DEVELOPMENT OF A MOTORCYCLE RIDER MODEL WITH FOCUS ON HEAD AND NECK BIOFIDELITY, RECURRING TO LINE ELEMENT MUSCLE MODELS AND FEEDBACK CONTROL

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ABSTRACT

Despite continuing improvements in vehicle safety, motorcyclist casualties are estimated between 13% and 17% of road fatalities. Looking at the last two ESV conferences for a tentative measure of the research effort that is geared towards motorcycle safety, oral/written papers referring to two-wheelers averaged 6%/3% of each group. This tendency is also identifiable in the clearly lagging development of experimental techniques and computational models for the study of crash scenarios involving PTWs. This *status quo* prompts further developments of PTW-specific design tools to stem from existing occupant (and pedestrian) tools, rather than already available motorcycle-specific solutions.

This paper aims at filling some of that gap by proposing developments in computational models for motorcyclists alongside real-world trials. The paper concludes that a MADYMO human body model, equipped with PID-controlled neck muscles, reasonably maintains its biofidelic erect posture in sample scenarios, under the assumption that riders attempt to maintain their head upright. Preliminary results yield activation levels of up to 50 and 55% during severe (± 1,7G and 0,8G) longitudinal and lateral loading scenarios, respectively.

Preliminary volunteer trials (N=8) were conducted to provide initial validation in the event of braking. Although not yet complete, the analysis suggests that the resulting head kinematics for an average aware volunteer is compatible with the simulated response.

This development focuses R&D efforts on preventing injuries to the head-neck-complex, the body's most vulnerable region, by providing biofidelic postures and reactions to developers of personal protective equipment and advanced occupant/rider restraint systems. It also allows the evaluation of a motorcycle active safety system's impact on human response, which directly influences the consequences of the potential subsequent pre-crash or crash event. Finally, it represents a first step towards fully active human models, which will provide life-like pre-crash behaviour to e.g. OEMs, equipment and barrier manufacturers, and policy makers.

INTRODUCTION AND BACKGROUND

When one specifically takes into account Europe's 8.5+ million motorcycles (mopeds excluded) and the estimated 5000 annual motorcyclist casualties [COST 327, 2001] (corresponding to somewhere between 1/6 and 1/8 of total road fatalities yearly), it is still not clear that the corresponding research effort and budget are allocated in proportion with PTW (Powered Two-Wheeled) vehicles' relevance within the road safety context. This can be unmistakably identified in the less-than-ideal development of human body models for the study of crash scenarios involving PTWs, regardless of the former's type: animal and PMHS trials have been almost unheard of, the appropriate ATD designed in the '90s (the "Motorcyclist ATD") does not reflect the latest biomechanical thinking mirrored in some car-specific modern alternatives (and also lacks multiple body sizes) [MATD ISO], and specifically-designed and validated computational models are generally limited as researchers have focused strongly on the occupants' and pedestrians' perspectives. The fact that the MATD achieved limited acceptance in the industry also lessened its effectiveness as a tool for the sharing of knowledge and solutions between PTW and car safety work. This context prompts the further development of PTW-specific design tools to stem from existing occupant (and/or pedestrian) models, rather than already available (and hence less developed) motorcycle-specific solutions or the models developed specifically for automotive applications. That is indeed the chief objective here. PTW accident scenarios are far more dependent on their pre-crash dynamics (herein called "trajectories") than their four-wheeled counterparts, and thus the relevant human body modelling approaches are limited to those which are compatible with relatively long simulated timeframes (at least a couple of seconds). Also, because the posture and movement of PTW occupants DOES significantly influence the global system's behaviour (PTW+riders), the modelling approach must allow relatively broad magnitudes of movement and interaction of the simulated human body in relation to its environment (at least, the motorcycle). Jointly considered, these two points imply that the implemented solution needs to be truly "dynamical", i.e. designed to simulate a complex sequence of events corresponding to the whole scenario under study, rather than focusing on any specific event with extensive detail.

