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Hybrid III anthropomorphic test device (ATD) response to head impacts and potential implications for athletic headgear testing

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ABSTRACT

The Hybrid III 50th percentile male anthropomorphic test device (ATD) is the most widely used human impact testing surrogate and has historically been used in automotive or military testing. More recently, this ATD is finding use in applications evaluating athletic helmet protectivity, quantifying head impact dosage and estimating injury risk. But ATD head–neck response has not been quantified in omnidirectional athletic-type head impacts absent axial preload. It is probable that headgear injury reduction that can be quantified in a laboratory, including in American football, boxing, hockey, lacrosse and soccer, is related to a number of interrelated kinetic and kinematic factors, such as head center of gravity linear acceleration, head angular acceleration, head angular velocity, occipito-cervical mechanics and neck stiffness. Therefore, we characterized ATD head–neck dynamic response to direct head impacts in a series of front, oblique front and lateral head impacts.

Key findings were: (1) impacts producing highest ATD resultant center of gravity linear acceleration resulted in the lowest resultant occipito-cervical spine bending moment/force. (2) Resultant ATD head angular velocity and angular acceleration did not appear coupled to impact direction at lower impact energy levels; these parameters were coupled at higher energy levels. (3) The ATD had progressively increasing occipito-cervical stiffness in extension, torsion and lateral bending, respectively. Because the ATD neck influenced head and neck impact dosage parameters, testing agencies, manufacturers and researchers should consider using the Hybrid III head form attached to a neck as a means to quantify head and neck injury risks as opposed to systems that do not utilize a neck. This heightened understanding of Hybrid III ATD head–neck response, and consideration of order of stiffest axes in the lateral, oblique and extension directions, respectively, should aid in the development of head and neck injury impact testing standards.

1. Introduction

Concussion, traumatic brain injury (TBI) and spine injuries continue to pose a threat to athletes engaged in contact sports. The Hybrid III 50th percentile male anthropomorphic test device (ATD) is a standard impact testing human surrogate that is finding increasing use in athletic injury analysis (Viano et al., 2006; Beckwith et al., 2007; Funk et al., 2009; Manoogian et al., 2006; Newman et al., 2005; Withnall et al., 2005; Rousseau and Hoshizaki, 2009; Walilko et al., 2005; Rousseau et al., 2010; Pellman et al., 2006). The Hybrid III ATD was originally developed and validated for United States Federal Motor Vehicle Safety Standard (FMVSS) 208 frontal crash testing (National Highway Traffic Safety Administration, 2008). As such, head–neck dynamic response was developed and validated in the sagittal plane using the pioneering studies of Mertz and Patrick (1967, 1971). Although the Hybrid III ATD has been well characterized in automotive-type impacts (Duma et al., 2003), with the exception of one recent study (Rousseau and Hoshizaki, 2009), the aforementioned athletic injury analysis studies have not specifically studied ATD head–neck biofidelity.

Hybrid III ATD head–neck characterization is of particular interest because the National Operating Committee on Standards for Athletic Equipment (NOCSAE) is investigating the use of Hybrid III

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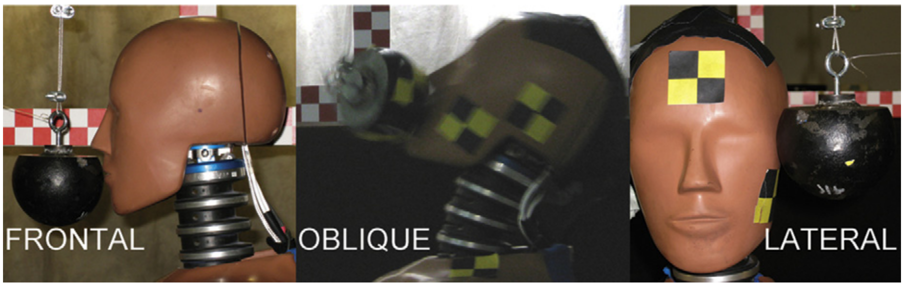

