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A method to model anticipatory postural control in driver braking events

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ABSTRACT

Human body models (HBMs) for vehicle occupant simulations have recently been extended with active muscles and postural control strategies. Feedback control has been used to model occupant responses to autonomous braking interventions. However, driver postural responses during driver initiated braking differ greatly from autonomous braking. In the present study, an anticipatory postural response was hypothesized, modelled in a whole-body HBM with feedback controlled muscles, and validated using existing volunteer data. The anticipatory response was modelled as a time dependent change in the reference value for the feedback controllers, which generates correcting moments to counteract the braking deceleration. The results showed that, in 11 m/s^2 driver braking simulations, including the anticipatory postural response reduced the peak forward displacement of the head by 100 mm, of the shoulder by 30 mm, while the peak head flexion rotation was reduced by 18°. The HBM kinematic response was within a one standard deviation corridor of corresponding test data from volunteers performing maximum braking. It was concluded that the hypothesized anticipatory responses can be modelled by changing the reference positions of the individual joint feedback controllers that regulate muscle activation levels. The addition of anticipatory postural control muscle activations appears to explain the difference in occupant kinematics between driver and autonomous braking. This method of modelling postural reactions can be applied to the simulation of other driver voluntary actions, such as emergency avoidance by steering.

1. Introduction

Numerical human body models (HBMs) are used for research and development of vehicle occupant protection systems [1,2]. Recently, an interest in simulation not only of the crash phase, but also of the pre-crash phase, of road accidents has led to implementation of active muscles and control strategies in HBMs. Feedback control is suitable to model occupant postural responses in autonomous braking interventions [3,4]. In volunteer experiments, it was found that during driver initiated braking, drivers more effectively maintained their initial posture than during autonomous braking interventions [5,6]. For example, forward head displacements for males ($n = 11$) were 35 (SD 37) mm on average in driver braking; this is significantly less ($p < 0.05$) than the 98 (SD 65) mm found for autonomous braking of the same magnitude, 11 m/s^2 [5]. Driver initiated braking differs from autonomous braking in that the driver performs a voluntary action. The driver rapidly shifts his foot from the accelerator to the brake pedal, extends the hip and thigh, and plantarflexes the ankle with relatively high muscle efforts [7].

For other types of voluntary actions, anticipatory postural responses are found before activation of the prime movers [8–12]. For instance, prior to step initiation, anticipatory postural adjustments initiate a forward and lateral movement of the body mass [9]; during falling anticipatory muscle activity prepares the body for impact [10]; lifting of the arm while standing generates leg muscle activation 50–100 ms prior to activation of the prime

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movers of the arm [11]. These anticipatory responses are generated by the central nervous system (CNS) in a feed-forward manner to generate approximate correcting muscle activations in various body parts prior to postural perturbations [12]. In the present study, a method to model anticipatory postural responses in HBMs for occupant simulation is investigated and applied to study maximum driver braking.

2. Methods

A whole body Finite Element (FE) HBM, the THUMS® AM50 v3.0 [2], was used in this study (Fig. 1). The model contains rigid bodies (e.g., the vertebrae) and deformable parts (e.g., the intervertebral discs, ribs, skin, and internal organs), totalling 68 100 solid elements, 75 700 shell elements, and 3400 one-dimensional elements. Some changes were made to the THUMS® for this study [4]. The hip joints were modelled with ball joints positioned in the femoral head [13]. As the current study included frontal loading only, the irrelevant hip degrees of freedom, abduction–adduction and medial–lateral rotation, were constrained by high passive stiffnesses (20 000 Nm/rad). The knee and ankle joints were modelled with revolute joints positioned according to [14,15], respectively. The FE solver LS-DYNA® version 971, release 6.1.0 (LSTC Inc., Livermore, CA, USA) was used. Pre- and post-processing were done with LS-PREPOST® v4.0 (LSTC Inc., Livermore, CA, USA) and MatLab® R2012b (The Mathworks Inc., Natick, MA, USA).

2.1. Musculoskeletal feedback control model

The THUMS® model has previously been complemented with 348 line muscle elements representing the muscles of the neck, lumbar and abdominal areas [4], and the upper extremities [16]. A Hill-type muscle material is used, in which the maximum isometric stress is 1 MPa for the upper extremity muscles [17] and 0.5 MPa [18] for the other muscles in the model. Two different values were chosen to give the model maximum isometric strengths of similar magnitude as that of volunteers. For example, in elbow flexion and extension, the model strength is 86 Nm and 48 Nm compared with volunteers 78 (SD 11) Nm and 50 (SD 11) Nm [19]. For cervical flexion and extension the model strength is 32 Nm and 48 Nm compared with volunteers 30 (SD 5) Nm and 40 (SD 8) Nm [5], measured relative to T1.

