Importance of Neck Muscle Tonus in Head Kinematics during Pedestrian Accidents

Victor S. Alvarez, Madelen Fahlstedt, Peter Halldin, Svein Kleiven

Abstract Unprotected pedestrians are an exposed group in the rural traffic and the most vulnerable human body region is the head which is the source of many fatal injuries. This study was performed to gain a better understanding of the influence that the neck muscle tonus has on head kinematics during pedestrian accidents. This was done using a detailed whole body FE model and a detailed FE vehicle model. To determine the influence of the muscle tonus a series of simulations were performed where the vehicle speed, pedestrian posture and muscle tonus were varied. Since the human reaction time for muscle activation is in the order of the collision time, the pedestrian was assumed to be prepared for the oncoming vehicle in order to augment the possible influence of muscle tonus. From the simulations performed, kinematic data such as head rotations, trajectory and velocities were extracted for the whole collision event, as well as velocity and accelerations at head impact. These results show that muscle tonus can influence the head rotation during a vehicle collision and therefore alter the head impact orientation. The level of influence on head rotation was in general lower than when altering the struck leg forward and backward, but in the same order of magnitude for some cases. The influence on head accelerations was higher due to muscle tonus than posture in all cases.

Keywords Pedestrian accident, muscle tonus, finite element method, head kinematics.

I. INTRODUCTION

Pedestrians are a vulnerable group in traffic and comprise a significant part of the total fatalities caused by traffic accidents. Between 9% and 38% of the road fatalities in the world are pedestrians in the report presented by IRTAD [1]. The majority of the accidents occur in urban areas [2] with relatively low vehicle speeds. Injuries sustained to the head region comprise a large portion of the most severe injuries [3].

These numbers are a concern to the vehicle industry which invests in understanding impact mechanisms in order to reduce the number of fatalities [4]. A vehicle design can be tested for pedestrian safety in several ways, such as through component level tests that give information about the interaction between vehicle and human body parts. Another method is full-scale testing, generally with pedestrian dummies, but also cadavers, that give more information about the entire interaction of the human body [5-6]. These tests can give more clues to understand the mechanisms of the whole body kinematics that will generate a certain impact and help in designing for enhanced pedestrian safety. Although it could be argued that cadavers are most accurate from a biofidelity aspect, even in the process of decomposition, there are no active components that might be an influencing parameter, especially at low speed impacts. Dummies are a more robust method of testing and though many are designed to include muscle tonus [4], there is no active muscle contraction possible.

Muscle tonus in the neck has been shown to be an important parameter influencing the head kinematics in certain impact conditions [7–9]. There is however a lack of understanding of its importance in pedestrian accidents [10] and, to the authors' knowledge, to date no studies have been performed that try to investigate this parameter. With the use of computer Finite Element (FE) models there is the possibility of describing the human body in more detail than a dummy, as well as applying muscle tonus.

The aims of this study were to evaluate the influence of muscle tonus on head kinematics in a pedestrian accident and to compare this with other parameters known to affect head kinematics.

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II. METHODS

The study was carried out using the FE-software LS-DYNA 971 Revision 5.1.1. A human body FE model was used in three separate simulation setups: one to measure the maximum forces generated by the neck muscles of the model, another to create an activation scheme and finally to simulate a collision with a detailed FE vehicle model.

Pedestrian FE Model

The FE model Total Human Model for Safety (THUMS, Toyota Central R&D Labs., Inc.) Version 1.4 [11] developed by Toyota has previously been validated at the component level and for full body kinematics [12-13]. The head and cervical spine of the THUMS was removed and replaced with the KTH head and neck model. The T1 vertebrae from both models were placed in the same position and tied with a rigid constraint. The KTH model has been validated at the component level [14] and against volunteer sled experiments on head kinematics using muscle activation [8]. The detailed head model has been validated against several relative motion experiments [15] and localized brain motion, intracerebral acceleration and intracranial pressure [16].

Muscle Model

The neck muscles are modeled as discrete springs with passive and active properties [8]. The active properties are modeled with Hill-type muscle contraction and the activation level is controlled with an activation-time curve going from 0 for no activation to 1 for 100% activation. The springs were remodeled slightly and attach to corresponding anatomical landmarks on the THUMS model. Each muscle group in the neck was given an appropriate number of springs in parallel to capture the anatomical insertion points [17] and to give realistic lines of action.

Muscle Force Validation

To validate the level of muscle forces generated by the model in the neck, simulations were done of experiments from the literature where volunteers pressed their head against a restraint in flexion, extension and lateral flexion [18–21]. The experiments had similar setups with the chest restrained and the Maximum Voluntary Contraction (MVC) measured as the force generated on a plate lying against the head.

Fig. 1. THUMS KTH model with head strap for MVC test.

The THUMS KTH model was set up in a similar simulation using a rigid head strap constrained from any motion (Fig. 1). The muscles were activated in three different ways, with as high activation levels as possible, so that the head was extended, flexed and flexed laterally if no strap was present. The MVC was then measured in the three directions as the contact force on the head strap.

Muscle Activation

To capture a theoretical large influence of the muscle tonus, the response to external stimuli such as loud noise or visible danger [22] was implemented. These stimuli create a startle response that tends to activate the muscle so that the head is moved forward and downward but maintained in an approximately horizontal position while the shoulders are raised [23-24]. It was hence assumed that the pedestrian would be aware of the oncoming vehicle. An activation scheme was manually developed that aimed at activating each muscle as much as possible and yielding a reasonable head posture (see Fig. 2). This muscle tonus was maintained during the entire collision event.

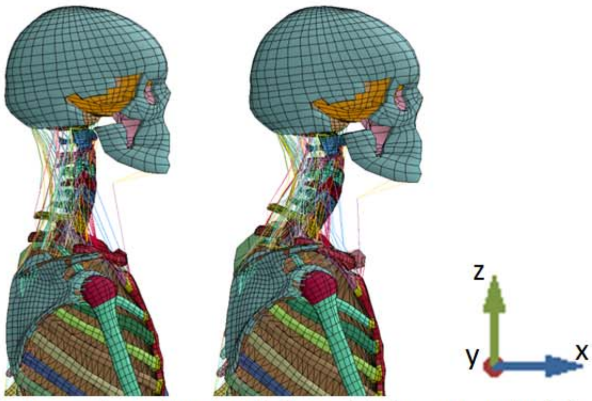

Fig. 2. THUMS KTH model before and after startle activation (left and right respectively).

