A NEW METHODOLOGY FOR BIOFIDELIC HEAD-NECK POSTURAL CONTROL

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ABSTRACT

Active safety systems require accurate modelling of human behaviour. The effect of bracing, together with an appropriate human head/neck stability control are essential in the study of pre-crash behaviour and neck injury, since they can account for changes in apparent joint stiffness in the neck. In this paper a biofidelic postural control of the head and neck with co-contraction ratio is implemented. The controller is implemented in Matlab and coupled with a MADYMO human body model. A validation is performed, comparing the new controlled system with experiments with test subjects in the pitch direction.

Keywords: Biofidelity, **Biomechanics**, **Kinematics**, **Models**, **Neck**.

OVER THE YEARS, a large number of protective measures avoiding or mitigating traffic accidents have been devised. Cars today have many systems that activate before, during, or after a crash in order to minimize the damage to its occupants. Helmets developed for motorcycles are able to resist very strong impacts and have helped the human head become less and less vulnerable. In particular, for low speed, urban collisions [Fraga, 2009], the protection offered by helmets has come to a point where the head tends to lose its place as one of the most fragile parts of the body. The neck on the other hand, generally unprotected, remains to be of critical importance during all type of collisions. Any traffic accident can have serious consequences in the neck, ranging from whiplash in car collisions, to severe injuries and death for motorcycle accidents. More than 35% of motorcycle riders involved in accidents suffer from head and neck injuries, and 12% of those injuries are located specifically in the neck [APROSYS, 2005]. Understanding the dynamics of the head and neck is then crucial to developing new protections that can improve the safety on the roads. In motorcycle drivers in particular, a correct modelling of the neck can prove to be highly important for the task of modelling the whole crash. The reason is that the actions of the driver have a considerable impact on the dynamics of the whole system, given that the inertia of the driver is, unlike with a car, a significant proportion of the inertia of the system. Moreover, having a biofidelic neck behaviour prior to crash allows for studying effects of active safety systems on head/neck response and may influence head/neck orientation at time of impact, which in its turn influences crash response.

The objective of the study is to obtain biofidelic responses of the model in the three rotational axes (roll, pitch, yaw), together with the implementation of a 'bracing' behaviour that can account for changes of stiffness and damping of the neck during collisions. A controller design is then proposed and implemented in Matlab, coupled with the MADYMO [MADYMO, 2009] software. The starting point is a passive human body model with Hill-type muscles in the neck [van der Horst, 2002]. Findings from an earlier attempt to implement posture control on a detailed neck model [Fraga, 2009] where used to design this completely new controller concept. The design is contrasted with human test subjects in order to analyse biofidelity in frequency behaviour and crash dynamics, in the pitch direction. Validation for roll and yaw is left for future work.

METHODOLOGY

The way that humans have to control the position of the neck is to this day not fully understood. Many research projects devote resources to understanding the impressively complex controller that the

human central nervous system can be. Aspiring to achieve such a complexity in a simulation-software would be nothing but too ambitious at this point, given the state of the art in human head and neck control [Bove, 2009 & Chih-Hsiu Cheng, 2009 & Gurses, 2005]. Moreover, in order to reach a sufficiently high level of complexity that can effectively account for human biofidelic postural control, a path has to be followed, from an initially simple model, to a more thorough and complete one.

Because the system is so complex in nature, a first step towards a complex model consists of making some simplifications. The obstacles to be overcome, either because of the nature of the task, or coming of the simplifications made, are then described. A design will then be proposed for the controller. Once the general structure of the controller is defined, a better understanding of the influence of the muscles in the neck's dynamics is needed, and the procedure to obtain it and the information learnt will be summed up in a Muscle Effect Simulation Database. Finally, the need arises to elaborate a strategy with respect to the muscles' roles from the controller's perspective. This will be defined as the Muscle Recruitment Strategy.

SIMPLIFICATIONS

Human Body Model (HBM). The chosen software platform was MADYMO [MADYMO, 2009]. Specifically, a detailed neck model was used, which has a thorough model of the skull and vertebras, inter-vertebral discs, ligaments, and muscles [van der Horst, 2002]. In particular, the muscles are modelled as Hill muscle models. Since this is the model used, the controller which is the scope of this paper is fully based on it.

Control loops: humans have several types of feedback in order to control movement. There is literature on the different feedbacks (force feedback, visual feedback, vestibular feedback, etc), and the relationship between them [van der Helm, 1998 & Verdaasdonk, 2004 & Lackner, 2005]. However, including all the feedback loops in a controller would enormously increase its complexity, and for this reason only a simple position feedback was used, in the form of three angular position sensors that indicated to the controller the angular displacement of the skull with respect to T1 vertebrae. These sensors refer to pitch, roll and yaw angles of the head.

Linearity*:* The head and neck complex (HNC) has several properties that make it non-linear. Muscle force varies with relative elongation, contraction speed or fatigue level, among other parameters, and the effect of the muscles varies according to the position of the HNC with respect to the torso. Moreover, joints and muscles have a complex non-linear stiffness function. In order to reach a first order solution, the present non-linearities of the model were not compensated for. This means that a linear controller was considered suitable enough, despite the presence of non-linear components.

