# Headform and Neck Effects on Dynamic Response in Bicycle Helmet Oblique Impact Testing

Megan L. Bland, Craig McNally, Steven Rowson

***Abstract*** The incidence of cycling-related injuries in the USA has increased in recent years, with the head being among the most commonly and seriously injured body parts. Current bicycle helmet standards present limited representation of real-world cyclist accidents. Researchers generally agree that evaluating helmets using oblique impacts, which are common in cyclist impacts, could enhance helmet design. However, the boundary conditions across various oblique impact rigs vary widely. The purpose of this study is to evaluate effects of the anthropomorphic test device (ATD) headform and neck on dynamic response in bicycle-helmeted, oblique impacts. A Hybrid III (HIII) and National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform were impacted in drop tests onto an angled anvil with or without a HIII neck. Both headforms produced similar linear response (0.4% average difference in peaks, p=0.76), while rotational velocity and acceleration peak responses were ~22% and 31% greater for the HIII (p<0.01). The headform-only tests produced 17-35% greater peak linear and rotational values than the tests with a neck, although impact location influenced neck effects. Trends in these results can be used to interpret differences across published bicycle helmet oblique impact studies and have important implications for injury risk.

***Keywords*** Biomechanics, Cycling, Head injury, Impact kinematics, Rotational

## I. INTRODUCTION

Cycling is a common activity and form of transit worldwide and is becoming increasingly popular in the USA, with many embracing its health and environmental benefits [1-3]. However, its increase in popularity is paralleled by increasing accident injury rates. The number of bicycle-related hospital admissions has grown 120% over the past 15 years [4], and in 2015 alone there were 818 cyclist fatalities – a 12% increase on the previous year [1]. The associated costs are considerable, with adult cycling-related injuries accounting for an estimated $24.4 billion in the USA in 2013 [5]. The head is among the most commonly injured body parts in these accidents, especially for more severe or fatal injuries [6-8].

Helmet use has been repeatedly demonstrated to reduce head injury risk in cycling [6-7][9-13]. During an impact, the expanded polystyrene (EPS) liner of a typical helmet crushes permanently, dissipating the energy that would otherwise be transferred to the head. Helmet impact performance is regulated by safety standards that impose a limit on peak linear acceleration (PLA) of an anthropomorphic test device (ATD) headform during prescribed impact testing [14-15]. This testing involves guided drop tests of the headform onto an anvil at an impact angle normal to the anvil surface. In the USA standard, the metal half-headform is rigidly attached to the drop mass so that rotation of the head is constrained upon impact [14].

Standards impact testing, although effective in limiting energy input into the head, is not completely representative of real-world cyclist accidents. In contrast to the solely normal impacts conducted in standards testing, it has been shown that cyclist head impacts involve both normal and tangential incident velocities (termed 'oblique') [8]. This occurs because cyclists typically approach the ground at an angle upon falling, often between 30° and 60° from the horizontal [16-20]. Standards also only assess PLA, while oblique impacts generate considerable rotational motion of the head in addition to linear. This has significant implications for injury risk, as the brain is known to be especially susceptible to diffuse injury from the relative motion and tissue strain that are associated with rotational impact [21-24]. Peak rotational velocity (PRV) and acceleration (PRA), often in conjunction with PLA, have been correlated with occurrence of traumatic brain injury (TBI) [23-28]. Lastly, the metal half-headform and rigid connection to the drop mass used in standards testing are not reflective of human properties and preclude realistic assessment of headform motion upon impact.

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Due to the limitations of standards testing in replicating real-world cyclist head impacts, more recent research has suggested evaluating bicycle helmets in oblique impacts as an avenue for further improvement of helmet design [12][29-31]. A variety of oblique impact test rigs have been developed for such testing. Several rigs involve a horizontally-moving plate to generate the normal and tangential inbound velocities characteristic of oblique impacts [30-32], while others make use of an angled anvil [29][33-34]. Choice of headform and use of an ATD neck also vary across these studies, as do prescribed impact velocities and locations. However, there is currently no universally accepted method for oblique impact testing, and the effects of varying several of these boundary conditions on resulting helmet performance are not well understood.

