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Biofidelity of Motorcycle Helmet Criteria

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ABSTRACT

To protect motorcyclists in accidents, Section 218 of the Federal Motor Vehicle Safety Standards (FMVSS) sets standards for motorcycle helmets based on the magnitude and duration of the acceleration of a vertically guided helmeted headform at impact. Worldwide, there are several other helmet standards. For example, the Economic Commission for Europe standard, ECE 22.4, limits both the Head Injury Criterion (HIC) and peak acceleration computed from a free falling helmeted headform; while the ANSI and Snell Memorial Foundation standards limit only the peak acceleration. Previous work established that the mechanical strain predicted by an anatomically based finite element model is a good biomechanical correlate to bare-headed, impact-induced, linear skull fracture. The purpose of this effort is to determine the biofidelity of motorcycle helmet standards using computed strain as the biomechanical basis. Preliminary analysis indicates that peak acceleration and SFC (the average acceleration over the HIC time interval) are biofidelic criteria to impact skull fracture for motorcycle helmets.

INTRODUCTION

Motorcycle helmets save lives. The National Highway Transportation Safety Administration (NHTSA) of the US Department of Transportation (DOT) estimates that motorcycle helmets saved 500 lives in 1998 and that 307 more could have been saved if all motorcyclists had worn helmets. Unfortunately in 1999, 2284 motorcyclists were killed in accidents and an additional 49,000 were injured. Per vehicle mile, motorcyclists are about 14 times more likely to die in a traffic accident than automobile occupants (NHTSA, 1999).

FMVSS 218: Motorcycle Helmet Standard

Motorcycle helmets are made of a hard protective outer shell lined with an energy absorbing material. The helmet is fastened to the head with a chinstrap. Each motorcycle helmet legally sold in the United States must comply with Section 218 of the Federal Motor Vehicle Safety Standards (FMVSS). FMVSS 218 prescribes a series of tests that a helmet must pass in order to meet DOT approval (US Code of Federal Regulations, 2004). FMVSS 218 establishes requirements for impact attenuation, penetration, and retention. The purpose of the impact attenuation requirement is to protect the head from impact shock. The penetration requirement establishes the ability of the helmet shell to resist penetration of sharp objects. The purpose of the retention requirement is to keep the helmet on the head during an accident. The primary protection against skull fracture and brain injury in nonpenetrating head impacts is offered by the impact attenuation requirement.

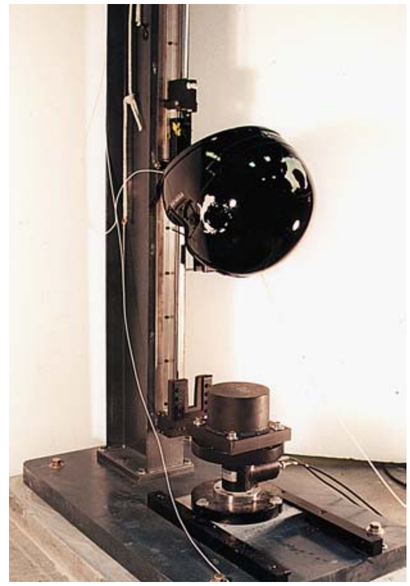


Figure 1: Helmet test drop tower.

In an FMVSS 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, Figure 1. The standard prescribes headforms in several sizes to fit a range of helmets. The vertical acceleration of the headform is measured during the drop of the headform/helmet combination onto a metallic anvil. Tests against both flat and hemispherical anvil are required. For each helmet model and size, a total of thirty-two such drops are conducted.

Table 1 is an example of a test matrix. Four helmets are used, each with a different environmental pre-conditioning: ambient, hot, cold and wet. Each of the four helmets is struck 8 times: two hits at each of four impact locations. The testing laboratory selects the impact locations, under the constraint that each location be separated by at least 60 degrees from the other three. To pass the impact attenuation requirement of FMVSS 218, the criteria,

$A_{max} \leq 400$ g,

$T_{150g} \leq 4$ ms, and

$T_{200g} \leq 2$ ms,

must be satisfied for each of the 32 drops. A_{max} is the peak acceleration. The dwell times, T_{150g} and T_{200g}, are defined as the cumulative time at which the acceleration vs time curve exceeds 150 g and 200 g respectively. A graphical interpretation of each of the above three attenuation criteria is given in Figure 2.

The FMVSS 218 criteria are based on the 1966 optional swing-away test of the American National Standards Institute (ANSI) Standard, Z90.1 (Preamble to CFR, 1973). In 1979 ANSI eliminated the swing-away test in favor of a fixed anvil drop test and omitted the dwell time requirements in favor of a single 300 g peak acceleration criterion (ANSI Supplement Z90.1b) FMVSS 218 retains the original dwell time and peak acceleration requirements of the 1966 ANSI standard, but requires a vertical guided drop onto a fixed anvil. The ANSI test conditions are also different than those of FMVSS 218. In ANSI Z90.1b, the first hit is at 6.9 m/s and the second is at 6.0 m/s for both anvils. FMVSS 218 requires that both hits are at 6.0 m/s against the flat anvil and both are at 5.2 m/s against the hemispherical one. In addition, ANSI Z90.1 requires a single 6.9 m/s hit against a 6.3 mm edge anvil, a requirement not included in FMVSS 218.