Fig. 1. Front, oblique front and lateral impact setups.

to supplement the existing testing standards (National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2008; National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2009; National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2006). It is probable that helmet and headgear protectivity in ATD head impacts are related, in part, to head kinetics and kinematics, like linear acceleration, angular acceleration, angular velocity, energy transfer, momentum transfer and contact forces, along with occipito-cervical junction mechanics and neck surrogate stiffness—as opposed to solely being a function of the helmet or headgear tested. Further, ATD head impact response (including the Hybrid III, NOCSAE, DOT and ISO head forms) may not fully reproduce human biofidelity (Agaram et al., 2001; Kang et al., 2001) during direct loading. Finally, the Hybrid III ATD head–neck was validated in the sagittal plane without direct head impact (Mertz and Patrick, 1967, 1971); the hollow ATD skull and absence of a deformable brain might influence the inertial loading parameters of the head–neck spine complex. This issue becomes especially complex when ‘crossover’ surrogate models are used, such as testing with a combination of a Hybrid III neck form and NOCSAE head form (Gwin et al., 2010; Bartsch et al., 2011) that could potentially over-compress standard Hybrid III neck rubber nodding blocks or limit range-of-motion.

Therefore, in order to characterize Hybrid III ATD head–neck response to direct head impacts absent axial compression, we conducted impacts in the front, oblique front and lateral directions absent preload at three (3) increasing impact magnitudes on par with those known to cause head and neck injury in athletes (Viano et al., 2006; Beckwith et al., 2007; Funk et al., 2009; Manoogian et al., 2006; Newman et al., 2005; Withnall et al., 2005; Rousseau and Hoshizaki, 2009; Walilko et al., 2005; Rousseau et al., 2010).

2. Methods

A Hybrid III 50th percentile ATD was instrumented with a triaxial linear accelerometer (Model EAS3-250 Measurement Specialties, Hampton, VA) rigidly mounted at the head center of gravity (cg) and a triaxial angular rate sensor (Model ARS-06S, ATA Sensors, Albuquerque, NM) mounted adjacently and in the same local coordinate system as the accelerometer. A six-channel occipito-cervical load cell (Model 1716A, Denton ATD, Rochester, MI) measured forces and moments in three directions. The moments measured by this supracondylar load cell were translated inferiorly to the occipital condyles by multiplying respective x- or y-axis shear forces by the appropriate 0.01778 m moment arm. All data were collected at 5000 Hz, filtered according to SAE J211 (Society of Automotive Engineers, 2003) and sign convention adhered to SAE J1733 (Society of Automotive Engineers, 1994). Angular acceleration was calculated by differentiating the filtered angular velocity signal. The coefficient of variation (COV) for independent variables was calculated by dividing the standard deviation from each trial set of six impacts by the trial set mean value. This COV was calculated in order to ensure each set of six impact trials was repeatable and that using the mean value for each dynamic parameter was justified.

A 3.6 kg steel sphere impact pendulum was used in all impacts. This rigid effective mass was chosen to simplify the test setup by minimizing impactor energy dissipation post-impact. The impactor was hung from a 6.4 mm steel braided cable approximately 5 m in length that was free to swing from a ceiling-mounted carabiner directly above the ATD head cg. The ATD was seated in a chair and secured with tie-down straps. Prior to testing, the lower extremities were removed. In order to minimize inertial responses below the cervical spine, the test stand was weighed down with 3500 N of sandbags and the ATD thorax was secured to the test stand with tie-down straps. For each impact trial, the pendulum was released from the desired swing height by cutting a lightweight nylon string attached to the impactor. In our study, a total of 54 head impact trials were performed, in sets of six trials for each impact direction (front, oblique front, lateral) and impact energy level (lowest, medium, highest) as shown in Fig. 1 and Table 1. The highest energy impacts were limited by the available swing height within the laboratory testing space. The front and oblique front impacts targeted the maxillar–mandibular junction, while the lateral impact targeted the temporal region. All impacts were in the same vertical plane as the head cg. The swing heights were selected within ranges of impact energy and momentum from prior head impact studies (Viano et al., 2005; Stojsih et al., 2010; Smith et al., 1988; Pincemaille et al., 1989; Ommaya and Hirsch, 1971; Margulies and Thibault, 1992; Thibault and Gennarelli, 1990; Newman et al., 2000; Ommaya et al., 2002; Pellman et al., 2003; Zhang et al., 2004; Greenwald et al., 2008; Newman et al., 2005; Duma et al., 2005; Casson et al., 2009; Rowson et al., 2009; Frechede and McIntosh, 2009; Walilko et al., 2005) that indicated higher risk of concussion or neck injury.

