Potential Effects of Deceleration Pulse Variations on Injury Measures Computed in Aircraft Seat HIC Analysis Testing

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Abstract

Aircraft seating systems are evaluated utilizing a variety of impact conditions and selected injury measures. Injury measures like the Head Injury Criterion (HIC) are evaluated under standardized conditions using anthropomorphic dummies such as those outlined in 14 CFR part 25. An example test involves decelerating one or more rows of seats and allowing a lap-belted dummy to impact components in front of it, which typically include the seatback and its integrated features. Examples of head contact surfaces include video monitors, a wide range of seat back materials, and airbags. The HIC, and other injury measures such as Nij, can be calculated during such impacts. A minimum test pulse, with minimum allowable acceleration vs time boundaries, is defined as part of the regulations for a frontal impact. In this study the effects of variations in decelerations that meet the requirements are considered. A series of Finite Element simulations of a generalized aircraft seat were performed to determine the variation in HIC and Nij observed based on variations in the deceleration pulse. The results indicate that the pulse characteristics affect the resulting head motions and can influence the ability to pass the HIC analysis test.

Introduction

Airworthiness standards have been created to provide protection to aircraft occupants under various conditions. For transport category airplanes, the airworthiness regulations have been defined in Title 14 of the Code of Federal Regulations Part 25 (14 CFR 25) [1]. Requirements have been defined in other sections for other types of aircraft, but 14 CFR 25 is applicable to commercial passenger aircraft. 14 CFR 25, and related advisories, identifies various impact conditions and test methods to be used for evaluation of the crashworthiness performance of seating systems that are authorized to be used for these planes. The impact conditions include definitions of the minimum decelerations time histories (pulses) that are to be applied to the seats in various impact modes. The minimum pulses are defined such that they represent the lowest acceptable test pulse allowed by the regulation. The regulations do not, however, limit the maximum values of deceleration pulses that can be used. It is, of course, in the interest of the seat manufacturer to conduct the tests at the lowest possible test pulse in order to reduce the peak loads generated.

These pulses can be applied to the seats using either physical or virtual testing methods. With physical testing methods, typically test sleds using either impact or reverse acceleration methods have been used. With virtual testing methods the decelerations are applied with highly advanced and sophisticated scientific engineering tools that are utilized almost universally across industries to design and evaluate crashworthiness performance of various types of transport and occupant protection systems [2]. The test devices (anthropomorphic dummies) that are position in the seats are able to measure, for example, accelerations in the head and forces and moments in the neck that can be used to quantify metrics called injury measures. These injury measures can then be compared against injury assessment reference values to determine whether the seat system is acceptable or not based on the specifications contained within 14 CFR 25. The relationship between such injury measures and human injuries depends on how well the test devices reflect responses that can be related to injuries for a given body area (biofidelity). Over the years anthropomorphic test device representations (both physical and virtual) have been introduced and with each new generation the biofidelity has been improved. Certification that a seat meets the requirements can be done with physical testing or using methodologies using virtual testing [3].

An example frontal impact test involves a dummy seated in an upright position, held with a two point belt that is decelerated from a defined speed and allowed to impact the components that are in front of the dummy. Examples of head contact surfaces involving a seat in front of the dummy can include video monitors, a wide range of seat back materials, arm rests, tray tables, and airbags.

While a minimum deceleration pulse is defined as part of the regulations for a frontal impact, variations in the allowable pulse are thought to significantly affect the resulting response of dummy prior to and during head impact. Such variations include the slope, shape, and duration of the pulse, which can all vary substantially will maintaining the same delta-V. Previous work by NASA provided comparative information on various crash pulse shapes on seat pan loads in general aviation seats [4]. This study quantifies the differences in selected dummy Head Injury Criterion (HIC) and a neck injury measure (Nij) [5] as a function of various deceleration pulse magnitudes, i.e. onset slope and peak value, that all meet the requirements of the test definition. A series of finite element simulations of a generalized aircraft seat were performed to determine the variation in observed injury measures based on incremental increases in the deceleration pulse. The results indicate that the pulse characteristics affect the resulting head motions and can influence the ability to pass the frontal impact occupant protection HIC analysis test.

Method

A diagram which summarizes the deceleration pulse shape requirements for a Zone C test HIC evaluation is provided in Figure 1 [6].