The muscles were activated almost instantly in order to reduce calculation times but with a specific timing in order to create the desired posture. The activated muscles were divided into five groups that were activated separately (shown in TABLE I) and the timing and level of activation for each group is shown in Fig. 3.

TABLE I
MUSCLE GROUPS USED IN STARTLE ACTIVATION

Muscle group 1	*Muscle group 2*	*Muscle group 3*	*Muscle group 4*	*Muscle group 5*
Longus Colli Rectus Capitis Anticus Minor / Lateralis Scalene Anterior/Medius/Posterior Hyoid muscles	Longus Capitis Rectus Capitis Anticus Major Iliocostalis Cervicis	Sternocleidomastoid	Longissimus Capitis Longissimus Cervis Splenius Cervicis	Trapezius

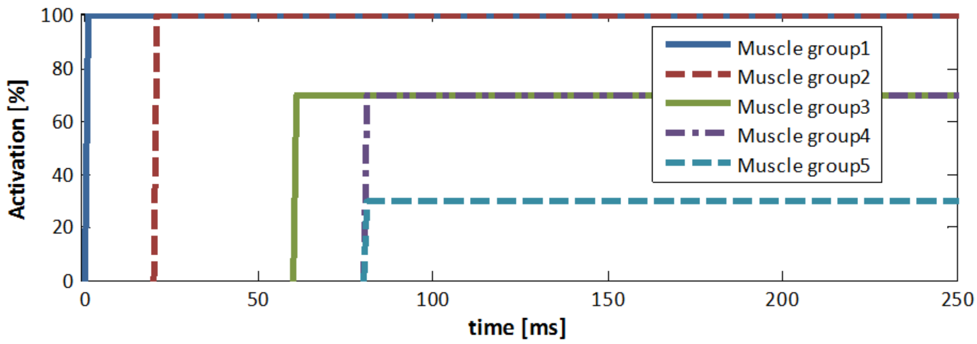

Fig. 3. Level of activation for the muscle groups used in startle activation scheme.

Vehicle Model

The collision simulations were performed using a detailed FE model of the front of a large passenger car containing all main structural parts including engine block and suspension. The model was compared with a NCAP upper leg impactor test [25] using a validated FE model of the same impactor [26], see Fig. 4.

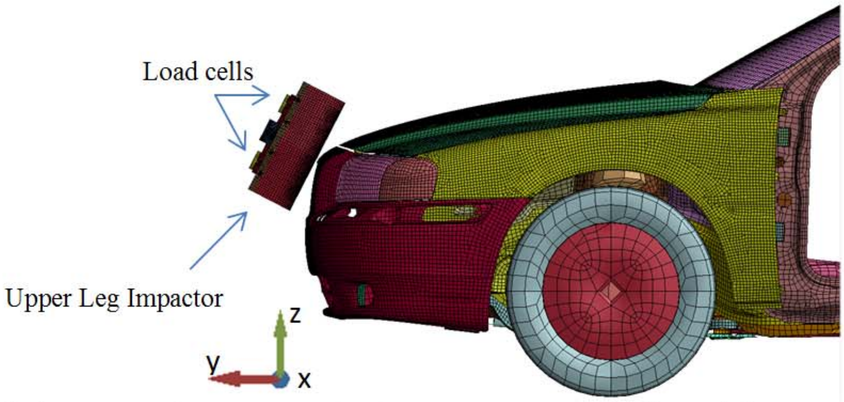

Fig. 4. Simulation setup of NCAP upper leg impactor test on vehicle model.

Pedestrian Setup

To be able to compare the influence of the neck muscle tonus to other parameters, the pedestrian model was set up in several different collision simulations where parameters that have been known to influence head kinematics were changed. An important parameter is the vehicle speed [27-28] as all kinematic parameters get a direct influence, but it also tends to inhibit or enhance other parameters' influence on the kinematics. In the GIDAS database [3], 70% of the pedestrian accidents were at a velocity of 40 km/h or less which is why 40 and 20 km/h were chosen.

Another parameter influencing the entire body rotation and hence also the head rotation is the pedestrian gait, previously found in rigid body simulations [28–32]. An especially large difference can be seen when comparing the Struck Leg Backward (SLB) and Struck Leg Forward (SLF) postures (see Fig. 5). This is because the entire body tends to rotate forward or backward respectively [29], which is why these two postures were chosen.

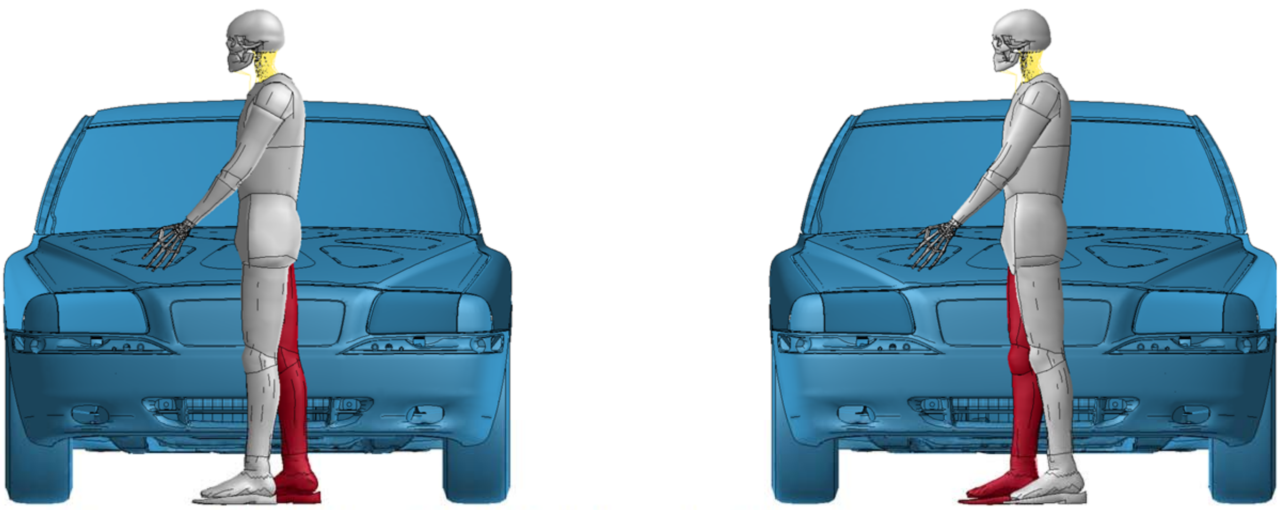

Fig. 5. Postures of THUMS KTH model with SLB (left image) and SLF (right image).