The mean dynamic results only are presented because the omnidirectional impacts precluded meaningful comparison among respective x-, y- and z-axes. Furthermore, the mean dynamic results were plotted against impact energy only, and were not plotted against impact momentum for purposes of brevity.

3. Results

Table 2 displays the respective coefficient of variation (COV) from the impact trials at each impact energy and direction. The highest COV was observed with occipital force, with a mean 7.7% variation among all trials and all impact directions. Resultant head cg linear acceleration was associated with the lowest mean COV with 4.6%.

Shown in Fig. 2, mean resultant linear cg acceleration was highest for lateral impacts, followed by front and oblique front impacts, respectively. For the mean resultant head angular velocity shown in Fig. 3, at low energy, all three directions produced similar resultant angular velocities. These responses remained tightly grouped at

Table 1
Impact matrix.

Impact direction	Impact energy (J)			Impact momentum (N s)		
	Low	Medium	High	Low	Medium	High
Front	27	54	76	14	20	23
Oblique front	27	50	89	14	19	25
Lateral	27	54	67	14	20	22

Table 2
Coefficient of variation (COV) for all impacts.

Percent coefficient of variation				
Impact direction	Head cg acceleration	Head angular velocity	Occipital force	Occipital moment
Front low	2.9	3.5	8.3	2.1
Front mid	2.5	1.8	5.7	2.7
Front high	3.9	4.8	5.9	7.1
Oblique front low	7.4	5.2	18.4	5.7
Oblique front mid	10.1	8.7	7.7	8.8
Oblique front high	4.5	4.7	8.3	6.1
Lateral low	4.7	12.8	5.0	9.1
Lateral mid	2.1	9.5	3.3	5.5
Lateral high	3.6	5.5	6.4	7.3
Mean	4.6	6.3	7.7	6.0

medium energy. In high energy impacts, front impacts had the highest angular velocity, followed by oblique front and lateral impacts, respectively. The mean resultant head angular acceleration shown in Fig. 4 displayed a similar trend to the angular velocity results. At low energy, all three directions had similar angular acceleration magnitudes. Again, a crossover was observed in medium energy impacts, as lateral magnitude dropped below the oblique front magnitude. In high energy impacts, front angular acceleration was the largest, followed by oblique front and lateral impacts. In Fig. 5, the mean resultant occipital moment had a clear trend for all three impact energies. The front impact moment was always highest, followed by oblique front and lateral impacts. Finally, in Fig. 6, the mean resultant occipital force was plotted against impact energy. The front impacts were always associated with the highest occipital force across all impact energies. The oblique front and lateral impacts demonstrated very similar results in low energy impacts.

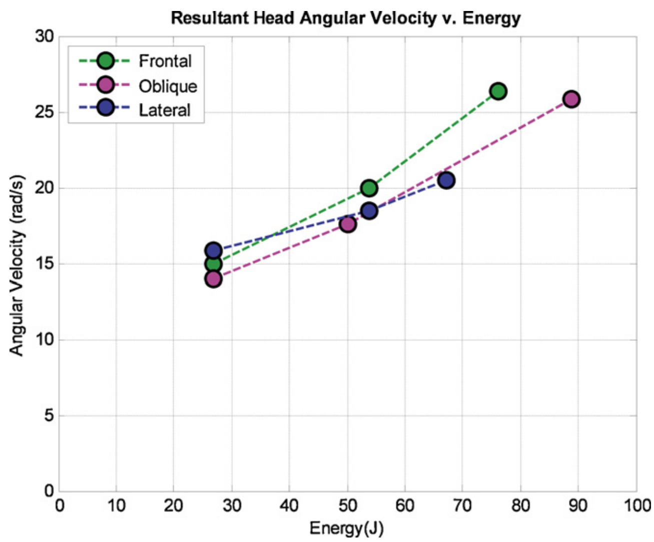

Fig. 3. Mean resultant ATD peak head angular velocity.