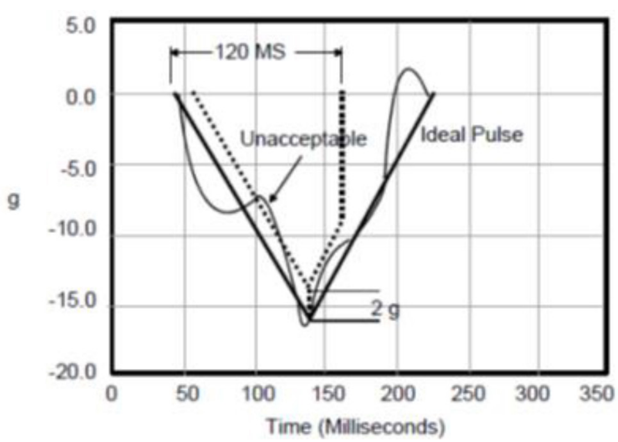


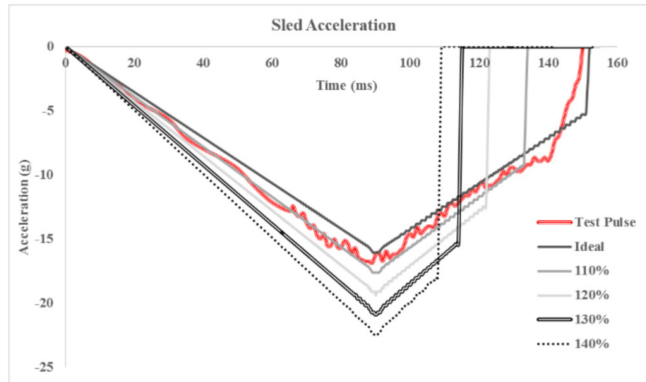
Figure 1. Frontal sled impact pulse requirement (abstracted from 14 CFR 25).

The requirements for this test are:

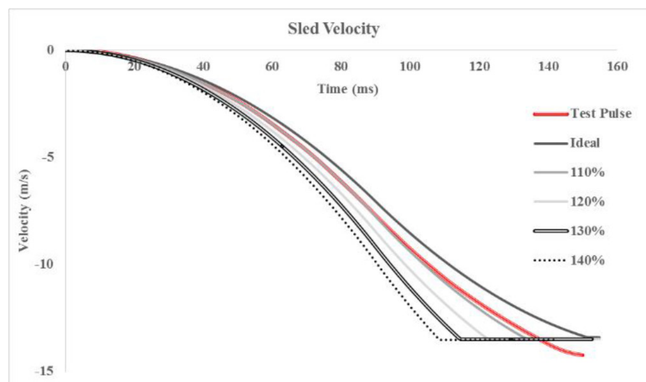
1. the magnitude of the pulse be greater than 16 g;
2. the rise time must be less than 90 ms;
3. one-half the required velocity change (6.7 m/s) be reached prior to 90 ms;
4. the total required velocity change be achieved prior to 207 ms after onset or prior to the acceleration reaching 0 g, whichever occurs first.

If the above requirements are met and the measured pulse is greater than the ideal pulse during the entire interval up to 16 g, then the pulse is defined as acceptable. Measured pulses are allowed to fall up to 2 g below the ideal pulse, however further requirements apply that are not applicable to the curves defined in this study.

Six different acceptable pulses were defined for the simulations and are given in Figure 2a. Five pulses were generated and scaled from the defined ideal pulse, and a representative test pulse was included for comparison. The ideal pulse defined in 14 CFR 25 (shown in Figure 1) was refined to provide a maximum sled velocity of 13.4 m/s (30 mph) by reducing the deceleration to zero at a predetermined time after the peak acceleration. Additional select pulse variations were derived by amplifying the ideal crash pulse to achieve peak decelerations equal to 110, 120, 130, and 140% of the minimum required deceleration pulse. All pulse variations, except the representative test pulse, were designed to limit the maximum change in velocity to 13.4 m/s, as demonstrated in Figure 2b.



a.



b.

Figure 2. Defined and experimental sled acceleration (a) and velocity (b) time histories, both zeroed at crash pulse initiation.

A non-linear explicit finite element solver, LS-DYNA 971 [7], was used to simulate the select tests with the prescribed deceleration pulses. The model included two rows of single seats with a pitch of 913 mm. The target seat had a video monitor included in its standard position. A seated 50th percentile Hybrid III dummy [8] was settled in the rear seat with the belt tightened evenly across its hips as shown in Figure 3.

Filter classes consistent with SAE J211 [9] were used for post processing. The HIC was calculated for the first head to seatback impact using a time interval of 36 ms [10]. A neck injury criteria, referred to as Nij, was also calculated for each impact according to the equations and thresholds outlined by Eppinger et al. [5]. All results were normalized to those calculated from the ideal pulse.

