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Evaluation of the Biofidelity of FMVSS No. 218 Injury Criteria

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ABSTRACT

The biofidelity of the injury criteria used by Federal Motor Vehicle Safety Standards (FMVSS) No. 218 were examined against biomechanically based injury metrics. The current FMVSS No. 218 for motorcycle helmets is a generalized safety standard based on acceleration peaks and dwell times. FMVSS 218 uses a rigid headform mounted with a unidirectional accelerometer for conducting helmeted drop tests at a specified speed. The biomechanical basis of FMVSS No. 218 is not known, however. This study builds on previous work which developed a technique for measuring the under-helmet pressure contour on the headform during a FMVSS No. 218 impact attenuation test. By measuring the pressure applied directly to the head, the need for including the helmet in finite element analysis was bypassed. The headform pressure data from 80 impact tests to the front, crown and side of a helmet were used in finite element model simulations to predict skull fracture. FMVSS No. 218 impact attenuation injury criteria were correlated against the calculated skull strain and the generalized Skull Fracture Correlate (Chan et al., 2007 ESV). The acceleration data from the drop tests were used in NHTSA's SIMon model to predict brain injury. The FMVSS No. 218 injury criteria were correlated against SIMon's injury metrics for diffuse axonal injury (DAI), brain contusion, and subdural hematoma. FMVSS No. 218 injury criteria were also correlated against HIC. It was found that peak headform acceleration was the best correlate for all injury metrics. Dwell times over 150g and 200g both had very poor correlation with injury metrics. This research has shown that peak head acceleration can be an acceptable injury metric for the FMVSS No. 218 test method. However, the current FMVSS No. 218 limit of 400g allows for a high probability of skull fracture and brain injury.

INTRODUCTION

The National Highway Traffic Safety Administration (NHTSA) of the US Department of Transportation (DOT) estimates that motorcycle helmets saved 1,546 lives in 2005, and that 728 more could have been saved if all motorcyclists had worn helmets. Unfortunately in 2004, 4008 motorcyclists were killed in accidents and an additional 76,000 were injured. The motorcycle fatality and injury rates have been increasing steadily since 1998 (NHTSA, 2005). Although motorcycles make up only 2 percent of all registered vehicles, in 2005 they account for 10 percent of total traffic fatalities.

COST 327, a 2001 study on motorcycle safety helmets by the European Commission, analyzed a set of motorcycle accidents where the operator wore a helmet (COST 327, 2001). It was reported that 67% of casualties sustained head injury and 27% sustained neck injury. The helmet damage was evenly distributed across the helmet, although impacts to the crown were less frequent. Out of 409 injuries investigated, 22.5% were skull fracture, 0.5% concussion, 10.3% contusion, and 12% subdural hematoma. The objective of this study is to evaluate the biomechanical basis of the impact attenuation tests of FMVSS No. 218 versus known biomechanically-based injury criteria.

FMVSS No. 218: Motorcycle Helmet Standard

FMVSS No. 218 prescribes a series of tests that a helmet must pass in order to meet DOT approval (Regulations, 2006) and establishes requirements for impact attenuation, penetration, and retention. The purpose of the impact attenuation requirement is to protect the head from impact shock. In an FMVSS No. 218 attenuation test, the helmet is fitted to a metallic headform that is instrumented with a single linear accelerometer. The headform is then attached to a vertical monorail guided drop assembly. The vertical acceleration of the headform is measured during the drop of the headform/helmet combination onto a metallic anvil.

To pass the impact attenuation requirement of FMVSS No. 218, the criteria,

$$A_{max} \leq 400 \text{ g}, T_{150g} \leq 4 \text{ ms, and } T_{200g} \leq 2 \text{ ms},$$

must be satisfied for each drop. A_{max} is the peak acceleration. The dwell times, T_{150g} and T_{200g}, are defined as the cumulative time at which the acceleration versus time curve exceeds 150g and 200g, respectively.

The biomechanical basis of the FMVSS No. 218 helmet criteria is not known. The acceleration and dwell time requirements were taken from the ANSI Standard. However, the method and speed of delivery were modified. Previous studies have recommended a decrease in the peak acceleration limit from 400g to 300g (Thom et al., 1997; Vander Vorst et al., 2000). However, there has not been a study involving an extensive investigation of the skull and brain injuries associated with these accelerations. The correlation between dwell times above 150 and 200g and head injury is also not well understood. There are, however, biomechanically validated methods for predicting head injury. By comparing the injury criteria used in FMVSS No. 218 against validated injury metrics for skull fracture and brain injuries, the biofidelity of FMVSS No. 218 can be evaluated.

Biofidelic Injury Criteria

For skull fracture, FMVSS No. 218 will be compared against the generalized linear skull fracture criteria developed over the past four years under NHTSA sponsorship. Vander Vorst (Vander Vorst et al., 2003; Vander Vorst and Chan, 2004) first presented the linear skull fracture criteria called skull fracture correlate (SFC). The head impacts in this study involved side, frontal, and crown hits; therefore, the generalized SFC injury curve is used.