The incorporation of a neck is a particularly debated boundary condition for oblique impact rigs. The Hybrid III (HIII) 50th percentile male neck is one of the most commonly-used surrogate necks for impact testing. Originally developed for use in the automotive industry and validated based on post mortem human subject (PMHS) flexion/extension response in frontal sled tests [35], it has also been used extensively in athletic equipment testing in recent years [12][32-33][36-40]. Despite its wide usage, the HIII neck has been shown to be less biofidelic in other loading scenarios and directions [41-46]. This has produced disagreement as to whether it is suitable to incorporate the neck (and effective torso mass) in sports ATD head impact testing and what the effects of testing with and without it may be. Some researchers have suggested that the neck does not play a role in the initial impact response of the head in humans, and therefore it is better to forego it due to its lack of biofidelity [31][47]. Others advocate for its use, stating that the neck in fact plays a key role in the impact response of the head [12][48-49]. Among those arguing that the neck does affect impact response, various theories on its exact effects on kinematic outcomes have been presented [34][42][47-48][50-52].

The choice of headform is another debated boundary condition for head impact testing. For sports helmet testing, the 50th percentile male HIII and National Operating Committee on Standards for Athletic Equipment (NOCSAE) are two of the most commonly used headforms [29][32][36-37][39-40][49][53-54]. The HIII headform was developed in tandem with the HIII neck for automotive testing [55], and has a large central cavity allowing for placement of a nine-accelerometer array (9AA) to enable calculation of linear and angular head acceleration [56]. The NOCSAE headform, on the other hand, was developed specifically for helmet testing [57]. It has a more biofidelic shape compared to the HIII [58] and is thought to have more humanlike construction, with a gel-filled cavity simulating the brain [57]. While the NOCSAE was not designed to be coupled to a neck and contains a smaller instrumentation channel, past studies have modified the headform to couple it to a HIII neck and used linear accelerometers and tri-axis angular rate sensors (ARS) to record linear and rotational kinematics [36][59]. Both the HIII and the NOCSAE headforms were validated based on PMHS linear acceleration curves in drop tests [55][57]. However, differences in geometry, material construction, instrumentation and inertial properties between the two headforms likely affect dynamic response. To date, two studies have investigated differences in dynamic response across headform in other loading scenarios, but neither evaluated oblique impacts or varying neck configurations, which involve differing and possibly more complex loading characteristics [36][54]. Since published bicycle helmet oblique impact studies have utilised varying headform and neck configurations, understanding the effects of these variations is imperative to interpreting results across studies.

The purpose of this study is, therefore, to investigate the effects of choice of ATD headform and neck on dynamic response in bicycle-helmeted, oblique impacts.

## II. METHODS

Impact testing was conducted using a linear drop tower with a sandpaper-coated 45° anvil (80 grit to mimic road surfaces [60]). A blue, medium NOCSAE headform and a 50th percentile male HIII headform were used in testing (Fig. 1, Table I), and were connected to a 50th percentile male HIII neck for half of the tests. The neck was also attached to an effective torso mass (16 kg) that was constrained to the drop tower. This torso mass represents the upper torso of a 50th percentile male HIII dummy, which has been used in a variety of head impact testing scenarios [61-64]. The drop mass was stopped just after impact to limit the load on the neck. For the No Neck tests, the headform was positioned in hoop of a larger diameter than the anvil such that it passed outside the anvil upon impact, separating it from the headform. The headform was secured in place using a lever arm that released just prior to impact, and was then free to rotate off the anvil upon impact. For the Neck tests, the NOCSAE headform required modification to render it compatible with the HIII neck. This modification has been outlined in previous studies, and resulted in a similar mass between headforms and a centre of gravity (CG) located 22 mm superior for the NOCSAE headform compared to the HIII (relative to the occipital condyle (OC) pin) [36][59].

