SPINAL KINEMATICS OF RESTRAINED OCCUPANTS IN FRONTAL IMPACTS

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ABSTRACT

This paper analyzes acceleration measured at different locations on the spine of a post-mortem human surrogate (PMHS) subjected to restraint loading in a frontal impact. The study was performed to assess the use of acceleration measured at the first thoracic vertebra (T1) in the development of thoracic injury criteria for use with anthropomorphic test dummies. This study has an experimental component involving sled tests with a PMHS and with a Hybrid III 50th percentile male dummy, and a computational component involving the MADYMO human body model and the MADYMO dummy model. Mass-scaled PMHS T1 acceleration is found to reflect changes in test conditions (airbag-only, belt-with-airbag, driver-side and passenger-side) differently than dummy chest cg acceleration. PMHS T8/9 acceleration-time history is found to reflect transient restraint loading better than T1 acceleration-time history, but PMHS T8/9 maximum acceleration, even scaled for differences in PMHS and dummy mass, is found to be significantly greater than dummy chest cg acceleration for some test conditions. Spinal curvature and differences between human and dummy spinal flexibility are shown to be significant factors in the interpretation of PMHS-based acceleration measurements, particularly for use as a proxy for dummy chest cg acceleration.

Key Words: spine kinematics, cadavers, dummies, frontal impacts, sled tests

ANTHROPOMORPHIC TEST DEVICES (ATDs), OR DUMMIES, are used to predict whether car occupants would suffer injury in a given crash condition. ATD parameters such as thoracic acceleration and deflection have been used as thoracic injury criteria, which are compared to injury threshold values. The threshold values are estimated using animal, human volunteer, and post-mortem human surrogate (PMHS) test results (Pike 1990).

The relationship between chest acceleration and injury has been studied since the early 1970s (Stapp 1970, Mertz and Gadd 1971). The current US Federal Motor Vehicle Safety Standard (FMVSS) 208 specifies that the 50th percentile male Hybrid III ATD thoracic center of gravity (cg) acceleration cannot exceed 60 g and that the chest deflection cannot exceed 75 mm in a defined frontal impact test. The US National Highway Traffic Safety Administration (NHTSA) is conducting an ongoing research and crash-test program that uses PMHS thoracic responses to evaluate ATD biofidelity and to develop injury criteria. NHTSA researchers Kuppa and Eppinger (1998) proposed a thoracic injury criterion derived from a linear combination of PMHS chest deflection and PMHS resultant spine acceleration recorded by instruments mounted to the first thoracic vertebra (T1). Eppinger, et al. (1999) then related this PMHS-based injury criterion to the chest deflection and chest acceleration measured on an ATD. Kent et al. (2001) determined that PMHS-based instrumentation introduces restraint-dependent bias into an injury criterion that is intended for use with an ATD. To ameliorate this bias, Kent used dummy measures to predict PMHS injury outcome in a nominally identical impact.

Recording PMHS thoracic acceleration at T1 and other sites on the periphery of the thorax was first described by Robbins, Melvin, and Stalnaker (1976). Measuring acceleration at T1 is technically easier than measuring other spinal sites because it avoids interference with chest deflection instrumentation (chest bands). Therefore, T1 has become the site most commonly instrumented and T1 resultant acceleration is the most commonly reported PMHS thoracic acceleration. We were

unable, however, to find published evidence that would suggest the exclusive use of T1 acceleration in the development of thoracic injury criteria.

Using the Kuppa-Eppinger ATD-based criterion to predict human (as represented by a PMHS) injury contains the inherent assumption that the ATD thoracic acceleration, commonly recorded at the chest cg, is reasonably approximated by PMHS T1 acceleration and that the trends in PMHS and ATD acceleration are consistent as test conditions are varied (e.g. when a belt is used versus when a belt is not used, when seating position is changed). Recent testing conducted at the University of Virginia Automobile Safety Laboratory (UVA) indicates that these assumptions should be examined. Substantial differences between mass-scaled PMHS spinal acceleration and ATD thoracic acceleration values have been observed in nominally identical collisions. Acceleration maxima, even when mass-scaled as suggested by Eppinger et al. (1984), have been as much as 26% higher at the PMHS T1 than at the ATD chest cg. Further, the acceleration vector typically has a different direction. Higher average T1 z-axis (vertical) accelerations have been observed for the PMHS. This paper investigates these experimental findings and presents a computational study performed to elucidate the effect of the restraint system on acceleration measured at different vertebrae. The methods used to record acceleration are analyzed, and the implications for the development and application of thoracic injury criteria are discussed.

EXPERIMENTAL EVALUATION OF PMHS AND DUMMY ACCELERATION

METHODS: Data from eight PMHS tests and eight Hybrid III 50th percentile male dummy tests were used to compare PMHS spinal acceleration measured at T1 to that measured at T8/9 and to evaluate the correlation between PMHS spinal acceleration and dummy chest cg acceleration. A mid-size 4-door sedan buck was used for all tests and all tests were performed with a 48-km/h change in sled velocity (ΔV). Test conditions included occupants seated in the driver position and in the right-front passenger (RFP) position. The driver-side and RFP tests were performed with slightly different acceleration pulse shapes, so comparison across seating positions is not valid. This factor does not affect the study, however, since the purpose of the study is to compare PMHS measurements with ATD measurements in nominally identical conditions. All ATD-to-PMHS comparisons involve subjects exposed to identical acceleration pulses, restraint conditions, and seating positions.

Driver-side occupants were tested with a pre-tensioned, force-limiting (FL) lap-and-shoulder (LS) belt system with an airbag (AB) (three dummies, three PMHSs) (Condition A_{Dr}) (Shaw et al. 2000). RFP occupants were tested with two restraint conditions: 1) a pre-tensioned, FL LS belt system with an AB (three dummies and three PMHSs) (Condition A_{RFP}) and 2) an AB restraint and no shoulder belt (two dummies and two PMHSs) (Condition B_{RFP}) (Table 1) (Kent et al. 2000, Kent et al. 2001). Condition A_{Dr} and condition B_{RFP} were chosen to span a wide range of conditions: condition A_{Dr} results in early coupling of the occupant to the vehicle, early airbag engagement, and well-controlled occupant kinematics, while condition B_{RFP} results in later coupling of the occupant to the vehicle, later airbag engagement, higher torso rotational rates, and higher occupant accelerations. Condition A_{RFP} is an intermediate condition since the pretensioned belt couples the passenger-side occupant with the vehicle early, but the larger airbag and greater initial distance between the occupant and the airbag result in intermediate values of spinal acceleration and T1 rotational rate.

