THE SAFER HBM – A HUMAN BODY MODEL FOR SEAMLESS INTEGRATED OCCUPANT ANALYSIS FOR ALL ROAD USERS

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SUMMARY

The development of the SAFER human body model (HBM) started in 2008 and is still ongoing. SAFER HBM is an omni-directional model that can be tuned and scaled (morphed) to correspond to humans of different age, sex, weight, and stature. The model can be positioned to the posture of occupants, pedestrians, bicyclists, motorcyclists etc to enable analysis of road users inside as well as outside the vehicle. SAFER HBM is capable of predicting human kinematics in evasive maneuvers (low-g) as well as in crashes (high-g). The capabilities also include injury risk predictions in crashes.

The model has been thoroughly validated and used in numerous studies. Some examples: The effect of reversible pre-tensioning of the diagonal belt on occupant kinematics and injury risk for pre-crash evasive maneuvers followed by a crash, the influence of different postures and anthropometries on occupant kinematics, and the injury reducing benefits of a helmet in bicyclist to car impacts have been evaluated. Based on results from these studies, the SAFER HBM is considered to be an efficient and biofidelic tool for development and validation of protection systems for road users inside and outside the vehicle.

INTRODUCTION

Development of road user protection systems in crashes requires human surrogate tools capable of evaluating the human response for a large variation of human characteristics and crash configurations, such as impacts from all directions. Road users include vehicle occupants, bicyclists, motorcycle riders and pedestrians, or very generally any human using the road transport system. The tools should optimally be able to mimic human responses during the pre-crash phase, such as for instance posture maintenance during braking, and in the crash to predict kinematics, loads, and injury risk irrespective of crash configuration.

Protection systems for car occupants have historically been developed by means of anthropometric test devices (ATD) as human surrogates in crash testing. The ATDs are mechanical models of the human body with the inherent difficulties of representing living tissue with metal and rubber materials. In addition, the ATDs have to be robust to enable repeated use and therefore the capability to predict human fracture by means of breakable parts is limited. Injury risk is predicted by means of global measures such as chest deflection for prediction of chest injury risk (Laituri et al. 2005) or force and moment for neck injury risk (Eppinger et al. 1999). Furthermore, the ATDs are developed to predict human kinematics and injury risk for a specific crash direction, such as frontal impact for the HIII and THOR ATDs or side impact for the WorldSID and SIDIIs. For evaluation of safety systems for other road users, only few ATDs have been developed. One such example is the POLAR dummy which was developed for pedestrian impact evaluations (Akiyama et al. 2001).

Mathematical human body models (HBMs) are tools that can overcome some of the limitations with ATDs and have proven to be valuable tools in injury biomechanics. The mathematical HBMs, can be built as anatomically accurate computer models from magnetic resonance imaging (MRI), computed tomography (CT) and other medical scanning techniques (Gayzik et al. 2009). The models can predict how the body's bones, organs and connective tissues react to external objects and forces. Using the Finite Element (FE) method, HBMs can also represent the actual human anatomy and material properties in detail, and internal stresses and strains of tissues in the human body can be evaluated. As such, HBMs have the potential to have advantages over ATD models also for injury prediction. An HBM can be made omnidirectional by design and can therefore predict human kinematics and injury risk for all crash directions, including those that are in-between the pure frontal, side, and rear impacts, usually referred as oblique crashes. In the HBM the injury risk can be evaluated for specific anatomical structures, which makes it possible to address injury with physical parameters related the injury mechanism and at a detailed level, such as predicting rib fracture risk by means of predicted strain in the ribs in an HBM (Forman et al. 2012).

The development of HBMs started with development of individual human body parts, such as the lower extremity, thorax, neck, and head/brain. For instance, Kitagawa et al. (2001) developed a lower extremity FE model in order to elucidate injury mechanism of the ankle and tibia in frontal impacts. Aiman et al. (1999) developed a human head/brain FE model in order to estimate the brain/skull relative displacement magnitude due to blunt head impact. The models were used to understand local injury mechanisms. However, in real-world crashes, especially fatal ones, various types of loads on the occupants may often cause multiple injuries in several body parts. For examples, occupants contact components in the vehicles such as the steering wheel, windshield, side doors, instrumental panel and/or even an adjacent occupant during the crash sequence. As a result, occupants can sustain multiple injuries, one of which sometimes can be fatal. Therefore, it is important to simulate gross motion and overall multiple injuries of the human whole body at the same time.

The first FE whole body human model was developed by Huang et al. (1994). The model had a detailed geometrical representation of the rib cage, whereas the contour of the head, neck, shoulder, pelvis. and limbs were greatly simplified. A more detailed FE whole body human model was presented by Ruan et al. (2003). The model had a much more detailed geometric representation compared to the model described above. Its geometry was primarily based on the Visible Human Project, and the topographies from human body anatomical texts. Detailed head, neck, ribcage, abdomen, thoracic and lumbar spine, internal organs, pelvis, upper and lower extremities were simulated in this model. In the European project HUMOS (Robin, 2001) a finite element whole body human model in a driving seating posture was developed. The skin, bones, muscles as well as the main organs (lungs, heart, liver, kidneys, intestine etc.) were represented in detail in the model.