Reference*:* Not taking into account non-linearities makes the task of the controller much more difficult. Especially when considering that the relative position of the HNC modifies most of the non-linear properties (muscle effects, joint and muscle stiffness's, etc). Therefore, it was chosen to have only one working reference position –which in turn means that the controller will have a fixed position to control-, as a starting point for the controller design. This position was chosen to be the most neutral and relaxed position possible, and its set-up will be described later.

Actuators*:* the Human Body Model (HBM) available in the software simulation package, MADYMO, has 32 Hill-type muscles in the neck, each of them with different fascicles to account for different origin/insertion points. Each of these fascicles can be independently activated. Therefore, the number of possible actuators available rise to an impressive 136. Considering that there are only three sensors for the controller, having so many actuators hinders the design process, causing a higher cost in effort, than the benefit it provides by allowing more degrees of freedom. For this reason, it was decided to activate each muscle's fascicles all together, bringing the actuator count down to a more manageable 32 (see Table 1).

Table 1: List of muscles present in the neck model. Each of these muscles is present both in the left and the right side of the neck, making the list 32 muscles long.

1	Lumped Hyoids	9	Scalenus Medius
2	Levator Scapulae	10	Scalenus Posterior
3	Longissimus Capitis	11	Semispinalis Capitis
4	Longissimus Cervicis	12	Semispinalis Cervicis
5	Longus Capitis	13	Splenius Capitis
6	Longus Colli	14	Splenius Cervicis
7	Multifidus Cervicis	15	Sternocleidomastoideus
8	Scalenus Anterior	16	Trapezius

Reflex reactions*:* There are several levels of reactions. An intended movement being the most conscious one and a direct muscle reflex being the least conscious one. However, in order to account for the different types of reflexes, all the control loops must be taken into account, and this has been ruled out (see above). However, a proper biofidelic controller must have some reflex behaviour included. One of the most common reflexes prior to a possible impact is bracing by co-contracting a large set of muscles. Thus, a generic Co-Contraction Ratio (CCR) was included in the controller. The activation of this behaviour is generic and will be described further in the design section.

Neural delay*:* between intended activation, and real activation of a muscle, there is a neural delay which has been found to be in the order of a hundred milliseconds [Siegmund, 1986]. In order to keep the controller as simple as possible, the neural delay was omitted in this design. It should be noted though that since reflexes are grouped into the CCR, the controller itself can be 'allowed to be slow', (i.e. by means of a low gain) while still holding a biofidelic behaviour

Controller: Due to its simplicity, effectiveness, and speed, a Proportional-Integrative-Derivative (PID) controller was the chosen control strategy. Such a controller was useful in previous applications [Cappon, 2007 & Fraga, 2009], and although it doesn't account for non-linear behaviour, it is a very efficient solution considering that a fixed reference position is to be used. Using this type of controller brings some new problems.

CONTROL PROBLEM DESCRIPTION. The main issue with controlling the HNC is that the system is completely coupled. There is no obvious and evident way to link three sensors to 32 actuators. None of the muscles generate a rotation on a standard axis (i.e., no muscle can generate only roll, only pitch, or only yaw, but always a combination of the three), and in order for the controller to be biofidelic, it should use all the muscles. Now, a PID controller can only handle one degree of freedom, so the controller needs three independent, uncoupled PIDs in order to control the three degrees of freedom. The outputs of the PIDs, however, have to be the activation levels of 32 muscles. This means that the problem is under-determined, as there are 32 unknowns, and only 3 equations.

Also, it was said that the PIDs need to be uncoupled. This means that the actuator from a PID should not generate a disturbance of the degrees of freedom controlled by the other two. In other words, the pitch controller should never generate roll or yaw movements, and similar restrictions apply to the other two controllers. Since all the muscles generate combinations of all three movements, independent muscles cannot be used as direct actuators, and a combination of them has to be used. Fortunately, because the muscles are (ideally) symmetrical to the sides of the sagittal plane, a solution for the pitch movement is quite simpler: when a specific muscle is activated in both left and right sides, the resultant torque generated actuates only in the sagittal plane, which is to say that roll and yaw are automatically compensated. However, muscle groupings that can generate only a roll torque, or a yaw torque are not so simple to find.

Finally, a PID is a linear controller. The system to be controlled, however, is not. On one hand, even a simple muscle model (Hill type muscle model) has a non-linear behaviour with respect to parameters such as relative elongation or elongation speed. On the other hand, as this is a dynamical system, the effects of a muscle in a specific position are very different to the effects of the same muscle, when the head is in a different position. This means that even when a muscle group is chosen for a specific condition, it will not be completely valid as the HNC rotates.

DESIGN**.** The distribution of the loads related to neck dynamics is often referred to as the 'load sharing problem'. This can be seen basically as the problem of how human beings use certain muscles for certain tasks. It has been stated [van der Helm, 1998] that human beings can solve the load sharing problem with the use of a specific strategy related to how motor units in the muscles are controlled, or 'recruited'. This muscle recruitment strategy is key to understanding how all the muscles in the neck activate together to perform a given task.

