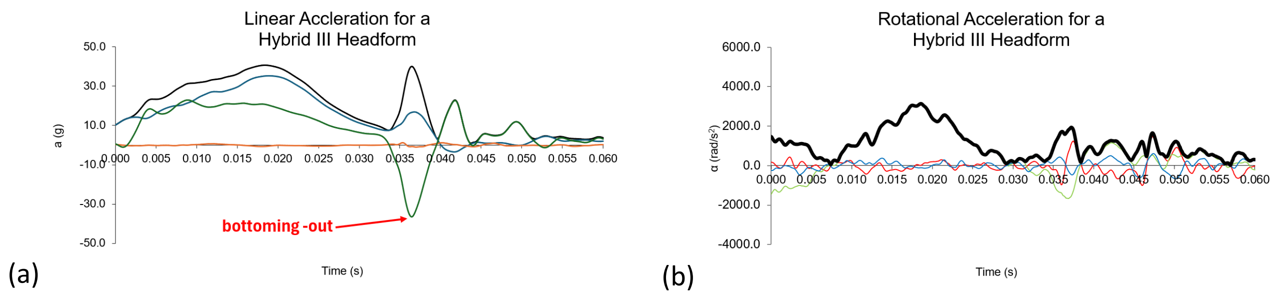

Fig. 5. Kinematic curves for 7.0 m/s, offset (7.6 cm) impact condition showing a 50 ms duration of (a) linear acceleration and (b) rotational acceleration.

IV. DISCUSSION

Finite element modelling using UCDBTM v2 showed that all impacts at 5.5 m/s and 7.0 m/s exceeded 50% risk levels for concussion [14]. The reported peak global MPS values fall within or exceed the 50% probability for concussion, which is generally reported between 19% and 30% MPS in the cerebrum [14].

The findings of this study show that moderate-speed rear-end collisions generate neural tissue strains sufficient to cause concussion in reconstructed events representing braced female occupants. The initial head position influenced the timing of the peak kinematics, but the impact velocity was the most consistently significant factor across the measured outcomes.

Across all conditions, peak head linear velocity exceeded the initial collision-simulated velocity (Table II), indicating that the seat–occupant interaction did not attenuate head motion but instead facilitated momentum transfer to the head–neck system.

Distinct patterns were observed between direct and offset head position conditions across both impact velocities. In the direct condition, linear acceleration curves demonstrated a 'bottoming-out' phase between approximately 45 ms and 50 ms, characterised by a reduction in acceleration in the z-direction. This phase corresponds to the head reaching maximal extension in contact with the head restraint (Figs 2 and 4).

Rotational acceleration curves revealed that the peak resultant rotational acceleration occurred following the 'bottoming-out' phase, during the rebound motion of the head (Figs 2 and 4). For example, at 5.5 m/s in Fig. 2, peak rotational acceleration reached 2561.7 rad/s², occurring shortly after the head reversed direction following restraint contact. A similar pattern was observed at 7.0 m/s, where peak rotational acceleration increased to 4978.4 rad/s² in Fig. 4, again occurring during the rebound phase.

In contrast, the offset condition demonstrated a different temporal relationship. Peak rotational acceleration occurred prior to the 'bottoming-out' phase, before the head contacted the head restraint. At 5.5 m/s, peak rotational acceleration reached 2355.8 rad/s², while at 7.0 m/s it reached 3153.8 rad/s² (Figs 3 and 5). This indicates that rotational motion was initiated earlier in the event, likely due to increased relative motion between the torso and head before restraint was engaged, resulting from the 7.6 cm offset.

These patterns were consistent across both velocity conditions, suggesting that head–restraint offset influences the timing of peak rotational loading, rather than solely its magnitude.

A consistent finding throughout the impact reconstructions was that the offset condition (7.6 cm) resulted in lower peak resultant rotational accelerations compared to the direct contact (0 cm) condition (Table II). This pattern in kinematics suggests that the 7.6 cm offset provides time for the neck springs on the uONSA to engage and stabilise the headform. It is hypothesised that the lower rotational accelerations are a result of the neck springs mitigating the energy transfer during the initial 'body-first' lag phase, thereby reducing the net energy transferred to the headform prior to head-restraint contact. Because the uONSA is a mechanical surrogate, this mechanism should be interpreted as a test-system-specific response rather than a direct representation of in vivo cervical spine mechanics.

The vulnerability of the female population, as represented by the 5th percentile HIII headform, is underscored by the high MPS values recorded (Table II). Factors such as lower neck strength and different effective mass distribution compared to 50th percentile males may exacerbate the kinematics observed in rear impacts [15].

Limitations

This study used a 5th percentile HIII headform mounted on the uONSA to provide a repeatable mechanical representation of a braced female occupant. Although this setup allowed controlled comparison of initial head position, the headform and neck apparatus do not fully capture the complex multidirectional responses of a living human [7], [8]. The uONSA spring configuration represents an active/braced neck condition and should not be interpreted as representative of passive or unaware occupants. As a result, the observed reduction in peak resultant rotational acceleration in the offset condition may be specific to the spring-mediated response of the uONSA and should not be generalized to all rear-end collision scenarios.

The spring forces were set to 100% maximum voluntary contraction based on female neck-muscle force data. This represents a braced or aware occupant condition rather than the full range of muscle activation that may occur in real-world rear-end collisions. Although awareness of an impending collision has been reported in a subset of occupants, this condition does not represent the dominant rear-impact scenario [9]. Future testing

should evaluate passive, intermediate, and active neck stiffness conditions to determine how occupant awareness and muscle activation influence head kinematics and predicted brain strain.
The finite element analysis was performed using the UCDBTM v2.0, which was geometrically scaled to approximate a 5th percentile female headform. A single global scaling factor was applied to all model components, and material properties were not modified to represent female-specific tissue properties. Therefore, the model should be interpreted as a geometrically scaled approximation of an adult female head rather than a validated sex-specific female brain model. The UCDBTM v2.0 is a validated FE model of the head, for which the brain stem has yet to be validated, therefore the analysis of this study was limited to the cerebrum [12], [13].

The number of repeated trials per condition was small and unequal, particularly for the 7.0 m/s direct condition (Table I). Statistical analyses were therefore considered exploratory and were used to support descriptive trends rather than definitive causal conclusions.

Future research should evaluate a broader range of neck stiffness and seating postures.

V. CONCLUSION

In moderate-speed rear-impact reconstructions using a 5th percentile Hybrid III headform mounted on the uONSA with a 100% MVC braced-neck spring configuration, both direct and 7.6 cm offset head-restraint positions produced MPS values exceeding commonly reported 50% concussion-risk levels when calculated using a geometrically scaled UCDBTM v2.0. These findings indicate that moderate-speed rear impacts pose a risk of concussion for braced female occupants. Under these specific experimental and modelling conditions, initial head position influenced the timing and magnitude of head kinematics, while impact velocity was the most consistent determinant of the measured responses.

VI. ACKNOWLEDGEMENTS

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