At the present, there are two major, available, occupant FE-HBM families: the Global Human Body Model Consortium (GHBMC) models and the Total HUman Modes for Safety (THUMS) models. The GHBMC includes 5th female, 50th male and 95th percentile male ‘detailed’ occupant models (Park et al. 2014; Davis et al. 2015; Davis et al. 2016) and ‘simplified’ occupant models Schwartz et al. (2015). THUMS comprises 5th female, 50th male and 95th percentile male models (Watanabe et al. 2012). In addition to these two commercial models, other initiatives to create Open Source (OS) license HBMs have been pursued. For instance, a 50th female HBM, the VIVA model was developed (Östh et al. 2017). A 50th percentile male model was created by morphing of the VIVA+ 50th percentile female model. An additional HBM is the SAFER HBM, which is the focus of the present paper. The evolution of the SAFER HBM from 2008 until today is summarized. The development, validation, morphing capability, as well as the capability of the model to represent all road users are discussed. Additionally, a ribcage subsystem response validation study, with focus on rib fracture prediction is included.

THE SAFER HBM

The overall objective with SAFER HBM, formulated in 2008 is to develop an “omnidirectional, tunable and scalable human body model capable of injury risk and biofidelic kinematics prediction in high-g as well as low-g events”. It means a tool that can be used for seamless integrated occupant analysis. Targeting this, several parallel research activities have contributed to taking steps towards fulfilling this objective. The development journey of the SAFER HBM is described in this chapter, with the development of the mid-size male model summarized in the first subchapter, including the steps from car occupant model to models for other road users. This is then followed by a summary of the developments of the injury prediction capabilities and a new study demonstrating the capability of SAFER HBM to predict rib fracture risk in oblique loading. The scaling and tuning (morphing) functionalities are addressed in the subchapter of morphing, whereafter its capability to address positioning, including postural control follows. Lastly, examples of the SAFER HBMs applications are summarized.

Design

The baseline SAFER HBM corresponds to a 50th percentile male, with a weight of 77 kg and stature of 175 cm (Schneider et al. 1983). The development of the model started with modifications to the THUMS v3, for instance material properties for low-g events (Östh et al. 2012) and the ribcage (Mendoza-Vazquez et al. 2013).

In the beginning of 2017 the SAFER HBM v8, or THUMS SAFER as it was then called, was created by major updates to the head and thorax of the model, see Figure 1. The Royal Institute of Technology FE head model (KTH model, Kleiven 2007) was integrated with the SAFER HBM. The KTH head model consists of 25,000 elements, of which approximately 4,000 are solid elements belonging to the brain which is compartmentalized and with specific material properties for the regions of the brain such as the white and gray matter, the corpus callosum, the cerebellum and the cerebrospinal fluid. The head model utilizes hyperelastic Ogden materials with Prony series viscoelasticity and has been validated with respect to Post Mortem Human Subject (PMHS) data for intracranial pressure as well as for frontal, occipital and lateral impacts (Kleiven and von Holst 2002a; Kleiven and von Holst 2002b; Kleiven and Hardy 2002; Kleiven 2006), and injury risk curves for mild traumatic brain injury in the form of concussion has been developed using accident reconstructions of NFL football head impacts (Kleiven 2007; Fahlstedt et al. 2022).

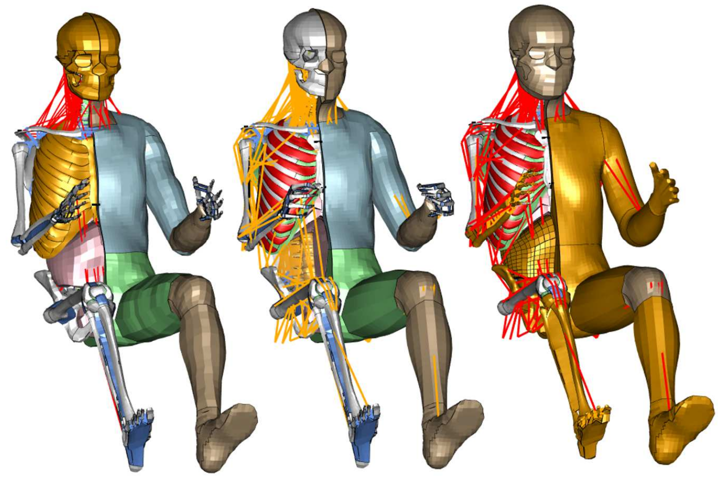

Figure 1. The three versions of the mid-sizes male SAFER HBM left to right; v8 (2017), v9 (2019) v10 (2020) with updates for each version highlighted in gold.