For brain injuries, FMVSS No. 218 will be evaluated using the NHTSA SIMon finite element head model (Takhounts et al., 2003). SIMon calculates three injury metrics from the three dimensional head kinematic data of the headform. These criteria are: cumulative strain damage measure (CSDM), a correlate for diffuse axonal injury, dilatational damage measure (DDM), an estimate of the potential of contusions, and relative motion damage measure (RMDM), a correlate for acute subdural hematoma.

For general head injury, FMVSS No. 218 will also be compared against the Head Injury Criteria (HIC). HIC was incorporated into FMVSS No. 208 (frontal impact protection in cars) and is widely used in the automobile industry. However, HIC was never incorporated into FMVSS No. 218 and is only used in the U.N. Economic Commission for Europe motorcycle helmet safety standard (ECE 22.05) with a limit of 2400.

METHODS

Instrumentation for Measuring Headform Pressure

To predict the efficacy of a particular helmet using finite element calculations coupled with the head would require a validated structural model of the helmet. This task is impractical for each helmet model to be tested. However, if during a drop test, the pressure applied by the helmet to the headform were measured, and this pressure applied to the anatomical finite element model to compute the skull strain, then the probability of skull fracture could be predicted for the specific helmet. To accomplish this, instrumentation to measure the pressure contours on the headform was developed.

TekScan's FlexiForce sensors were chosen to measure the contact pressure between the helmet liner and the headform. For further details on the preparation of the FlexiForce sensors, see the author's paper from the 34th International Workshop on Injury Biomechanics Research entitled "Measurement of Under-Helmet Force Distribution on FMVSS No. 218 Headform." A total of 36 to 46 FlexiForce sensors were used to cover the impact area of the headform.

The FlexiForce sensors were distributed in a regular grid pattern, and it was assumed that the pressure measured by a sensor was uniform over the sub-grid area with the sensor at the center. For each impact configuration (crown, side or frontal drop), it was assumed that the contact load would primarily be borne by the impact side of the headform and tangential loads were negligible. Therefore, the sensors were placed only out to the edge of the impact side of the headform. The total impact area was estimated for each impact configuration and distributed evenly to each sensor sub-grid area for inputs to the anatomical finite element model for skull fracture prediction.

Impact Attenuation Tests

One hundred and twenty drop tests were conducted to gather input data for the finite element model simulations. Impact attenuation tests were performed according to specifications given in FMVSS No. 218. The helmets were dropped against a flat anvil at a speed of 6 m/sec or against a hemispherical anvil at a speed of 5.2 m/sec. Half the tests were conducted against the flat anvil, the other half were against the hemispherical anvil. There were three impact sites selected for each helmet: crown, front, and left side.

Twenty helmets were used in the tests. Helmet types consisted of a mixture of full face helmets, open face helmets, and half helmets. Helmets were randomly placed in a testing matrix consisting of impact location and anvil type. As per FMVSS No. 218 protocol, each helmet was secured to the headform, so that they did not shift before impact. Each helmet was struck at each of the three impact locations with two successive identical impacts. All attenuation tests were conducted at ambient conditions. Tests at the other environmental conditions identified by FMVSS No. 218 were not conducted.

Skull Fracture Finite Element Model

The maximum principal skull strain was calculated for each impact attenuation test using a refinement of the anthropomorphic, medical imaging-based, finite element model of Vander Vorst et al. (2004). The baseline model was composed of 24,000 elements and resolved the outer and inner tables, diploe, brain, scalp, and face. The mass of the baseline model was 4.54 kg. The skull components were modeled using fully integrated thick shells and the brain, scalp, and face were modeled with fully integrated bricks. Since this model was based on CT imaging of a PMHS, the skull shape and thickness are anatomically correct. The thickness of the compact skull tables was set to be 1 mm uniformly, since they were too thin to be resolved from the CT scan. The 1-mm value was based on measurements of photographic cross-sections from the Visible Man project (NIH, 2000). The properties of the biological materials were taken from the open literature. The elastic properties of compact skull bone were from

Wood (1971). Diploe was taken to be linear elastic (Khalil and Hubbard, 1977). The linear viscoelastic properties of the brain were from Takhounts et al. (2003). Scalp was assumed to be viscoelastic with properties calibrated by Vander Vorst et al. (2003). Material properties are shown in Table 1. All finite element model simulations were performed using Version 9.70 of LS-Dyna3d software.

Table 1. Material Properties of Finite Element Model.

Part – Material Property	Value
Brain – Viscoelastic	
Specific gravity	1.0
Bulk modulus	0.55 GPa
Short term shear	10 kPa
Long term shear	5 kPa
Decay time	0.01 sec
Diploe – Linear elastic	
Specific gravity	1.8
Young's modulus	0.74 GPa
Poisson's ratio	0.05
Scalp – Viscoelastic	
Specific gravity	1.3
Bulk modulus	6.8 MPa
Short term shear	2.5 MPa
Long term shear	0.68 MPa
Decay time	0.17 ms
Tables – Elastic	
Specific gravity	3.06
Young's modulus	15.8 GPa
Poisson's ratio	0.35

The sensor locations on the headform were mapped directly to the scalp elements of the skull fracture FEM. For example, if the headform had a line of seven sensors equally spaced from the anterior to posterior reference line, the location of the reference plane on the skull fracture FEM would be determined and the seven sensor locations would be equally spaced similar to the headform. The maximum strain in either the inner or outer table of the skull for each helmet attenuation test was found and used in the statistical analysis.