Towards biofidelic active human models

As researchers develop methodologies and tools to understand driving-related injury, both known statistics on the subject [IRCOBI Future Research 2006] and the knowledge of human anatomy and physiology points to the head-neck complex (HNC) as a priority in terms of which regions to model and what dynamics are of crucial understanding.
From the perspective of computational simulations, it has been clearly demonstrated over the past decade that the contribution of actively controlled human muscular action in automotive safety situations can not be neglected. Van der Horst resorted to computational modelling in MADYMO to define realistic lines of action and insertion points for the cervical region's multi-segment muscles [vd HORST 1997 and 2002]. It was thus possible to analyse the effects of muscular activation as the resulting force was exerted along a complex path surrounding the vertebrae, and hence study the former's influence on the HNC's global kinematic behaviour. Some authors even argue that muscle activation is unequivocally important and fundamentally changes the behaviour of an otherwise unrealistically passive model, based on an approach which also provides the moments of inertia and the forces produced by the cervical musculature [Brelin-Fornari 1998]. More recently, Lopik and Acar developed a model of the human HNC in visualNASTRAN 4D and used MatLab's Simulink to control the corresponding muscles. Rear impact scenarios then suggested that the influence of muscle activation in an unaware occupant's kinematics was small, but nevertheless the authors concluded that the forces recorded on the neck's soft tissues (and presumably the injury potential) were considerably influenced by the activation of the muscles [van Lopik 2004]. Even more recent work has demonstrated that neck muscle contraction stabilizes the head and neck during whiplash and reduces soft tissue deformation in aware impact situations [Stemper 2006].
From the experimental perspective, on the other hand, studies regarding HNC muscular responses are usually related to out-of-position automobile occupants. In one approach, the authors recurred to ElectroMyoGraphy (EMG) and subjected human volunteers to mild whiplash-like rear and lateral impacts while their torso and head were flexed out of the normal stance inside an automobile [Kumar 2004 and 2005]. The measured signals pointed towards a set of muscles which seem to be of primary relevance in the body's attempt to stabilize its posture and avoid injuries.

SCOPE AND OBJECTIVES

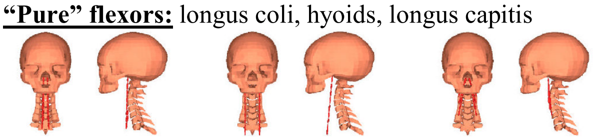
Whether one focuses on automobile occupants or motorcycle riders, it is consensual to state that severe decelerations ("high-G") will cause more destructive structural injuries than low- and medium-G scenarios. Given the apparent reflexive time delays and intrinsic limitations of active muscle force, muscular action is obviously much more relevant in low- and medium-G scenarios than it is in high-G, and thus only for impact will human response be completely passive. Therefore, the widely available passive human body models (of either the actual human body or ATDs) CAN accurately emulate human response under impact conditions, if necessary with stiffened joints. However, potentially perilous low-G scenarios may still arise (for instance) when an occupant swerves abruptly between two steering directions and his head is projected from one side to the other in a short amount of time, or in the event of a rollover.
The development of advanced crash-avoidance systems and impact restraint mechanisms depends on sensing and acting upon the pre-crash reactions of the human body. Throughout a potentially hazardous event, both the vehicle (with its safety systems) and the external environment will interact in real time with the human-in-the-loop, so their influence on the latter, and the human's reactions, may in turn impact the way the situation unfolds.
For motorcyclists specifically, it was already discussed how actively-controlled muscular action considerably influences the outcome of virtually all hazardous road situations, since the dynamics of motorcycle-specific scenarios are even more

dependent on the actions and posture of the rider (both at the level of pre-crash and actual crash) than their automobile counterparts.
Also, when it comes to low-G or pre-crash simulations, the time lapses involved are significantly larger and any chosen models for an alert driver/rider should exhibit some degree of muscular activation and the ability to react and adapt to the ongoing events in real time. Consequentially, any passive model displays a clear disadvantage in terms of its dynamic biofidelity, and truly dynamical models of the actual human body are unmistakably required.
Because of its particular relevance in this issue, recent research has focused on the HNC, establishing where the quest for active muscles and posture should receive the most effort. Naturally, as with all simulation-based approaches, any proposed advances will require motorcycle-specific validation trials, probably using EMG to identify the relevant muscles.
In conclusion, active human body models permit the study of how posture and movement influence the simulated response, and eventually the injury potential, in scenarios typically addressed within the vehicle safety field. For that end, they require the implementation of muscle models (for instance of the Hill type) and a dynamic mechanism to control their voluntary and/or involuntary activation. All but the last of these requirements are nowadays fulfilled with MADYMO's facet human body model with a detailed head and neck [vd HORST 2002].

