The effect of crash pulse shape on AIS1 neck injuries in frontal impacts

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ABSTRACT

Crash data from real-life frontal car collisions, where the crash pulses have been measured with crash pulse recorders and where the influence of pulse shape on the risk of both short- and long-term disability from AIS1 neck injuries, have been studied. The risk of long-term consequences was especially influenced by the shape of the crash pulse. To understand how the shape of the crash pulse affected occupant motion, a series of computer simulations of frontal impacts were conducted where the information from the crash pulse recorder data has been used in the simulations. Several dummy response parameters, such as neck loads and accelerations, were compared with the injuries in 143 real-life collisions using the recorded crash pulses and the injury records. The results showed that for a specific change of velocity the pulse shape could significantly influence some of the dummy response parameters, such as angular head acceleration and neck bending moments. It was also found that there was a correlation between these dummy response parameters and the long-term consequences. The results may help to explain the injury mechanism of the AIS1 neck injury in frontal impacts and may have implications on the design of the seat belt system.

THERE IS ONE INJURY that has increased substantially, both in terms of risk and in number (Morris and Thomas 1996, Krafft 1998). In Sweden between 1990 and 1995, it has been shown that approximately 60% of the injuries causing a disability were AIS1 neck injuries (Krafft 1998). Nygren (1984) found that 10% of the occupants in rear impacts and 5% of the occupants in frontal impacts that reported initial whiplash symptoms suffered disability at least one year after the collision. In most research on AIS1 neck injuries, only the initial symptoms have been studied. It is, however, important to separate the analyses for initial and residual symptoms since there may be different injury types and different injury mechanisms (Krafft 1998, Kullgren 1998).

Injuries to the neck are often regarded as a problem in rear-end impacts. However, about 1/3 of the AIS1 neck injuries occur in frontal impacts (Galasko et al. 1993, Krafft 1998). The AIS1 neck injury mechanisms in different collision modes are still not known. Different hypotheses exist concerning injury mechanisms in rear-end impacts, attributing the injury to the flexion (v Koch et al. 1995) or the extension (McConnell et al. 1995) motions of the neck. In frontal collisions, Larder et al. (1985) found that no head contact with the interior of the vehicle compartment had been noticed, thus the forward flexion of the neck was assumed to be the injury-causing motion. Walz and Muser (1995) and Ewing et al. (1975) described the motion of the head relative the neck in a frontal collision with no head contact. For a restrained occupant, the initial phase of a collision, results in a purely translational head motion producing a S-shape of the cervical spine followed by flexion of the neck. Walz and Muser (1995) suggested that neck injury may occur in this predominantly inertial loading of the head and neck.

Several studies show that the neck injury risk is associated with seat-belt use (Larder et al. 1995; Otremski et al. 1989). Galasko et al. (1993) found an increase in neck injuries from 8% to 21% associated with an increase in belt wearing rates in the UK. In the studies mentioned above, it is unknown if the increase occurred for all impact directions.

Regarding impact severity, studies have shown a correlation between change of velocity and initial neck injury symptoms in rear impacts (v Koch et al. 1995, Ryan et al. 1994). However, Krafft (1998) found that a higher change of velocity of the struck cars did not increase the risk of long-term consequences compared to short-term consequences. The results indicated that the acceleration levels seemed to influence the risk of long-term consequences to a higher extent. Olsson et al. (1990) also found similar results. Furthermore, results from real-life rear-end impacts where the crash pulse has been measured with a crash pulse recorder indicate that acceleration levels seem to better explain the neck injury risk than change of velocity (Krafft et al. 1998, Krafft 1998).

In frontal impacts, Kullgren et al. (1999) has found that the shape of the crash pulse particularly influences the risk of long-term consequences to the neck. The study by Kullgren et al. was based on results from real-life impacts where the crash pulse was measured with a crash pulse recorder. Crash pulse characteristics were related to the risk of both short- and long-term consequences to the neck. The crash pulses were divided in intervals of 33 ms: $\Delta v1$ 1-33 ms, $\Delta v2$ 34-66 ms, $\Delta v3$ 67-99 ms, etc. It was found that a large change of velocity in the second part of the pulse followed by a smaller velocity change in the third part, resulted in a high risk of long-term consequences. In Figure 1 it can be seen that the mean crash pulse for the impacts with occupants that sustained long-term consequences differed significantly from the impacts with occupants that sustained short-term consequences and the occupants that did not report an AIS1 neck injury (Kullgren et al. 1999). It is also important to notice that the mean pulse for the impacts with occupants that sustained short-term consequences was almost the same as for the occupants that did not report an AIS1 neck injury.

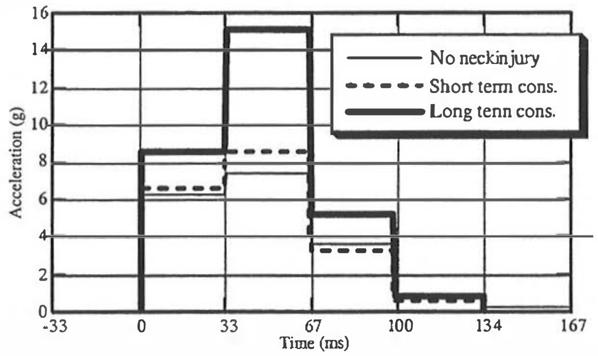


Figure 1. Mean crash pulses for the occupants with no Reported neck injury and for those with short- and long-term consequences (from Kullgren et al. 1999).