PMHS spinal acceleration was measured using a triaxial array mounted midsagittally in a coordinate system nominally similar to the dummy coordinate system. The mounting scheme used for the PMHS T1 accelerometers resulted in the center of the array being mounted approximately 15 mm posterior of the skin over the T1 spinous process. Lateral X-ray views of the PMHS mounting sites were interpreted by an orthopedic surgeon to determine the distance from the vertebral body center (assumed to be a reasonable representation of the joint center) to the center of the externally mounted accelerometer array. This distance was found to be approximately 70 mm to 75 mm. Magnetohydrodynamic (MHD) angular rate sensors, mounted at the same location as the accelerometer array, were used to resolve the externally measured T1 resultant acceleration to the vertebral body center. The purpose of this data manipulation was to determine whether rotation of the spinal vertebrae resulted in significant differences between externally measured acceleration and acceleration at the vertebra center. An investigation, summarized in Appendix A, found that the

resultant spinal accelerations were similar but that the individual component accelerations could be substantially different between the sensor mount and the vertebral body locations.

In order for a fair comparison of T1 and T8/9 to be made, it was necessary to verify that differences in instrumentation mounting schemes were not significant. Compared to the accelerometer array at T1, the array used at T8/9 was not mounted as far posterior from the joint center. Based on this proximate accelerometer location, and the fact that rotational rates are not as large at T8/9 as they are at T1, it was assumed that the externally measured acceleration at T8/9 was equivalent to the acceleration at the T8/9 joint center.

Due to spinal compliance and instrumentation mounting differences between PMHSs and dummies, only the resultant accelerations will be used in the experimental comparison study. For comparison with the dummy, all PMHS acceleration values have been mass-scaled to 75-kg using the scaling technique described by Eppinger et al. (1984).

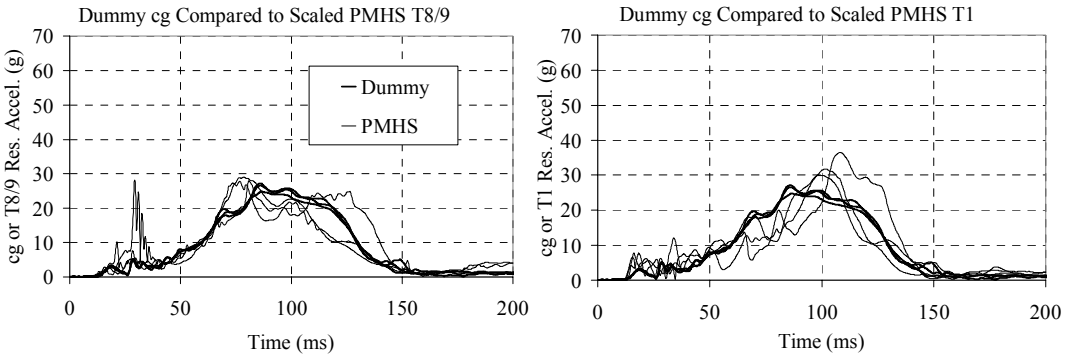
PMHSs were preserved until the time of testing by refrigeration, freezing, or customized embalming. Pre-test radiographs were taken to verify that no subjects had existing pathology or other skeletal abnormalities. All test procedures were approved by the University of Virginia institutional review board. Pulmonary and cardiovascular systems were pressurized to typical *in vivo* levels (approximately 10 kPa) immediately before testing.

Table 1 –Summary of Test Conditions

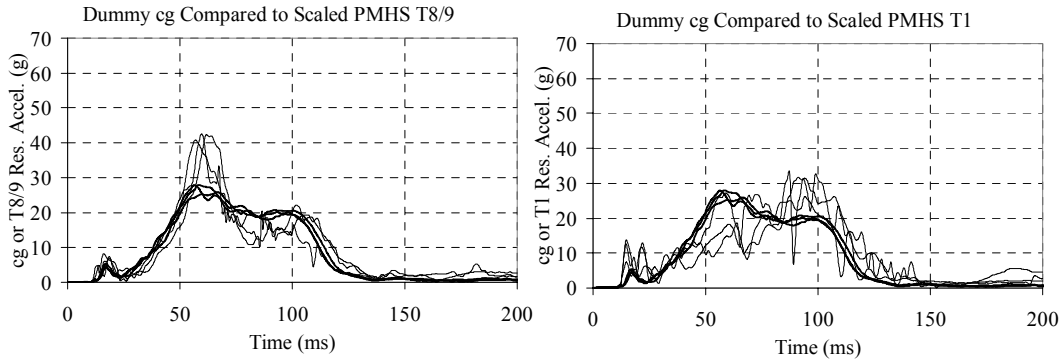
Test	Subject	Position	Restraint	Test Code
532	Hybrid III 50th male	Driver	Pre-tensioned, FL LS Belt and Airbag	A_{Dr}
537	Hybrid III 50th male	Driver	Pre-tensioned, FL LS Belt and Airbag	A_{Dr}
538	Hybrid III 50th male	Driver	Pre-tensioned, FL LS Belt and Airbag	A_{Dr}
534	47 male, 51 kg, 170 cm	Driver	Pre-tensioned, FL LS Belt and Airbag	A_{Dr}
544	59 female, 56 kg, 168 cm	Driver	Pre-tensioned, FL LS Belt and Airbag	A_{Dr}
545	67 male, 74 kg, 176 cm	Driver	Pre-tensioned, FL LS Belt and Airbag	A_{Dr}
571	Hybrid III 50th male	RFP	Pre-tensioned, FL LS Belt and Airbag	A_{RFP}
572	Hybrid III 50th male	RFP	Pre-tensioned, FL LS Belt and Airbag	A_{RFP}
576	Hybrid III 50th male	RFP	Pre-tensioned, FL LS Belt and Airbag	A_{RFP}
577	57 male, 70 kg, 174 cm	RFP	Pre-tensioned, FL LS Belt and Airbag	A_{RFP}
578	69 female, 53 kg, 155 cm	RFP	Pre-tensioned, FL LS Belt and Airbag	A_{RFP}
580	57 female, 57 kg, 177 cm	RFP	Pre-tensioned, FL LS Belt and Airbag	A_{RFP}
648	Hybrid III 50th male	RFP	Lap Belt and Airbag	B_{RFP}
649	Hybrid III 50th male	RFP	Lap Belt and Airbag	B_{RFP}
650	40 male, 47 kg, 150 cm	RFP	Lap Belt and Airbag	B_{RFP}
652	46 male, 74 kg, 175	RFP	Lap Belt and Airbag	B_{RFP}

RESULTS: Comparison of T1, T8/9, and Dummy Chest CG Acceleration Measurement: The tests performed on the driver side (A_{Dr}) resulted in reasonably good similarity between the dummy chest cg acceleration and the mass-scaled PMHS T8/9 acceleration (Figure 1a, left plot). The acceleration-time histories for both occupant types were nominally trapezoidal in shape with a peak of approximately 25g and a duration of approximately 150 ms. The PMHS T1 acceleration is less similar than T8/9 to the dummy chest cg acceleration (Figure 1a, right plot).

