Rollover Simulation Using an Active Human Model

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ABSTRACT

Today's anthropomorphic test devices (ATD) derive their behavior from cadaver test data. This same statement also applies to numerical models of these physical ATDs, and equally to the more sophisticated numerical human body models. Across the wide spectrum of automotive and aerospace crash scenarios, the prediction of occupant responses relies mainly on joint properties that are inherent in such behavioral representations. These do not account for the muscle reflexes of tensing and bracing. Most ATDs, and especially the Hybrid IIIs, are relatively rigid and their response will effectively represent occupants subjected to high speed impacts.

A series of numerical active human (AH) body models have been developed for the 5th, 50th and 95th percentile of human subjects using multi-body modelling that incorporates joints with active torque behavior. In addition to the standard joint torque resistance, active joint behavior is implemented numerically in these AH-models using proportional-integral-derivative (PID) control methods to deliver torque resistance representative of active muscle responses. Active torque behavior for selected human body joints is achieved by optimizing PID gain parameters to correlate with test responses of human volunteer test data. The result of this work was applied to this first generation of active joint human models.

The potential of human body models with active joints is demonstrated in a vehicle rollover situation. The specific case of vehicle rollover provides a crash scenario where the occupant's accident awareness response is likely to influence tensing and active joint behavior at various stages during the accident. These simulations highlight the influence of muscle tensing and joint bracing on potential injury risk.

This method of modelling the active joint torque seeks to mimic the complex behavior of muscles. It provides an efficient modelling technique that can be used to simulate long duration events (such as vehicle rollover) that in the past may have been considered less than optimal for the more complex human models. The ability to activate or deactivate the joint behavior to account for conscious muscle tensing will allow the analysis of various occupant awareness states during a rollover accident.

It is anticipated that the addition of active joint behavior will provide a more accurate numerical representation of human body kinematics and hence improve the quality of the prediction of the risk of injury that can be deduced from simulations. The ability to activate joint behavior to account for conscious muscle reflexes will also extend the range of crash scenarios which can be modelled effectively.

INTRODUCTION

Occupant safety regulations for the crashworthiness testing of vehicles rely on the physical representation of a human occupant by an anthropomorphic test device or ATD. These ATDs, more generally referred to as crash test dummies, exist in various customized forms that are designed to perform biomechanically for specific types of crash configurations. ATDs derived their behavior from cadaver test data [1,2] which can be considered similar to a flaccid human. However, if one manipulates the various joints or bend the neck, it becomes very clear that the dummy is very stiff. This is because ATDS are tuned to represent humans in high speed (about 60km/h), potentially injury-causing crashes. In a low velocity impact test, an ATD's response is unrepresentative because of their overly stiff nature.

The Hybrid III dummy is the current international standard for frontal crashes, while various other ATDs have been specifically designed for alternative test configurations such as side impact. To adequately represent the range of human sizes and proportions, these ATDs also exist as a series of anthropometric scales. Hence, such ATDs are implicitly limited by their characterization of actual complex humans in terms of their own geometric representation and mechanical response. "ATDs are mechanical surrogates designed to represent a particular demographic according to gender, size, and age. In addition, they are designed to exhibit a biofidelic response for specific loading conditions (e.g. principal direction of force and severity). The responses of these devices are not validated for alternate loading conditions and thus may not produce biofidelic responses beyond their intended design specifications."[3]

Simple but computationally efficient multi-body models as well as very detailed finite element (FE) models of the various ATDs have been developed over the years to simulate real world ATDs. Having numerical equivalents of the various ATDs has complemented real world tests by allowing more cost effective, efficient, and detailed analyses of ATD behavior in a wider ranges of test scenarios. The limitations of these models however, is that they can only be as accurate as the ATD's representation of real humans. For these reason, there have been extensive parallel development of numerical human body models that attempt to simulate real humans rather than ATDs.

Detailed numerical computational models of the human body have been developed by various research groups to allow more detailed study into biomechanical issues. Unlike ATDs, humans have functioning circulatory and respiratory systems, resting muscle tone and active bracing capabilities, continuous neural responses, and the ability to perform cognitive functions [3]. Numerical human models offer not only modelling flexibility but more exact characterizations of both varying human anthropometry and their biomechanical responses in a wide range of loading conditions. Some of these human models that have been developed include the H-MODEL (Hongik University's Human Body Model) [4], THUMS (Toyota's Total Human model for Safety) [5], and GHBMC's human body models (Global Human Body Models Consortium) [13].

In more recent years, attention has also been directed towards modelling active muscle behavior to account for an occupant's bracing reflex in the event of awareness of the approaching accident. It has been shown that tensed muscles can change the initial posture, kinematics, and subsequently the kinetics during an automotive collision and as a consequence, the resulting injury patterns may be altered based on muscle activation [8]. This study illustrated that muscle activation has a significant influence on the biomechanical response of human occupants in low-speed frontal sled tests.