SIMon Simulations

The time-acceleration profile of the headform during impact was imported into the SIMon finite element analysis package and used to evaluate traumatic brain injury. To calculate the full angular and translational motion of the head, SIMon uses a nine accelerometer package or angular velocity sensor data as inputs. The FMVSS No. 218 protocol, however, restricts headform motion to one axis. The single axis accelerometer data was imported to the corresponding axis of motion in the SIMon coordinate system, while a dummy waveform (+/- 0.5g sine wave) was used as input data in the other restricted axes. HIC and SFC were also calculated from the time-acceleration profile of the headform.

Each test was computed out to 20 msec. The injury measures recorded were: cumulative strain damage measure (CSDM), dilatational damage measure (DDM), and relative motion damage measure (RMDM). SIMon reports CSDM values at various tolerances of strain. Takhounts et al. reports that a CSDM with a tolerance of 15% strain in the brain achieved the best correlation with diffuse axonal injuries compared to other tolerances. Therefore, CSDM with a tolerance of 15% strain was used in this study. The RMDM threshold for injury was established using only sagittal impact data. Data from side hits were not used to evaluate RMDM; only crown and frontal results are reported as suggested in the SIMon documentation.

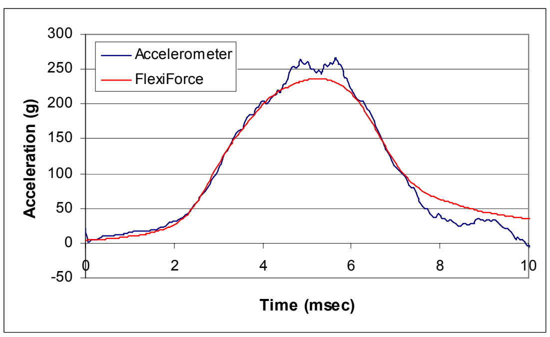
RESULTS

Impact Attenuation Tests

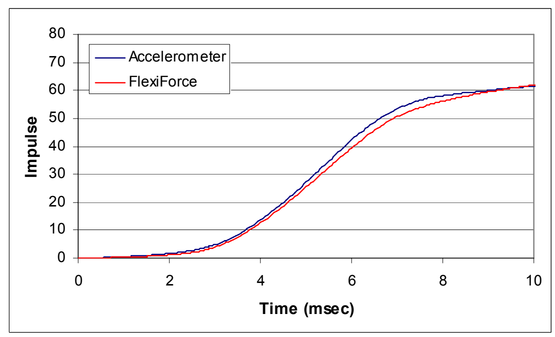
A total of 80 tests were used in analysis. Twenty percent of the helmets tested did not pass the current criteria of FMVSS No. 218. Six percent failed the 400g peak acceleration limit, 1% failed the 150g dwell time limit, and 13% failed the 200g dwell time limit. All the helmets that failed the 400g peak acceleration limit were from side impacts. It was also noticed that a majority of side impact tests against the hemispherical anvil resulted in an area of concentrated high pressure directly under the impact location. This is in contrast to the crown and front impacts against hemispherical anvils where the load is better distributed to the surrounding area.

The FlexiForce data were validated against the headform accelerometer data. The total vertical component of the force from the FlexiForce sensors was computed and divided by the mass of the head to get the acceleration. The headform accelerometer data and the headform acceleration calculated from the FlexiForce sensors were in good agreement, including the impulse comparison, as shown in Figure 1.

Head Acceleration



Head Impulse



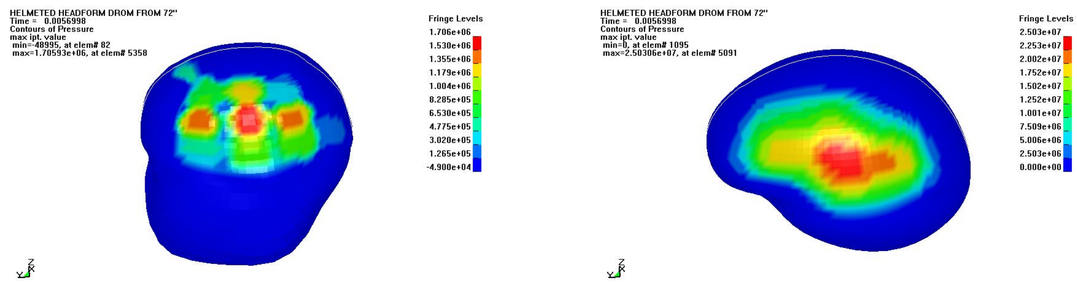
(a) Side Drop D1

Figure 1: Sample Head Acceleration and Impulse Data Comparison.