A simulation matrix including eight simulations with these parameters, with and without muscle tonus (passive vs. startled) was set up and compared with respect to head kinematics (see TABLE II). In the simulations using muscle tonus the vehicle was set to have the first impact with the pedestrian at 120 ms giving the model time to retract the head. The position of the nodes at 120 ms where saved and used in the simulations without tonus, so the model had the same position as the startled model at time of impact for easier comparison. In all simulations the coefficient of friction between pedestrian and vehicle was set to 0.3.

TABLE II

SIMULATION SETUP MATRIX

Posture	*40 km/h*		*20 km/h*	
SLB	Passive	Startled	Passive	Startled
SLF	Passive	Startled	Passive	Startled

III. RESULTS

Muscle Force Validation

The data extracted from the different studies on MVC are all for male adults and these are compared with the results from the MVC simulations. The results are given as the contact force on the head strap in the direction of the motion, as well as the equivalent moment about the T1 vertebra shown in TABLE III. The results show that the model is within the range of the experimental data.

TABLE III
MVC FROM LITERATURE, MEAN (SD), AND THE THUMS KTH MODEL SIMULATIONS

Reference	*# Subjects*	*Extension*	*Flexion*	*Lateral flexion (Left/right)*
[18] [Nm]	60	52 (11.4)	29 (5.1)	-
[19] [N]	17	96.17 (48.38)	90.61 (35.86)	63.66 (25.79) / 68.91 (30.20)
[20] [N]	29	278 (50)	151 (47)	-
[21] [N]	21	100 (28	72 (18)	76 (26) / 76 (23)
THUMS KTH				
Force [N]		200	68	84
Moment [Nm]		40	13.6	16.8

Vehicle Model

Fig. 6 shows the result from the NCAP upper leg impactor tests. The result from simulation is presented as the sum of the forces in the y-direction in the coordinate system in Fig. 4. The forces are extracted from the upper and lower load cells in the FE simulation and the experiment.

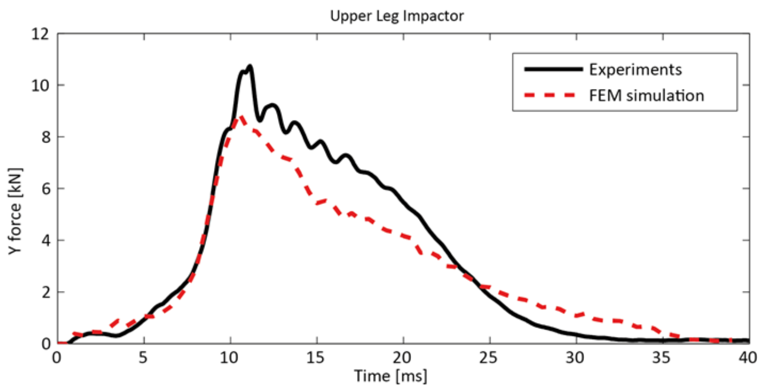

Fig. 6. Results from the upper leg impactors FE-simulations compared to the experiments.

Startle Activation

In Fig. 7, the displacement of the head CG relative initial position in the coordinate system in Fig. 2 as a result of the startled muscle activation is plotted. It can be seen that the head starts to stabilize at 120 ms.

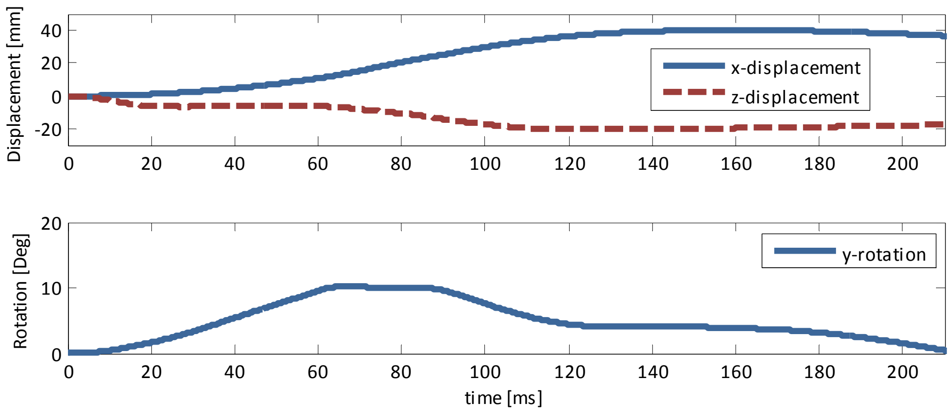

Fig. 7. Global head CG displacement during startle activation simulation.

Pedestrian Collisions

The 2D trajectories of the head CG, T1 and T8 vertebrae and pelvis were plotted as the displacement relative to the vehicle. There were, however, very small differences between the simulations with same vehicle speeds. Only between SLB and SLF postures could any larger differences be seen, but only in the pelvis (see Appendix).

In Fig. 8 - Fig. 10 the head CG local x-, y- and z-rotations are plotted, integrated from angular velocities. In Fig. 8 the simulations at 40 and 20 km/h without muscle tonus are plotted and compare the SLB and SLF postures. For 40 km/h simulations the x-rotations start to deviate at 40 ms and differ with approximately 4 degrees fairly constantly between 60 and 100 ms and deviate more towards the end of the simulation. The y-rotations start to deviate 1-2 degrees and cross each other three times and at 96 ms they separate towards the end. The z-rotations deviate at 50 ms and the difference increases during the whole simulation to a maximum at the end.

In the 20 km/h simulations the x-rotations start to deviate at 50 ms and have a very similar behavior as in 40 km/h, increasing to approximately 7 degrees, and more towards the end of the simulation. The y-rotations start to deviate only slightly at 40 ms but more after 100 ms, growing towards the end. The z-rotations deviate at 60 ms and the curves appear almost inverted with the SLF posture being entirely positive and the SLB posture entirely negative. The curves move apart during the entire simulation.

