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A Study Into The Kinematic Response For Unbelted Human Occupants During Emergency Braking

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Abstract: Since January 2014 more vehicles are being fitted with new active safety systems facilitating vehicle collision avoidance or mitigation by reducing the kinetic energy prior to impact, the most common being Autonomous Emergency Braking systems (AEB). Although beneficial in reducing crash energy these safety features may influence the occupant's posture within the cabin, and require special attention for the design and development of restraints system should the occupants be out of position or unbelted. This paper reviews the current reference volunteers' physical test, proposed by Ejima, used to validate the unbelted kinematics of active human computer model and proposes a new set of generic biomechanical kinematic responses based on OM4IS sled test data, which are judged more representative to an unbelted occupant's reaction. New kinematics corridors for head and torso angular change in a typical 1'g' frontal scenario are provided.

Keywords: Reflex; Kinematics, Biomechanics; Occupant protection; Active Safety; Human Modeling; Extreme braking

1.0 Introduction

Passive safety has for many years reduced the number of fatalities on the roads, however, its effect on occupants' safety has now stabilised [1], meaning that new safety features, like active safety are needed to reduce further the number of causalities. These active safety features vary from Autonomous Emergency Braking (AEB), stopping the vehicle before collision against another vehicle or a pedestrian, to alternatively automatically generating swerving manoeuvres, reducing the collision speed (kinetic energy) or avoiding the collision itself (zero energy). The remaining reduced energy (during the pre-crash phase) has to be managed by load case optimized (stiffness & shape) passive safety restraint systems (crash phase). These vehicle manoeuvres could have an influence on the occupant's posture within the cabin which could subsequently be cause for concern as the occupant is Out of Position (OoP) [2]. From research undertaken in the PRISM project [3][4], it has been reported that Hybrid III dummies and human volunteers behaved differently under low 'g' braking with respect to the kinematics, leading to the conclusion that only human body models appropriately describe the occupant's motion during the pre-crash phase of a vehicle fitted with active safety features. Further studies [5][6] have subsequently shown, that in a low 'g' braking scenario, human body model kinematics were the same prior to contacting the dashboard regardless of the occupant's driving posture, hence leading to the conclusion that occupants' bracing behaviour, not being considered in commercial simulation software, was necessary. Some studies into occupant numerical behaviour undertaken in side impact [7] have also clearly shown the influence of occupant muscle activity in low 'g' events, and the importance of modelling this biomechanical feature in order to better predict occupant injuries and position within the cabin [7]. In order to support this endeavour, biomechanical kinematics corridors or targets need to be derived, which is the aim of this paper.

In the unbelted loadcase, the worst kinematic scenario considered in this paper, it was noticed that the occupants kinematics were strongly influenced by the seat friction [6] which was preventing any pelvis forward motion during the first second of the pre-braking stage [7]. After 1.0s, the occupant's arms are taking the main load while the pelvis is moving forward. This suggests that a volunteer test using a lap belt to restrain the pelvis for the first second would be totally reasonable to assess an unbelted condition whilst protecting the volunteer from any potential dangerous excursions [7]. It has been also reported that occupants could undergo serious airbag injuries when positioned too close to the airbag module [9], usually less than 250

mm, or when the airbag was not present, disabled or out of reach [10], hence the importance of this research.
Looking at the current population and vehicle types, it can be observed [11] that occupants of 5th percentile anthropometry are already within the airbag stipulated danger zone. In the case of the 50th unbelted percentile occupants, the sternum is 120 mm away from the stipulated danger zone, which may be reached should the vehicle be subjected to an emergency braking.
The reference dataset which is currently used for the development of future active human models, like the THUMS Finite Element based human model, [12][13] in an unbelted scenario is based on the work of Ejima based on "*five 22 to 26 year-old male volunteers in good health*" having participated in the series of frontal braking experiments and set in a lap belt environment [14]. During the Ejima tests, the volunteers were supposed to stay fully relaxed during the 0.8 g deceleration scenario, as displayed in *Figure 1*.

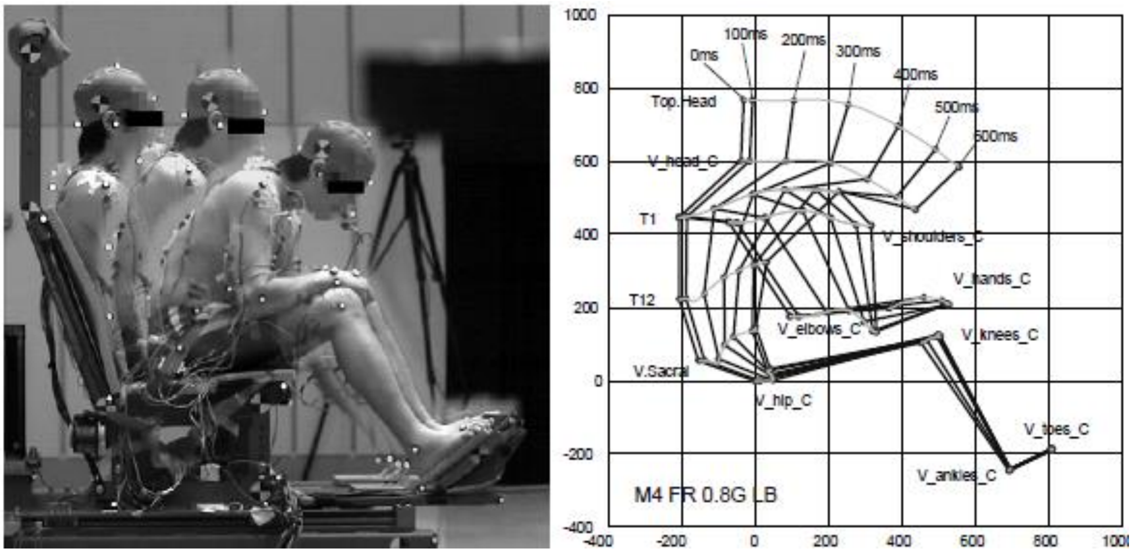

Figure 1: Physical motions from the 3D motion capturing system (Male, 0.8G: Relaxed) [14]

The test performed does not represent what happens in the real world, as the volunteers would in real life tense, as displayed in *Figure 2*. It has been documented that in an emergency braking scenario [4] occupants' arms moved forward to prevent contacts with the vehicle interior, which is contradictory to the standard scenario setup displayed in *Figure 1*.


Figure 2: Pictures of passengers bracing when subjected to pre-braking [4]

This paper will compare both test kinematic results and propose a new set of biomechanical kinematics corridors for low 'g' frontal human physical based on a OM4IS (Occupant Model For Integrated Safety) consortium dataset [18][19][20].

2.0 Method

The study presented here was based on the information gathered from the OM4IS consortium, which had undertaken sled tests under low 'g' deceleration conditions [19][20]. The tests involved a number of volunteers, from whom 7 complete data sets were extracted, examining volunteers' joint kinematics extracted from an 8-camera Vicon motion-tracking system [21].
The OM4IS tests undertaken with 7 male volunteers [19][20], corresponding to a 50th percentile human population range (average height 179 cm (+/-5 cm) and mass 74 kg (+/-4 kg)), focused on frontal and lateral sled tests. The sled test manoeuvres represented typical lane change swerving manoeuvres and emergency braking scenarios [3][4] and were converted into average displacement/time functions [8](maximum acceleration level of 0.90 g in frontal and 0.55 g in lateral load directions). The data provided by Prüggler to the author (Bastien) detailed in this paper were based on 4 volunteers with 3 or 4 repeat tests subjected to each to a frontal braking scenario [20]. The volunteer is not made aware when the braking deceleration is engaged and his kinematics behaviour, which includes a reflex behaviour, is captured. In this OM4IS test, the occupant had to hold straight a light object horizontally, aiming to replicate some of the effects of holding the steering wheel. Each test recorded 70 location points using motion capture (*Figure 3*).