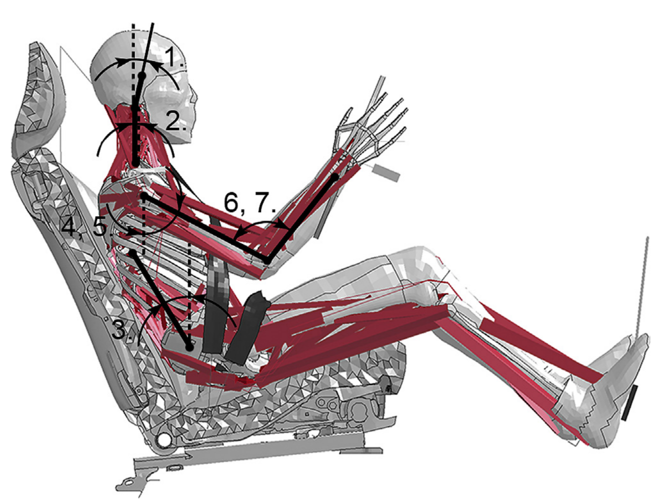

Fig. 1. The controller angles for the (1) head, (2) neck, (3) lumbar spine, (4) left and (5) right shoulder all use the angle of the body part with respect to the vertical axis. The (6) left and (7) right elbow controllers utilize the relative angle between the humerus and ulna. Soft tissues of the trunk, neck, and upper extremities and half the seat are not shown to disclose the musculoskeletal structure of the model.

To model postural control and response to external loads, seven proportional, integral, and derivative (PID) controllers were implemented. The control signals are defined as the angle in the sagittal plane between the vertical axis and a vector defined by two nodes in the model, and for the elbow controllers as the angle between vectors spanning the humeri, from the centre of the glenohumeral joint to the elbow, and ulnae, from the elbow joint to the distal end of the ulna (Fig. 1). The lumbar vector extends from the sacrum to the vertebral body of T10, the cervical vector from the vertebral body of T1 to the mid occipital condyles, and the head vector from the mid occipital condyles to the head centre of gravity. Head centre of gravity is determined according to the mass distribution of the models skull, flesh, and brain. The PID controllers are hypothesized to represent vestibular and proprioceptive feedback; they generate a control signal, $u(t)$, computed according to:

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

$$u(t) = k_{\mathrm{p}} * e(t) + k_{\mathrm{i}} * \int_0^t e(\tau)d\tau + k_{\mathrm{d}} * \frac{de(t)}{dt} \tag{2}$$

The joint angle, $y(t - T_{de})$, is compared with the reference, $r(t)$, and the control signal is proportional to the error, $e(t)$, between the two, Eq. (2), with proportional feedback gain, k_{p}, integrative feedback gain, k_{i}, and differential (velocity) feedback gain, k_{d}. The proportional and velocity gains can be considered as generic representations of reflexes responsible for the maintenance of posture, i.e. muscle spindle feedback [20] and vestibular reflexive stabilization [21], while the integrative controller corrects any residual error and maintains the desired posture in the presence of gravity. The transport delay, T_{de}, accounts for the time needed for the neural signal to be conveyed to and from the CNS. T_{de} was 34 ms for the elbow and 30 ms for the shoulder [22] controller. For the head and neck T_{de} was 20 ms, i.e. a shorter delay was estimated due to the proximity to the spinal cord, matching the 18 ms delay reported for the cervicocollic reflex in cats [21]. For the lumbar controller, T_{de} was 25 ms, which is relatively close to the 30 ms that has been reported for the lumbar spine muscles [23]. The control signal, $u(t)$, is converted to a muscle activation request by scaling with the maximum isometric strength of each controlled muscle group. The scaled activation request is passed through a muscle excitation–contraction dynamics model consisting of two coupled first order filters [24], giving a muscle activation level, $N_{\mathrm{a}}(t)$. A generic muscle recruitment strategy divides the muscles of each controlled joint into either flexors or extensors, with the same activation level. Co-contraction of muscles around the controlled joints is implemented as a lower bound on the muscle activation, i.e. all muscles always have a prescribed minimum activation level as selected below.

2.2. Lower extremity muscle implementation

For the lower extremities, Hill-type line muscles were added, see Table 1. To account for the curvature of the gluteus maximus around the pelvis and of the quadriceps and patellar tendons over the knee, the Hill-elements were coupled in series with stiff (10 000 N/engineering strain) "seat belt" elements. These elements were fed through slip rings attached to the pelvis for the gluteus maximus and to the distal head of the femur and proximal head of the tibia for the quadriceps and patellar tendon.

2.3. Maximum driver braking simulations

Volunteer kinematics, interaction forces, and muscle contraction levels in 11 m/s^2 driver braking events from 70 km/h to a