The first two problems described in the previous section can be solved by using a reliable muscle recruitment strategy for the load sharing problem [Damsgaard & Dul, 1984 & Hawkins, 1992]. Or, in other words, the system can be both simplified and decoupled by correctly choosing a combination of muscles and the degree to which they are individually activated for every specific task. Without an adequate muscle recruitment strategy, the actuators of the PIDs are very likely to interfere with each other. This would result in normal disturbances that are present in the system to be greatly increased due to the coupling between the controllers. The outcome would then be a highly over-activated system with a chance of being very oscillatory, or even unstable. For this reason, it is critical to choose a proper muscle recruitment strategy that can minimize the coupling of the three PIDs. The practical result of such a recruitment strategy would be no more and no less than three vectors. Each of these vectors can contain activation signals for each muscle, in such a way that activating the muscles with the coefficients of a vector, would result in a simple one degree of freedom rotation. In other words, applying the coefficients of the left roll vector directly as activation signals would then result in a pure left roll. A coefficient multiplying the entire vector would then determine the degree of activation of this new "left roll actuator".

Since a normal human being can easily generate torques in any standard rotation direction, it is clear that this proposed muscle recruitment strategy that has only one component of torque is possible. Although it is unreasonable to assume that this vector would remain the same for different positions of the HNC, given the need of a fixed reference position, only one position was analyzed to simplify the problem. It will be shown later that such a simplification does not stand in the way of developing a more general controller that can work in more reference positions.

Once a position is chosen for the analysis, a muscle recruitment strategy that is both biofidelic, and can independently control roll, pitch and yaw can be sought. However, in order to analyse the effect of possible muscle combinations, in-depth knowledge of the effects of every single muscle in the movement of the HNC is needed. This was obtained by simulating a scenario in which only one muscle is activated, and the torques in the HNC are computed. If this simulation is repeated for every muscle of the neck, a thorough database of the muscles' effects on the HNC can be obtained.

A mathematical optimization algorithm can now be used for choosing the right muscle recruitment strategy: by applying a certain set of activation signals in a certain group of muscles, for example, a minimum-fatigue, only-roll movement can be obtained. Repeating the procedure for the three rotations results in the appropriate muscle groups that can be used directly as actuators for the PIDs. More precisely there are four rotations, considering that the muscle recruitment strategy for pure flexion will not be equal to the one used for extension, since there is no symmetry between flexor and extensor muscle action. However, in this study flexion and extension are considered one rotation degree of freedom.

CCR on the other hand can be thought of as the degree of activation of another muscle group that has, however, no resultant torque in the neck. The CCR can then be handled independently of the PIDs, with a Variable Contraction Setpoint (VCS): a reference that can vary dynamically according to a pre-defined value or curve. In other words, the CCR can be set to follow any desired signal: a low, fixed value in order to simulate a 'sleepy' person, whereas a mathematical time-varying function depending on, for instance, the accelerations to which the model is subjected would be in accordance to an 'aware' person adjusting his or her level of co-contraction as a function of that function.

This way, the HNC would be controlled in four different ways: three PIDs would, independently, control the degree of roll, pitch and yaw rotation, while a user-defined variable, the VCS, would control the CCR.

A schematic of the controller can be seen in figure 1.

The controller can be divided into four sections:

- The roll controller
- The pitch controller
- The yaw controller
- The CCR control

Each of these parts can be also subdivided. Each PID consists of a reference –the desired position–, the PID parameters themselves (K_P, K_I, K_D), and a vector, containing the coefficients obtained with the load sharing optimization algorithm. The CCR control on the other hand also has a muscle vector, but no 'PID parameters'. The VCS on the other hand is equal in practical terms to the reference.

Each of these components provides 32 activation signals for the 32 muscles, and since in this model the force exerted by the muscles grows linearly with respect to the activation signal, they can be added individually to come up with the final 32 activation signals that are inputted in MADYMO. Finally, the loop is completed when three position sensors coming from MADYMO are fed back into the system, and subtracted from the references.

