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Response of an Impact Test Apparatus for Fall Protective Headgear Testing Using a Hybrid-III Head/Neck Assembly

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Abstract

A test method based upon a Hybrid-III head and neck assembly that includes measurement of both linear and angular acceleration is investigated for potential use in impact testing of protective headgear. The test apparatus is based upon a twin wire drop test system modified with the head/neck assembly and associated flyarm components. This study represents a preliminary assessment of the test apparatus for use in the development of protective headgear designed to prevent injury due to falls. By including angular acceleration in the test protocol it becomes possible to assess and intentionally reduce this component of acceleration. Comparisons of standard and reduced durometer necks, various anvils, front, rear, and side drop orientations, and response data on performance of the apparatus are provided. Injury measures summarized for an unprotected drop include maximum linear and angular acceleration, head injury criteria (HIC), rotational injury criteria (RIC), and power rotational head injury criteria (PRHIC). Coefficient of variation for multiple drops ranged from 0.4 to 6.7% for linear acceleration. Angular acceleration recorded in a side drop orientation resulted in highest coefficient of variation of 16.3%. The drop test apparatus results in a reasonably repeatable test method that has potential to be used in studies of headgear designed to reduce head impact injury.

Keywords

Head Impact; Helmet Impact Testing; Protective Wear; Traumatic Brain Injury; Linear/Angular Acceleration; Falls; Biomechanics

Introduction

This paper presents a methodology based upon a twin wire drop apparatus that includes measured angular kinematics in the test procedures used to assess the performance of protective headgear for persons subjected to falls. The current work presents the response of a test apparatus based upon the twin wire drop test as described in ASTM F1446¹ retrofit with a Hybrid-III anthropomorphic test dummy (ATD) head and neck assembly. The test

protocol includes measure of angular acceleration using a nine-accelerometer array in addition to the more common linear acceleration measure. Implementation of the device will potentially allow for the development of fall protective designs that simultaneously targets both angular and linear acceleration reduction.

Head injury due to impact from falls represents a significant and growing problem. Falls are a leading cause of head injury, especially in elderly, followed by pedestrian road accidents.² When left unprotected during a fall, head impact levels can reach upwards of 300 g (gravitational acceleration), which is the acceleration at which significant injury or even death can occur.³ A protective system for reducing head injury due to falls needs to be designed with specific criteria including injury protection level, thickness, stiffness, weight, and cost, among others. Use of higher impact energy levels in the design typically requires the resistive system to be thicker and/or stiffer.⁴ Systems that are too thick or stiff can be objectionable to the user owing to the lack of comfort and aesthetics. Unattractive fall protection helmets are often stigmatizing, which leads to nonuse and lack of compliance when their use is directed, for instance when prescribed by a medical doctor for a patient at risk for falls. Therefore, appropriate methods for prescribing impact levels and performance of product testing are required so that functional and desirable designs can be furthered.

The legacy in helmet design has focused on protection from normal impact forces and reduction of the linear acceleration component; however, studies have shown that resisting angular acceleration may be of equal or more importance in the reduction of diffuse types of head injury.⁵ Angular acceleration of the head can cause concussive types of injuries such as subdural hematomas and diffuse axonal injuries (DAI).⁶ Diffuse brain injuries such as DAI are the result of exceeding a critical strain of the axons due to excessive angular motion. Severity of the damage depends on the magnitude of the angular acceleration and duration of the impulse. As the duration and magnitude of the rotational motion increase, high strain occurs deeper into the brain causing axonal damage.⁷ A study by DiMassi⁸ demonstrated that a pure rotational motion of the head will generate considerably more strain than a pure translational motion, and combinations of linear and rotational acceleration will induce more strain than in cases with only rotational motion. DAI can cause loss of consciousness along with potential for permanent loss of physical function. Yoganandan et al.⁹ reported that the shape of the angular acceleration pulse has a local influence on brain strains. While linear and angular acceleration are typical indicators of the input to the head during contact, some researchers, for example King¹⁰, advocate the use of local brain responses such as brain strain and strain rate. However, at present standard testing methods incorporating brain strain are unavailable. Kimpara et al.¹¹ recommended the use of injury predictors based upon angular acceleration in addition to the traditional head injury criteria (HIC) that is used to primarily predict skull fracture and brain contusion. Accordingly, applying a test methodology for fall impact that accounts for both linear and angular acceleration in a repeatable manner will hopefully provide a pathway for improved, non-stigmatizing designs that significantly reduce brain injury.

Head Injury Predictors

Many head injury predictors have been suggested and studied over the years to determine which one most accurately and consistently predicts head injury in humans.¹² Some of these include the head injury criterion (HIC),¹³ Gadd severity index (SI),¹⁴ peak resultant translational acceleration of the center of gravity (CG) of the head,¹⁵ peak resultant rotational acceleration,¹⁶ linear impact velocity,¹⁷ angular impact velocity,¹⁸ generalized acceleration model for brain injury threshold (GAMBIT),¹⁹ head impact power (HIP),^{20,21} peak force,²² including time duration limits of several of the above. The abbreviated injury scale (AIS) to classify the severity of injuries²³ was first created in 1969, and is one of the most common anatomic scales for traumatic injuries.