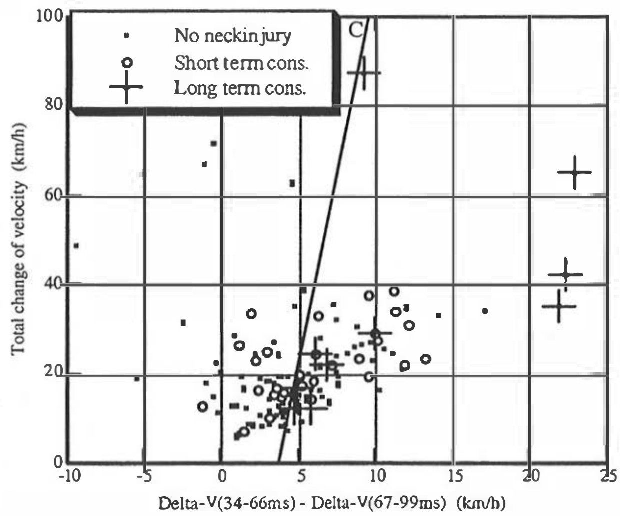


Figure 2. Total change of velocity versus the difference in change of velocity between the second and third part of the pulse (from Kullgren et al. 1999).

The correlation between different collision parameters can be studied to determine their relation to injury risk. Kullgren et al. (1999) compared total change of velocity (ΔV) and difference in change of velocity between the second and third parts of the pulse ($\Delta v2$-$\Delta v3$) as predictors of injury. They found that the variation of velocity change between the pulse segments (pulse shape) influenced the risk of long-term consequences to a higher extent than the total change of velocity (pulse magnitude). An example of their study is shown in Figure 2.

To be able to understand how the pulse shape influences the trajectory of the occupant's head and torso and the resulting influence on the neck injury risk, simulations based on real-life impacts could be helpful. There is a shortage of information regarding the relation between laboratory dummy readings from crash tests or computer simulations and injuries in real-life impacts.

The aim of this study was to compare the results of computer simulations of occupant motions with reports of AIS1 neck injuries. Using recorded crash pulse information in the simulations, the occupant response in a specific collision could be correlated to the reported injury outcome. This may provide a better connection between occupant response parameters and the risk of long-term consequences from AIS1 neck injuries.

MATERIAL/METHODS

THE REAL-LIFE IMPACTS

An accident data set containing the vehicle acceleration data recorded by a crash pulse recorder (CPR) was used in the study. The same data set, with the exception of one additional long-term disability (see definition below), was used in the study by Kullgren et al. (1999). The crash pulses were measured along the longitudinal axis of the vehicle and were filtered at approximately 60 Hz. The CPR and the analysis of the crash records have previously been described by Aldman et al. (1991) and Kullgren (1998).

The impact sample included 187 restrained front seat occupants in 143 frontal collisions with an overlap exceeding 25%. Belt use was verified from inspection of the seat-belt system. The front seat passengers consisted of 133 males and 54 females. The injuries were classified according to the 1985 revision of the Abbreviated Injury Scale (AAAM 1985). Only AIS1 neck injuries were considered in this study. The neck injuries were divided in short- and long-term consequences. The injury reports were collected directly after the accident, questionnaires were sent to the occupants. A follow-up after at least 6 months was also done by telephone interview. When the occupants recovered within 6 months, the injuries were classified as short-term consequences. To be classified as long term consequences, the occupant had to continuously have symptoms (at least every second week) for longer than 6 months. From the 187 total occupants, 32 occupants were classified as short-term and 11 as long-term consequences. In Sweden it may take several years after an accident before a medical disability can be verified by a doctor. In 3 cases a medical disability was verified by a doctor and in 8 cases it was assessed by telephone interview.

SIMULATION APPROACH

A study of crash pulse parameters was conducted within the simulation environment MADYMO from TNO. The objective was to compare the occupant response resulting from different crash pulses. To facilitate the comparison, simulations were conducted with identical model parameters except for the crash pulse definition. The rigid body model for a Hybrid III dummy was selected to represent the occupant. A simplified seat and seatbelt was included in the model to represent the vehicle interior. The crash loading was applied to the model as a time dependent acceleration function which is described below.