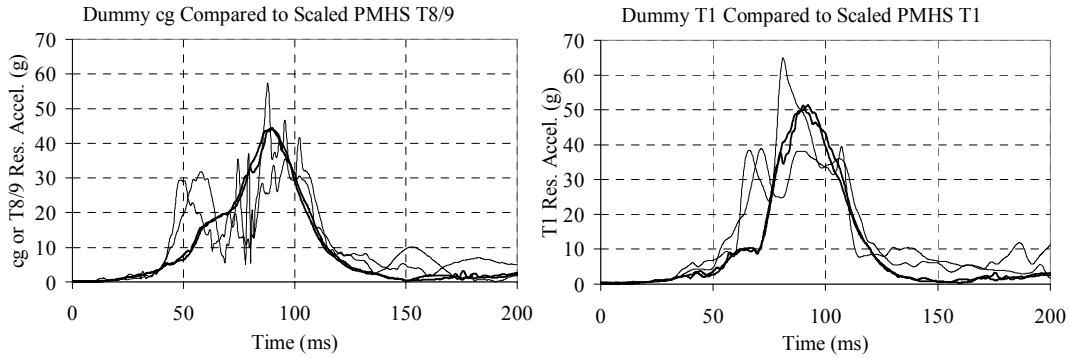
For the tests performed with the A_{RFP} condition, the maximum acceleration measured at the PMHS T8/9 was much greater than the dummy chest cg maximum acceleration. The T8/9 acceleration reached a peak of approximately 40g, while the dummy maximum was approximately 25g (Figure 1b, left plot). In this test condition, the magnitude of the T1 acceleration was a better representation of the dummy chest cg acceleration magnitude (Figure 1b, right plot) (Figure 2). It is interesting to note,



a) Condition A_{Dr} (Pretensioned, Force-Limited Belt, Airbag, Driver Side)



b) Condition A_{RFP} (Pretensioned, Force-Limited Belt, Airbag, Passenger Side)



c) Condition B_{RFP} (Lap Belt, Airbag, Passenger Side)

Figure 1 – Comparison of PMHS (thin lines) and Dummy (thick lines) Acceleration-Time Histories for Three Test Conditions

however, that, despite the difference in the magnitudes of the maxima, the restraint interaction with the torso is described similarly by both the dummy chest cg acceleration and the PMHS T8/9 acceleration: with a pre-tensioned, FL belt and airbag on the passenger side, there are two distinct loading regimes, which result in a bimodal acceleration-time history. An initial and global maximum is generated at approximately 60 ms under primarily belt loading, followed by a second, sub-maximal acceleration peak at approximately 100 ms when the occupant is at maximum engagement with the airbag. This bimodal response is captured by both the dummy chest cg acceleration and the PMHS T8/9 acceleration (Figure 1b, left plot), but is not seen in the PMHS T1 acceleration (right plot). So, while the PMHS T1 acceleration magnitude is closer to the dummy chest cg acceleration magnitude,

the PMHS T8/9 acceleration is a better representation of the transient restraint loading on the anterior thorax.

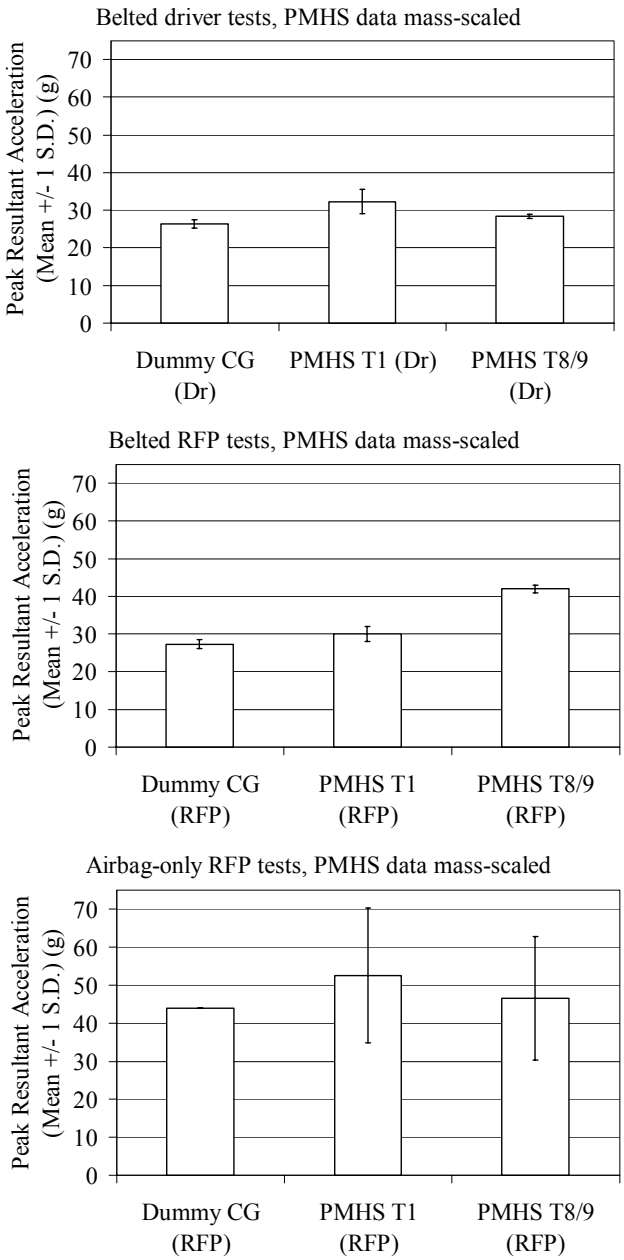


Figure 2 – Comparison of Dummy cg, PMHS T1, and PMHS T8/9 Acceleration Maxima for all Three Test Conditions

For the tests performed with the B_{RFP} condition, the PMHS test-to-test variability was much larger than that observed for the other two test conditions, particularly for the T1 acceleration (Figure 1c). Dummy chest cg acceleration maximum was approximately 43g. The mean PMHS T1 acceleration maximum for the two tests was higher than the mean PMHS T8/9 acceleration maximum, and both T1 and T8/9 were higher than the dummy chest cg acceleration maximum. The PMHS T8/9 acceleration bounded the dummy chest cg acceleration for most of the pulse.

For all restraint conditions, the PMHS test-to-test variability was less for T8/9 acceleration than for T1 acceleration.