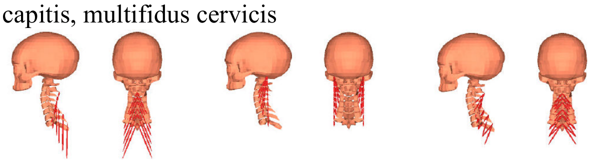
AN ACTIVE HUMAN MODEL FOR PTWS

If it is to be effective as a design tool for research and industry alike, the desired end product must build on a validated and widely accepted model for the system being studied, and thus the chosen basis was Van der Horst's detailed multibody head and neck as integrated with the MADYMO facet human body model [vd HORST 2002, MADYMO HBM Manual]. It consists of a rigid head and vertebrae, (non)linear viscoelastic discs, frictionless facet joints, nonlinear viscoelastic ligaments and segmented contractile muscles which follow the curvature of the neck, thus allowing realistic lines of action. Literature data provided the mechanical properties of the tissues involved, and the model is capable of outputting their local loads and deformation. A more extensive description does not fit within the scope or focus of this paper, except for a specific note regarding the muscle modelling itself. Van der Horst resorted to MADYMO's implementation of the Hill-type muscle model: it comprises a "contractile element" (CE), which describes the actively generated contractive force, and a passive "parallel element" (PE), which describes the elastic force arising from the elongation of the muscle tissue. Total force is therefore the sum of these two contributions, which are described extensively in [vd Horst 2002]. Some representations of the Hill-type muscle model also include another passive elastic element in *series* with the CE, meant to introduce a spring-like "delay" when the CE is producing force, but in MADYMO that contribution is built into the latter.
Specifically on the contractile element's behaviour, its contribution to the total force depends on an "activation state" which describes the normalized activation level of the muscle and adopts values between 0 (rest state) and 1 (maximum activation). It is precisely this parameter that will become the control variable in order to attain the desired posture or movement of the HNC.
In total, the van der Horst HNC model comprises 16 muscle pairs divided in 68 muscle segment pairs. Their activation signals were initially arranged in three groups: flexors, extensors, and the single-member sternocleidomastoids. In order to isolate the muscles with relevant contribution to lateral (roll) motion, each individual muscle pair was activated on one side to analyse the head's response. This procedure led to a new set of muscle groups, each with a relatively clear biomechanical function:

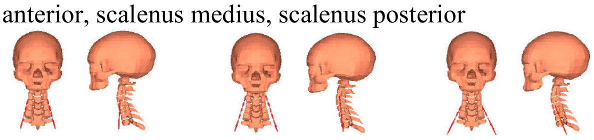
"Pure" flexors: longus coli, hyoids, longus capitis



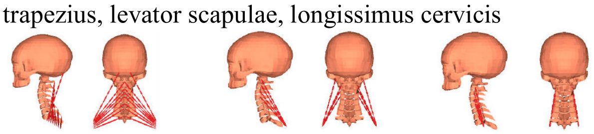
"Pure" extensors: semispinalis cervicis, longissimus capitis, multifidus cervicis



Rollers with secondary flexion function: scalenus anterior, scalenus medius, scalenus posterior



Rollers with secondary extension function: trapezius, levator scapulae, longissimus cervicis



Yaw with secondary extension function: splenius capitis, splenius cervicis (above), sternocleidomastoid, semispinalis capitis (below)

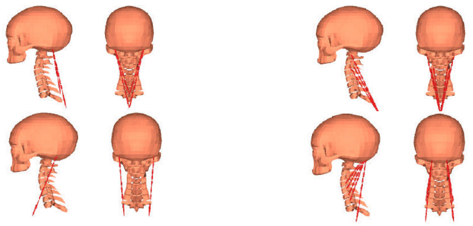


Figure 1. HNC muscle groups divided in flexors, extensors, rollers, and yaw.

