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Finite Element Modelling of Equestrian Helmet Impacts Exposes the Need to Address Rotational Kinematics in Future Helmet Designs

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Jockey head injuries, especially concussions, are common in horse racing. Current helmets do help to reduce the severity and incidences of head injury, but the high concussion incidence rates suggest that there may be scope to improve the performance of equestrian helmets. Finite element simulations in ABAQUS/Explicit were used to model a realistic helmet model during standard helmeted rigid headform impacts and helmeted head model (UCDBTM) impacts.

Current helmet standards for impact determine helmet performance based solely on linear acceleration. Brain injury related values (stress and strain) from the UCDBTM showed that a performance improvement based on linear acceleration does not imply the same improvement in head injury related brain tissue loads. It is recommended that angular kinematics be considered in future equestrian helmet standards, as angular acceleration was seen to correlate with stress and strain in the brain.

Keywords: Head impact; traumatic brain injury (TBI); concussion; sports helmets; horse racing.

1. Introduction

The use of helmets by equestrian jockeys is widespread, and it is mandatory equipment for a jockey in professional competitive racing. Using a helmet considerably reduces the risk of sustaining serious head injury (Harrison et al. 1996; Turner et al. 2002). Equestrian jockeys, especially jump racing jockeys, fall very frequently and have a high risk of suffering head injury. Data from Britain, France and Ireland (Forero Rueda et al. 2010) shows that of all the injuries in both jump and flat racing populations of amateur and professional jockeys, 15% are concussive head injuries, more than half of which involve loss of consciousness.

Research into equestrian helmets considerably lags behind the technologies which are commonly used for other types of helmets. Nevertheless, due to the high incidence and risk of head injury to jockeys, horse riding helmets merit dedicated research. Research techniques such as FE modelling, which have been employed on other types of helmets, could be applied to analyse existing equestrian helmet standards and helmets in order to identify improvements to the next generation of head protective measures for horse riders.

Very few studies have considered evaluating helmet performance based on injury related parameters other than linear acceleration, such as rotational acceleration and brain tissue loads (Aare et al. 2003; Deck and Willinger 2006). These studies were done on motorcycle helmets. Particular attention has been paid to the energy absorbing foam liner and the shell for helmet optimization. As computer processing power and Finite Element Modelling (FEM) techniques improve with time, virtual testing is increasingly being used to assess motorcycle helmet performance.

There have been many technical studies regarding bicycle helmets (Mills 1990; Andersson et al. 1993; Mills and Gilchrist 2003; Depreitere et al. 2004; Van Lierde et al. 2005; Mills and Gilchrist 2006a,b; Mihora et al. 2007). These studies focussed only on ways to reduce linear acceleration and on the performance of the foam liner, in terms of its energy absorption by measurement of force-displacement. Recently, there has been some consideration given to oblique impacts (Verschueren 2009), but acceleration has been the only variable which has been used to assess head injury risk for bicycle helmet evaluation.

Recently, in the studies by Forero Rueda et al. (2009) and Cui et al. (2009a), equestrian helmets have begun to be analysed using computational methods. The helmet performance was analysed not only by headform linear acceleration as in the current equestrian helmet standards (NSAI 1997, 2005), but also in terms of the internal behaviour of the helmet liner material. This gave a more complete picture of how a helmet absorbs impact energy, and how the liner of a helmet can be improved to reduce linear acceleration.

1.1 Finite Element head modelling

Equestrian helmet designs are still only evaluated in terms of their ability to reduce linear acceleration. However, a criterion based on bulk mass kinematics cannot quantify the stresses or strains within the brain, which are the actual sources of brain injury. The use of FEM techniques allows the behaviour of brain tissue loads to be observed during a helmeted impact instead of only kinematic outputs such as head linear acceleration, as if the human head were only a rigid body, which can easily be computed using multibody dynamics simulations (O'Riordain et al. 2003; Doorly and Gilchrist 2006, 2009; Forero Rueda and Gilchrist 2009).