Skull Fracture Evaluation

A finite element model simulation was performed for each impact attenuation test using the headform pressure data as input. The simulation was run to 20 msec. From these simulations, the maximum principle strain out of the inner and outer tables was determined. The FEM head acceleration and head impulse closely matched the experimental results. Characteristic contour plots of the pressure applied to the scalp and the resulting pressure on the skull are shown in Figure 2.

Scalp Pressure (FlexiForce Input)	Skull Pressure (Predicted)

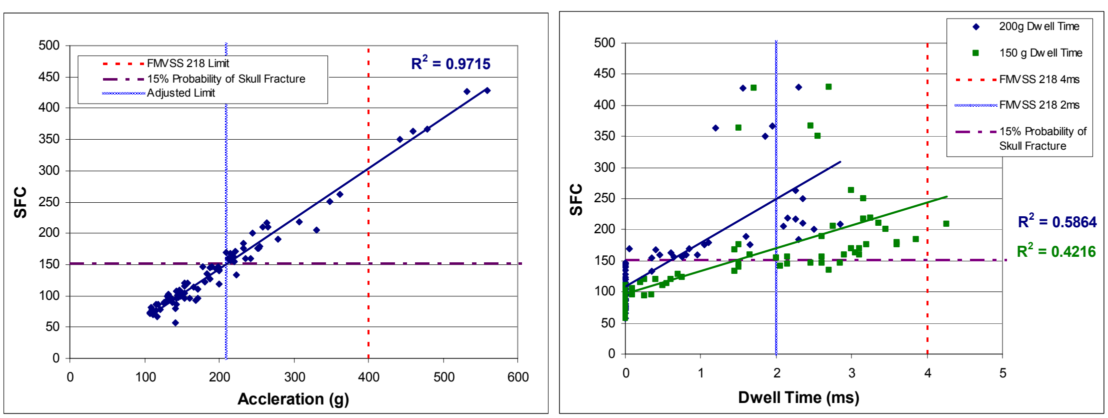


(a) Side Drop (viewed from left side)

Figure 2: Sample Pressure Contour Plots of Scalp and Skull.

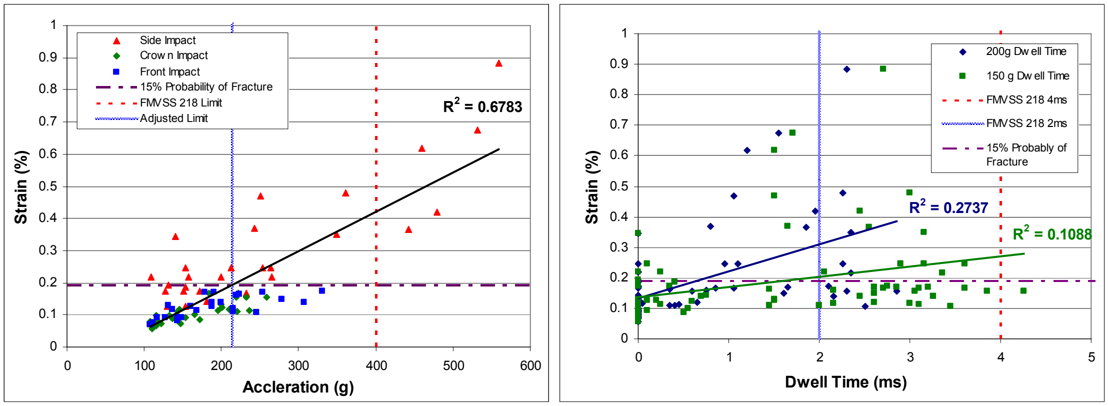
The finite element model did not reach the termination time of 20 msec for a few of the side impact cases. This was due to excessive deformation of the scalp from the very high pressures recorded during these drop tests. In these cases, the model was run out until failure and the last recorded skull strain was used. Failure usually occurred within a millisecond of the peak input pressure; therefore, it was assumed that the strain values are close to the actual values.

An adjustment of the SFC risk curve for the rigid FMVSS No. 218 headform was made using the finite element results. The standard SFC risk curve was originally established for the Hybrid III headform with a rubber skin, with 15% probability of skull fracture predicted by SFC=124g (Chan et al., 2007), but this value will change for the rigid headform with no skin. Fortunately, skull fracture can be predicted using the skull strain calculated from the anthropomorphic FEM. Therefore, the SFC risk curve was adjusted for the FMVSS No. 218 headform by correlating the SFC values calculated from the headform acceleration with the skull strain calculated from the FEM. For 15% probability of skull fracture, which corresponds to 0.19% of skull strain, the SFC value will be 151g for the FMVSS No. 218 headform, and this value will be used for comparison with the FMVSS No. 218 criteria. The skull fracture results correlated against FMVSS No. 218 injury criteria are shown in Figures 3, 4, and 5.