Additionally, for SAFER HBM v8 a detailed generic rib cage model was created based on a combination of in-vitro and in-vivo datasets (Iraeus et al 2020, Iraeus et al. 2019). Generic rib models were developed by defining elliptical cross sections along the rib center line. Cortical thickness, varying along and around each rib, was assigned based on the dataset presented by Choi et al. (2011). The ellipses were connected to define the periosteal and endosteal surfaces. Thereafter the ribs were meshed and morphed into the final shape based on a statistical ribcage model (Shi et al. 2014). Additionally, the sternum was re-modelled based on a geometry from another statistical shape model, (Weaver 2014). The parameters for the statistical model were adjusted to be representative of a male aged 40 years with a stature of 1,770 mm and a BMI of 25. The ribs were connected using three layers of solid elements representing the intercostal muscles (Figure 2). Validation of ribcage kinetic, kinematics and strain distribution were carried out at three levels of complexity: anterior-posterior rib bending tests; rigid impactor table-top test; and a 40km/h frontal sled test (Iraeus et al. 2019).

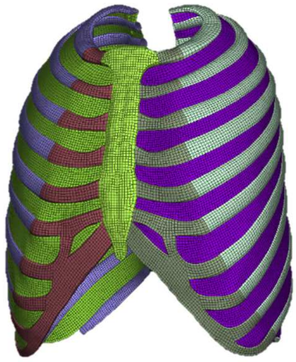

Figure 2. The new generic ribcage developed for SAFER HBM v8.

Further updates for the SAFER HBM were made in 2018, when the feedback postural control package (Östh et al. 2012b; 2015; Ólafsdóttir et al. 2019) developed for the model was integrated with the up until then passive HBM, creating the SAFER HBM v9 which can be used as both an active and a passive HBM through seamless control with switching parameters in the model. Feedback postural control is applied for the cervical and lumbar spine using a spatial tuning pattern for omni-directional control which was later validated for braking and lane change evasive maneuvers (Larsson et al. 2019). Soft tissue material updates were made to accommodate a combined active and passive model. For instance, the skin was converted from a linear elastic material to a non-linear material model using. Additional updates for v9 were updates of the lumbar and cervical spine, which was given new vertebral discs and ligaments, and for the cervical spine the facet joints were converted from shearing solids to sliding contact joints (Östh et al. 2020), and validation of the spinal properties were carried out with respect to functional spine unit properties, subsystem and whole-body responses in lumbar flexion-extension.

For version 10 of the SAFER HBM, created in 2020, the model had updates to the pelvis, head and brain, lower arm, and lower leg, plus the soft tissues around the torso were remeshed. The pelvis was updated based on a new statistical shape model (Brynskog et al. 2021). The KTH head model was also updated to a version also encompassing the brainstem. Additional updates were remeshing of the torso soft tissues, to create a continuous mesh over the shoulder and hip joints. The torso soft tissues were adapted to an average 50th percentile male shape (Reed and Ebert 2013; Pipkorn et al. 2021) and a major update of the contacts of the model was made to improve model reproducibility and computational efficiency (Östh et al. 2021). Furthermore, the radius, ulna, carpals, metacarpals, phalanges, and ligaments were replaced with a new model (Bayat and Pongiaporte 2020) with skeletal part based on medical images, and ligaments and joints based on literature. The radius, ulna and hand were positioned in a driving posture matching that of the existing lower arm in the SAFER HBM, and soft tissues and skin were added on top of the skeletal structure of the lower arms. The lower arm model was validated by means of impact tests (Forman et al. 2014; Dumas et al. 2003). In addition, the leg was updated. The mesh of the tibia, fibula, calcaneus, and talus of the lower legs were replaced by a pure hexahedral mesh representing the trabecular bone and a quadrilateral shell mesh representing the cortical bone. The updated lower leg geometry was based on CT-Scans from 5 females (Roberts 2020) rescaled to fit the 50th percentile male. The cortex of the intermediate bones, metatarsals and phalanges of the feet was modelled using rigid shells, replacing the original mesh, while the original ligaments and soft tissues of the lower legs were kept.

To enable evaluation of pedestrian safety the SAFER HBM v10 was positioned as a pedestrian in a walking gait (Figure 3). For bicyclist and motorcycle safety the feet of the SAFER HBM were positioned on the pedals and foot support respectively and the spine curvature was modified for the model to reach the handlebar. For the pedestrian posture the pelvis flesh was morphed to accommodate for the modification of pelvis shape when a human changing posture from a standing posture to a sitting.

Figure . The SAFER HBM v10 pedestrian

Injury Prediction

The SAFER HBM has the capabilities to predict injury risks by means of global measures such as cross-sectional forces. However, detailed biofidelic prediction of injury risk using injury metrics that are physically related to the injury mechanism is in focus. Such injury metrics can be used to predict injury risk equally well regardless of loading directions with the benefits that the same injury metric can be used to predict injury risk for all road users.