Tolerance levels of neural tissue are determined by the mechanical load on the tissue. These loads are determined by stresses and strains. Experimental research has been done to determine the tolerance of brain tissue (Margulies and Thibault 1992; Galbraith et al. 1993; Miller et al. 1998; Anderson et al. 1999; Bain and Meaney 2000), and to correlate neural tissue loads to actual injury types through FE modelling (Shreiber et al. 1997; Miller et al. 1998; Anderson et al. 1999; Willinger and Baumgartner 2003; Zhang et al. 2004; Kleiven 2007; Doorly, 2007). It is important to emphasize that FE-based thresholds are limited by any model's choice of constitutive parameters for the strain and stress based measures and the modelling strategy of the skull-brain interface for all measures. With careful selection of material properties, current FE brain models are able to output biofidelic orders of magnitude for brain loads (Kleiven and Hardy 2002; Horgan and Gilchrist 2003, 2004). In this manner, it is possible to establish comparative trends between different external load cases.

1.2 Brain tissue loads

1.2.1 Von Mises stress

Willinger and Baumgartner (2003) reconstructed real-life accidents involving motorcyclists, pedestrians and football players and found that Von Mises stress levels of 18 kPa and 38 kPa correspond to a 50% risk for moderate and severe neurological lesions, respectively. Their model however, was validated against limited experimental data, which limits the reliability of these predicted stress levels.

Von Mises stress levels were also analysed from experiments done on miniature pig heads and FE modelling (Miller et al. 1998). Depending on the particular modelling approach, it was found that peak Von Mises stresses of 9.1-44.5 kPa gave a 50% probability of axonal injury and 7-8.6 kPa for a 50% chance of contusion. Anderson et al. (1999) suggested levels of 8-16 kPa for diffuse axonal injury based on an experimental and FE study with sheep heads. A FE model of cerebral contusions in the rat was developed and compared to experimental injury maps demonstrating blood-brain barrier breakdown (Shreiber et al. 1997). The values for Von Mises stress were in the range of 6.1 to 10.8 kPa, but no statistical significance was found relating these quantities to injury.

In a more recent FE study by Kleiven (2007), the region of the corpus callosum showed the highest correlation with injury, with a 50% probability of concussion to be related to a Von Mises stress of 8.4 kPa. These levels are within the values observed by Shreiber et al. (1997).

1.2.2 Longitudinal strain

Margulies and Thibault (1992) proposed human injury tolerance curves for diffuse axonal injury (DAI) and milder forms of axonal injury such as cerebral concussion. They suggested critical strain for moderate to severe DAI ranges from 5% to 10%.

Galbraith et al. (1993) carried out experiments to determine the levels of elongation required to damage the squid giant axon. They showed that a stretch ratio of 1.12 resulted in reversible injury, that axons subjected to elongation above 20% never fully recovered, and that structural failure resulted when axons were stretched by more than 25%.

Bain and Meaney (2000) carried out stretching experiments on the right optic nerve of an adult male guinea pig. The liberal threshold, intended to minimize the detection of false positives, was a strain of 0.34, and a conservative strain intended to minimize the detection of false negatives, was 0.14. The optimal threshold strain criterion that balanced the specificity and sensitivity measures was 0.21. Similar comparisons for the electrophysiological impairment produced liberal, conservative and optimal strain thresholds of 0.28, 0.13 and 0.18 respectively. Zhang et al. (2004) also investigated mild TBI with respect to strain, and found brain strain levels of 25%, 50% and 80% probability of mild TBI to be about 0.14, 0.19 and 0.24 respectively. Zhang et al. (2004) also pointed out that these limits would vary depending on the material properties of the model used. Kleiven (2007) reported similar levels of maximum principal strain to those previously mentioned values for DAI and contusions. He reported a 50% probability of concussion is found for a level of 0.21 in the corpus callosum and 0.26 in the grey matter.