MODEL DEVELOPMENT

Active human body models (aH-Model) are currently being developed with body size and weight representative of accepted automotive industry standards for occupant safety testing. A series of three aH-Models currently under development include the 5th percentile female (aH-F05), the 50th percentile male (aH-M50) and the 95th percentile male (aH-M95). The motivation for developing these aH-models is to study the contribution of active reflexive human responses to accident events and its potential to affect injury risk. The computational implementation takes advantage of the simpler but extremely efficient modelling techniques of multiple rigid body segment connected by joint elements with defined moment resistance. This traditional technique of modelling a joint is complemented by incorporating active joint behavior via a torque actuator at the joint. The torque actuator is implemented using a numerical closed loop proportional-integral-derivative (PID) controller. The efficiency of the aH-Models with its moderate CPU demand will allow the simulation of more complex and long duration crash and test scenarios where active responses are more likely to play a role in the kinematics of the occupant.

Human Geometry

For each of the human models, the outer surface geometry of their body segments were meshed from 3D surface data. The 3D human surfaces were produced from geometric scans of selected human subjects postured in a seated position. The three scanned human subjects were selected by size and weight, to be representative of the previously mentioned size categories. The selection criteria were based on the Size USA 2002 datasets.

Parts Segmentation and Joint Locations

The human geometric scans were discretized into a finite element (FE) mesh, and then subsequently segmented into the fifteen commonly-accepted anatomical body parts. For each mesh of the human model category, a consistent segmentation method was applied across the three aH-models to create these rigid body segments. Figure 1 shows the three human models and the segmentation scheme used to create the various anatomical parts.

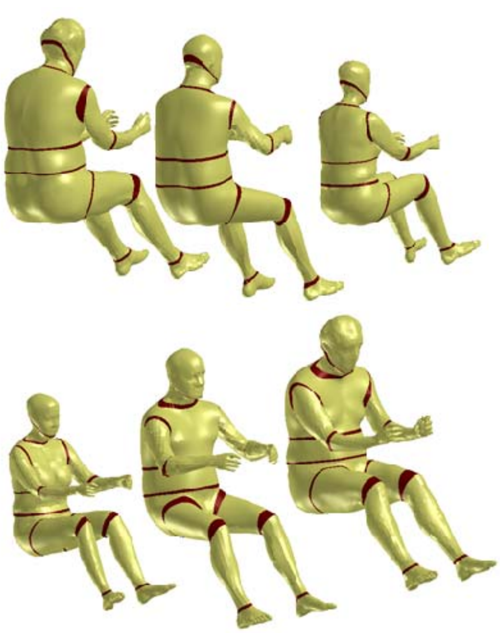


Figure 1. Human models meshed and segmented from the Size USA 2002 data

A skeletal mesh within the surface mesh, as illustrated in figure 2, provides a reference to aid in the body segmentation process. In particular, while the head/neck/trunk is represented by five body segments that are separated by defined spinal positions, they can only be physically located using the spine as a reference. The skeletal articulation also assists with the location of the joint position of the shoulder, hip, and joints of the limbs. Table 1 summarizes the modelled joints and their anatomical positions.

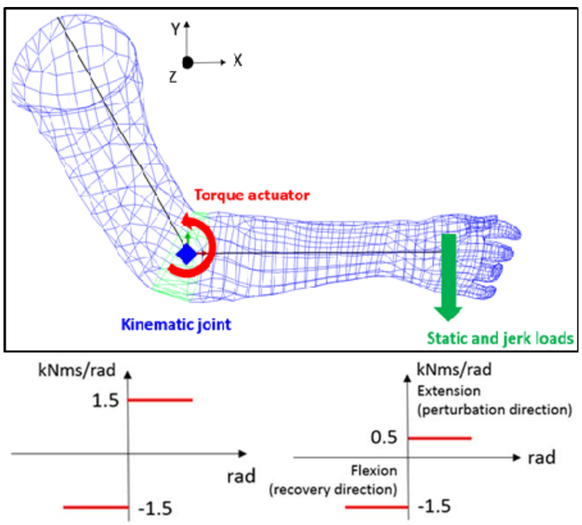
Figure 2. Human models including skeletal structure

Table 1.
Joints and body segments.