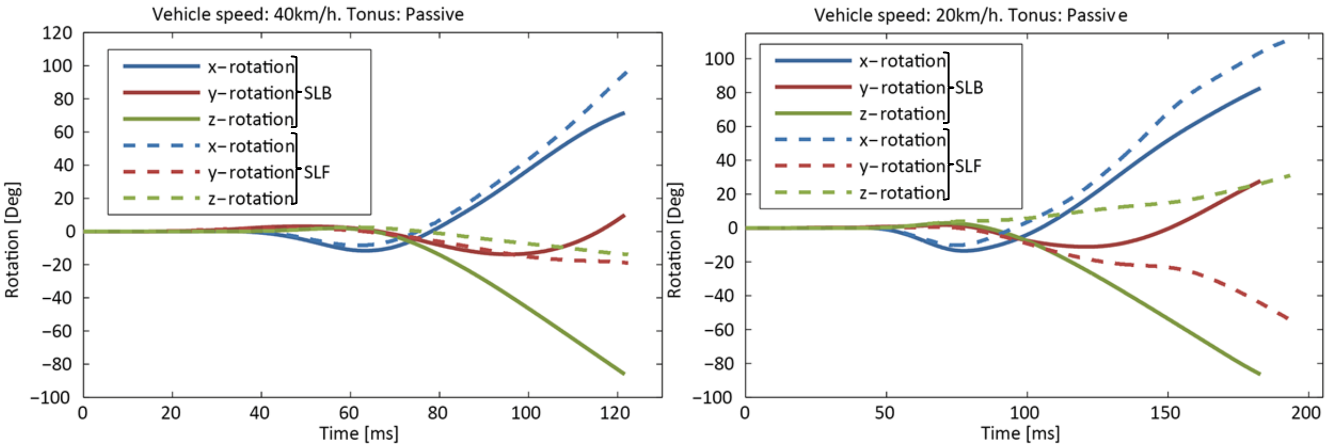

Fig. 8. Head CG rotations for 40 km/h (left) and 20 km/h (right) vehicle speed with passive muscles, comparing SLB and SLF.

Fig. 9 shows the head rotations for the simulations with SLB and SLF postures at 40 km/h, comparing passive and startle. It can be seen that in the SLB posture the x-rotations only deviate slightly for the majority of the simulation (max 1 degree) but they deviate more notably towards the end. The y-rotations start to deviate at 30 ms up to 4 degrees and the curves then cross at 90 ms, after which they separate more but follow the same shape. The z-rotations deviate at 55 ms, slowly growing continuously towards the end but following the same shape.

In the SLF posture the x-rotations deviate notably at 60 ms and the curves separate slightly but coincide again at the end. The y-rotations start to deviate at 30 ms up to 5 degrees and the curves cross at 95 ms, after which they continue to deviate for the rest of the simulation. The z-rotations deviate slightly from 30 ms and curves cross at 58 ms after which the startle continues to grow positive and the passive grows negative for the remainder of the simulations.

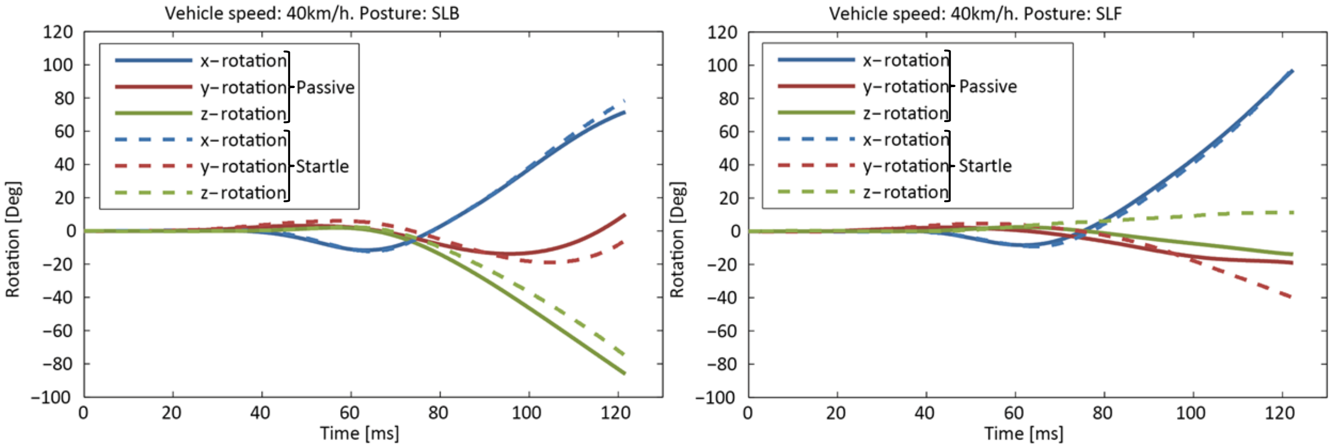

Fig. 9. Head CG rotations for SLB posture (left) and SLF posture (right) with 40 km/h vehicle speed, comparing SLB and SLF.

In Fig. 10 the head rotations for the simulations at 20 km/h with SLB and SLF postures are plotted, comparing passive and startle muscle. In the SLB posture the x-rotations start to deviate to a noticeable extent after 80 ms and slowly move apart during the whole simulation, except just before head impact. The y-rotations deviate at 65 ms to about 1 degree and the curves cross at 105 ms after which they separate rapidly but follow the same shape. The z-rotations deviate from 35 ms up to 10 degrees but finally the curves cross just 5 ms before head impact.

In the SLF posture the x-rotations are almost equal during the majority of the simulation, only deviating maximum 1 degree, but at 156 ms the curves suddenly move apart and the difference increases for the rest of the simulation. The y-rotations start to deviate mostly after 108 ms and the difference increases fast and stabilizes after about 50 ms and for the rest of the simulation. The z-rotations start to deviate at 50 ms and the difference increases up to 19 degrees, but the curves cross 10 ms before head impact.