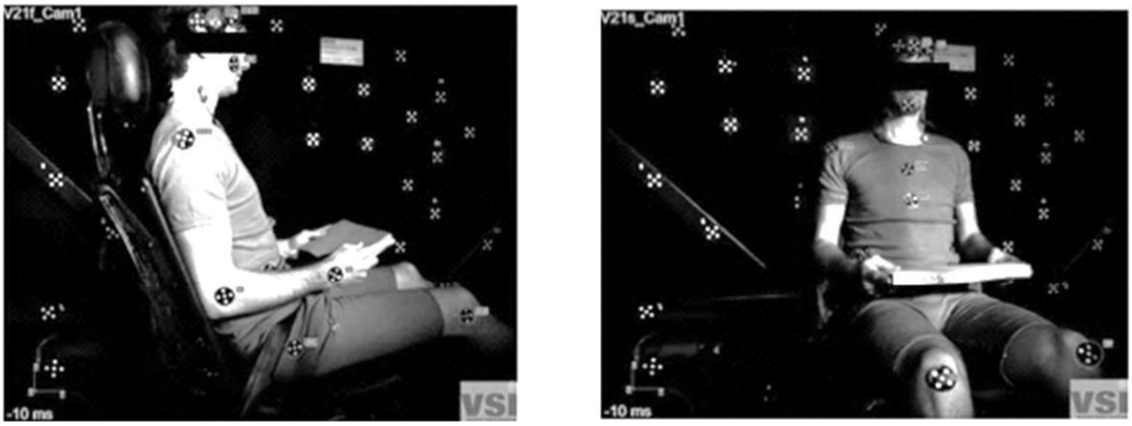

Figure 3: Typical posture setup for OM4IS sled tests [20]

This test setup varies greatly from previous research, as the occupant will tense due to muscle reflex compared to staying relaxed during the whole duration of the emergency braking [14].
In order to capture the volunteers' kinematics, it was decided to investigate the relative angle change of the head and the thorax using the marker points listed in *Table 1* with the point locations displayed in *Figure 4*.

<table>
<tr><th>Member</th><th>Marker Label</th><th>Marker Name</th><th>Member</th><th>Marker Label</th><th>Marker Name</th></tr>
<tr><td rowspan="4">Head</td><td>LFHD</td><td>Left Front HeaD</td><td rowspan="4">Torso</td><td>LBAK</td><td>Left upper BAcK</td></tr>
<tr><td>LBHD</td><td>Left Back HeaD</td><td>RBAK</td><td>Right upper BAcK</td></tr>
<tr><td>RFHD</td><td>Right Front HeaD</td><td>LPEC</td><td>Left PECtoralis</td></tr>
<tr><td>RBHD</td><td>Right Back HeaD</td><td>RPEC</td><td>Right PECtoralis</td></tr>
</table>

Table 1: Relevant Markers on Subjects on Head and Torso

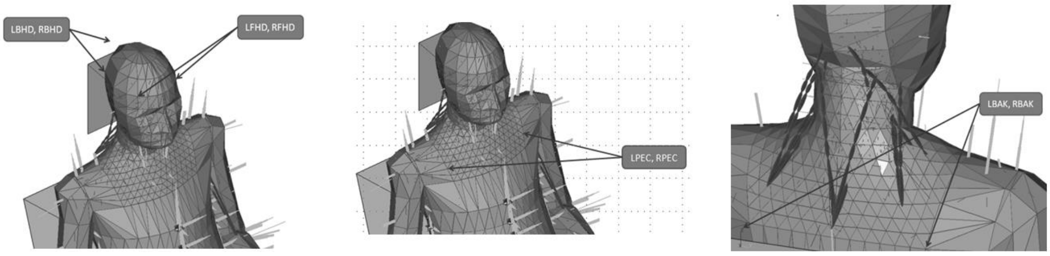

Figure 4: Location of key Vicon points on the occupant head and torso [16]

3.0 Test Results

Signal Nomenclature

The relevant marker coordinate output from the OM4IS test are plotted against time and are provided in the Appendix (Figure 14 to Figure 21). The output relates to 4 volunteers which were subjected to repeat tests. The volunteers are code named Proband1, Proband2, Proband3 and Proband4, and the repeat test indices are labelled "*_Front XX dyn*", where *XX* is the test number.

Signal Analysis

It can be observed that, in the channel output of RBHD Z, which relate to the Z vertical output from the marker RBHD, the channel "Proband04_front_02dyn" looks very linear for 1 s, which is suggestive of a faulty channel. As a consequence, the volunteer response "Proband04_front_02dyn" will be ignored from now on in the study [15]. The same action is taken for the RBHD X and Z channels where the following head signals "Proband1_Front 02 dyn", "Proband1_Front 03 dyn", "Proband2_Front 03 dyn" and "Proband3_Front 01 dyn" were incomplete. Consequently, only nine complete sets of data will be considered in our study.
It can be noticed and already documented that there is noticeable "*kinematics variability"* within the same percentile [20]. This comment is confirmed by the test results available in the Appendix. As such, it may be suggested that the conditions for each volunteer may not have been equal, for example pulse, belt forces and friction parameters may have been different. Maybe the physiological state of each volunteer was different, i.e. maybe they were stressed or maybe relaxed.
A typical start and maximum posture position represent the type of kinematics recorded in the OM4IS tests (*Figure 5*). It has to be noted that the camera was not mounted directly on the sled.
It can be observed in *Figure 5* that the occupant tends to keep his line of sight horizontal during the duration of the test, as well as creating tension in his arms, like tension when holding a steering wheel, which is very different from the Ejima test displayed in *Figure 1*.

Figure 5: A typical OM4IS frontal sled test [20]

4.0 Test Results Analysis and Comparison with Ejima's tests

These OM4IS test results can be compared to the results previous obtained in by Ejima. Some markers have been selected such that their location on the volunteers (*Figure 4*) was comparable and listed in Table 2.

OM4IS		Ejima	
Marker name	Description	Marker name	Description
C7	Base of cervical spine	T1	Top of thoracic spine
RSHO	Right Shoulder	Shoulder	Shoulder
RFHD	Right head temple	Head Top	Top of head

Table 2: Marker position comparison between OM4IS and Ejima

From the kinematic plots extracted from these 2 independent series of tests (*Figure 6, Figure 7* and *Figure 8*) it can be noted that the response trend tends to be comparable, except that in the OM4IS test, the volunteer's motion is less pronounced. It has to be noted that the Ejima's sled test pulse was not provided in the literature; hence some differences may be present between the 2 series of tests. It is assumed in this paper that the Ejima pulse to be representative of a typical emergency braking scenario and comparable to the OM4IS one. This suggests that the results from OM4IS are credible, as they have some similarities with the work conducted by Ejima. Ejima's occupant kinematic results values were evaluated graphically from *Figure 1* and offset in Z and X were added to align the occupant's initial position with the OM4IS test data as the reference for each test was different.

In the Ejima tests, the occupant is more in a crouched position at 600 ms (*Figure 1*) whereas in the OM4IS tests the volunteer is still very straight (*Figure 5*), in spite of the fact that in both tests only a lap-belt is used.

This is clearly showing that the 2 test setups and conditions are very different, relaxed against keeping a determined posture. The outcomes are different; even if the overall shape of the kinematics is comparable, the Ejima tests present an exaggerated motion (*Figure 6, Figure 7* and *Figure 8*), particularly for the head excursions. The head excursion between tests varies from 183 mm to 513 mm between the tests, which would suggest that, should Ejima's test aim to replicate a 'sleeping' passenger compared to an awake and alert one, the Ejima tests under-predict the occupants kinematics behaviour (*Table 3*).