1.3 What this study offers

In this study, impact simulations using a FE model of the human brain, the University College Dublin Brain Trauma Model (UCDBTM) (Horgan and Gilchrist 2003, 2004), in conjunction with various equestrian helmet models, were performed and compared with impact simulations done with a standard headform. This was done to determine whether and how a rigid headform could reflect brain tissue loads. As seen before, previous study has shown that there are relationships between brain tissue loads (Von Mises stress and longitudinal strain) and brain injuries. The output parameters of the UCDBTM are compared to the rigid headform outputs to determine how a rigid headform could be used to reflect actual brain tissue loads within the human brain.

2. Method

2.1 Head Models

2.1.1 Rigid headform model

Two different types of head models will be used in this study. The first model is the rigid headform model, which was used to record linear and angular acceleration. The headform (size designation 575, mass 4.7 kg) solid geometry was generated from spherical coordinate points specified in the European headform standard EN 960:2006 (NSAI 2006). The size of the headform had a good fit with the interior of the helmet liner. In the present study, a rigid material definition was used by establishing a rigid body constraint between the headform mesh and a reference point at the centre of mass of the headform. This was considered to be a good approximation to the actual laboratory test as the headform is considerably more rigid than any other helmet component; therefore energy absorbed by the headform can be ignored. Additionally, acceleration measurements are more convenient as the centre of mass acceleration can be retrieved easily from the centre of mass reference point. Computational time is greatly reduced, and acceleration measurements can be done at exactly the centre of mass. A material section was assigned to the headform in order to assign the adequate density to accurately represent the mass and inertial properties of the headform.

2.1.2 UCDBTM

The UCDBTM (Horgan and Gilchrist 2003, 2004, (Figure 1)) was developed in University College Dublin to simulate real life impact scenarios and relate injury types and severities to various engineering values that have been found to correlate with different types of head injury. The model was compared with cadaver tests, showing good agreement with the results (Horgan and Gilchrist 2004).

FIGURE 1 NEAR HERE

The resulting 3-dimensional finite element model of the skull-brain complex consists of scalp, three-layered skull (cortical and trabecular bone), dura, cerebrospinal fluid (CSF), pia, falx, tentorium, cerebral hemispheres, cerebellum and brain stem. The material properties used in the model were defined by Horgan and Gilchrist (2003, 2004) (Table 1). The interaction between the skull and the brain in the UCDBTM has been approximated in the present simulations by modelling the CSF as an incompressible solid of low stiffness and modelling the interface between the CSF and the skull as a penalty contact surface pair with no separation with a friction coefficient of 0.2.

TABLE 1 NEAR HERE

The brain tissue parameters analysed in this study using the UCDBTM were Von Mises stress and maximum principal strain, which have been verified to cause brain injury.

2.2 Helmet modelling

2.2.1 Foam liner material

The foam liner is modelled in ABAQUS/Explicit (2009) using the crushable foam model with a volumetric hardening rule in conjunction with the linear elastic model

(ABAQUS 2009). The hardening behaviour for the polymeric foam is defined by the constitutive stress-strain relationship. The stress-strain curve for the polymeric foam is a function of foam density. The constitutive model for EPS foams of various densities used in the current study have been tested and determined, with good agreement with laboratory tests done up to 85% strain (Cui et al. 2009b). The foam curves were defined up to 95% strain to account for a larger portion of the densification region, to model the “bottoming out" of the foam.

2.2.2 Shell and other components

The outer helmet shell is modelled as a linear elastic material and the rubber ring at the lower edge of the helmet is modelled as a rubber elastomer with Poisson's ratio approaching 0.5 (almost incompressible). When it is not otherwise mentioned, the shell stiffness used in most of this study was of 7.25 GPa, typical of that of most equestrian racing helmets in the marketplace. The foam block between the shell and foam liner is modelled as a hyperelastic elastomeric compressible foam with material constants specified by experimental test data using the ABAQUS hyperfoam model (ABAQUS 2009). The foam block in the actual helmet is made of low density elastomeric PE foam of 21 kg/m^3 density. The purpose of the foam block is to bond the shell and liner together and also to leave a small gap between the shell and liner to allow the shell deform and absorb some energy before the foam liner crushes in an impact. The shell-liner contact interface was defined as a general contact interaction, all with self (available in ABAQUS/Explicit), with a penalty friction formulation (coefficient of 0.2). This same contact interface was defined throughout the whole model with the exception of the interaction between the UCDBTM skull and brain. The individual parts of the helmet are shown in Figure 2.

