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Modelling Head Injury due to Unmanned Aircraft Systems Collision: Crash Dummy vs Human Body

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Recent developments in the concept of UAS operations in urban areas have led to risk concerns of UAS collision with human. To better understand this risk, head and neck injuries due to UAS collisions have been investigated by different research teams using crash dummies. Because of the limitations in biofidelity of a crash dummy, head injury level for a crash dummy impact may differ from the human body impact. Therefore, the aim of this paper is to investigate differences in head and neck injuries subject to UAS collision with an often used crash dummy and a human body. To perform such investigation, multibody system (MBS) models have been used to simulate UAS impacts on validated models of the crash dummy and the human body. The findings confirm the moderate risks of head and neck injuries that have been reported. However, neck load differs significantly between the crash dummy model and the human body model, and the human body model sustains larger head injury but smaller neck injury compared to the crash dummy model.

Nomenclature

AIS	=	abbreviated injury scale
ASSURE	=	Alliance of System Safety of UAS through Research Excellence
AOA	=	angle of attack
ATD	=	anthropomorphic test devices (crash dummies)
CG	=	center of gravity
EASA	=	European Aviation Safety and Agency
FMVSS	=	Federal Motor Vehicle Safety Standards
HIC	=	head injury criteria
LNL	=	lower neck load criteria

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MBS	=	multibody systems
NCAP	=	New Car Assessment Programme
NHTSA	=	National Highway Traffic Safety Administration
N_{ij}	=	neck injury criteria
N_{km}	=	neck force criteria
N_{TE}	=	neck tension/extension injury criteria
N_{TF}	=	neck tension/flexion injury criteria
N_{CE}	=	neck compression/extension injury criteria
N_{CF}	=	neck compression/flexion injury criteria
UAS	=	unmanned aircraft system

I.Introduction

Unmanned aircraft systems (UAS) are expected to operate in low-level airspace in an urban environment where population density is high. The risk from such implementation has given rise to the question of safety of people on the ground. This motivates efforts to understand the impact severity of drone collision on human through analytical or experimental approaches. In impact experiments, an anthropomorphic test device (ATD), or crash dummy, is widely used as a representative substitution of a real human body. The Alliance of System Safety of UAS through Research Excellence (ASSURE) research group conducted a series of impact drop test using DJI Phantom III UAS on the Hybrid III crash dummy at various UAS impact attitudes and speeds.[1–3] These tests provide valuable insights into head and neck injury from UAS collision. Campolettano[4] also performed a series of live flight test and impact drop test using three different UAS weight classes on an instrumented Hybrid III crash dummy. The aim of the test was to estimate the range of head injury risks to humans due to UAS impact.

Even though crash dummies are based on the human body, for road accidents it has been shown that limitations in biofidelity of the dummies can result in different biomechanical head and neck responses comparing to the real human.[5] Human body neck complex is the spine which is a mechanical structure composed of bony vertebrae, ligaments and intervertebral discs.[6] It is a flexible structure with a primary function to protect the spinal cord and nerve roots while carrying loads and perform the physical motion. The Hybrid III neck is designed to represent the cervical human spine by connecting the head and torso through a rigid attachment. The neck itself is a one-piece column made of rubber separated by aluminium discs and there is no inherent curvature to the Hybrid III neck column.[6]

Based on experimental work by Sances[7], a comparison of inverted drops on the Hybrid III crash dummy and on human cadavers showed that the dummy neck was two to four times stiffer than human cadavers. Additionally, an experiment by Sances[8] indicated that the crash dummy system transmits about 70-75% of the applied force from the head or upper neck to the lower neck area. On the other hands, only about 20-30% of the applied force was transmitted from the head to the lower neck in the study on a human cadaver. Such differences can lead to a discrepancy in head injury level between a crash dummy used in testing and a human.

In any investigation to determine an impact severity of a particular vehicle, it is vital that the measuring instrument is appropriate to serve the investigation objective. In this case, it is important to know whether the Hybrid III crash dummy is a suitable measuring instrument for an investigation on UAS collision severity and can realistically represent the human body. If the discrepancy between the Hybrid III crash dummy and the real human body is significant, then it is important to address the scale of such difference and an appropriate conversion method should be proposed. Therefore, the primary aim of this paper is to investigate the different head and neck injury levels between the Hybrid III crash dummy and the human body due to UAS collisions by using validated models of the Hybrid III crash dummy and the human body.

This paper is organized as follows. Section II describes the analysis methods including the models used in the simulation and analysis. Section III presents the comparative results from the models developed and simulated in MADYMO. Section IV and V presents the discussion of the results and the conclusion, respectively.

II.Modelling and Simulation Approach

A. UAS, Crash Dummy and Human Body Models

For a comparison of injuries from UAS collision impact of the Hybrid III crash dummy and the human body, the numerical simulation models have been developed and implemented within the software called MADYMO. A typical UAS model chosen for this study was the DJI Phantom III UAS with a take-off weight (W_0) of 1.28 kg. For this UAS model, a multibody system (MBS) model, as shown in Figure 1, was developed and validated[9]. The MBS UAS model was validated for an impact case on the Hybrid III crash dummy using the crash test data from the ASSURE research group. An example of the validation results is shown in Figure 2.

To simulate a crash test, the UAS MBS model is coupled with the 50th percentile Hybrid III crash dummy model and the 50th percentile human body model in MADYMO as shown in Figure 3. "50th percentile" refers to the size of the crash dummy or human body which is equivalent to the average North American male. The Hybrid III crash dummy is represented by the multibody model with facet surface and distributed with MADYMO (filename: d_hyb350el_Q, version 2.0). This Hybrid III crash dummy model has been validated against a real Hybrid III crash dummy at various load directions.[10,11] The human model is also distributed with MADYMO (filename: h_occ50fc, version 5.2) and was originally published by Happee[12,13]. The human body model is also a multibody system model with passive muscle model and the skin is modelled using a facet surface which is a mesh of shell-type massless contact elements. The skeleton of the human body model consists of chains of rigid bodies connected by kinematic joints. The biomechanical data including joint characteristics and mechanical properties are based on biomechanical data and are validated using volunteer and post mortem human subject (PMHS).[14]

Figure 1. UAS model for impact modelling: (a) a real-world system and (b) DJI Phantom III Standard multibody system model developed in MADYMO by the author.[9] The two landing skids are neglected in the model since they are not in contact with the human head in impact cases that are investigated.