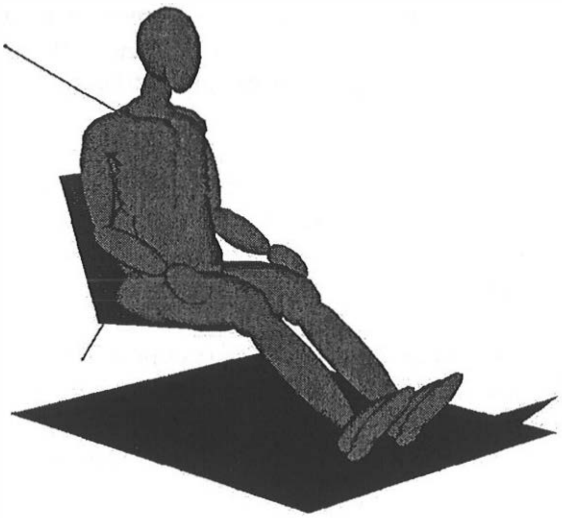


Figure 3. Computer representation of occupant

The starting point of the model was a reference model provided with the MADYMO software. A model of a sled test with a 50^{th} percentile male Hybrid III dummy is provided with results of an actual test, identifying the model's accuracy. This reference model was modified so that seatbelts were more representative of automotive applications. A review of other sled tests was used to determine an appropriate belt slack to be included in the model. Because the objective was to compare the outcome from different crash pulses, specific features of restraint systems, retractors, and occupant stature were not modelled. An illustration of the model is shown in Figure 3.

Two separate simulation phases were conducted. A parametric study of crash pulse shapes and magnitudes was conducted using the idealised square wave pulse components proposed by Kullgren et al. (1998) shown in Figure 1. The purpose of this was to observe the relative influence of pulse shape and pulse magnitude in terms of occupant kinematics. Another objective was to try to identify occupant response parameters that could be used as injury predictors. Only a 100 ms pulse duration was used in this simulation series. Three pulse shapes, representing Kullgren et al's (1998) description of No-Injury, Short-term only, and Long-term consequences, were compared to various combinations of $\Delta v1$, $\Delta v2$, and $\Delta v3$ magnitudes. In addition, pure square wave pulses for a range of ΔVs were included as references shapes.

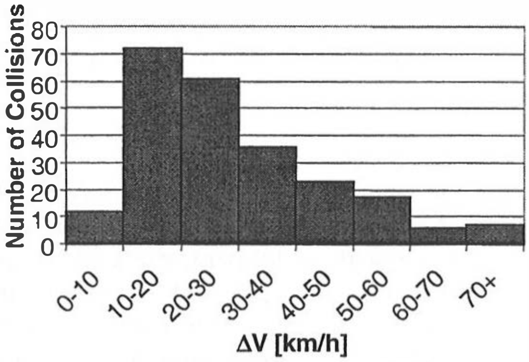


Figure 4. Speed Distribution from CPR

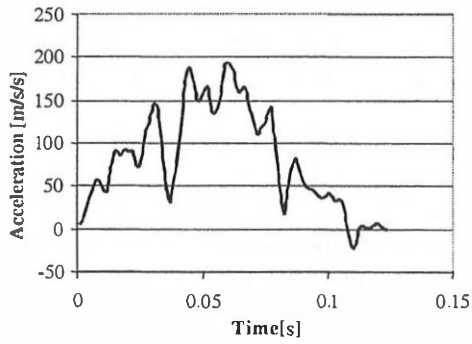


Figure 5. Example of Crash Pulse

A subsequent simulation study investigated the occupant responses for 143 crash pulse shapes, as recorded by the Folksam Crash Pulse Recorder. These pulses were not limited to the 100 ms duration and represented the actual recording from the CPR. The distribution of simulations with ΔV is shown in Figure 4. An example of one of these pulses, as input to the computer model, is shown in Figure 5.

The computer simulations were configured to provide various kinematic and loading data. Significant parameters that were monitored were: kinematics of the head center of gravity; kinematics and joint forces/moments for upper and lower neck locations representative of C1 and T1 vertebrae; motion of the sternum relative to the vehicle; and tension loads in the seatbelt. Various combinations of absolute and relative kinematics were plotted against pulse characteristics to detect occupant responses that may indicate injury. These output data from the simulations were cross-referenced to the injury data recorded by Folksam for the crash recorder pulse of interest.

SIMULATION RESULTS

FIRST SIMULATION STUDY

The first simulation series was designed to investigate the hypothetical performance of the idealised pulses shown in Figure 1. This exercise was aimed at discriminating factors that would highlight the differences between injury and non-injury conditions due to the magnitude or the shape of the crash pulse. In terms of the magnitude (ΔV) of the pulse, a summary of maximum head accelerations can be seen in Figure 6. From the results in Figure 6, we can observe the influence of crash pulse on occupant response. For each discrete ΔV simulated, a range in occupant responses is possible. This variation in output can only be attributed to the different crash shape for each ΔV. As the ΔV increases, the possible variation in occupant response also increases.