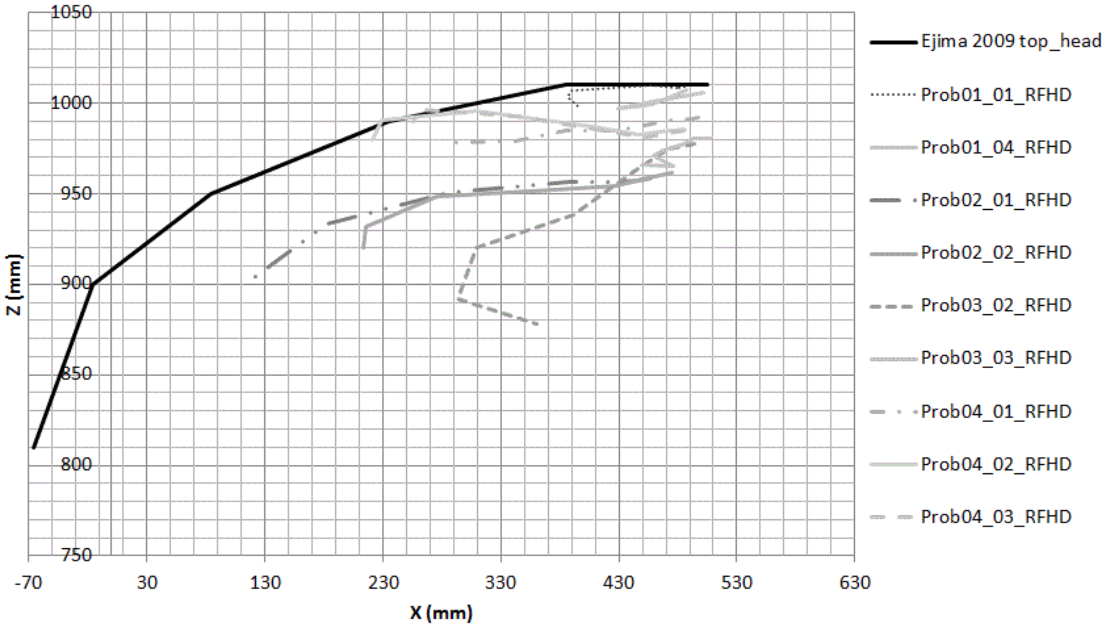

Figure 6: Comparison between Ejima and OM4IS head motions (based on the Right Front Head Marker) [16]

Tests	X coordinate (mm)	Excursion OM4IS vs Ejima (mm)
Ejima test	-64	N/A
OM4IS Prob_01_01_RFHD	381	445
OM4IS Prob_01_04_RFHD	430	494
OM4IS Prob_02_01_RFHD	119	183
OM4IS Prob_02_02_RFHD	210	274
OM4IS Prob_03_02_RFHD	289	353
OM4IS Prob_03_03_RFHD	449	513
OM4IS Prob_04_01_RFHD	285	349
OM4IS Prob_04_02_RFHD	218	282
OM4IS Prob_04_03_RFHD	247	311
	MAXIMUM Head Excursion (mm)	513
	MINIMUM Head Excursion (mm)	183

Table 3: Difference of head excursion between Ejima and OM4IS tests

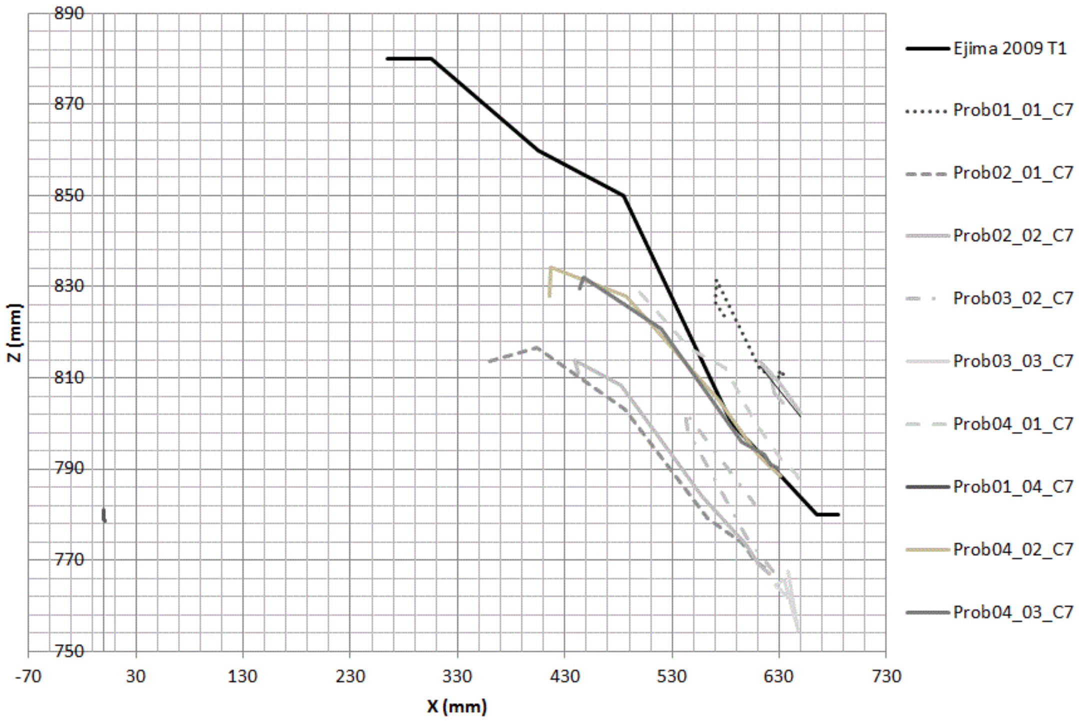

Figure 7: Comparison between Ejima and OM4IS base of neck motions (based on the C7 cervical) [16]

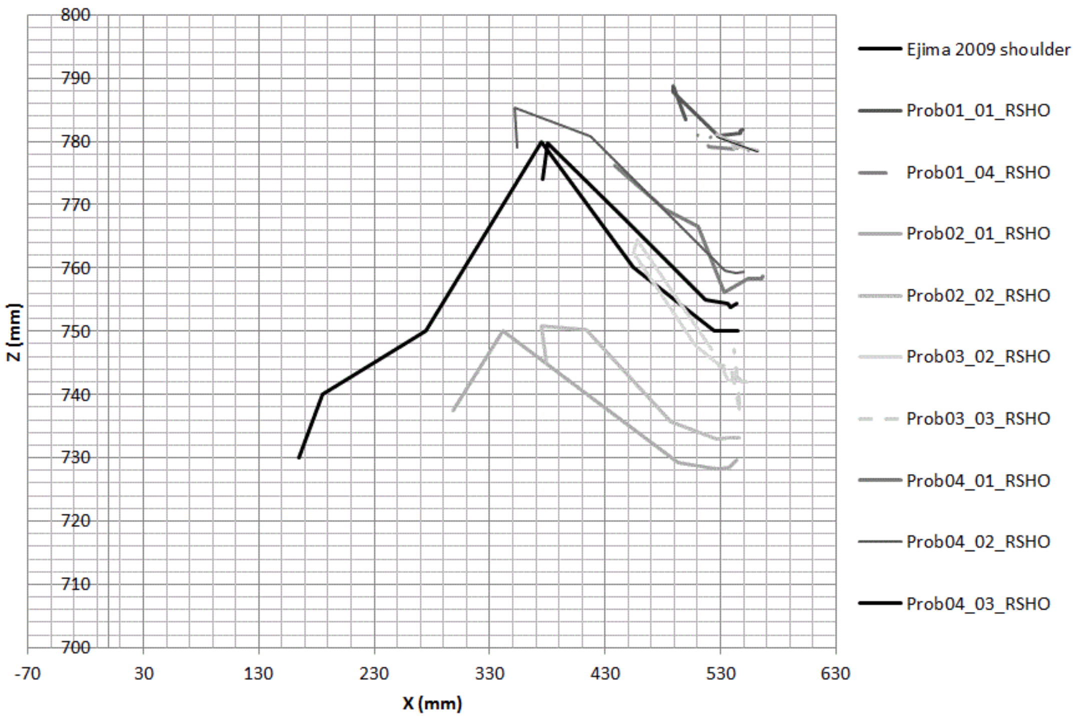

Figure 8: Comparison between Ejima and OM4IS shoulders motion (based on the Right Shoulder marker) [16]

Using the comparison graphs (*Figure 6*, *Figure 7* and *Figure 8*), it can be noted that each volunteer's kinematics usually tends to stay within a defined envelope, i.e. there is little scatter of kinematics for the same volunteer tested, which reinforces the issue that each human beings can behave differently from another one. Looking at the shoulder and the lower neck cervical, it can be noted that if the coordinate values are different between the volunteers, their slope is similar. There seem to be more differences in the head motion, where the head coordinate changes and the slope between the volunteers are different. This is suggesting that the difference comes from the neck stiffness from C7 to C1 and the difference of head masses (which have not been measured).

5.0 Derivation of the Head and Torso kinematics median response

Creating kinematics reflex corridors is a challenge as there can be a scatter between the coordinates of each marker on the volunteers' bodies. As such, matching the coordinates between each volunteer and performing comparison will be problematic, as body marker locations need to match exactly between volunteers.