NHTSA originally intended that “the performance levels for the impact attenuation requirement be upgraded to that of the Head Injury Criterion (HIC) required by Motor Vehicle Safety Standard No. 208” (Preamble to CFR, 1973).

Table 1. FMVSS 218 Impact Attenuation Test Matrix.

Physical Helmet	**Conditioning**	**Anvil Type[1]**	**Hit Number**	**Impact Location[2]**	**Velocity (m/s)**
1	Ambient	flat	1	Left	6.0
		flat	2	Left	6.0
		flat	1	Rear	6.0
		flat	2	Rear	6.0
		hemisphere	1	Right	5.2
		hemisphere	2	Right	5.2
		hemisphere	1	Front	5.2
		hemisphere	2	Front	5.2
2	Hot	. . .	. . .	. . .	. . .
3	Cold	. . .	. . .	. . .	. . .
4	Wet	. . .	. . .	. . .	. . .

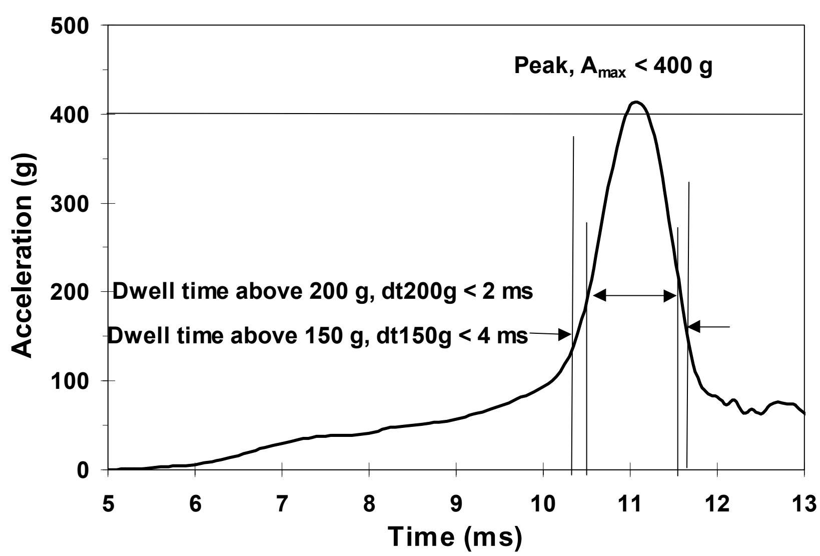


Figure 2: FMVSS 218 impact attenuation criteria.

ECE 22.4: European Motorcycle Helmet Standard

The Economic Commission for Europe uses a different motorcycle helmet standard, ECE 22.4. The ECE 22.4 test device uses a guided free fall but an unrestrained rebound, as opposed to the constrained monorail device prescribed by FMVSS 218.

Table 2 gives the test matrix for a helmet. Impacts are 7.5 m/s onto both flat and curb anvils. Impact locations on the helmet are front, lateral, rear and crown. Each helmet location receives a *single* hit, not two as in FMVSS 218. There are 5 different environmental preconditions. In all, 10 helmets are used (two for each environmental condition), and each helmet undergoes 4 impacts. ECE 22.4 uses the metallic ISO headform, which is different from that prescribed by FMVSS 218. The headform is fitted with a triaxial accelerometer array at its center of gravity and the resultant acceleration is computed from the three waveforms. To pass ECE 22.4, the conditions,

$A_{max} \leq 275$ g and
HIC ≤ 2400,

must be satisfied for each drop (United Nations, 1994).

Table 2. ECE 22.4 Test Matrix.

Physical Helmet	Conditioning	Anvil Type	Hit Number	Impact Location[2]	Velocity (m/s)
1	Solvent & Ambient	Flat	1	Front	7.5
				Left or right	
				Rear	
				Crown	
2		Curb		Front	
				Left or right	
				Rear	
				Crown	
3 & 4	Solvent & hot	…		…	
5 & 6	Solvent & cold	…		…	
7 & 8	Solvent & UV	…		…	
9 & 10	Water spray	…		…	

The Snell Standard

The Snell Memorial Foundation promulgates standards for many different types of protective headgear. Like the ECE standard, Snell prescribes the ISO headforms. Like FMVSS 218, the impact attenuation part of the Snell standard requires a guided drop onto both flat and hemispherical anvils. In addition, the Snell standard requires an impact against an edge anvil. Two hits are performed against the flat and hemispherical anvils and one against the edge one. Snell prescribes an impact energy for each impact, but ideally, this is equivalent to impact velocities of 7.7 m/s on the first hit and 6.6 m/s on the second. In addition, a single drop at each of four impact sites is done at 7.6 m/s against an edge anvil. The drop weight for each test is 5 kg, independent of the helmet size. A helmet passes if the peak acceleration does not exceed 290 g (Snell Memorial Foundation, 2005).