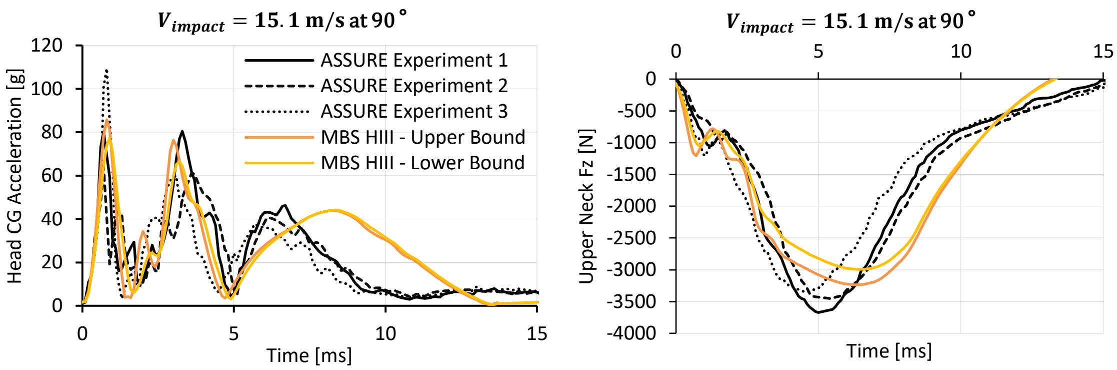

Figure 2. Example of the validation results of the UAS MBS model at various impact angles and velocities impacting the Hybrid III dummy[9].

B. Simulation Setup

This paper focuses on an impact to the head of the crash dummy and the human body as it is the most vulnerable part of the body. In the model set up, both the crash dummy and the human body models are seated on non-smooth rigid seats with full back support. The velocity vector of the UAS model is aligned with the head centre of gravity (CG) of the crash dummy and the human body. The UAS angle of attack was fixed at 0º from the horizon axis for all impact case.

Crash simulations were performed by varying two main parameters; impact velocity (V_{impact}) and impact angle (θ_{impact}). V_{impact} is varied from 2 to 20 m/s with an increment of 2 m/s. θ_{impact} is set to 0º (horizontal impact), 45º (angle impact) and 90º (vertical impact). The horizontal and angle impact cases represent a loss of control failure model in which the UAS flies directly onto the human head. The vertical impact case represents a failure mode in which a UAS falls to ground uncontrollably due to the complete loss of power. The simulation was run on a 2.6 GHz processor, resulting in a computational time of approximately 60 s and 120 s for the human body model and the crash dummy model, respectively.

To assess the risk of serious head injury such as traumatic brain injury or skull fracture, the head injury criterion of 15 ms impact time (HIC_{15}) was used.[15,16] The HIC, which is a measure of the likelihood of head injury due to impact, can be calculated using equation 1. It is an integral of the resultant head acceleration within a time range that maximizes the HIC value. The time range limit is often 15 or 36 ms – 15 ms time range limit is chosen as a more appropriate choice for this short-duration impact study. Functionally, the HIC represents the peak average power delivered to the head.[17] Based on FMVSS and NCAP, the HIC value of 700 is considered to be a minimum safety standard where the probability for skull fracture ($AIS \geq 2$) for mid-sized male is 31%.[18] To measure head acceleration, both the crash dummy and the human body models are instrumented with 3 single-axis accelerometers positioned at the CG of the heads. A low-pass filter with a channel frequency class (CFC) 1000 is applied to linear acceleration curves from the head CG accelerometers.

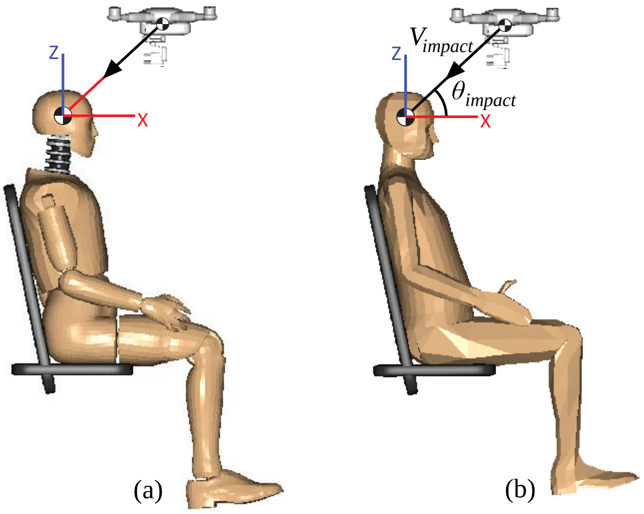

Figure 3. Simulation setup in MADYMO of UAS collisions on (a) the Hybrid III crash dummy model and (b) the human body model.

$$HIC = \left\{ (t_2 - t_1) \left[\frac{1}{t_2 - t_1} \int_{t1}^{t2} a(t)dt \right]^{2.5} \right\} max \quad (1)$$

Furthermore, the N_{ij} is a neck injury criterion which considers the upper neck force and moment proposed by the National Highway Traffic Safety Administration (NHTSA).[19] The "ij" represents indices for the 4 injury mechanisms; namely N_{TE}, N_{TF}, N_{CE} and N_{CF}. The first index represents the actual load (tension or compression) while the second represent sagittal plane bending moment (neck flexion or extension). The current performance limit of the N_{ij} is 1 which represents a 22% risk of greater than the Abbreviated Injury Scale (AIS) level 3.[20] The equation for the N_{ij} is:

$$N_{ij} = \left| \frac{F_z}{F_{int}} \right| + \left| \frac{M_Y}{M_{int}} \right| \quad (2)$$

III. Modelling Results for Hybrid III Crash Dummy vs. Human Body

C. UAS Impact Injuries

1. Overall Kinematic of Head/Neck System

From the simulation, an overall kinematic of the head/neck of the crash dummy and the human body is presented in Figure 4 for different impact cases. By comparing the head trajectory in the horizontal impact case, the Hybrid III head/neck complex can realistically mimic the movement of the human body head/neck. The motion observed in this impact case is mostly rotational in the extension direction and insignificant on the translational motion. The difference starts to be observable when the applied load direction increases toward the vertical direction (90°) as shown in angle and vertical impact cases. In angle impact case where applied load direction is approximately 45°, the downward travel of the crash dummy head is small comparing to the human body as shown in Figure 4(b). Head rotational direction also differs where extension rotation occurs in the crash dummy and flexion rotation on the human body. This is because the Hybrid III crash dummy is designed primarily for frontal impact analysis and the head/neck construction holds anatomical difference comparing to the human body head/neck construction.