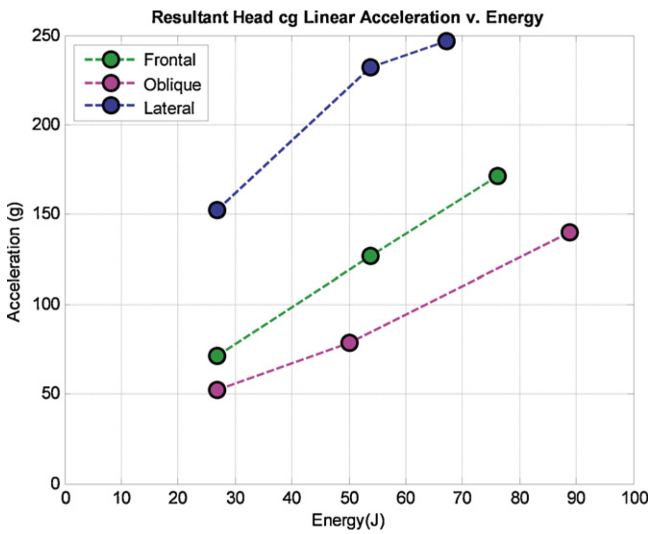

Fig. 2. Mean resultant ATD head center of gravity (cg) peak linear acceleration for the three impact directions compared with impact energy.

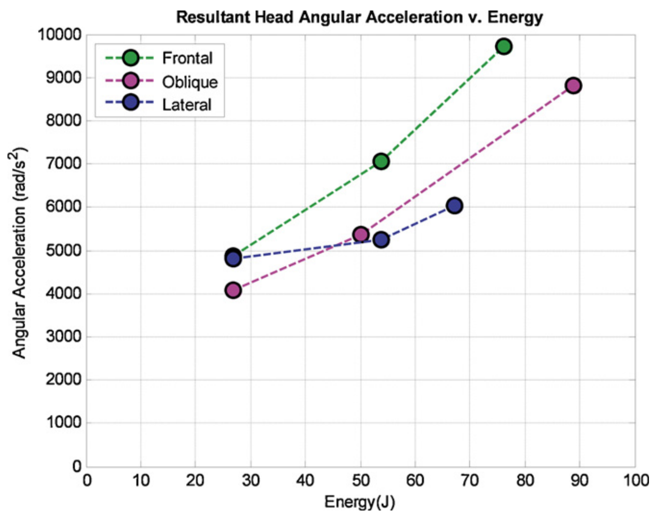

Fig. 4. Mean resultant ATD peak head angular acceleration.

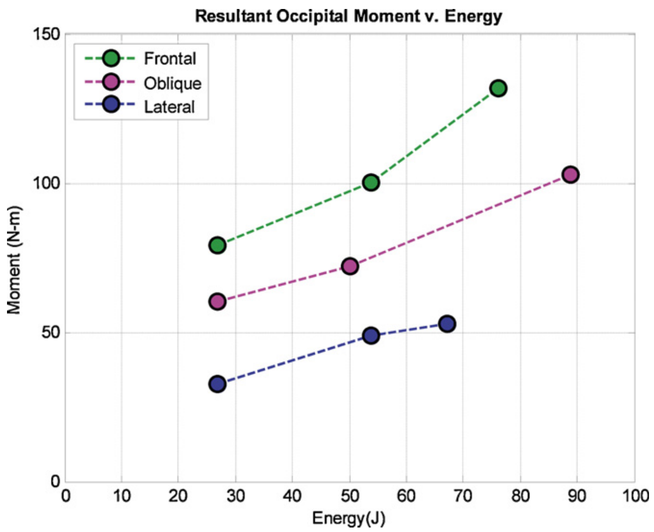

Fig. 5. Mean resultant ATD peak occipital bending moment.