#	Joint	DOF	Anatomical position
1	Head-neck	3	OC joint
2	Neck-Upper trunk	3	C7/T1
3	Upper-Center trunk	3	T12/L1
4	Center-Lower trunk	3	L5/S1
5	Upper trunk-arm, R	3	Right Shoulder
6	Upper-Lower arm, R	1	Right Elbow
7	Upper trunk-arm, L	3	Left Shoulder
8	Upper-Lower arm, L	1	Left Elbow
9	Lower trunk-leg, R	3	Right hip joint
10	Upper-Lower leg, R	1	Right Knee
11	Lower leg-foot, R	3	Right Ankle
12	Lower trunk-leg, L	3	Left hip joint
13	Upper-Lower leg, L	1	Left Knee
14	Lower leg-foot, L	3	Left Ankle

Active Joint Modelling

Each joint is modelled using a kinematic FE joint element consisting of an angular stiffness function, damping, and a PID torque actuator. All of these joint constraints act in parallel to represent various muscle conditions. Using Choi's hypotheses [6,7] of active joint responses, the rotational stiffness of a joint is complemented by different joint damping profiles depending on whether muscles are tensed or relaxed, and a PID torque control is applied to the joint to model whether the subject is aware or unaware during loading. The active torque of the joint represents the resultant actions of all muscles, ligaments, tendons, and other human tissues which affect that joint behavior.



Tensed (co-contraction) Relaxed (single contraction)

Figure 3. Elbow jerk loading test by Choi[8]

To show this, volunteer tests were conducted [6] to measure the response of the elbow joint to the application of an initial static load followed by a jerk load while the muscles are tensed or relaxed. The experiment produced different joint damping resistances as shown by the graphs in Figure 3. The tensed state of muscle co-contraction produces both a flexion resistance (-1.5kNms/rad) to the static load and an equivalent extension resistance (1.5kNms/rad) when the jerk loading is applied. When the subject is relaxed with the muscles only in the single contraction state to resist the static load, the same flexion resistance is observed (-1.5kNms/rad), although the extension resistance to the jerk loading is greatly reduced (0.5kNms/rad). Therefore, the various states of muscle tensing can be reasonably modelled by a damping moment.

In addition to the muscle being relaxed or tensed, there are also the potential joint responses to the cognitive states of being aware or unaware of both an impending load or the actual loading 'per se'. Awareness generally leads to muscle tensing but during long duration loading events (such as a rollover accident), human instinct and reflexes mean that the subject cognitively tries to correct their posture and limb configurations to counter the forces acting on them. This reflex reaction is modelled by applying a torque actuator to the joint via a PID closed loop control method. At every cycle of the computation, the PID function makes an assessment of the proportional, integral, and derivative behavior of the joint relative to the target position. A very general explanation of PID control would say that the *proportional* is the current behavior, the *integral* is the historical behavior, and the *derivative* is the projected future. Based on this, the PID controller attempts to correct the system based on the error calculated at each cycle multiplied by the gain constants (k_p,k_i,k_d). The PID function is summarized by the function below.

Muscle model with Closed-Loop Control (PID)

$$e(t) = r(t) - y(t)$$

$$u(t) = k_p \cdot e(t) + k_i \cdot \int_0^t e(\tau)d\tau + k_d \cdot \frac{de(t)}{dt}$$

y(t): current state k_p: proportional gain
r(t): reference k_i: integral gain
u(t): control signal k_d: derivative gain

The gain parameters k_p, k_i , and k_d are obtained through an optimization processes by correlating the model's response to the active response exhibited by an aware subject in a volunteer test.

The joint stiffness properties of the aH-models exploit the knowledge gained from past human body model studies [6,7,8]. Each aH-model consists currently of 15 rigid body segments connected by 14 articulated joints. All joints are modelled with three rotational degrees of freedom with the exception of the elbow joints and the knee joints which only have one rotational degree of freedom. Active torque capability is modelled for all joints in all their degrees of freedom, for the aH-models similar to the elbow model described.

Joint Stiffness Scaling

The 50th percentile male is the most common human size used for compliance standards. This has led to it being the most widely documented and tested. For this reason, the aH-M50 model is the first to be calibrated and correlated with test data. The aH-M50 serves as the reference for the stiffness of the various body joints for this first generation of aH-models. To estimate the stiffness of joints for the aH-F05 and aH-M95, some general scaling can be made from the aH-M50. The stiffnesses of human joint articulations are dependent on the gender, body size, muscular structure, anatomical proportions and hence, some scaling assumptions can be used. The same can be said for the joint damping which models the muscle tensing strength. Therefore, the joint stiffness of the various joints for the aH-F05 and the aH-M95 can within reason be initially scaled and interpolated from the aH-M50 until additional data for calibration becomes available. The aH-M50 joint properties will be re-tuned regularly as more up-to-date data becomes available with the same being done on the aH-F05 and aH-M95.

Joint reflex modelled by PID controls are less scalable however, as instincts are not dependant on body size or muscle strength. Nonetheless, similar scaling of PID gain parameters can be used as a good first estimate.