In the current study, the HIC, rotational injury criteria (RIC), and power rotational head injury criteria (PRHIC)¹¹ are the three measures employed to assess head injury in addition to the maximum linear and angular acceleration. These measures will be described briefly herein. The HIC is one of the most widely accepted predictors of head injury and is based upon the translation acceleration magnitude. The mathematical expression for the HIC is given in Eq. 1 as follows:

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

where t_1 and t_2 are times in seconds during the acceleration-time history, $a(t)$ is the resultant translational acceleration of the head in g 's, and t_1 , t_2 are selected so as to maximize the HIC. In 2000, the national highway traffic safety administration (NHTSA) evoked limits that reduced the maximum time interval ($t_2 - t_1$) for calculating the HIC to 15 ms and is called HIC₁₅.²⁴

Kimpara et al.¹¹ presented a study of head injury predictors based upon angular accelerations. Included in their study are methods based on head kinematics and on finite element analyses. Two kinematic measures presented that use angular acceleration as injury measures are the RIC and the PRHIC. The RIC given in Eq. 2 is a rotational version of HIC with the translational acceleration $a(t)$ replaced by the angular acceleration $\alpha(t)$.

$$\text{RIC} = \left\{ \left[\frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} \alpha(t) dt \right]^{2.5} (t_2 - t_1) \right\}_{\max} \quad (2)$$

The PRHIC is based upon the HIP but includes only the rotation power terms in the HIP expression, which is called HIP_rot(t), given as:

$$\text{HIP}_{\text{-rot}} = I_{xx} \times \alpha_x \int \alpha_x \times dt + I_{yy} \times \alpha_y \int \alpha_y \times dt + I_{zz} \times \alpha_z \int \alpha_z \times dt \quad (3)$$

The PRHIC is then computed similar to the HIC with $\text{HIP}_{\text{-rot}}(t)$ replacing the linear acceleration magnitude $a(t)$.

$$\text{PRHIC} = \left\{ \left[\frac{1}{(t_2 - t_1)} \int_{t_1}^{t_2} \text{HIP}_{\text{-rot}} \times dt \right]^{2.5} (t_2 - t_1) \right\}_{\max} \quad (4)$$

It was observed that longer duration was typically required for the angular calculation and the maximum time increment ($t_2 - t_1$) was increased from 15 (as done for the HIC) to 36 ms. Use of HIC_{15} , RIC_{36} , and PRHIC_{36} for prediction of head injury in pedestrian accidents was proposed, but Kimpara et al.¹¹ concluded that more work is required to develop injury protection thresholds for the angular acceleration-based measures.

Probabilities of brain injury from several studies using the above kinematic measures are presented in Table 1 including the data from Fenner et al.²⁵ who presented injury standards related to helmets. Also included are injury predictors based upon sports-related injuries that were investigated by King et al.,¹⁰ Newman et al.,²¹ and Zhang et al.²⁶ In contrast, Funk et al.^{27,28} reported that curves previously created to assess mild traumatic brain injury (MTBI) risk for the national football league (NFL) are too conservative and proposed considerable higher limits. Rowson and Duma²⁹ reported a maximum acceleration and HIC indicator that was similar to Funk.

Rowson and Duma³⁰ presented a brain injury indicator to predict the probability of MTBI using a combination of linear, a , and angular, α , acceleration. Their risk function, CP, presented in Eq. 5 represents the combined risk in terms of regression coefficients, β_0 , β_1 , β_2 , and β_3 .

$$\text{CP} = \frac{1}{1 + e^{-(\beta_0 + \beta_1 \times a + \beta_2 \times \alpha + \beta_3 \times a \times \alpha)}} \quad (5)$$

Their evaluation of the coefficients is based upon Head Impact Telemetry System (HITS) and NFL data. The NFL data set was from impact reconstruction using ATDs where concussive impacts had an average of 98 ± 28 g and 6432 ± 1813 rad/s². The HITS data included 63,011 impacts where concussive impacts averaged 104 ± 30 g and 4726 ± 1931 rad/s². The regression coefficients were determined to be $\beta_0 = -10.2$, $\beta_1 = 0.0433$, $\beta_2 = 0.000873$, and $\beta_3 = -0.00000092$. Their analysis concluded that linear, angular, or combined data were all good predictors with linear acceleration and combined methods being significantly better than angular acceleration alone. Similar measures are currently in need for risk assessment of injury due to falls, especially in the elderly population.

Assessment of Head Impact

A major task in the development of a method to evaluate fall protection devices is to assess the input that occurs during an unprotected fall. Outside of actual human testing which is not viable for head impact due to a fall, methods used to assess human fall response must be interpreted with caution. O'Riordain et al.³¹ performed multibody dynamics study of falls using the MADYMO™ program for four cases of persons ranging from 11 to 76 years old with injury ranging from contusion to skull fracture. Results were shown to be heavily dependent on head contact characteristics, with default contact conditions resulting in linear accelerations ranging from 311 to 1015 g compared with a range from 243 to 435 g for alternate contact characteristics, where the angular accelerations ranged from 17,600 to 43,500 rad/s². Doorly et al.³² investigated 10 real life cases where a fall occurred with the subject standing and reconstructed each case using MADYMO™. It was observed that the multibody model gave fine representations of the real life cases; however, when the case was complex, the results were not ideal and showed a high variance due to the boundary and initial conditions. The calculations resulted in linear acceleration between 189 and 456 g, angular acceleration of 7400 to 49,200 rad/s² and HIC₁₅ ranging from 511 to 5951.