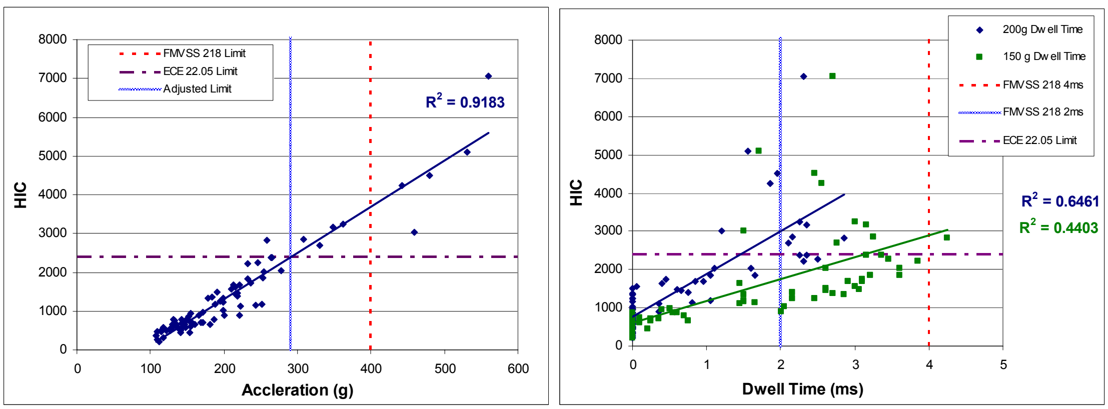
(a) SFC vs. Peak Head Acceleration (b) SFC vs. Dwell Times over 150 and 200g

Figure 3: SFC Comparison.



(a) Skull Strain vs. Peak Head Acceleration (b) Skull Strain vs. Dwell Times over 150 and 200g

Figure 4: Skull Strain Comparison.



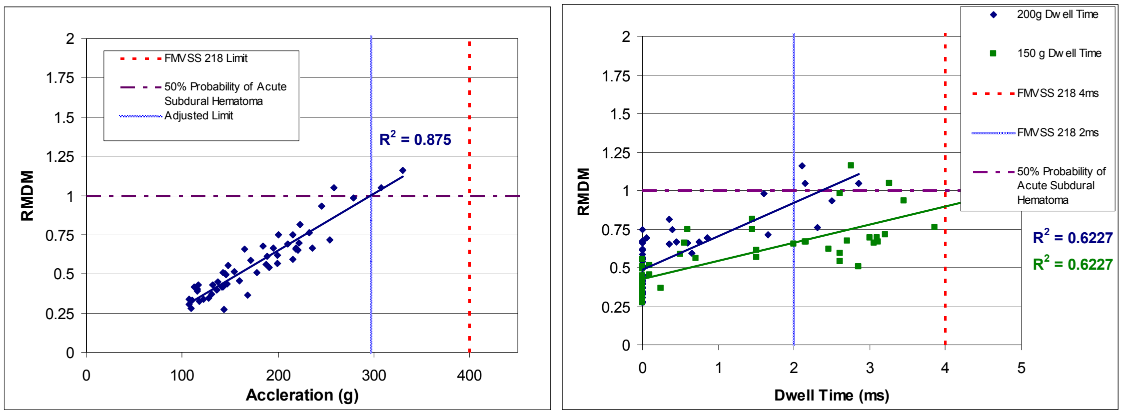
(a) HIC vs. Peak Head Acceleration (b) HIC vs. Dwell Times over 150 and 200g

Figure 5: HIC Comparison.

There is good correlation between SFC and peak head acceleration and HIC and head acceleration. There is fairly good correlation between peak skull strain and head acceleration. In each case, there is poor correlation between the dwell times and the injury criteria. This poor correlation is evidenced by the low R^2 values as well as the fair amount of false positive and false negative data points.

