# MUSCLE ACTIVATION STRATEGIES IN HUMAN BODY MODELS FOR THE DEVELOPMENT OF INTEGRATED SAFETY

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## ABSTRACT

Human Body Models (HBMs) have been used in crash safety research for some time, and are now emerging as tools for the development of restraints systems. One important challenge in the development of advanced restraint systems is to integrate sensory information about the pre-crash phase (time to collision, impact speed and direction, occupant position) to alter restraint activation parameters. Restraint activation can begin even before the beginning of an impact, providing additional time to reposition or restrain the occupant. However, any such pre-crash intervention would invoke a muscle response that needs to be taken into account in HBMs used in simulation of integrated restraints.

The objective of this paper is to provide an update on state-of-the-art modeling techniques for active musculature in HBMs. Examples of applications are presented, to illustrate future challenges in modeling of car occupants muscle responses to restraint activation.

The most common approach for modeling active muscle force in HBMs is to use Hill-type models, in which the force produced is a function of muscle length, shortening velocity, and activation level. Active musculature was first implemented in cervical spine models. These models were applied to study occupant kinematic responses and injury outcome in rear-end, lateral, and frontal impacts; it was found that active musculature is essential for studying the response of the cervical spine. One approach utilized to represent muscle activity in HBMs is to use experimentally recorded muscle activities or activity levels acquired through inverse optimization in open-loop. More recently, in order to represent car occupant muscle responses in pre-crash situations, closed-loop control has been implemented for multibody and finite element HBMs, allowing the models to maintain their posture and simulate reflexive responses. Studies with these models showed that in addition to feedback control, anticipatory postural responses needs to be included to represent driver actions such as voluntary braking.

Current HBMs have the capacity to model (utilizing closed-loop control) active muscle responses of car occupants in longitudinal pre-crash events. However, models have only been validated for limited sets of data since as high quality volunteer data, although it exists, is scarce. Omni-directional muscle responses have been implemented to some extent, but biofidelity of the simulated muscle activation schemes has not been assessed. Additional experimental volunteer muscle activity measurements (with normalized electromyogram recordings) in complex 3D-loading scenarios are needed for validation and to investigate how muscle recruitment depends on occupant awareness and varies between individuals. Further model development and validation of muscle activations schemes are necessary, for instance startle responses, and individual muscle control. This could improve assessment of restraint performance in complex accident scenarios, such as multiple impacts, far-side impacts and roll-over situations.

## INTRODUCTION

Human Body Models (HBMs) have been used in crash safety research for some time. Compared to Anthropomorphic Test Devices (ATDs) they can be made more humanlike, and usually incorporates an omnidirectional design; which makes them suitable for a wider range of crash situations. Future safety development challenges include more oblique and complex crash scenarios, in contrast to standardized crash test scenarios (usually pure lateral, frontal, or rear-end impacts) with ATDs in upright seated postures. Recently, small overlap frontal crashes have been included in the standardized test procedures (IIHS 2012) and one of the challenges identified is the ATDs' limitations in recreating occupant kinematics within these crashes, with a combination of longitudinal and lateral movement (Jakobson et al. 2013). Small overlap frontal crashes are an example of crash situations where the benefit of an omnidirectional HBM is clear. Other situations include oblique or angled impacts, multiple events, rollover, and run off road events. In a run off road event, the vehicle kinematics is complex and posing significant challenges for occupant simulation (Jakobsson et al. 2014). During run off road events not only the directions of impact can vary, but also the occurrence of multiple impacts connected with low acceleration amplitude events influencing  occupant posture during the events. Such scenarios could perhaps be viewed as the ultimate challenge from an occupant simulation perspective. Developing HBMs to address this challenge would be beneficial from a real world safety perspective.

Pre-crash maneuvers can influence occupant posture prior to an impact (Heiter et al. 2005; Hault-Dubrulle et al. 2011). In an analysis of accident data in Japan for 1993–2004, Ejima et al. (2009) reported that approximately 50% of drivers made an evasive maneuver in the form of braking or combined braking and steering before an impact. The development of Collision Avoidance Systems (CASs) is likely to increase the occurrence of pre-crash braking, by adding autonomous braking to the cases where drivers did not brake prior to the accident. CASs help drivers avoid or mitigate collisions through warnings and/or interventions, based on information about the traffic environment (Ljung-Aust et al. 2015). This information can be obtained by radar, laser, camera and other sensors. Intervention can include automatic braking and/or steering as well as means for improving occupant protection by triggering/adapting restraints. The development of CAS has been rapid over the last decade. While quite exotic 10 years ago, today most vehicle manufacturers offer some form of CAS in their vehicles, at least as an extra option. This has been made possible through developments in sensing and threat assessment, as well as improved actuators in production vehicles like steering control and differential wheel braking. The first generation systems were introduced in early 2000 with functionality restricted to provide brake support by tracking objects moving in the same direction as the host vehicle (Coelingh et al. 2006). In 2015, the state of the art systems includes tracking of cyclists, pedestrians and vehicles in front of the host vehicle as well as a first step into addressing intersection situations (when turning in front of an oncoming vehicle) (Ljung-Aust et al. 2015). The results from the few available real world follow-up studies indicate that CAS provide a substantial safety benefit (IIHS 2011; Isaksson-Hellman and Lindman 2012; IIHS 2013; Rizzi et al. 2014). The development of CAS creates a need for tools to evaluate the occupant response during the pre-crash phase, combined with possible subsequent impacts.