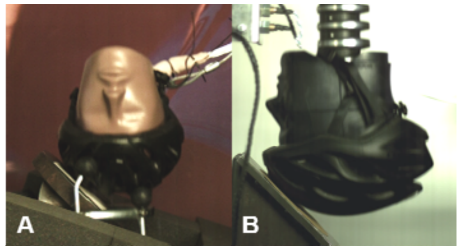


Fig. 1. Representative configurations for oblique impact testing. A HIII, No Neck, parietal impact is shown in (A), while a NOCSAE, Neck, frontal impact is shown in (B).

TABLE I
PROPERTIES OF BOTH THE HIII AND THE NOCSAE HEADFORMS ALONG WITH ASSOCIATED HUMAN HEAD VALUES [54-55][65-67], WHERE MOMENTS OF INERTIA ABOUT THE X, Y, AND Z AXES THROUGH THE HEAD CG (SAE J211 COORDINATE SYSTEM) ARE DENOTED USING *I* AND THE RELATED SUBSCRIPT. HUMAN VALUES REPRESENT A RANGE IN AVERAGES REPORTED FOR MALE SPECIMEN. THE INERTIAL CHARACTERISTICS OF THE NOCSAE HEADFORM ARE NOT WELL-DOCUMENTED

| | **Human** | **HIII** | **NOCSAE** |
|---|---|---|---|
| **Circumference (cm)** | 55.0 – 59.0 | 57.2 | 57.6 |
| **Mass (kg)** | 3.5 – 4.8 | 4.54 | 4.40 |
| **$I_x$ (kg cm$^2$)** | 130 – 249 | 159 | 190 |
| **$I_y$ (kg cm$^2$)** | 164 – 244 | 240 | - |
| **$I_z$ (kg cm$^2$)** | 136 – 201 | 220 | - |

Each headform-neck configuration was tested at 6 m/s at both frontal and parietal impact locations (commonly impacted velocity and locations in cyclist accidents [16][19-20]). Consistent headform positioning between tests for the Neck impacts was ensured using defined Y- and Z-axis rotation increments on the drop mass (SAE J211 standard coordinate system). For the No Neck tests, matching impact locations were defined using a dual-axis inclinometer (DMI600, Omni Instruments, Dundee, UK). Five trials were conducted for each of the eight headform-neck-location configurations, totalling 40 tests in all. Prior to testing, headforms were fitted with a Bell Draft bicycle helmet, with a new helmet used per test. This helmet was selected to be representative of a classic road helmet design, containing an expanded polystyrene liner, a polycarbonate shell, an elongated shape with venting throughout, and no other notable technologies that might influence impact performance compared to conventional road helmets. Helmets were fitted according to manufacturer recommendations on both headforms by adjusting the dial fit system in the rear of the helmet and tightening the retention straps around the jaw/neck region until snug. High speed video was taken for each configuration (1000 frames/s, Phantom v9, Vision Research, Wayne, NJ).

Linear acceleration and rotational velocity data were collected at 20 kHz for all tests using three linear accelerometers (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA) and a tri-axis ARS (ARS3 PRO-18K, DTS, Seal Beach, CA) located at the CG of both headforms. The ARS was selected as it is more readily compatible with the NOCSAE headform than the traditional 9AA system [36][68]. Rotational accelerations were calculated as the derivative of rotational velocity traces. Linear acceleration was filtered in accordance with SAE J211 using a channel frequency class (CFC) of 1000, while rotational velocity was filtered at a CFC of 175. The latter CFC was optimised in previous studies to minimise the error between ARS-calculated PRA and 9AA-calculated PRA in pendulum impacts to a football-helmeted HIII headform [68]. This filter choice was also validated in the present study by instrumenting the HIII with a 9AA as well as the ARS. The average difference in PRA between the two systems was 0.23%, so the ARS-calculated PRA was used in analysis.

Resultant PLA, PRV and PRA were determined per test, as well as duration, time to PLA, and time to PRA.

Duration was defined as the time between the resultant linear acceleration curve first exceeding 5 g and first falling below 5 g after the peak (5 g threshold determined to reliably detect endpoint of impact pulse). All metrics were compared across headform-neck-location configurations using 3-way ANOVA with Tukey's HSD post hoc tests.