The approach undertaken is to consider a relative change of the head and the torso angles as a more universal measure of understanding the difference in kinematics between volunteers, as markers' coordinates are volunteer specific, due to their different anthropometry, their relative head and torso angle variations are not [8]. The approach is based from the observations from the kinematics traces in *Figure 6, Figure 7* and *Figure 8.*

This approach is suitable, because the occupant is attached to the sled about the pelvis, which suggests that the torso will rotate about the hip and that the position of the feet and legs will remain unchanged during the low 'g' test event. Had the volunteer been unbelted (fully unrestrained), this approach would not have been adequate, as there may be some relative movements due to a sliding motion. The sliding would have been small, nevertheless present [5] [6].

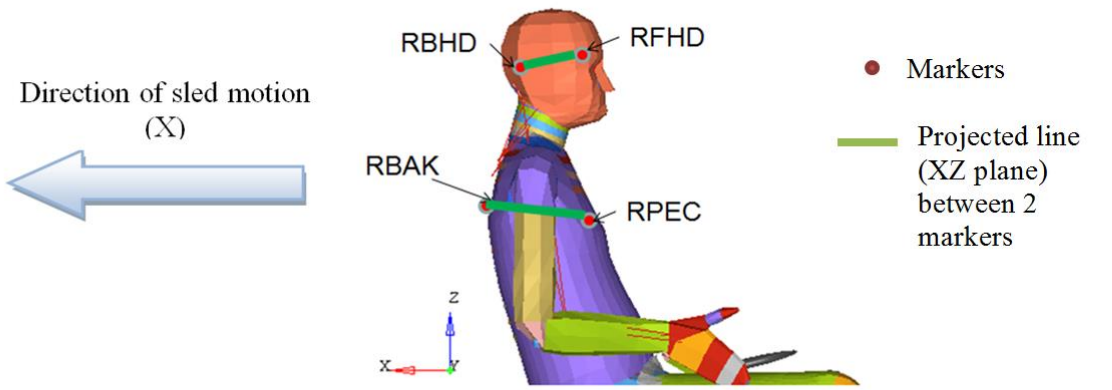

Figure 9: Markers for frontal load case and projection in XZ plane [16]

It was noticed that the sled motion was aligned with the volunteers' sagittal plane, where biomechanics extension and flexion motions occur. Using the sagittal plane's geometrical properties, it is suggested that either side of this plane would provide a similar kinematics response, hence that the projection of the markers from 1 side of the volunteer (left or right) onto the sagittal plane would be representative of the motion of both sides of the volunteer (*Figure 9*). Consequently, RBHD, RFHD, BAK and RPEC's y coordinates are assumed to be 0 (projection) and the relative angle change is computed from *Equation 1*.

$$\theta^{\circ}_{frt}(t) = \tan^{-1}\left(\frac{\Delta z(t)}{\Delta x(t)}\right) \times \frac{180}{\pi} - \theta^{\circ}_{frt}(0)$$

Equation 1: Derivation of frontal angle calculation as a function of time (for head and torso angle computation)

Using this methodology for the markers located in *Figure 9* using the frontal test information from the OM4IS tests, a set of responses curves were calculated as well as the target curve computed using the median value of the tests (*Figure 10* and *Figure 11*).

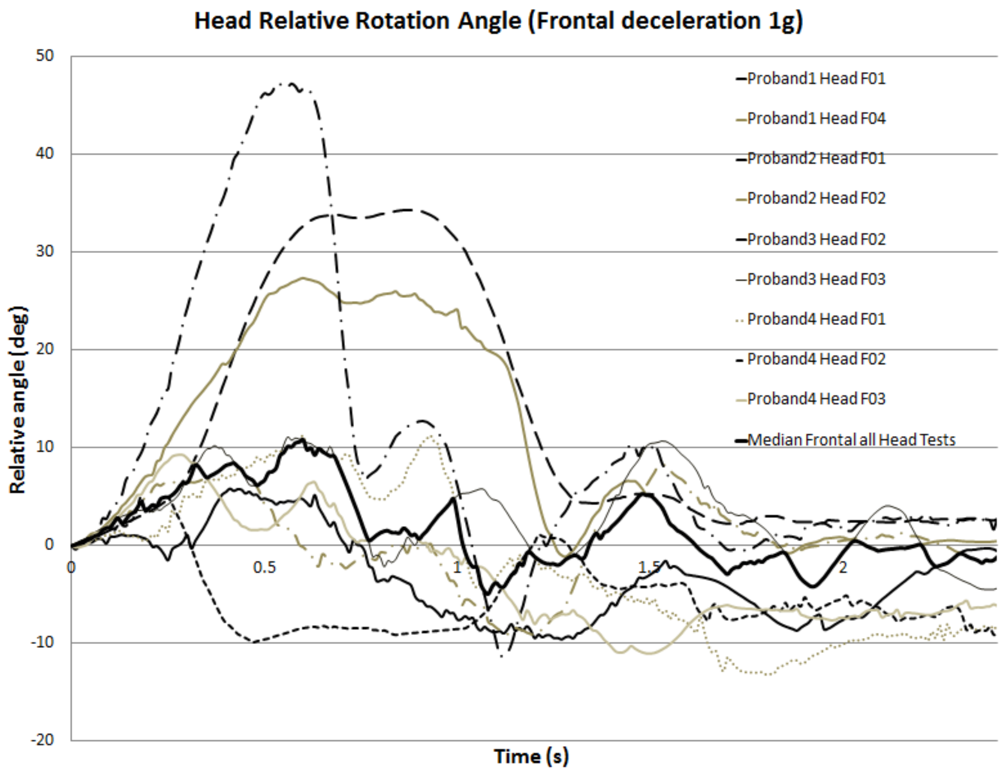

Figure 10: Head frontal motion relative angle for all 9 responses

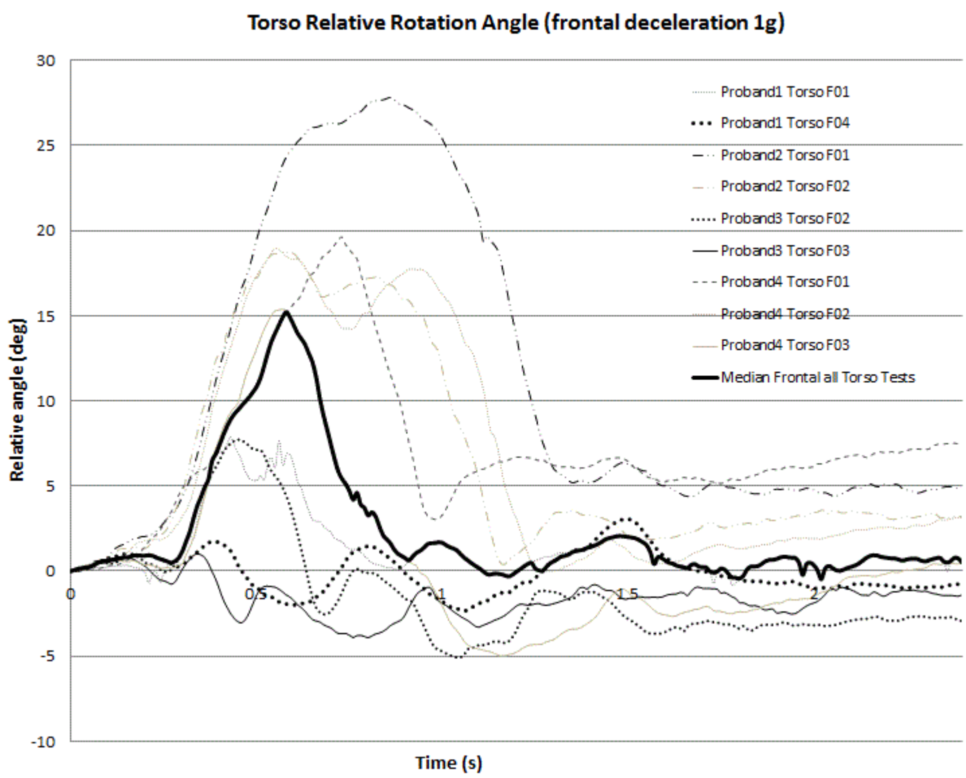

Figure 11: Torso frontal motion relative angle for all 9 responses

From *Figure 10* and *Figure 11*, it can be observed that there is a very large spread of responses amongst the 4 volunteers (Proband1 to Proband4), where "Head Fxx" and "Torso Fxx" relate respectively to a test number 'xx' with head and torso motions outputs. The positive sign indicates a downward rotation. The maximum relative head rotation is reached at around 0.5 s with a value of 47 degrees, while Ejima reaches 70 degrees in 0.6 s.
The OM4IS median of the relative head rotation is around 10 degrees which is indicating that the occupant tries to keep the line of vision unchanged as before the test, which is also observed in the PRISM tests (*Figure 5*) in real-life scenarios.