To improve restraint functionality, and prepare occupants for impact, reversible pre-tension systems have also been used together with CAS (Schöneburg et al. 2011); this allows the occupant to be more tightly coupled to the seat during autonomous braking, and has the potential to reduce forward displacement of occupants as a result of the pre-crash braking (Antona et al. 2010). Advanced restraint systems, that utilize reversible pre-tensioning of seat belts during the pre-crash phase, are emerging and some studies on the effect of pre-tensioning on occupant kinematics have been published, using either volunteers (Mages et al. 2003; Good et al. 2008a; Schöneburg et al. 2011; Östh et al. 2013; Ólafsdottir et al. 2013; Develet et al. 2013; Ito et al. 2013) or ATDs (Good et al. 2008b; Woitsch and Sinz 2014). However, extending these studies to also assess the injury reduction potential of restraints active in both the pre- and in-crash phase is difficult. Volunteers cannot be subjected to injurious loads while ATDs, developed to predict injury in high energy impacts, are too stiff to represent relaxed vehicle occupants under low loading conditions (Beeman et al. 2012).

The limitations associated with ATD and volunteer testing can be addressed with mathematical HBMs that can represent occupant responses in pre-crash as well as crash loading conditions (Schöneburg et al. 2011, Mages et al. 2011). To represent an attentive occupant and the influence of the occupant's muscle reaction on the kinematic response during the pre-crash phase, active muscle response and a human-like control strategy of the muscles must be included in future HBMs. In the past, several models with one dimensional or solid elements only representing the passive elastic and damping response of the neck musculature have been developed (Jost and Nurick 2000; Robin 2001; Ejima et al. 2005; Toyota 2008). However, the active force generated by muscles has a different order of magnitude than the passive muscle stiffness and damping at physiological muscle lengths. Early implementation of active muscle properties in HBMs were made in cervical spine models (Deng and Goldsmith 1987; de Jager 1996; Wittek 2000; Brolin et al. 2005). Since then, models with active musculature have been included in numerous HBMs, summarized in Table 1. The most common method for implementing active muscle properties in HBMs is to utilize 1D Hill-type muscle elements. In some models 1D Hill-type elements have been super-positioned with a passive bulk material to provide 3D muscles geometry with existing material models in the FE solvers (Behr et al.

2006; Hedenstierna et al. 2008; Iwamoto et al. 2009; 2011; 2012). Another study implemented the Hill-model with local fiber directions in a continuum FE material model (Khodaei et al. 2013).

The aim of this paper is to provide an update on state-of-the-art modeling techniques for simulation of muscle activity in HBMs, and to highlight future challenges and benefits with modeling of car occupants muscle responses to restraint activation.

## REVIEW OF ACTIVE MUSCLE CONTROL IN HBM

Several methods for regulating muscle activity in HBMs have been proposed. These methods can be divided into two main categories: open-loop control, where muscle activation functions are defined prior to simulation, and closed-loop control, where muscle activities are regulated based on sensory information about the current state of the model, for instance the position of a limb.

**Table 1. Summary of HBM studies that have included active musculature. MB = Multibody; FE = Finite Element; 1D = 1 dimensional; 3D = 3 dimensional. EMG = Electromyogram; PID = Proportional, Integral, and Derivative; T1 = First thoracic vertebra.**

| *Model* **Type / Solver** | **Reference** | **Body part** | **Actuators[1]** | **Control** | **Activation scheme** | **Application** |
|---|---|---|---|---|---|---|
| *TNO Active Human Model* | Cappon et al. 2007 | Spine | Torque actuators | Closed-loop | PID controllers | Reversible belt pre-tension, roll-over |
| | Budziszewski et al. 2008 | Upper extremity | 1D muscles | Closed-loop | PID controllers | Elbow flexion |
| MB / MADYMO | Meijer et al. 2008 | Spine, left arm and legs | Torque actuators, 1D muscles | Open and closed-loop | PID controllers, engineering judgment | Far-side impact |
| | Fraga et al. 2009 | Cervical spine | 1D muscles | Closed-loop | PID controllers | Motorcycle braking and cornering |
| | Nemirovsky and van Rooij 2010 | Cervical spine | 1D muscles | Closed-loop | PID controllers | Rear-end impacts |
| | van Rooij 2011 | Spine | Torque actuators, 1D muscles | Closed-loop | PID controllers | Autonomous braking |
| | Meijer et al. 2012 | Whole body | 1D muscles, torque actuators | Closed- and open-loop co-contraction | PID controllers, variable co-contraction | Autonomous braking, frontal, lateral, and rear-end impact |
| | Meijer et al. 2013b | Whole body, hip and elbow added | 1D muscles, torque actuators | Closed- and open-loop (co-contraction) | PID controllers, variable co-contraction and reaction time | Pendulum impacts, car braking, sled impacts |
| | Meijer et al. 2013a | Whole body, new neck and elbow | 1D muscles, torque actuators | Closed- and open-loop (co-contraction) | PID controllers, varied levels of co-contraction | Anterior-posterior T1 perturbations, elbow flexion impulses, and autonomous braking. |
| | de Bruijn 2014 | Cervical spine | 1D muscles | Closed-loop | Vestibular and muscle spindle feedback | Anterior-posterior T1 perturbations |

