

# SAFETY ASSESSMENT OF AUTONOMOUS EMERGENCY BRAKING SYSTEMS ON UNBELTED OCCUPANTS USING A FULLY ACTIVE HUMAN MODEL

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## ABSTRACT

This paper assesses the safety benefits of a typical Autonomous Emergency Braking System (AEBS) followed by a subsequent 25mph rigid wall impact using a 50<sup>th</sup> percentile active human model including full muscle activity behaviour. Occupant kinematics as a function of various postures and states of awareness are investigated to determine the degree of out-of-position and their respective chest, neck and head injuries.

The study concludes that the Madymo Active Human Model is suited to model active safety scenarios and that the generated kinematics and injuries provided are plausible.

The study has established that, within the active safety scenario investigated, the occupant's kinematics depend on the seat friction coefficient, arms' kinematics and the level of awareness. Overall, it has been observed that for a reflex delayed response of less than 120ms that chest, neck and head injuries values for gripping the steering wheel with 2 hands were comparable for a given value of seat friction. Alternatively, occupants with 1 hand on the steering wheel (holding a mobile phone for example) were out of the airbag deployment zone after 1.1s of extreme braking regardless of their state of awareness and seat friction value.

## INTRODUCTION

Passive safety has for many years reduced the number of fatalities on the roads. However, its effect on occupants' safety has now stabilised, meaning that new safety features, like active safety

are needed to further reduce the number of casualties [1]. These active safety features vary from Autonomous Emergency Braking Systems (AEBS) aiming at reducing vehicles speed prior to collision occurring in order to reduce the vehicles kinetic energy on impact to a minimum [2] to automatic lane change.

An initial method of assessing the effect of active safety involving an improved airbag model [3] and 1g pre-brake scenario on an occupant was undertaken [4] using a passive human model. Further implementations and details were incorporated and published [5], including a controlled spine allowing the occupant to sit straight and balance its own weight. This study however has shown that muscle activation was necessary as the occupant's kinematics during the pre-braking phase was independent of its stance. In order to remedy this, volunteer physical tests were performed under low 'g' sled tests [5] to derive human occupant target muscle activity curves as well as an active human computer model [7]. This new active human computer model still required some improvements as some of the muscle activity [9] timing and controller stiffness' needed further developments. It did however show some important findings; in some cases the duration of the AEBS pre-braking phase moved the occupant away from the airbag effective envelope in case of a subsequent impact.

This paper will initially focus on the validation of the new Madymo 7.4.1 Active Human Model in an unbelted scenario which will be used throughout this paper. Occupant kinematics due to various postures, state of awareness will be investigated to

determine the degree of out-of-position and their respective chest, neck and head injuries.

### VALIDATION OF AN UNBELTED ACTIVE HUMAN MODEL IN LOW 'G' SCENARIO

The first stage of the study required the calibration of the latest active human model in a lap-belt scenario under low deceleration [10][11]. It is believed that this scenario is the closest to an unbelted scenario without potentially injuring the volunteers.

Utilising the published sled model, it was possible to calibrate the muscle activity controller values in order to correlate the lap-belt sled tests under 1g frontal motion. The controller scalar value is set such that the limbs' position are maintained under the influences of external disturbances.

The seat friction value used between the seat and the human model was assumed to be 0.5 [8].

It was discovered that the spine, hips and arms controller were already suitable for replicating this test (*Figure 1*).

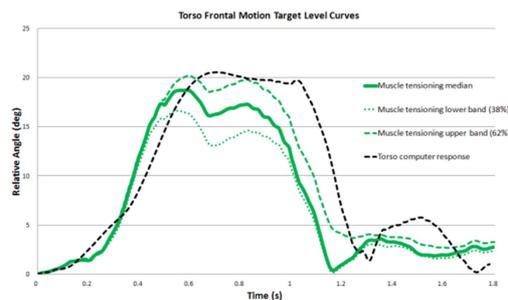


Figure 1. Correlation of torso motion in 1g and lap-belt scenario.

In a lap-belt scenario, the torso rotates more than in a 3 point belt configuration. Consequently, the neck muscle activity controller had to reproduce the fact that the neck flexion was increased (*Figure 2*) from the original model calibrated for a 3 point-belt [10].

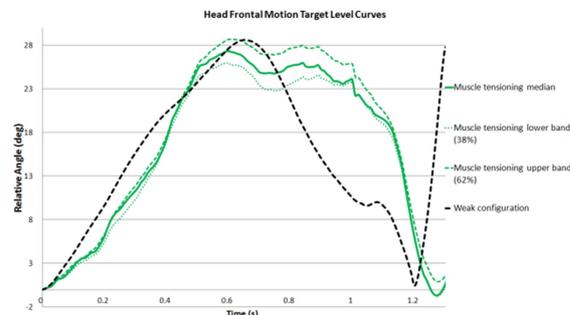


Figure 2. Correlation of head motion in 1g and lap-belt scenario.

The neck controller value was therefore modified to 0.6, whilst keeping all the original controller values to '1', as indicated in Table 1.

The proposed lap belt model's kinematics have a similar response shape and amplitude to the test results, especially for the torso (*Figure 1*). The head's relative angular rotation magnitude was comparable with the physical tests (*Figure 2*). Nevertheless the head relative angular motion returns to its original position faster than in the test. The model controllers being encrypted, it was not possible to make any further investigations into the head controller scalar value.