The ANSI Z90.1 Motorcycle Helmet Standard.

The ANSI Z90.1 motorcycle helmet standard is another voluntary standard. The impact attenuation part of ANSI Z90.1 requires two hits at each of four impact sites, two onto a flat anvil and two onto a hemispherical one. The first hit is at 6.9 m/s and the second at 6.0 m/s. In addition each helmet must receive a single hit against an edge anvil at 6.9 m/s. Impact sites must be separated by at least 1/6 of the maximum circumference of the helmet. Four samples are tested using ambient, cold, hot and wet environmental conditions. The headform can be either DOT or ISO. As in the Snell standard, a helmet fails if Amax > 300 g for any impact.

FMVSS 208: Automobile Crash Test Standard

FMVSS 208, the standard to protect passengers in a car crash requires a different head injury criterion than the motorcycle helmet criterion. FMVSS 208 uses the resultant acceleration at the center of gravity of the head of a Hybrid III anthropomorphic test dummy. To pass the FMVSS 208 head injury criterion "the expression:

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

shall not exceed 1,000 where a is the resultant acceleration expressed as a multiple of the acceleration of gravity, and t_1 and t_2 are any two points in time during the crash of the vehicle which are separated by not more than a 36 millisecond time interval" (US CFR, 2004). For future reference, we denote the time interval which maximizes HIC and the change in velocity over that interval by:

$\Delta t_{HIC} = (t_2 - t_1)$, and

$\Delta V_{HIC} = \int_{t_1}^{t_2} a\,dt$.

Skull Fracture Correlate

To predict linear skull fracture, Vander Vorst et al. (2003, 2004) developed and validated an anatomically based finite element model of the human head derived from a human CT medical image (Figure 3). The model is composed of 24,000 elements and resolves the outer and inner tables, diploe, brain, scalp, and face. The mass of the model is 4.54 kg. The skull components are modeled using fully integrated thick shells and the brain, scalp, and face are modeled with fully integrated bricks. Since this model was based on CT imaging of a postmortem human subject, the skull shape and thickness are anatomically correct. The thickness of the compact skull tables was set to be 1 mm uniformly, since they are 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 (National Library of Medicine, 2000). The properties of the biological materials were taken from the open literature. The model was validated against postmortem human specimen tests to correlate skull strain with skull fracture (Figure 4). Using skull fracture data from tests with postmortem human test subjects, head acceleration data from the Hybrid III anatomical test device, and skull strain from finite element simulations, all under the same conditions, they derived correlates to the probability of skull fracture versus strain and to several kinematic risk factors. SFC, the average acceleration over the HIC time interval,

$$\mathrm{SFC} = \Delta V_{HIC} / \Delta t_{HIC} ,$$

was the best kinematic correlate to linear skull fracture. The 15% probability of skull fracture occurs at peak maximum principal strain of 0.20% and 50% probability of fracture is at a strain of 0.29%.

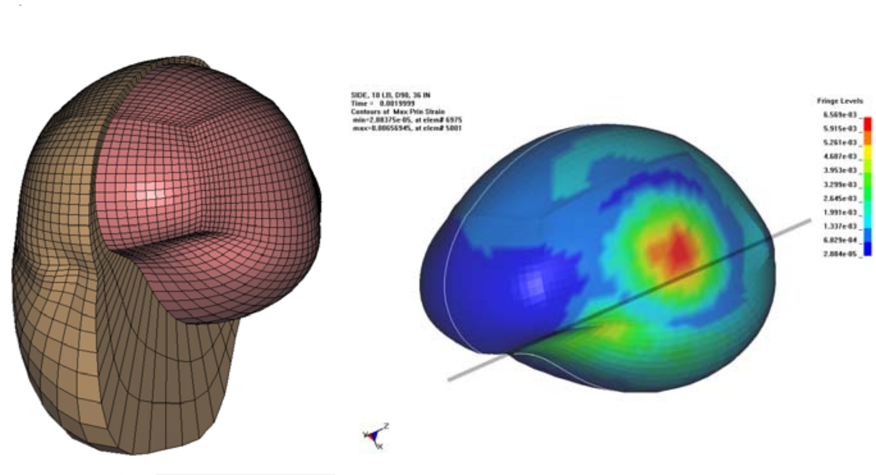


Figure 3: Anatomically based head model for skull fracture prediction.