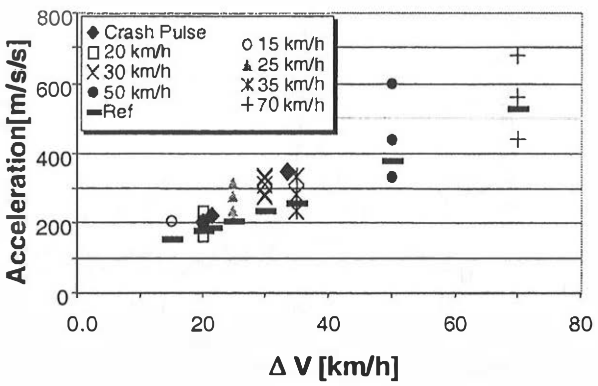


Figure 6. Peak Head Acceleration as a function of **ΔV.**

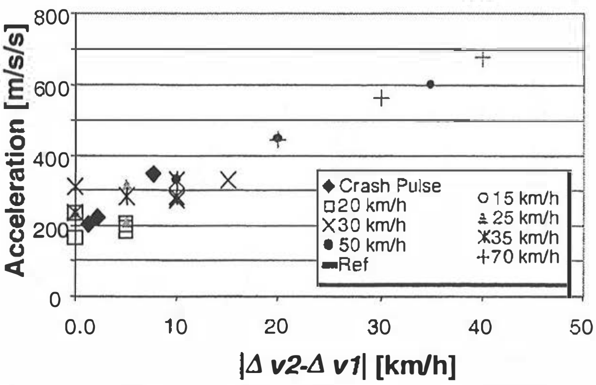
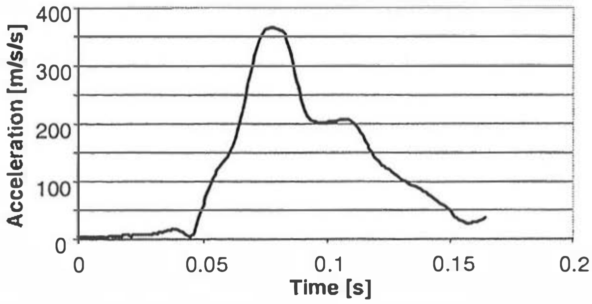


Figure 7. Peak Head Acceleration as a function of $\Delta v2$ and $\Delta v1$.

When a pulse shape parameter, such as the difference between $\Delta v2$ and $\Delta v1$, is used to plot the occupant results, a plot like Figure 7 is produced. This plot exhibits a more linear result for the occupant response and thus a better correlation between vehicle response and occupant response. This type of plot has two significant features: 1) there is the potential to define a more discriminating injury threshold and/or injury criteria. 2) a better correlation between the parameters can provide a better diagnostic tool for research and epidemiological investigations.

SECOND SIMULATION STUDY

The second simulation study involved the use of actual crash pulses recorded in injury, and non-injury producing collisions. This should produce a more realistic description of actual crash response. The simulations resulted in occupant response curves similar to those seen in Figure 8, resulting from the pulse in Figure 5. These plots are typical of the simulated occupant responses.



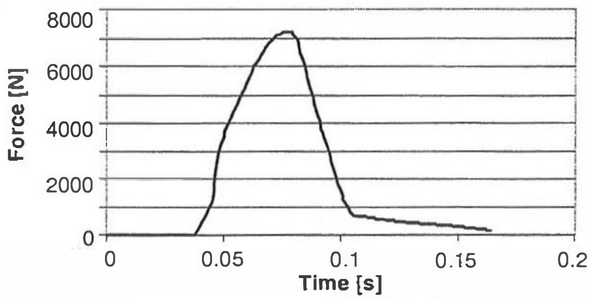
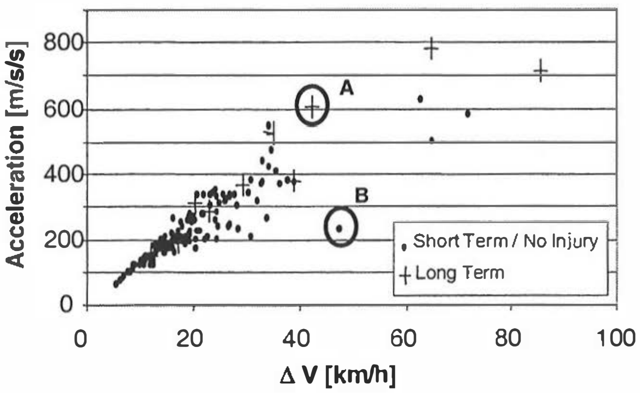


Figure 8 a). Simulated Head Accelerations b). Simulated Shoulder Belt Loads

As in the first simulation series, the influence of the pulse shape on occupant kinematics was observable. In Figure 9, the resultant head acceleration is plotted as a function of ΔV. A significant variation in occupant output was noticeable. Of particular interest were the two cases, A and B, highlighted on the graph. Both cases represent collisions that had ΔVs around 45 km/h, but entirely different occupant responses arose. In one case, A, a long-term consequence was identified, albeit at a lower ΔV. The differences between these two pulses are obvious when plotted together in Figure 10. The injury producing pulse (A) can be seen to have a shorter duration with a more pronounced peak acceleration. The pulse content after 66 ms is very small, and supports Kullgren et al.'s (1999) hypothesis that a large discrepancy between $\Delta v2$ and $\Delta v3$ has a higher injury risk.