Table 1. Updated Madymo 7.4.1 controller values for unbelted occupant modelling.

Spine	Neck	Arms	Hip	Co-contraction	Reaction time (s)
1	0.6	1	1	0	0

It has to be noted that this current model is not capable of leg bracing. The "Co-contraction" field in Table 1 refers to neck bracing, which was set to "0" as the volunteers had no cognitive input of the test, i.e. were unaware of the start of the sled motion.

The active human model configuration listed in Table 1 was used throughout the entirety of this paper.

### DRIVERS' KINEMATIC STUDY

#### Study setup

The aim of this section is to evaluate the kinematics of an occupant under unexpected 1g emergency braking in different states of awareness, in a generic vehicle environment. As such, no bracing is applied prior to the braking event taking place.

The parameters which will be considered in this study are the seat friction parameter and the state of awareness of the occupant. Seat friction accounts for the evaluation of tendencies in lower extremities load paths, i.e. footrest and pelvis.

The seat friction coefficient in vehicles is of the order of 0.8 [7]. Nevertheless, values utilised for the correlation 0.5 and lower (0.3) were studied to evaluate the spread of the response to this variable. The seat model was constructed from planes with a stiffness characteristics extracted from an accident reconstruction technical report [12]. In all instances, it was assumed that the friction between the feet and the floor to be 0.85 representing rubber sole shoes to a carpet.

The awareness level can vary greatly. A "very aware" person has a reflex response time of 30ms; an "aware" occupant of 120ms, which can be modelled as a 'motor reflex delay' in the human model [11].

The hand is attached to the steering wheel using a RESTRAINT.POINT command in Madymo with a maximum grip force level of 400N [15][16], to simulate the hand release.

The list of normal awareness computer runs are listed in Table 2.

**Table 2.**  
Normal awareness computer setups

Run number	Stance	Reflex time (s)	Human to seat friction
Run 11	FMVSS208	0.030	0.5
Run 11a	FMVSS208	0.030	0.3
Run 11b	FMVSS208	0.030	0.8
Run 21	FMVSS208	0.120	0.5
Run 21a	FMVSS208	0.120	0.3
Run 21b	FMVSS208	0.120	0.8

Prior to performing the kinematics study, the occupant was positioned in the vehicle using a 1g vertical 'Z' for the duration on 1.5s to balance the occupant with its environment. During the 1g emergency braking, the 1g vertical gravity field was maintained.

The forward braking pulse was applied on the human model with the cabin and airbag system set as static. The pulse shape had a well documented characteristic as illustrated by Figure 3 [5].

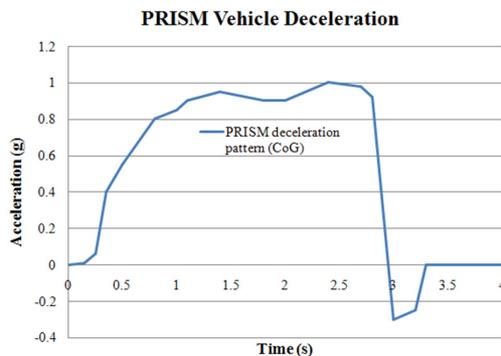


Figure 3. PRISM project. Straight line braking. Vehicle deceleration.

### Results of the occupant kinematics' study of standard grip stance ("very aware" or "aware")

The first results concerning the seat with very low friction indicated that the driver's pelvis was sliding forward until the leg contacted the dashboard, as shown in Figure 4. The pelvis is sliding because of the lowest resistance provided by the seat relative to the direct loading of the arms.

The torso (solar plexus) almost stayed still (+0.05m forward motion from original position, +X in Figure 4) due to the resistance of the arms, which consequently moved the head away from the airbag

(-0.05m rearward motion from initial position, -X in Figure 4).

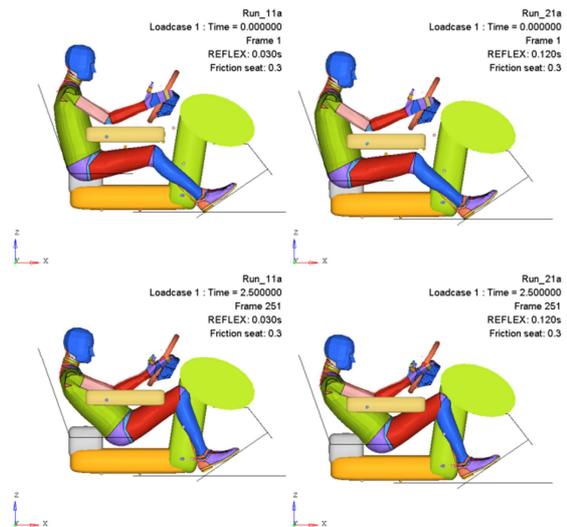


Figure 4. Scenario with seat with friction set at 0.3 (30ms and 120ms awareness displayed Left to Right) at time 0s (top) and 2.5s (bottom).

It was noted that for a very low seat friction, the occupant kinematics was very similar for a "very aware" and "aware" person, especially after 0.5s for the top of the head as well as the solar plexus, as can be noted in Figure 5 and Figure 6, where the displacement curves mostly overlap during the duration of the event.