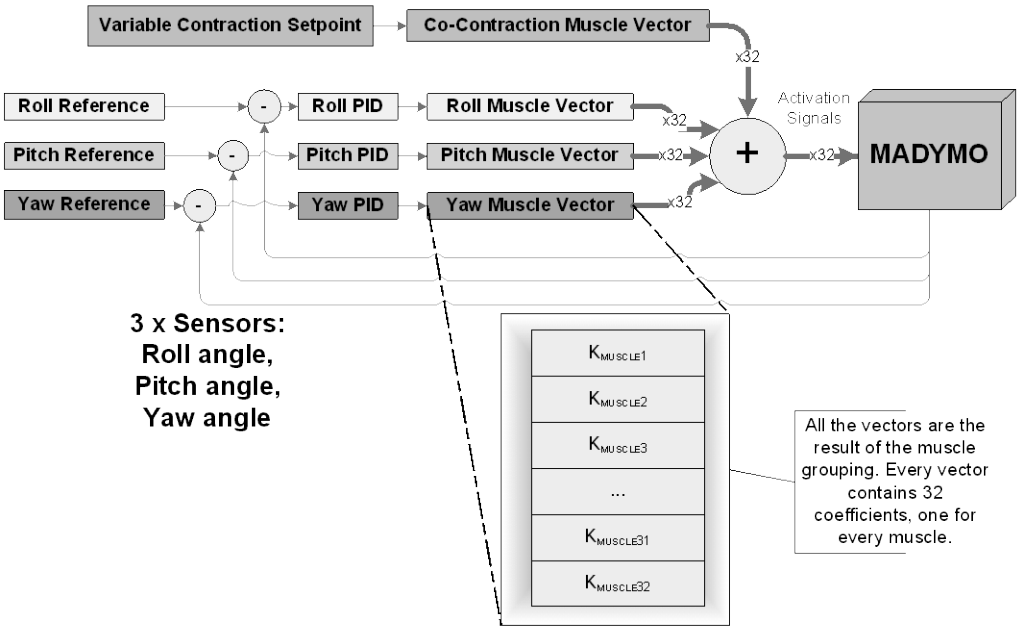

Fig. 1: General schematic depicting the controller strategy.

MUSCLE EFFECT SIMULATION DATABASE. The concept is based on a static simulation. The objective is to obtain a vector containing for the 32 muscles the torque that each muscle can exert in a given condition. So, simulations need to be performed in order to compute the forces and torques of the muscles in a specific position, for a specific case. Thus, it is imperative to carefully choose the initial conditions for the simulations, since it will be also the choice of the optimal condition for the controller. Therefore, there are several issues to take into account before performing the actual measurements, and these will be described in the following sections.

Position. The position chosen for the cervical joints was the most relaxed possible: the system was simulated without gravity, and let free until it reached a stable position. The angles of the joints in this position were then used as the fixed position for the simulations. This orientation corresponded to the muscles being completely relaxed –only possible given the lack of gravity in the simulation– and the posture of a human looking straight ahead. The position of the whole human body model was automotive seated, and based on a position used before with the model [van der Horst, 2002]

Activation Signal. The activation signal is modelled as a variable that ranges from 0 to 1, and gives the degree of activation of a muscle, where a 0 as an activation signal means a passive muscle, and an activation signal of 1 means that the muscle will contract to its maximum capacity. An activation signal needs to be chosen for determining the torques. In principle, the activation signal chosen for the measurements should have no consequences on the activation-torque relationship, since the muscle model behaves linearly with respect to it. Nonetheless, it is also true that some of the forces present in the system are not coming from the muscles themselves: passive forces from tendons, ligaments, etc. Thus, a too small activation signal would be counterproductive since these 'other' forces would introduce a high 'noise' in the simulation. Also, for position stabilization in normal scenarios, the activation is rarely high. For these reasons, an activation signal of 50% was chosen, since it is enough for the muscles to generate forces at least ten times higher than the other forces present in this neutral head orientation, while still being a round number that facilitates calculations in the following procedures. It is worth mentioning though that such an activation signal would only be used sporadically. Moreover, the accelerations generated by activation signals this high are not manageable by normal head stabilization. That is to say, these activation signals would not be present when a precise stabilizing control is needed; they would only be appearing when movement control as a result of large acceleration field or force applied to the head is needed, or, more likely, as a result of a reflex as a result of this same external load.

Muscle Tested. All the muscles present in the HNC of the model were tested, and a complete list of them can be seen in table.

Joints. Since the goal of this analysis was to compute torques in a static position, all the joints were locked. This avoided having a displacement in the HNC by the torques generated by the muscles. However, because the controller will of course work in a dynamical environment, a static measurement overlooks some issues. Specifically, there is no consideration for the stiffness of the joints, or the length/velocity behaviour of the muscles, and this will certainly be present during normal situations. So, to be sure that these effects were not too relevant for this simplified model, an analysis on the influence of these factors was made.

Interaction between muscles. In normal human beings, muscles are in contact with each other. Friction and viscosity forces are present when a muscle moves with respect to another, therefore reducing the total available force. However, the model used has line-element muscles. This means that forces between the muscles are not taken into account. Although it can be a factor, it only implies that the total available force of a numerical muscle will be more than the force that same muscle in a real human. The difference between the case of a normal human and the model will therefore only impact slightly on biofidelity, and in a much lower degree than the simplifications stated before. And while this is an obvious disadvantage, the greatly reduced computation time renders the system much more usable than if 3D FE muscles that allow interaction were to be used [Hedenstierna, 2008].

The stiffness of the joints for a sufficiently relevant deviation and speed was analysed. In a normal scenario, there should ideally be very little movement, and almost no deviation from the central position, since the controller would be taking care of maintaining that equilibrium. In spite of this, the effects of the joints properties were still measured in order to account for the low speeds and deviations that in a normal case would exist.

Also, the Hill muscle model has two characteristics that will add other forces and torques to the active ones. An elastic behaviour is present when the muscle elongation is other than the one in resting position, and a viscous behaviour is present during when there is a non-zero lengthening velocity.

The forces present with the joints unlocked were measured for a 50% activation signal (and the speeds and accelerations that result), and contrasted to those corresponding to the locked joints, arriving to the conclusion that those effects were not significant enough to take them into account for rotations below 5°.