4. Discussion

When comparing head impacts at similar energy levels, but different impact vectors, ATD resultant head acceleration was highest with lateral impacts, followed by front and oblique front impacts, respectively. At the highest impact energy levels, resultant front head angular velocity and angular acceleration were the highest, followed by oblique front and lateral impacts, respectively. At lower impact energy levels, angular velocity and angular acceleration had variable behavior dependent upon impact direction. Across all impact energy levels ATD occipito-cervical bending moment was highest with front impacts, followed by oblique front and lateral impacts, respectively. Resultant ATD occipital force was similar in the lateral and oblique front impact directions but highest in response to front impacts across all impact energies.

The post hoc coefficient of variation (COV) presented herein (Table 2) indicated that low intra-test variation was present in all of the high energy impacts. At lower impact energy, more variability was present. Hence, $n = 6$ impacts for each condition was appropriate, although future studies under similar impact conditions should also assess COV in a similar post hoc manner to quantify variability.

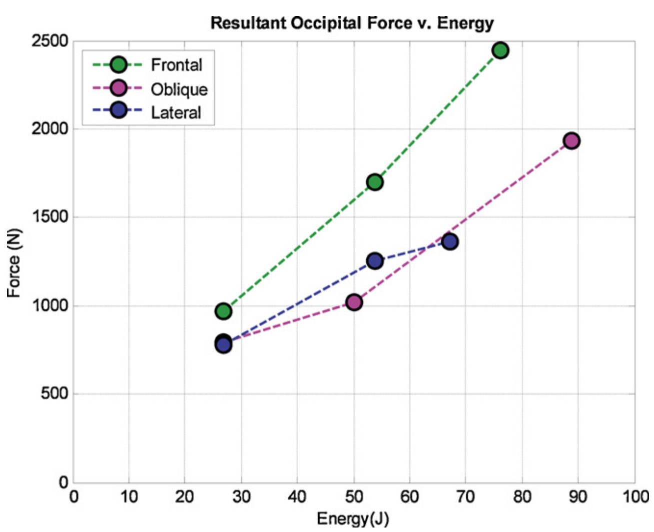

Fig. 6. Mean resultant ATD peak occipital force.

The results indicated that ATD head–neck stiffness plays a role in determining the relative magnitudes of impact dosage and injury risk parameters as a function of impact direction. The mean resultant ATD cg linear acceleration was always highest for lateral impacts. This might have been due to the lateral impact location, which was closer to the head cg, as opposed the oblique front or front impacts. This response may also have been attributed to the ATD head–neck complex being stiffer in the lateral direction, causing less head motion during impact and allowing for the linear acceleration to increase. The mean resultant ATD head angular velocity and angular acceleration results were very similar. At the lowest impact energy levels, both of these parameters were marginally affected by differences in impact direction. As impact energy increased, front angular velocity/acceleration increased faster than with either the oblique front or lateral impacts, respectively. It is interesting that the lateral impact direction was associated with a slower increase in such angular responses as energy increased. This is perhaps related to the ATD head–neck being stiffest in the lateral impact direction.

It is likely that ATD head–cervical spine stiffness influenced the occipital moment results, since both the front and oblique front impacts were associated with a more rapid increase in occipital moment than the lateral impacts. In the lateral direction, the smaller bending moment, compared with the oblique front and front impacts, might have indicated that the spine sustained overall bending as opposed to predominance of bending occurring at the occipito-cervical junction. When examining mean occipital force results, front impacts were always associated with the greatest resultant force. The oblique front and lateral impacts demonstrated a similar behavior at the lowest energy, while oblique front impacts had a larger force magnitude at the highest energy impact level.