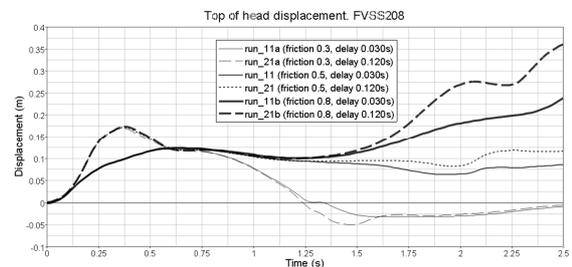


Figure 5. Summary of displacement of top of occupant's head.

It can be noted in Figure 5 that the head has a flexion motion due to the 1g braking pulse which is greater for a motor reflex delay of 120ms than 30ms, as the neck muscles are activated later. When a slower reflex occurs, it takes 500ms to match the head motions of an occupant with a faster reflex.

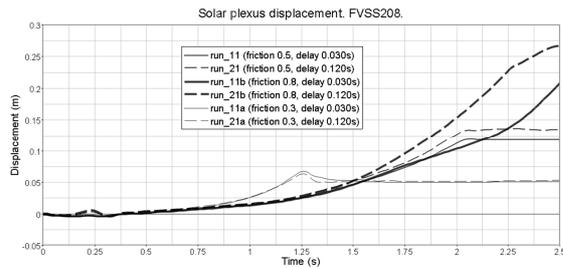


Figure 6. Summary of displacement of occupant's solar plexus.

It can be noted in Figure 6 that, for the same seat friction value, the solar plexus has a greater linear motion the longer the motor reflex delay.

Increasing the seat friction parameter increases the sliding force responsible for the occupants' motion. As illustrated in Figure 7, increasing the friction from 0.5 to 0.8 increases the resistive force to motion from 500N to 650N.

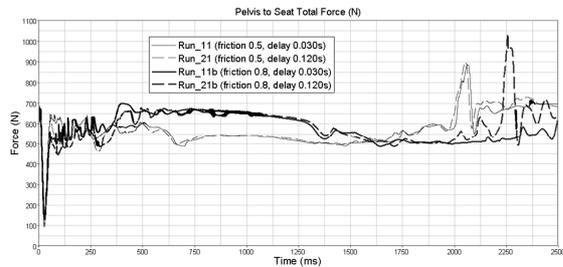


Figure 7. Occupant force on pelvis a function of seat friction and awareness.

Figure 7 also suggests that a motor reflex delay less than 120ms does not have an influence on the seat force due to friction. The human model used has a stabilising spine which will naturally keep the occupant seating straight and hence transfer the load onto the seat. The mass transfer looks very noisy early in the seat force readings (Figure 7), which may be caused by the repositioning of the human model from the initial gravity positioning as well as the early muscle activity which affects the heads' forward motion (flexion). Looking at scenarios with greater seat frictions, i.e. 0.5 and 0.8, it can be observed that the kinematics are different initially due to the fact that the seat friction resists the pelvis motion and forces the torso to rotate towards the steering column. This is clearly illustrated in Figure 8 and Figure 9.

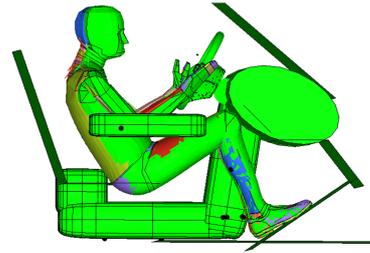


Figure 8. Comparison of occupant kinematics for seat friction 0.5 (30ms and 120ms awareness displayed in blue and green respectively). Pre-braking duration of 2.3s.

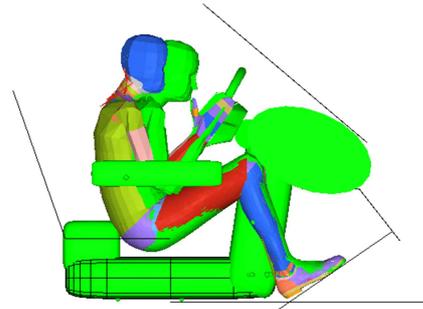


Figure 9. Comparison of occupant kinematics for seat friction 0.8 (30ms and 120ms awareness displayed in blue and green respectively). Pre-braking duration of 2.3s.

This suggests that the higher the seat friction the greater the torso rotation as the relative speed between the pelvis and the torso increases.

Looking at the reflex levels, Figure 8 and Figure 9 suggest that a slower reflex leads to a closer thorax position relative to the steering column.

Indeed, with a slower reflex, the velocity of the torso (measured at the solar plexus) is higher from approximately 1.2s, as the muscle activity in the human model is lagging. With increased velocity, the momentum is increased.

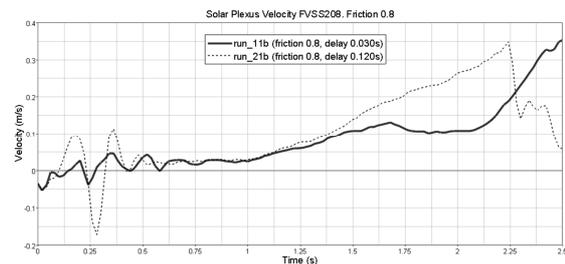


Figure 10. Velocity of human model's solar plexus.

As the hands are restrained on the steering wheel by a RESTRAINT.POINT command with no torque reaction, the arms rotate at the steering wheel attachment to compensate for the momentum.