CALIBRATION AND OPTIMISATION

Calibration and verification of the aH-M50 is currently being undertaken by correlating the models response to available test data and published experimental data.

Published data of post mortem human subject (PMHS) tests [10] and volunteer tests [3,11] provide some excellent references for correlating and assessing the performance of the aH-M50 model.

Some basic calibrations were initially performed on selected joint groups of local anatomy to optimize PID gain values that represent those of an aware subject. The joints were optimized locally for the head/neck system, the thoracic/lumbar spinal system, and the limbs. This was done by applying a relative low acceleration/deceleration pulse in selected loading directions, to the various anatomical systems. When the joints are defined only by angular stiffness functions, and using the head/neck system as an example, the segments would oscillate indefinitely in the direction of loading. When some damping was applied to the joints, the oscillations would come to rest after 1 to 2 cycles. A further application of torque actuators to the joints, through the PID closed loop control is applied to the system to represent natural human reflexes of an aware subject to stabilize oneself and resist the loading. The PID gain parameters were optimized with the objective of damping out the oscillations within approximately one cycle. This was considered a good first approximation of the human reflex contributions to resist such loading. Figure 3 shows some time history frame grabs of the head/neck system for non-active and active joints when the local joint systems are loaded in lateral bending. When the joints are active, the maximum lateral bending of the head/neck is reduced and is stabilized quicker, as well as returning to the neutral target position defined by the PID function.

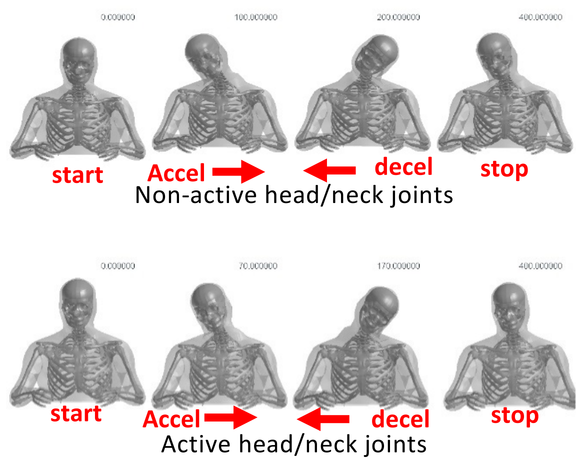


Figure 3. PID calibration of active joints in the head/neck system

Figure 4 shows a similar calibration where the lateral pulse loading is applied to the pelvis. Only the thoracic and lumbar joints are free to deform. As with the head/neck system calibration, PID activation of the thoracic/lumbar joints produced lower maximum deformations, earlier stabilization, and the model is returned to the target neutral posture.

Similar active torque calibrations were carried out for the hip/lower-limb and the shoulder/upper-limb systems. As data becomes available, particularly those regarding non-injurious loading of volunteers, such active joint torque calibrations can be refined to better represent bracing and reflex responses.

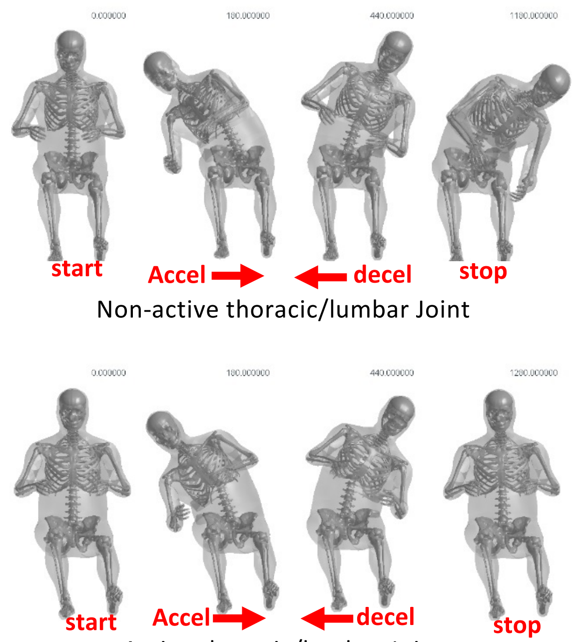


Figure 4. PID calibration of active joints in the Thoracic/lumbar system

HUMAN MODEL PEFORMANCE IN DYNAMIC ROLLOVER TEST SYSTEM (DRoTS)

DRoTS tests of PMHS [10] were used as an initial reference test to assess the performance of the full aH-M50 model. Several defined tests can be performed on this rollover test system. They include, a quasi-static test with 180^{o} rotation, an upside down drop and catch with 0^{o} rotation, a pure dynamic roll with 360^{o} rotation, a leading-side drop with 360^{o} rotation, and a trailing-side drop with 360^{o} rotation.