Hardy et al.³³ investigated the response of the human head to impact using cadavers and their results quantified head kinematics, internal pressures, and strain. Impact conditions were varied to produce differences in the relative angular acceleration and about half of the tests were helmeted and the others unprotected. Most of the tests were performed in the median plane with impact to the occipital region. When using an SAE CFC class 1000 filter,³⁴ the peak linear acceleration for the unprotected tests occipital impact ranged from 153 to 408 g with average of 280 g. The angular acceleration for these tests ranged from 7396 to 39,433 rad/s² with an average value of 20,113 rad/s². The HIC₁₅ ranged from 372 to 2540 with an average value of 1073. Significantly lower peak values were reported when computed subsequent to implementation of a CFC class 180 filter.

A study to ascertain head impact values for falls was performed using a pedestrian version of the ATD by Lloyd.³⁵ The peak value of linear acceleration magnitude and HIC₁₅ were recorded using a minimum of five trials per fall type. Table 2 presents a summary of the results of this study. ATDs were allowed to fall by dropping them from various configurations. An unprotected standing fall was accomplished by standing the ATD erect with its arms to the side and dropping the ATD onto a vinyl composition tile (VCT) covered concrete floor by releasing a cable that holds the ATD vertical. In the crumple fall, the ATD was placed in a kneeling position and was allowed to impact the floor. The crumple fall with protective arms was accomplished by placing the arms of the ATD in front of the head so that the arms break the fall.

Although not directly fall related, Walsh et al.³⁶ studied the influence of impact angle and location on head impact using a Hybrid-III headform and neck connected to a sliding table. The head was impacted using a pneumatic arm that was driven to a velocity of 5.5 m/s at impact. Data were sampled at 20 kHz and filtered using the SAE class 1000 protocol. Impact occurred at five different locations with the impactor impinging at four different angles to introduce differences between the linear and angular kinematics. Linear acceleration ranging from an average peak of 42.2 to 133 g resulted in angular acceleration

ranging from 3,840 to 12,860 rad/s². Increased relative angular acceleration compared with linear acceleration was shown to occur with the headform struck at a 45° angle toward the back in the transverse plane. This test method resulted in mean coefficient of variation for linear and angular measures of 4.2 and 6.4% respectively.

Impact Testing Methods

Most impact tests are designed to achieve a repeatable input condition and may not necessarily need to reproduce actual fall conditions. To the author's knowledge, at present, there are no standards that directly address head impact protection devices to mitigate fall injury. On the other hand, much work on impact testing protocols has been accomplished for motorcycle helmets and in the sports equipment industry using drop mechanisms,^{1,37–39} projectiles,⁴⁰ and sliding tables,^{41,42} for example. A drop type test is typically accomplished using either a twin wire drop tower or monorail system as described in the ASTM F1446 standard.¹ Head forms are sized to the helmet being tested and are made of rigid materials such as magnesium or titanium or, in some cases, a relatively stiff urethane may be used.

Anthropomorphic test dummies (ATDs) are finding increasing use in impact injury studies and are also being used increasingly in sports injury analysis and by the US military.⁴³ Bartsch et al.⁴⁴ recently studied the response of a 50% Hybrid-III head/neck assembly for use in testing of sports helmets using a head/neck assembly mounted rigidly to a base impacted by a swinging pendulum. The biofidelity of the ATDs has been extensively studied for use in vehicular crash tests.^{45–48} Kendall et al.⁴⁹ compared the response of the Hybrid-III head to the Hodgson-WSU head form typically used in NOCSAE testing and observed significant differences in peak linear and peak angular accelerations. In their study the nine-accelerometer array was used for both linear and angular measures. The Hybrid-III was instrumented internally, whereas the Hodgson-WSU head form was instrumented externally as it is not typically set up for angular acceleration measure.

Methodology

A test apparatus was fabricated for the purpose of assessing fall protective headgear with guidance from ASTM F2349⁵⁰ the standard for headgear used in soccer. The apparatus is based upon a Hybrid-III head/neck assembly provided by Humanetics™, Plymouth, MI. The Hybrid-III was selected for the purpose that it is readily instrumented with a nine-accelerometer array and its response has been extensively quantified. The drop mechanism consists of a twin wire fall system equipped with a drop arm that includes a 50th percentile male Hybrid-III head/neck assembly. The twin wire drop tower, shown in Fig. 1, was designed and constructed at the University of Maine and has a 5.5 m maximum drop height. It was originally built for an ASTM F1446 type test¹ but has been retrofit with the current head/neck apparatus. The coordinate system used for the data acquisition is shown in Fig. 2(b) and designated according to SAE J1733.⁵¹ The system incorporates a string potentiometer to measure vertical position as the test flyarm is lifted to its desired height. An automated release mechanism is software activated through a digital relay. Both a standard (70-80 Shore A durometer) and lower stiffness (35 Shore A durometer) neck were integrated into the system. The 35 durometer neck was procured from Humanetics and its basic design layout was essentially the same as the standard neck with lower rubber stiffness. The 35

durometer design, originally developed for side impact testing, was selected to provide a significant reduction in neck stiffness where one could reasonably assess the influence of the neck on the response of the test apparatus. With respect to automotive studies, Herbst et al.⁴⁷ reported on the biofidelity of the Hybrid-III neck and concluded that it was in general stiffer than observed in cadaver studies and does not provide meaningful data in situations where load comes in multiple directions such as in vehicle rollovers. Accordingly, a study to assess the influence of the neck stiffness on the apparatus response was included.