FIGURE 2 NEAR HERE

2.2.3 Helmet configurations

Once the material models were defined, it was possible to generate different helmet geometries. The headform and UCDBTM were used with four different helmet configurations (Figure 3), to also compare how the headform and UCDBTM measure relative performance between helmet types. The helmet models used in this study were a baseline helmet with a common equestrian racing helmet design and uniform density liner (Figure 3a, named “Baseline"), a functionally graded foam (FGF) liner helmet model (Figure 3b, named “FGF") and two helmet types (without and with an air gap) with a thicker liner and discrete layer configuration (Figures 3c and 3d, named “Type I" and “Type II" respectively). Material properties for the helmet shells and liners are shown in Table 2. The properties for the foam block and rubber ring were kept constant for all helmets.

TABLE 2 NEAR HERE

FIGURE 3 NEAR HERE

The Baseline helmet (Figure 3a) solid geometry was generated based on curves and CAD models obtained from commonly available equestrian racing helmets. The geometry of the helmet resembles a typical helmet design, but is not an exact copy of any particular commercial helmet. The helmet is a typical size 57 (2 ½), which is one of the most common helmet sizes used by equestrian jump jockeys. The liner density for the baseline helmet is a typical 64 kg/m^3.

The FGF helmet model (Figure 3b) had the same geometry of the baseline helmet model, with the difference that the foam liner was assigned a FGF liner configuration as had previously been done by Cui et al. (2009a). The FGF liner gradients used were the same as in Cui et al. (2009a), which were seen to have the best performance with respect to the baseline helmet (for a given impact speed and position) when evaluated using headform linear acceleration. The gradient foam was simulated by dividing the foam liner into 10 layers and assigning a density to each layer according to the desired gradient. The FGF used in the current simulations has its density increased or decreased through the thickness, *y*, according to a power-law gradient function as

$$\rho(y) = \rho_1 + (\rho_2 - \rho_1)(\frac{y}{d})^n$$

where ρ_1 and ρ_2 are the densities at the exterior surfaces of one foam specimen and *d* is the depth of the foam in the thickness direction.

Based also on the baseline helmet model, the Type I helmet configuration generated (Figure 3c) is that of a helmet with no air gap but with a thicker liner. This helmet model does not have a rubber ring or a foam block. A layered density liner configuration was added to capitalize on the added thickness and improve performance at both low and high impact energies. The helmet liner was divided in three equal thickness layers as had been done by Forero Rueda et al. (2009). The density layer configuration used was 50-50-25 kg/m^3 from the inside of the helmet liner to the outside.

The Type II helmet model (Figure 3d) was made by scaling the thickness of the helmet by an additional 50%, while keeping the dimensions of the foam block and the air gap the same as in the baseline helmet. The liner was divided into three layers of equal thickness as in Forero Rueda et al. (2009), and the layers were assigned a 64-25-25 kg/m^3 density configuration from the inside to the outside. In this case, the shell stiffness was increased to 50 GPa (the stiffness of a typical carbon fibre reinforced polymer) to allow the load to spread more on the large portion of low density foam present in the liner of this helmet.

2.3 Flat anvil impact simulations

To study the impact performance of the helmet design according to the headform and UCDBTM, helmet performance was measured at a range of impact severities and impact positions. Vertical impacts were simulated with three impact speeds (45° Side, 45° Front and Crown as in Figure 4). The helmet impacts were done at three impact speeds by applying them to the helmet-head assembly which was brought to within 1 mm away from the flat anvil. The impact speeds were 4.4, 5.4 and 7.7 m/s as specified by the EN1384 and EN14572 (NSAI 1997, 2005) flat anvil test specifications. These impact speeds correspond to laboratory drop heights of 1, 1.5 and 3 m. In total, 36 simulation results were analysed, from simulations done with four helmet types (Figure 3), three impact positions (Figure 4), and three impact speeds, low, medium and high (4.4, 5.4, 7.7 m/s).

