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Head motions while riding roller coasters: Implications for brain injury

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Abstract

The risk of traumatic brain injury (TBI) while riding roller coasters has received substantial attention. Case reports of TBI around the time of riding roller coasters have led many medical professionals to assert that the high gravitational forces (G-forces) induced by roller coasters pose a significant TBI risk. Head injury research, however, has shown that G-forces alone cannot predict TBI. Established head injury criteria and procedures were employed to compare the potential of TBI between daily activities and roller coaster riding. Three dimensional head motions were measured during three different roller coaster rides, a pillow fight, and car crash simulations. Data was analyzed and compared to published data using similar analyses of head motions. An 8.05m/s car crash lead to the largest head injury criterion measure (HIC_{15}) of 28.1 and head impact factor (HIP) of 3.41, over six times larger than the roller coaster rides of 4.1 and 0.36. Notably, the linear and rotational components of head acceleration during roller coaster rides were milder than those induced by many common activities. As such, there appears to be an extremely low risk of TBI due to the head motions induced by roller coaster rides.

Keywords

Roller Coaster; Head Injury; G forces; Injury Risk

Introduction

It is well recognized that many everyday activities involve a risk of traumatic brain injury (TBI) such as bike riding, roller blading, or playing contact sports. An accidental yet serious blow to the head can cause brain injury due to extremely rapid head motions that translates to damaging deformation of brain tissue^{1–6}. Injury thresholds based on the biomechanics of head motion have been extensively characterized and used as standards to determine the effectiveness of head protection measures^{7–13}. However, in both the general press and medical literature there has been confusion over how head motions are linked to TBI.

Gravitational force (G force), a directionless quantity of linear acceleration, is often inappropriately reported as the sole risk factor for brain injury. For instance, a series of medical case reports have described a potential causal relationship between intracranial hemorrhage and riding 'high G force' roller coasters^{14–18}. A recent review in a medical journal concluded that "emergency physicians should consider amusement park rides a possible cause of unexplained neurologic events", related to "dangerously high G forces"¹⁹. Accordingly, news stories that quote physicians frequently attribute high G forces induced by roller coasters to causing TBI in some riders^{20–23}.

Misperceptions of the relationship between high risk activities, G forces and TBI have had a surprisingly broad impact in our society. Without regard to years of scientific head injury research, news reports and anecdotal medical case reports have inspired legal suits for substantial monetary damages on behalf of roller coaster riders. As such, legislative acts have proposed to limit the level of G forces on amusement park rides at the state level^{21, 24, 25} and regulation of the amusement park industry at the federal level^{20, 21, 24, 26}. It is important to note, however, there is no scientific evidence demonstrating that G forces induced by amusement park rides pose any risk of TBI. To the contrary, two independent scientific panels and an engineering consulting firm failed to find a connection between brain injury and roller coaster riding^{27–29}. In addition, by using a simple mathematical model, our group previously calculated that the peak head accelerations during roller coaster rides were far below standardized thresholds for injury³⁰.

The source of confusion appears to be a fundamental misunderstanding of how G forces play a role in the biomechanics of TBI. A long history of brain injury and motor vehicle safety research has shown that a peak G force measurement alone is a very poor measure to determine the probability of injury to the brain^{1, 4–8, 11, 12, 34, 35}. Rather, all the kinematic parameters of head motion must be considered. The direction (linear and rotational in three dimensions), duration and magnitude of the motion are all important parameters to accurately determine if TBI thresholds have been exceeded.

To address the general misunderstanding of the role of high G forces and TBI in typical daily activities, we examined real-time 3-D head motions of volunteers during roller coaster rides, low speed car crashes and strikes with a pillow. The collective data were interpreted using established head kinematic parameters and compared to known thresholds of brain injury.

Methods

Selection of Participants

Four volunteers were selected ranging in age and weight to assess individual differences. Subject 1 was a 27 year old male, 165 lbs, 70 inches tall; subject 2 was an 11 year old male, 100 lbs, 57 inches tall; subject 3 was a 13 year old male, 86 lbs, 55 inches tall; and subject 4 was a 24 year old female, 150 lbs, 62 inches tall. Volunteers were recruited by Six Flags Great Adventure to ride three roller coasters, participate in a pillow fight and a 5mph car bumper hit. A Six Flags review board of Professional Engineers familiar with amusement ride test protocols approved the experimental procedures and participant consent. Each volunteer was asked to sign an informed consent form. In the case of the minors, the consent forms were also signed by the minors' guardians, who were present during the testing.