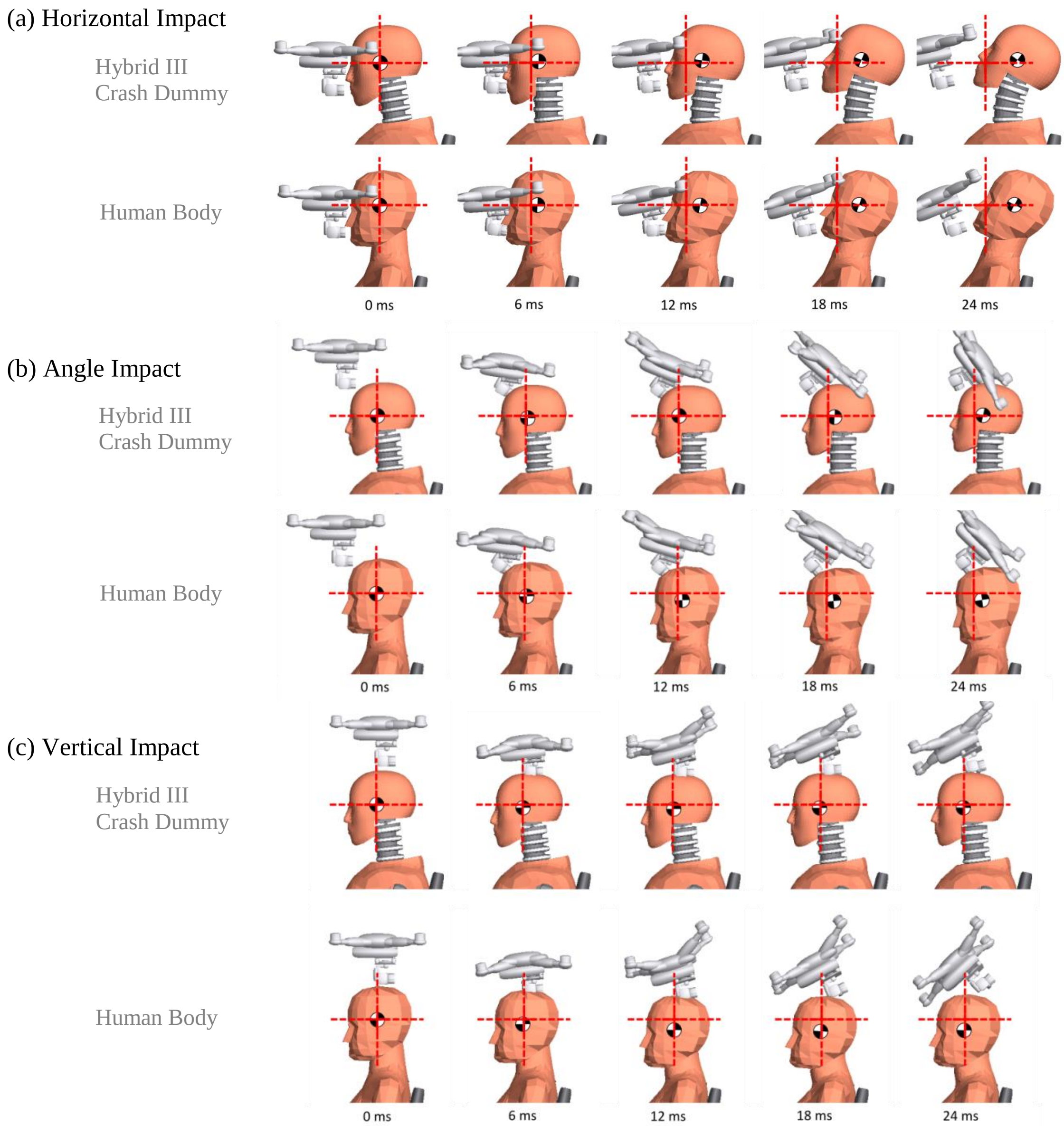

Figure 4. Comparison of impact sequences between the Hybrid III crash dummy and the human body for (a) horizontal, (b) angle and (c) vertical impact at 18 m/s impact velocity.

The effect of the stiff dummy neck system is apparent in vertical impact case as shown in Figure 4(c) where the downward travel of the head is significantly different between the crash dummy and the human body. Trajectory comparison shows the human head travels further down and over a longer period of time, while the crash dummy head vertical displacement is small and with a faster rebound. In addition to larger head displacement, the human head also rotates in extension direction when full vertical neck compression is reached, while such rotation is minimal in the crash dummy head.

2. *Head Injury*

Head injury criteria or HIC_{15} is an integral of head CG acceleration of a crash dummy or the human body heads. Before any difference in HIC_{15} can be realized, the difference in head acceleration between the crash dummy and the human body has to be addressed. Figure 5 shows a comparison of head CG acceleration between the crash dummy and the human body models at various impact angle and at an impact velocity of 18 m/s. Head CG acceleration for horizontal, angle and vertical impact cases at various impact velocities are presented in the appendix. Horizontal impact in Figure 5(a) shows that the crash dummy and the human body models produces similar head acceleration characteristics. The maximum peak acceleration matches well with only 3% difference, and the phase corresponds well between the two models.

For angle impact case, both models produce similar trend and phase as shown in Figure 5(b). The first impact peak matches well with one another with a 5% peak difference. This first peak represents the contact force generated at the initial impact moment. Since both crash dummy and human skin shares similar surface stiffness, these contact forces are similar in magnitude. However, the second peak is higher for the human body comparing to the crash dummy with a peak difference of 25%. The second peak occurs when the entire UAS fully compresses and impact energy is fully transferred to the head. The last peak, which represents the restitution phase of the neck which is compressed, also shows a significant difference of 90%.