**Table 1. Continued.**

| *Model* Type / Solver | Reference | Body part | Actuators[1] | Control | Activation scheme | Application |
|---|---|---|---|---|---|---|
| *SAFER A-HBM* | Östh et al. 2012b | Upper extremity | 1D muscles | Closed-loop | PID controller | Elbow flexion-extension impulse load and posture maintenance |
| FE / LS-DYNA | Östh et al. 2012a | Cervical and lumbar spine | 1D muscles | Closed-loop | PID controllers | Autonomous braking for car passengers |
| | Östh et al. 2014a | Cervical and lumbar spine, upper extremities | 1D muscles | Closed-loop | PID controllers | Autonomous braking with reversible pre-tension |
| | Östh et al. 2014b | Cervical and lumbar spine, upper and lower extremities | 1D muscles | Closed-loop with anticipatory component, and open-loop | PID controllers and based on normalized EMG (lower extremities) | Driver maximal emergency braking postural response |
| *Active THUMS* | Sugiyama et al. 2007 | Lower extremity | 1D muscles | Open-loop | Inverse dynamics model | Brake pedal impacts |
| FE / LS-DYNA | Iwamoto et al. 2009 | Upper extremity | 3D muscles | Open-loop | engineering judgment | Lateral impact to elbow |
| | Iwamoto et al. 2011 | Whole body | 3D muscles | Open-loop | Normalized EMG | Frontal impact |
| | Iwamoto et al. 2012 | Whole body | 3D muscles | Open-loop | Reinforcement learning model | Frontal and rear-end impacts |
| | Iwamoto and Nakahira 2014 | Whole body | 3D muscles | Open-loop | Normalized EMG, engineering judgment | Pedestrian impacts |
| FE /PAM-CRASH | Wittek 2000 | Cervical spine | 1D muscles | Open-loop | Reflex activation | Rear-end impacts |
| FE / LS-DYNA | Brolin et al. 2005; 2008 | Cervical spine | 1D muscles | Open-loop | Reflex activation, optimization | Frontal and lateral impact, helicopter crash |
| FE / LS-DYNA | Hedenstierna 2008 | Cervical spine | 3D muscles | Open-loop | Reflex activation, Optimization | Frontal, lateral and rear-end impacts |
| MB / LS-DYNA | Chancey et al. 2003; Dibb et al. 2013 | Cervical spine | 1D muscles | Open-loop | optimization | Tensile neck loading, frontal impact, child HBM |
| FE/LS-DYNA | Chang et al.2008; Chang et al. 2009 | Lower extremity | 1D muscles | Open-loop | Normalized EMG | Knee impacts |

[1] In all HBMs which used muscle elements as actuators, the active behavior was modeled with a Hill-type material model. All models with 3D muscles employ the super-position of a passive continuum bulk material and Hill-type line muscle elements (Hedenstierna et al. 2008).

### Muscle Models with Open-loop Control

In models that use open-loop control, muscle activities are defined as a function of time prior to the simulation, based on know-how from previous simulations, experimental data, or optimization in static load cases. The outcome is observed afterwards, and the activation function may be iteratively adjusted to achieve a more biofidelic model response in upcoming simulations.

## Reflex Activation

Several cervical spine models (de Jager 1996; Wittek 2000; van der Horst 2002; Brolin et al. 2005; Stemper et al. 2006) have accounted for the influence of active behavior by the application of a maximum activity starting at a specified time in the simulation. With this approach in a multibody (MB) neck model, de Jager (1996) showed the importance of active muscles to capture the human head-neck response in frontal and lateral impacts; the same model was later refined and employed in rear-end impacts, and yet again the importance of active muscles was shown by van der Horst (2002). Brolin et al. (2005) found that including muscle activity for an FE neck model, with 1D Hill-type muscle elements, improved the kinematic correlation with volunteer data for frontal and lateral impacts. The muscle activation properties, i.e. the shape of the curve defining muscle activity, were varied and the best correlation with experimental data was found when neck flexors and extensors were assigned different activity levels. In addition, Brolin et al. (2005) also found altered injury patterns as an effect of neck muscle activity with respect to cervical ligament strain.

Maximum muscle activation at a specified time in the simulation is a straightforward way to introduce the effect of reflexive muscle responses and that of eccentric muscle lengthening. If simultaneous activation of all muscles is modeled, the implicit assumption is that a reflexive startle response is present. A startle response is a rapid response to stimulation of mechanoreceptors, acoustic stimuli, visual stimuli, or a combination thereof. It is characterized as a bilateral response which includes closing of the eyes, extension of the neck, elevation of the shoulders, and extension of the lumbar back (Yeomans et al. 2002). This might very well be present in many impact-like scenarios, and this method may suffice for short duration impacts with a clear loading direction, but not for more complex scenarios or for scenarios in which posture must be maintained for a period of time (Brolin et al. 2008).