COMPUTATIONAL STUDY USING MADYMO HUMAN AND DUMMY MODELS

In order to gain a better understanding of the experimentally observed differences between ATD and PMHS acceleration results, a series of simulations was run using the MADYMO program (TNO Automotive, MADYMO 1999). The models were based on a previous model of a Hybrid III sitting in the occupant compartment of a common, midsize four-door sedan (1997 Honda Accord) (Thacker et al. 1999). The Hybrid III is a validated model from the MADYMO database, and the combined dummy/compartment model was validated against a 57-km/hr Accord US New Car Assessment Program (NCAP) test.

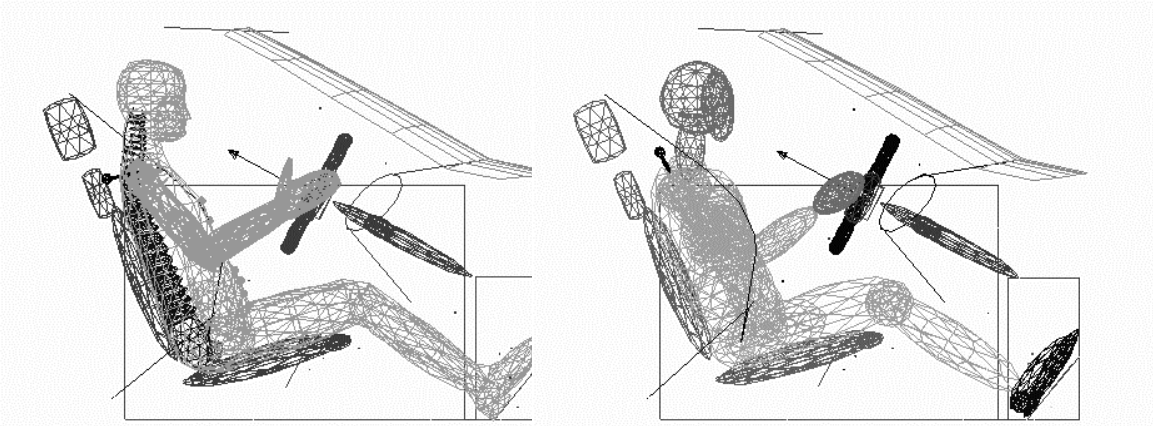
METHODS: To simulate the PMHS tests, a recently developed MADYMO 50th percentile male "human" model was placed in the model of the Accord compartment. The human model is significant in that it contains detailed representations of rib and spine flexibility. Neither of these components is represented with full biofidelity in the Hybrid III model, particularly the spine. Spinal properties in the human model are based on data from the RAMSIS database (Happee et al. 1998).

These two models were refined further using two 57-km/hr UVA sled tests. The tests were similar to the A_{DR} experimental test condition with the exception of the higher ΔV (57 km/hr vs 48 km/hr). The seat position, seat stiffness, belt properties, and airbag timing were adjusted in the models until good qualitative agreement was obtained between the validation tests and the model kinematics. Additional adjustments were made until the quantitative pelvic kinematics of the human and dummy models were nearly identical. This was done to ensure that any differences in thoracic measurements

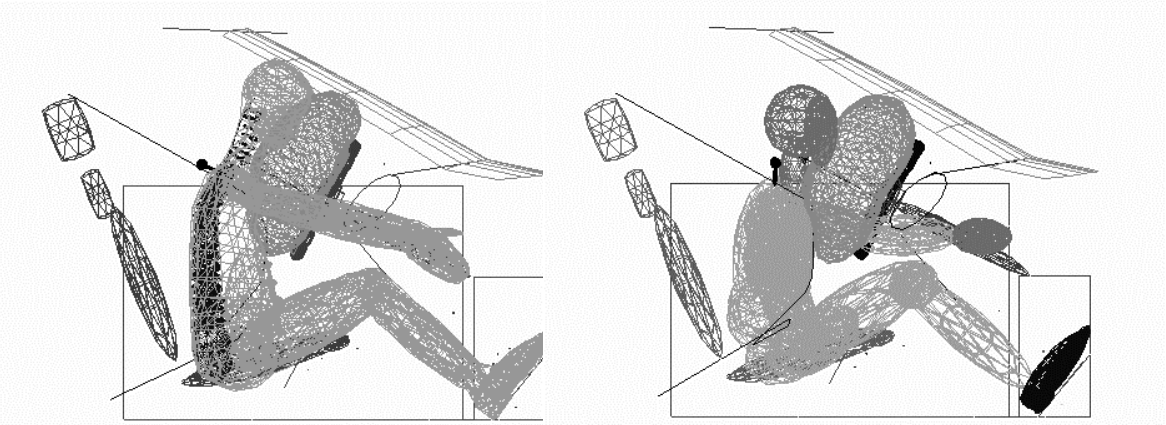
would be due to thoracic or spinal structural differences rather than differences in initial position. Figure 3 compares human and dummy body positions at 0 ms and 120 ms for restraint condition A_{Dr}.

The validated dummy and the human models were used to explore observed differences in test results recorded for dummy and PMHS occupants and to suggest possible mechanisms for these differences.

RESULTS: Effects of Spinal Flexibility: The T1 investigation demonstrated that the human spine undergoes local deformations that are significantly different from the gross motion of the torso. Obvious technical limitations, however, preclude mounting accelerometers at the PMHS torso cg. There also are limits to the number of accelerometers that can be mounted to the PMHS spine. The models were therefore used to determine whether measurements at other vertebral bodies are better indicators of torso kinematics than T1. Lacking PMHS torso cg acceleration data, the dummy chest cg acceleration was used as the standard with which to compare the acceleration results from the human T9. Accelerometers mounted to the T9 vertebra are approximately at the superioinferior location of the human torso cg, although several inches posterior.



a) Human (left) and Dummy (right) Positions at 0 ms



b) Human (left) and Dummy (right) Positions at 120 ms

Figure 3 – Comparison of Human and Dummy Kinematics with Restraint Condition A_{Dr} and 57-km/hr ΔV

Figure 4 shows the resultant acceleration at T1 and at T9 for the human model compared to the dummy chest cg. In this case, the T9 measurement is a better match than the T1 measurement. The magnitude is closer and the general shape of the curves is similar. In agreement with the experimental findings with this restraint condition on the RFP (Figure 1b), dual peaks are observed at T9 and at the

dummy chest cg (Figure 4, right plot), while the human T1 response is a unimodal peak that lags the dummy cg bi-modal response (Figure 4, left plot).