Torques. Once computed, the torques were decomposed in the three desired components: pitch, roll and yaw. The main results are summarized in the table 2, in which it can be seen, for every muscle, the amount of torque generated in each direction. It should be noted that these torques are generated with an activation signal of 50%. Due to the linear relationship between activation and force, an activation signal of 100% would produce twice the torques present in table 2.

Table 2: Torques (M) generated by every muscle, total and for each standardized axis of rotation.

Muscle	M (Total) (N.m.)	Mx (Roll) (N.m.)	Type	My (Pitch) (N.m.)	Type	Mz (Yaw) (N.m.)	Type
Hyoids	20.86	12.50	Left	16.37	Flexor	3.28	Left
Levator Scapulae	26.11	22.58	Left	12.76	Extensor	3.06	Right
Longissimus Capitis	7.65	6.63	Left	3.44	Extensor	1.65	Left
Longissimus Cervicis	5.86	5.63	Left	1.61	Extensor	0.03	Right
Longus Capitis	3.76	2.35	Left	2.88	Flexor	0.57	Right
Longus Colli	7.89	3.18	Left	7.20	Flexor	0.48	Left
Multifidus Cervicis	11.57	1.60	Right	9.11	Extensor	6.95	Left
Scalenus Anterior	8.37	7.91	Left	1.67	Flexor	2.19	Left
Scalenus Medius	8.62	8.52	Left	0.22	Flexor	1.27	Left
Scalenus Posterior	3.62	3.60	Left	0.17	Extensor	0.33	Right
Semispinalis Capitis	39.97	19.39	Left	31.44	Extensor	15.26	Left
Semispinalis Cervicis	11.19	0.11	Right	10.27	Extensor	4.43	Left
Splenius Capitis	45.92	22.67	Left	38.12	Extensor	11.93	Right
Splenius Cervicis	17.14	11.39	Left	11.67	Extensor	5.30	Right
Sternocleidomastoideus	60.86	51.09	Left	15.12	Extensor	29.41	Left
Trapezius	27.41	8.90	Left	16.64	Extensor	19.89	Left

MUSCLE RECRUITMENT STRATEGY. The main objective of the muscle recruitment strategy is to obtain groups of muscles, that when activated together and in the proper proportion, a pure rotation (roll, pitch or yaw) is achieved. However, this would mean that there are 32 coefficients, and only 3 conditions for every group. For example, in order to obtain the coefficients for the roll group, the conditions would be:

$M_X > 0$ for left roll $\qquad$ $M_X < 0$ for right roll
$M_Y = 0$
$M_Z = 0$

It is straight forward that there are many more degrees of freedom to control than equations. Therefore, it is necessary to use some criteria that can be used both to find a unique solution, as well as to ensure that the solution found has an acceptable level of biofidelity.

There is literature on different muscle recruitment strategies, and the load sharing problem [Hawkins, 1992 & Dul, 1984]. Most of the literature refers to minimizing a function. In this case, as a first approach, the algorithm is as simple as possible:

To begin with, given a specific rotation, only the muscles that can positively participate in that rotation are activated. Then, the first approximation for the activation signals is to make all of them equal. For flexion or extension, this works, since the muscles are completely symmetric. Therefore, if all the pitch flexors activate simultaneously and in equal degree, the left and right movements (roll and yaw) are automatically compensated by the same muscles on the other side of the neck (e.g., if the left longus colli is actuated simultaneously and in the same degree as the right longus colli, the result is pure pitch flexion).

Because extensors are stronger than flexors [van der Horst, 2002], an equal activation of all muscles contributing to left roll (both front and back, i.e., pitch flexors and extensors) will result in left roll and to some degree, extension. The most logical solution is to activate the muscles contributing to left roll (here called left rollers) with an extension function to a lower degree than the left rollers with flexion. Since the left rollers are activated all simultaneously, the strength difference between *all* extensors and *all* flexors was computed. Then, this strength difference was included as an extra coefficient for all movements that could compensate the difference in strength between flexors and extensors. In other words, since the flexors are significantly stronger than the extensors, they were activated significantly more to compensate.

There was no compensation for roll present in yaw, or yaw present in roll. This was a consequence of a small degree of roll present with activation of “yawers”, and viceversa, together with the fact that achieving such compensation would escalate the complexity of the muscle recruitment strategy.

This approach provides a very simple muscle recruitment strategy that makes the load sharing equal for every muscle in terms of exerted forces with respect to maximum force.

With respect to CCR, the strategy was just as simple. Roll and yaw can be compensated by activating simultaneously and equally the muscles from the left and from the right, and the remaining degree of freedom, pitch, can be compensated with the ‘extra coefficients’ mentioned above. If 100% of CCR is the maximum level of contraction that the model can have on its muscles without having a resulting torque, then the CCR is limited by the maximum force of the flexors. That is, with a 100% CCR, flexors are activated to maximum strength, whereas extensors are activated till flexors are compensated.

Finally, the whole controller was implemented in Matlab/Simulink. A coupling was established between Matlab/Simulink and MADYMO in such a way that for every time step of the simulation MADYMO provides Matlab with the dynamics of the system, and Matlab/Simulink provides MADYMO with the activation signals of the muscles.