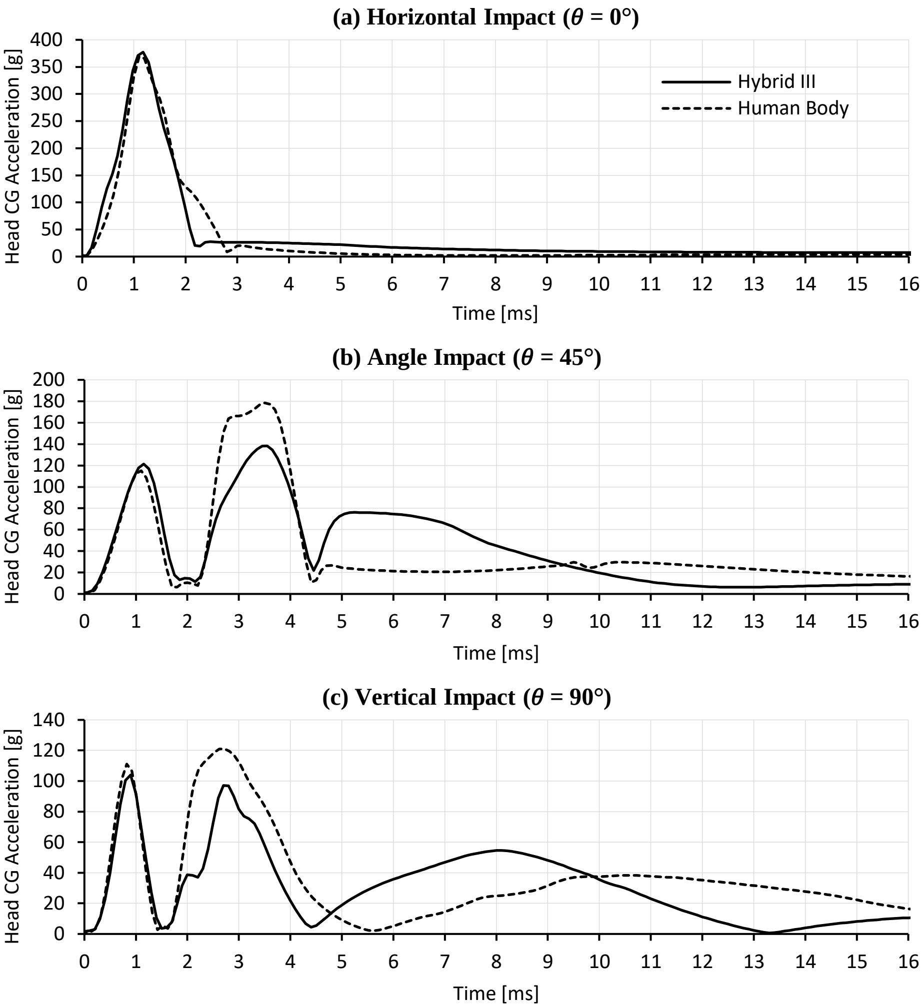

Figure 5. Comparison of head CG acceleration between the Hybrid III crash dummy and the human body models at (a) horizontal, (b) angle and (c) vertical impact cases at an impact velocity of 18 m/s.

For vertical impact case, both models share a similar trend with three observable peaks as shown in Figure 5(c). The phases of the first two peak match well between the two models. However, the third peaks are 2.5 ms out of phase with one another. For the peak differences, the first peak is 3% different and the second peak is 13% different between the models. The last peak of the vertical impact case also shows a similar result to the angle impact case where the neck of the human body rebound less and slower compared to the crash dummy.

Figure 6 shows a comparison of the calculated HIC_{15} between the Hybrid III crash dummy and the human body model. For horizontal impact angle shown in Figure 6(a), the crash dummy produces similar results comparing to the human body model. For both models, the graph shows that the HIC_{15} increases non-linearly as impact velocity increases. The maximum difference is less than 5% between the two models at an impact velocity of 18 m/s. For angle impact case, both models produce similar trends of the HIC_{15}, but 33% difference of the HIC_{15} at an impact velocity of 18 m/s. Such percentage error increases as impact velocity increases. Percentage difference is also significant in the vertical impact case where the maximum difference is 21% at an impact velocity of 18 m/s.

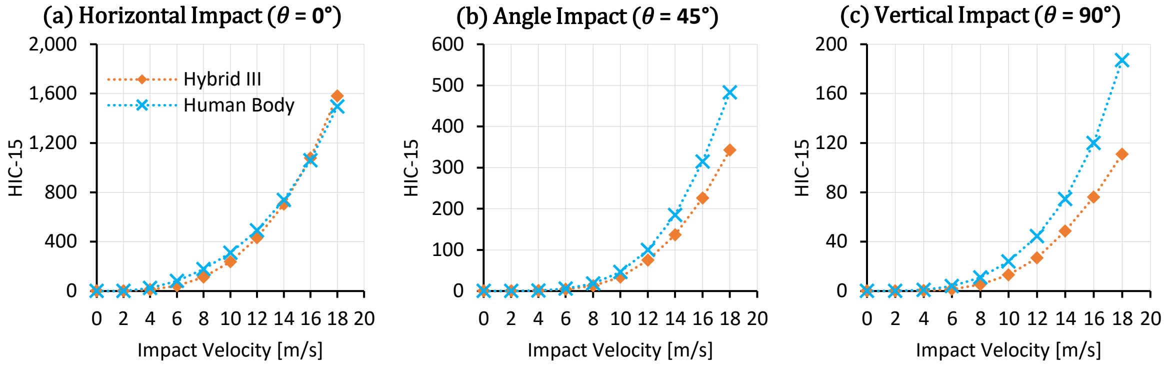

Figure 6. Comparison of HIC_{15} between a crash dummy and the human body models at (a) horizontal, (b) angle and (c) vertical impact cases.

Furthermore, HIC_{15} can be interpreted into a physical form of injury which is represented by the probability of the Abbreviated Injury Scale (AIS) level. The AIS is an anatomical-based coding system created by the Association for the Advancement of Automotive Medicine to classify and describe the severity of injuries.[21] By using the conversion chart by Mertz[22], the HIC_{15} can be converted to the probability of AIS level 2 and 3 which corresponds to the probability of skull fracture and brain injury, respectively. Table 1 shows the conversion of HIC_{15} to AIS injury level for horizontal, angle and vertical impact cases at 18 m/s impact velocity. For horizontal impact case, the percentage probability of AIS level 2 and 3 reaches 68.2% and 91.4%, respectively – this means that skull fracture and brain injury in highly probable. For angle and vertical cases, the AIS level 2 and 3 are low and considered to be within the safe threshold.

Table 1. HIC_{15} and AIS levels of the Hybrid III and the human body at various impact angle and at 18 m/s impact velocity.

θ_{impact}	Model	HIC_{15}	$AIS \geq 2$ (Skull Fracture)	$AIS \geq 3$ (Brain Injury)
(θ = 0°) Horizontal	Hybrid III	1580	68.2%	91.4%
	Human Body	1493	65.8%	88.4%
(θ = 45°) Angle	Hybrid III	342	9.1%	6.3%
	Human Body	482	17.7%	11.8%
(θ = 90°) Vertical	Hybrid III	110	0.4%	0.8%
	Human Body	187	2.0%	2.3%

3. Neck Injury

Neck responses between the Hybrid III crash dummy and the human body are different due to the difference in biofidelity of the neck anatomy. In a crash dummy, the neck complex is a segmented rubber and aluminium construction.[23] This results in the dummy neck to be less compliance comparing to the human neck in a vertical direction. The difference can be seen in force/moment transferred to the neck system from the head. Figure 7 shows the difference in upper neck force in Z-direction (F_z) and moment about the Y-axis (M_Y) between the Hybrid III crash dummy and the human body. The full set of upper neck forces and moment for various impact angle and velocities is presented in the appendix. The crash dummy peak upper neck force in the Z-direction is higher than that in the human body by approximately 87%, 85%, and 56% for horizontal, angle and vertical impact cases, respectively.