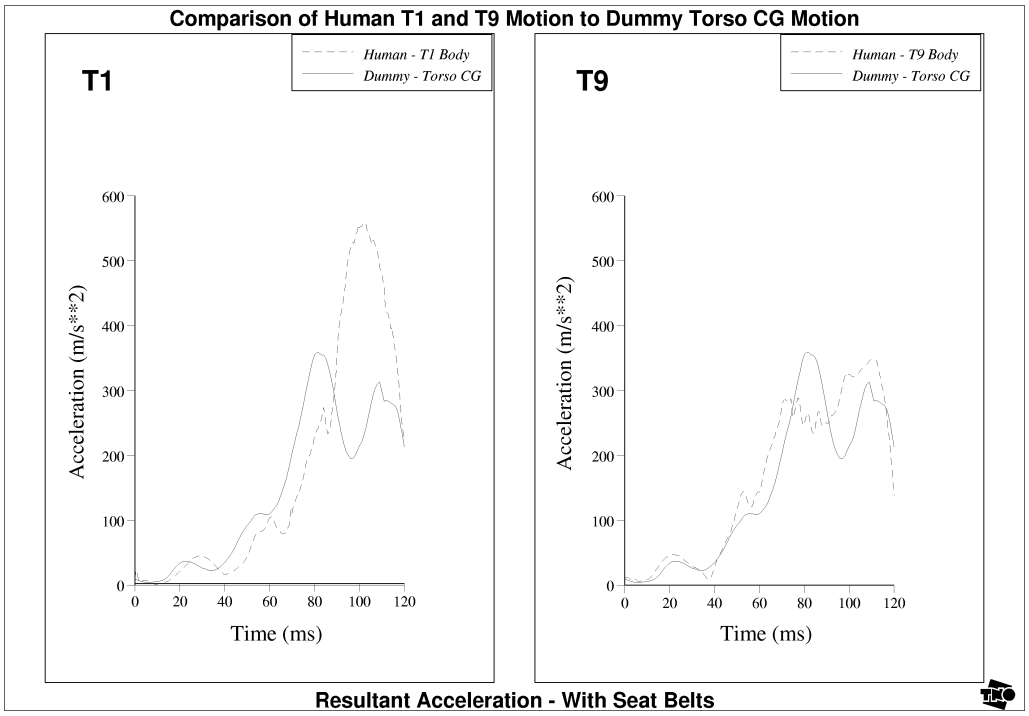


Figure 4 – Human T1 and T9 Acceleration Compared to Dummy Chest cg with Restraint A_{Dr}, 57-km/hr ΔV

Additional simulations were performed using unbelted occupants in order to determine if T9 was also a favorable instrument location for a different restraint condition. Belts limit torso motion considerably and reduce the effects of spinal flexibility. With belt restraints in place, the human and dummy kinematics are similar (Figure 3). When the belt is removed, however, kinematic differences are substantial (Figure 5); particularly evident is substantial spinal curvature for the human, which is not present in the dummy model.

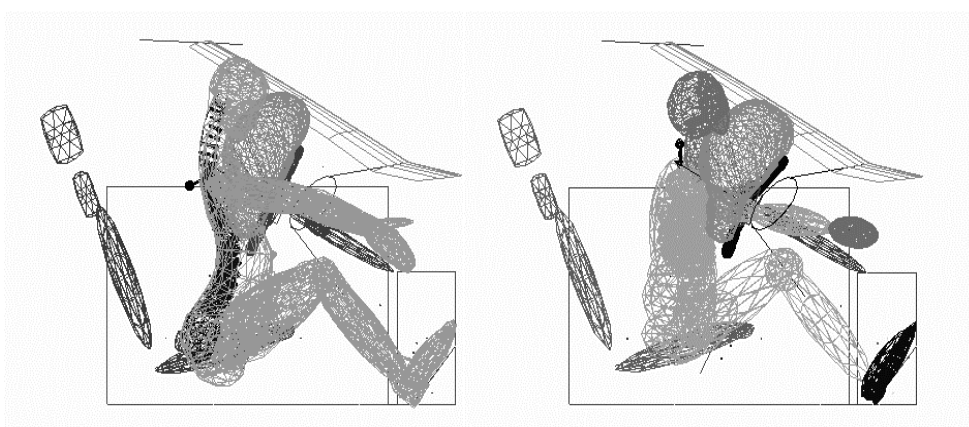


Figure 5 – Comparison of Unbelted Human (left) and Unbelted Dummy (right) at 120 ms (note spinal curvature)

Figure 6 shows how the acceleration measurements at T1 and at T9 have changed. With the belt removed, T1 acceleration is now in close agreement with the dummy torso cg, while T9 acceleration

has developed a large, sharp spike near 110 ms. The spike appears to be the result of the lumbar and lower thoracic vertebrae reaching their range-of-motion limit; effectively causing a sudden stiffening of the spine. It is not known whether this response is biofidelic, though unexplained spikes similar to this have been observed experimentally.

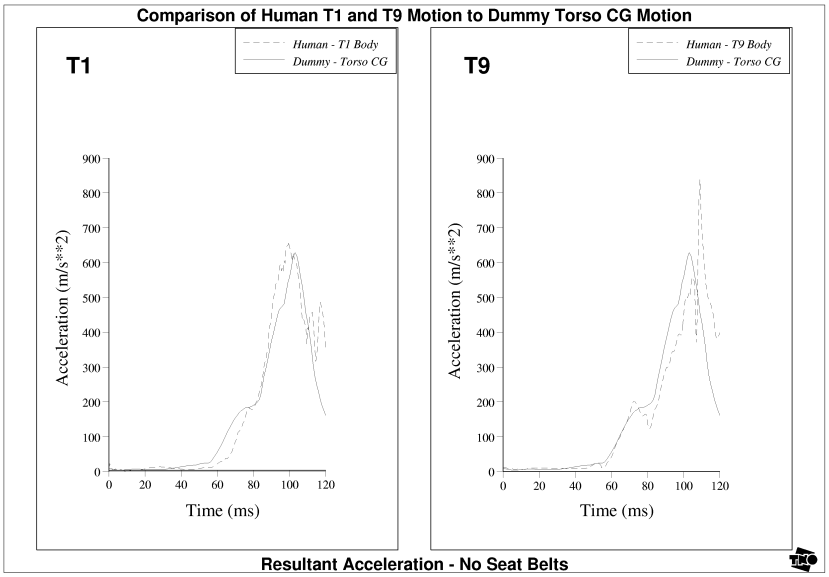


Figure 6 - Human T1 and T9 Resultant Acceleration Compared to Dummy Chest cg with No Seat Belt , 57-km/hr ΔV

Figure 7 shows the unbelted results for the z-axis acceleration component. While the magnitudes are similar, the sign of the peak is now reversed for both T1 and T9. With the belts removed, the flexibility of the human spine has introduced a significant qualitative difference between human and dummy behavior.