A feedback approach to muscle activation control

As discussed before, at least one recent human body model possesses most of the features required to begin the work at hand. What is lacking is a method of controlling muscle activation so that a certain target posture or movement is dynamically displayed. A computational control system (or control loop) is commonly defined as encompassing the computational model (in this case, the human body's HNC), various sorts of sensors, one or more controllers, and various actuators (applying for instance forces, torques or displacements) which in this case are implemented as line element muscles. Within linear control systems, the ones that include some sensing of the results they are trying to achieve are making use of feedback and so can, to some extent, adapt to dynamically varying circumstances. This feedback control method was chosen amidst the classes available to design an active system. The *rationale* for this choice is fairly straightforward: no other method allows such tuneable and swift design. Furthermore, feedback control's ease of implementation is unrivalled directly through MADYMO or resorting to a coupling with Matlab/Simulink.
For the activation of the muscles, one needs to characterize the dynamic behaviour of the physical system so that the control loop's features can be properly designed. The first step would be the definition of what exactly are the relevant control parameters. The following picture illustrates the very first thoughts on the subject:

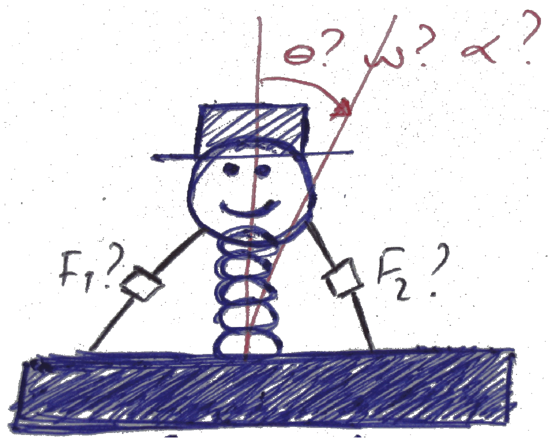


Figure 2. First sketch of what would eventually become the model's HNC control variables.

The following steps led to a deeper analysis but the burgeoning idea was retained, and the next section describes how the angular displacements between the head and the reference space became control inputs. Actually, in any feedback controlled system, both the relevant input and reference signals must be specified as numerical functions of time (and eventually other parameters). In this case, the direct modeling method was used: the fundamental features of the system (the human body's HNC) were analyzed as to their physical principals and desired behaviour, and appropriate control variables were identified. Considering that the model's range of movement was conceived to emulate human biomechanics, the articulation and the kinematics of the HNC joints and bodies should be reasonable proxies for their human counterparts. These are described below:

Neck Pitch – anteroposterior flexion and extension of the HNC, occurring in the sagital plane. This movement is not a unitary one, as it is permitted by the composition of small movements between adjacent cervical vertebrae with the help of the intervertebral discs. The downwards pitch is considered the positive direction for this displacement.

Neck Roll – the lateral abduction (away from the body's longitudinal axis) and adduction (towards the axis) equivalent of the Neck Pitch, occurring in the frontal plane. The rightwards roll is considered the positive direction for this displacement.

Head Yaw – the head's rotation about the neck's vertical axis. In actual fact, the head and the atlas rotate together on top of the axis (the second cervical vertebra) using the axis's dens (a tooth-shaped process) as a pivot. The leftwards yaw is considered the positive direction for this displacement.

It was decided that Head Yaw should be residual at all times, not because of humans being unable to perform the equivalent movement (which is clearly untrue) but because of modelling difficulties associated with the individual control of the model's facial orientation while the muscles are balancing the other two degrees of freedom (pitch and roll). Near forward-looking orientation was achieved for all attempted simulations as a direct consequence of the active control of the head's roll and pitch, because muscle tension intrinsically stabilizes the neck.

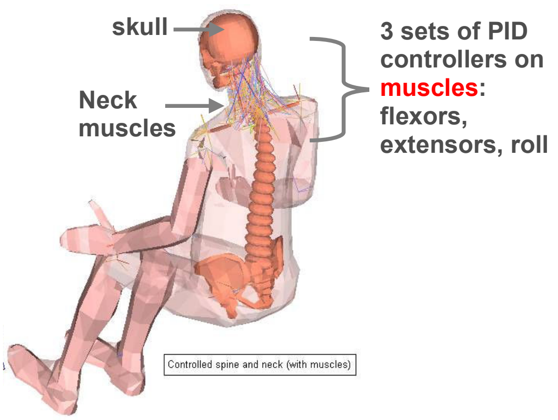


Figure 3. MADYMO facet human body model with detailed neck and optimised muscle groups.