6.0 Derivation of the Head and Torso kinematics bracing level response

The aim of this section is to extend the findings from the head and torso median responses in order to extrapolate the behaviour of the occupant when reacting in a pre-braking scenario and propose a set of responses for when the occupant produces a faster or slower reflex response.
An initial approach would be to cluster the data according to their position above or below the median [15][16][22]. This proposal would be suitable if more samples were available either side of the median. As the test sample is small (9 complete samples for the head and the torso respectively) and that in some cases responses may have been influenced by the conditioning of earlier experiences, i.e. repeat tests, this suggests that the test results distribution for this small number of tests may be skewed. Consequently, the approach using the median is more appropriate for such number of samples. It is however reasonable to expect for an infinite (or very large) number of tests the distribution to be Gaussian and centred. This can already be observed in Figure 12 and Figure 13 in which the mean (or average) and the median follow a similar trend, and this for just 9 samples, confirming the Gaussian centred distribution hypothesis.
In order to capture the spread of occupant motion responses and classify them according to their intensity, a standard deviation technique is used. This standard deviation evaluates the minimum and maximum response corridors representing a spread of motions observable by a population; the tighter the corridor, the tighter the range of amplitudes about the mean value.
The sign of the standard deviation will reflect an increase in motion, in the case of a positive standard deviation, and a reduction for a negative value of a standard deviation. The sign convention is aligned with the graphs in *Figure 12* and *Figure 13*, where any increase in head and thorax rotation is positive. Because of the small number of samples, the standard deviation techniques can be used to capture a range of occupants' reflex behaviour about the median, therefore representing an average human bracing at 1'g'. It is proposed to use 1 and 2 standard deviations (σ and 2σ) to capture the collection of motions, hence representing kinematics +/- 18.3% and +/- 45.5% about the median central position. Consequently, the median represents an average response, $+1\sigma$ represents a response with a slow reflex, $+2\sigma$ represents a response with a very slow reflex, -1σ represents a response with a fast reflex and -2σ a response with a very fast reflex.
Observing the target kinematic curves presented in *Figure 12* and *Figure 13*, it can be noted that the faster the reflex the smaller the amplitude of the occupant, which is to be expected. Currently, a Gaussian centred distribution is expected and as the mean and the median curves differ, it can be concluded that more data points are required to capture accurately the occupants motions. The proposed method suggests a robust and consistent methodology to extract a qualitative understanding of the occupants' kinematics envelope about the median; however it is understood that the corridors generated still need refining with more physical tests.

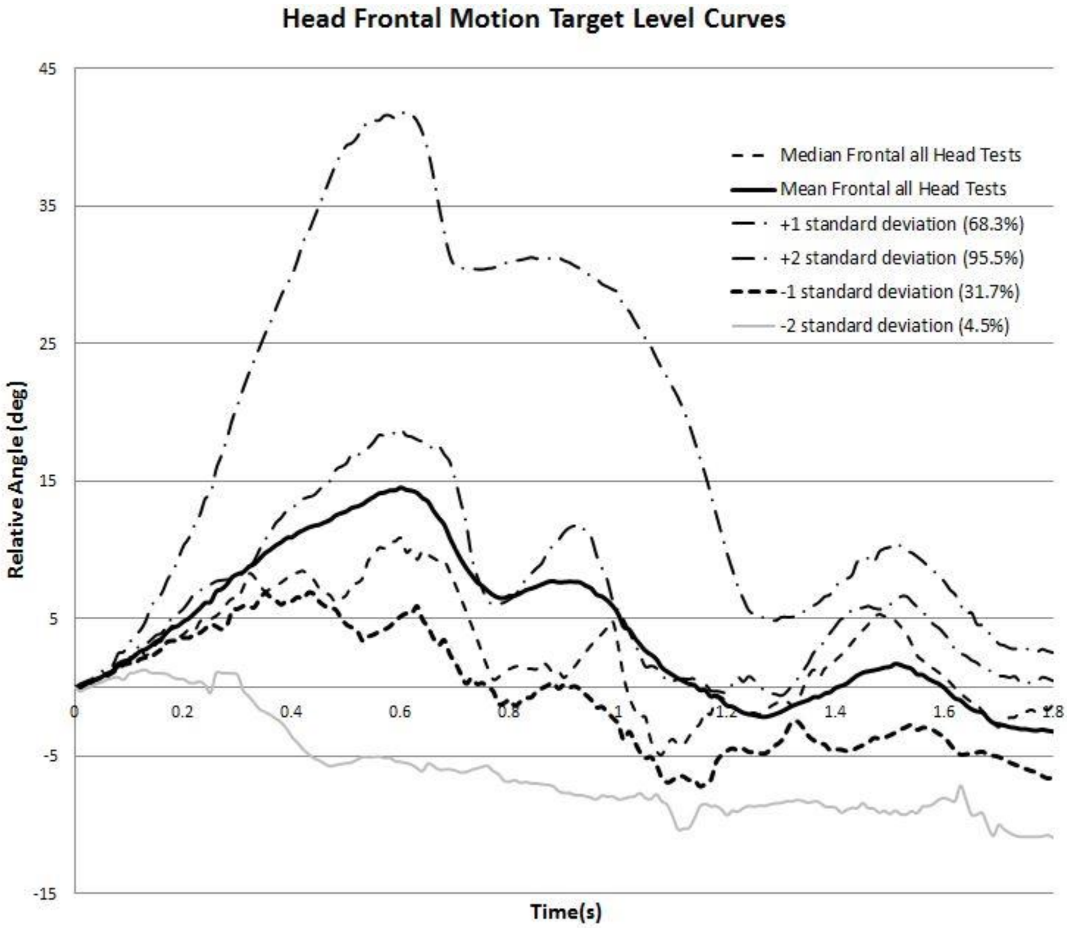

Figure 12: Head Relative Angles motion targets

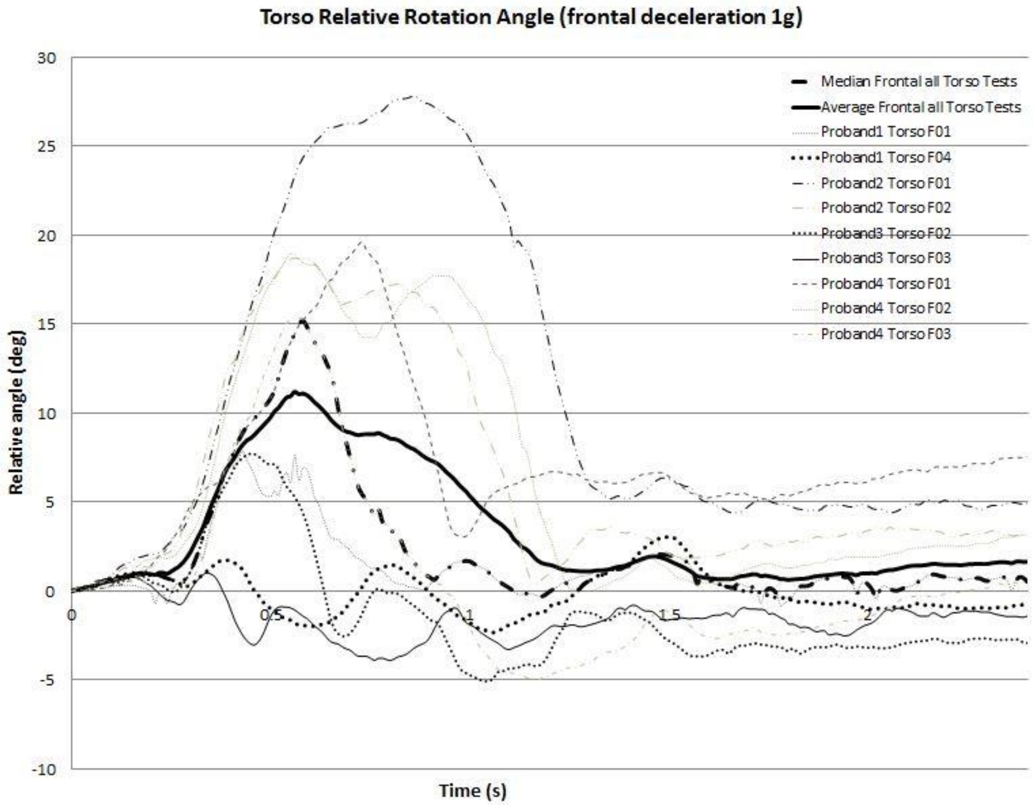

Figure 13: Torso Relative Angles motion targets

7.0 Discussion

The results suggest that the human response to tests have a large variability, even for the same volunteer as well as within the same percentile. Due to the small number of test samples, a median has been computed and has shown considerable difference compared to the slow and fast responses. The reflex only affects the timing for when the muscles are activated and does not influence the strength of the muscle activity, consequently using the median of these tests would not capture all potential responses, hence the proposed added standard deviation (σ).