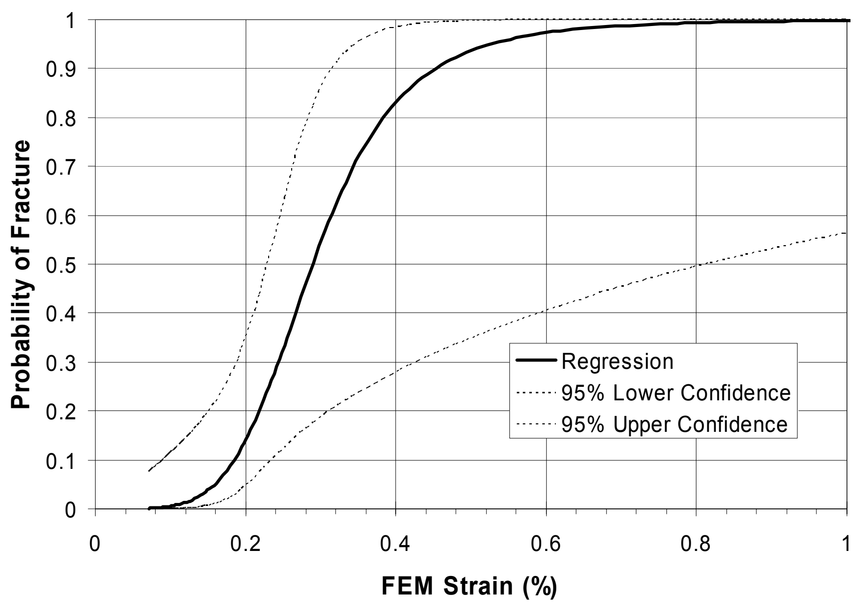


Figure 4: Probability of skull fracture vs. computed strain.

Issues

The biomechanical basis of the various helmet criteria is not known. To predict the efficacy of a particular helmet using the anatomical model purely from finite element calculations 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 peak strain, then the probability of skull fracture could be predicted for the specific helmet.

Future refinement and harmonization of criteria, test procedures and headforms should be based on biomechanical understanding of injury mechanisms.

Objective and Approach

The objective of this research is to evaluate the biofidelity of helmet criteria. The approach used to meet this objective is:

1. Predict helmeted skull fracture with finite element model (FEM) simulations, and
2. Develop instrumentation to measure headform surface pressure during drop tests.

This paper reports results from preliminary assessment using flat target impact simulations, and instrumentation development.

METHODS

To determine the biofidelity of the motorcycle helmet criteria, each of the kinematic risk factors, peak acceleration, SFC, 150 g dwell time, 200 g dwell time and HIC were correlated with skull strain as computed from the anatomical finite element model of Vander Vorst et al. (2004). CG head acceleration data were from the finite element simulations, after verifying that the acceleration from the simulation matches that measured from a drop test. Finite element simulations were done using LS-Dyna (LSTC, 2003). Drop tests of a size medium DOT headform from 1.83 meters (72 inches) onto 50 mm (2 inch) thick, 2 pcf encapsulated polystyrene (EPS, trade name Styrofoam) blocks were conducted using a United States Testing Service guided monorail drop tower and the time history of the CG headform acceleration as well as the force from a load cell were measured. EPS was modeled with the LS-Dyna crushable foam material using stress-strain curve from Liu et al. (2003). To validate the EPS material properties, the CG acceleration of a rigid head model impacting the EPS block was computed under the same conditions. This rigid head model was constructed from the anatomical biological model by setting each of the tissue properties to rigid. Calculations using the anatomical model with biological tissue properties were performed under the same conditions to determine the difference in acceleration response between the rigid and biological head models. Simulations were then performed using the anatomical model, varying the foam elasticity and drop height to generate head acceleration data covering a range from 100 to over 400 g. The risk factors, peak acceleration, SFC, 150 g dwell time, 200 g dwell time and HIC, were calculated for each test and correlated with peak maximum principal strain using Microsoft Excel.

Instrumentation for measuring the interfacial pressure between the helmet liner and headform was prototyped by gluing an array of 24 thin film pressure sensors (Tekscan Flexiforce gauges) to a size medium DOT headform (Figure 5). The sensors were calibrated on the headform using a Kistler Model 9724A5000 force hammer and the headform was dropped from 1.83 meters (72 inches) onto a 50 mm (2 inch) thick, 2 pcf encapsulated polystyrene (EPS) block.