PID parameters were tuned using impulse responses of the model, and assessing visual biofidelity, in several iterations. Seemingly human-like oscillation times, overshoot, and time to stabilization was all considered during the determination of the controller parameters.

RESULTS

Because the controller is clearly separated into two very different components (the PIDs and the CCR), an initial validation was performed following a similar criteria: in order to assess the behaviour of the CCR, the system was compared with a rear impact sled test with maximum acceleration of 3.6g performed with healthy volunteer test subjects [Ono, 1999], in which the stabilizing controlling behaviour is assumed to have limited influence (in comparison to CCR). In the same way, the behaviour of the PID control was compared to a test performed in the past in which the frequency response of healthy subjects was measured [Keshner, 2003]. The test was chosen so that the CCR influence would have minimum influence (the fact that the test lasted for several minutes for every subject, and used low-g would be an indicator that CCR is likely to be both low and constant). Both test setups can be seen in figure 2.

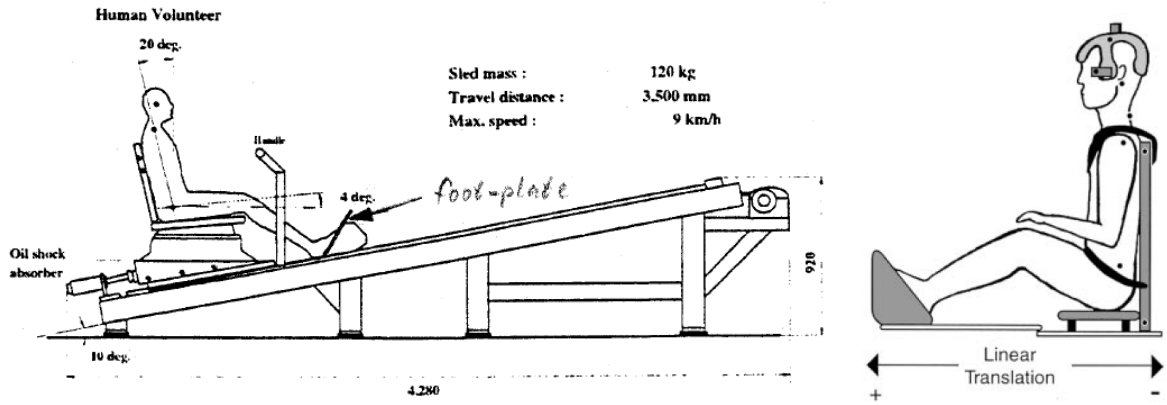

Fig. 2: Left: JARI test apparatus. Right: Keshner test apparatus.

It was said before that the controller does not account for quick reactions, rather than show the behaviour of the human stabilisation. Therefore, given an impact scenario, that impact takes so little time, that it allows for no real stabilizing control of the human neck, rather than a reflex in which most likely muscle bracing or stiffening occurs. Having said this, it becomes clear that an impact test can be useful to validate the CCR settings with respect to HNC kinematics, whereas the controller itself would not have enough time to fully act.

At JARI, low speed rear impact sled tests were performed with 9 male adult volunteers [Ono, 1999]. A rigid seat was used mounted at an angle of 10 degrees with respect to the horizontal. The maximum acceleration of the sled was of 3.6 g. The volunteers were fitted with film targets and accelerometers to record head and T1 kinematics and dynamics.

The scenario was replicated with the software MADYMO, in order to validate a human body model [Happee, 2000] though without a controller for the neck. The same set-up was used in this case in order to assess the behaviour of the model with varying degrees of CCR.

The simulation was performed for 0%, 20%, 40% 60% and 80% levels of CCR. The controller was turned off in these simulations and all other parameters where equal. The rotation of the head was measured in the base of the skull, with respect to T1, with positive rotations corresponding to flexion. Since the test and simulation consist of movement in one dimension, only pitch was computed. The corridor generated in the original tests can be seen in figure 2, together with the curves corresponding to the different levels of CCR.

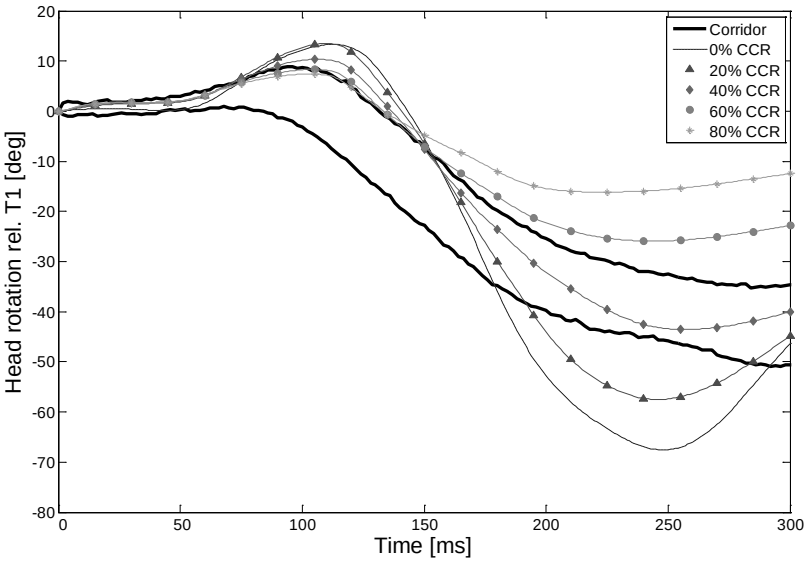

Fig. 2: Simulation of the JARI sled tests for different CCR and the controller turned off, against the corridor generated in that test.