The assembly was developed to impart and measure both linear and angular acceleration components of the headform simultaneously during impact. The fly arm, shown in Fig. 1, consists of aluminum tubing that articulates the head. Special adapters were fabricated that can be placed between the neck and flyarm support so that the head can be configured to strike in a frontal (Fig. 1(c)), rear (Fig. 1(d)), or side (Fig. 1(e)) impact orientations. Impact locations are shown in Fig. 2 that also includes a $\pm 7^\circ$ rotation around an axis parallel to Y going through the neck mount. These rotated positions were studied as they can be easily set up using the adjustment mechanism in the standard frontal mount of the ATD. The total mass of the arrangement outfitted with the head/neck assembly is 8.2 kg including the flyarm mass.

Anvil selection is an important consideration and will greatly influence the impact dynamics. A stiffer anvil will typically result in a shorter impulse period. For the purpose of this fall study an anvil was constructed to imitate fall onto a concrete floor, simulating the floor of a building such as a hospital. Accordingly, an anvil was formed from a 102 mm $\times$ 204 mm $\times$ 406 mm long solid concrete masonry unit with a VCT bonded to the top using standard construction practices. Additional anvils include a flat, 25-mm-thick steel plate and a Modular Elastomeric Programmer (MEP) anvil consisting of a 60 durometer, 25-mm-thick neoprene rubber mounted over a 25-mm steel plate. The flat steel and MEP anvils are often used in impact attenuation studies^{37,38}; therefore the impact response onto the concrete/VCT anvil was investigated in comparison to these more standard conditions. This selection of anvils results in a wide range of anvil stiffness characteristics to adequately bound the influence of the anvil on the system response.

The response of the test system was measured using the Hybrid-III ATD head/neck assembly in several configurations. Some of the parameters are: (1) the drop height; (2) the angular position and orientation of the head; (3) the neck stiffness; and (4) the anvil material. For each study a minimum of five drop repetitions per set were averaged. The associated linear and angular acceleration for these tests are captured from the acceleration array, and the HIC_{15} , RIC_{36} , and $PHRIC_{36}$ are calculated accordingly for each data set. The testing was performed by indexing through the drop heights for each data set and then repositioning the apparatus for subsequent data sets.

Data Acquisition—A specially written computer program is used to control the drop test system. Data are recorded through a measurement computing simultaneous sampling 16-bit data acquisition system operating at 20 kHz. A velocity gate used to record impact velocity is comprised of a pair of photodiodes spaced at 38 mm apart. Impact signals from four triaxial accelerometers were arranged in an array so that angular acceleration is calculated

directly from the filtered linear acceleration signals using a method as described by Padgaonkar.⁵² The accelerometers used were PCB model 356B21 triaxial accelerometer with a peak acceleration magnitude of 500 g. The HIC_{15} , RIC_{36} , and $PRHIC_{36}$ are computed within the data acquisition program using the magnitude of the center of gravity (CG) acceleration signal and angular acceleration-time histories, where appropriate. Impulse period of the primary impact is also estimated from the CG linear acceleration magnitude.

Filter Selection—Filtering of data needs to be carried out with care, especially if peak values are included in the reduced data set and as there is little legacy data for the current setup, a study of the filtering influence on the data processing was performed. An anti-aliasing four-pole Butterworth filter with a cutoff frequency of 8 kHz is employed prior to frequency domain computations of noise estimates. Instrumentation noise in the system was observed to be predominately white noise with a maximum root mean square value of 0.06 g across the input channels, and a maximum and minimum value of 0.34 and -0.37 g, respectively, across the input channels. Figure 3(a) shows a typical raw time history of the X , Y , and Z acceleration at the CG for a frontal drop test of approximately 300 g peak acceleration. It is observed that the primary impulse is predominantly made of the Z component acceleration; the nonzero X -direction component is owing to the lack of perfect symmetry about the Y - Z plane. Figure 3(b) presents the Fourier spectrum given in dB of the three linear acceleration components. A fourth-order low pass Butterworth filter was implemented with various cutoff frequencies ranging from 250 Hz to 4 kHz to study the effect of the filtering on the resultant peak amplitude. Filtering below approximately 1000 Hz was observed to cause a reduction in peak linear acceleration amplitude. A similar trend was observed in the filtering of the angular acceleration, which has a relatively flat plateau from 1250 to 2750 Hz, after a steady rise in peak value is observed. The HIC_{15} computation is not as sensitive to filter frequency, as expected and plateaus at 750 Hz. If filter cutoff is too low (below 800 Hz in this case) significant error in the impact period estimate is also prevalent. Accordingly, acceleration signals are filtered using a CFC class 1000 filter per SAE J211 which is a fourth-order Butterworth low pass filter with a cutoff frequency of 1650 Hz.³⁴

Results

The first study presented is the influence of the angular position of the head in a frontal drop onto the VCT/concrete anvil from -7° , 0° , $+7^\circ$ using the HIII neck mount. In the -7° position it was observed that the impact occurs on the forehead and nose simultaneously. Figure 4 shows the results of the maximum linear and angular acceleration for each case versus the impact energy, which is computed as the kinetic energy at impact based upon the mass and impact velocity. Drops were terminated when the maximum linear acceleration exceeded 400 g. Error bars in the figure indicate ± 1 standard deviation. An analysis of variance (ANOVA) indicates a similarity in the results between the various orientations for both the linear and angular acceleration with P -values of 0.75 and 0.49 for linear acceleration and 0.59 and 0.19 for angular acceleration comparing the $+7$ and -7 orientation to the 0 baseline for linear acceleration, respectively. The larger differences in the -7° orientation are attributed to contact of the nose that occurred with this orientation.