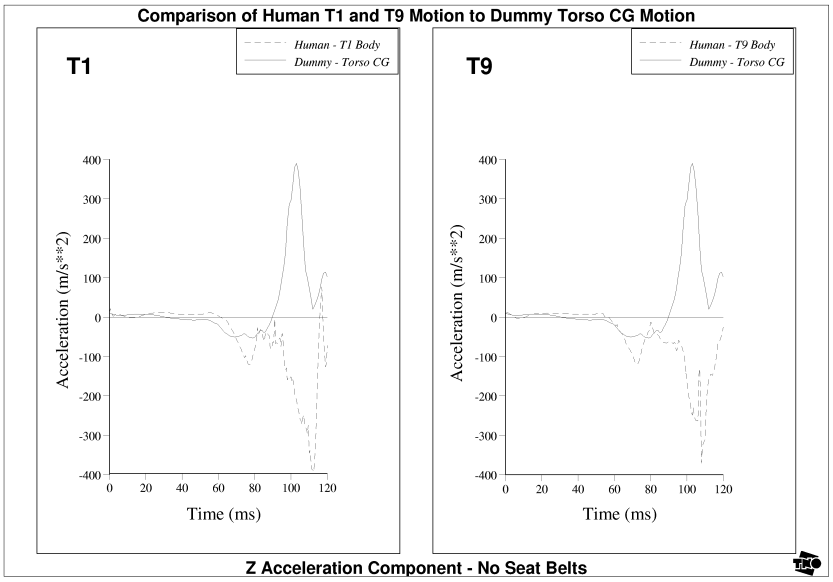


Figure 7 - Human T1 and T9 z-axis Acceleration Compared to Dummy Chest cg z-axis Acceleration with No Seat Belt , 57-km/hr ΔV

In order to explore the relationship of spinal kinematics to spinal flexibility, the human model was modified so that all of the thoracic and lumbar joints were locked, and the cervical joints were constrained to be rotational only (spherical) as opposed to rotational and translational (free). Note that qualitative agreement between dummy and modified human could only be achieved by locking all of the human lumbar joints. Leaving even one of them free to rotate produced more extension than is permitted by the dummy's stiff lumbar rubber. The modified human spine model produced qualitative kinematics that more closely approximate those of the dummy (compare Figure 8 to Figure 4). The numerical results also show better agreement when the human spine is constrained. The modified human z-axis T1 and T9 acceleration peaks, while still substantially different in magnitude from the dummy peaks, now have the same sign as the dummy peaks (Figure 9).

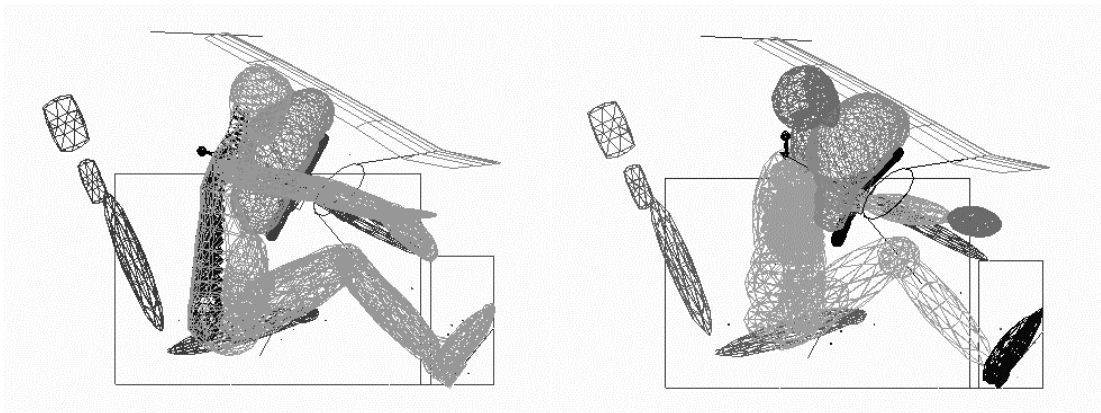


Figure 8 – Comparison of Unbelted Human (left) and Dummy (right) at 120 ms (Human Spine Modified)

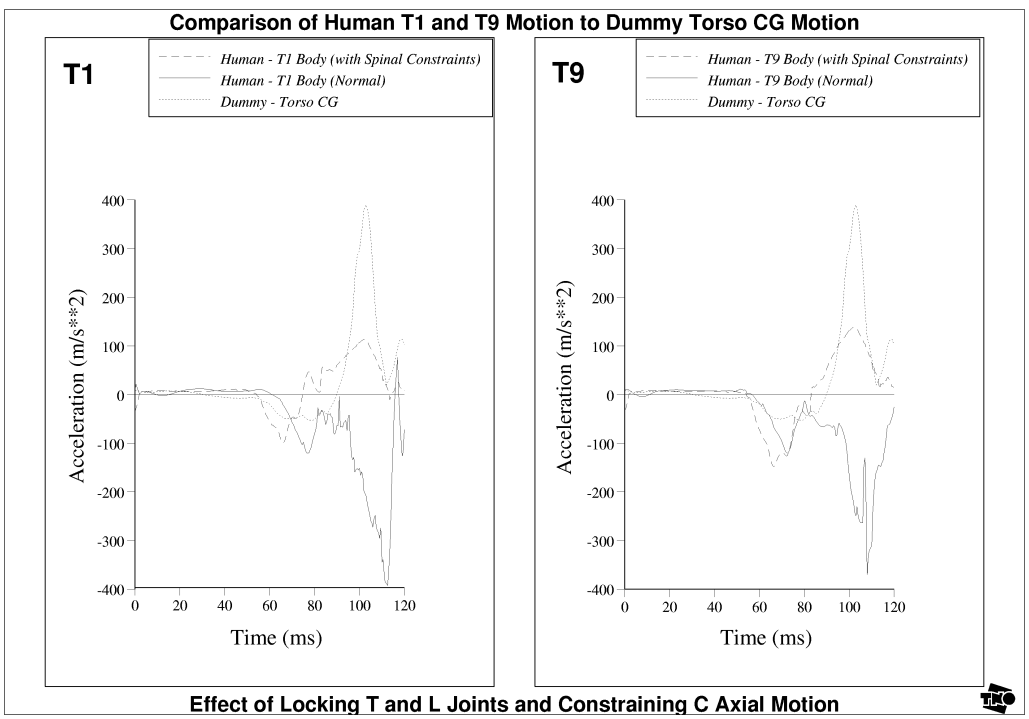


Figure 9 – Modified Human T1 and T9 z-axis Acceleration Compared to Dummy Chest cg z-axis Acceleration with No Seat Belt , 57-km/hr ΔV

DISCUSSION OF EXPERIMENTAL AND COMPUTATIONAL FINDINGS

For frontal sled tests, it is necessary to discuss two distinct phenomena in assessing spinal kinematics: 1) the ability of spinal acceleration to characterize the occupant response regardless of the surrogate and 2) the ability of dummy response measurements to represent those of the PMHS, and vice versa.