It can be seen that a 0% CCR presents a behaviour well outside of the corridor, and the increase of the CCR results in a higher damping of the movement. The curve with CCR equal to 40% fits the corridor reasonably well, considering that the shape of the curve reassembles to some degree the shape of the corridor, and it lies within it most of the time. It can be seen that already from 60% the movement becomes too damped.

The task to test the controller with as little influence as possible from the CCR is not so trivial. Whereas the CCR has a strong influence on the general stiffness of the neck, the controller can be tested with an experiment with frequency perturbations. In principle, the frequency behaviour of the controller should be somewhat independent from the CCR. However, the CCR changes the overall stiffness of the neck, therefore changing the system on which the controller actuates. The frequency behaviour is then always influenced by the amount of CCR. At the same time, it is not unreasonable to assume that during normal stabilizing behaviour a healthy human being will have some level of CCR, however small it may be. On the other hand, it is likely that different healthy human beings can have control strategies that differ slightly, that in time correspond to different CCR levels when responding to the same perturbations. This in term implies that there is no unique level of CCR for all human beings.

In order to do this study, a series of experiments performed in the past were used [Keshner, 2003], where twelve test subjects sat on a sled, with the torso restrained. The sled was activated so that the resulting acceleration was a sum of sines, of eight different frequencies, from 0.35 Hz to 4.05 Hz.

The experiment was replicated in the simulation environment. The input used was the same as the one carried out in the original test, and the conditions were similar only up to a certain point: a thorough reproduction of the restraints was not possible, so what was reproduced was the intended result of the restraints; everything below T1 was immobilized with respect to the sled in the simulation, i.e. a perfect restraint was assumed. Although there is in principle not one single ‘correct’ value for CCR, a level of 20% was used as a first approximation, and it was compared to a CCR level of 0%.

The frequency response of the simulation was computed and compared to the one obtained with the test subjects. This can be seen in figure 3.

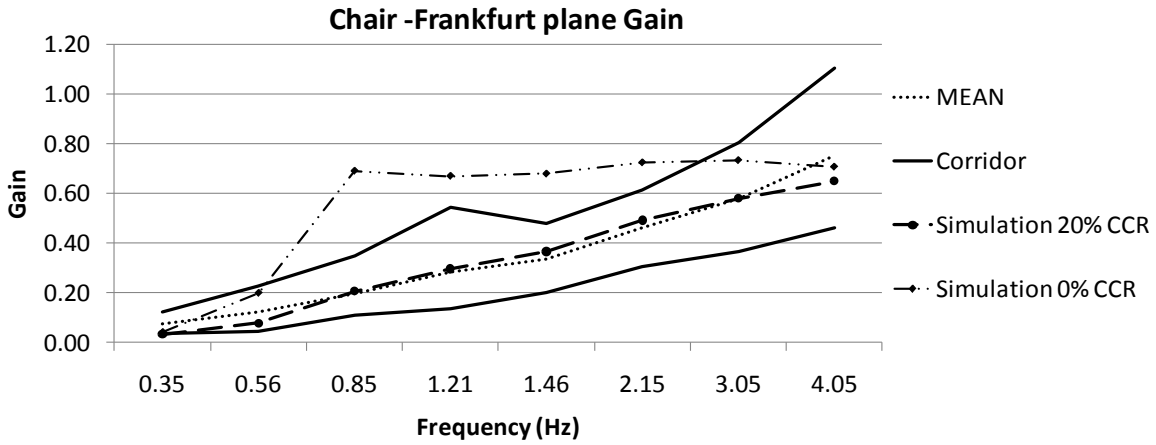

Fig. 3: Frequency response of the controller against the corridor generated in the Keshner tests.

The figure above shows the comparison between the frequency response of the test performed in healthy human subjects, and that of the controlled HBM with a 0% and 20% CCR. The straight lines show the corridor of the tests, with the thin dotted line being the mean of those tests, whereas the dashed line represents the response of the simulation. It can be seen from the figure that the controller's response with 20% CCR fits generally well with respect to the experiment. It does however stand lower than the corridor for 0.35 Hz. This means that the controller handles low frequencies better than a normal human being, which is to say that the head gets better isolated from the disturbances. It would be logical to assume that such behaviour can be tuned with the P, I and D parameters to better fit these curves.

The case of 0% CCR on the other hand, is quite outside of the corridor. It is not difficult to see first of all, the lack of biofidelity of a model that has no co-contraction whatsoever when subjected to a moving chair for a few minutes. From a more biomechanical perspective, it is logical to see that a lower CCR leads to a decreased stiffness and viscosity, which increases the difficulty of the controller to maintain a low gain.