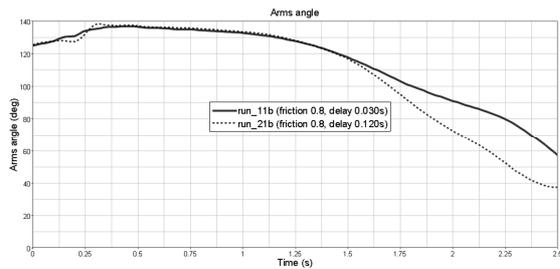


Figure 11. Occupant's arm angle change (deg).

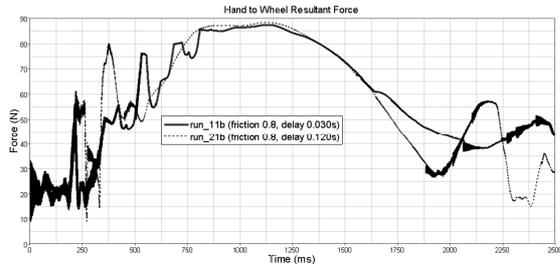


Figure 12. Hand force on the steering wheel.

As illustrated in Figure 11, the occupant's arm's angle starts to reduce from run\_11b ("very aware") to run\_21b ("aware") at time 1.5s. The force exerted on the steering wheel is reduced (Figure 12) and the occupant is therefore closer to the steering wheel, as illustrated in Figure 9.

### Results of the occupant kinematics' study of Mobile Phone stance

Using the same model setup as the 2 hand grip, removing the left hand from the steering wheel and raising to the ear level, it has been shown that the coefficient of friction had a great importance in the position of the occupant using a mobile phone.

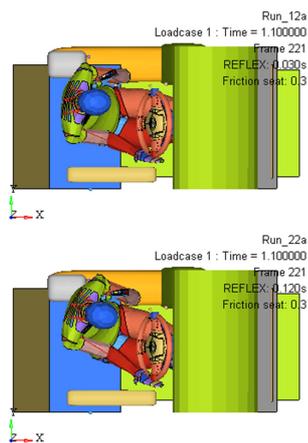


Figure 13. Kinematics results of mobile phone stance friction 0.3. "Very aware" (top), "aware" bottom.

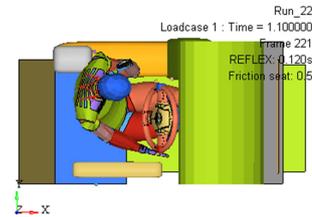
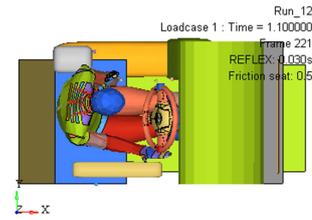


Figure 14. Kinematics results of mobile phone stance friction 0.5. "Very aware" (top), "aware" bottom.

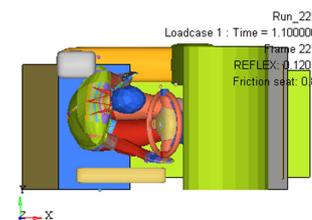
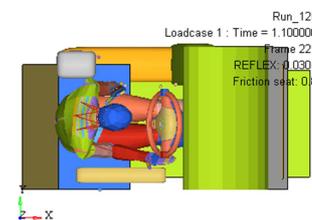


Figure 15. Kinematics results of mobile phone stance friction 0.8. "Very aware" (top), "aware" bottom.

Figure 14 and Figure 15 suggest that:

- The greater the friction, the more 'central' the occupant is situated (Y direction in Figure 15)
- The faster the motor delay, the more 'central' the occupant is situated (Y direction in Figure 15)

Using this latest Madymo human body model, it was possible to re-confirm that a braking duration in excess of 1.1s positions the occupant out-of the airbag zone of influence, as previously reported [9]

### Conclusions on the driver kinematics study

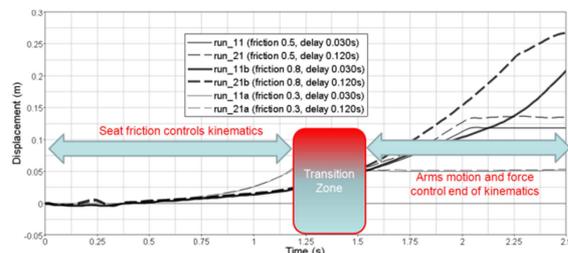
The driver kinematics suggest that the occupant needed to have a good level of awareness (motor reflex delay < 120ms) to resist an unexpected AEBs with no Frontal Collision Warning (FCW).

An occupant using a mobile phone will be in the airbag envelope up to 1.1s of AEBs braking

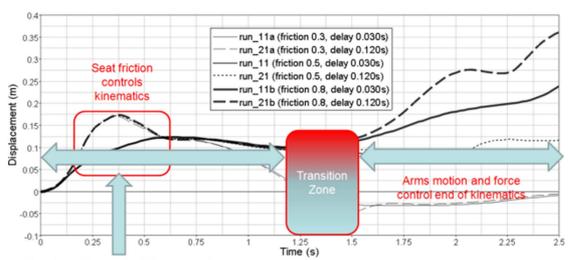
duration with no FCW. Should the duration last longer and should an impact occur, then the occupant will not have any restraint systems to protect it.

In all standard grip starting positions, i.e. with the 2 hands on the steering wheel, the study shows that the human model will resist the deceleration up to a braking duration of 2.5s (computed for all runs) even though it will move closer to the steering wheel. No hand loads has exceeded the 400N threshold level.