The effect of neck durometer was assessed by comparing the maximum linear acceleration versus the maximum angular acceleration as illustrated in Fig. 5. The dark markers in the figure represent the response with the lower durometer neck as opposed to the non-filled markers representing the standard neck. The neck stiffness was quantified in testing by the manufacturer, Humanetics, and the 35 Shore A durometer neck is approximately 25 and 27% the stiffness of the standard neck (70-80 Shore A durometer) in extension and flexion, respectively. Testing was performed in the front, rear, and side configurations. The front drop orientation resulted in similar response when the neck durometers are compared to each other resulting in P -values of 0.99 and 0.25 for linear and angular acceleration, respectively. A similar observation is made in the case of a side drop with a P -value of 0.93 for linear acceleration and 0.18 for angular acceleration. In the rear orientation, a different observation was made as the responses are significantly different with $P < 0.05$ for both linear and angular acceleration cases.

The peak linear acceleration magnitude is shown in Fig. 6, which compares the front, rear, and side impacts onto the VCT/concrete anvil. A sample set of time history traces is presented in Figs. 7 and 8 for the linear and angular acceleration, respectively. Filtered output acceleration components are given for the three drop orientations from a drop height of 50 cm, which delivers approximately 39 J impact energy. Using a frontal drop as the baseline, similarity is shown in the linear acceleration peak response with P -values of 0.85 and 0.40 comparing frontal to rear and frontal to side, respectively. Front compared with rear angular acceleration results are not significantly different statistically as indicated by a P -value of 0.79. On the other hand, angular acceleration results of front compared with side response were found to be significantly different with $P < 0.05$.

Figure 9 gives the RIC_{36} and $PRHIC_{36}$ versus peak angular acceleration for a front and a side drop. The rear drop is similar to the front drop case and not included in the figure for clarity. Computation of CP as given in Eq. 5 was also performed, and a >90% probability of concussion was indicated for all but the 20-cm drop height for a frontal drop at 57% and the 20- and 25-cm drop height for the rear case which are 53 and 86%, respectively. Influence of anvil material is compared using the HIC_{15} values versus the peak linear acceleration as presented in Fig. 10. Drops onto a VCT/concrete, steel, and MEP anvil were compared in a normal frontal impact. Figure 10 also includes the average drop test data in the study performed by Lloyd.³⁵

Discussion

The results presented demonstrate that the test setup described can be used to impart a relatively well-controlled combination of linear and angular acceleration using the Hybrid-III head and neck assembly. Desired impact velocity and input energy are controlled by the drop height. Recorded impact velocity deviated from theoretical values calculated using the drop height by a maximum of 0.83% and a minimum of -2.5% with a root mean square deviation of 0.85%.

Head orientation

When drop orientations are compared at a given input energy level (Fig. 6), the linear accelerations were essentially the same with the average percent difference of -1 and 0.02% comparing the side and rear impacts to the front impact, respectively. Maximum angular acceleration caused by the front and rear impacts were comparable with the rear impact being 17% less on the average than the frontal impact. The angular acceleration recorded during the side drop was significantly greater owing to the side loading of the neck and was on the average 124% greater than on a frontal impact. The lateral loading of the neck that greatly increases the angular motion compared with that when the neck responds in flexion or extension. Differences in angular response between front and rear orientations are mainly attributed to the bias in neck stiffness between flexion and extension. Although this response is not thought to be biofidelic, it may be a useful means to apply a wider variety of test conditions during protective gear testing. Overall, the test results show that the linear acceleration minimally influenced by the orientation and direction of the impact. The neck stiffness (lateral versus flexion versus extension) in linear acceleration had little influence on the response. On the other hand, angular acceleration is highly sensitive to impact orientation and direction. It is at a peak for the stiffest neck axis that is in the lateral direction.

In a frontal drop, the X component of linear acceleration is the primary response signal and on the average represents approximately 90.3% of the peak acceleration magnitude compared with the Z and Y components, which are 9.6 and 0.1% of the peak respectively. The low Y component indicates a relatively symmetric impact response along the X - Z plane. This average was computed over the total data set consisting of 10 drop heights with five repeats per height. In the rear drop case the relative percentage of the component response are 91.4 , 0.2 , and 8.4% for the X , Y , and Z directions, respectively. The Y linear acceleration is the primary component for the side drop case and is on the average 84.2% of the peak, while the Z and X direction signals are 14.0 and 1.8% , respectively. The angular acceleration in front and back drop orientations has the rotation around the Y axis as the primary signal resulting in contribution to the peak of 90.4 and 78.4% , respectively. The X and Y components for the front drop are 6.5 and 3.1% , respectively, and for the rear drop they are 10.9 and 10.7% , respectively. The X direction rotation is the primary for the side drop case at 92.9% of the total on the average, when the Y and Z components are 4.0 and 3.1 , respectively.