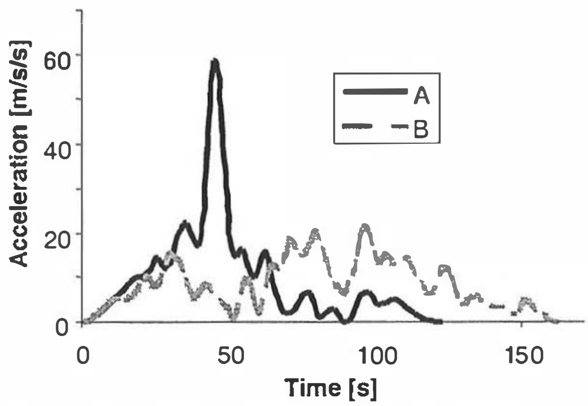


Figure 9. Head Accelerations as a Function of ΔV Severity

Figure 10. Crash Pulses of Similar Collision

A further comparison of these two simulations was made by comparing the animations of occupant response. Figure 11 shows occupant positions near the ends of the 33ms intervals used in analysing the pulses (such as seen in Figure 1). The overlaid images from the two collisions illustrate the different occupant motions due to different pulse shapes. After 32 ms, the two simulations produce identical occupant motions. However, after 66 ms, Case A has resulted in a larger occupant translation and head rotation. The last image at 90ms represents the most significant difference between the two cases. The injury case is reflected by a more pronounced head rotation and forward displacement of the occupant. The non-injury case exhibited lower values for all occupant parameters recorded.

The results of this simulation study were also summarised into plots of maximum occupant response parameters. The plots summarising the second simulation series are stratified to indicate the conditions producing long-term consequence, short-term, or no injury. This should allow particular conditions that could be attributed to an injury, in effect, an injury criteria to be identified.

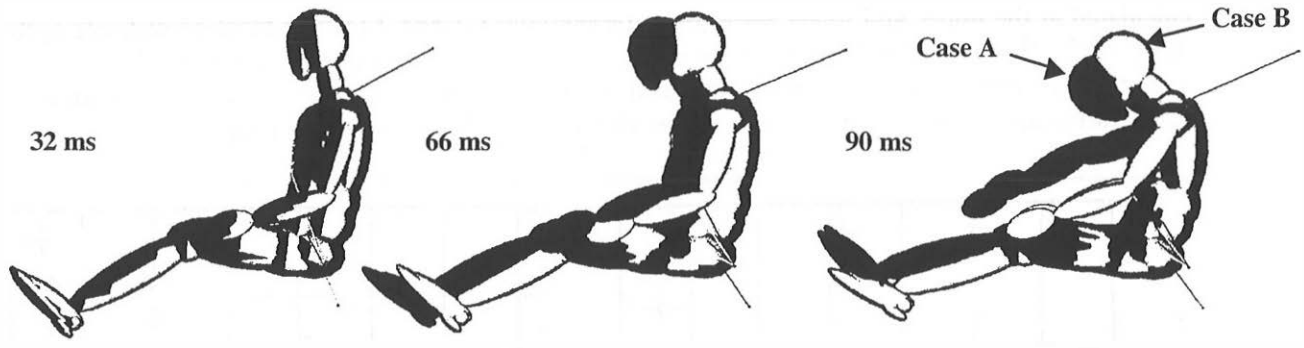


Figure 11. Simulated Occupant Response During 42 km/h (Injury) and 47 km/h (No Injury) Collisions.

From Figure 12 and 13, the difference in change of velocity between part 2 and 3, $\Delta v2$-$\Delta v3$, is seen to be a better predictor of injury than the change of velocity in part 2, $\Delta v2$. In Figure 13, all long-term consequences can be seen lying to the right of a threshold of 5 km/h. From Figures 12 and 13 it can be seen that the simulated head peak acceleration correlates with the risk of long-term consequences. Most of the crosses, representing the long-term consequences, are in the upper part of the plot. Figure 12 also shows that the head peak accelerations are well correlated with the change of velocity in part 2 ($\Delta v2$) of the pulse. Since the difference in change of velocity between part 2 and 3 ($\Delta v2$-$\Delta v3$) seems to distinguish the risk of long-term consequences, it has been used as impact severity parameter in the following plots.

From the data in Figure 12, the peak head accelerations for 8 of 11 occupants with long-term consequences were above the average peak head acceleration calculated for the entire long-term injury subset sample. This was the case for all the following plots for different occupant responses.

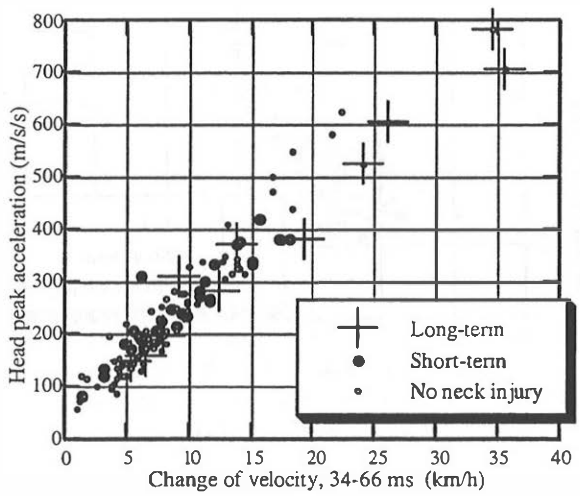


Figure 12. Head peak acceleration versus $\Delta v2$.