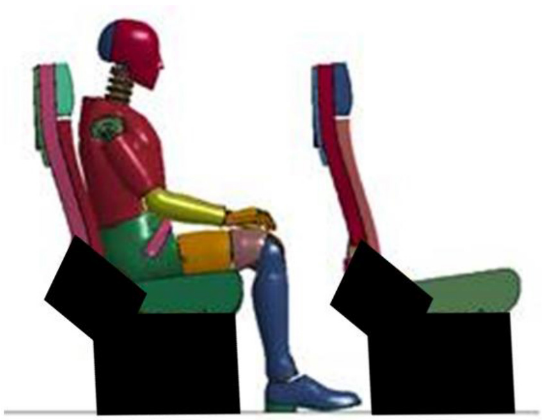


Figure 3. Example of initial test setup

Results

All results are normalized to the ideal case. The calculated HIC varied substantially between tests as shown in Figure 4. The HIC values did not appear to be directly associated with the peak sled deceleration magnitude. The simulations with the greatest or least peak accelerations did not result in the greatest and least HIC values, respectively. The time interval over which the HIC was calculated for each impact generally decreased with increasing pulse magnitude with the exception of the 'ideal' pulse which had the shortest interval (5.8 ms).

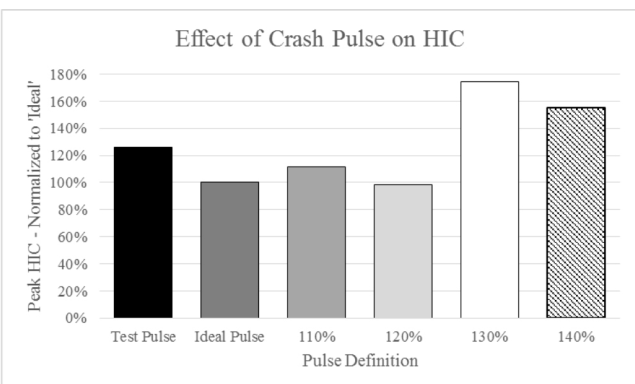


Figure 4. Effects of sled pulse variations on HIC

The peak head acceleration values, normalized to the ideal case, for each pulse definition are shown in Figure 5. A general upward trend is observed.

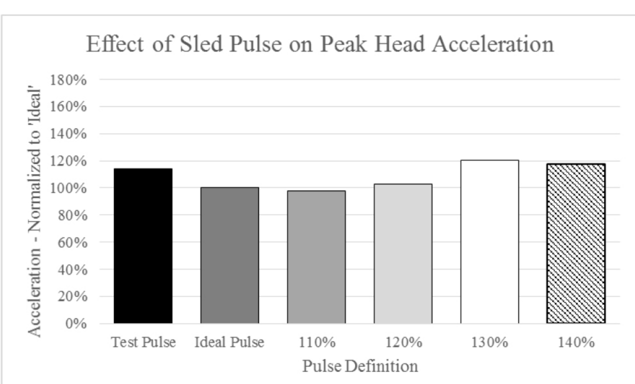


Figure 5. Effects of sled pulse variations on peak head acceleration

The Nij values the tests also varied somewhat erratically and did not appear to be related to the crash pulse as shown in Figure 6. Four of the six simulations resulted in maximum Nij values for the tension-extension orientation, while the 'ideal' and the 120% cases had Nij maximums in the compression-extension orientation.

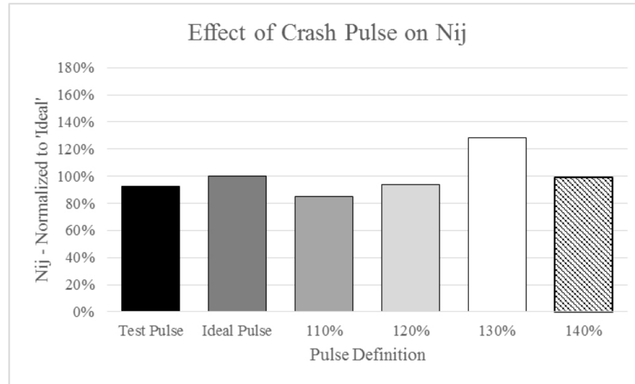


Figure 6. Effects of sled pulse variations on Nij

The motion of the dummy head, forward seatback, and video monitor was affected by variations in crash pulse magnitude. As the sled began to decelerate, the dummy moved forward relative to the seat and the hands contacted the target seatback. Interaction between the hands and target seatback altered the motion of the seatback, video monitor, and dummy. Greater deceleration pulses resulted in earlier and longer contact between the hands and seatback. This generally also resulted in the monitor base rotating further out of the shroud prior to head impact. As the lap belt reached its peak force, the upper torso began to rotate forward and downward, with the head and neck following in slight extension. Depending on the timing and deceleration pulse the neck may move into flexion resulting in increased angular velocity of the head just prior to impact, such as in the 10% case.