## III. RESULTS

Distinct kinematic properties were observed for each of the headform and neck configurations. Average parameter values are given in Table II, with all significant comparisons provided in the Appendix (Table A1). Both headforms produced similar time-based metrics and PLA, with average differences less than 6.5%. PLA was within 0.4% across headform on average (p=0.76), and resultant linear acceleration curves for the HIII and NOCSAE were very similar across the entire time series (Fig. 2). This was contrasted by the resultant rotational velocity and acceleration curves (Fig. 3 and Fig. 4), which were markedly different across headform and produced peak values for the NOCSAE that were 20-30% less than those for the HIII (p<0.01). The Neck curves showed a secondary peak corresponding with the drop mass being caught following impact. This peak was ignored, and only the first peaks were used in analysis. All trends in headform differences were generalisable across impact location, with the exception of PRV for the parietal Neck impacts. PRV was not significantly different across headform for this configuration (p=0.39), while the HIII produced greater PRVs for all other configurations. Variance was low for both headforms, with coefficients of variance (CV) averaging 5.3% and 6.3% for the HIII and NOCSAE, respectively.

TABLE II

PARAMETER AVERAGES ± STANDARD DEVIATIONS ACROSS ALL IMPACT CONFIGURATIONS. THE HEADFORMS PRODUCED MARKEDLY DIFFERENT ROTATIONAL KINEMATICS, AND THE ADDITION OF A NECK LOWERED ASSOCIATED ACCELERATIONS AND VELOCITIES

| | **HIII** | | | | **NOCSAE** | | | |
|---|---|---|---|---|---|---|---|---|
| | **Neck** | | **No Neck** | | **Neck** | | **No Neck** | |
| | **Frontal** | **Parietal** | **Frontal** | **Parietal** | **Frontal** | **Parietal** | **Frontal** | **Parietal** |
| **Duration (ms)** | 10.6 ± 0.3 | 9.7 ± 0.7 | 10.3 ± 0.6 | 9.4 ± 0.4 | 9.5 ± 0.5 | 11.4 ± 0.9 | 9.7 ± 0.2 | 8.7 ± 0.3 |
| **Time to PLA (ms)** | 4.9 ± 0.1 | 3.9 ± 0.8 | 5.0 ± 0.0 | 4.0 ± 0.2 | 4.8 ± 0.2 | 3.8 ± 0.1 | 4.4 ± 0.2 | 4.0 ± 0.3 |
| **Time to PRA (ms)** | 4.5 ± 0.1 | 2.1 ± 0.4 | 4.6 ± 0.2 | 3.7 ± 0.6 | 2.8 ± 1.0 | 5.1 ± 2.0 | 3.9 ± 0.3 | 4.0 ± 0.2 |
| **PLA (g)** | 128.6 ± 4.6 | 105.9 ± 8.5 | 148.7 ± 2.8 | 155.5 ± 7.0 | 115.4 ± 3.4 | 109.2 ± 3.5 | 152.7 ± 4.9 | 163.5 ± 5.4 |
| **PRV (rad/s)** | 35.5 ± 0.6 | 22.0 ± 0.9 | 35.9 ± 0.4 | 34.5 ± 1.1 | 27.0 ± 0.3 | 20.6 ± 2.4 | 28.9 ± 0.3 | 23.6 ± 0.5 |
| **PRA (rad/s$^2$)** | 8860 ± 509 | 6011 ± 602 | 7680 ± 284 | 6011 ± 602 | 5570 ± 293 | 3164 ± 432 | 6974 ± 60 | 5558 ± 246 |

The two neck conditions generated comparable time-based parameters (differences generally within 10%), although duration was shorter for No Neck (p<0.01). Differences in PLA, PRV and PRA were larger (No Neck 17–35% greater, p<0.01), reflected in the time series curves of resultant linear acceleration and rotational velocity and acceleration (Fig. 5 and Fig. 7). All trends were again generally similar across impact location. However, the rotational velocity curves appeared much more similar at the frontal location than at the parietal location (Fig. 6). Additionally, variance was greater for the Neck tests, producing an average CV of 8.0% compared to a CV of 3.6% for No Neck tests.