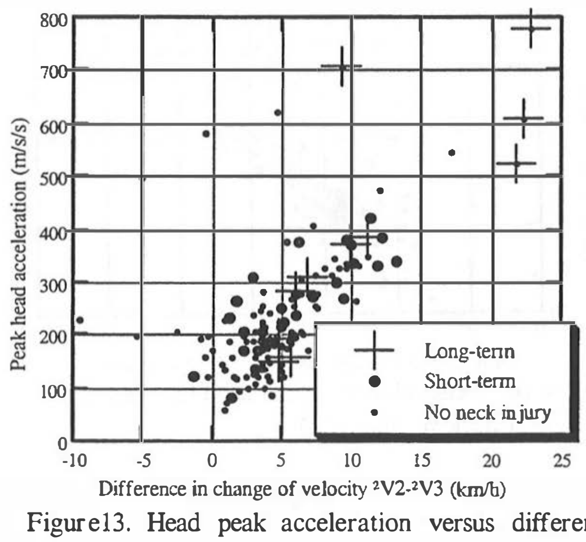


Figure13. Head peak acceleration versus difference $\Delta v2$-$\Delta v3$

Figures 14 and 15 shows that there are correlations between both neck horizontal force and torque (calculated in the saggital plane at the occipital condyles) and the long-term consequences. Figure 16 shows that the relative acceleration between upper and lower neck also seems to correlate with the long-term consequences. This relative acceleration is the difference between the linear accelerations calculated at the upper and lower neck joints (essentially C1 and T1). The local co-ordinate systems are used for these accelerations and thus simulate potential accelerometer recordings. Figures 17, 18 and 19 shows that the relative velocity between sternum and the vehicle interior and head peak angular acceleration and velocity seems to correlate with the risk of long-term consequences.

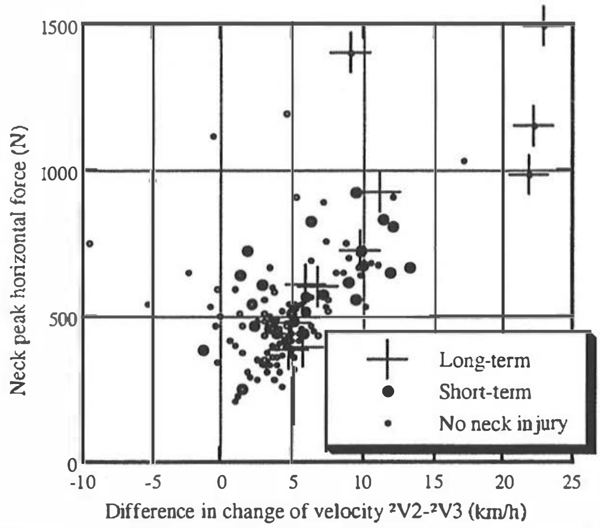


Figure 14. Neck peak horizontal force versus $\Delta v2$-$\Delta v3$.

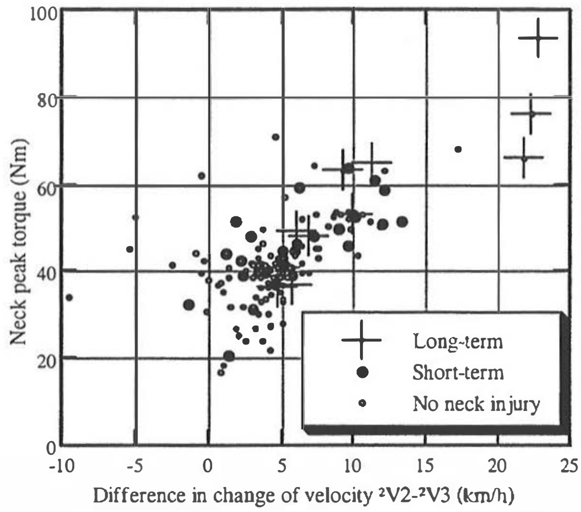


Figure 15. Neck peak torque versus $\Delta v2$-$\Delta v3$.

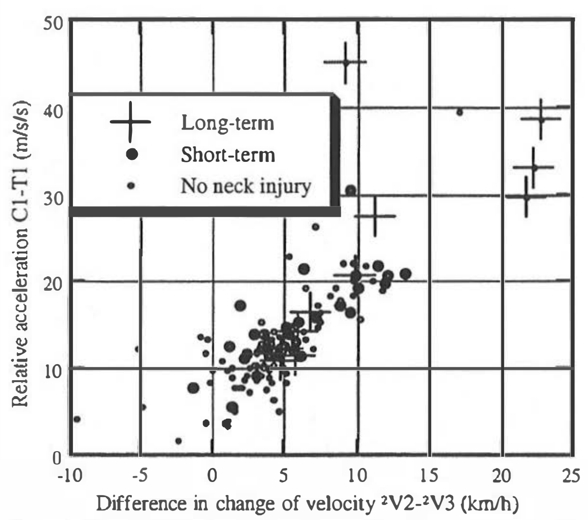


Figure 16. Peak relative acceleration between upper and lower neck versus $\Delta v2$-$\Delta v3$.