For the SAFER HBM, a strain-based probabilistic method to predict rib fracture risk with whole-body FE models was developed (Forman et al. 2012). An age-adjusted ultimate strain distribution based on cortical bone coupon testing (Kemper et al. 2005, Kemper et al. 2007) was used to estimate local rib fracture probabilities within SAFER HBM. These local probabilities were combined by means of a poisson binominal distribution to predict injury risk and severity within the whole ribcage. The strain based probabilistic rib fracture method was refined with new risk functions (Larsson et al 2021). The capability of the probabilistic rib fracture method to predict rib fracture risk was validated by means of detailed accident reconstructions and population-based reconstructions (Pipkorn et al 2019, Larsson et al 2021. Overall SAFER HBM version 10 with probabilistic rib fracture prediction method was capable to predict rib fracture risk based on the reconstructions.

For the present paper, an additional demonstration of the capability of the SAFER HBM v10 model to predict rib fracture risk was carried out by a ribcage subsystem response validation study. Oblique pendulum PMHS impact tests were reconstructed and the predicted rib fracture risk with the number of fracture ribs sustained by the PMHS were compared (Viano, 1989) (Figure 4).

In the pendulum impact simulations, the impact velocity was 4.4m/s (low severity) and 6.5m/s (mid severity), the mass of the impactor was 23.4kg and the diameter was 150 mm. The center of the impactor was aligned with the xiphoid process and rotated 30 degrees. In the PMHS test study, five tests were carried out for each impact velocity, and one female and four males were tested for each velocity. Acceleration was measured in the impactor and force was calculated from the acceleration. Force-Time corridors for oblique impacts at 4.4 and 6.5m/s were generated.

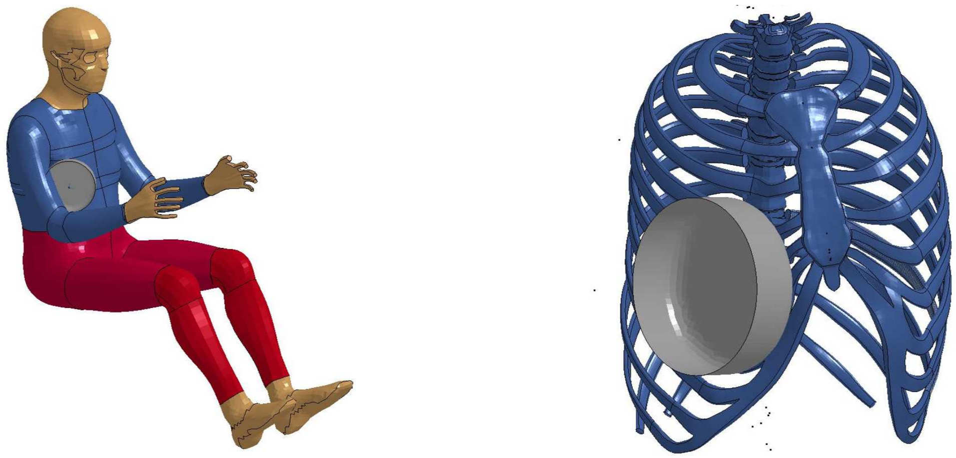

Figure 4. Validation of ribcage subsystem response by means of oblique pendulum impact tests according to Viano (1989)

The forces predicted using SAFER HBM v10 were inside or close to the experimental force-time corridors for both impact velocities (Figures 5a and 5b). In the 4.4m/s impact tests the average age of the PMHS was 46 year old and the number of rib fractures were 0.4. In the 6.5m/s impact tests the average age was 54 and average number of rib fractures were 5.4. In the corresponding simulations the SAFER HBM predicted a 2% risk for a 46 year old, and a 94% risk for a 54 year old to sustain more than 3 fractured ribs (NFR 3+) in 4.4 and 6.5m/s pendulum impacts respectively (Table I).

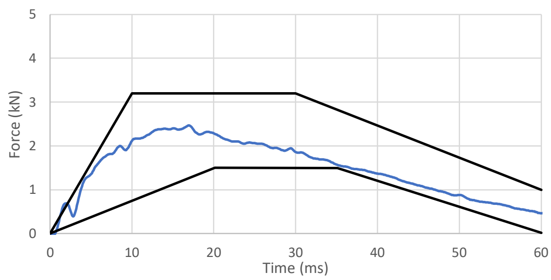

Figure 5a. Impactor force vs time for the 4.4m/s oblique pendulum impact test, corridor from Viano (1989)

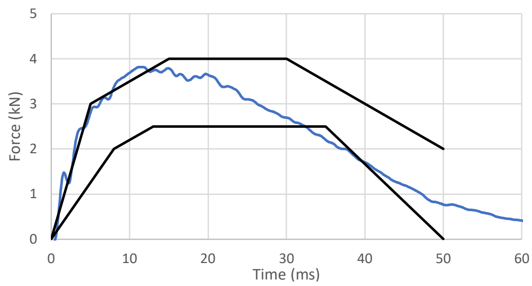

Figure 5b. Impactor force vs time for the 6.5m/s oblique pendulum impact test, corridor from Viano (1989)

Table 1.
Fractures in PMHS tests (Viano 1989) vs predicted rib fracture risk from the simulations in the oblique pendulum impact tests

Average Age	**Average Impactor Speed**	**Average No. of Fractures**	**Predicted NFR3+ Risk**
46 years	4.4m/s	0.4	2%
54 years	6.5m/s	5.2	94%

To compare various predictors for mild traumatic brain injuries 58 NFL head impact cases were reconstructed with the head model. A statistical correlation between numerous predictors with injury was found (Kleiven 2007). However, strain, was found to be the preferred predictor to be used to predict the risk for mild traumatic brain injury (Fahlstedt et al. 2022). The value for 100th, 99th, 95th, 90th, and 50th percentile for element and nodal averaged element strain was evaluated. It was found that 100th percentile strain with element value could give high strain due to only a few elements in a model. Therefore, the 99th percentile strain was selected to predict brain injury risk.