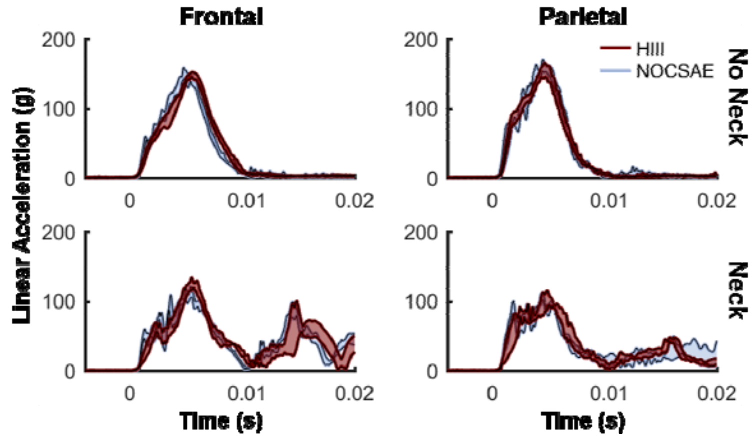


Fig. 2. Time series linear acceleration corridors (min. and max.) across headform at the No Neck (top), Neck (bottom), frontal (left), and parietal (right) conditions. Curves were similar across headform with low variance.

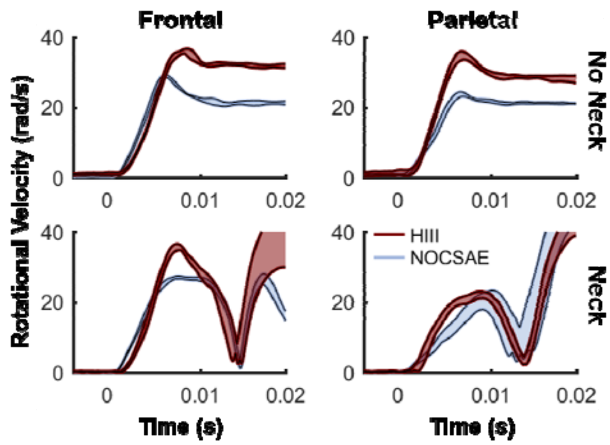


Fig. 3. Rotational velocity corridors (min., max.) across headform. The HIII PRV was generally greater.

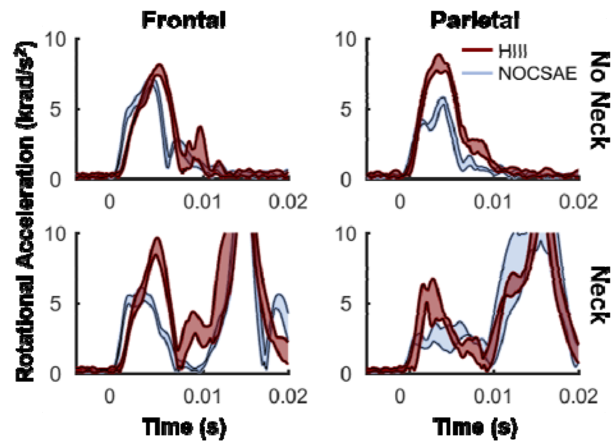


Fig. 4. Rotational acceleration corridors (min., max.) across headform. PRA was greater for the HIII.

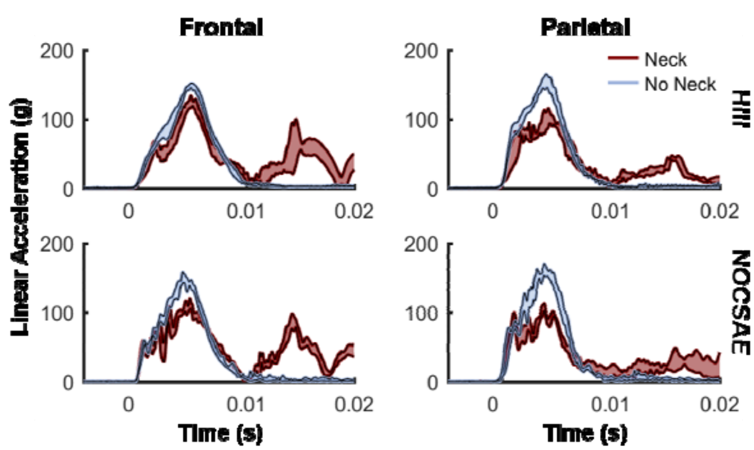


Fig. 5. Time series linear acceleration corridors (min. and max.) across neck condition for the HIII (top), the NOCSAE (bottom), frontal (left), and parietal (right) conditions. The No Neck tests produced greater PLA.