In the model, the head of both the crash dummy and the human body models are modelled as a rigid sphere without any internal deformation such as the skull or brain deformation. This means that the force transfers from the head to the neck system in the crash dummy are substantially higher than in the human body. Furthermore, upper neck moment M_Y in the crash dummy is significantly higher than the human body in horizontal impact case by 114%. Figure 4, which illustrates the head/neck movement at different time steps, shows the difference in the initial movement of the head and neck between the crash dummy and the human body.

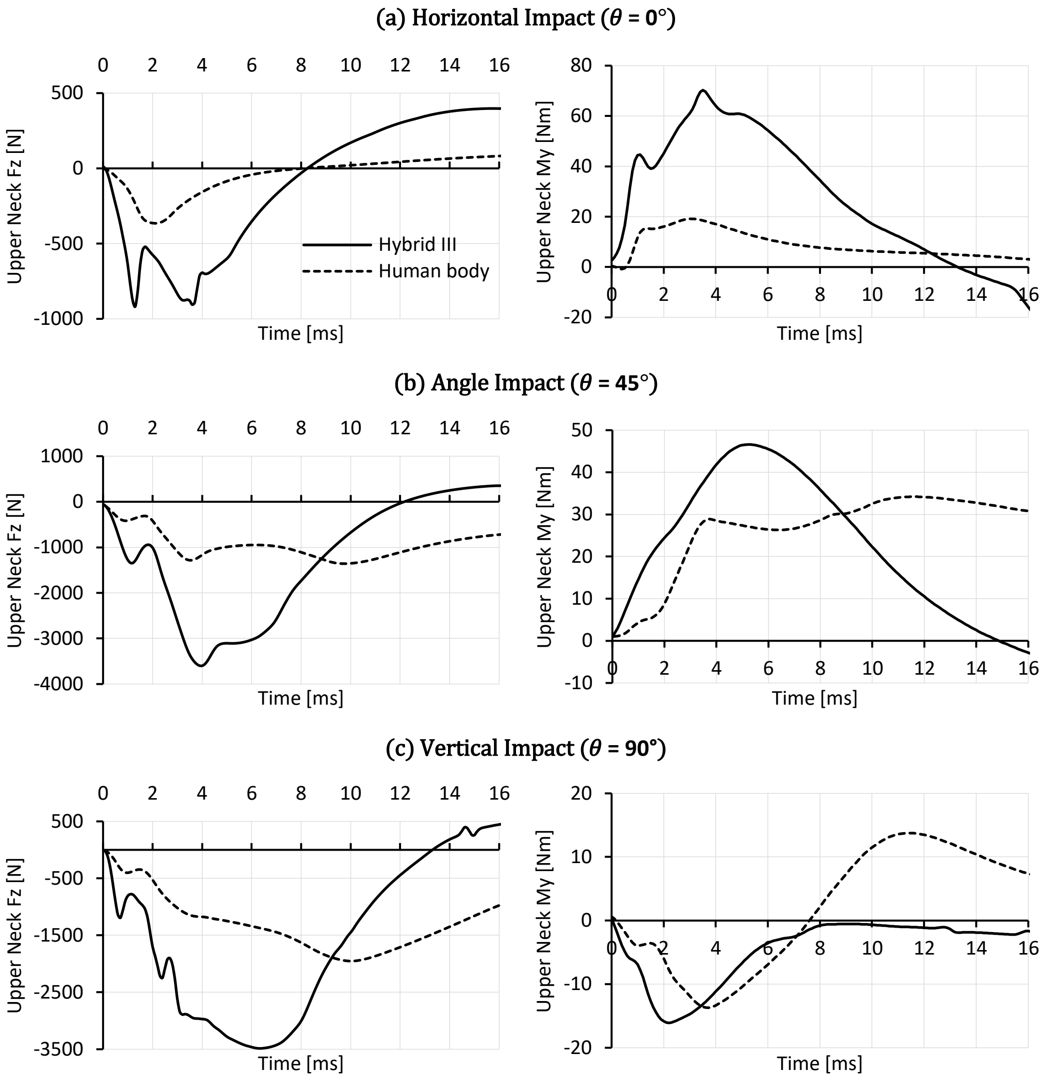

Figure 7. Comparison of upper neck force in Z-direction and moment about Y-axis between the Hybrid III and the human body models at (a) horizontal, (b) angle and (c) vertical impact cases at an impact velocity of 18 m/s

Since the human body neck is made of small vertebrae, it allows more initial translational motion between inter-vertebral disc along the horizontal line before rotation when comparing to a crash dummy. The crash dummy neck, on the other hand, is made of rubber and aluminium discs that allow rotation. This does not permit any translation between disc in the neck system. Therefore, the upper neck moment M_Y of a crash dummy is larger than the human body. In angle impact case, a similar response to the horizontal impact case is observed.

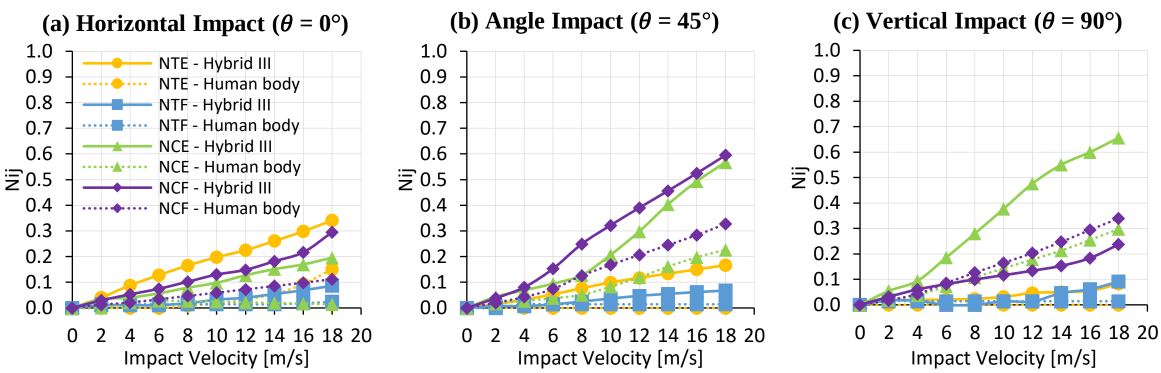

Figure 8. Comparison of neck injury criteria (N_{ij}) between the Hybrid III crash dummy and the human body models at (a) horizontal, (b) angle and (c) vertical impact cases.