FIGURE 4 NEAR HERE

2.3.1 Analysis of results

The flat anvil impact test results for both the headform and UCDBTM for the baseline helmet, the FGF helmet and helmet Type I were compared between each other. This was done to determine if the headform outputs reflected a similar relative performance between helmet models and the UCDBTM outputs. From the simulation results, presented in Appendix A, twenty-seven pairs of data were obtained for the baseline helmet, helmet Type I and helmet Type II, three impact speeds and three impact positions. The relationship between the headform outputs and the UCDBTM outputs was analysed by fitting a linear regression for each of the sets of 27 points and calculating its linear parameters and the least squares coefficient R^2. The headform outputs were taken as the independent variable and the UCDBTM outputs were taken as the dependent variable.

3. Results

3.1 Comparison between Baseline, FGF and Type I helmet

The Baseline, FGF and Type I helmets are compared in terms of headform linear acceleration and UCDBTM stress and strain outputs. This is to investigate whether helmets that were seen to improve on linear acceleration reduction with respect to the baseline improved in terms of brain tissue loads, which are the actual causes of brain injury.

Table 3 shows the corresponding headform linear acceleration results. Table 4 shows the peak and average Von Mises stress within the cerebrum for three helmet types, three impact positions, and three impact speeds. The average Von Mises stress for the whole cerebrum was calculated at the moment where the peak Von Mises stress was reached. When calculating the average Von Mises stress, the stress for each element is weighted by its volume, by multiplying the stress value on each element by the corresponding element volume, then dividing the sum of all the element stress-volume products by the total brain volume:

$$\frac{\sum_{i=1}^{n} v_i x_i}{\sum_{i=1}^{n} v_i}$$

where x_i is the Von Mises stress value on element i and v_i is the volume of element i.

TABLE 3 NEAR HERE

TABLE 4 NEAR HERE

For the FGF helmet case, it can be seen that the change in peak Von Mises stress and average Von Mises stress varies considerably from case to case. The peak Von Mises stress increases by up to 5.2% while it decreases by up to 18.3% when compared to the values of the baseline helmet. There is an increase of peak Von Mises stress for five of the nine impact cases. This contrasts with the linear acceleration results, where linear acceleration always decreased when compared to the baseline helmet. A similar behaviour is seen for the average Von Mises stress. Five out of nine impact cases exhibited an increase, while linear acceleration always decreased.

For the Type I helmet, it is seen that the peak Von Mises stress decreases for all nine impact cases; this decrease varies from 3.9 to 24.3%. Linear acceleration always decreases for helmet Type I when compared to the baseline helmet, but the decrease tends to be larger than the peak Von Mises stress decrease. The Type I helmet peak Von Mises stress decrease averaged 11.5% while its average linear acceleration decrease was 27.7%.

When looking at the average Von Mises stress, only five out of nine impact cases showed a decrease. This contrasts with the constant decrease in linear acceleration. The observed reductions were considerable (up to 26.6%) but so too were the increases (up to 16.2%).

Table 5 shows the peak and average maximum principal strain results. As mentioned before, longitudinal strain has been seen to be one of the important brain loads which lead to brain injury along with stress. Refer to Table 3 to compare with the corresponding headform accelerations.

TABLE 5 NEAR HERE

The FGF helmet does not reduce peak maximum principal strains in the cerebrum as it does with linear acceleration. Five out of nine cases showed an increase in peak maximum principal strain. For the average maximum principal strain only two cases showed an actual decrease. For both peak and average maximum principal strain results, the reductions present were comparatively smaller than those for the linear acceleration cases.

The Type I helmet manages to reduce peak maximum principal strain for eight out of nine impact cases, but the reductions tend to be lower than those of linear acceleration (range of 2.6-21.2% compared to 23.6-48.5%). For the average maximum principal strain case, only two out of the nine impact cases showed a decrease.