## Optimization of Static Posture-Maintaining Activities

Chancey et al. (2003) developed an MB neck model with detailed muscles and studied the effect of muscle activity on tensile loading of the neck for two sets of muscle activities. The muscle activities evaluated were determined with an optimization scheme that gave initial stable postures for low-level and maximal muscle activation. More recently the same method was applied to find posture maintaining muscle activation schemes for six and ten-year-old pediatric cervical spine models (Dibb et al. 2013). The neck stabilizing muscle activity levels reported by Chancey et al. (2003) were used as a starting point to find load case specific stabilizing activities in a study with an FE neck model conducted by Brolin et al. (2008). The model was thereafter applied to evaluate the influence of tensed muscles on spinal injuries for helicopter pilots, and it was found that stabilizing neck muscle activation reduced the risk of ligamentous injuries. Bose and Crandall (2008) and Bose et al. (2010) used an MB HBM which was varied in size from the approximate $20^{th}$ to $80^{th}$ percentile male anthropometry, nine different initial postures, and 0–100% muscle co-contraction activity in 1D Hill-type elements. They performed optimization simulations to generate static stabilizing co-contraction activity levels and evaluated the influence of initial muscle co-contraction on a whole body injury metric in a simulated 57 km/h impact. They found that the initial posture was the most significant factor in determining the injury outcome, although the initial muscle co-contraction also had some influence. In particular, an increased risk for injury in the lower extremities with increasing muscle co-contraction was reported.

In open-loop control, optimization is a systematic way to balance a set of muscle activity levels. However, actual stabilizing activity levels are difficult to achieve. For instance, Chancey et al. (2003) defined posture maintenance as less than 5° rotation or 10 mm translation of the head over a period of 100 ms and Brolin et al. (2008) used a similar criteria in combination with a threshold of ±1 mm and 0.01 radians head motion for the subsequent 300 ms as a requirement for postural stability. In addition, any reflex responses due to the perturbation are not accounted for in an impact scenario.

## Optimization of Dynamic Activities

Iwamoto et al. (2012) presented a version of the THUMS HBM with a detailed 3D representation of muscles for all body parts. For the head and neck, a simplified model using 1D Hill-type elements was also developed. Using the simplified neck model and an optimization process called reinforcement learning, tabulated muscle control functions that account for both joint angles and velocities were derived. The optimization provided individual muscle activation functions that were applied in the detailed model in a rear-end impact test case. With the reinforcement learning muscle activities the THUMS with 3D muscles appeared to perform better than with deactivated muscles in the initial phase of the impacts compared with volunteer data, but then it overestimated the effect of the muscle activity on kinematics.

Multi-dimensional, pre-determined, activity tables can resemble actual human motor control, as the joint angles and velocities would correspond to stretch and rate of change of muscle lengths and correlate with input to the vestibular organs. If the model is accurate in reproducing the muscle activity for a wide range of combinations of angles and velocities, the resulting activation patterns could be compared with experimental data and the muscle response of the model would be human-like. However, a drawback of the study performed by Iwamoto et al. (2012) appears to be the large number of simulations (660) needed to generate the dynamic activity tables, and that these tables were derived for a simplified model, that potentially have different dynamic properties than the actual active HBM.

**Estimation based on Experimental Data**

Behr et al. (2006), Sugiyama et al. (2007), and Chang et al. (2008; 2009) all modeled emergency braking with active muscles in the lower extremities. The muscle activity levels were taken from normalized electromyogram (EMG) measurements in emergency braking experiments. They studied the injury risk in frontal impacts (Behr et al. 2006), brake pedal impacts (Sugiyama et al. 2007) and knee impacts (Chang et al. 2008), and concluded that the inclusion of active musculature changes the injury risk in these situations. Chang et al. (2008) predicted that the external force producing a fracture in the knee-thigh-hip area decreases when muscle tension is taken into account, although a limitation of the study was the lack of detailed muscle activity data for the lower extremities. Therefore, a second study was made (Chang et al. 2009), in which an inverse dynamics musculoskeletal model was used to derive detailed individual muscle activity levels from experimental data. The same approach applying inverse optimization was used by Choi et al. (2005), i.e. an optimization in which muscle activity levels are derived using a musculoskeletal model, measured forces, and limb positions, together with hypothesized optimization constraints. They simulated occupant bracing in sled impacts with active muscles in the upper and lower extremities. Iwamoto and Nakahira (2014) simulated pedestrian impacts with a whole body HBM with 3D musculature without muscle activity, with relaxed activity and 20% activity in all muscles, as well as only neck muscle activity based on volunteer data. They concluded that muscle activity affected pedestrian kinematics and could have an influence on the injury outcome predicted in such simulations.

If maximum voluntary contractions are performed for the specific experimental setup, it is possible to derive muscle activity levels for measured muscles under experimental conditions. These levels can then be used in simulations, such as in the studies summarized in the present section, either directly or through optimization with inverse dynamics models. This approach will generate muscle activities in the model which are in good temporal agreement and of the right magnitude, if the EMG is appropriately processed. However, for each scenario simulated, volunteer experiments must be performed and the models would not be able to predict occupant responses in other conditions. Therefore, this practical approach has limited applicability for safety restraint development, since interaction with new restraint systems may change muscle activity, as will more severe loading than what can be used in volunteer experiments.