The capability for the lower arm model to predict fracture risk was developed by means of reconstructing two different human subject lower arm impact tests (Bayat 2020). In one configuration the arm was mounted to a force transducer and a reaction plate assembly free to move along linear guide rails. The impact load was provided by a guided mass dropped from various heights (Forman et al. 2014). In the other configuration a horizontal arm supported by two cables was impacted with a pneumatic piston at various impact velocities (Dumas et al. 2003). Based on these two test configurations with human subjects the capability to predict lower arm fracture risk by means of predicted lower arm force at the elbow was developed.

Morphing

A morphing method to enable creation of the individuals in the diverse population was selected and implemented for the SAFER HBM (Hwang, et al. 2016). The morphing method involves several steps. For a given age, sex, stature and body mass index (BMI) the external shape geometries are predicted by means of statistical geometry models developed by Reed and Parkinson (2008), while the skeleton geometries are based on other statistical shape models, ribcage (Wang 2016), pelvis femur and tibia (Klein 2015). These geometry models are further aligned to anatomical landmarks on each bone by means of a rigid registration algorithm. The predicted geometries are then used to define target landmarks, with corresponding source landmarks defined according to the baseline SAFER HBM geometry. The morphing is carried out by means of radial basis function (RBF) interpolation. By using this method, SAFER HBM can be morphed to populations of vehicle occupants to be used to develop protection systems that provides all road users with good protection.

Positioning

Starting from version 9, positioning of the SAFER HBM v9 as an occupant is carried out by the use of a dummy mechanism and a javascript in PRIMER (Figure 6) (Oasys Ltd, Sollihull, United Kingdom). The PRIMER dummy mechanism enables manipulations of the head, neck, torso, as well as the upper and lower extremities using rigid body joint definitions in PRIMER. Based on the final position 1D cable elements are created which then are used in a Marionette-method pre-simulation to position the HBM in the desired posture (Figure 6) (Poulard et al. 2015). The pre-simulation is typically 300 milliseconds long and run with global damping to generate a close to velocity free end-state for the model. After the pre-simulation, the nodal positions are read out and carried over to the baseline model, thus omitting any initial stresses and strains created in the positioning pre-simulation.

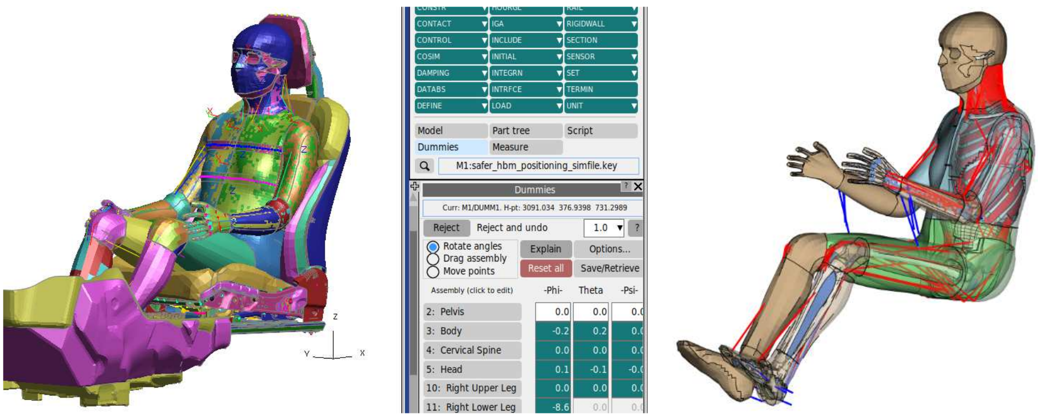

Figure 6. SAFER HBM v10 positioning using the Marionette-method and PRIMER. Left: Dummy positioning interface in Oasys Primer. Right: Initial position in Marionette-positioning pre-simulation to a passenger posture. Blue lines represent the positioning beams.

The tool is capable of modifying the posture of SAFER HBM occupant to the posture of all road users. The Marionett-method described above is used in steps to enable positioning of SAFER HBM to a pedestrian, bicyclist, or motorcyclist posture. For the lower extremities, initially the femurs are pulled to the desired position followed by the tibias.