3.2 Relationship between headform and UCDBTM outputs

In this section it is investigated whether there is a relationship between outputs that could be obtained from a laboratory setting using a helmeted headform (linear and angular acceleration) and outputs from the helmeted UCDBTM (stress and strain). The intention of this is to find whether it is possible to use a standard headform with added instrumentation to measure quantities that better reflect the behaviour of the brain material as seen by a head finite element model such as the UCDBTM. In the same fashion, it is also possible to study the potential of using other additional headform outputs to analyse comparative helmet performance which reflect the changes observed in the UCDBTM when applying it to other helmet models.

3.2.1 Linear acceleration

From the regression results, it can be seen that even though UCDBTM outputs tend to increase with increased linear acceleration, they showed low correlations ranging between 0.20 for the average maximum principal strain (Table A1) and 0.52 for peak Von Mises stress. For the average Von Mises stress vs. linear acceleration it is possible to see that there could be two distinct linear relationships with two distinct slopes, one above and one below the fit line of Figure 5. When the source of the points above the fit line (Figure 5) was investigated, it was seen that all the points above the fit line belonged to the 45° Side impact position (Figure 6).

FIGURE 5 NEAR HERE

FIGURE 6 NEAR HERE

Significant correlations for the maximum Von Mises stress and maximum principal strain were found for the 45° Side impact position separated from the other impact positions (Table A1). In some cases, when the 45° Side position was considered separately, the other two positions had a good correlation with linear acceleration without being separated, as in the case of average Von Mises stress or average maximum principal strain (see Table A1).

Further analysis was done in an analogous manner separately for each impact position. In the cases of maximum Von Mises stress and maximum principal strain, considerably different slopes were observed for the different impact positions (Table A1). This was seen especially for the 45° Side impact position, which was usually higher than for the rest of the positions. This indicates the higher sensitivity of the head model to side impacts due to its higher sensitivity to changes in linear acceleration. The better correlation obtained when separating impact positions indicates that there seems to be a considerable influence of the impact position on the correlations between the headform output linear acceleration and the UCDBTM output variables.

3.2.2 Angular acceleration

When the rotational acceleration results from the standard headform are analysed with the UCDBTM variable outputs by keeping all impact positions together, it is seen that angular acceleration has a good correlation with maximum and average Von Mises stress and maximum and average maximum principal strain (Table A2). These results contrast against the linear acceleration case, which shows relatively low correlations with UCDBTM stress and strain outputs. The correlation of angular acceleration with Von Mises stress is shown in Figure 7.

FIGURE 7 NEAR HERE

For most of the UCDBTM variables, separating the impact positions does not significantly improve the correlation with angular acceleration (see Table A2). Rotational acceleration seems to correlate better with the UCDBTM outputs for all combined impact positions better than linear acceleration. An increase or decrease of angular acceleration seems to better reflect an increase or decrease of a good proportion of the studied UCDBTM output variables. Rotational acceleration does this without having to differentiate between impact positions, which means that it inherently accounts for impact position sensitivity. This shows that angular acceleration has a similar sensitivity to impact position as do the UCDBTM output variables.

4. Discussion

A reduction in linear acceleration in the helmet standard tests does not necessarily mean a reduction in injury related UCDBTM outputs. This confirms the findings in this study where it was seen that linear acceleration alone did not have a good correlation with most of the UCDBTM output variables. Helmets which were seen to improve on linear acceleration did not necessarily improve on brain tissue loads.

Linear acceleration was seen to have a low correlation with stress and strain in the brain. When separating the data by impact position, correlations with the UCDBTM outputs improved dramatically. This indicates that linear acceleration sensitivity of the UCDBTM outputs is different for each impact position, which in turn indicates the impact position sensitivity of the UCDBTM.

Angular acceleration could be used as a measure to predict UCDBTM stress and strain values with less need to consider impact directions individually. The results obtained for angular acceleration show a high correlation with stress and strain UCDBTM outputs which suggests that angular acceleration is important when actual injury mechanisms of human brain tissue are considered. Angular acceleration was seen to encompass the sensitivity of the human head to impacts in different directions, because it shows impact direction sensitivity similar to that of many of the UCDBTM output variables. In contrast, linear acceleration seemed to always have a different correlation with UCDBTM outputs depending on impact direction. This indicates that it is more difficult to predict brain tissue loads, and hence injury risk, based on linear acceleration alone.