The case of pure dynamic roll, over 360^{o} was simulated with the aH-M50 in the leading and trailing seat positions. The loading conditions of this test allow the simulation of a rollover of a full rotation, which occurs over an extensive period of more than 1.5 seconds, where active joints are expected to play a role in the kinematics of the aH-model. Figure 5 shows the rollover test rig developed by the University of Virginia's Center for Applied Mechanics with occupants in the leading and trailing seat positions.

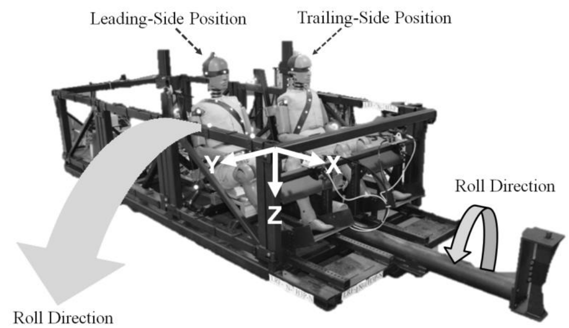


Figure 5. DRoTS system from the University of Virginia Center for Applied Biomechanics

Figure 6 shows some frame grabs at 0^{o}, 90^{o}, 180^{o}, and 270^{o} of the simulation for pure dynamic roll, with both the leading and trailing occupants. As in the PMHS tests, the aH-M50 models are each restrained with a three-point lap sash belt. In addition, the left and right hand are strapped to their respective left and right upper leg with the lower legs and feet restrained to the test rig.

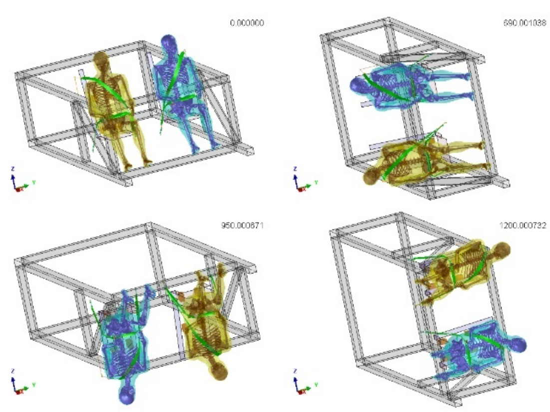


Figure 6. DRoTS simulation with two aH-M50 models

Of interest in the DRoTS test of pure roll is the kinematics of the spine during the entire rolling event. The overall performance of the aH-M50 can be assessed relatively simply by comparing the lateral bending of the head/neck system during a

PMHS physical test and its simulation using the aH-M50 model. Images at 45° intervals of angular rotation during the rollover, taken from high speed camera footage, were used to make these qualitative comparisons with the simulation.

Pure Roll Test (360°) Leading-Side Position

Figure 7 shows the comparison between the PMHS leading-side position test (column1) and its model equivalent of the aH-M50 with non-active joints (column2). Simulation results of an aH-M50 with active joints representing an aware human that is tensed with reactive reflexes (column 3) is also presented to study variations in kinematic response compared to an aH-M50 with non-active joints. The amount of lateral bending of the head/neck cannot be clearly observed from the photo frames at 225° and 270° for the leading-side position due to what appears to be visual obstruction of the camera view. A comparison of the head/neck lateral bending response between the leading-side PMHS test and the aH-M50 model with non-active joints show good general qualitative agreement in terms of the amount of head/neck angular rotation in lateral bending as well as being in phase. When the joints are activated in the aH-M50 model to simulate tensing and reflexive behavior, the amount of lateral bending deformation is reduced in all the frames shown. This reduction in lateral bending deformation of the head/neck system is most obvious at the frame rotations at 90°, 180°, and 360°. The probable reason for this is because the largest variation between an active and non-active model is observed at the higher loads levels, when the joint PID torque actuator is most affective. At lower load levels where the head/neck lateral bending displacements are lower, less variation is also observed between the aH-M50 with and without active joints.

Pure Roll Test (360°) Trailing-Side Position

Figure 8 shows the same comparison of the DRoTS pure roll test for the trailing-side position. The PMHS test results are shown in column 1 with the non-active aH-M50 in column2 and the active aH-M50 human model in column 3. For this trailing-side position test, the view of the PMHS head/neck kinematics is obscured in the 270° and 315° photo frames.