Postural Control and Active Muscles

Real-world crashes occur in a variety of situations, often including a combination of low- and high-g events. Exemplified by braking followed by a frontal impact, a rear-end impact followed by a frontal impact, or a complex run off road crash. Creating a HBM capable of this includes several challenges, such as developing a model that is capable of predicting occupant kinematics for the low load and long duration events and injury risk for the high-g events. To enable time efficient analysis of the combined events requires careful selection of the modelling detail. The modelling detail is a compromise between the number of elements to avoid long run time while have a sufficient level of detail to enable reliable prediction of injury risk. To enable simulation of whole-sequence crash events including a low-g pre-crash phase, the first active musculature implementation for the SAFER HBM was made for an isolated arm to show proof of concept of using feedback postural control in an FE HBM (Östh et al. 2012a). The existing elbow joint in the THUMS v3 arm was replaced with a revolute joint, and the upper extremity muscles spanning the shoulder and elbow joints were modelled using line muscle elements. The updated model's passive properties were verified with respect to human subject data, and both posture maintenance and elbow perturbation experiments were simulated with the model.

The elbow model with active control was later implemented into the whole body HBM, together with postural flexion-extension controllers for the head, neck, and trunk (Östh et al. 2012b; Östh et al. 2015). The head, neck and trunk controllers were first iteratively tuned to the occupant response in light braking interventions and sled tests (Östh et al. 2012b), and some modifications to the baseline THUMS v3 were made to soften the passive model's response. Nodal constraints for skin nodes were removed, Young's modulus for the skin was softened, spinal ligament and intervertebral discs stiffness were reduced and buttocks soft tissue material was changed to the same as for the torso (Östh et al. 2012b)

To enable simulation of both driver and passenger occupants, the elbow model complemented with flexion-extension controllers for the shoulder were integrated (Östh et al. 2015) into the modified THUMS v3 which was the basis for the feedback postural control HBM at the time. The controllers were tuned through an optimization of the controller gains in a sled test series with volunteers in a driver environment, and later validated with respect to both driver and passenger volunteer responses to autonomous braking interventions (Östh et al. 2013; Ólafsdóttir et al. 2013). The head and T1 trajectories of the HBM was within one standard deviation of the average volunteer response for the simulated conditions (driver and passengers in autonomous braking, with and without a reversible pre-tensioned belt), but for most conditions the model was on the more flexible side of the volunteer responses.

In the analysis of volunteer data from both driver initiated and autonomous braking, a different type of response was found. When braking on their own, drivers extended their arms and neck, and braced prior to the braking, while during autonomous braking drivers reacted with postural response first when braking was applied (Östh et al. 2013). This was simulated with the HBM by the inclusion of open loop controlled leg musculature (Östh et al. 2014), which was prescribed published muscle activations from volunteer tests (Behr et al. 2010), in combination with a hypothesized anticipatory postural response which was modelled by using a time dependent reference signal for the postural controllers as a function of the applied acceleration. This method was successful in reproducing the driver responses, while the postural feedback control which worked well for autonomous braking was not able to do so.

To extend the analysis capability of the HBM to lateral and oblique planar movements, spatial tuning patterns for the cervical muscles were derived from volunteer tests with intramuscular electrodes in seated perturbation tests (Ólafsdóttir et al. 2015). The spatial tuning patterns were implemented into the feedback postural control scheme for the SAFER HBM (Ólafsdóttir et al. 2019), and the response of the model for both longitudinal and lateral evasive maneuvers was validated by means of volunteer data (Figure 7) (Larsson et al. 2019).

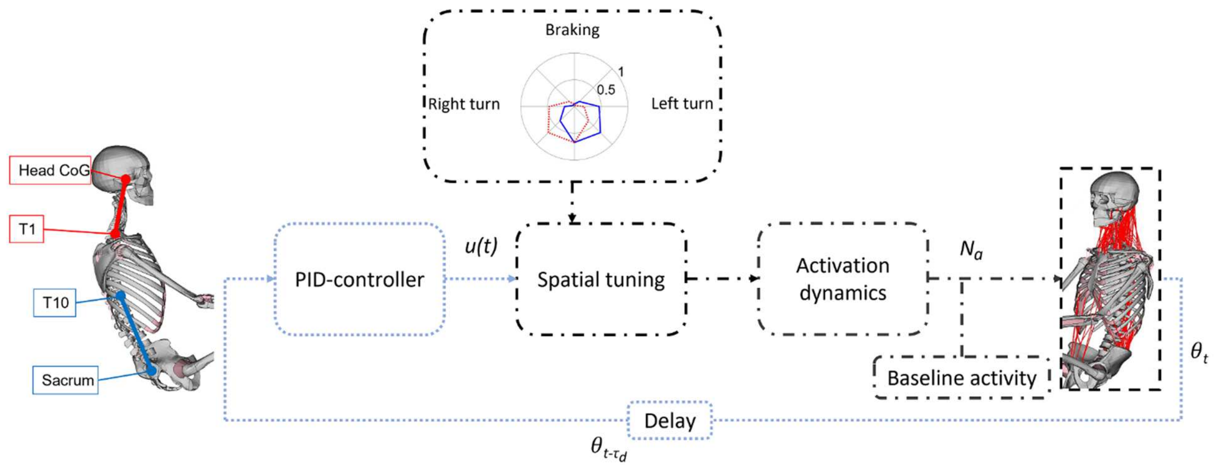

Figure 7. Schematic representation of the postural feedback control model for the SAFER HBM v10. The blue dotted lines represent signals common for the whole control system and the black dash-dotted lines represent signals on the muscle specific activation level. Adapted from Östh et al. (2022).