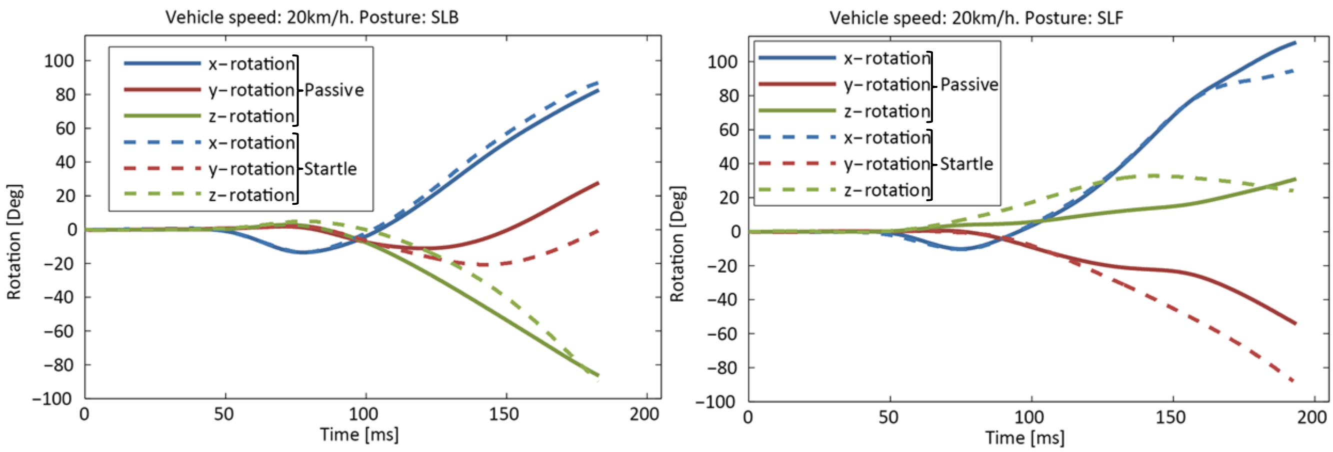

Fig. 10. Head CG rotations in global coordinates for 20 km/h vehicle speed comparing passive and startle.

Fig. 11 shows the time history of the accelerations in the head CG for the simulation in 20 km/h, SLB posture and passive muscles. The largest peaks from all plots correlate with head impact. In the beginning there are some smaller peaks in the angular acceleration, but not in the linear acceleration. In Fig. 12 - Fig. 13 the peak value of the accelerations from all simulations are plotted. The linear acceleration in Fig. 12 is lower for passive than startle in the SLB posture for both velocities. In the SLF posture the acceleration is lower with startle than passive. The angular accelerations are lower for all cases with passive compared to startle. Between postures the resultant angular accelerations are very similar when using passive muscles. Between the SLB and SLF postures a larger change in resultant angular acceleration can only be seen when both have the startle muscle tonus.

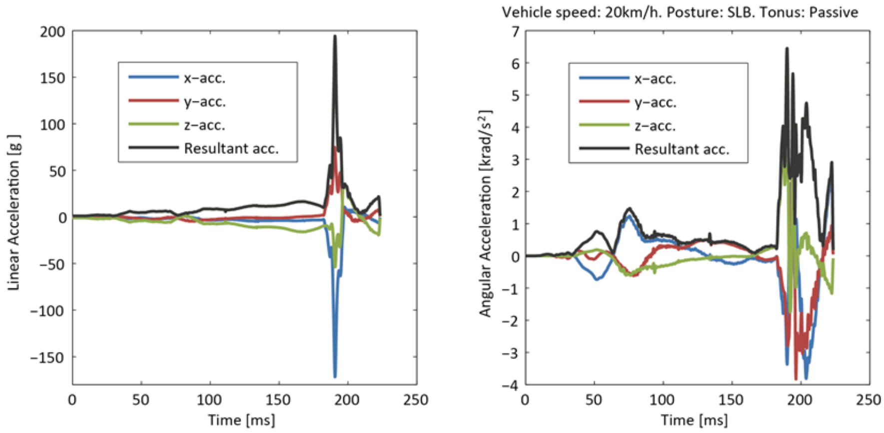

Fig. 11. Acceleration time histories for simulation in 20 km/h, SLB posture and passive muscles.

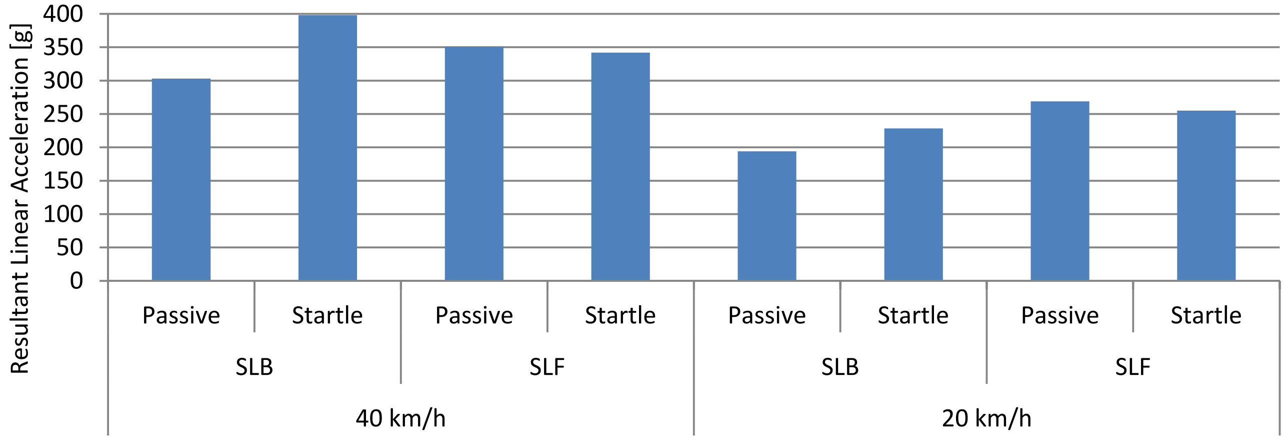

Fig. 12. Peak resultant linear accelerations in head CG from all simulations.

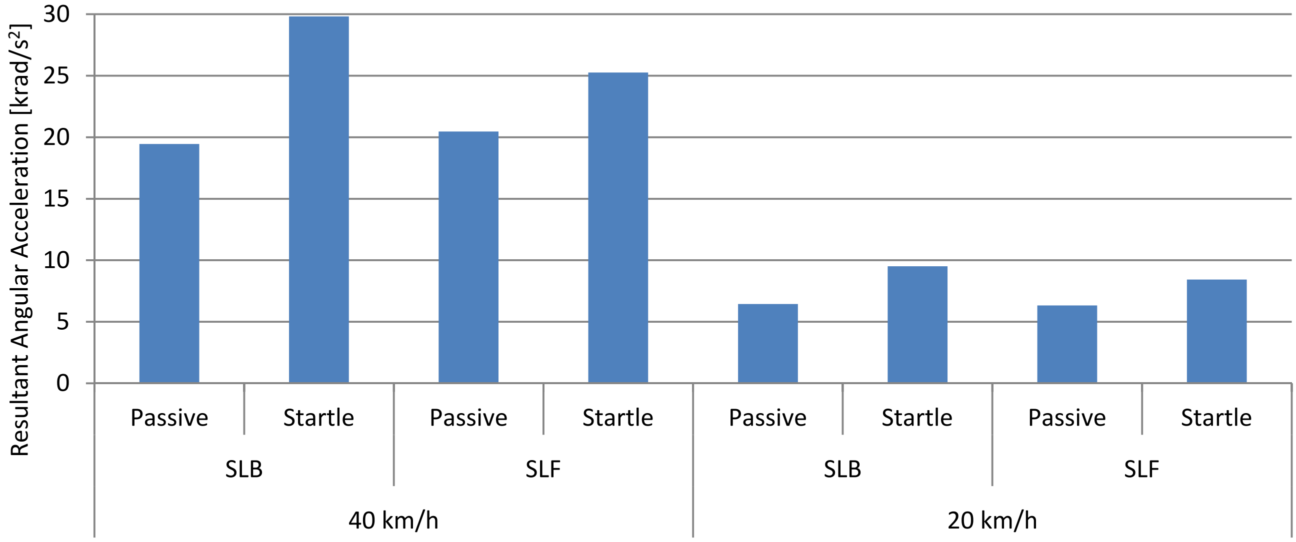

Fig. 13. Peak resultant angular accelerations in head CG from all simulations.