Cadaver Test | **non-active (cadaver)** | **Active (Human)**

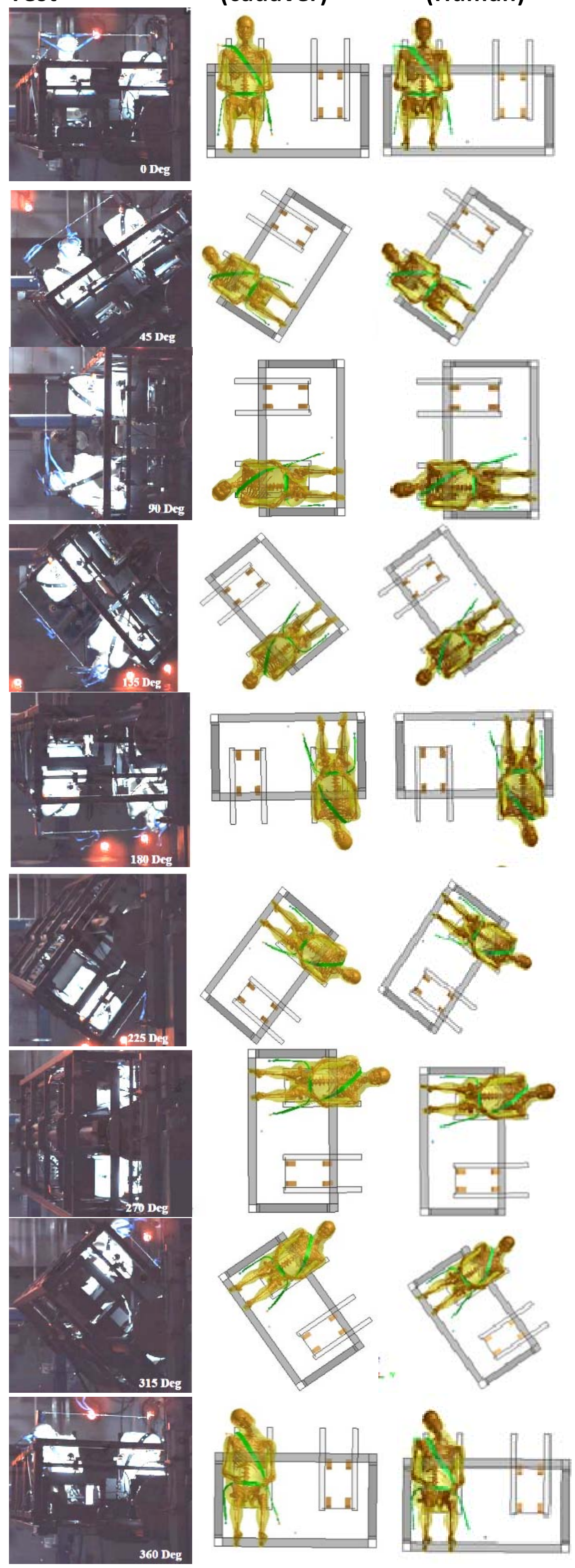


Figure 7. Pure dynamic roll of PMHS and aH-M50 in the leading-side position

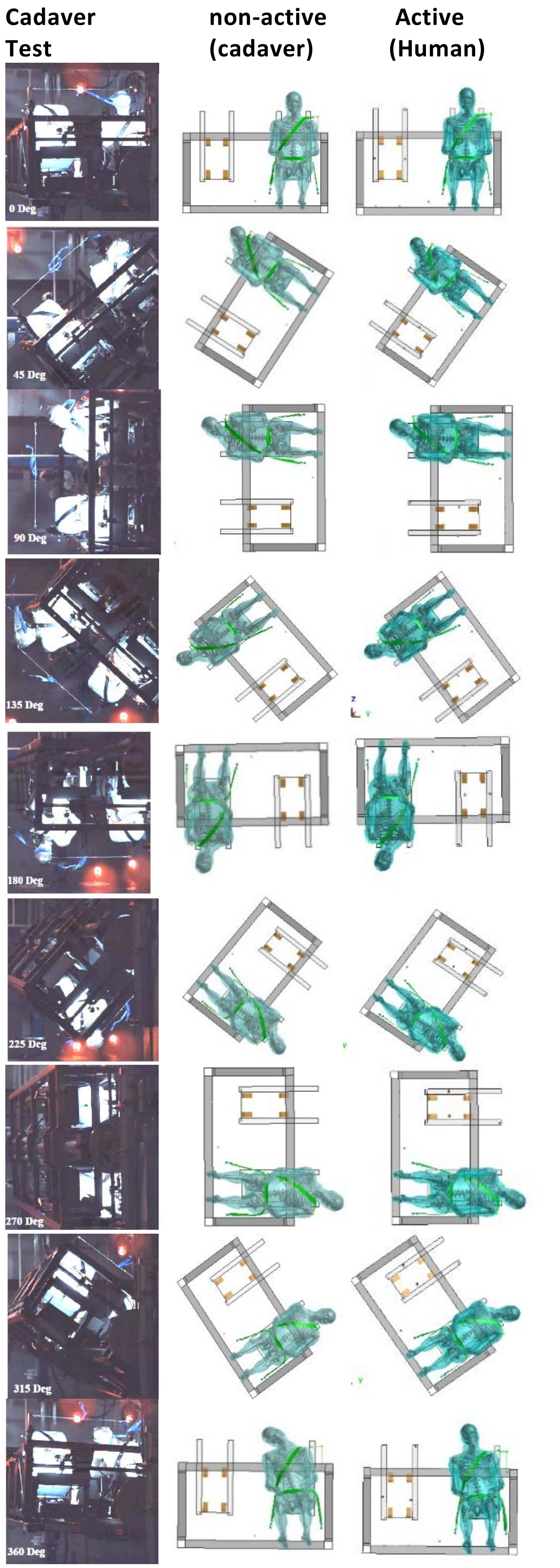


Figure 8. Pure dynamic roll of PMHS and aH-M50 in the trailing-side position

Qualitative comparison between the test results and the aH-M50 with non-active joints show good agreement of the lateral bending of head/neck in terms of general magnitude and timing. Unlike the leading-side position analyses, when the aH-M50 joints are activated for the trailing-side position analysis, the reduction in head/neck lateral bending is less pronounced. The only frames that showed an obvious difference were the 45° and 360° frames. What this highlights is that the loads experienced in the leading and trailing positions are quite different.

These two controlled pure rollover cases illustrate is that the aH-M50 model with active joints generally shows lower maximum bending of corresponding joints. Not obvious in these frame grabs is the earlier recovery of the joint angular displacements displayed by the active human model to try to return to its defined neutral (or target) position defined by the PID controller. Earlier recovery of joint angular displacements leads to changes in human body kinematics during the loading event. Taking the case of a vehicle rollover as an example, these kinematic changes can result in different levels of maximum human body joint deformations, timing of head impact or human body contact to the vehicle interior, and can change the location of the impact which can ultimately change the type of injury potential.

VEHICLE ROLLOVER