Recently, the active SAFER HBM v10 was used to study the effect of the feedback postural controllers on injury prediction in whole-sequence crash simulations (Östh et al. 2022), that is a pre-crash maneuver followed by a crash. The study showed that for frontal impacts, the muscle activation did not have any considerable effect on the kinematic response, however HBM injury predictions were affected in a higher acceleration frontal impact. For far-side impact, there was a moderate effect on kinematics in the form of reduced peak inboard head excursion with active muscles. The recommendation from the study was that muscle activations should also be included in the crash-phase of whole-sequence simulations to enable HBM simulation to better represent live occupants. However, as the postural control algorithm appears to give high muscle activations during the crash-phase, a strategy to hold the muscle activations constant during the crash was recommended and is used with the SAFER HBM for now.

Applications

As the first model to seamlessly predict human response in an evasive maneuver followed by a crash, the SAFER HBM was positioned in the driver seat in a vehicle interior model of a mid-sized passenger car, with an integrated restraint system in the form of a reversible pre-tensioned seat belt (Östmann and Jakobsson 2016; Saito et al. 2016). Demonstrating the feasibility to use the model, one study reported small effects on the occupant crash response from the reversible pre-tensioned seat belt (Östmann and Jakobsson 2016), while reducing the impact velocity as an effect of the pre-crash braking reduced occupant accelerations in the crash considerably. The other study showed reduced forward excursions during the pre-crash phase with increased reversible pre-tension force, as well as reduced chest deflections in the frontal impact simulated, as chest contact to the steering wheel rim was avoided when applying the reversible pre-tension (Saito et al. 2016). The capability of the SAFER HBM was further demonstrated in a more recent study (Wass et al. 2022), in which the model was used to study whole-sequence events consisting of a pre-crash braking followed by a far-side side-impact pulse. The braking intervention led to a more forward occupant position at the start of the crash phase, influencing results in the crash phase, compared to the impact only. A more forward impact point on the vehicle side resulted in 50 mm lowered lateral head excursion, compared to a mid-compartment impact configuration point.

Another strength of HBMs is that they can be used to represent a larger range of occupant positions and sizes. Several studies with the SAFER HBM have explored the influence of non-nominal seating postures and occupant sizes. Thirty-five postural variations for the 50th percentile male SAFER HBM v9 were simulated in three crash configurations representative of possible future crash scenarios (Leledakis et al. 2021). It was found that occupant postures such as crossing the legs increased pelvis excursions compared to a nominal position, that leaning inboard and outboard affected side impact kinematics considerably, as well as that resting the arm on the center console can increase inboard excursions in far-side impacts, compared to if the arm is on the outboard side of it. Expanding this study also to eleven male and female anthropometries, by morphing of the SAFER HBM v9 showed that seat fore-aft position had considerable effects on occupant speed relative to the vehicle interior in frontal impacts. Furthermore, knee restraint (depending on seat adjustment and occupant size) had large effects on femur, pelvis, and lumbar spine loads (Figure 8) (Leledakis et al. 2022).

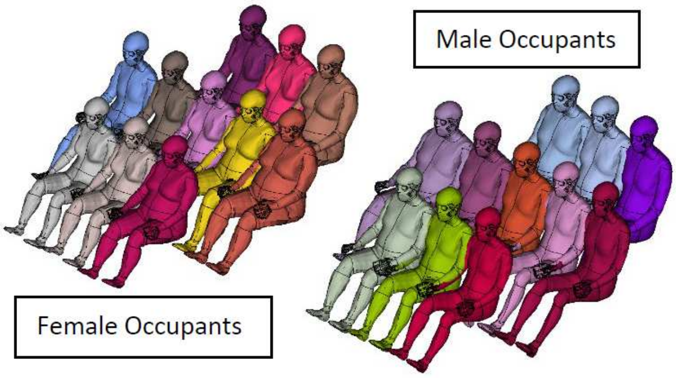

Figure 8. Eleven female and male anthropometries in the range from 1.48–1.90 m and 50–119 kg achieved through morphing of the SAFER HBM v9 and used to study occupant crash responses in four impact scenarios by Leledakis et al. (2022). Figure adapted from Leledakis et al. (2022).

SAFER HBM was also used to evaluate bicyclist safety. The potential injury reducing benefits of a bicycle helmet and bicyclist airbag mounted on the vehicle was evaluated (Pipkorn et al. 2020). SAFER HBM positioned on a bicycle model and impacted from the side in 40km/h (Figure 9). Brain injury risk was predicted by means of strain in the brain. It was found that generally head injury risk was reduced for a bicyclist wearing a helmet when impacted by a passenger vehicle in 40km/h. Additional reductions was obtained for a vehicle with a bicyclist airbag.