Study design and setting

Instrumented volunteers took part in five intense activities while the three dimensional kinematics of their heads was measured. Experiments took place on three distinctively different roller coasters at a Six Flags Amusement Park: (1) an inverted (track is overhead)

looping coaster with a top speed of 55 miles per hour (mph), (2) a linear induction motor launch coaster that rapidly accelerates the riders to 65 mph and has several intense loops, and (3) a tall and fast non-inverting coaster with significant drops and a top speed of 85 mph. Each volunteer rode in the middle of the train under normal operating conditions. Two additional tests were performed to gather relative data from two common experiences that involve rapid accelerations to the head: (4) a pillow fight and (5) a common car bumper hit of approximately 5 mph into a barrier.

Methods of measurement

An instrument biteplate was built for each subject that could be placed in the mouth and held in place by biting, Figure 1. Three linear accelerometers (Crossbow, model #CXL25M3, San Jose, CA) and three angular rate sensors (Murata, model #ENC-03J, Tenjin Nagaokakyoshi, Kyoto, Japan) were mounted onto the biteplate in an orthogonal fashion. The x-axis lies in the anterior-posterior direction, y-axis in the lateral direction and z-axis in the axial direction. Accelerometer and rate sensor data was collected on a TDAS Pro Acquisition Unit (DTS Inc., Seal Beach, CA) at a 10,000Hz sampling rate with a 3,000Hz anti-aliasing filter. A total of 6 channels of data were collected for the entire duration of the test. All DTS data acquisition hardware meets the requirements of SAE J211 and is certified to the National Highway and Traffic Safety Administration (NHTSA), the Federal Aviation Administration (FAA), and ISO 6487 standards.

Data Analysis

To remove measurement noise, linear accelerations were filtered with a 1,650Hz low pass filter as set forth in SAE J211 for CFC 1000 data. Angular velocities were filtered with a low pass filter of 600 Hz³⁶. Amusement rides produce rigid body accelerations in the frequency range up to 1.5 Hz, therefore angular velocity measurements were also filtered with a high pass filter of 1.5 Hz to remove DC offsets in the data per SAE J211 and J1727. Filtering was performed only on raw data prior to further calculations.

Angular accelerations of the head were calculated by differentiating the angular velocity measurements (three point centered difference) and linear velocities were obtained by integrating the linear acceleration measurements (Euler's Method) using Matlab Software (Mathworks, Inc., Natick, MA). The directional velocities and accelerations were then combined into resultant vectors and peak accelerations and velocities were identified.

The maximum accelerations and velocities were compared to reported tolerance levels for concussion 37–39, subdural hematoma (SDH)^{40, 41}, and DAI 8–35. In addition, we related our results to other reported measurements of head kinematics that occur during: daily living activities⁴², heading a soccer ball, football hits, hockey contact^{13, 43–46}, boxing⁴⁷, and an 18MPH rear end car crash test⁴⁸ (M. Kleinberger and A. C. Merkle, Johns Hopkins University Applied Physics Laboratory, personal communication, February 11, 2005).

Head Injury Assessment Functions

The results in this study and comparative published data were used to calculate, where possible, mathematical predictors of head injury developed for use in automobile safety testing¹¹. The head injury criterion (HIC), a federally mandated motor vehicle safety standard, was calculated by the expression:

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

Here, the conservative HIC_{15} was used where the time interval (t_2-t_1) was 15ms. Thresholds for the HIC_{15} range from a maximum value of 390 in car crash safety standards⁴⁹ to a value of 151 for a mild injury¹³.

Since the HIC only evaluates linear accelerations of the head, the head impact power (HIP)¹², a function of both linear and rotational accelerations and velocities in three dimensions, was also evaluated by the expression:

$$HIP = 4.5a_x \int a_x dt + 4.5a_y \int a_y dt + 4.5a_z \int a_z dt + 0.016\alpha_x \int \alpha_x dt + 0.024\alpha_y \int \alpha_y dt + 0.022\alpha_z \int \alpha_z dt$$

Calculation of the Probability of Concussion from Football data

The occurrence of concussion as a result of football tackles was used to approximate the probability of sustaining a concussion in other recreational activities⁴⁶. In this study, the diagnosis of concussion in National Football League (NFL) players strongly correlated to the value of the HIC_{15} . A probability curve of concussion vs. HIC was constructed using the Consistent Threshold method, a non-parametric method for ranking censored data^{50, 51}.