All the run comparisons are listed in *Figure 16* and *Figure 17*.



*Figure 16.* Summary of Solar Plexus displacement (all runs).



*Figure 17.* Summary of Top Head displacement (all runs).

Following the study it can be concluded that the first component of the occupant motion is the seat friction, which controls the first 1.0s of the motion. The second part of the motion is due to the motion of the arms which is a result of the own increased kinetic energy as well as possibly the torso's kinetic energy.

Higher friction leads to higher relative velocities from the pelvis to the thorax, hence greater energy of rotation of the thorax relative to the pelvis, leading to increased rotation of the arms, reduction of grip force and consequently closer proximity to the steering wheel.

Considering *Figure 16* and *Figure 17*, the change of energy transfer comparing 2 occupants having different motor reflex delay, i.e. awareness, is difficult to pinpoint accurately.

From the results obtained, a 'transition zone' has been evaluated from the graphs which bands the possible starts of the increase in kinetic energy from the torso. This zone starts around 1.0s and finishes at 1.5s.

As a summary, considering all the variables in this posture study (seat friction, reflex delay and braking duration), it can be concluded that the active human model's kinematics are, in a 1 hand and 2 hand steering wheel grip, reasonable.

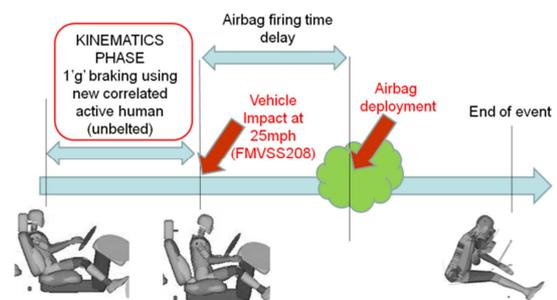
## DRIVERS' INJURY STUDY

### Background and study setup

The aim of this section is to investigate the effects of occupant injuries when vehicles are fitted with AEBs and assuming that full brake is applied on an unbelted occupant with no FCW. The occupant will have a reflex behaviour and not a bracing one as the braking event is sudden and unforeseen.

This study will compare the occupant protection level based on a standard FMVSS208 rigid wall crash test (25mph) against a 1g vehicle deceleration until the vehicle reach 25mph, then followed by a rigid wall impact.

This deceleration will cause the occupant to be out-of-position (OoP) before the impact takes place, as illustrated in *Figure 18*.



*Figure 18.* Severe braking scenario followed by rigid wall impact (airbag fire time set to 10ms).

Following the kinematics study, injuries have been extracted for chosen scenarios. The scenarios with the friction parameter of 0.3 have not been further analysed as the occupant's head has moved away from the airbag (*Figure 13*).

Cases where head and thorax positions lie within the airbag envelope were favoured.

The crash phase will utilise the pre-braking occupant position which will be re-mapped into an environment with 1g vertical gravity loading and the vehicle crash pulse.

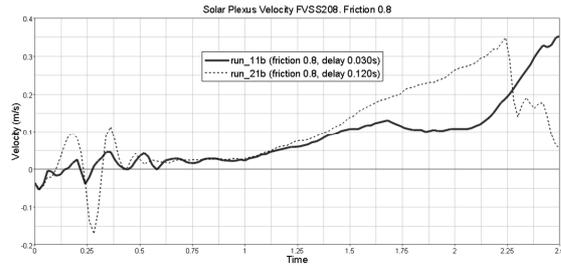


Figure 19. Solar Plexus Velocity for a seat with a friction of 0.8.

Due to the fact that the solar plexus velocity is low velocity (0.35m/s), as illustrated in Figure 19, it was assumed that the occupant was in a state of equilibrium relative to the steering wheel before the accident event starts.

This split-run strategy was used to improve the CPU runtime and allow this study to be performed.

The scenarios with the friction parameters of 0.5 and 0.8 were selected, considering 3 braking durations aiming to reduce the vehicle from its original cruising speed down to 40km/h (25mph) [7], as listed in Table 3.

**Table 3.**  
Braking duration to reduce vehicle speed back to 25mph

Vehicle start velocity (km/h)	60	80	100
Time to reach 25mph (s) under extreme braking	1.1	1.7	2.3

The vehicle crash pulse information utilised has been obtained from previous research [9]. To allow a qualitative comparison between the injuries, it has been necessary to scale the magnitude of the input crash pulse by 65% in order to generate a safe vehicle under FMVSS208 unbelted criteria.

The scaling of the crash pulse was done as the origin of the vehicle from which the airbag computer model was not known, as well as its engine variant, interior type (friction), ergonomics, crash pulse and steering column characteristics [18][19].

In the study undertaken, each occupant starts from a slightly different position due to seat friction values and braking durations before the crash pulse is applied. As a consequence, it is not possible to categorically state the exact cause of each injury value recorded in Figure 20 to Figure 25.

### Results for the standard grip stance

All Neck Injury values ( $N_{ij}$ ) are well below the legal limit of 1, hence have not been plotted.