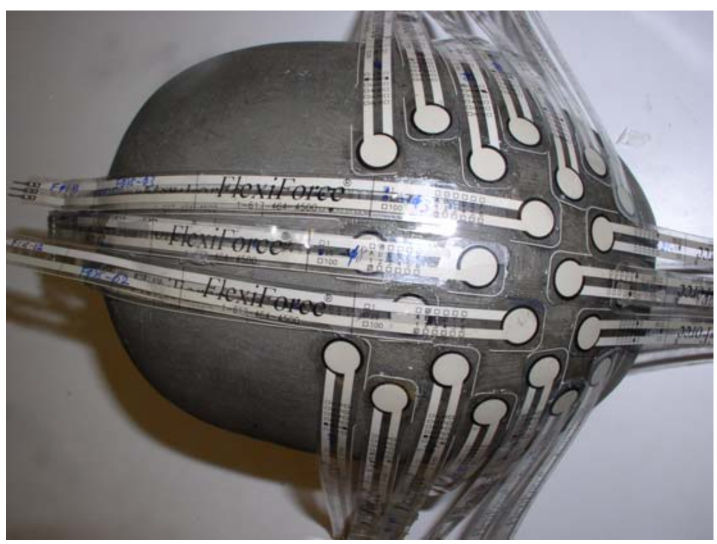


Figure 5: Thin film sensors attached to DOT size medium headform.

RESULTS

The CG acceleration time histories from the headform (Figure 6), the rigid head anatomical model, and the biological anatomical model (Figure 3), impacting the EPS are similar (Figure 7). This validates the EPS model against drop test data and demonstrates that the biological model acceleration response is similar to that of the rigid headform.

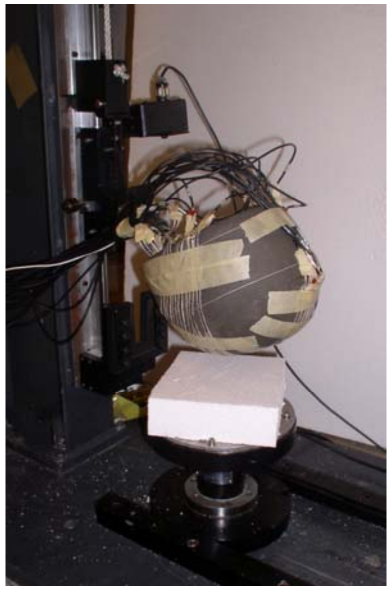


Figure 6: Test setup of instrumented headform impacting EPS.

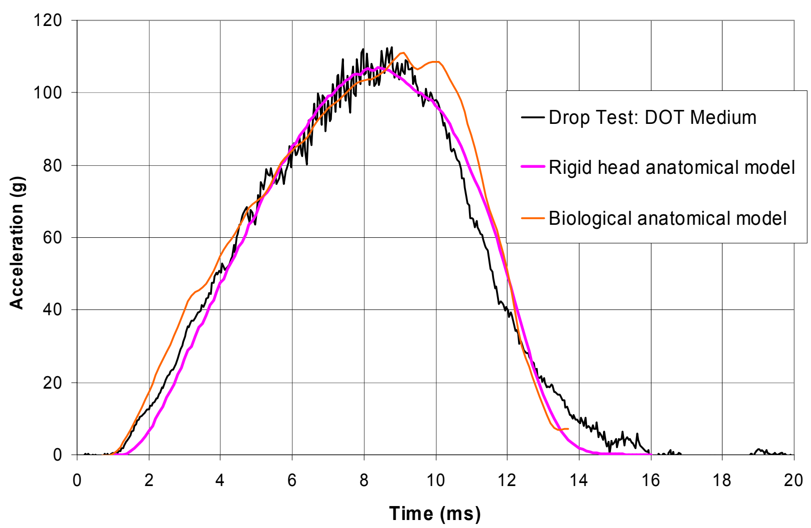


Figure 7: Comparison of CG head acceleration from head form drop test and anatomical finite element models.

Peak acceleration and SFC are both good correlates to strain with R^2 of 0.86 and 0.87, respectively (Figure 8); while the 150 g and 200 g dwell times as well as HIC are poor correlates with R^2 of 0.49, 0.18 and 0.52, respectively (Figures 9).