In both the PMHS sled tests and human models, we observed significant differences between the behavior of the upper and middle spine accelerations. These deviations appear to depend on both the restraint configuration and the occupant's seating position. These differences suggest that T1 and T8/9 accelerations are not equivalent measures for characterizing occupant and restraint interaction. Furthermore, it suggests that maximum accelerations at these spinal locations are not comparable assessment tools for predicting the likelihood of occupant injury. Statistical correlations between observed injury and recorded spinal acceleration have used the T1 data since it has traditionally been the only location uncomplicated by complementary devices recording chest deformation. Given the dissimilarities between the T1 and T8/9 accelerations for certain restraints, data in this paper suggest that if spinal acceleration is to be used as a comparative measure that consideration be given to using the T8/9 rather than T1 location. The T8/9 behavior for passenger PMHS tests (A_{RFP}) showed better qualitative agreement with the restraint interactions than did the T1 acceleration-time histories. In particular, the increased distance from occupant to airbag in the passenger tests allowed distinct phases of belt and airbag loading to be characterized using the T8/T9 accelerations. These phases were corroborated using film analysis and chest band data (Kent 200). The T1 accelerations did not characterize these loading regimes. We hypothesize that, in the case of restraint loading that causes relatively high-frequency changes in the force on the anterior thorax, the relative motion among spinal vertebrae acts as a low-pass filter. As a result, kinematic response to belt loading followed by airbag loading is described fairly well at T8/9, while the acceleration signal is filtered as it moves up the spinal column and the restraint effects are not captured at T1. This same filtering results in artifactual acceleration spikes from, for example, head strikes to be more pronounced at T1 than at T8/9 (Hassan and Nusholtz 2000).

Although T8/9 acceleration appears to characterize occupant-restraint interactions, this alone is not sufficient to conclude that spinal accelerations are either causative or correlative factors in thoracic injury. Additional examination of the available frontal sled test data (cf. Kent et al, 2001) and supplemental testing may be necessary to determine a relationship between spinal acceleration and the likelihood of injury.

An additional step in assessing the viability and practicality of spinal acceleration as an injury predictor or as a descriptor of restraint interaction is the relationship between the dummy and PMHS accelerations. Given the considerable differences in flexibility, construction, and segmental mass, studies correlating the gross spinal response of dummy and PMHS have shown surprising similarities (e.g. Shaw et al. 2000). If polarities of individual component accelerations are considered, however, the models show a difference in sign between the z-axis dummy accelerations and the PMHS accelerations for some restraint conditions. This may be attributed to local differences in spinal flexion (Figure 5).

In this paper, the simulations of belted drivers showed similar behavior for the dummy chest cg and human T9 accelerations. For the unbelted occupants, maximum T1 resultant acceleration closely matched the dummy maximum chest cg resultant acceleration while T9 accelerations exhibited a similar trend until a spike in the data occurred. At the time of this paper, it is unclear whether this spike is related to computational errors or actual spinal phenomena. The authors have observed similar behavior in sled tests where a sudden over-center or joint-contact phenomenon has been hypothesized as the mechanism.

The correlation between dummy and PMHS accelerations is less clear from the experimental test results. The T8/9 maximum accelerations more closely approximated those of the dummy chest cg for the A_{DRV} and B_{RFP} tests while maximum T1 accelerations were closer for the A_{RFP} tests. It is important to remember that the peak accelerations simply represent one instant in time for the entire impact event and do not in themselves constitute similar biomechanical response. The dummy accelerometer is located at the chest cg and is attached to a structure that is both massive and rigid.

By comparison, the PMHS accelerations are recorded on a single vertebral structure that is part of a flexible spinal column posterior to and, in the case of T1, significantly superior to the chest cg location. Given these differences, mass scaling of PMHS spinal acceleration data is likely insufficient to provide an equivalent measure to the dummy chest cg, especially when restraint conditions are varied. We found a substantial difference in the spinal kinematics between the dummy and human models due to the flexibility of the human torso. When joint definitions of the human spine model were changed to eliminate or reduce spine flexibility, the simulation results approximated those of the dummy model. The flexibility of the human spine compromises experimental determination of overall thoracic kinematics from instruments mounted to individual vertebrae. The ability to approximate overall torso acceleration with individual vertebral data varies with the restraint condition. Given these considerations, magnitude similarities between recorded dummy and PMHS accelerations may be coincidental and more emphasis should be placed on characterizing the general acceleration-time history behavior.

Although use of the T8/9 spinal acceleration qualitatively conforms to observed restraint interactions, the fundamental question remains – whether it is better to characterize gross motion (i.e., a dummy equivalent measure of overall chest response) or localized spinal response (i.e., PMHS vertebral measures) to assess occupant response, injury, and interaction with restraints. While measurement of PMHS spinal motion is a better indicator of human spinal motion, the dummy chest cg measurement is a better descriptor of the overall thoracic response to restraint loading and therefore is probably a better indicator of restraint loading than is the acceleration measured at a single vertebra. Unfortunately, extrapolation of the dummy measure to the case of a human is difficult because of the obvious lack of spinal biofidelity in the dummy. We have shown that correlating PMHS-based acceleration measurements to dummy-based acceleration measurements is not valid when restraint conditions and seating position are varied, especially if PMHS-based acceleration is measured at T1. In the absence of widespread utilization of improved anthropometric test devices, therefore, direct comparisons of measured dummy response to observed PMHS injury in the same impact environment provide the only viable assessment tool (e.g. Kent et al. 2001). This approach assumes, however, comparable occupant kinematics between dummy and PMHS, so caution must be used to ensure that the dummy's kinematics are representative of the PMHS's.

The introduction of dummies with improved biofidelity (Shaw et al. 2000) will at least provide more comparable measures between surrogates. The advanced Test device for Human Occupant Restraint (THOR) dummy includes a flexible spinal unit intended to improve spinal biofidelity. The BioRID dummy, which was developed for rear-impact studies, incorporates individual vertebral components for the entire cervical, thoracic, and lumbar spine (Davidsson et al. 1999). It is recognized that complex spinal motions occur in rear impacts and that these motions affect overall body kinematics. Further, these motions may be important indicators of injury risk. The results of our studies suggest that a detailed dummy spinal model may be necessary in order to attain biofidelic kinematics in a frontal impact as well if the restraint condition is varied. Neither the THOR dummy or the BioRID dummy are currently in widespread use, however, so the short-term solution may involve developing assessment tools based on the particular dummy, restraint, and seating position involved.