The reflex component of a human comes from the sensing of the environment. When the sensing is activated, a message is sent to the brain for processing and then the muscles areas are activated with the following delays [23]. This signal takes 40 ms to reach the neck, 70 ms for the spine area, 70 ms for the arms and 100 ms for the legs.

This information is compatible with the results found in *Figure 12* and *Figure 13,* as the head and torso motions, measured in the OM4IS tests, are not modified before 0.2 s, in the fastest reflex response cases.

The speed of reflex response of a human depends on the cognitive input to the brain and as such active human models should have cognitive input features to automatically use the 5 response curves for the head and torso relative rotation angles.

While cognitive features are being considered in the new active human models [13], it would be beneficial in the meantime to consider the 5 proposed reflex responses and include them within unbelted scenarios to investigate future automotive safety systems, accident reconstruction situations as well as aircraft and bus/ coach safety as passengers are wearing a lapbelt.

8.0 Conclusions

This work builds on earlier work of Ejima by introducing a novel consideration of the influence of occupant muscle activation on occupant kinematics in an extreme braking scenario.

A series of sled tests from the OM4IS consortium were conducted which included the effect of holding a steering wheel, by requesting the volunteer to hold a light mass steady and keeping a posture as constant as possible during the deceleration duration of the test. The study has shown the importance of being able to determine the position of the occupant within a vehicle cabin in case of unbelted/ lap-belted scenarios. The difference of test setup between OM4IS and Ejima, based on 9 tests, indicated that the Ejima's test was under-predicting the head forward excursion between 183 mm and 513 mm.

Kinematic response curves have been computed based from 9 OM4IS tests, proposing 5 levels of reflex activation speed: very slow, slow, median, fast and very fast. These kinematics curves were approximated from standard deviation techniques based on the global set of test data, due to the small number of tests available as well as considering the volunteers' response variability. It is proposed to include these 5 reflex curves in the development of future active human models by either replicate 5 separate and discrete occupant's reflex behaviours or by coupling the proposed response curves to a sensory brain model.

9.0 LIMITATIONS AND FUTURE WORK

It is believed that the methodology described in this paper is suitable to generate kinematics curves for humans in frontal braking scenario. However it is evident that more data is needed to refine the proposed response curves, as 9 samples is not sufficient. Considering the kinematic response variability between subjects observed in this research, it is proposed to conduct more tests to refine further the head and torso relative angle target curves.

Should more data be available, it would be interesting then to compare a cluster approach to the global approach undertaken in this paper and test which method is more accurate.

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Appendix: OM4IS Frontal sled tests markers outputs.

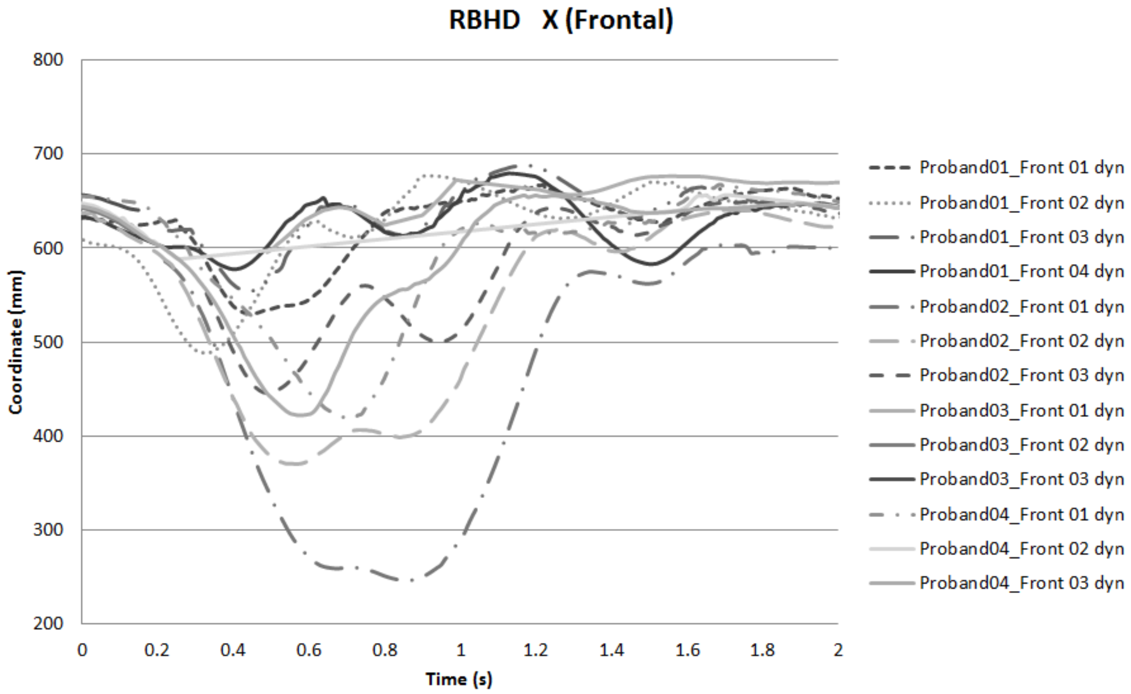

Figure 14: X Coordinates of RBHD (Right Back of Head) as function of time [16]

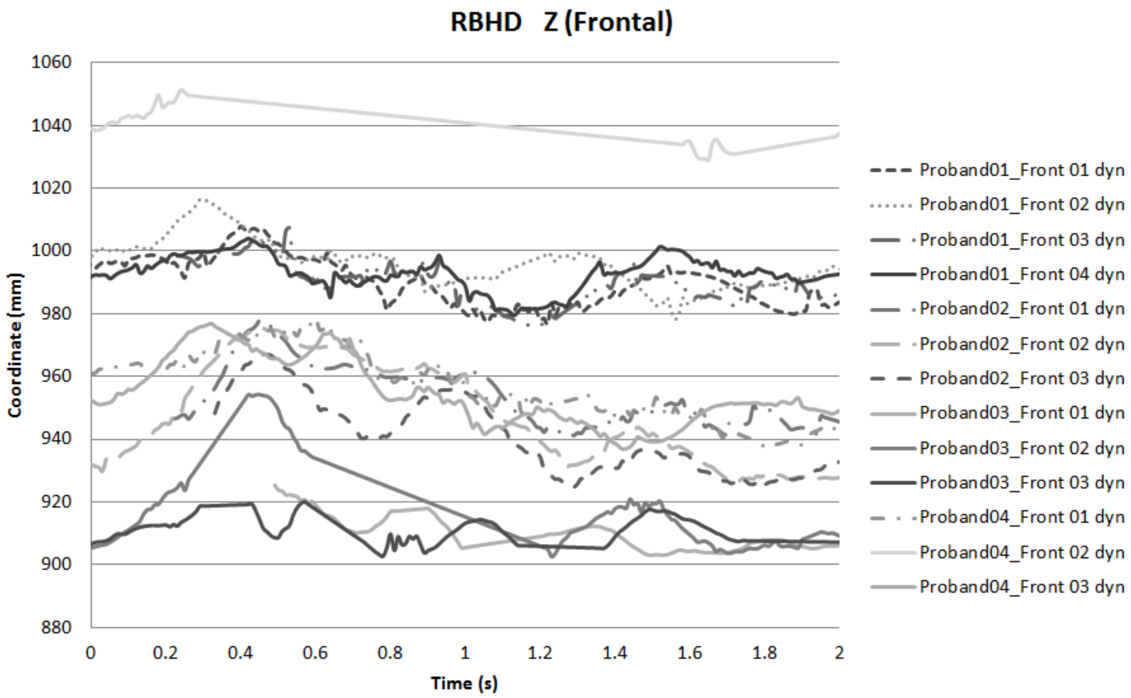

Figure 15: Z Coordinates of RBHD (Right Back of Head) as function of time [16]