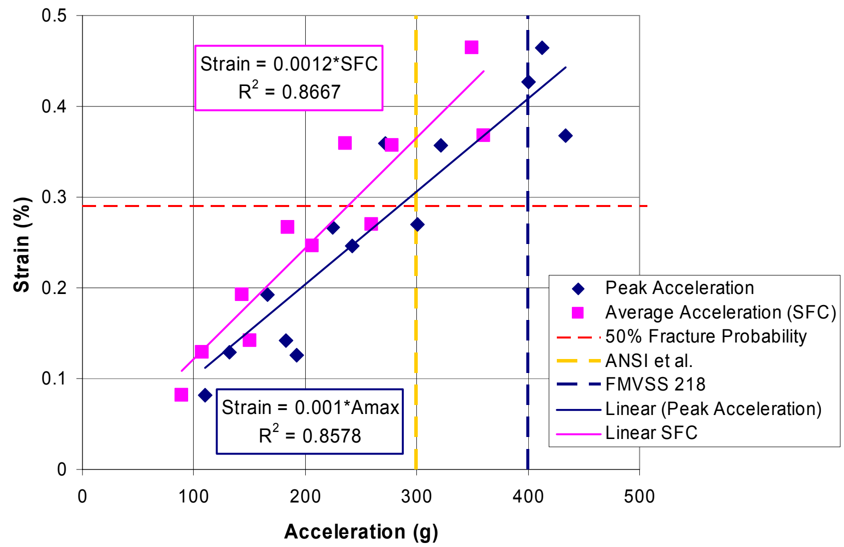
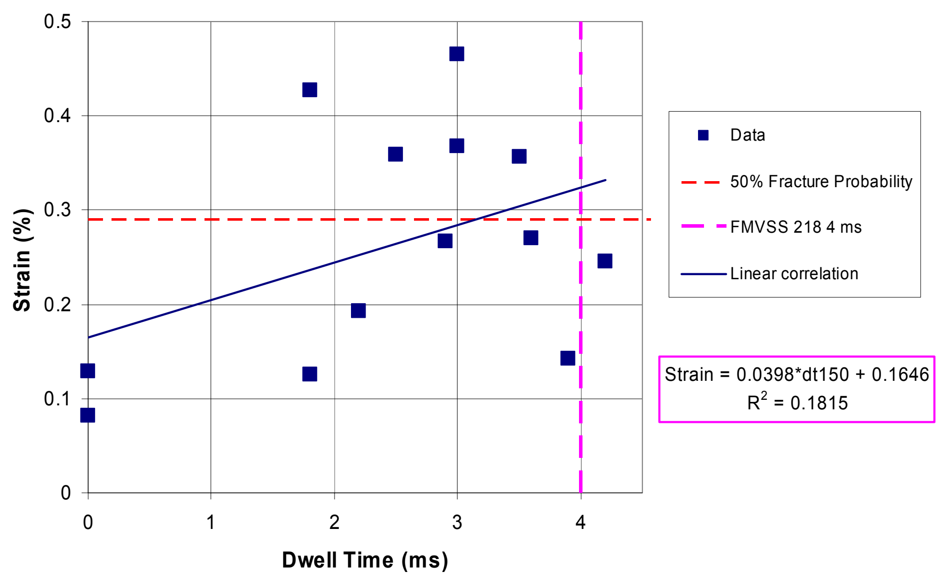
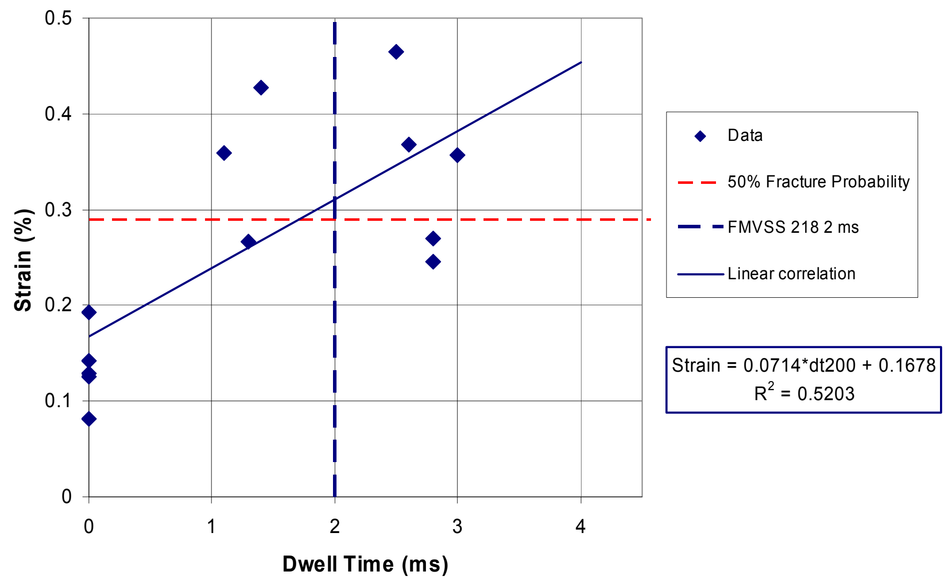


Figure 8: Computed strain from anatomical model vs. CG head acceleration and SFC.