Comparisons can be made with between the data presented here and prior dynamic human and ATD tests. In terms of omnidirectional load application, non-injurious dynamic human volunteer direct bending tests demonstrated the tolerable lateral cervical spine force to be nearly twice that of the tolerable oblique front force (Patrick and Chou, 1976). In a dynamic cadaveric helmeted impact study, front impacts were associated with occipital accelerations that were 50% higher than temporal region impacts (Got et al., 1978). When examining the Got et al. (1978) study, the cadaveric head acceleration behavior was in opposition to the results published herein, with the cadaver in the Got study producing anterior head acceleration approximately 50% of the lateral acceleration. But this cadaveric finding may have been due to the absence of active muscle tensing, with only passive osteoligamentous resistance providing occipito-cervical and cervical stability. Hence, their cadaveric response was more robust in the lateral plane and less robust in the sagittal plane.

When considering loading in a single plane, dynamic inertial human volunteer motion studies indicated that occipital bending moments up to 42 N m could be sustained without injury (Nahum et al., 1968). Dynamic cadaveric inertial and directly applied extension studies demonstrated that the onset of occipito-cervical AIS2+ 'moderate' injury (Association for the Advancement of Automotive Medicine, 2005) occurred between 43 N m and 100 N m (Pintar et al., 2005; Hu et al., 1977; Nightingale et al., 2002, 2007). For lateral motion tolerance during dynamic inertial tests, volunteers endured resultant occipital moments up to 55 N m without injury (Wismans and Spenny, 1983)—while dynamic inertial cadaveric specimens have shown AIS2+ occipito-cervical injury at resultant moments of 247 N m (McIntosh et al., 2007). In torsional loading, volunteers endured inertial occipito-cervical bending moments up to 20 N m without injury, while dynamic cadaveric specimens were injured at 17–23 N m (Wismans and Spenny, 1983; Chang et al., 1992; Myers et al., 1991). Cadaveric specimens tested in pure torsion have endured bending moments up to 14 N m prior to failure (Goel et al., 1990). All of these omnidirectional and single-plane loading studies, with the exception of Got et al. (1978), indicated that the occipito-cervical junction stiffness is the greatest in response to lateral loading, followed by oblique front and extension loading, respectively.

With the above-referenced published data in mind, the ATD head–neck was the stiffest in response to lateral impacts, which mirrored the lateral motion tolerance findings from the aforementioned literature. However, the ATD had a stiffer response to oblique front loading and less stiff response to extension loading. This differed from the literature. Regardless, it is very important to note that the Hybrid III ATD head–neck was validated solely in the sagittal plane for flexion-extension responses. Biofidelic occipito-cervical response out of sagittal plane was never specifically optimized for this ATD. In future work, as the Hybrid III ATD is investigated as a possible testing surrogate for direct head impacts in NOCSAE athletic helmet testing (National Operating Committee on Standards for Athletic Equipment (NOCSAE), 2006) or elsewhere, great care should be taken to ensure that the head–neck response is quantified and that the ATD performs as intended. Further, any test methodology that incorporates a combination of two types of testing surrogates, such as using a Hybrid III ATD or Hybrid III neck form attached to a NOCSAE head form (Gwin et al., 2010; Bartsch et al., 2011), should ensure that such a non-standard head–neck attachment does not influence testing results by potentially over-compressing standard Hybrid III neck rubber nodding blocks or producing unrealistic post-impact motions. Finally, the proposed STAR football helmet evaluation system, that utilizes combinations of NOCSAE drop testing with on-field impact exposure data (Rowson and Duma, 2011), cannot be directly compared with our testing of the Hybrid III because of differing impact interfaces, head forms, post-impact trajectory, force-deflection characteristics and our lack of real-world exposure data. However, it is possible that any future evaluation system that incorporates a head–neck form surrogate may result in theoretical injury risk safety ratings different than STAR because of influences from the end conditions present at the head–neck form attachment. These potential differences provide ample opportunities for future study.