**Muscle Models with Closed-loop Control**

In closed-loop applications the response of the controlled system is continuously monitored and the control signal is adjusted in accordance with the actual model response. In the human body the reflex arc, is the simplest closed-loop structure in the neuromuscular control system. In current active HBMs, closed-loop control has mainly been implemented with proportional, integral, and derivative (PID) controllers defined as:

$$e(t) = r(t) - y(t) \tag{1}$$

$$u(t) = k_p \cdot e(t) + k_i \cdot \int_0^t e(\tau)d\tau + k_d \cdot \frac{de(t)}{dt}. \tag{2}$$

The current state of the system, $y(t)$, is compared with the reference, $r(t)$, and the control signal, $u(t)$, is proportional to the difference between the two according to Equation (2). The characteristics of the PID controller are determined by the proportional gain, $k_p$, integral gain, $k_i$, and derivative gain, $k_d$. The PID feedback control can be applied to model human postural responses; the proportional and derivative feedback then models the effect of muscle spindle and vestibular reflexive stabilization, while the integrative controller removes any steady state error due to constant loads such as gravity.

**Closed-Loop Postural Control with Torque Actuators**

One of the first implementations of closed-loop control to model occupant postural responses was realized by Cappon et al. (2007) who utilized PID controllers to apply torque at each individual vertebral joint in a MB HBM.

The model was applied to study the phase preceding a roll-over accident and a static application of a reversible pre-tensioned restraint. The addition of the active spine improved the model kinematics in the roll-over scenario but was less successful in capturing the volunteer response to the reversible pre-tensioned restraint. The spine with active torque actuators was later utilized in several publications on the *TNO Active Human Model* (Meijer et al. 2008; van Rooij 2011; Meijer et al. 2012; Meijer et al. 2013a; 2013b). In addition, the torque actuator approach was used to simulate child kinematics with a 6-year old MB child model (Brolin et al. 2014). Torque actuators with closed-loop control are suitable for modeling scenarios where kinematic responses are of primary interest, but less so for crash events or study of injury outcome.

**Closed-loop Postural Control with 1D Muscle Elements**

Budziszewski et al. (2008) implemented closed-loop control of 1D Hill-type elbow flexor and extensor muscles in a MB arm model. A PID controller was implemented for the elbow joint angle and the muscles were grouped into flexors and extensors and assigned equal activity levels from the controller. The model was tested and compared with experimental data of voluntary elbow flexion and extension; it was concluded that the kinematic performance of the model matched that of the volunteers but that predicted muscle activity levels were over-estimated. Fraga et al. (2009) used feedback PID control of line muscle elements to stabilize the head of a motorcycle rider in lateral and longitudinal maneuvers for MB simulations. They concluded that their model appeared to capture resulting head kinematics of a volunteer with average awareness who applied the brakes on a motorcycle. Furthermore, they stated that the model was promising for the development of advanced restraint systems for motorcycle riders, and that it was a step towards active whole body HBMs.

The head-neck model used by Fraga et al. (2009) was further developed by Nemirovsky and van Rooij (2010) by the implementation of a postural controller for the head-neck complex, with the aim of regulating flexion-extension, lateral flexion, and rotation of the head. The motions were decoupled by a muscle recruitment strategy, which would ensure that only one degree of freedom was influenced by each controller; however, only the model response in flexion-extension was evaluated. Along with three PID controllers for the three head rotational degrees of freedom, a variable co-contraction ratio was implemented. The co-contraction ratio was important for the resulting closed-loop response, as muscular co-contraction makes a significant contribution to the damping of the closed-loop system. The model was later used by van Rooij (2011), who hypothesized that the attentiveness of drivers is reflected by the gains used in the control model. He simulated the influence of different levels of awareness on driver kinematics in autonomous braking interventions.

Meijer et al. (2012) combined and extended the work presented in the previous publications on the *TNO Active Human Model* (Cappon et al. 2007; Meijer et al. 2008; Fraga et al. 2009; Nemirovsky and van Rooij 2010; van Rooij 2011) to form an active whole body model. The feedback loop was complemented with a reaction time for events that produce a larger controller error than the preceding ones in the simulation. A low-pass filter function representing the neural transmission time from the CNS to the distal muscles was also added. The signal from each controller was converted to the muscles or torque actuators through multiplication by a constant defined in a recruitment table, to ensure that only the degree of freedom being regulated was affected (Nemirovsky and van Rooij 2010). Furthermore, muscle co-contraction was defined prior to the simulations, i.e. open-loop control, to generate muscle tension without any net moment around the joints, contributing to the intrinsic stiffness. Kinematic responses of the model were evaluated for autonomous braking, frontal, lateral, and rear-end impacts. It was concluded that both feedback control and muscle co-contraction is needed to predict volunteer responses in these types of events.

In Meijer et al. (2013b), feedback controlled elbow and hip muscles were introduced and a muscle recruitment approach similar to that described by Nemirovsky and van Rooij (2010) was used to decouple hip flexion-extension, medial-lateral rotation and abduction-adduction. Utilizing 50% co-contraction of the muscle actuators, the model was reasonably well able to capture forward displacements of the chest and neck in 1 g driver braking events, and in 3.8 g, and 15 g volunteer impact tests. Meijer et al. (2013a) introduced new and more biofidelic neck muscle geometry in the TNO Active Human Model, and evaluated the response of the head-neck complex to low level random perturbations of the T1 vertebra. Furthermore, force pulse perturbations were applied to the hand, inducing flexion and extension of the elbow and the model response was compared with that of one volunteer. Finally, the difference between a braced state and a relaxed state for the model was evaluated by simulating a braking event. It was concluded that the model response for the relaxed condition is different from the braced condition.