Figure 9. SAFER HBM v9 with helmet on bike impacted by a passenger vehicle travelling at 40km/h.

DISCUSSION

Starting with the overall objective to create a human surrogate model which is “omnidirectional, tunable and scalable with injury prediction capability, and biofidelic kinematics for high-g as well as low-g events” the SAFER HBM has become a capable tool to predict human kinematics in evasive maneuvers and in crashes, and injury risk in crashes. Validation of the capability of SAFER HBM models morphed to correspond to humans of various sex and sizes to predict human kinematics and rib fracture risk for both frontal and lateral loading is ongoing. Additional validations include the capability of the lumbar spine model to predict human lumbar spine vertebrae kinematics. Ongoing developments include tissue based risk functions for lumbar spine vertebrae and pelvis iliac wing fracture risks predictions. Additional development and validation efforts are to enhance the capability of SAFER HBM to predict occupant submarining.

The combination of the capabilities to predict human kinematics in evasive maneuvers, and to predict injury risks in addition to kinematics in crashes gives a model which is unique. The model can be used to evaluate whole-sequence crashes consisting of a pre-crash and crash phase efficiently. The level of modelling detail has been selected to be a compromise between short simulation run times and detailed representations of parts important for injury risk prediction. A pre-crash evasive maneuver can be up to 2 seconds in duration and the computer run time for an FE simulation with an HBM is roughly proportional to the number of elements. Therefore, for time efficient analysis of the pre-crash phase, an HBM comprising a minimum number of elements is preferred, still enabling biofidelic prediction of human kinematics in the pre-crash phase and injury risk prediction in the crash phase.

The latest version of SAFER HBM (v 10) contains approximately 410,000 elements. This is significantly less than the latest version of the GHBMC 'detailed' and the THUMS models which contain 2,3M and 1,9M elements, respectively. The relatively low number of elements in SAFER HBM makes the model an efficient tool for analysis of integrated pre-crash and crash events.

The SAFER HBM strategy is to develop a model that can be morphed to a family of models based on sex, weight, and stature that covers the population of road users. This strategy is different from the GHBMC and THUMS model development strategies in which individual models for vehicle occupants and pedestrians as well as models with fixed percentiles, such as a 5th female model, are developed (Park et al. 2014; Davis et al. 2015; Davis et al. 2016; Watanabe et al. 2012). Developing one model and morphing it to different sex, anthropometries and posture is an efficient method to create these models, as compared to developing models for fixed sex and percentiles. One of the main advantages of this strategy is that a variety of morphed models can be used to assess robust protection, by adapting for the situations and addressing equity of the population exposed.

The THUMS v3 was in 2008 capable of predicting in crash gross human kinematics (Toyota, 2008). The development of the capabilities to predict human kinematics in pre-crash events was initiated based on THUMS v3. The first studies were published in 2012, and by 2015 the SAFER HBM model was developed and validated for predicting human kinematics in autonomous and driver initiated braking maneuvers (Östh et al. 2015). The development of the capability of the model to predict pre-crash lateral evasive maneuver was the next step. For the validation, several series of volunteer tests were carried out (Östh et al. 2013; Ólafsdóttir et al. 2013; Ghaffari et al. 2018; 2019; 2021; Larsson et al. 2022). A model capable of predicting human kinematics for longitudinal as well as lateral maneuvers was available in 2017. Development of the capability of the model to predict human response for vertical load is ongoing and is expected to be finished by 2023.

Euro NCAP has announced inclusion of HBMs for virtual testing in the rating program. To enable inclusion, building trust in the models is an important challenge. Another important challenge is getting acceptance of the model in the community. SAFER HBM is a research-based tool and in order to build trust and get acceptance, the developments are documented and published in scientific journals.

HBMs have been mainly developed and used for vehicle occupant analyses. Development and validations, but not as extensive as for occupant analyses, have also been carried out for prediction of the response of road users outside the vehicle. These developments are important due to the fact that there are no commercially available mechanical human substitutes developed for analysis of road users outside the vehicle. Therefore, validated and biofidelic HBMs are sensible alternatives for analysis of road users outside the vehicle.

SAFER HBM is today an efficient tool capable of predicting human kinematics and injury risk for road users inside and outside the vehicle. In addition to the further developments and validations needed for enhancements addressing road users outside the vehicle, the refinements also include the occupants inside the vehicles, while improving the tool's capability to predict occupant kinematics and injury risks. Challenges for occupant protection include a larger variety of seat positions, including reclined, and sitting postures, also driven by the relative increase of vehicle passenger (non-drivers) with the trends of shared mobility and potential future autonomous vehicles. A human body model designed to replicate human biomechanics and kinematics in a variety of scenarios can help assess and develop safety systems for human-centered real-world protection.

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