The relevant movements can be sensed by the human vestibular apparatus and as such their equivalents should be sensed in a similar manner to render controller inputs. The corresponding analogues are presented shortly. It must however be said that the chosen methodology consists of an approximation, for the sensed parameters (angular deviations) bare resemblances to the biological system but are still not actual counterparts to it – in fact, the vestibular apparatus senses angular accelerations and is also able to "predict" the dynamic loading to some extent. This is one of the reasons why the authors believe the control method will require a different approach in the near future, as the validation progresses to more demanding scenarios.
As for the control references, maintaining overall head verticality (*i.e.*, keeping the head's longitudinal axis orthogonal to the ground) was the chosen criterion. This idea's stems were twofold: knowledge of the biological semicircular canals' arrangement, which suggest that the postural control mechanisms work at their best in said position, and the notion that the cortical processing of visual perceptions is strongly dependent upon the horizontal reference provided by the horizon. It is nevertheless debatable whether this criterion is valid regardless of one's role on a motorcycle or inside an automobile. That is, not only is the verticality approach an approximation in itself, but it could also be reasonably expectable that a distracted and/or relaxed rider/occupant (unwary to the perception acuteness necessary for the safety of the driving process) will allow significantly broader HNC displacements even if at the expense of visual and postural references.
The output signals are of arguably trivial choice: since an active system requires some sort of actuation applied to appropriate joints or bodies in order for the system to follow a reference, the controller outputs need to be fed to such actuators. As aforementioned, in this case line element muscles embraced that role: they present realistic force points of application and direction vectors, and as such their combined effects influence the whole system. This design decision thus correlates with accepted biomechanical evidence: the body's muscles apply contractive forces to the bones in order to generate torques and thus rotations of the same bones about the body's joints.

Implementing a controlled head-neck-complex

The chosen control approach implies that the true spatial status of the head-neck complex – how "vertical" it is at any given moment, regardless of its position in relation to the thorax – must be known at all times. That said, two sensors were implemented to provide the absolute (spatial) pitch and roll of the head. This formulation allows for the measurement of the head's deviation from verticality in any chosen direction and throughout the simulation, which would be untrue if joint angular displacement sensors had been used with the same purpose (since these would yield relative pitch and roll). In fact, experiments with low level random perturbations illustrated the importance of vestibular feedback in neck stabilization [Guiton 1986, Kesher 2000/2003] and the combined visual and vestibular feedback can be assumed to register head orientation in space as well as rotational velocity and acceleration.
The signals from the sensors are sent to PID controllers, which at every moment attempt to determine the "error correcting" signal: the one which nullifies the difference between the sensor and a reference signal supplied by the user (which represents the abovementioned "vertical" head position). The outputs from the PIDs are then sent to the muscles, specifying the activation state that is necessary to maintain the desired position (in this case, the head's verticality) against external stimulation – the trajectory-induced inertial forces. Naturally, each of the previously defined "pitch"

muscle groups (flexors and extensors) requires its own PID and associated parameters because the corresponding muscles and model geometry are not symmetrical at all. The completely symmetrical "roll" muscle groups, though, are activated on the left or right side depending on whether the difference between sensed and reference angles is respectively positive or negative, allowing the use of a single (rectified) PID. This leads to 3 separate PID controllers and control parameters for the HNC.

Tuning of the control parameters – Due to the absence of pre-existing biomechanical data, the control parameters were determined with the Ziegler-Nichols method, educated guesses, and trial-and-error, taking into account these notions:

Table 1.
Expected effect of increasing control parameter

Parameter	Rise Time	Overshoot	Settling Time	Steady-state Error
P	Decrease	Increase	Small Change	Decrease
I	Decrease	Increase	Increase	Eliminate
D	Small Change	Decrease	Decrease	Small Change

Neural delay and the activation dynamics time constants have so far been ignored in this study.
The controller output, which is the muscle activation state necessary to combat the angular deviation from the reference, is expressed in arbitrary units and sent to the corresponding muscle after normalization.
An early set of control parameters (obtained through Ziegler-Nichols) yielded a very satisfactory and credible HNC behaviour in most situations, but sudden shifts in the input trajectory led to non-physiological reaction times between the reversion of the previous trend and the adequate response to the next, along with occasional resonant oscillatory results. Trial-and-error was then used to fine tune the parameters until the response was adequate.