The head-neck complex from *TNO Active Human Model* was taken a step further by modeling muscle spindle and vestibular feedback in more detail (de Bruijn 2014). The controller included a model of the dynamics of each reflex loop and their respective neural delays. Muscle activity regulation was based on inputs on muscle length and lengthening velocity (spindle feedback) and head angular velocity and gravitational force (vestibular feedback). Each muscle was activated individually based on a muscle recruitment scheme obtained from isometric optimization simulations. Although the model was not used to simulate the head-neck response in vehicle impact scenarios it has potential to be applied in such studies in the future.

Östh et al. (2012b) implemented PID control for the elbow joint of the THUMS v3 (Toyota 2008) FE HBM, actuated by 1D Hill-type muscles. This study showed the feasibility of using closed loop muscle activation with FE HBM. In this particular case the PID controller was implemented in LS-DYNA with use of the solution control subroutine (Erhart 2010) in LS-DYNA, which allows for users to program their own material models and element formulations, Figure 1. The model was able to maintain its posture in a field of gravity. With 5 % extensor and 3 % flexor co-contraction and experimentally determined controller gains the model was able to reproduce the response of one volunteer to a 15 N force impulse. Furthermore, it was concluded that the detail of the original contact based joints in the THUMS v3 did not provide biofidelic passive joint properties and had to be replaced with revolute joints. An alternative approach for performing feedback control in LS-DYNA was presented by Prüggler et al. (2011), who used external software for the feedback control algorithm coupled to the FE solver for simulations with a simplified FE HBM.

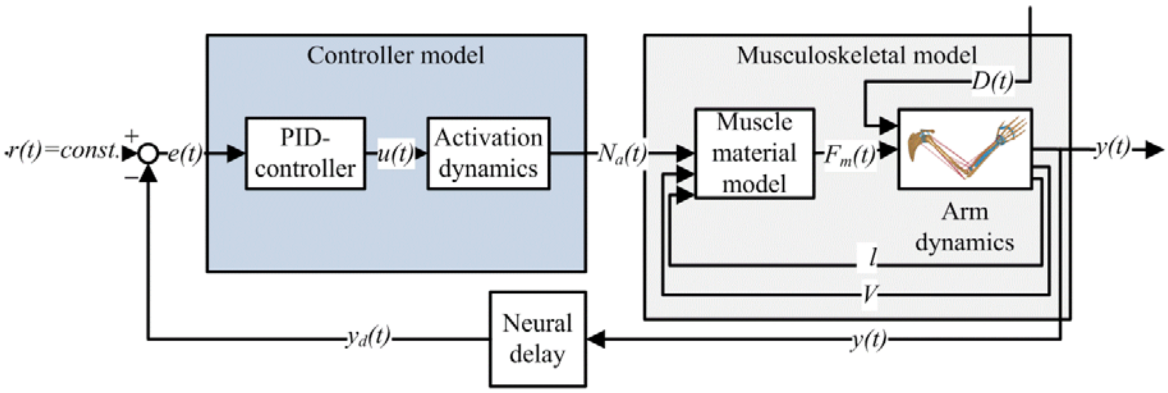


**Figure 1. Illustration of the neuromuscular feedback control loop used for the elbow joint by Östh et al. (2012b). Reprinted with permission from Taylor and Francis.**

The work by Östh et al. (2012b) was continued by adding 1D Hill-type muscles to the cervical and lumbar region and three PID controllers for the angles of the head, neck, and lumbar spine relative to the vertical axis. The model was used to study passenger forward head displacements in autonomous brake interventions of 6 $m/s^2$ compared with volunteer data (Östh et al. 2012a). It was reported that for passengers wearing a three-point seat belt, the support of the belt reduced the importance of the lumbar controller to match volunteer kinematics. Hence, to validate a lumbar controller experimental data from volunteers not wearing a shoulder belt is needed; instead wearing a lap belt or travelling unrestrained is needed. Moreover, in order to be able to simulate driver postural responses to autonomous braking, feedback control and linked 1D Hill-type muscles were added to the shoulders and combined with the previously developed elbow, trunk, and neck controllers (Östh et al. 2014b). This version of the model, denoted the *SAFER A-HBM* (Table 1), was then validated for four loading conditions: in driver and passenger positions for autonomous braking of 11 $m/s^2$ with two different three-point seat belt configurations. A seat belt with a reversible pre-tensioned retractor that provided 170 N of pre-tension to the shoulder belt prior to the braking was compared to a belt with a standard inertia reel retractor. The model compared well for kinematics, timing, and boundary forces between the occupant and vehicle, and reasonably for muscle activity levels. The validated model was employed in a parameter study of belt activation parameters, which showed that the largest reduction of peak head displacement was found for 570N pre-tension 0.15 s before deceleration onset, for both the driver and passenger positions.