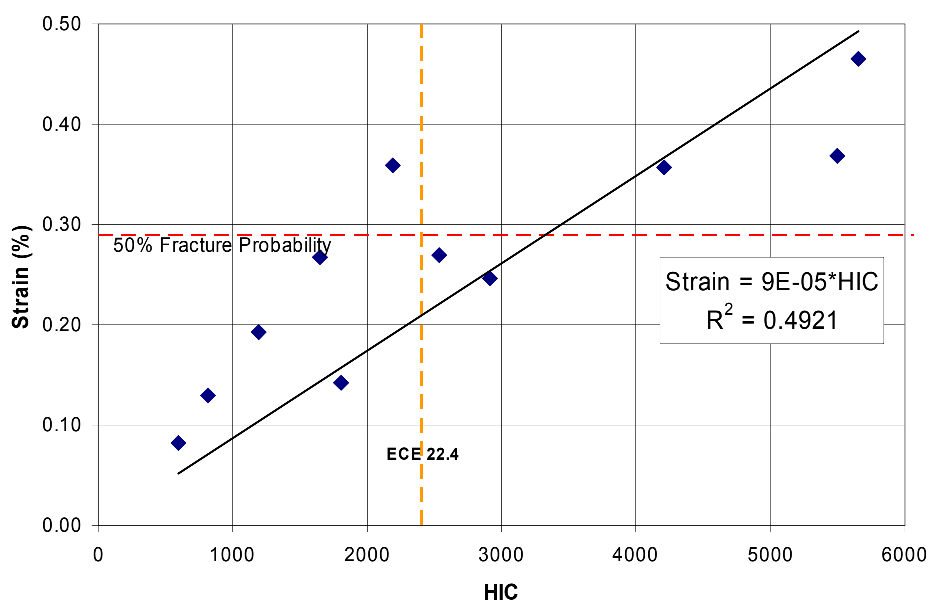


(a) 4 ms dwell time

Figure 9: Computed strain from anatomical model vs. dwell times and HIC.



(b) 2 ms dwell time



(c) HIC

Figure 9: Computed strain from anatomical model vs. dwell times and HIC (continued).

Integrated force from the 24 thin film sensors agrees with load cell and accelerometer measurements (Figure 10), validating the consistency of the measured total pressure distribution. Pressure contours show peak values at the center with a local maximum at the edges (Figure 11).

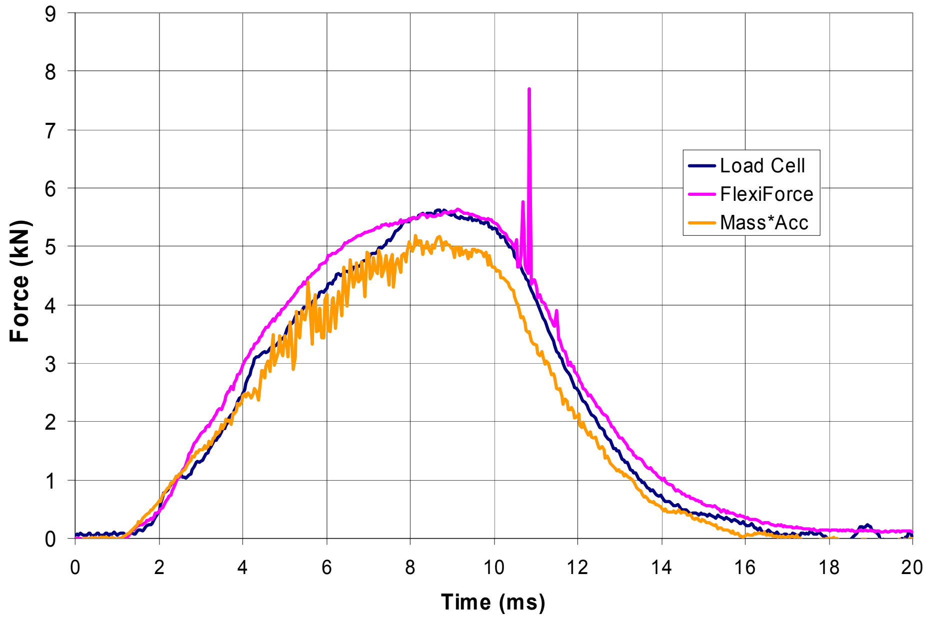


Figure 10: Comparison of force vs. time from headform impact as measured by load cell, CG head acceleration and integration of thin film pressure sensor data.

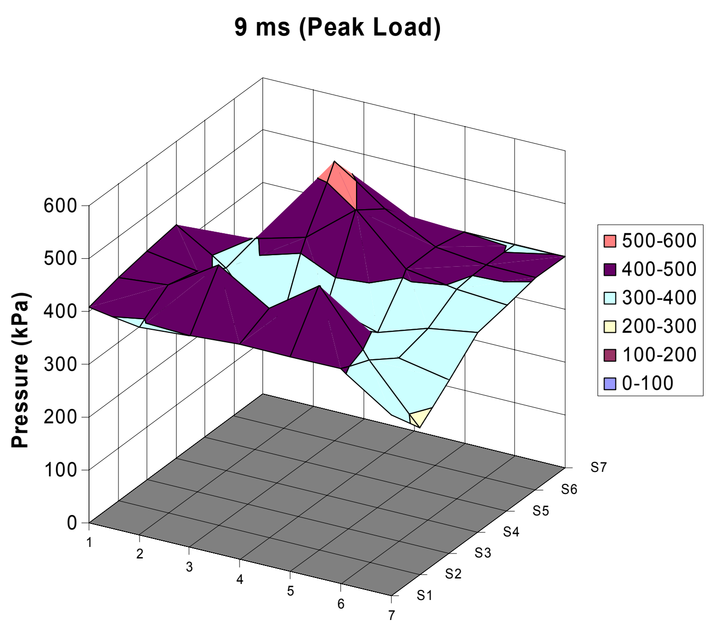


Figure 11: Spatial distribution of measured pressure data between EPS block and headform.