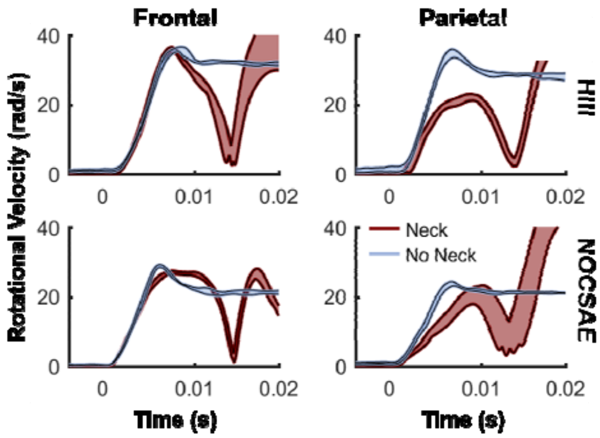


Fig. 6. Rotational velocity corridors (min., max.) across neck condition. No Neck PRV was greater at the parietal location.

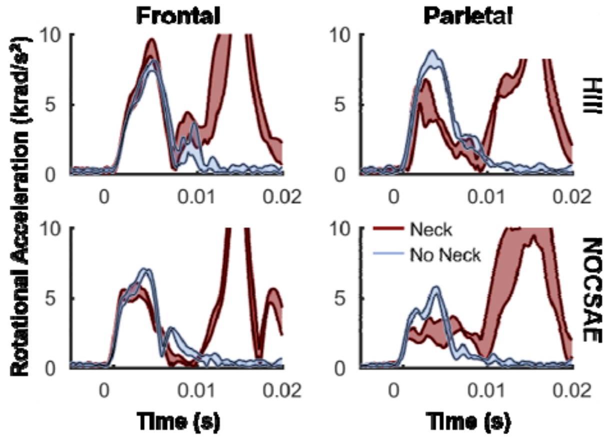


Fig. 7. Rotational acceleration corridors (min., max.) across neck condition. PRA was generally greater for No Neck.

## IV. Discussion

Each of the headform and neck conditions evaluated produced distinct kinematic profiles. Across all configurations, the greatest similarities in response curves were observed in linear acceleration between headforms (Fig. 2). This is likely attributable to both headforms being designed to reflect linear acceleration characteristics in PMHS head drop tests [55][57]. Conversely, rotational kinematic parameters varied considerably between the headforms, with the HIII averaging 22% and 31% greater PRV and PRA than the NOCSAE, respectively. This may suggest that the moments of inertia (MOI) about the X and Y axes for the HIII are smaller than for the NOCSAE. Literature values indeed indicate a larger MOI for the NOCSAE about the X axis (Table I), although MOI about other axes are not as well documented for this headform, and experimental methods used across these studies are not clearly defined [54-55][66]. Human head MOI vary considerably across the population [65], so the NOCSAE MOI values may still fall within a biofidelic range. Future studies are needed to fully characterize MOI of both headforms using matched methodology.

Differing frictional properties of the headforms may have also contributed to variation in rotational kinematics. This is especially likely in the present impact scenario, where the high friction of the simulated road surface causes the helmet to stick initially and exacerbates effects at the helmet-headform interface. The HIII headform is thought empirically to have an unrealistically high surface friction and may therefore produce greater peaks in rotational kinematics [64]. However, analysis of relative angles between the helmet and headforms during impact using high-speed video did not indicate increased motion at this interface for one headform than another (relative angles within ~1.5° across headforms). Nonetheless, future studies are needed to validate dynamic frictional properties of each headform. Importantly, both headforms produced little variance and thereby good repeatability, which is essential for comparative testing.