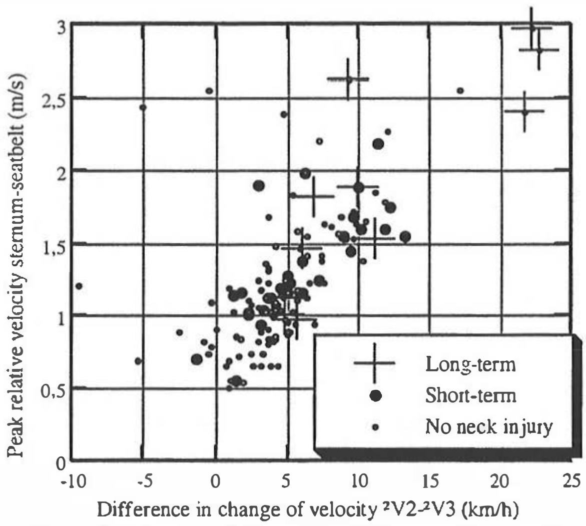


Figure 17. Peak relative velocity between sternum and vehicle interior versus $\Delta v2$-$\Delta v3$.

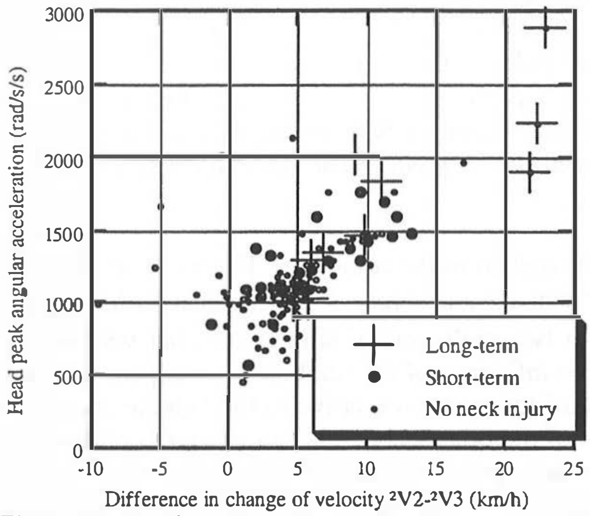


Figure 18. Head peak angular acceleration versus *Δv2-Δv3*.

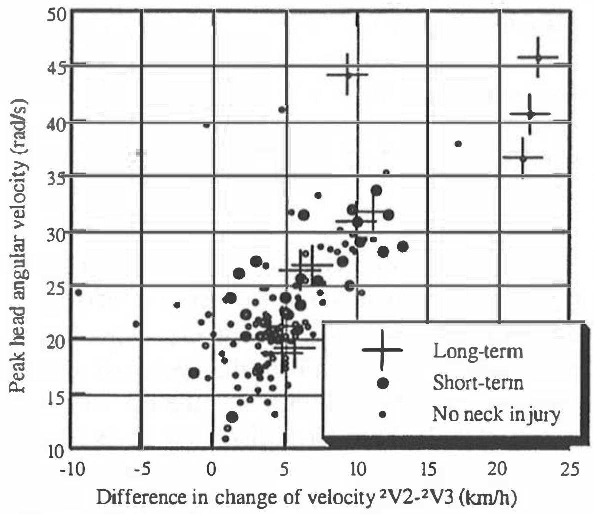


Figure 19. Head peak angular velocity versus *Δv2-Δv3*.

DISCUSSION

The material used in this study is unique. The impact severity parameters were calculated based on measured crash pulses in real-life collisions and the injuries have been followed up to 5 years from the time of impact. The recorded impact severity was used as an input for computer simulations and the output could then be compared to the injuries in the real-life accident sample. This combination of valid and reliable data from real-life impacts with computer simulated output made it possible to study relations which would otherwise be impossible to obtain. The use of crash pulse characteristics, instead of total **ΔV**, allowed more separation of injury and non-injury cases to be achieved and represents an effective tool in future epidemiological studies. Employing composite pulses instead of complete pulse histories can reduce the data and processing requirements for analysing crash trauma without losing too much information.

The first simulation phase was used to identify significant occupant response parameters and pulse characteristics. The results indicated that pulse shape had a significant influence on occupant response. As the pulse magnitude of 50 km/h is approached, the occupant response is almost equally affected by crash magnitude (**ΔV**) and pulse shape. Also observable in Figure 6 is the relation between occupant output to a crash pulse and a reference square pulse shape. The square wave represents the least severe loading shape for most conditions investigated. This is an important point as the square pulse shape appears to be the least severe of all possible pulse configurations. Occupant protections systems developed to protect against this pulse will thus be sub-optimised for real pulses that have been recorded.

An important feature that was also seen in the simulations was the strong correspondence between a parameter, like head acceleration, and the change of velocity in the second pulse segment. As seen in Figure 12, this relation is very useful for future accident studies, because this relationship between occupant and vehicle response is much better than for just **ΔV** as seen in Figure 9. This linearity is important when correlating severity of impact to occupant response, with or without an injury.

The results of all the simulations showed that the shape of the pulse had a significant affect on the response parameters of the occupant. Many of occupant results (accelerations, forces, etc.) parameters had similar trends when plotted against the different pulse shapes. A preliminary statistical analysis, comparing the mean values for all injury simulations and the subset of long-term injury simulations was conducted. However, the number of long-term consequences was too small to make statistically significant conclusions. Thus the ranking of parameters that predicted long-term consequences was not conducted.