Previous work (Marjoux et al. 2008, Kleiven 2007, Greenwald et al. 2008) using both experimental and computational methods, have also found similar conclusions which confirm the findings, methods and the capabilities of the head FE model used in this study. Current finite element computing power that is readily available means that high-resolution, two-dimensional models (Gilchrist & O'Donoghue 2000, Gilchrist et al., 2001) of sectional planes through a head could offer useful insights to complement lower-resolution three-dimensional models. The work by Marjoux et al. (2008) showed that finite element modelling of the head can model the mechanical behaviour of the intracranial contents more effectively than just considering the head as a single mass. The studies by Kleiven (2007) and Greenwald et al. (2008) showed that injury criteria which used angular kinematics have the potential to predict brain injury more effectively than only considering linear acceleration. Through the use of different research methods and tools, the consensus that is being reached in terms of the importance of the influence of angular kinematics on brain injury is the first step towards changing future helmet designs and certification procedures.

There have been some attempts to devise methods that could used to introduce angular acceleration in the helmet design process (Willinger and Baumgartner 2003 and Halldin et al. 2001). Willinger and Baumgartner have devised a mechanical model of the brain which shows the need to consider variables beyond linear acceleration to assess head injury. Halldin et al. (2001) have devised a testing method for oblique impacts, using a dummy headform to record both linear and angular acceleration. These studies show how it could be possible to use what is currently known regarding the mechanics of head injury to develop helmet tests capable of gauging actual impact injury. Nevertheless, there is still considerable work to be done to determine the criterion that is capable of predicting brain injury accurately. This requires extensive forensic, experimental, and simulation work to establish accurate thresholds for head impacts, and reach an agreement on the testing methods to design and certify protective headgear.

Even though the UCDBTM brain properties were derived from experimental tests and have a mechanical behaviour similar to that of actual human brain tissue (Horgan and Gilchrist 2004), the model cannot be used to determine a specific injury risk as it still needs to be validated against a large number of real world head impacts (both with and without injuries), to generate adequate injury risk curves. More accurate constitutive models and material property data would also serve to make model predictions of value for specific impact cases. Nevertheless, it is still possible to use the UCDBTM to establish comparative trends between impact scenarios, to determine how a change in external loads affects changes in brain stress and strain. From the trends given by the UCDBTM in this study, it can be seen that future head protection mechanisms could be designed in order to minimise angular acceleration in order to reduce stress and strain in the brain.

5. Conclusion

The results of this study show that recording head angular accelerations in helmet impact tests could prove useful for predicting injury risk and designing helmets to prevent brain injury more effectively. A change in linear acceleration would not reflect the same change in injury related loads to neural tissue such as stress or strain, while rotational acceleration is seen to reflect these changes more accurately. It was seen that angular acceleration seemed to reflect the values and sensitivity of the side impact stress-strain response of the brain material. Therefore, angular acceleration measurements could be included in headform impacts even when there is no evident angular acceleration source as in the case of glancing impacts; in this case angular acceleration was generated as the headform load does not pass through the centre of mass and is not concentrated on one point. Angular accelerations were generated due to the inertial properties of the headform. The headform inertial properties and its shape contributed to increasing the impact sensitivity of the headform to different impact positions (especially the side impact), which, in turn, reflected the impact position sensitivity of the UCDBTM output variables.

The head injury related tissue loads from the UCDBTM are sensitive to impact position. Therefore, it is possible to say that the reduction of UCDBTM outputs with respect to a helmet design change depends heavily on impact position. The headform parameter that best reflected impact position influence was angular acceleration. This indicates that changes based on angular acceleration would best reflect changes in UCDBTM stress and strain outputs. All of this suggests that angular acceleration is more important for the purposes of predicting brain injury than linear acceleration, which, in turn, suggests that angular acceleration needs to be considered when designing protective headgear.

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Appendix A. Summary results for correlation analyses

TABLE A1

TABLE A2