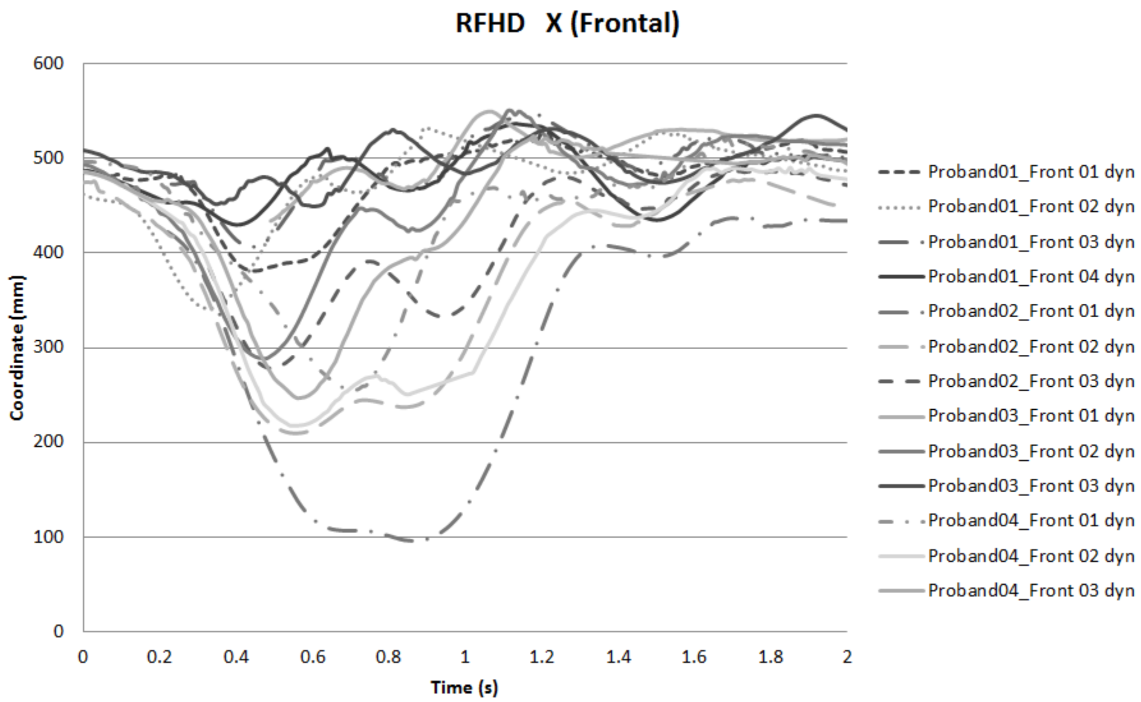

Figure 16: X Coordinates of RFHD (Right Front of Head) as function of time [16]

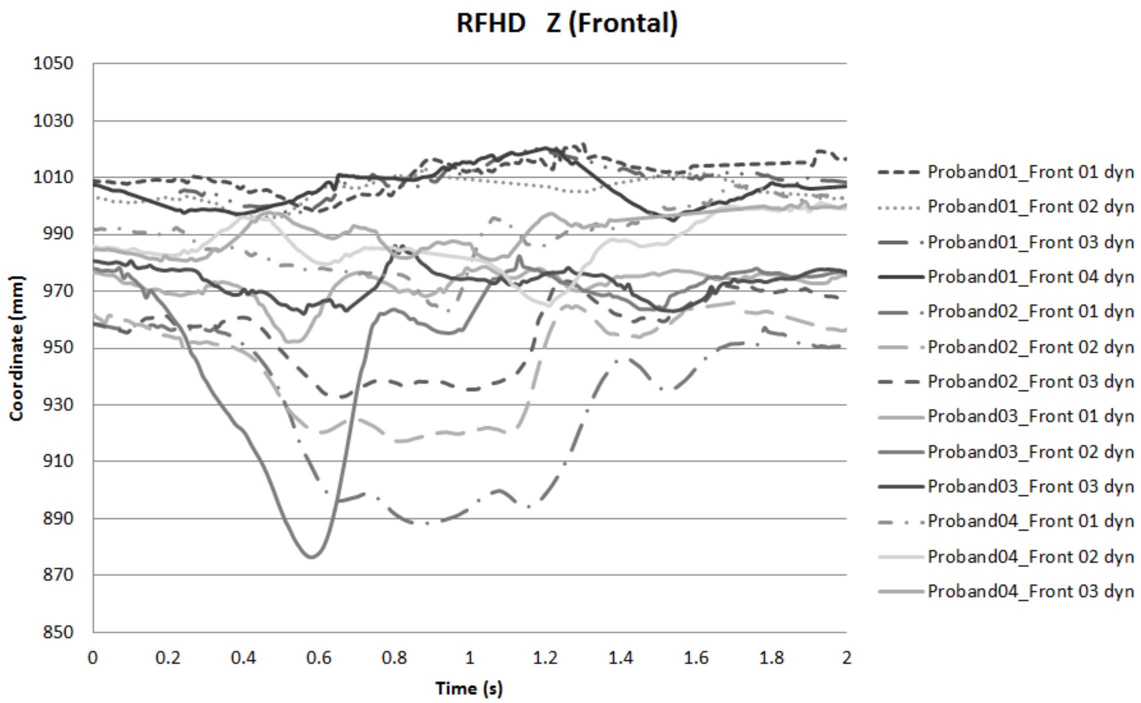

Figure 17: Z Coordinates of RFHD (Right Front of Head) as function of time [16]

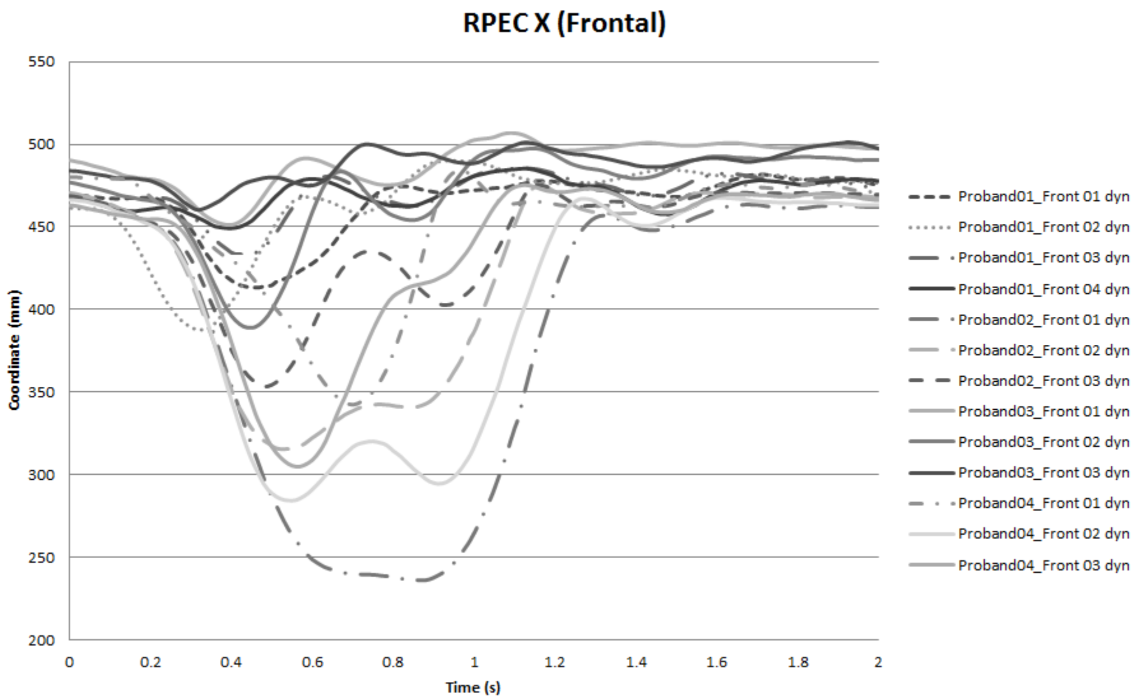

Figure 18: X Coordinates of RPEC (Right Pectoral) as function of time [16]