Vehicle rollovers by their nature are complex crash events that can be triggered by various combinations of driver behavior, road surface type and its interaction with the vehicle, and the size/height/weight of the vehicle. Rollover crashes occur over a long duration measured in seconds as opposed to general car crash durations of only a few hundred milliseconds. For these reasons, the numerical analysis of occupant behavior in vehicle rollovers have been less viable due to the high computing demands required to simulate such long duration events.

The numerical efficiency of the aH-M50 model has meant that simulating a long duration rollover analysis is attainable. Trial analyses were performed to assess the viability of simulating a full vehicle rollover with the aH-M50 human occupant model. The rollover arrangement simulated, involves the vehicle flipping over at 48km/h inducing approximately three rolls of the vehicle before coming to rest over a duration of five seconds. Figure 9 shows the various vehicle rollover states during one of the rollover simulations. Two simulations were carried out using this loading configuration, one with the aH-M50 with active joints and the other with non-active joints. The aH-M50 human models are seated in the driver side position for these simulations.

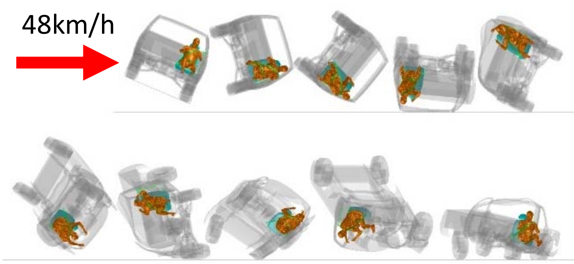


Figure 9. Whole vehicle rollover simulation with the aH-M50 model.

Two main factors directly affect the risk of injury to an occupant in a rollover. The first is the ability of the vehicle to maintain its structural integrity during the crash thereby preserving the occupant space. The second is the performance of the vehicle's internal safety and restraint systems such as seatbelts that restrain occupants and keep them away from hard surface impacts. With occupant ejection from the vehicle being a concern in rollover accidents, seatbelts also play a role in preventing this type of phenomena. More recently, the prevention of occupant ejection has been a secondary consideration in the design of curtain airbags [12].