Table 2. N_{ij} and AIS levels of the Hybrid III and the human body at various impact angle and at 18 m/s impact velocity.

θ_{impact}	Model	N_{ij}	$AIS \geq 2$ (Broken Neck)	$AIS \geq 3$
(θ = 0°) Horizontal	Hybrid III	0.341	16.2%	7.2%
	Human Body	0.1509	13.3%	5.1%
(θ = 45°) Angle	Hybrid III	0.595	20.7%	11.4%
	Human Body	0.3277	15.9%	7.0%
(θ = 90°) Vertical	Hybrid III	0.656	21.9%	12.6%
	Human Body	0.298	15.5%	6.7%

A neck injury can be assessed through various neck injury criteria, such as N_{ij}, N_{km}, or LNL criterions. However, in this paper, since the impact direction induces head motion mainly in X and Z-directions (along sagittal plane), only the assessment and comparison of N_{ij} criterion will be investigated. As explained in the earlier section, N_{ij} criterion takes into account upper neck force F_z and upper neck moment M_Y. In horizontal impact, N_{ij} criterion shows no significant neck injury. However, the difference between a crash dummy and the human body is quite significant. By looking at the highest N_{ij} value, the difference in peak value of neck tension/extension criterion (N_{TE}) is approximately 77% in horizontal impact. In angle impact case, both the N_{CE} and N_{CF} are prominent and the difference in maximum values between a crash dummy and the human body is 119% and 86%, respectively. Lastly in vertical impact case, N_{CE} is the highest N_{ij} criterion and with the maximum difference of 75% between the crash dummy and the human body.

Furthermore, N_{ij} is converted to AIS level to assess the percentage probability of AIS level 2 and 3 as shown in Table 2. The maximum N_{ij} value is converted into the AIS probability using injury risk curve.[24] Based upon the consensus that no more than a 22% risk of AIS 3 or greater neck injury was acceptable, NHTSA applied the AIS 3 curve to select N_{ij} of 1.0 as the performance limit.[25] For horizontal impact case, AIS level 2 and 3 are not significantly different for the horizontal impact case. For angle impact case, the AIS level 2 from the Hybrid III model is 20.7% while the human body model is 15.9%. Vertical impact case also shows significance in the AIS level 2 and 3. For the Hybrid III, the AIS level 2 is 21.9% and the AIS level 3 is 12.6%. For the human body, the AIS level 2 is 15.5% and AIS level 3 is only 6.7%. Since in the Hybrid III, the impact force is transferred to the neck system and torso higher than in the human body, the neck injury shows to be much higher with the AIS level 2 almost exceeding 22%.

IV.Discussion

This research examined the difference in injury level between the Hybrid III crash dummy and the human body due to UAS collisions. A multibody system (MBS) model of the DJI Phantom III UAS is implemented to simulate collisions on the Hybrid III crash dummy and the human body models in MADYMO[9]. The MBS modelling technique allows fast simulation time with accurate results comparing to the finite element modelling technique and can capture accurately the overall kinematics of the system.

For the investigation on the difference injury level between the Hybrid III crash dummy and the human body, the results show that the crash dummy can produce a similar response and predict similar injury level to the human body in horizontal impact case. This reaffirms other works which show that the Hybrid III crash dummy is a well-designed ATD for horizontal load direction.[26,27] When load direction changes towards the vertical direction, the ability of the crash dummy to produce force response in the neck system similarly to the human body reduces. As can be seen in angle and vertical impact cases, the head injury prediction (HIC_{15}) can differ by almost 33% in which the human body sustains a higher injury. This stems from the difference in head acceleration of the crash dummy and the human body which is a result of difference neck compliance. The head models are identical since the heads of both models are represented by a rigid body with contact deformation, but without any internal deformation. Therefore, the differences in neck complexes of the two models affect the neck maximum deformation and resistance to head acceleration. The neck system in the Hybrid III is constructed by a straight column in which a higher impact force from the head is transferred to when comparing to the human head. The more compliance human body neck system is modelled realistically to represent the vertebrae structure with passive muscle force. This allows the head to travel faster in a downward direction with a less resisting force upward, resulting in larger head acceleration and lower neck force.

Based on a qualitative analysis of the impact sequences in Figure 4, head displacement and neck deformation in the human body is larger than the crash dummy's. The Hybrid III crash dummy has a stiffer neck system comparing to the human body which limits the neck deformation and resists downward head motion. A lack of biofidelity in the Hybrid III neck is attributed to high resistance to compressive force and bending of the neck and the torso[6], this leads to the tendency to over-represent axial compression injuries. This is confirmed by the neck injury analysis using the N_{ij} criterion which shows the dummy predicts 75% higher in maximum N_{ij} values comparing to the human body.

From the impact severity analysis perspective, choosing the Anthropomorphic Test Device (ATD) or crash dummy for a particular load case is vital to the accuracy of injury prediction. A wide range of crash dummies has been developed to account for various load cases, such as frontal, side, rear or vertical. The Hybrid III is a highly improved dummy with a realistic response to the real human especially in a frontal impact, but still has limitation to certain load cases such as vertical load case. With the human body model that has been validated against real human and cadavers, it is possible to realistically simulate various impact cases in all load direction.

Furthermore, the analysis results reflect the importance of choosing the impact scenario that represents the worst case scenario before the finding of impact severity can be concluded. To extend the analysis to cover larger scenario, other parameters need to be incorporated and investigated in future works, for example, off-set between UAS CG and head CG, UAS initial rotational velocity or yaw and roll angles. More importantly, the variation of mass, size and shape of UAS are influential parameters on injury severity.

V. Conclusions

When conducting an impact testing research, it is important to account for the type of crash dummy model used and recognize the accuracy limitation. The primary aim of this paper is to investigate the differences in head and neck injury levels between a 50th percentile Hybrid III crash dummy and a 50th percentile human body due to UAS collisions. The DJI Phantom III UAS was chosen as a representative UAS model used in this study. Impact modelling and simulation have been conducted to compare head and neck injury levels from UAS impact on the Hybrid III crash dummy versus the human body. The impact simulation and analysis use a validated multibody system (MBS) UAS model and validated MBS models of the crash dummy and the human body.