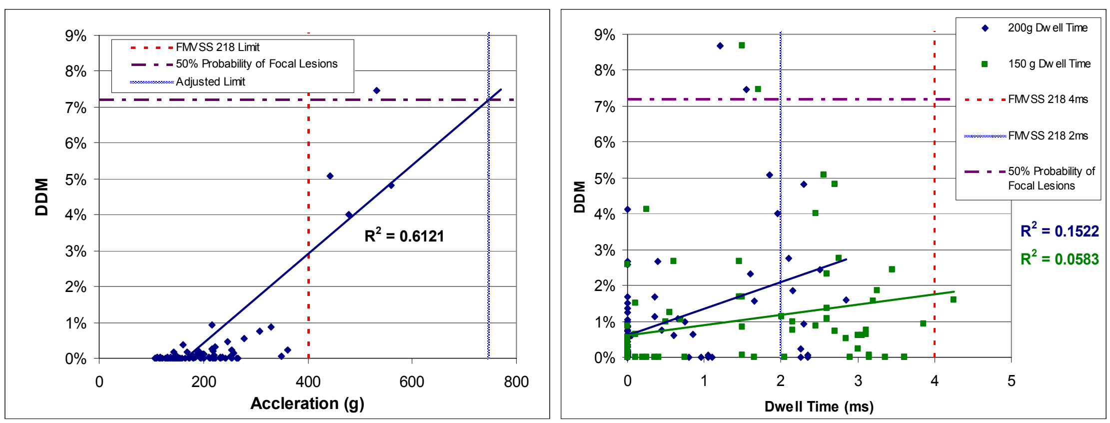
Brain Injury Evaluation using SIMon

SIMon analysis was performed for each helmet attenuation test. Acceleration data was entered in the Z-axis for crown drops, in the Y-axis for side drops, and in both X- and Z-axes for frontal drops. For each drop, SIMon reported no angular acceleration as expected for a rigid nonrotating headform. The SIMon head motion was verified visually using LS-Dyna's PrePost Processor. The peak values of CSDM (tol. = 0.15), DDM, and RMDM were recorded. Correlation plots of SIMon injury metrics plotted against FMVSS No. 218 injury criteria are shown from Figures 6-8.



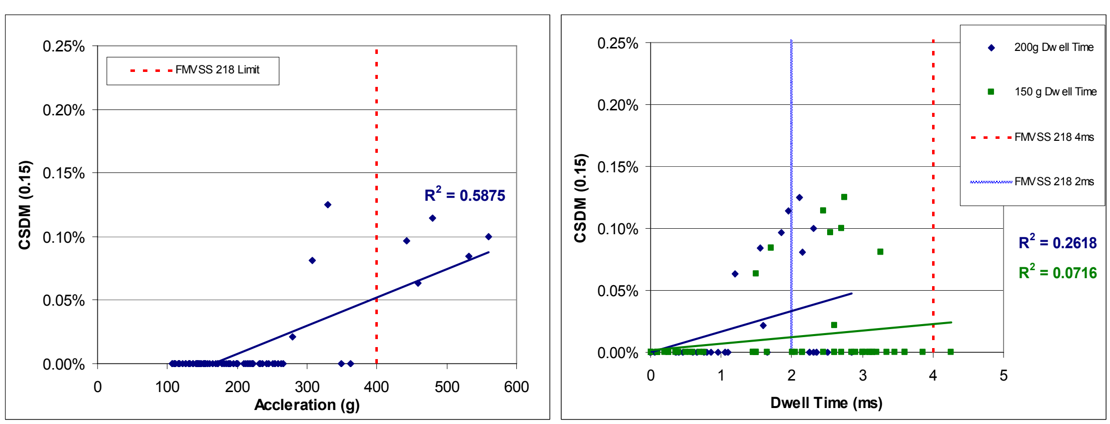
(a) RMDM vs. Peak Head Acceleration (b) RMDM vs. Dwell Times over 150 and 200g

Figure 6: RMDM Comparison.



(a) DDM vs. Peak Head Acceleration (b) DDM vs. Dwell Times over 150 and 200g

Figure 7: DDM Comparison.



(a) CSDM vs. Peak Head Acceleration (b) CSDM vs. Dwell Times over 150 and 200g

Figure 8: CSDM Comparison.

There is good correlation between RMDM and peak head acceleration but moderate correlation between RMDM and dwell times (Figure 6). There are some false positive and false negative data points for the dwell time comparison with RMDM (Figure 6b). Both the DDM and CSDM risk factors are grouped near zero in the peak head acceleration and dwell time plots. The data is inconclusive and no approximation of the acceleration values needed for a 50% probability for brain contusion or concussion will be conducted.

The overall results show that the peak acceleration is a good correlate with the risk factors for all the damage measures compared, but both dwell times are not. A summary of the peak acceleration comparison results is shown in Table 2, which also indicates the needed adjustments according to the published limits for the various damage measures. As shown in Table 2, the current 400g limit greatly underpredicts skull fracture (skull strain and SFC), RMDM and HIC. The adjusted peak acceleration values matching the published limits for skull fracture, RMDM and HIC are within the range from 210-297g.

Table 2. Summary of Damage Measures.

Damage Measure	Injury at 400g		Published Injury Limit		Adjusted G for published injury limit	R^2
	Measure	Probability	Measure	Probability		
Skull Strain	0.41%	88.0%	0.19%	15%	214	0.6783
SFC	304	92.0%	151	15%	210	0.9717
RMDM	1.37	80.0%	1	50%	297	0.875
DDM	2.90%	23.0%	7.20%	50%	n/a	--
CSDM (.15)	0.12%	1.0%	55.0%	50%	n/a	--
HIC (ECE)	3694	--	2400	--	290	0.9183

DISCUSSION

The percentage of helmet failures in this study is consistent with previous helmet tests. Out of the 20% that failed in the current study, the majority failed on the 200g dwell time requirement (65%). A USC study of 576 impact attenuation tests had a failure rate of 7%, with 85% of the failures due to failing the 200g dwell time requirement (Thom et al., 1997). A study by Vander Vorst et al. (2000) had a FMVSS No. 218 helmet failure percentage of 41% with 80% of failures due to the 200g dwell time requirement.

Results from this study have shown that both the 150g and 200g dwell time requirements of FMVSS No. 218 do not correlate well with any biomechanically-based injury damage metrics. There seems to be a moderate correlation between dwell times and RMDM and HIC; however, the peak acceleration is the much better correlate in all cases. The helmets that fail due to the 200g dwell time requirement are therefore failing due to a criterion which is a poor correlate to injury.