Statistical data for linear and angular acceleration versus drop height for the front rear and side orientations include mean peak values and coefficient of variation for each of the test conditions and are given in Tables 3 and 4, respectively. The front and back drop tests show a relatively low coefficient of variation ranging from 0.4 to 2.9% for linear acceleration and from 0.8 to 5.1% for angular acceleration. The angular acceleration in the side drop condition shows a substantially higher variation likely due to the sensitivity to load eccentricity in side position with coefficient of variation (COV) ranging from 4.3 to 16.3% , where the linear acceleration COV ranged from 1.4 to 6.7% .

The response of the apparatus in a frontal drop orientation is only slightly sensitive to the head angle in the range tested. In the -7° orientation, the linear acceleration was greater

than the 0° baseline with percent difference ranging from 1.3 to 7.7% depending on drop height. In this orientation the nose impacted the anvil during the primary impulse. The angular acceleration was also, in general greater, with differences ranging from -1.1 to 14.5%. The +7° angle reduced peak angular acceleration and caused, in general, a slight reduction in the linear acceleration ranging from 0.7 to -5.6% with the exception of the 53J energy level that resulted in an increase of 0.3%. The angular acceleration also showed a reduction ranging from -0.3 to -13.2% again with the exception of the 53J input which showed a 31% increase. In summary, no definitive advantage is observed in switching the head angle in the ±7° range and the test apparatus should be set up to avoid the -7° angle where a nose strike occurs during the initial impulse.

Neck durometer

The relationship between linear and angular acceleration for cases comparing different neck durometer is modeled using a second-order curve passing through the origin that relates the linear acceleration, $A_{\max}$, in g's to the angular acceleration, $a_{\max}$, in rad/s² where:

$$a_{\max} = b_1 A_{\max}^2 + b_0 A_{\max} \quad (6)$$

Table 5 presents a summary of the coefficients b_1 and b_2 determined using the test data. The coefficient of determination, R^2 , with a minimum of 0.991, indicates a good fit of the data to this model. When response with different neck durometer is compared, the stiffer standard neck results in a significantly higher relative angular to linear acceleration when subject to a side drop. Conversely, on a front drop the lower durometer neck shows a slightly higher ratio and the rear drop case alternates. A change in neck durometer resulted in statistically the same peak linear acceleration for a frontal drop with percent difference ranging from -4.3 to 8.8% depending on drop height. The angular acceleration for the front drop case is on the average 7.3% greater for the lower durometer neck with a range from 18.0 to -1.4%. For the side drop, the linear acceleration recorded was independent of neck durometer with an average difference of 1.3% and a range from 10.5 to -3.0% for the various drop heights. The angular acceleration of the standard neck shows a response that is increasingly greater than that produced by the low durometer neck with impact energy. At 20 J, the response is nearly identical at -1.9% difference, whereas at 33 J the response of the lower durometer neck is -16.9% less than the standard neck. Overall, the neck durometer was shown to play a significant role in the response of the test apparatus especially for the side drop case where the stiffer neck attracted more flexural and torsional load causing higher angular acceleration. The significant difference in response during a rear drop is attributed to the neck geometry of the ATD where the rubber disks separate and a gap opens with the headform hanging in the rear orientation. In the rear orientation, the response of the lower durometer neck is more flexible and caused the extremity of the flyarm to contact the base support during the primary impulse.

Anvil materials

Variation of anvil material can be used to primarily control the relation of peak linear acceleration to HIC_{15} . When different anvil materials are compared, as summarized in Table 6, a high correlation with $R^2 = 0.98\text{--}0.99$ was observed between the maximum linear acceleration and HIC_{15} using a second-order curve passing through the origin with coefficients as given in Eq. 7.

$$HIC_{15} = c_1 A_{\max}^2 + c_0 A_{\max} \quad (7)$$

Observed in Fig. 10 is that the relative ratio of HIC_{15} to maximum linear acceleration can be increased by using an anvil with a softer impact surface. Results of ATD testing by Lloyd³⁵ lie between the cases of a very rigid anvil, such as steel or VCT/concrete, and the MEP anvil. If desired, matching the ratio observed in the ATD testing would require an anvil surface that has rigidity between the two cases and could be readily made from a stiffer and/or thinner rubber than the 25 mm, 60 durometer rubber used in the MEP anvil.

Comparisons to other methods

Impact attenuation tests of helmets used for motorcycles, bicycling, and other sports equipment currently use a twin wire or monorail drop test device as described in ASTM F1446.¹ These devices are designed to impart a single component of translational acceleration onto the test piece and typical impacts are onto a relatively rigid anvil surface such as a flat steel plate, hemisphere, or curbstone. Therefore, comparison of the angular acceleration response to these devices is not applicable as they are not designed to measure angular motion. The drop assembly fly arm for traditional bicycle helmet testing, for example, has a mass specified as 5.0 ± 0.1 kg excluding the helmet. This is compared with the higher mass of 8.2 kg in the current apparatus primarily owing to the inclusion of entire Hybrid-III head and neck assembly. Comparing to Snell standards,¹⁵ impact energy levels of 110 J are typical for certification testing of a bicycle helmet against a flat anvil, and this is the case of having a stiff impactor and impact surface as compared with the current setup where the impactor is relatively flexible. To that end, an unprotected drop calibration typically uses an MEP pad which is a 25-mm-thick 60 Shore A elastomer that is impacted at a velocity of 5.44 m/s (74 J) with the recorded acceleration to be in the range of 380 – 425 g. The current setup impacted onto a similar MEP pad results in a peak acceleration of approximately 195 g at 74 J input that is 51% of the minimum response in the Snell calibration. The difference is mostly attributed to a more flexible impactor in the current setup. In addition, the current setup was calibrated up to approximately 52 J without the use of an MEP pad with resulting peak acceleration of approximately 412 g. Testing under the NOCSAE standard³⁸ uses a similar apparatus with a NOCSAE headform having anthropomorphic features and calibrates with a 7.6-cm- and 1.3-cm-thick MEP pad. Expected response is 232 g for the thick MEP pad when the impactor dropped onto that at an impact velocity of 5.44 m/s, and is 385 g for the thinner MEP pad when the impactor dropped at an impact velocity of 3.9 m/s.