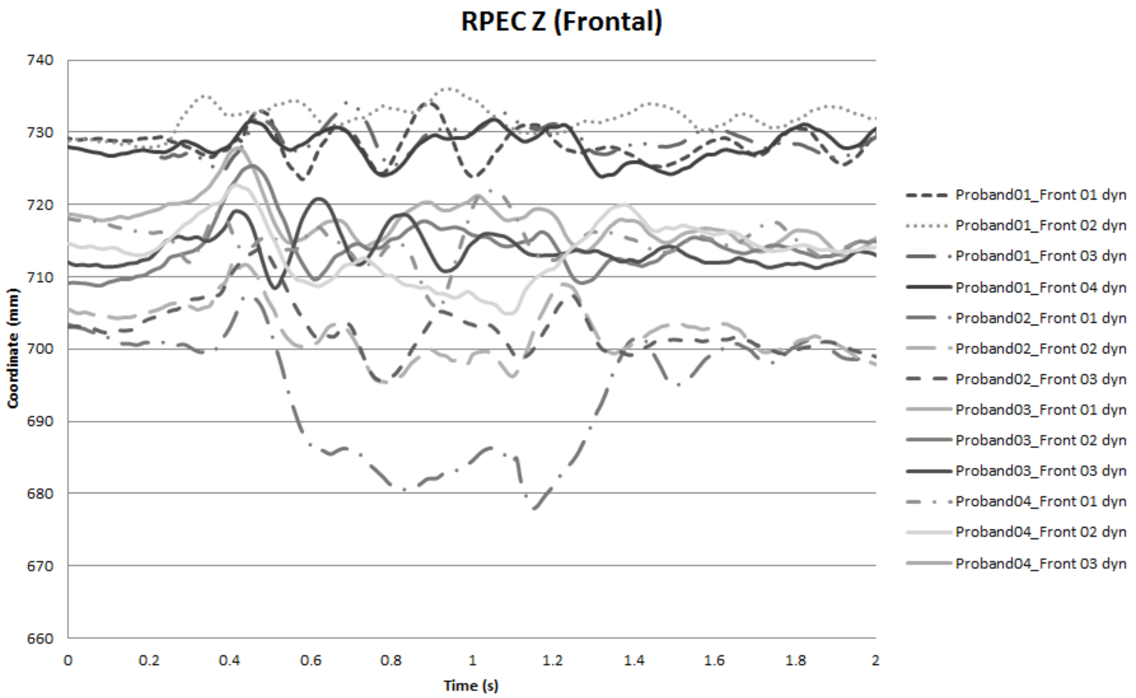

Figure 19: Z Coordinates of RPEC (Right Pectoral) as function of time [16]

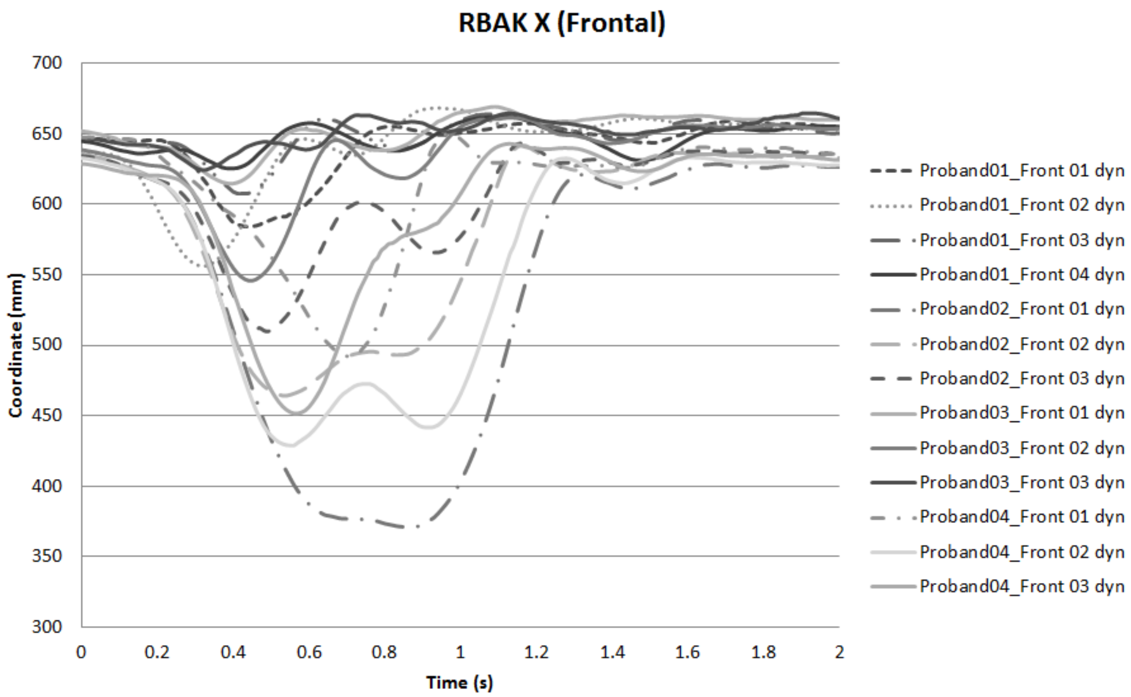

Figure 20: X Coordinates of RBAK (Right Back) as function of time [16]

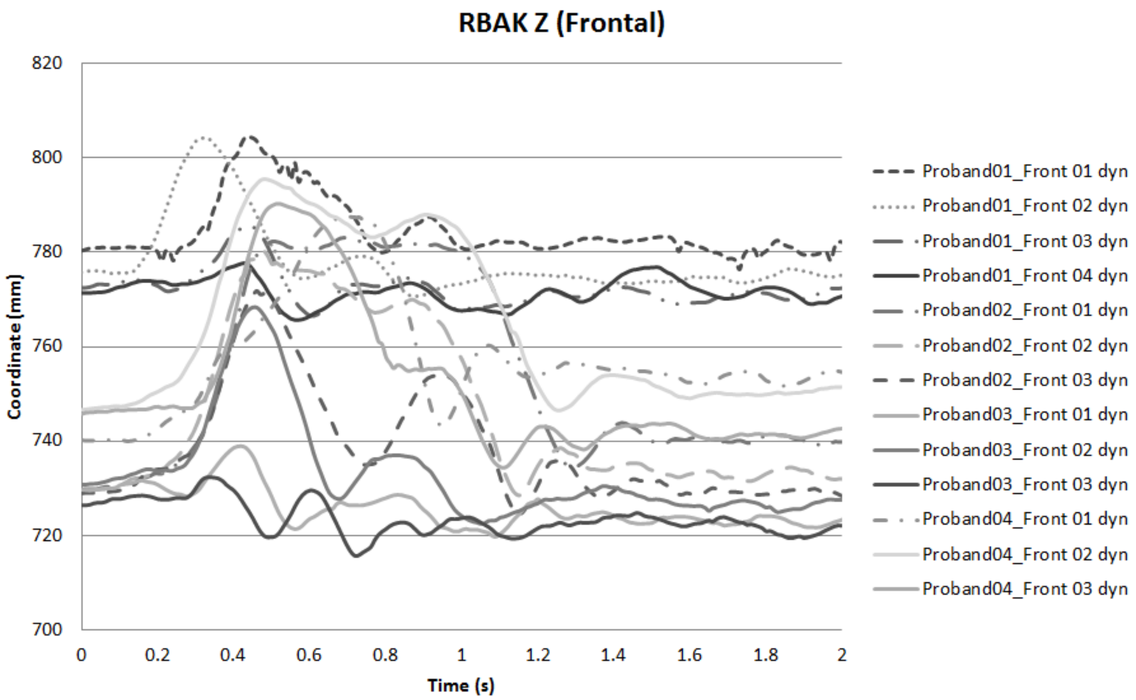

Figure 21: Z Coordinates of RBAK (Right Back) as function of time [16]