The Head Injury Criterion (HIC) for all simulations is presented in TABLE IV. At 40 km/h HIC is increased by 1721 in the SLB posture as a result of the startle muscle tonus. In the SLF posture the HIC value is instead reduced by 161. The two postures have a difference in HIC value of 1203 with passive muscles and 679 with startle muscle tonus. At 20 km/h the HIC value is increased by 35 in the SLB posture as a result of the startle muscle tonus. In the SLF posture the HIC value is instead reduced by 151. The two postures have a difference in HIC value of 247 with passive muscles and 61 with startle muscle tonus.

The position of the head and cervical spine at time of impact for the 40 km/h simulations is shown in Fig. 14

and for the 20 km/h simulations in Fig. 15. It can be seen that the impact orientation changes most between the two postures, going from frontal to parietal head impact. But the difference is clearly noticeable also between passive and startle, especially in the SLF posture at 40 km/h.

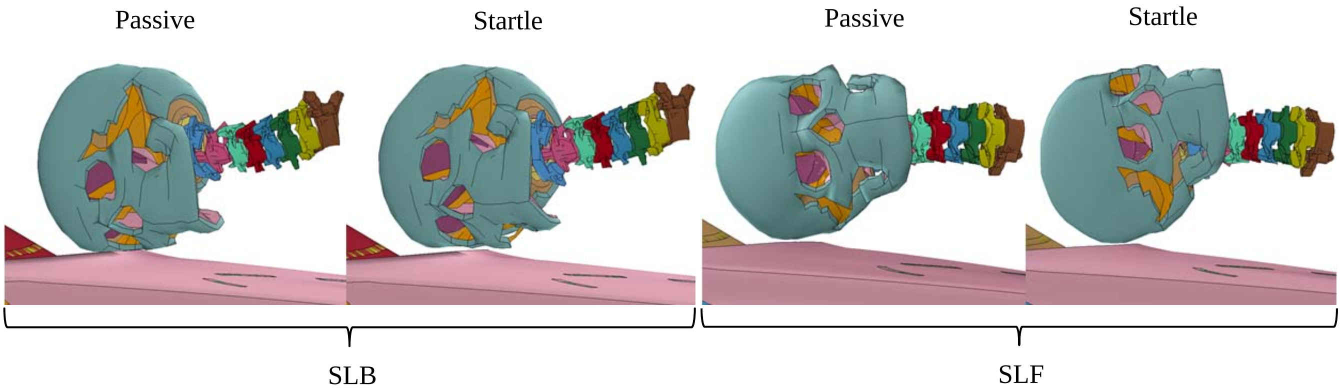

Fig. 14. Impact orientation of the head at time of impact for all simulations at 40km/h.

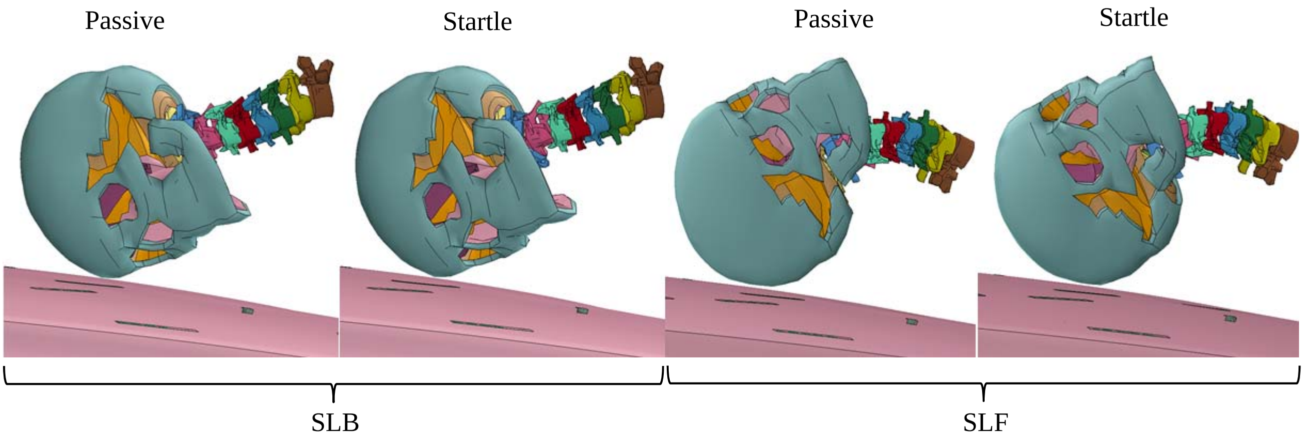

Fig. 15. Impact orientation of the head at time of impact for all simulations at 20km/h.

In TABLE IV the time between first pedestrian-to-vehicle contact and head impact (impact time) is shown for all simulations. Listed are also the resultant impact velocity and head rotation at head impact. The impact times are only changed notably for the two postures in the 20 km/h simulations (in the order of 10 ms) while for the change in muscle tonus it is 1 ms or less for all simulations. The same tendency is seen for the impact velocities where the largest change is between the postures in 20 km/h with passive muscles (1.2 m/s).

In the 40 km/h simulations the muscle tonus has a relatively small influence on the x-rotation compared to the posture. The influence on the y-rotation is in the same order of magnitude when changing tonus and posture. When the two posture simulations have startle muscle tonus the difference becomes larger. The difference in z-rotation is notably larger for the change in posture than muscle tonus, but the magnitude is still in the range of some of the largest rotational changes.