Unprotected acceleration of the head with impact onto a hard surface due to falls is highly variable and more work is currently required to categorize the bounds. Studies have been conducted numerically by O'Riordain et al.³¹ and Doorly et al.,³² using cadavers by Hardy et al.,³³ and using ATD's by Walsh et al.³⁶ Peak acceleration in an unprotected fall case is shown to range up to 1340 g by Lloyd³⁵ using ATDs for an unprotected backward fall. Cases of cadaver impact by Hardy³³ show more modest peak acceleration above 400 g. Angular acceleration indicates a wide range of variation with values up to 50,000 rad/s² in the numerical studies reported by Doorly.³² Figure 11 presents the linear versus angular acceleration response of the Hybrid-III setup in the front and side drop orientations compared with the results of these other studies. The solid lines in the figure represent the Hybrid-III setup and the dashed line is the average results of the other studies. The studies presented represent a wide variety of input conditions that included variation in methods, impact velocity, orientation at impact, and impact surface, and are meant to envelope the realm of possible fall conditions. The numerical studies by Doorly³² and O'Riordain³¹ were reported as susceptible to input conditions and showed little correlation to a second-order polynomial model with R^2 of 0.05 and 0.43, respectively. The cadaver study by Hardy³³ was the most closely correlated to a second-order curve and resulted in $R^2 = 0.84$. In a loose sense, the frontal and side drop of the current Hybrid-III setup can be used to represent the bounds of the physical studies presented and indicate that the setup can be used to implement an appropriate level of relative linear to angular acceleration by varying the test conditions. More work is required to develop appropriate input levels for differing fall conditions.

Conclusions

The current study quantifies, on a preliminary basis, the performance of a test apparatus that is intended for the evaluation of protective headgear for persons at risk for falls. Recent studies have indicated that reduction in angular acceleration may be of equal or more importance than linear acceleration in reducing incidences and severity of brain injury, especially the diffuse type. Accordingly, the system presented in this study is based on a twin wire drop tower and uses a Hybrid-III ATD head/neck assembly that was outfitted with an array of triaxial accelerometers to quantify both linear and angular acceleration.

It was demonstrated herein that the device can be set up to reproduce a wide range of input conditions. Coefficient of variation in resultant linear and angular acceleration was measured using the device at various drop heights and impact orientations for five trials under each condition. Variation in linear acceleration ranged from 0.6 to 6.73% depending on drop height and impact orientation. Angular acceleration in the front and rear impact orientations shows a similar variation with a minimum of 0.8% to a maximum of 6.8%. Angular acceleration measured in the side impact orientation shows a higher variation with a worst case of 16.3%. This is likely attributed to a method of reaction of the rotational components of force in the twin wire apparatus and the characteristics of the HIII neck. Unprotected impact conditions such as maximum linear acceleration and angular acceleration are shown to be related to the input energy (drop height) along with head orientation and anvil material, which can be used to control the input parameters. The ratio of linear to angular acceleration will vary depending on the type and conditions of the fall and can be controlled, to some

degree, by impact orientation, with side orientations resulting in a higher ratio of angular to linear acceleration than front or rear drop cases. Other drop orientations not addressed in the current study are also possible. The relationship of the HIC_{15} to the peak linear acceleration shows a repeatable trend that is predicted by a second-order expression and the ratio can be controlled by adaptation of the impact anvil surface material. Softer impact surfaces result in higher values of HIC_{15} relative to maximum linear acceleration. The average conditions achieved in an ATD fall study by Lloyd³⁵ can potentially be simulated using an anvil with stiffness in between that of steel and the MEP.

Methods used to assess human fall response must be interpreted with caution and more work is currently required to quantify human falls so that proper input conditions can be developed leading to innovative and highly functional designs. Injury measures are currently in need for risk assessment of injury due to falls, especially in the elderly population. With proper input conditions, the described test apparatus can be used to assess both linear and angular performance of fall protection headgear for a given set of parameters. In so doing the effectiveness of a headgear design can be evaluated on a preliminary basis with respect to its translational and rotational response. In summary, the Hybrid-III test apparatus is potentially suitable for the evaluation of linear and angular accelerations associated with biomechanical impact events, such as falls.

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Figure 1. Photograph of the University of Maine drop test setup. (a) Drop tower with assembly; (b) coordinate system, (c) front drop; (d) rear drop; and (e) side drop.

Figure 2.
Impact locations.

Figure 3.

Linear acceleration components for a frontal drop from 50 cm height. (a) Raw data time history for CG acceleration components. (b) Frequency composition of raw data time history.