One injury predictor was related to seatbelt loads applied to the occupant. Figure 11 illustrates the occupant kinematics and the relation between belt contact and injury. The figure also illustrates the different motions an occupant may experience from two collisions of similar $\Delta\mathbf{V}$, but with different pulse characteristics. From this series of pictures, the influence of the seatbelt on occupant motions is obvious. As seen in Figure 17, the maximum speed of the sternum relative to the vehicle can be used to separate the long-term disability cases from most of the other cases. This parameter is related to the amount of restraining effort required of the restraint system. The occupant loading arising from seatbelt contact is the predominant source of head-neck motions leading to injury. By association, the time at which occupant makes contact with the seatbelt and begins loading the shoulder belt should thus be related to the injury risk. However, this parameter was not as good a predictor as the actual occupant kinematics. This may be partially due to the fixed occupant size and position used in each situation which did not reproduce the characteristics of the actual occupants in the sample. In addition, the time of belt contact does not directly infer the severity of the occupant loading by itself.

A parameter that was found effective for predicting injury was the relative C1 and T1 accelerations. Since this parameter is part of the NIC criteria calculation, effective for rear impacts (Boström et al 1996), it should also provide some indication of neck injuries in frontal impacts. The model used in the study did not have a sufficiently detailed neck structure to simulate the "S" shape phases of neck motion. As a result, the full NIC calculation incorporating relative velocity and acceleration was not pursued in this study. Parameters such as NIC should be investigated for frontal impacts in further investigations of AIS 1 neck injuries.

All the parameters presented in Figures 12-19 exhibit similar characteristics. This is expected as they were chosen as potential indicators of neck injury and should thus be correlated. Parameters indirectly related to the head neck response (relative velocity of the occupant and vehicle, Figure 17) did not predict long-term injuries as well as directly related variables (relative neck accelerations, Figure 16). No single variable stood out as a definitive neck injury predictor. This information will only come with improved simulations including occupant characteristics and larger sample sizes.

The main limitations of the present study were the simplified vehicle and occupant representations. No attempt was made to reproduce the different sitting positions or seatbelt systems represented in the field data. These variables are difficult parameters to use effectively. Occupant positioning information is only available from the occupant and witness reports and is not reliable nor truly quantitative. Accurate, vehicle specific, belt models can be used, but their dependence on occupant positioning decreases their effectiveness in this type of study and only introduces more variables to the analysis.

When plots of occupant response versus $\Delta\mathbf{V}$ were investigated, they all had similar characteristics as Figure 9. For impact speeds below 20 km/h, the results were directly related to $\Delta\mathbf{V}$, as seen in this figure. This may arise from a similar response of most vehicles below this collision threshold, essentially elastic vehicle behaviour. Vehicle crush mechanisms are more active above 20 km/h and the various structural designs should be observed by different acceleration responses. The performance of the Hybrid III dummy may also be limited below this threshold and may not be suitable for discriminating between different crash pulses below 20 km/h. Above 20 km/h, the dummy may be providing better sensitivity to pulse characteristics. A comparison of the Hybrid III mechanical dummy and human volunteer response for a 15g (approximately 30 km/h ΔV) impact was made by Seeman et al (1986). The dummy was found to have a stiffer response for this loading level. As the computer model used for the simulations is based on the mechanical Hybrid III dummy, it is assumed to exhibit the same trends. Thus, results obtained for the lower speed collisions will be suspect until better neck models can be used for the analysis. The development of more bio-fidelic multibody models will increase the sensitivity and applicability of future studies. As a result of the coarse occupant model, not all of the injuries that occurred at low speeds could be separated from other non-injury cases. This can be attributed to both the limitations of the model at low speeds and the use of one generic occupant stature for modelling a broad occupant database.

The occupant simulation is limited in this preliminary study because of the selection of the 50th percentile Hybrid III male dummy. The accident data represents different occupant characteristics from this occupant definition. This mathematical representation must also be qualified as a model of a mechanical dummy and thus has a less human like response than is desired for this type of study. However, since dummies are still used to measure occupant response in crash tests, it is useful to understand what measurement parameters are available and how the dummy responds under these crash conditions. As more bio-fidelic models become available, these simulations can be repeated to obtain better, human like responses. The use of one occupant configuration allows an easier comparison between loading conditions and occupant results. The sample set can still be considered reasonably small, especially when the limited number (11) of long-term consequences is considered.

The injury mechanism for AIS 1 injuries in frontal impacts is not identifiable from the results presented herein. However, criteria for the injury, in terms of crash-pulse characteristics and occupant kinematics, may be defined using this type of study.

CONCLUSIONS

Computer simulations of a Hybrid III dummy showed that occupant kinematics and inter-segmental loadings were significantly influenced by the shape of a crash pulse. Using pulse shape characteristics, loading situations that were more likely to cause injury could be identified as opposed to using a $\Delta \mathbf{V}$ threshold.

All occupant response parameters presented in this study corresponded well with the occupants classified as long-term consequences in the real-life collision sample. The sample size was too small to calculate the effectiveness of individual parameters to predict long-term injuries. However, there are qualitative differences between variables that can still be observed.

The work presented herein represents a significant step forward in accident analysis and injury characterisation. Future studies that refine the model and increase the database will not only improve the quality of the results, but also increase its applicability.

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