The findings conclude that the Hybrid III crash dummy accurately represents the human body in a horizontal impact case. As the angle of the load direction increases towards the vertical axis, the crash dummy tends to under-predict the head acceleration and over-predict the neck compressive force and moment when comparing to the human body. Therefore, caution should be taken when using the Hybrid III crash dummy for UAS impact testing in certain load directions. Nevertheless, UAS impact testing is understandably difficult to be performed on a human cadaver, and the use of a crash dummy is inevitable. In follow-on research, an approach to convert crash dummy injury levels to human body injury levels will be investigated. Complementary to this, differences between UAS impacts on human bodies of child and woman versus male will be investigated.

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VI. Appendix

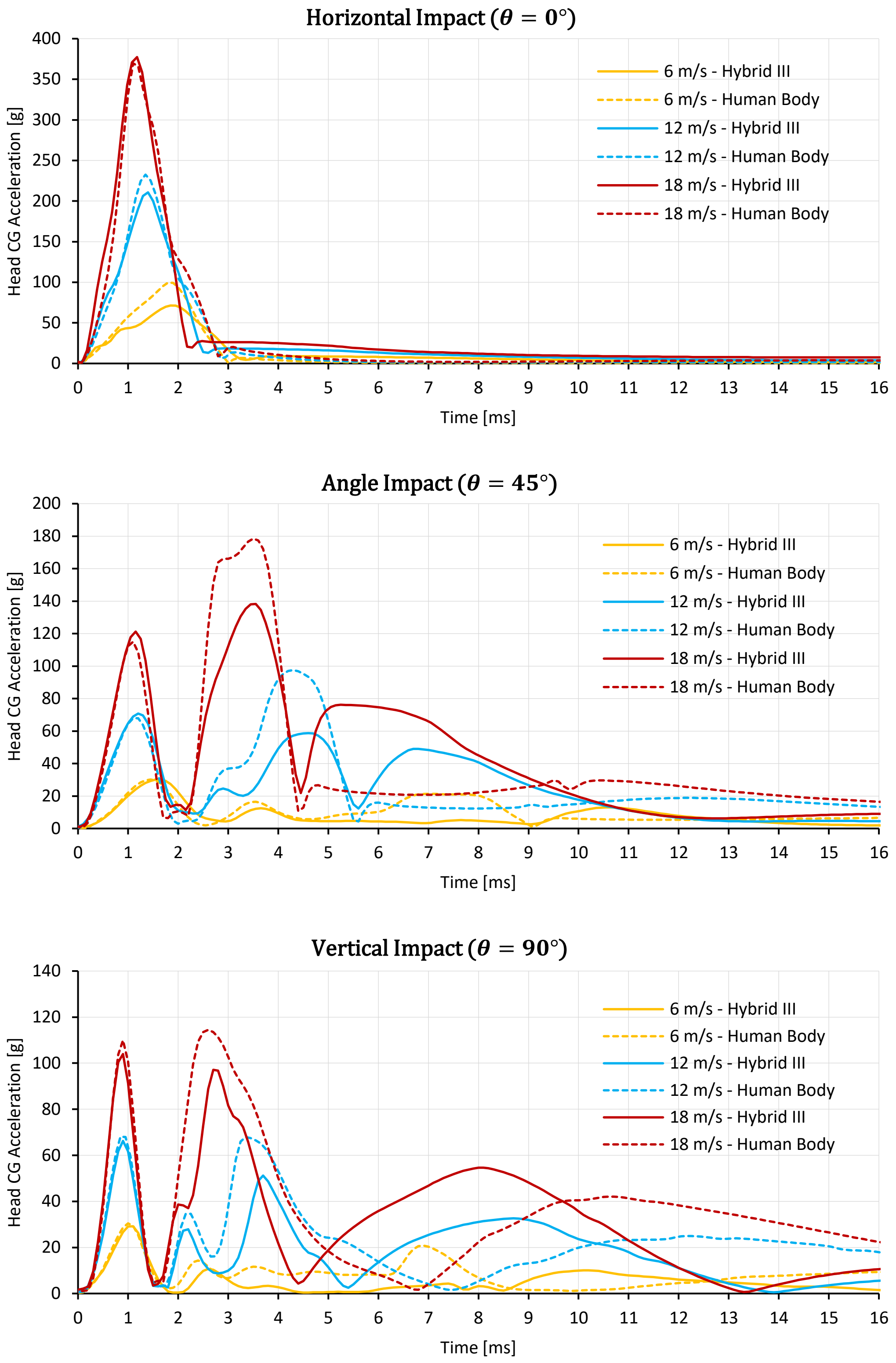

Figure 9. Comparisons of head CG resultant acceleration between the Hybrid III crash dummy and the human model at impact speeds of 6, 12 and 18 m/s for horizontal, angle and vertical impact cases