Human body model on APROSYS motorcycle

MADYMO allows complex models to be "driven" by means of supplying the positions and angular motions that they should follow over time. As a result, one can observe the model's reactions (both in terms of animations and the time evolution of several key parameters, like the angles between the neck and several spatial references) when it follows any trajectory which is considered relevant to understand the HNC's behaviour. The "dynamical" nature of the target scenarios requires a trajectory (as opposed to the traditionally used impact acceleration pulses) that can be fed to a proxy for a motorcycle which the human model is "riding", so this motion will be completely prescribed for the model to follow. This approach will ensure that the human model's posture and external loading profile is consistent with real (or at least plausible) road situations, which naturally include gravity in all simulations. For simplicity, and also because it was not the focus of this work, no detailed description of the motorcycle's "banking" when cornering was developed, so its motion involves just the three planar degrees of freedom. The chosen model was developed within [APROSYS], representing a "touring-style" vehicle that can be considered typical of one of the most common classes of PTWs in Europe.
The MADYMO facet/multibody human model [Lange et al 2005] was adopted in conjunction with the MADYMO detailed neck model [vd Horst 2002] as the basis for the work described in this paper.
To ensure that the rider followed the motorcycle, "point restraints" were implemented between the wrists and handlebar, the feet and feet rest, and pelvis and motorcycle seat. These restraints apply supportive forces (between the corresponding bodies) which increase quickly with distance (up to 30 kN for 10 cm), so the human body model is adequately secured to its "surroundings" which is what is actually being driven with the trajectory. Finally, the HBM spine joints were locked to ensure the torso stayed upright, focusing the analysis on the HNC.

Figure 4. MADYMO human body model positioned on APROSYS motorcycle.

DYNAMICAL SCENARIOS

Having devised an actively controlled computational model of the human HNC (along with the rest of the still passive human body and motorcycle), the system should be put to the test under simple conditions so that relevant outputs can be obtained and analysed in the quest for biofidelity. Two simple trials were thus conceived to assess whether or not the model was reacting adequately, one for longitudinal accelerations and the other for lateral (roll) ones. In these straightforward examples, the rotation in the sagital plane (linear sled test) and in the frontal plane (circular test) can be studied separately. In both test cases, for each time step chosen for the multibody calculations, MADYMO requires a global (i.e., referenced to the reference space) XX and YY position (the motorcycle follows planar trajectories) as well as an angular heading so that it moves along the desired trajectory. Computing these (XX, YY) pairs is trivial and will not be described here.

Simulated linear sled test

The first test is a softer version of the rocket sled ridden by Colonel John Paul Stapp. In addition to being a somewhat historic experiment, it is a simple trial that may be reproduced in real life albeit if only with very high performing vehicles. In this test, the modelled motorcycle is accelerated from naught to about 180 km/h at 1.7G for three seconds, then it retains its speed for two seconds before finally braking for three more seconds at a constant 1.7G (Figures 5-8). This sort of acceleration is attainable with top sport motorcycles (so-called “superbikes”), while the braking deceleration is limited to very special roads cars or racing cars. The global scenario is therefore a sequence of intensive conditions, illustrating how demanding the non-impact settings this work is aiming for can be. During the course of the test, the vehicle travels about 240m.

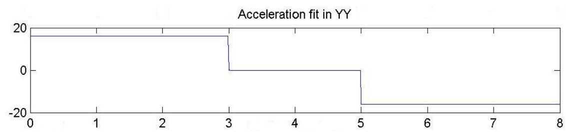


Figure 5. Acceleration felt during the sled test ((m.s-2) vs. time (s)).

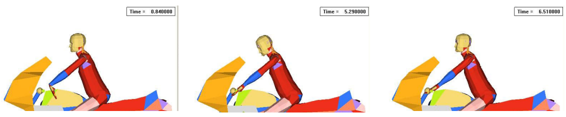


Figure 6. Screenshots from the linear sled acceleration test, corresponding to accelerating, constant speed and decelerating (left to right).

The three stages of this experiment are very distinguishable when one analyses the model’s behaviour. Three screenshots were taken at key moments of the simulation (previous figure) to help visualize and understand the simulated response. They all allow the observation of the hands of the model: because only the wrist is restrained to the handles, the hands themselves display their inertial response by pointing back, down and to the front as the motorcycle accelerates, cruises and then brakes.

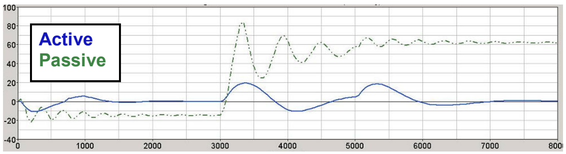


Figure 7. Active and passive HNC pitch angle for the sled test (degrees vs. time (ms)).