In the 20 km/h simulations the postures give the largest differences in all directions. However the change in y-rotation due to muscle tonus is relatively large. The largest change caused by muscle tonus is in the SLF posture (33.6 degrees) compared to the SLB posture (28.4 degrees). The muscle tonus changes the influence of the posture to some extent. The difference in impact times and impact velocities are slightly reduced as well as the x- and z-rotations, 20.9 and 3.9 degrees respectively, but the y-rotation is increased with 5.2 degrees.

TABLE IV
COMPARISON OF IMPACT TIMES, IMPACT VELOCITY, IMPACT ANGLES AND HIC FOR ALL SIMULATIONS

Velocity [km/h]	*Posture*	*Muscles*	*Impact time [ms]*	*Impact velocity [m/s]*	*X-rot [Deg]*	*Y-rot [Deg]*	*Z-rot [Deg]*	*HIC*
40	SLB	Passive	121.6	13.9	71.6	10.0	-86.1	3428
		Startle	121.4	14.1	78.5	-6.2	-74.8	5149
	SLF	Passive	122.4	14.3	97.0	-19.0	-13.9	4631
		Startle	122.6	14.2	98.5	-40.2	11.4	4470
20	SLB	Passive	182.8	8.7	82.6	27.8	-86.5	819.2
		Startle	182.6	8.2	86.9	-0.61	-89.7	854.5
	SLF	Passive	193.4	7.5	111.4	-54.3	31.1	1066
		Startle	192.4	7.4	94.8	-87.9	24.0	915.1

IV. DISCUSSION

This study investigates the effect of the muscle tonus on head kinematics in pedestrian to car collisions. The study is based on eight simulations comparing the head kinematics for accidents in 20 and 40km/h. The head kinematics was measured by comparing the head-to-car impact velocities, the head trajectories (translational motion in the x and z-directions) and the head rotation around the x, y and z-axis. In order to benchmark the effect of the cervical muscle forces on the head kinematics with different leg postures were also conducted.

The models used were validated for impact situations that are relevant in this study but not the exact impact situation simulated due to lack of input data. The overall validation of each part in the simulations should however give a platform for realistic results. The curves from the upper leg impactor tests correlate well, especially in the initial slope, and the peak is slightly lower in the FE simulation but lower values were also seen in the calibration tests by [26]. The most uncertain parameter in the simulation setup is the muscle activation scheme. The muscle activation used in this study was an activation scheme that could simulate a person who is aware of an impacting car compared to a person who is not aware of the impact.

Both for the simulations in 40 km/h and 20 km/h the largest change in total impact orientation was seen when comparing the SLB and SLF postures using passive muscles. These postures resulted in different whole body kinematic behavior in the beginning of the impact giving the pedestrian a rotation forward or backward, which is in agreement with previous studies [28–32].

The largest influence of muscle activation on head impact angles was seen in the SLF posture, indicating that the posture is an important parameter not only to general kinematic variables but also the possibility for muscle tonus to have a measurable influence. Conversely, it could be said that the muscle tonus changes the level of influence the posture has on the head rotations.

In the simulation at 40 km/h, only in the y-rotation the influence of the muscle tonus was of the same order of magnitude as the change in posture, giving a relatively small total change in head impact orientation in SLB posture. In the SLF posture the change in z-rotation was smaller than for a postural change but relatively large compared to other directions, giving a very notable change in impact orientation seen in Fig. 14.

The change in z-rotation in the SLF posture due to muscle tonus is, surprisingly, significantly larger in 40 km/h than 20 km/h. This could be due to the fact that the curves with and without tonus do not have the same shape. The curves separate to a larger extent in the 20 km/h than in the 40 km/h simulation, but in the 20 km/h simulation the curves cross each other very close to the time of head impact (Fig. 10) whereas in the 40 km/h simulation they move apart in the whole simulation (Fig. 9). This change in timing indicates that the muscle tonus alters the stiffness of the neck and therefore the rotation history of the head. The timing is also affected by other variables, such as vehicle geometry, speed, pedestrian height etc. and could therefore also affect to what extent the muscle activation influences the head impact orientation.

However, the posture did not have the largest influence on head accelerations. The largest change in accelerations was seen for the passive vs. startle in the SLB posture at 40 km/h. The HIC value was affected more by the muscle tonus in the SLB posture but less in SLF posture at 40 km/h and reversed at 20 km/h. The HIC was higher with the startle tonus than passive muscles in all but one case. These findings indicate that the change in impact orientation is not the only factor affecting the accelerations in the head. It could be suggested that it is also influenced by the stiffened neck. Even though it could be argued that many parameters are changing between the simulations, such as slight change in impact point on the vehicle, making it hard to say to what extent the tonus is influencing the angular accelerations. It was however expected that the posture would have a larger influence in the 40 km/h simulations because the head impacted at the base of the windscreen and the posture gave a more noticeable change in impact on the head and linear acceleration.

It has previously been found that the influence of impact direction has a substantial effect on the intracranial response [16, 33]. In a study by Hodgson [34], impacts on the side produced longer periods of unconsciousness than at any of the other locations. In the study by Gennarelli et al. [35] traumatic coma in monkeys was produced by accelerating the head without impact. It was found that the majority of the animals that were enduring a coronal motion suffered coma lasting longer than 6 hours, while all animals that were accelerated in the sagittal plane had coma lasting less than 2 hours. Therefore it could be of interest to further investigate how large an influence the impact orientation has on the strain in the brain and see if this could be related to muscle tonus.

An assumption made in this study was that the pedestrian would have enough time to react to the oncoming vehicle and tense the muscles in an instinctive way giving the startled posture described above. This was considered realistic if considering reported reaction time of approximately 70 ms [36–38] and the time from activation onset to full activation of approximately 60 ms [39]. This would mean that for a vehicle going at 40 km/h, the pedestrian would only have to be 1.4 m away from the vehicle in order to reach full activation.

Another assumption was that the muscles would activate as much as possible with the condition to retract the head and raise the shoulders. This activation was found iteratively, as well as the division of muscle groups, until a combination of activations was found that gave a suitable contraction seen in Fig. 2. It is therefore possible that higher levels of activation could be obtained with some more advanced optimization routine, and that it is higher in the real human neck. It is however not clear to what level the muscles would actually activate in such a reaction, but since a MVC in extension flexion and lateral flexion falls within the range of experiments (see TABLE III), it is assumed that the forces produced in the neck muscles during the impact are realistic. It could however be noticed that the strength in flexion is in the weaker range. It could therefore be argued that the level of activation is an underestimate since the activation levels might exceed the MVC due to the added movement of the head during the collision [40]. It is however hard to determine how the human body would react to such a fast and catastrophic event, even if there would be enough time for muscle activation theoretically. But it could be of interest to investigate if an augmented dynamic activation during collision could alter the kinematics further.