The skull fracture results suggest that the helmets are less effective for side impact protection than for frontal or crown impacts. The FEM results show that the helmeted side impacts produce higher skull strains than frontal and crown impacts. For those side impacts that generate high skull strain exceeding 0.4%, FlexiForce data indicate a concentrated high pressure zone under the impact point. The headform pressure data suggest that the helmets deform more significantly against side impacts than against frontal and crown impacts. This evidence is strong since pressure data were taken using a rigid headform with no skull deformation coupling. No attempt was made to evaluate the helmet shell and padding properties.

Skull fracture analysis results show that the peak acceleration measured by the FMVSS No. 218 protocol is a good indicator for skull fracture, although the limit should be adjusted lower. There is excellent agreement between the skull strain and SFC analysis in what the adjusted acceleration limit for a

15% skull fracture should be (214 and 210g respectively, Table 2). This close agreement gives high confidence in setting the peak acceleration limit for skull fracture.

The FMVSS No. 218 method is likely not adequate for prediction of concussion due to its use of a nonrotating rigid headform. The Wayne State Tolerance Curve, based on a linear acceleration criterion, predicts a threshold of 60 to 80g for concussion. Pellman et al. (2003) found the peak acceleration in concussion-causing impacts in professional American football to be 98 ± 28g. Using small primates data obtained from Ono et al. (1980) with scaling to humans, Vander Vorst et al. (2007) estimated a 175g peak linear acceleration limit for 10% probability of concussion. The Vander Vorst estimate was somewhat higher than the others because concussion was defined as dizziness in the human studies but unconsciousness in the small primate study.

The use of the nonrotating rigid headform is probably why the current SIMon model does not predict much of any concussion using FMVSS No. 218 data. The experimental protocol does not allow any rotation of the headform, but the SIMon model was validated using both translational and rotational acceleration. A study using SIMon showed that rotational acceleration contributes more than 90% of total strain seen in the brain (Zhang et al., 2006). It is known that DAI is highly dependant on rotation, as also indicated by other researchers (Margulies and Thibault, 1992). It is known that in real crash trauma conditions, it is very unlikely to have pure translation without rotation, and both linear and rotational accelerations are usually strongly correlated with each other.

The use of nonrotating headform may also have affected the prediction of RMDM by SIMon, even though RMDM showed a strong correlation with peak linear acceleration. Results from this study predicted that for the current FMVSS No. 218 peak acceleration limit of 400g, there is an 80% probability of subdural hematoma. An adjusted peak acceleration limit of 297g would result in an injury probability of 50%. It has been estimated with finite element analysis that rotational acceleration contributes more than 2.5 times as much force to the deformation of bridging veins than translational acceleration. Therefore, thresholds for subdural hematoma injury are usually correlated with rotational acceleration. Using PMHS, critical thresholds for injury have been suggested to be 4500 rad/s^2 for durations 15 to 50 msec (Lowenhielm, 1974) and 10,000 rad/s^2 for durations under 10 msec (Depreitere et al., 2006). Since the FMVSS No. 218 protocol does not allow rotation of the headform, it is difficult to compare published injury thresholds with injury probability results based on linear head acceleration from this study. To accurately measure subdural hematoma injury metrics, impact attenuation tests using a free rotating headform is necessary.

The SIMon contusion results seem to agree with other real world estimates. SIMon predicted a 23% probability of brain contusion if the head experiences 400g. The accident analysis in COST 327 reports that 10% of the total injuries in a motorcycle accident were brain contusions. COST 327 does not supply the relationship of contusion occurrence and peak head acceleration for each of the documented contusion injuries; however, it does provide contusion details for some of the accidents. Analysis of accidents in which the peak head acceleration ranged between 105g to 204g resulted in no brain contusions. An accident with a peak head acceleration of 447g did result in brain contusion. The DDM-acceleration curve computed from SIMon predicts a 27% probability of contusion for an impact with a peak linear acceleration of 447g. Other research using data from small primates scaled to humans predicted a 10% probability of contusion at a linear acceleration of 350g (Vander Vorst et al., 2007). Using the DDM-acceleration curve computed by SIMon, a peak head acceleration of 350g predicts a DDM value of 2.2% corresponding to 15% probability of contusion. Thus the contusion evaluation results are in agreement with the primate study and COST 327 analysis.

HIC has been used as a correlate for head injury for automobile crash tests and is included as a test criterion in the European Helmet Safety test ECE 24.05. HIC does not characterize any one particular type of injury, but is a general measure for head injury. Vander Vorst et al. (2003) demonstrated that HIC correlates poorly with skull strain due to its high sensitivity to target compliance. Although, if the contact area in considered, HIC correlates well with strain. The COST 327 study concluded that HIC correlated better with the Abbreviated Injury Scale (AIS) for the head than peak acceleration or impact speed. Consistent with previous research, COST 327 found a HIC of 1000 predicted an AIS of 2 and a HIC of 1500 predicted a AIS of 3. Analysis for the current study based on the headform peak linear acceleration of 400g suggests a HIC value of 3694. A value this high would result in critical injury or death. Reducing the linear peak acceleration limit down to 290g would correlate to a HIC value of 2400 (the current HIC limit in ECE

22.05). Compared to FMVSS No. 218, it should be noted that the ECE 22.05 headform is different and the test protocol allows for free head rotation.