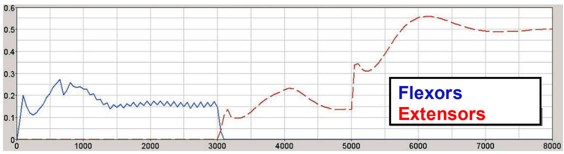


Figure 8. Muscle activation curves for the sled test (activation fraction vs. time (ms)).

As shown in Figure 7, the passive HNC extends about 15 deg backwards while accelerating, flexes about 70-80 degrees forward during cruising, and stabilizes at a 60 degree flexed posture while braking. The results from this 8 second long trial suggest that, based on the limited data available, the active HNC’s response can be judged to be biofidelic while enduring significant longitudinal accelerations: the HNC never tilts more than roughly 20º from verticality under either positive or negative acceleration. The distribution of intervertebral rotations through the neck is quite homogeneous. The flexor muscle group (of which the *longus colli* is an example), exhibits very moderate activation (15 to 25%) only during the first phase as it is enough to maintain the desired posture. For the second phase a *Semispinalis Cervicis* was chosen to represent the “extensor” muscle group. The extensors’ initial response displays some overshoot but eventually the signal stabilizes within the 2 seconds of the “constant speed” phase, at roughly 15% of the full activation potential. The next overshoot, from the onset of braking, is dealt with less smoothly (activation peaks at 55%) but the extensor muscles reach equilibrium with the external stimulation at 50% after 2 seconds. Owing in part to the locked spinal joints, T1 rotation never exceeds 10 deg in this simulation, which implies that the results actually relate to head and

neck stabilisation and are not compounded by spinal motion below T1. The maximum displacements occur with the 2 deceleration “initiations”: forward acceleration stops at 3.0 sec and the head is pushed forward as a result of inertia, and later (5 sec) the actual braking again propels the head (and indeed the whole body, to a lesser extent) forward.
The reaction time needed to counteract all of the accelerations and nullify the angle never exceeds one second, even though the two instances mentioned above are very demanding.
All the lateral roll outputs are null throughout, as they should be since there is no lateral acceleration.

Simulated uniform circular motion test

The second experiment consists of a uniform circular motion that gives rise to a constant lateral acceleration, much like the centrifuge used to test the maximum g-force that a fighter pilot can withstand. The simulation is carried out with a lateral acceleration of 0.8G over a 5m radius circular trajectory for the same eight seconds, which corresponds to an angular velocity of about 1,26 rad.s-1. These acceleration values are attainable in most everyday cars, but might actually not be easy to reproduce in a motorcycle – at least without tilting it laterally. This motorcycle-specific “limitation” is immaterial to the work being developed on the model’s HNC, and more realistic motorcycle trajectories are needed.

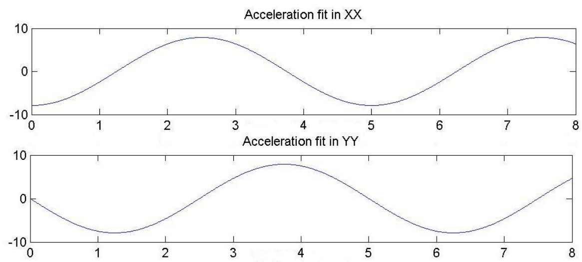


Figure 9. Constant lateral acceleration XX and YY (global) projection vs. time (s).

In this figure one may observe the projected acceleration patterns used to simulate the uniform circular motion in MADYMO. Since lateral acceleration is constant, its projections in the (global) XX and YY axis are sinusoidal and their phase difference 90º, which should be expected for such a movement. The model responded as follows:

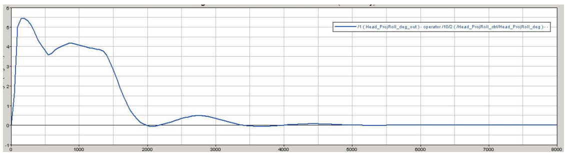


Figure 10. Active HNC roll angle for the circle test (degrees vs. time (ms)).

Figure 11. Muscle activation curves for the circle test (activation fraction vs. time (ms)).