4.1. Limitations

There were three main drawbacks to this study. First, impact location on the ATD head could have played a role in the response. While front and oblique front impacts struck the same elevation on ATD head, the lateral impact struck approximately three inches superior to the front impact. This was closer to the transverse cg plane and might have affected rotational responses by generating higher lateral occipital bending moment and head angular velocity and angular acceleration. Unfortunately, the ATD in lateral impacts could not have been struck in the same transverse plane as the front and oblique front impacts because this would have required the impactor to strike the ATD neck and head simultaneously.

The second drawback is related to the fact that the total head–neck rotational position response could not be quantified. While prior extension, oblique front, torsional and lateral bending studies have investigated total (Mertz and Patrick, 1967, 1971; Hu et al., 1977; Myers et al., 1991; Patrick and Chou, 1976; Wismans and Spenny, 1983; Wismans et al., 1986; Yoganandan et al., 2000; Seeman et al., 1986; Clemens and Burow, 1972) and local (Panjabi et al., 1988, 2001; Nightingale et al., 2002, 2007; Pintar et al., 2005; Goel et al., 1990; Lysell, 1969; Camacho et al., 1997; Dibb et al., 2006; Chang et al., 1992; Shea et al., 1991) head–cervical spine rotational position, we could only monitor head rotation via the angular rate sensor. Thus, a true value of head–cervical spine stiffness, with comparison between force or moment and segmental rotation, was not possible.

Third and finally, the ATD was impacted numerous times over the course of a single day, without a predetermined rest time or temperature/humidity monitoring between impacts. While the effects of conducting these impacts in a high-throughput manner versus over the course of many days would likely be very small, the effects of repeated impacts, nonetheless, were not quantified.

5. Conclusions

The Hybrid III 50th percentile male anthropomorphic test device (ATD) is the most widely used human impact testing surrogate and has historically been used in automotive or military testing. More recently, this ATD is finding use in applications evaluating athletic helmet protectivity, quantifying head impact dosage and estimating injury risk. But ATD head–neck response has not been quantified in omnidirectional athletic-type head impacts absent axial preload. It is probable that headgear injury reduction that can be quantified in a laboratory is related to a number of interrelated kinetic and kinematic factors, such as head center of gravity linear acceleration, head angular acceleration, head angular velocity, occipito-cervical mechanics and neck stiffness. Therefore, we characterized ATD head–neck dynamic response to direct head impacts in a series of front, oblique front and lateral head impacts.

Pendulum impact energy ranged from 27 J to 89 J and momentum varied from 14 N s to 25 N s. For all impacts, Hybrid III ATD resultant head center of gravity (cg) linear acceleration was 52–247 g, resultant head angular velocity was 14.0–26.4 rad/s, resultant head angular acceleration was 4080–9710 rad/s^2, resultant occipital condyle force was 778–2440 N and resultant bending occipital moment was 20.6–111 N m, respectively.

Key findings were: (1) the impacts producing highest ATD resultant cg linear acceleration resulted in the lowest resultant occipito-cervical spine bending moment/force. (2) Resultant ATD head angular velocity and angular acceleration did not appear coupled to impact direction at lower impact energy levels, while these parameters were coupled to impact direction at higher energy levels. (3) And the Hybrid III ATD had progressively increasing occipito-cervical stiffness in response to extension, torsion and lateral bending, respectively.

In future studies with a Hybrid III under similar impact conditions, protective padding effects on impact dosage and injury risk should be separable from impactor effects. Additionally, in future biofidelic head impact testing using Hybrid III ATD or non-standard 'crossover' surrogates combining NOCSAE head form with Hybrid III neck form, the surrogate head–neck response must be well characterized and appropriate for such testing.

Because the ATD neck influenced head and neck impact dosage parameters, testing agencies, manufacturers and researchers should consider using the Hybrid III head form attached to a neck as a more biofidelic means to quantify head and neck injury risks as opposed to systems that do not utilize a neck. This heightened understanding of Hybrid III ATD head–neck response, and consideration of order of stiffest axes in the lateral, oblique and extension directions, respectively, should aid in the development of head and neck injury impact testing standards.

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