This study is limited in some aspects and it is difficult to grade the significance of the changes in the computed parameters. Whether a certain change in degrees about some axis is significant or not is hard to determine. There is no clear answer to this and it is probably not that simple as we need to understand the complete impact situation. What is known is that a change in rotation will change the impact location on the head. We also know that the impact location together with the impact direction and speed gives the total initial impact situation for the head and will result in some strain in the brain. So, the sensitivity for the head in different impact speeds and impact directions and locations needs to be investigated further in order to define what is significant and what is of minor importance. But what can be said is that when compared to another parameter known to influence the impact situation, the influence is in some cases in the same order of magnitude. So if this parameter needs to be accounted for, maybe so does the neck muscle tonus.

V. CONCLUSIONS

The objective of this study was to evaluate the importance of neck muscle tonus during pedestrian accidents. The simulated muscle tonus resulted in minor influence on the impact time and impact velocity. It was however concluded that the muscle tonus shows an influence on the head impact location and direction. The influence is in general larger for a lower speed of 20 km/h than at 40 km/h, but not in all cases and directions. The comparison to the Struck Leg position in this study showed that this parameter had substantially larger influence in the z-rotation but not always in the other directions. The influence on angular accelerations of the head was larger due to muscle tonus than posture in all cases.

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VII. References

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VIII. APPENDIX

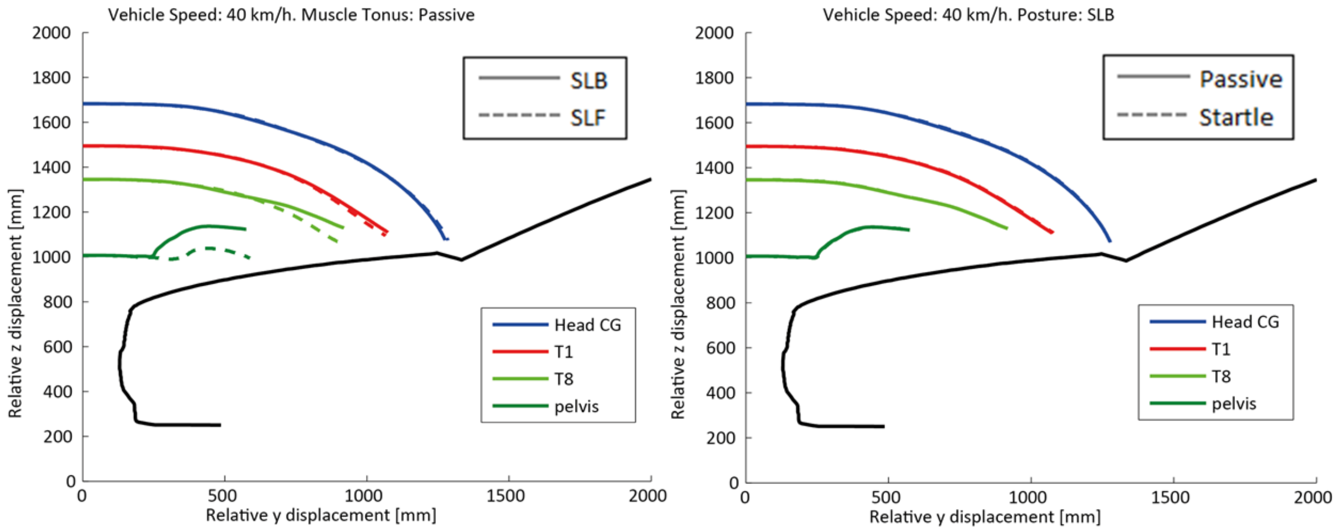

Fig. A1. Trajectories for vehicle speed 40 km/h comparing SLF and SLB (left figure) and passive vs. startle muscle tonus (right figure).

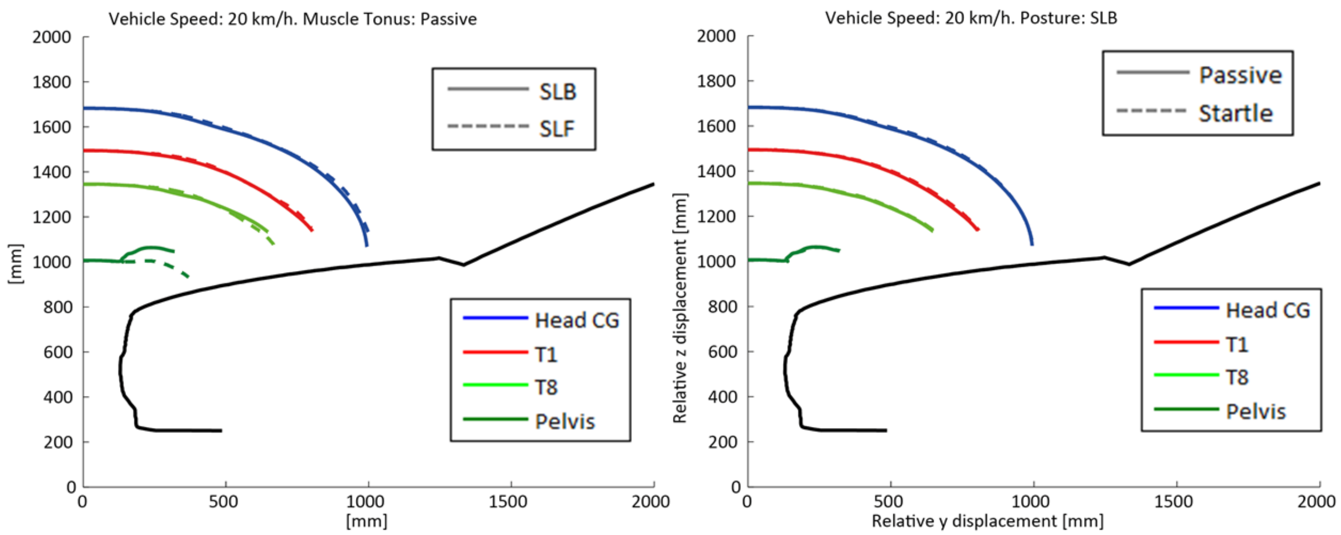

Fig. A2. Trajectories for vehicle speed 20 km/h comparing SLF and SLB (left figure) and passive vs. startle muscle tonus (right figure).