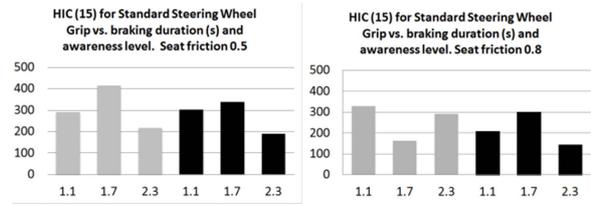


Figure 20. HIC (15) for Standard Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

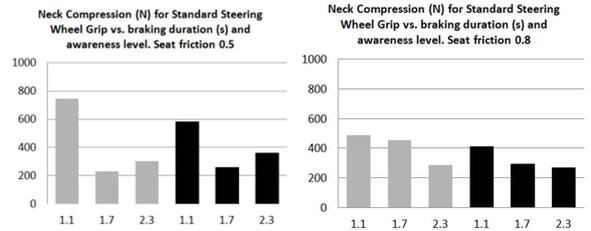


Figure 21. Neck Compression for Standard Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

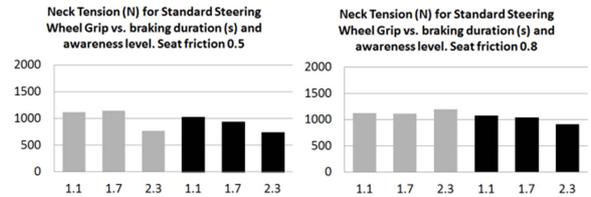


Figure 22. Neck Tension for Standard Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

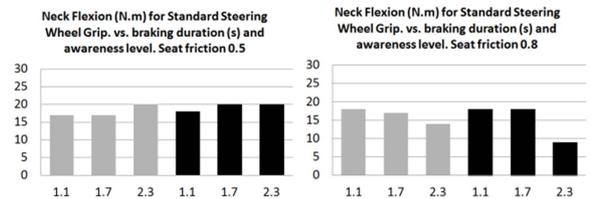


Figure 23. Neck Flexion for Standard Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

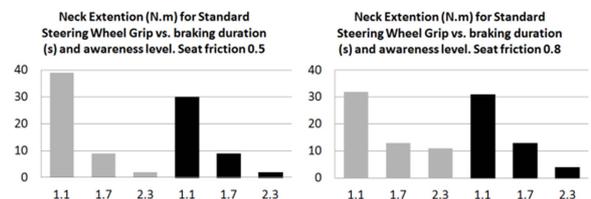


Figure 24. Neck Extension for Standard Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

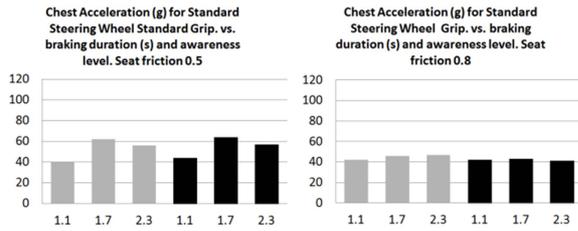


Figure 25. Chest Acceleration for Standard Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

### Discussion for the standard 2 hand grip stance

Considering the chest acceleration values from Figure 25, it can be observed that the injury values are comparable within a given friction and pre-braking duration value. They are also in the same order of magnitude, between 41g and 64g.

Looking at Figure 25 it can be observed that the maximum chest deceleration values are below 64g, when the braking duration equals 1.7s, i.e. 64g for the 30ms reflex delay and 62g for the 120ms reflex delay one.

Considering Figure 6, which is the solar plexus' forward motion, it can be observed that the longer the braking duration, the closer the thorax is to the steering wheel, hence interacts with the airbag sooner.

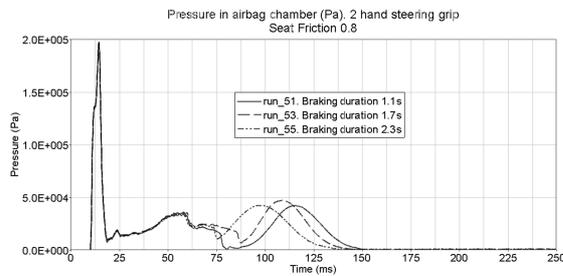


Figure 26. Airbag pressures for a 2 hand steering grip and seat friction of 0.8.

This is illustrated by the airbag pressures (Figure 26) showing the burst-out phase (15ms) followed by the membrane loading phase (from 75ms) which is starting earlier the closer the occupant sits relative to the restraint system.

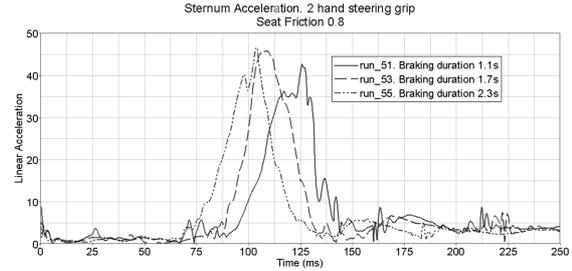


Figure 27. Sternum acceleration for a 2 hand steering grip and seat friction of 0.8.

The delay in the restraint system occupant loading is illustrated by the timing of the chest injuries, especially the chest to steering wheel interactions as illustrated in Figure 27.

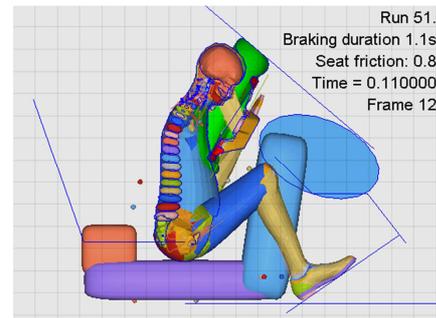


Figure 28. Run51. Start of interaction between chest and steering wheel at time 110ms

The interaction between the chest and the steering wheel can be observed and correlated to the increased chest acceleration starting at time 110ms. This is illustrated in Figure 28, where the occupant's thorax contacts the steering wheel in a 25mph wall impact prior to a 1.1s pre-braking phase.