CONCLUSIONS

Resultant spinal accelerations showed minimal differences between sensor mount and vertebral body locations. If examining individual acceleration components, additional degrees of freedom and instrumentation (i.e., additional accelerometers or angular rate sensors) need to be added to transform spinal accelerations from the sensor mount location to the vertebral body.

T8/9 accelerations demonstrate better characterization of the restraint interactions than do T1 acceleration measures. While no injury correlative or causative relations exist for T8/9 information, the improved qualitative agreement of T8/9 over T1 for characterizing restraint effects suggests examination of the mid-spine in addition to the upper spine data in all PMHS sled tests.

Local loading effects of the body by restraints and occupant interactions with the vehicle interior structures may appear as localized accelerations of the spine. Regional compliance and curvature of the spine itself may also contribute to recorded local accelerations.

Significant differences exist between the current dummy spine design and human anatomical structures. These differences preclude dummy assessment of accelerations that would be recorded in the human. Current techniques to compensate for these differences are limited to conducting PMHS and dummy tests in identical test environments and then developing correlations between measured dummy and PMHS responses for specific restraint and occupant position.

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Appendix A
Effects of External T1 Measurement

An investigation was undertaken to determine whether movement of the spinal vertebrae resulted in significant differences between externally measured acceleration and acceleration at the vertebra center. We examined the resultant acceleration maxima recorded in the sled tests. A computational study using the previously described MADYMO computer models explored the effects of sensor position on acceleration resultant and components.

Sled Test Resultant Acceleration

The externally measured PMHS T1 resultant acceleration ($T1_{ext}$) was found to represent the resultant acceleration at the T1 vertebra body center ($T1_{joint}$) better for some test conditions than for others. For all three test conditions, however, the effect of the external measurement was small compared to the test-to-test variability in peak $T1_{ext}$. For condition A_{Dr}, the test-to-test variation in

peak T1ext was approximately 18% (largest difference in peak acceleration as a percentage of mean peak acceleration for all tests), while the difference between T1ext and T1joint varied from 0g to –1g (0% to 3%). For condition ARFP, the difference between T1ext and T1joint was more pronounced, varying from +3g to –3g (6% to 11%). This difference was still small, however, compared to the test-to-test variability in peak T1ext (23%). The tests run with condition BRFP resulted in the highest difference between peak T1ext and peak T1joint (–4g to –8g) (6% to 17%), but also resulted in the greatest test-to-test variability (36%). These findings indicate that, for a range of restraint conditions and occupant sizes, the effect of externally mounted accelerometers at T1 is small compared to test-to-test variability in resultant acceleration maxima (Figure A1).

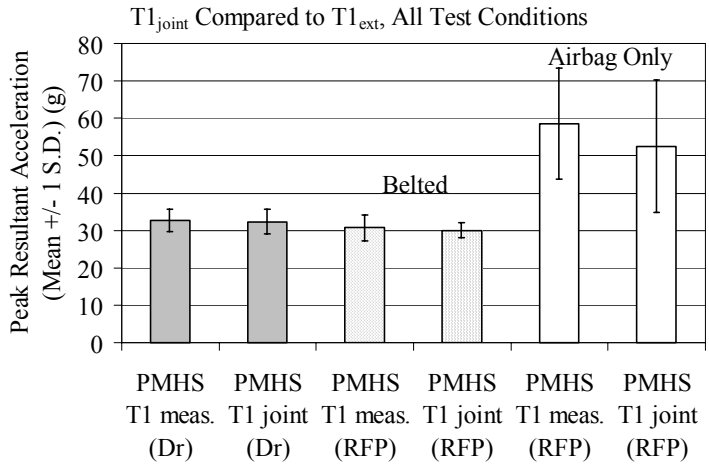

Figure A1 – Variation Between $T1_{ext}$ and $T1_{joint}$ Compared to Test-to-Test Variability For All Three PMHS Test Conditions

Computational Study of T1 Acceleration The experimental offset between the sensor and joint center locations was simulated with an element offset from the human and dummy models at the nominal location of T1, and attached to T1 using a rigid element (Figure 3). Figure A2 shows the sensitivity of acceleration measurements to the offset distance. Similar to the experimental findings, instrument offset shifts the peak resultant acceleration by only about 5%. The z-axis acceleration (Figure A2, right plot), however, illustrates that the offset can have a substantial effect on the magnitude of the vector components (note that in Figure A2, "body" refers to the nominal joint center acceleration and "cube" refers to the externally measured acceleration).

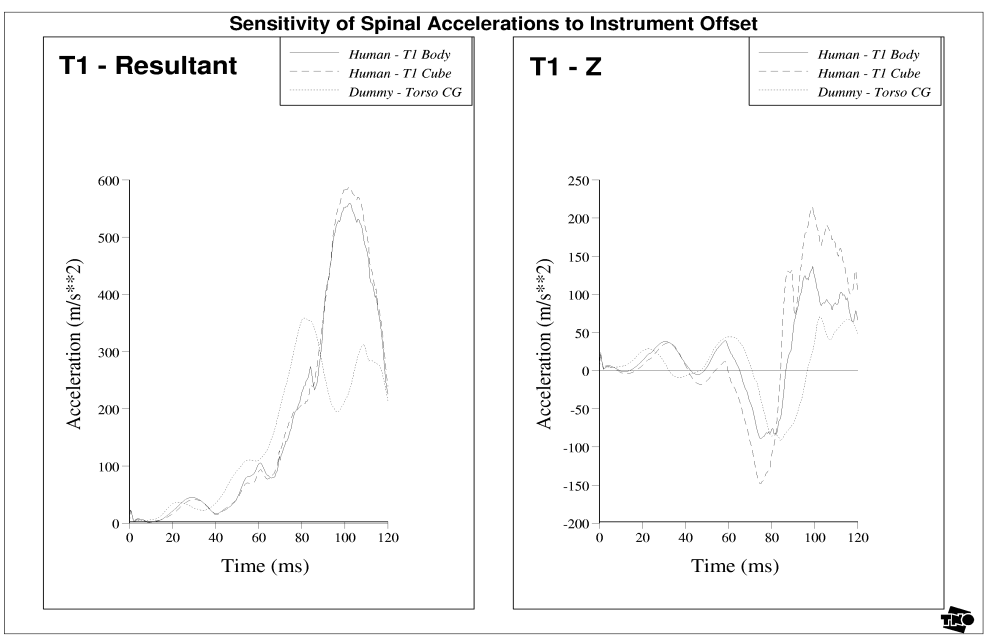

Figure A2 – Sensitivity of Human T1 Acceleration to Instrument Offset