Lastly, lower extremity muscles were added to the *SAFER A-HBM* (Östh et al. 2014a) and maximum driver braking was simulated. Based on experimental observations, an anticipatory postural response was hypothesized and implemented by changing the reference angles of the PID controllers for the head, neck, elbows and lumbar spine proportionally to the deceleration pulse. In general, most drivers will have extensive experience from braking events of varying magnitude, and, hence, will have inherent knowledge of the acceleration pulse that will follow after voluntary brake application. Without the anticipatory postural response the model predicted 100 mm longer forward displacements and 16° greater head rotations than were measured for braking volunteer drivers. With the anticipatory postural response, the model head displacement was within the volunteer corridors. Therefore, the hypothesized approach seems feasible and has the potential to be enhanced for other loading modes.

## DISCUSSION

HBMs will play an important role in future safety development of vehicles. Although certification procedures most likely will take a number of years to include virtual testing, it is not unlikely that it will become part of consumer information testing to a greater extent than today. Robustness and repeatability are important requirements for HBMs to be used for verification, with or without simulated muscle activity. This will be a challenge for HBMs with muscle control that changes their response with the level and direction of loading. From a real world safety development perspective there is a considerable need for active HBMs. Maneuvers prior to an impact are a reality (Ejima et al. 2009; Bohman et al. 2011) and are necessary to simulate in order develop protective systems for these situations. Evasive braking as well as steering maneuvers, represent the first priorities. However, ultimately HBMs should be designed to simulate a whole event of combined intensity, such as multiple impacts or run off road events. We would like to stress the importance of developing active HBMs that can be used to study both the pre-crash and crash in one or coupled simulations in a straight forward and simple methodology, preferably with the same HBM.

To date, one FE (Östh et al. 2014b) and one MB (Meijer et al. 2013b) whole body occupant HBM with muscle activity regulated by closed loop control, have been developed to simulate driver and passenger kinematics in pre-crash and emergency events. Both of these models are of average male anthropometry. They have been evaluated with respect to volunteer data in longitudinal loading situations. Future needs for safety development require HBMs with omnidirectional biofidelity and therefore there is a need to further enhance and validate these HBMs for oblique loading with lateral components. Also, to study long duration and complex crash scenarios FE HBMs are particularly needed for the crash simulations, as explicit FE is the industry standard. Furthermore, FE HBMs have the ability to predict injuries in more detail than MB HBMs. MB HBMs have the main strength for kinematics simulation, which is why they can have a benefit for applications like motorcycle events, pedestrian impacts, and for studying how design of sensors and signals and their systems will influence occupant kinematics in autonomous events without subsequent impacts. To conclude, FE HBMs are needed to study small overlap frontal crashes, oblique/angled impacts, multiple events, rollover, and run off road events, for example.

### Future Development of the *SAFER A-HBM*

The next step in the development of the *SAFER A-HBM* (Östh 2014a; 2014b) will be to implement muscle activity control for lateral and oblique load cases. The muscle activity varies between individual muscles and with the direction of loading (Ólafsdóttir et al. 2015). Therefore, refined recruitment strategies are needed, especially for the head-neck complex to capture head motion in various loading events. This means regulating the activity of each muscle individually. One of the major challenges in developing a controller for individual neck muscle regulation is the definition of load sharing and muscle recruitment patterns. So far this has been based on optimization in simulation of isometric conditions (Nemirovsky and van Rooij 2010, de Bruijn 2014) and not based on in vivo data from dynamic events, because such data has not been available previously. This year, Ólafsdóttir et al. (2015) published the first study providing such data by analyzing volunteer neck EMG data during seated perturbations and presenting spatial patterns of muscle recruitment for acceleration pulses in 45 degree intervals from 0 to 315 degrees.

More physiological muscle activity control in HBMs can be developed for many reasons; which are to be prioritized based on the simulation requirements for development of safety systems. Postural control by the central nervous system, for example of the head on the torso, can either be relative to space or relative to the torso depending on the loading conditions, low and high frequency perturbations respectively (de Bruijn 2014). We speculate that muscle control is modulated differently during driving (high frequency) and in autonomous interventions (low frequency). With these two postural control strategies muscle activity can be triggered by feedback from either muscle spindle, i.e. either stretch of the muscle, or from the vestibular system, i.e. balance in space. For vehicle occupants both systems are likely important and should ideally be implemented in active HBMs. For instance, during an emergency event or a long-duration crash, where an occupant tries to maintain a posture while interacting with advanced restraint systems that may cause local high muscle stretch (repositioning for example), both control loops are likely triggered. Future active HBMs might therefore benefit from including both reflex loops. Furthermore, as the contribution of the two control strategies varies with perturbation bandwidth (de Bruijn 2014) the controller parameters in the HBM need to be modulated depending on the simulated loading condition. In addition, safety systems activated before the event, whether by a crash or avoidance maneuver, can trigger startle-like responses (Östh et al. 2013; Ólafsdóttir et al. 2013). Omnidirectional HBMs with detailed neuromuscular control models that can simulate the startle reflex would be useful to further study how increased muscle activity due to startle affects the injury risk. Further, enhanced models for anticipatory postural response for driver initiated maneuvers, provide the potential to study how drivers interact with autonomous systems and how that changes the occupant response.