It must also be noted that in the model, the steering wheel is mounted to a rigid bracket, with no steering column collapse possible. Consequently, the chest acceleration values quoted in this study are comparative values and not absolute ones.

It can be observed from Figure 20 to Figure 25 that contacting the airbag earlier does not generally increase the occupant's injury levels.

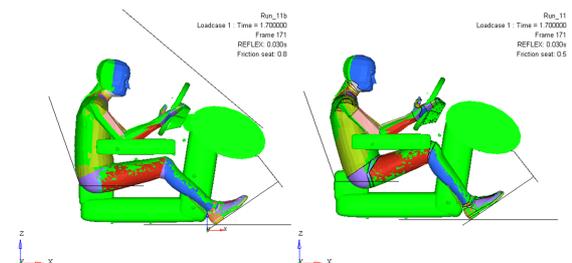


Figure 29. Comparison of occupant kinematics for seat friction 0.5 left, 0.8 right (30ms and 120ms awareness displayed in blue and green respectively). Pre-braking duration of 1.7s.

Looking at Figure 29, it can be also observed that the knees have not yet contacted the dashboard at a time 1.7s before the subsequent impact occurs, compared to a braking duration of 2.3s (Figure 8 and Figure 9). As a consequence, more mass will be accelerated, hence more energy transferred into the airbag/ steering wheel assembly. This is suggesting that the braking duration is influencing the occupant's kinematics in the subsequent crash scenario. A longer braking duration (2.3s) would mainly cause a rotational torso momentum in the crash phase because the knees are already in contact with the knee bolster at the time of impact, while a smaller braking duration (1.1s and 1.7s) would allow a legs and torso translation followed by a rotation. As a consequence, this will influence the restraint system ride down performance.

Overall, it can be noted that all the injury values do not vary significantly between a motor reflex delay of 30ms and 120ms, as results are comparable in magnitude in Figure 21 to Figure 25, considering the same seat friction parameter. This suggests that a "very aware" and "aware" occupant will withstand comparable levels of injuries on a secondary impact after an unexpected 1g pre-braking.

In all cases the occupant is, after an unexpected pre-braking phase, aligned with the airbag system before the subsequent impact occurs. As a consequence, in this safety scenario and configuration, it can be seen that for a standard 2 hand grip stance the system analysed in this study provides a good level of protection, bearing in mind that the airbag utilised meets OoP1 and OoP2 for a 5<sup>th</sup> percentile female as well as for a 50<sup>th</sup> percentile human model.

Considering all the variables (seat friction, reflex delay and braking duration) in this injury study, it can be concluded that the active human model is, in a 2 hand steering wheel grip, very stable in all conditions and that its injury responses are plausible.

### Results for the mobile phone stance

Following the kinematics study, the occupants' injuries are calculated up to a pre-braking duration of 1.1s, as it is judged that afterwards the occupant will miss the airbag, as illustrated in Figure 30 and Figure 31.

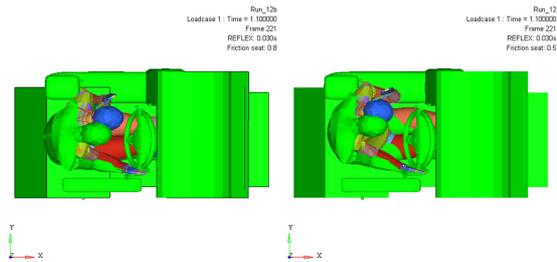


Figure 30. Comparison of occupant kinematics for seat friction 0.5 left, 0.8 right (30ms and 120ms awareness displayed in blue and green respectively). Pre-braking duration of 1.1s.

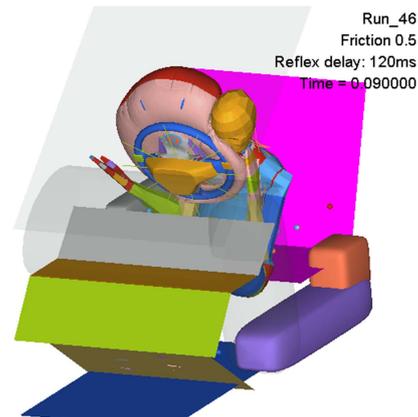


Figure 31. Impact with steering wheel at 90ms. Friction 0.5, Delay reflex 120ms.

Injuries are plotted in Figure 32 to Figure 37, focussing on cases with a seat friction coefficient of 0.5 and 0.8.

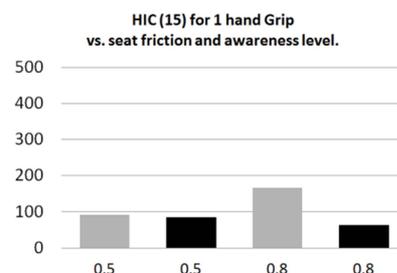


Figure 32. HIC (15) for 1 hand Steering Wheel Grip vs. seat friction. Reflex 30ms (light), 120ms (dark).