The passive HNC presents a trivial response: the roll angle shoots to its maximum value of 30 degrees and remains there throughout the simulation. It was not included in the graphic to facilitate readability and avoid representation issues due to disparate scaling. The trajectory provided here is far more homogeneous than in the previous case: the longitudinal and lateral accelerations are null and constant, respectively. Consequentially, the active HNC needed only to counteract the outward-propelling centrifugal acceleration (~ 0.8 G) and within little more than two seconds it had endured the maximum angular displacement (< 6º), forced a very slight inward overshoot, and attained dynamic equilibrium with the centrifugal force at 0º. The previous figure shows that the left “roller” muscle group (of which the *trapezius* is an example) displays a quick activation spike to 20% in the first quarter of a second while the controller stabilized the HNC against the external stimulation (which pushed the head to the right). The activation state then rose steadily over the next 2 seconds, and once the control response was in steady-state 55% of this muscle group’s activation potential was eventually required to counter the constant lateral acceleration. As expected, the right *trapezius* (and the other “right” rollers) did not display any noticeable activation.

EXPERIMENTAL BRAKING SLED TRIAL

In order to preliminarily validate the model response for a typical riding scenario, volunteer trials were conducted using an inverted braking sled setup [Symeonidis et al 2008]. Eight volunteers participated in the experiment. Steady-state decelerations of 0,2G and 0,4G were employed in two modes: “aware” (the volunteer triggered the sled motion) and “unaware” (the sled was launched by the researcher, unbeknownst to the volunteer). The riders’ kinematics and muscle activation patterns

were captured with an optoelectronic motion capture system and electromyography equipment.
The corresponding analysis is still not complete and will be adequately published in the near future, but in order to include some initial insight for this paper, one set of data was chosen for visualization. The selected case was the 0,4G aware run of a volunteer with average (“middle of the corridor”) responses. Perhaps because of these factors, the resulting HNC kinematics was trivial: very slight oscillation of the head around verticality, never exceeding 2 degrees. Using the acceleration data measured at the sled, the model presented in this paper was subjected to the same trial and yielded the following response, in which the deviation reaches a maximum of 6 degrees.

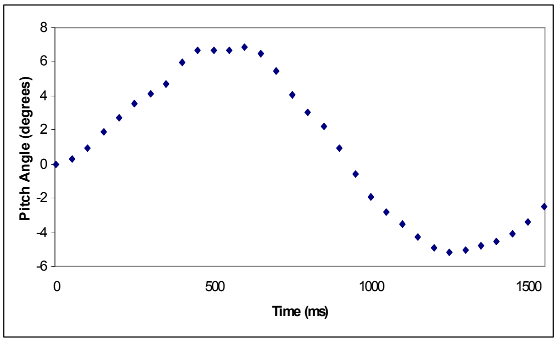


Figure 12. Active HNC pitch angle for the inverted braking sled trial (degrees vs. time (ms)).

Although further analyses will still be developed, it seems that the simulated HNC kinematics may be compatible with the response of this aware volunteer.

CONCLUSIONS

This paper allows for one foremost conclusion: a computational model of a human body, equipped with a feedback-controlled HNC, does seem to reasonably maintain its target erect posture in simple, one-degree-of-freedom loading scenarios. A very preliminary comparison with an experimental sled braking study seems to confirm that same conclusion. The authors acknowledge, however, that unlike a real human subject this model is not able to predict future events, which would be especially relevant in a test with changing trends like the first one, and so the proposed computational solution is (at least at this stage) simply reactive. Some sort of prediction or learning may become possible in the future, since the authors believe a new control paradigm is required (e.g. neural networks) and will attempt to implement. Even so, the outputs presented so far suggest that the active HNC is able to mimic expected human reactions in an acceptably biofidelic manner, at least if one assumes riders attempt to maintain their head upright at all times. In fact, a rider’s (or driver’s) priority would probably not be his comfort but rather his ability to maintain kinematic stability between his visual senses and the vehicle’s behaviour, thus emphasizing the need for an adequate posture. A vehicle’s passenger, however, will probably not forcefully maintain his head’s verticality but instead minimise effort or possibly balance several strategies. This conclusion mainly draws upon the reactions (and other selected outputs) provided by such an active model, both at “pitch” and “roll” levels, when it went through a preliminary analysis in a couple of scenarios. Time-dependent position data was used to build those scenarios, devising a general procedure that can be applied to more elaborate situations.
The controlled HNC assembly was itself the outcome of applying active control methodologies to a multibody model that was augmented to comprise a facet human body model and a touring motorcycle that was propelled along the chosen trajectories. Sensors and muscles groups were thus selected, tested, and adequately implemented in the model.

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