For these trial simulations, a simple analysis of the results was undertaken to compare the effect of using a human body model with fully active joints against one that is non-active. The active joints were maintained throughout the rollover simulation in the first rollover case. This represents one extreme of the occupant being fully aware for the whole rollover duration. The second rollover case with non-active joints simulate the other extreme of the occupant being unaware throughout the accident duration. The graph in Figure 10 shows the contact force magnitude of the head to the interior surfaces of the vehicle for the first 2.5 seconds of the rollover. It can be seen that when the joints are modelled as active, the force of the first major contact (~600ms) made between the head and the interior of the vehicle is about 30% lower in comparison to the non-active model. This can be attributed to the resistive reflex nature of the active joints at the neck to correct its posture and reduce the amount of head/neck motion, all of which leads to a lower contact force. It is also noteworthy that the timing of the initial head contact force between the 2 cases is already slightly out of phase due to variations in the active joint response of the head/neck system. Beyond this first impact event, the head contact forces are significantly higher for the active joint case. The reasoning for this is that after the first main head impact, the kinematics of the two cases have varied enough that subsequent head contacts (between the two cases) are no longer in phase and are contacting the vehicle interior at different locations as well as from different velocity vectors from the head. A general hypothesis that can be made is that relatively reliable quantitative comparisons between models can be made up to the first major contact event. Beyond this, only qualitative comparisons are reasonable due to growing variations of the occupant kinematics with time after the crash event. This is not to say that these later occupant kinematics, are any less important as they determine the interaction of the occupant with the interior environment, interactions with other occupants, or even occupant retention.

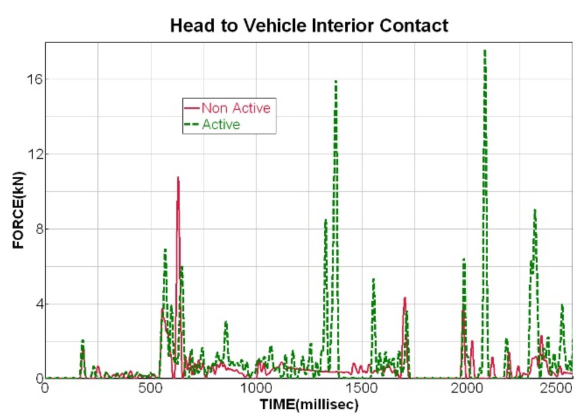


Figure 10. Contact force between the aH-M50 head with the vehicle interior

The example presented is an ideal extreme of active and non-active joints on the aH-M50 human model. In a real-world rollover, it is more likely that an occupant will experience different states of awareness throughout the rollover. One possible scenario could involve the driver bracing (active joints) at the beginning of the accident, but losing consciousness (non-active joints) during the rollover due to a head impact to the vehicle interior. Alternatively, a passenger may not be aware of an impending rollover and will brace much later during the rollover. These are just two examples of many possible scenarios in this type of long duration accident.

FUTURE MODELLING

This first series of aH-Models has taken advantage of past human body models and their joint stiffnesses. In addition, the models have been correlated to available muscle tensing studies, and optimized for projected active reflexive behavior. Although still relatively early in its development, their performance in these initial correlation studies show promise, especially in regard to the implementation of active joint torque behavior. Incorporating translational degrees of freedom (with stiffness properties) to selected spinal joints is the next logical step to refining the biomechanical response of these aH-models.

To better understand the reflexive behavior of aware subjects, it would be essential to calibrate such responses to human volunteer data. Volunteer experiments to measure joint reflexes would need to be performed at loads that are non-injurious. Although not ideal, such data will still be invaluable for projecting the anticipated reflex behavior under genuinely injurious loads using mathematical interpolation techniques.

The current models have been developed in a manner that increased sophistication can be retrofitted to the model. Possible retrofit options may include the addition of spinal complexity through additional joint articulations, or whole cervical or thoracic spine replacements. Some deformability of the pelvis or torso may also be developed as a retrofit option to better model seat and seatbelt interactions. However, the model's efficiency, which has been achieved by using simple modelling techniques would gradually be sacrificed as more complexity is added to the model. Recalling the motive for developing these active human models, the model's efficiency should be maintained when possible to set it apart from existing complex and very detailed human models.

Calibration and correlation is ongoing and will be updated as new data becomes available. It is anticipated that a database of tensing parameters and optimized PID gain values for reflexive strength will eventually be developed for specific loading conditions and severity.

CONCLUSIONS

Three active human models with active joint behavior have been developed. The series of active human models consist of the 5th percentile female, the 50th percentile male, and the 95th percentile male.

The performance of the 50th percentile model has been correlated to a limited selection of published PMHS and human volunteer test data with good general agreement. The models will need to undergo further correlation and calibration to extend their validity over a wider range of loading severities and loading types. This will involve further optimization of the active joint behavior with existing and future PMHS and human volunteer test data.

The aH-M50 model has been used to simulate a vehicle rollover undergoing three full rotations over a relatively long duration of five seconds.

The implementation of active joints in the active human models allows the simulation of human joint tensing and reflex behavior resulting from different states of human awareness. Active joints coupled with the overall model efficiency will allow the analysis of longer duration accident scenarios that account for complex human awareness reactions and ultimately will broaden potential fields of application.

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