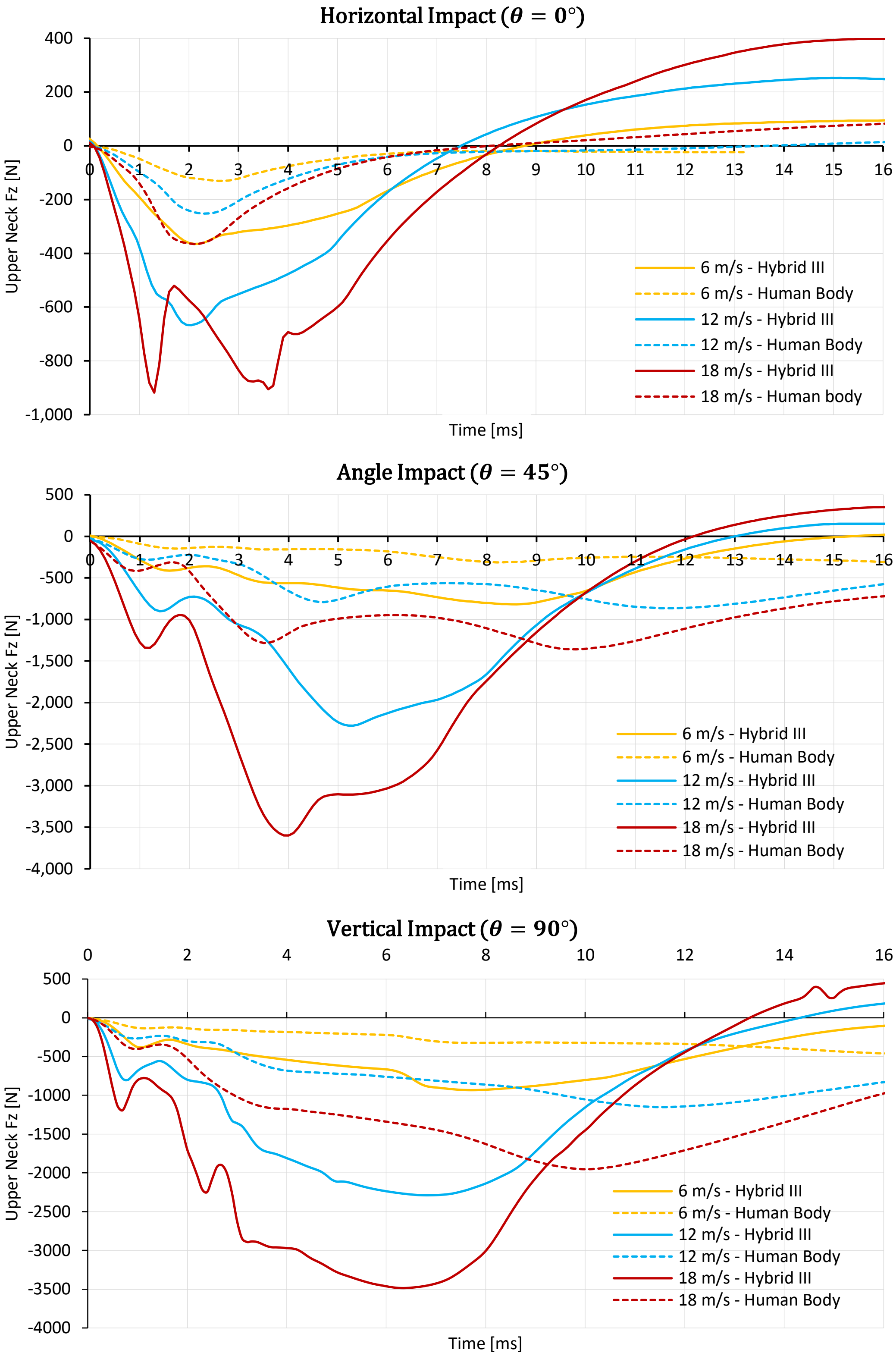

Figure 10. Comparisons of upper neck force in Z-direction between the Hybrid III crash dummy and the human model at impact speeds of 6, 12 and 18 m/s for horizontal, angle and vertical impact cases.

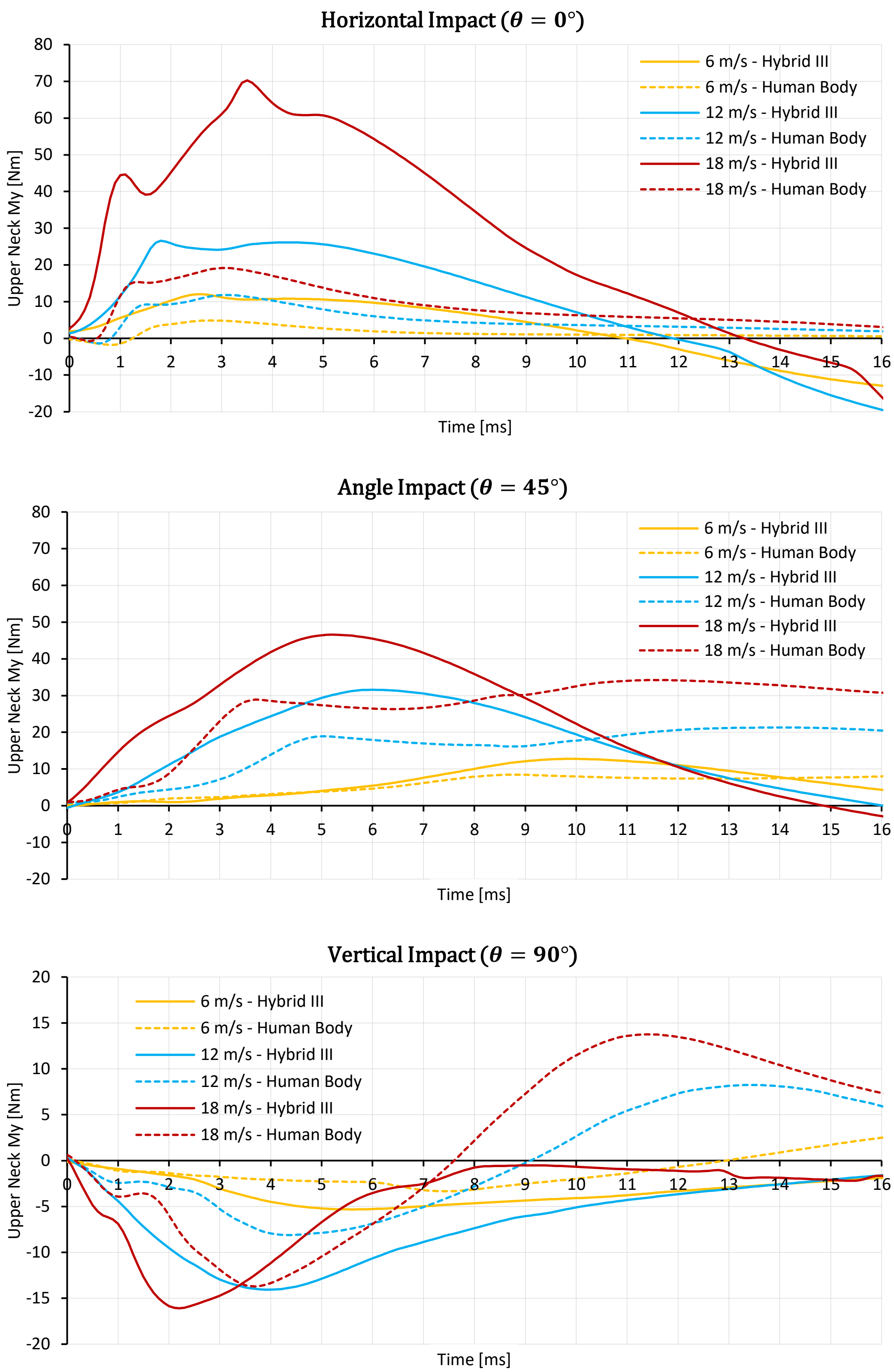

Figure 11. Comparisons of upper neck moment about Y-axis between the Hybrid III crash dummy and the human model at impact speeds of 6, 12 and 18 m/s for horizontal, angle and vertical impact cases.