**Validation Data for Future Active HBM**

The evaluation of the biofidelity of active HBMs requires experimental data from volunteers in scenarios that replicate pre-crash conditions. In these tests a non-injurious, but representative, acceleration pulse is applied to seated volunteers whose muscle activity can be measured through EMG, the kinematics can be recorded with a camera, a motion capturing system, and/or accelerometers, and the boundary conditions can be recorded with load cells mounted to the steering wheel, seat, pedals, etc. The experimental data obtained in these experiments can also provide an estimate of the muscle activation schemes adopted by occupants in actual pre-crash events, aiding the development of methods for simulating muscle recruitment strategies, as outlined in the previous section. Hence, future volunteer experiments should be prepared and carried out carefully in order to generate useful data on muscle activity, the kinematic response, and boundary conditions.

**Scenarios**

Volunteer data that represent the effect of autonomous braking are readily available. Ejima et al. (2007; 2008) measured EMG, kinematics, and boundary conditions in volunteers in frontal loading conditions with a sled configuration and accelerations ranging from 0.2 – 1.0 g. In addition, a number of volunteer experiments with autonomous braking events using passenger cars and driving in regular traffic are available (Carlsson and Davidsson 2011; Östh et al. 2013; Ólafsdóttir et al. 2013). By driving or riding in a regular vehicle on regular roads the experimental conditions mimicked the targeted scenario to a higher degree than sled tests conditions; it was expected that these experiments provided data more representative of a scenario for which brake systems are autonomously activated.

Systems that avoid a collision by steering are being researched (e.g. Eidehall et al. 2007) and therefore there is an urgent need for validated omnidirectional active HBMs that can mimic the human response in these scenarios. Volunteer experiments with lateral and oblique loading have so far received less attention than longitudinal loading. Volunteer kinematic responses and EMG data during pure lateral and lane change type loading have been provided by van Rooij et al. (2013) and Ejima et al. (2012). In both these studies the volunteers were seated in a rigid seat and the experiments were carried out inside a laboratory. Huber et al. (2012) presented upper torso, arm and head kinematics, activation timings and absolute EMG levels during 1 g lane change maneuvers for front row vehicle passengers. Future volunteer experiments are to be carried out in conditions that more closely resemble an actual driving event.

Other scenarios for which volunteer test data will be required are multiple events, rollover, and run off road events. These scenarios include complex vehicle kinematics and are difficult to simulate on a test track using a regular passenger car. Instead they could be replicated in advanced vehicle simulators or robot test rigs

**EMG**

EMG signals from various muscle groups were recorded and normalized to the maximum EMG value recorded during the event for several muscle groups in the experiments conducted by Ejima et al. (2007; 2008) and van Rooij et al. (2013). These experiments provided a valuable insight into which muscle group is activated during pre-crash braking and steering events, the respective activation timings and overall kinematics. The magnitude of the EMG signals, however, cannot be used to directly compare the level of muscle activity between different muscles or volunteers nor with the simulated muscle activity in active HBMs; hence, their applicability for model development are limited. EMG signals normalized to maximum voluntary contractions (MVCs), as were provided by Östh et al. (2013) and Ólafsdóttir et al. (2013), are more appropriate for active HBM development and validation where the signals are represented as a percentage of a maximum activation, which can more easily be compared to or defined in an active HBM. For the simulation of test scenarios that result in lateral occupant motions there is a need for muscle data for the muscles surrounding the pelvis, legs and for the muscles stabilizing the lower spine. Few studies have yet presented such data; one study has presented leg muscle activations when the volunteer was performing emergency braking (Behr et al. 2010).

Several studies have measured the activation levels of deep muscles in the spine using intra-muscular EMG when the volunteer were subjected to perturbations (Siegmund et al. 2007; Ólafsdóttir et al. 2015). Such data provide neuromuscular parameters, muscle synergies and an understanding of neck stabilization in dynamic events and are essential for the development of muscle recruitment models for the head-neck complex. However, additional data is required for loading conditions matching those that would occur in vehicles fitted with systems that employ active steering.

In conclusion, access to MVC normalized EMG data from various loading conditions is imperative for development of muscle recruitment strategies and as validation data for active HBMs. Future studies should preferably include measurements of deep muscles activity and include muscles that stabilize the trunk.

**Kinematic response**

Proper recording of kinematics in volunteer experiments are a necessity for the development and evaluation of active HBMs. Recording volunteer kinematics with traditional video recordings in regular passenger vehicles when longitudinal acceleration is deployed have been carried out successfully (Carlsson and Davidsson 2011; Östh et al. 2013; Ólafsdóttir et al. 2013). The visibility of targets has sometimes been limited; targets mounted to the lower neck region can be obstructed by the seat and chest targets can be obstructed by clothing and other occupants. Motion capturing systems have also been successfully adopted (Huber et al. 2012) although these systems encounter similar visibility limitations as traditional video recordings. For scenarios including lateral vehicle acceleration, the volunteer response would be more complex and would require multiple video cameras or other systems to capture occupant kinematics. Although traditional video system and motion capturing systems can be adopted and installed we encourage the development of new methods to measure occupant kinematics during dynamic events.

**Boundary conditions**

It is important to measure boundary conditions, such as restraint, pedal, and steering wheel forces and seat contacts, as the occupants will be interacting with their environment. In past volunteer studies the boundary conditions have been recorded (Ejima et al. 2007; 2008; 2012 Östh et al. 2013; Ólafsdóttir et al. 2013) while in others the main purpose of volunteer studies has not been to provide validation data (Carlsson and Davidsson 2011) and boundary conditions have thus been omitted. Such boundary conditions are important in validations of active HBMs and should be recorded in future studies.

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