Results

A time history of the three dimensional kinematics of the adult and child head was acquired during three roller coaster rides, a pillow fight and a 5mph (2.2m/s) car bumper hit. Rotational velocities and linear accelerations were measured during each activity, from which, rotational accelerations and linear velocities were calculated. Directional x, y, and z components were combined into resultant vectors and the peak values were identified, Table 1. In addition, head kinematic test data was collected for an 18mph (8.1m/s) car crash simulation using test dummies from the Johns Hopkins University Applied Physics Laboratory (M. Kleinberger and A. C. Merkle, personal communication, February 11, 2005). The results were compared to published kinematic data from peer-reviewed studies on non-penetrating brain injury and to head motions measured in contact sports.

The 18mph (8.1m/s) car crash simulation resulted in the highest measurements of linear acceleration (29.3 xG), linear velocity (4.0 m/s) and rotational velocity (17.1 rad/s) of the head, Table 1. The highest level of rotational acceleration (2054 rad/s²) was measured during the pillow fight. Interestingly, the pillow fight generated peak head accelerations and velocities greater than the three roller coaster rides. Despite the difference in the three roller coaster rides (i.e. speed, turns, loops), they lead to similar head motions. It is important to note that variations in head motions were small between the roller coaster rides, pillow fight and 5mph (2.2m/s) car bumper hit.

It is well accepted that individual peak values of acceleration or velocity of the head alone are not adequate to predict the risk of a brain injury^{1, 4-8, 11, 12, 34, 35}. The time interval over which they occur must also be analyzed. Two well-established head injury assessment functions, the head injury criterion (HIC) and head impact power (HIP), were calculated to evaluate the head motion as a function of time, Table 1. The 18mph (8.1m/s) car crash lead to the largest values of $HIC_{15} = 28.1$ and $HIP = 3.41$. In contrast, the accelerations and velocities associated with the roller coaster rides and pillow fight occur over shorter time frames and therefore much smaller HIC_{15} and HIP values. This highlights the importance of both the magnitude of the head motions and the time frame over which they occurred. For instance, the male adult in the pillow fight experienced the same linear acceleration as the male child #1 on roller coaster #2 (10.5 and 10.2 xG respectively), however, the HIC_{15} values are different (1.3 and 4.1 respectively). The transient component of these head motions can also be seen on inspection of individual recordings shown in Fig 2. When

compared to Federal Automobile Safety standards, we find that the HIC_{15} values are two to three orders of magnitude below the minimally accepted values of 390 for infants and 700 for adults⁴⁹.

Test data from studies on non-penetrating brain injury were collected from peer-reviewed literature to compare the potential risk of sustaining an injury from the intense activities of this study, Table 2. All data analyzed in this study are substantially below the lowest reported kinematic parameter that resulted in a measurable level of brain injury including: diffuse axonal injury⁸, coma and concussion^{37, 38}, tearing of bridging veins⁴¹, and concussion from football tackles^{13, 46}. Importantly, these thresholds are not limited to one measured parameter. They were based on all reported measurements. Included in Table 2 are data from head motions in several common activities that did not lead to injury. The data in this study are consistent with peak values of rotational accelerations and velocities seen in soccer heading, and linear accelerations similar to plopping in a chair, yet far below what a boxer or a football player experience.

Recently published data on the occurrence of concussion in football provides a reference database to evaluate the probability of sustaining a concussion in similar intense activities⁴⁶. The results demonstrated that the HIC_{15} correlated closely with the diagnosis of concussion. We used this data to construct a non-parametric probability curve for concussion, Fig 3a. This curve indicates that the probability of sustaining a concussion is zero up to a HIC_{15} value of 77, corresponding to the first diagnosed concussion as a result of a football tackle. The roller coaster, pillow fight and 5mph (2.2m/s) car crash were more than 19 times below this threshold. Even the 18mph (8.1m/s) car crash was three times below this threshold. Comparing the calculated HIC_{15} values of this study and from other contact sports in the literature, figure 3b illustrates that only boxing and football tackles fall within the probability of suffering a concussion.