The present trends in headform results have varying levels of agreement with other studies comparing the two headforms. Cobb *et al.* [36] used pendulum tests to impact football-helmeted headforms attached to a HIII neck. In contrast with the present study, the NOCSAE showed greater PLA and PRA than the HIII. Despite this, PLA differences were much lower and often not significant, similar to the results of this study. Kendall *et al.* [54] also found greater PLA and PRA for the NOCSAE rather than the HIII in drop tests of both headforms onto a flat anvil through the CG. However, both studies utilised different loading scenarios than the oblique, helmeted impacts assessed herein, and neither study assessed differences across neck configuration. The differing resultant force vectors and transfer of momentum involved in the present study complicate the generalisability of trends across studies. Nonetheless, Cobb *et al.* and Kendall *et al.* also demonstrated good repeatability of both headforms.

The use of a neck produced markedly different kinematics for both linear and rotational response curves. The No Neck tests generated significantly shorter durations than the Neck tests, likely owing to a delayed response of the neck and to the headform being constrained to the drop tower for the Neck tests. The No Neck tests also produced significantly greater PLA, PRV and PRA. Linear acceleration differences suggest that the neck and effective torso mass play into the overall effective mass during impact, lowering PLA for the Neck tests. The increase in overall effective mass would also contribute to inertial differences between the Neck and No Neck tests, affecting rotational characteristics as well. These results suggest that the neck and body do, in fact, affect the dynamic response of the head in ATD testing. Several simulation studies have predicted similar results for humans as well [50-51]. Lastly, it is also important to note that use of the neck produced considerably higher variance under these loading conditions.

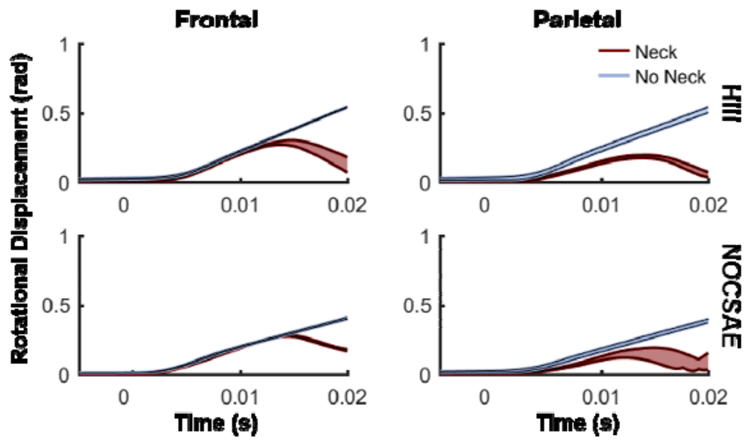


Fig. 8. Time series rotational displacement corridors (min. and max.) across neck condition. The frontal location produced similar curves during the duration of the impact, while the parietal impacts differed more notably.

Trends in headform and neck condition differences were generally consistent across impact location. One notable exception was that PRV was significantly different between headform types for all configurations except the parietal, Neck configuration, as evidenced by the rotational velocity curves (Fig. 3). This may be attributable to the design of the HIII head-neck junction, which rotates in flexion-extension about a pin joint, but is constrained (and therefore much stiffer) in the lateral direction. In lateral impacts, the motion of the head is thus translated to a point lower in the neck, and it is possible that this neck motion was a controlling factor in the dynamic response. This is supported by the linear acceleration loading curves, which exhibit a plateaued region for parietal neck impacts. Further, when comparing across neck condition, rotational velocity curves at the frontal location are much more similar than those at the parietal location (Fig. 6), suggesting the neck plays a larger role in loading characteristics at the parietal location. This is also reflected in PRV values, which were significantly different between Neck and No Neck tests at the parietal location ($p<0.01$) but not at the frontal location ($p>0.13$). Differentiating these rotational velocity curves demonstrates the differences in rotational displacement paths (Fig. 8). At the frontal location, the Neck and No Neck curves are very similar up to ~12 ms, at which point the parietal curves are more distinctly different between neck conditions.