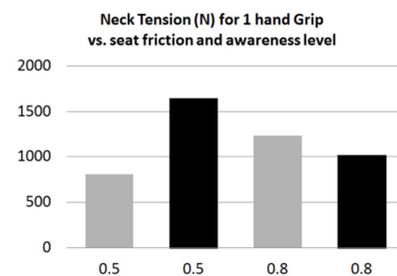


Figure 33. Neck Tension for 1 hand Steering Wheel Grip vs. seat friction. Reflex 30ms (light), 120ms (dark).

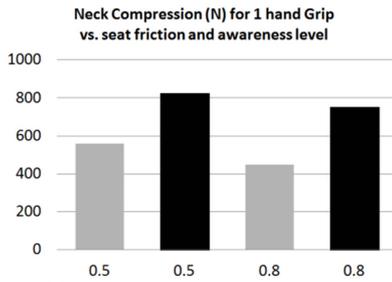


Figure 34. Neck Compression for 1 hand Steering Wheel Grip vs. seat friction. Reflex 30ms (light), 120ms (dark).

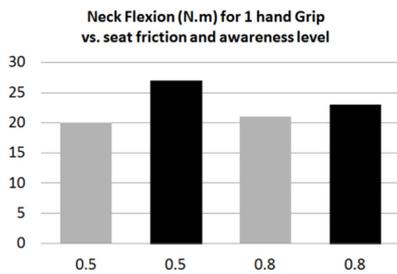


Figure 35. Neck Flexion for 1 hand Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

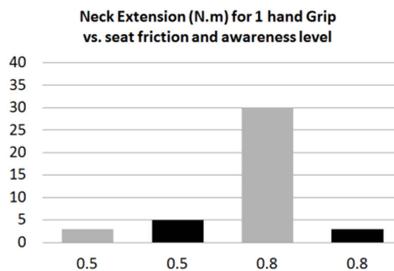


Figure 36. Neck Extension for 1 hand Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

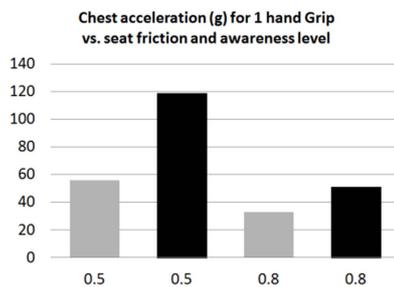


Figure 37. Chest Acceleration for 1 hand Steering Wheel Grip vs. braking duration. Reflex 30ms (light), 120ms (dark).

It can be seen that for a seat friction value of 0.5 that the reflex delay has a greater influence on the injury levels, as the posture of the occupant is away from the airbag zone of influence (Figure 30).

Observing the occupant position at 1.1s from

Figure 14 and Figure 15, it can be noted that the occupant relationship relative to the airbag and steering wheel assembly is less favourable to protect the driver before the airbag triggers than in the case of Figure 8 and Figure 9.

In a 1 hand steering wheel grip, the torso and the head are in line with the lateral steering wheel rim, which as a consequence leads to considerably higher chest acceleration (119g), as illustrated in Figure 31. It has to be noted that in all the models, the steering wheel is mounted to a rigid bracket, with no steering column collapse possible. As a consequence, the values quoted are comparative values and not absolute ones.

It has to be can be noted that head injury criteria as well as the Nij (Neck Injury Criteria) values met their legal limit in all cases. This suggests that the airbag adequately protects the occupant's head, but not sufficiently to stop the thorax hitting the steering wheel rim.

For a friction level of 0.8, results suggest that there is not a clear injury level pattern between the 2 levels of reflex delays. This maybe be caused by the fact that the occupant movement is a forward and lateral motion which causes a different interaction with the airbag than what it is designed to do, i.e. protecting for a forward motion. Nevertheless, in this study, for a friction level of 0.8, injury values are all under the legal requirements.

Overall, it seems that seats with higher friction tend to keep the unbelted occupants using a 1 hand steering wheel grip more aligned with the airbag.

Considering all the variables (seat friction, reflex delay and braking duration) in this injury study, it can be concluded that the active human model is, in a 1 hand steering wheel grip, very stable in all conditions and that its injury responses are also plausible.

## CONCLUSIONS

The study has initially correlated the Madymo Active Human Model to sled tests involving volunteers wearing a lap-belt and then successfully applied it in an active safety scenario. This scenario involved an unexpected vehicle pre-braking phase of 1g with an unbelted occupant followed by a subsequent 25mph rigid wall impact.

Overall, the study concludes that the Madymo Active Human Model used in this research provides believable kinematics and injury response behaviours. This model is very stable and has responded in a plausible manner when numerous variables, like seat friction, reflex delay and braking duration, were introduced.

The research has suggested that, within the active safety scenario investigated, the occupant's kinematics depend on the seat friction coefficient, arms' kinematics and the level of awareness.

By and large, it has been observed that for a reflex delayed response of less than 120ms that chest, neck and head injuries values for gripping the steering wheel with 2 hands were comparable for a given value of seat friction. Alternatively, occupants with 1 hand on the steering wheel (holding a mobile phone for example) were out of the airbag deployment zone in this research scenario after 1.1s of extreme braking regardless of their state of awareness and seat friction value.

## FURTHER WORK

This next step of this study would be to perform the same investigation with a more defined vehicle interior including refined seat as well as adding a collapsible steering column in order to explore restraint system design in active safety scenarios.

The kinematics study could also in the future be replicated with a FCW scenario which would then generated occupant neck and legs bracing behaviour. This study would be possible when a new Active Human model includes these needed leg, spine and arms bracing features.

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