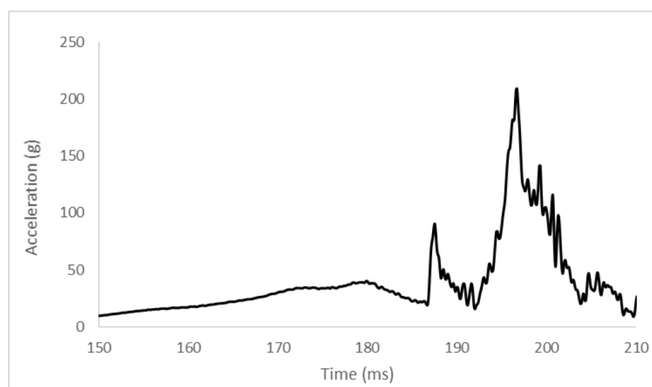
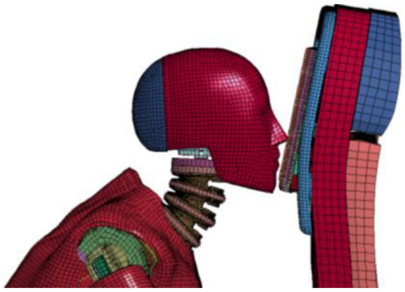
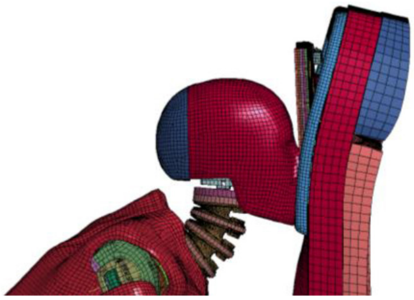


Figure 7. Typical head acceleration response

The relationships were complicated and suggest that interactions between the dummy, seat, and seat components affect the resulting response and subsequent injury measures. While the monitor is coupled to the seatback, it has the ability to rotate and typically contacts the dummy head prior to the seatback. This impact between the dummy head and video monitor can be seen in the head acceleration curve, Figure 7, at 186 ms. The head typically first contacted the lower portion of the video monitor as displayed in Figure 8a, resulting in a local head acceleration peak. The head then pushed the monitor into the seatback until they both interacted with the seatback roughly 7-10 ms later as shown Figure 8b. This results in a bimodal acceleration curve such as that presented in Figure 7.



a.



b.

Figure 8. Head-to-monitor impact (a) and head-to-seatback (b) impacts

Discussion

The magnitude of the deceleration pulse clearly affected the kinematics of both the dummy and the target seat prior to head impact. Interaction between the dummy hands and target seat prior to head impact complicated the resulting response of the dummy head. The effects ultimately manifested as differences in closing speed between the dummy head and seatback/monitor.

The 110% case resulted in higher injury measures than the 120% case which indicates that the dummy response is not simply directly related to the magnitude of the deceleration pulse. Examination of this phenomena revealed that in the 110% case the closing speed between the head and seatback was greater than the 120% case. This resulted in more seatback shroud interaction than the 20 % case for the time duration reflected by the HIC.

Head impact response is bimodal with an initial impact occurring first between the head and video monitor followed by an impact to the lower shroud. These two distinct impacts can be seen as distinct local peaks in the head acceleration trace with a relatively low acceleration due to impact with the monitor followed by the much higher acceleration as the head interacts with the shroud and the monitor is displaced toward the seatback structure.

The effects of pulse variation are primarily attributed to differences in the forward seatback displacement and dummy motion occurring prior to head contact. The resulting HIC is sensitive to both the impact velocity and head orientation as well as the behavior of the seatback and video monitor assembly.

Conclusion

The results show that variations in the sled deceleration pulse, with equal total velocity change, can affect the resulting HIC. There are virtually an infinite number of sled deceleration pulses that could occur in a crash or from test facility to test facility. The effects of these variations should be understood when considering the design implications from test results.

Since the potential exists for large differences in the observed HIC from a compliance viewpoint, the effects of a selected crash pulse should be considered. Probabilistic methods or design of experiment approaches would likely enable identification of deceleration pulses that result in the best HIC results for a given seat design.

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