There are several limitations to the present study that should be taken into account when interpreting the results. First, only one helmet model, one velocity, one anvil angle, two locations, and one neck orientation relative to the anvil were tested. Results are specific to these prescribed impact conditions, although general trends may hold true in other conditions as well. Varying orientations of the neck relative to the anvil may have a larger effect on neck trends, as this may change the contribution of the neck/effective torso mass to the overall effective mass. A study with the body oriented horizontally in drop tests indeed showed varying trends [48]. Further, varying relative neck orientation influences the type of loading that results in the neck. In the present study, the HIII neck was likely subject to considerable axial loads, under which this neck is thought to behave differently than a human neck [41][44]. Thus, the HIII neck in this orientation may not be representative of a human response, whereas it may be more realistic in other loading scenarios. Additionally, the choice of effective torso mass could also have affected kinematics; however, previous head impact studies considering a variety of loading scenarios have suggested that the torso may not play into the impact until the primary head impact forces have subsided [52][61]. Finally, effects of hair and a scalp were not considered in the present study, which likely affect the helmet-headform interface and resulting dynamics. Despite these limitations, the prescribed impact conditions reflect those common in published bicycle helmet oblique impact studies, and as such these results provide a basis upon which to evaluate differences in results across other studies.

## V. CONCLUSIONS

The present study demonstrates that choice of ATD headform and use of a neck have significant effects on dynamic response during bicycle-helmeted, oblique impacts. The NOCSAE headform produced similar PLA but lower PRV and PRA compared to the HIII, while the incorporation of the HIII neck generally decreased PLA, PRV and PRA compared to headform-only tests. These trends provide a framework by which to interpret results across published bicycle helmet oblique impact studies and have important implications for injury risk assessment. Future research optimising testing boundary conditions around replicating human responses can contribute to improved helmet design for cyclist safety.

## VI. ACKNOWLEDGEMENTS

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## VIII. Appendix

TABLE A1

Significant comparisons ($p < 0.05$) per kinematic parameter, with H = HIII, N = NOCSAE, F = Frontal, P = Parietal. Only comparisons that differ by one factor are included (e.g. H-Neck-F not compared to H-NoNeck-P or N-Neck-P)

| **Duration (ms)** | **Time to PLA (ms)** | **Time to PRA (ms)** | **PLA (g)** | **PRV (rad/s)** | **PRA (rad/s$^2$)** |
|---|---|---|---|---|---|
| H-Neck-F<br>N-Neck-F | H-Neck-F<br>H-Neck-P | H-Neck-F<br>N-Neck-F | H-Neck-F<br>N-Neck-F | H-Neck-F<br>N-Neck-F | H-Neck-F<br>N-Neck-F |
| H-Neck-P<br>N-Neck-P | H-NoNeck-F<br>H-NoNeck-P | H-Neck-P<br>N-Neck-P | H-Neck-F<br>H-Neck-P | H-NoNeck-F<br>N-NoNeck-F | H-Neck-P<br>N-Neck-P |
| N-Neck-F<br>N-Neck-P | N-Neck-F<br>N-Neck-P | H-Neck-F<br>H-Neck-P | H-Neck-F<br>H-NoNeck-F | H-NoNeck-P<br>N-NoNeck-P | H-NoNeck-P<br>N-NoNeck-P |
| N-Neck-P<br>N-NoNeck-P | | N-Neck-F<br>N-Neck-P | H-Neck-P<br>H-NoNeck-P | H-Neck-F<br>H-Neck-P | H-Neck-F<br>H-Neck-P |
| | | | N-Neck-F<br>N-NoNeck-F | N-Neck-F<br>N-Neck-P | N-Neck-F<br>N-Neck-P |
| | | | N-Neck-P<br>N-NoNeck-P | N-NoNeck-F<br>N-NoNeck-P | N-NoNeck-F<br>N-NoNeck-P |
| | | | | H-Neck-P<br>H-NoNeck-P | H-Neck-F<br>H-NoNeck-F |
| | | | | N-Neck-P<br>N-NoNeck-P | H-Neck-P<br>H-NoNeck-P |
| | | | | | N-Neck-F<br>N-NoNeck-F |
| | | | | | N-Neck-P<br>N-NoNeck-P |