Discussion

Real time 3-D motions of the heads of volunteers were measured during rides on three different roller coasters, strikes with a pillow, and low speed car crash simulations. It was found that the peak head accelerations and velocities from these intense events were all comparable and fell far below the established biomechanical thresholds for TBI.

When predicting TBI thresholds, all the parameters of head motion must be considered. Specifically, injury to the brain is dependent upon 1) the direction of head motion, 2) the magnitude of velocity and acceleration, and 3) the time frame over which it occurs^{7, 11, 12, 30}. For instance, linear motions result in focal injuries such as skull fracture, cerebral contusions and hematomas at the site of impact to the head. Rotational motions, on the other hand, causes extensive deformation of the brain and vasculature, inflicting diffuse axonal injury throughout the white matter. Overt damage such as tissue tears in the white matter and intraparenchymal hemorrhage only occur from deformations caused by exceptionally high levels of rotational acceleration, over very short time periods^{6, 40}. The results of this study show that the linear and rotational head accelerations produced by riding a roller coaster are similar to the well tolerated head motions experienced during a pillow fight or heading a soccer ball, Table 2 and Fig. 3.

The perception that riding roller coasters presents a risk of TBI is currently not supported by epidemiological or scientific data. Rather, this misperception appears to stem from a fundamental misunderstanding of the role of G forces in TBI linked with a handful of case reports of patients suffering brain bleeding around the time of riding a roller coaster^{14, 15, 17, 18, 63–70}. The human body can withstand very large G-forces when they occur over very

short time periods. For instance, a sneeze generates linear accelerations of up to 10 G's, but occurs over only 0.002 seconds with no ill effect ⁴². Likewise, a boxer can withstand approximately 3 msec punches of 100 Gs or more with no overt signs of injury ⁴⁷. In contrast, a fighter pilot will lose consciousness at 5–9 G's if these forces are sustained for over 40 seconds. In this case, loss of consciousness results from restriction of blood flow rather than mechanical injury to the brain ⁵⁵. These collective studies illustrate that scalar measurement of linear acceleration (G-force) alone is a poor measure to assess risk to injury. Indeed, this limitation led to the development of the HIC₁₅ and HIP by automobile safety researchers to integrate the linear and head rotational motions over their respective durations ¹¹.

A probability curve for mild levels of TBI was created to relate the degree of head motions during a roller coaster ride to those that can occur in contact sports, Fig. 3a. This curve statistically correlates the injury risk of concussion injury in football players with the HIC₁₅ value caused by a football tackle. The probability of concussion in football was compared to the HIC₁₅ values from roller coaster rides in this study and from published reports of other intense activities ^{13, 43–47, 56–62}, Fig 3b. The HIC₁₅ values that occur during a roller coaster ride fall below the probability of injury in all contact sports using current data from the literature, demonstrating a lower risk of TBI than from playing a sport.

It is important to note that it is currently difficult to establish biomechanical thresholds that will cause mild levels of TBI. In particular, established head injury criteria consider the entire kinematics of head motion, however, most peer reviewed experimental studies do not report all these parameters. In addition, it can be difficult to diagnose brain injury and injury severity clinically. The diagnosis of concussion, for example, is very controversial ⁵². Accordingly, there is a lack of published experimental and clinical data to directly correlate head motion to minor injuries to the brain.

Another important limitation in assessing the cause of brain injury is the unknown presence of pre-existing conditions that could augment a person's susceptibility to injury. For example, a pre-existing brain aneurysm might rupture during or near the time of a roller coaster ride, as has been demonstrated in at least one fatality case ^{53, 54}. However, rupture of the aneurysm could occur from many factors other than head accelerations, such as hypertension due to excitement. Even if these individuals had preexisting conditions such as cerebral vascular malformations, it is unknown whether hemorrhage is more likely to be induced by the level of head motions during roller coaster rides as opposed to other daily activities. The current study does not address this possibility.

Our current empirical data supports two scientific panels' opinions ^{27, 28} as well as previous results from a computational model ³⁰. Specifically, head motions during roller coaster riding fall within the range of normal activities and are far below thresholds of TBI in normal individuals.

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Figure 1.

Volunteer instrumentation. The instrument biteplate consists of three linear accelerometers and three angular rate sensors. The x-axis lies in the anterior-posterior direction, y-axis in the lateral direction and z-axis in the axial direction. Head rotations about the x-axis lie in the coronal plane, rotations about the y-axis lie in the sagittal plane and rotations about the z-axis lie in the axial plane.