Figure 4.
Influence of neck angle on peak linear and angular acceleration in a frontal drop.

Figure 5.

Influence of neck durometer on peak angular acceleration versus linear acceleration.

Figure 6.
Impact orientation on peak linear and angular acceleration.

Figure 7. Linear acceleration components for a frontal drop from 50 cm height using CFC class 1000 filter. (a) Front, (b) rear, and (c) side impacts.

Figure 8. Angular acceleration components for a frontal drop from 50 cm height using CFC class 1000 filter. (a) Front, (b) rear, and (c) side impacts.

Figure 9.
RIC and PRHC versus angular acceleration for front and side impact orientations.

Figure 10.
Influence of anvil material comparing maximum linear acceleration and HIC for a frontal drop.

Figure 11.
Relationship of linear to angular acceleration in various studies.

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Table 1
Some published threshold values for various brain injury predictors

References	Injury risk, <i>P</i> percentile	Max Lin. Acc., g	Max Ang. Acc., rad/s ²	SI	HIC ₁₅	HIP, kW	RIC ₃₆	PHRIC ₃₆
Fenner ²¹	50	80	6200	300	230			
Newman ¹⁵	50	77.6	6322	291	240	12.8		
Zhang ²²	50	82	5900		240			
Kimpara ¹⁹	50				700		1.03×10^7	8.7×10^5
King ⁶	50	79	5757		235			
King ⁶	25	57	4384		136			
Funk ²³	10	165	9000		400			
Funk ^{*24}	10	199			689			
Funk ^{*24}	50	264			1030			
Rowson ^{*25}	10	149						
Rowson ^{*25}	50	192						

* Prediction for collegiate football MTBI.

Table 2

Summary of fall study using a Hybrid-III ATD

Condition	Linear acceleration (g)		HIC ₁₅		Estimated energy (J)
	Mean	Maximum	Mean	Maximum	
Standing fall					
HIII—5% female	202	(243)	591	(686)	59
HIII—50% male	302	(518)	1487	(3756)	87
HIII—95% male	1153	(1340)	n.r.		113
Crumple fall					
HII—5% female	158	(245)	447	(1066)	30
HII—50% male	226	(409)	705	(2235)	42
HII—95% male	591	(618)	n.r.		48
Protected crumple fall					
HII—5% female	80	(184)	89	(269)	30
HII—50% male	207	(358)	661	(1631)	42
HII—95% male	240	(415)	1686	(5660)	48

n.r., not reported.

Table 3**Mean value of peak linear and angular acceleration**

Drop height, cm	Average velocity, m/s	Nominal input energy, J	Linear acceleration, g			Angular acceleration, rad/s ²		
			Front	Rear	Side	Front	Rear	Side
20	1.94	16.1	134.8	136.0	131.6	6,203	5,918	13,701
25	2.20	20.1	165.5	162.6	163.7	7,844	6,917	16,230
30	2.42	24.1	198.1	189.1	202.7	9,417	7,647	20,881
35	2.62	28.2	231.6	219.9	232.0	10,790	8,645	27,043
40	2.79	32.2	259.0	254.6	261.9	12,164	9,618	30,575
45	2.96	36.2	289.4	293.1	291.0	14,128	11,159	32,024
50	3.12	40.2	324.2	328.5	308.9	15,306	12,525	34,215
55	3.29	44.2	361.8	362.7	346.9	16,527	13,818	35,206
60	3.44	48.3	389.4	395.9	388.8	18,612	14,745	37,636
65	3.58	52.3	410.4	422.2	404.9	18,332	16,135	38,571

Table 4**Variation of peak linear and angular acceleration**

Drop height, cm	Linear acceleration, COV in %			Angular acceleration, COV in %		
	Front	Rear	Side	Front	Rear	Side
20	1.8	1.8	2.8	2.7	2.7	3.9
25	1.6	0.4	2.4	3.8	1.2	4.3
30	1.3	0.7	3.4	5.1	4.2	16.3
35	1.9	0.8	4.6	3.9	3.3	7.4
40	2.9	0.5	3.5	3.8	2.2	14.2
45	1.6	1.7	6.4	0.8	2.5	9.6
50	2.3	0.6	5.9	1.9	2.2	12.4
55	2.1	0.9	3.7	1.3	2.8	6.9
60	1.6	1.1	6.7	4.1	1.4	11.2
65	1.6	1.5	1.4	6.8	2.3	9.9
Average	1.9	1.0	4.1	3.4	2.5	9.6

Table 5
Coefficients relating angular to linear acceleration for different neck durometer and drop orientation

Case	b_1	b_0	R^2
Front-standard	-0.0072	49.04	0.993
Front-35 durometer	-0.0164	54.63	0.991
Rear-standard	-0.0115	42.21	0.994
Rear-35 durometer	-0.0437	54.23	0.977
Side-standard	-0.0682	125.75	0.961
Side-35 durometer	-0.0636	113.8	0.983

Table 6
Coefficients relating HIC_{15} to linear acceleration in several frontal drop cases

Case	c_1	c_0	R^2
Steel anvil	0.0105	0.0331	0.999
VCT/conc anvil	0.0109	-0.1222	0.999
MEP anvil	0.0256	-0.6585	0.998
Pad under skin	0.0122	0.1205	0.999
ATD tests	0.0172	-0.3924	0.985