CONCLUSION

The biofidelity of the injury criteria used by FMVSS No. 218 were examined against biomechanically based injury metrics. Helmet drop tests were conducted using the FMVSS No. 218 protocol to obtain acceleration and pressure data on the headform during impact attenuation tests. The data were used in finite element models to predict injuries for skull fracture, concussion, brain contusion, and subdural hematoma. The predicted damage measures were then correlated against the injury criteria used in FMVSS No. 218.

It was found that peak head acceleration is a good correlate for all injury metrics. Dwell time above 150g and 200g showed poor correlation with the biomechanical based injury metrics. The current linear acceleration limit of 400g predicts a high probability for skull fracture and subdural hematoma. Reducing the limit to 210g would drop the probability of skull fracture to 15% and a reduction to 300g would predict a 50% probability for subdural hematoma. SIMon predicted a small probability for concussion at the linear acceleration limit of 400g. It is thought that SIMon underestimates the CSDM values because of the noninclusion of rotational data in the analysis. The FMVSS No. 218 protocol is not expected to be adequate for predicting acceleration-induced brain injuries. Previous research has shown that angular acceleration is a significant factor in motion related brain injuries. The current test protocol for FMVSS No. 218 restricts movement of the headform to one axis with no rotation allowed. Allowing the headform to rotate during impact should provide a more biofidelic test method. A single value of peak acceleration is insufficient in characterizing a large number of injury modes. Results suggest a need for multi-mode criteria.

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DISCUSSION

PAPER: **Evaluation of Biofidelity of FMVSS 218 Injury Criteria**

PRESENTER: ***Paul Rigby, L-3 Communications/Jaycor***

QUESTION: *Nick White, Wayne State University*

Nice presentation. I just have a question on your FlexiForce system that you've used. I've used, extensively, the I-Scan system from Tech Scan for the pressure mass and found that shearing plays a tremendous role in your results as in that it will—any shear involved will actually lower your peak results and can actually time-shift it: the peaks. And I was just wondering: You mentioned that you had to do a few things to, basically, help prevent shear. I was wondering what that was and I was wondering about your scale factor that you were saying: the 1.1, 1.2 that you were using for some of your results, just because we tried to find a scaling factor to correct for some shearing. And after doing a lot of testing, we found it really wasn't anything linear. If you're going to use the Tech Scan or some kind of pressure sensor like the FlexiForce just to try to eliminate shear as a whole. And I was just wondering what your thoughts on that were.

ANSWER: Okay. Yes. Actually, we were talking about what we did to the FlexiForce to eliminate shear - that was our presentation last year. So if you look at the CD, it goes into pretty good detail of what we did. But in short, what we found out—I did extensive tests on hitting the FlexiForce at different angles and different types of shear. What we found the best was we coated the FlexiForce with a thin layer of Vaseline, then we put two thin layers of Teflon on top of it and pretty much stuck them in place. Without that, we would see lower values, lower voltage values. But by doing that, we were able to eliminate a large percentage of that problem.

For the scaling factor, what we did on that was: That was scaling for the impulse and so we took the impulse calculated from the FlexiForce data and imported calculations from the acceleration data, and we were mainly looking at the point of peak acceleration. And if those didn't match up—this was pretty much on a case-by-case basis—we would adjust it so we would have conservation of impulse.

Q: Thank you.

Q: *Stephan Duma, Virginia Tech*

So basically, you're saying 218 works for skull fracture but not for brain injury. And I am wondering: If you look at the field data, is that what you see—a percentage of head injuries in motorcycles? I mean, I don't get the impression that skull fracture's a problem; it's really brain injury. I'm wondering if the field data, basically, supports what you're showing.

A: Yes. Well, we haven't really looked at the field data.

A: *Erik Takhounts, NHTSA*

I can answer that question.

A: *Paul Rigby, L-3 Communications/Jaycor*

Thank you.

A: *Erik Takhounts, NHTSA*

The only data that we have is very sparse data from FARS. There is no injury data on motorcycles and we unfortunately don't collect CDS-type data on injuries. And whenever we have head injury, there are all sorts of other injuries present. And there is no actual way to attribute death/fatality to head injury or chest injury or impact because everything happens at almost the same time. Well, that's part of the reason we wanted to initiate research to see how actual injuries occur in real life. That's the short answer.

A: *Paul Rigby, L-3 Communications/Jaycor*
And just to add one part to that: With the 218, it works pretty well with the skull fracture. But since we don’t have a rotating head form, the dwell time criteria actually may come in and that might help alleviate some of the brain injury. We found that didn’t really work with the skull fracture. But with a rotating head form, maybe the dwell time does correlate well with brain injury.