Figure 2.

Resultant peak linear accelerations for the male adult in the pillow fight and the male child #1 on roller coaster #2. While the peaks are the same, (10.5 and 10.2 xG respectively), the time over which each of these head motions occur is different. Accordingly, the HIC values are different, 1.3 for the pillow fight and 4.1 for the roller coaster demonstrating the importance of the time over which the head motion occurs.

Figure 3.

The probability of concussion versus the calculated Head Injury Criterion (HIC₁₅). a) Probability curve of concussion in football tackles vs. HIC₁₅ constructed using the Consistent Threshold method. The first occurrence of concussion was at a HIC₁₅ of 77. b) Comparison of the probability of concussion in football tackles to other intense activities and contact sports. Considering other activities for which data was available, only football tackles and boxers fell within the probability of concussion.

Table 1

Head kinematic parameters from measured and calculated data from each subject in each activity. Values are the peak values of the resultant vectors. Bolded values indicate the overall maximum values discussed in the text.

	Linear acceleration xG	Linear Velocity m/s	Angular acceleration rad/s ²	Angular velocity rad/s	HIC ₁₅	HIP
Coaster 1						
Male adult	6.4	1.2	918	7.2	1.2	0.16
Male child 1	7.8	0.6	948	7.7	2.1	0.13
Male child 2	6.2	0.8	775	6	1.1	0.08
Female adult	5.4	0.7	637	7.6	0.9	0.06
Coaster 2						
Male adult	7.9	1.4	721	7	1.7	0.22
Male child 1	10.2	1.2	1547	10.5	4.1	0.36
Male child 2	9.2	1.0	1575	8	1.7	0.21
Female adult	6.3	1.3	643	8.2	1.3	0.18
Coaster 3						
Male adult	6.8	0.8	719	5.1	1	0.1
Male child 1	6.9	0.8	1847	10.9	1.6	0.36
Male child 2	6.8	0.7	833	6.4	1.4	0.12
Female adult	7.8	0.9	1318	7.3	1.8	0.18
Pillow fight						
Male adult	10.5	0.8	2054	10.5	1.3	0.21
Male child 1	5	1.0	711	15.2	0.6	0.32
Female adult	4.7	0.5	1537	6.8	0.4	0.2
5mph bumper hit						
Male adult	5.2	1.2	555	9.6	0.8	0.2
Male child 1	4.9	0.9	558	7.3	0.7	0.09
Male child 2	3.5	1.3	358	10.9	0.3	0.12
Female adult	4.2	0.9	176	4.6	0.5	0.05
18mph car crash						
	29.3	4.0	1,261	17.1	28.1	3.41

Table 2

Test data from studies on non-penetrating brain injury collected from peer-reviewed literature. Data measured from the intense activities of this study are well below the lowest reported kinematic parameters that resulted in a measurable level of brain injury and below studies of other activities, such as contact sports, that result in no injury to the brain.

	Author	Test Subjects	Test/Activity	Injury	Angular acceleration rad/s ²	Angular velocity rad/s	Linear acceleration xG
Injury	Margulies	Primates	Rotational injury	DAI	9000	70	---
	Ommaya, 1967	Primates	Rotational injury	99% concussion	7500	---	---
	Lowehielm, 1975	Human Surrogates		Bridging veins	4500	50	
	Zhang, 2004	Human Volunteers	Football Tackle	Mild TBI	7354	---	103
No Injury	Ommaya, 1971	Primates	Whiplash	concussion	1800	45	---
	Pincemaille, 1989	Human Volunteers	Boxers	None noted	6155	30	54
	Zhang, 2004	Human Volunteers	Football Tackle	No Injury	4204	---	55
	Naunheim, 2003	Human Volunteers	Soccer Heading	None noted	1457	11	20
	Ewing, 1975	Human Volunteers	Whiplash	No Injury	1223	25	20
	Naunheim, 2000	Human Volunteers	Soccer Heading	No Injury	---	---	54.7
	Naunheim, 2000	Human Volunteers	Hockey Check	No Injury	---	---	35
	Allen, 1994	Human Volunteers	Plop in a Chair	No Injury	---	---	10
	This study, maximum values	Human Volunteers	Roller Coaster Riding	No Injury	2054	15.2	10.5