DISCUSSION

Preliminary investigation of the biofidelity of motorcycle helmet criteria was completed; and instrumentation was developed to measure the surface pressure distribution between a helmet and headform during drop tests. Using finite element simulations and headform drop tests, the peak acceleration requirement of the FMVSS, ECE, ANSI and Snell motorcycle helmet injury criteria was shown to be biofidelic for head impacts onto encapsulated polystyrene, the material most often used as the attenuation liner in motorcycle helmets.

The limitation of this work is that the impacts studied were not helmeted, but were onto flat sheets of helmet liner material. In the next phase of this work, the instrumentation will be expanded from 24 to 48 thin film sensors to extend the coverage of the pressure sensors around the head; and the pressure distribution time histories will be acquired for helmeted headforms using several qualities of helmets. The measured pressure for each helmet and impact location on the helmet will be applied as a boundary condition to the finite element head model in order to predict the skull strain and hence the probability of injury for the impact. These results will be used to fully establish the biofidelity of helmet standards.

Based on computed strain, the 50% probability of skull fracture occurs at a peak acceleration slightly below 300 g, which is similar to the ANSI, ECE and Snell criteria. The peak acceleration requirement in the FMVSS criterion has a higher limit of 400g. Independently, Thom et al. (1997) recommended lowering the peak acceleration requirement to 300g. These preliminary results also suggest that the dwell time requirement of FMVSS 218 and the HIC requirement of ECE 22.4 are not biofidelic. However, since both the FMVSS and ECE criterion use peak acceleration as a component, the other requirements could be considered as ancillary and may serve to make the standard more conservative.

CONCLUSIONS

Preliminary analysis shows that peak acceleration and SFC (the average acceleration over the HIC time interval) are biofidelic criteria for helmeted impact skull fracture. The 50% probability of fracture occurs when the peak acceleration is 290 g; a value which is similar to that used in the current ECE, ISO and Snell standards, but lower than that in the FMVSS standard.

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DISCUSSION

PAPER: **Evaluation of Biofidelity of Motorcycle Helmet Criteria**

PRESENTER: ***Phil Chan, Titan Corporation***

QUESTION: *Mike Kleinberger, Johns Hopkins University*
Since the main reason for using helmets is not necessarily to prevent skull fracture but to prevent brain injury, is there any reason why you're not looking at strains inside the brain material rather than just on the skull?

ANSWER: That would be additional injury components that we could look into at a later time. Right now, we're focusing on skull fracture, which can be considered as a good line of defense, but your question is correct. We're not considering the other components yet at this time, but we're focusing on skull fracture for now.

Q: *Guy Nusholtz, Daimler Chrysler*
How are you calculating the strains? Is it a result of the contact forces?

A: The strain on the skull? It's calculated at the peak strain, which is: Run the calculation and search for the peak strain.

Q: But how is—Is it the result of contact forces from something striking the head?

A: Yeah. It would be the skull impacting on the Styrofoam.

Q: It would be local.

A: Yeah. It would be local at a point.

Q: So the strains are a local phenomenon.

A: Yeah. Correct.

Q: With the confidence intervals that you have on your fracture, your risk curve—

A: Yeah. Right, right.

Q: Have you taken that into account in trying to estimate whether one is a better predictor than the other? The only thing you've got there is the correlation coefficients.

A: For the models?

Q: For the models, but there's a lot of –a large error in the risk. So statistically determining whether one is better than the other should take into the account those error terms.

A: Yeah. Well, we went against the skull fracture or the model prediction of skull fracture and we would consider the confidence interval and right now, which tends to be greater at the larger end of the strain. So if we stay on the lower end of the strain level, I think that confidence interval is pretty good, which is where we want to focus on. And, you're correct that we should run more tests and then consider statistical variations between repeated drops when we run the actual drop tests with a helmet on. That ought to be factored in. Yeah.

Q: One of the interesting aspects of your results is that you're getting a better correlation with acceleration.

A: Acceleration.

Q: Than you are—significantly better—than you are with HIC. Do you have a rationale as to why that would be?

A: Right now, we just let the results stand. We recognize that we are just dropping on flat target and we want to complete our more—using actual helmet drop tests to complete the assessment before we, you know, make a more definitive assessment of that.

Q: One of the things you might want to consider as you're doing your analysis is whether the correlation's a result of the testing process?

A: Um hm.

Q: Because as you drop—With a well-distributed load, as you have more acceleration, you'd expect to see more strain.

A: Yeah. Correct. Yeah. Good suggestion. Thank you!