The CCR can also be adjusted to obtain a better response, or other simplifications in the model cause the difference. All in all, further tests remain to be done. Frequency behaviour above 4.05 Hz on the other hand was not analysed during these tests, and this can have a strong influence on the biofidelity of the controller. It should be noted that in order for the controller to correctly imitate human behaviour, it should be consistent with it in all frequencies. However, because typical peak head accelerations in a EuroNCAP test are no faster than 10 ms duration from being small to peak [van der Laan, 2009], frequencies above 100Hz need no validation, as long as the controller is stable. At the same time, considering typical delays in the human control system [Siegmund, 1986], it is not possible to control perturbations of frequencies above 20 Hz.

CONCLUSIONS

A strategy for a controller for a numerical model of the human head-neck complex (HNC) with line-element muscles has been proposed. The strategy consists of three PIDs operating over roll, pitch and yaw angles of the head with respect to T1. On top of that, a co-contraction ratio is used as an approximation to various levels of 'bracing'.

The control strategy was implemented in Matlab/Simulink, which was coupled with a MADYMO human body model with a detailed neck that contains 32 Hill-type muscles models.

The system was compared against two different tests: rear impact sled tests with volunteers, performed by JARI, were used to assess the behaviour of the model with different levels of co-contraction (CCR). On the other hand, a frequency response experiment, limited from 0.35 Hz to 4.05 Hz, was used as comparison of the frequency behaviour of the controller, with an assumed level of co-contraction. All was done in the pitch direction of movement.

The results of both tests were encouraging. The comparison against the sled tests showed that adding a CCR to the model can help account for the difference in stiffness and damping in the neck model. The frequency behaviour on the other hand proved to be reasonably accurate, although the controller was seen to apply too much control in lower frequencies, when compared to the human subjects present in the experiment.

DISCUSSION

Results suggest that the task of developing a biofidelic controller for the human HNC is going in the right direction. The biofidelic frequency response achieved implies a biofidelic stabilizing behaviour, and the addition of the CCR can help model the stiffness/damping changes in the neck due to bracing. Considering that the controller does not include neural delay or different type of reflexes, the results obtained are overall very positive.

In spite of this, the results are only partial. The two validations done are partial in themselves, and the tests used for contrasting the controller are not nearly enough for a thorough validation of it.

The first and perhaps most important part of the controller that remains without validation is its possibility to control all three rotations of the HNC. The tests performed were only in the direction of pitch, leaving out of the picture both roll and yaw.

Both comparisons were intended to test only one parameter, by leaving the other outside. The frequency response test was performed assumed a fixed, ‘reasonable’ CCR, and the linear sled tests were simulated assuming no influence of the human control. Both assumptions are too simplistic in nature, leaving room for some variation in the strategies to validate just as well. For the sled tests, the duration of the experiment is of 300ms, time more than enough for the stabilising behaviour to begin [Siegmund, 1986].

Therefore, after the first hundred ms (at most) the controller should become active, changing the outcome of the dynamics of the system. This doesn’t annul the positive finding: an increasing CCR does appear to improve the similarities with the human tests. However, the validation performed is not enough to properly prove this. For the frequency response test, the problem is more subtle. The CCR was set at 20%, although there is no proof that that is a representative value. The results encourage the belief that a normal value cannot be far beyond that 20% (especially considering the sensitivity observed when CCR goes to 0%). However, a lower CCR with a higher gain may also fit the corridor properly. Moreover, for both the CCR behaviour and the stabilising control, the muscle grouping was also done following a very simple approximation for which there is no specific validation. Finally, the controller hasn’t been validated for higher frequencies. Although human beings are less capable of controlling higher frequencies, it is important to validate whatever control, or lack of it happens at higher frequencies.

Considering that several simplifications were made at the beginning of the design, it would be desirable to understand the impact of each simplification on the results obtained. It is however very difficult to ascertain, the effect of some of the simplifications: already the use of the MADYMO human body model carries a great deal of simplifications (as it is but a model). It is clear though that simplifying what is known as a very complex controller with a PID will generate an overall restriction

in the validity of the controller: only by manually changing the parameters can differences between individuals be accounted for. If, for example, a position is reached where a normal human being would experience pain, a change in the control strategy could occur that would be ignored in the current model.

None of these drawbacks make the result less positive. On the contrary, this means that the chosen design can indeed be made to behave in some tests in a biofidelic way. It means that although there is a lot of room for improvement, the path chosen points in the direction of a controller that can replicate normal human behaviour and be validated on many different scenarios. Moreover, a successful fine tuning of the controller can result in a biofidelic controller for a HBM, but may also help understand not only how a normal, healthy human being controls his neck, but it may also be of use when understanding of impairments is lacking.

However, it is clear now that a lot more work remains to be done with respect to validation. Optimal parameters need to be found for both the stabilising control and the CCR level, and in order to do this, more tests are needed. The future work that remains is mainly testing and tuning.

ACKNOWLEDGEMENTS

The help of Dr. Emily Keshner with validation data is gratefully acknowledged.
We would also like to specifically thank the European Commission (EC 6th Framework Programme for the support provided through the Marie Curie project MYMOSA (MRTN-CT-2006-035965) on motorcycle and motorists' safety.

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