Muscle Activation Onset Latencies and Amplitudes during Lane Change in a Full Vehicle Test

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Abstract Vehicle occupant kinematics in the pre-crash phase is strongly influenced by active muscle contribution. In order to determine the human response to low-load situations prevalent in the pre-crash phase 21 healthy male volunteers (co-driver, age: 33.4±8.8 y, mass: 78.5±6.3 kg, height: 179.2±4.6 cm, sitting height: 91.0±2.0 cm) were subjected to a series of lane change maneuvers with peak accelerations of around 1 g in lateral direction. Three awareness conditions were realized: an unaware, an anticipated and an informed condition.

Surface Electromyograms (EMG) were recorded bilaterally on three neck muscles (m. sternocleidomastoideus, m. trapezius p. cervicalis and p. descendens) and four trunk muscles (m. rectus abdominis, m. obliquus externus abdominis, m. latissimus dorsi, erector spinae (lumbar region)) and are supplemented with occupant and vehicle kinematics data.

In the subsequent EMG analysis mean onset latencies between 0.106±0.020 s and 0.171±0.047 s were found, where neck extensors responded with the shortest latency. For selected conditions and muscles significantly shorter onset latency was recorded on the right side.

Significantly higher activations of the right muscles were observed in the first part of the maneuver, while only a limited number of muscles showed an effect of the awareness condition.

Keywords EMG, integrated safety, lane change, onset latency, occupant kinematics

I. INTRODUCTION

One of the declared goals of the EU's policies for making roads safer is to reduce traffic fatalities by a half in the decade between 2010 and 2020 [1]. Passive safety systems, i.e. systems designed to mitigate the severity of a crash for the occupant, have been present in vehicles over the last several decades and have helped to reduce the number of casualties and the medical and socio-economic consequences of accidents drastically. These systems, however, have been refined to a point where further improvements are hard to realize.

A promising and economically viable path is to equip vehicles with active safety systems that are capable of influencing the vehicle's kinematics before impact. Thereby crash severity can be mitigated or collisions can even be avoided. In 2014 EuroNCAP will include for the first time the performance of active safety systems, namely the autonomous emergency braking, in their ratings [2]. In order to assess the interaction between the vehicle and the occupant and to detect possible out-of-position events there is a need for numerical human body models that are not only capable of describing the crash phase correctly, but also of simulating the pre-crash phase, where active and reactive contributions from muscles play a role.

Recently increasing effort has been put into generating data about the occupant's behavior in conditions where only mild loading occurs. Although anthropomorphic testing devices (ATD; dummy) allow estimating the occupant's kinematic response and also injury risk in certain loading conditions, the kinematic response during low load situations is significantly different between occupant and dummy [3]. Human response to the very

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specific loading conditions occurring in driving maneuvers is typically studied in either a sled test setting or more realistic driving situations.

In [4] a frontal braking maneuver was simulated using a cable-guided sled on an inclined plane that was decelerated by a damper. In this study five subjects (two female, three male) took part and the kinematic response to the braking maneuver was recorded using an optical motion tracking system combined with Electromyography (EMG). Two muscular conditions (relaxed and tense) were tested in three different load levels (0.2 g, 0.6 g, 1.0 g). In subsequent studies the maneuvers were simulated by a cable-guided sled on a plane, where the sled was accelerated from a static position. Both frontal and lateral load cases were tested [5-6]. In all three studies the strong influence of muscular conditions on occupant kinematics was underlined.

A series of sled tests with 11 male subjects was performed in [7], where both a frontal and a lateral maneuver was simulated using a cable-guided sled. Each maneuver was performed three times and EMG as well as occupant kinematics were recorded. Large inter-subject variability in the movements was observed for both maneuvers, but large intra-subject variability was mainly detected in the frontal maneuver.

In [8-11] five volunteers were subjected to peak loads of 2.5 and 5 g on a sled that was propelled backwards by a hydraulic cylinder. Two muscular conditions were also tested, a braced and a relaxed one. Additionally a comparison to an ATD as well as post mortem human surrogates (PMHS) was performed. Significant differences in the biomechanical responses were observed for human volunteers, PMHSs and ATDs. It was concluded that the lack of muscle activity in PMHSs and the reduced muscle activity in the relaxed subjects contributed to a significantly larger forward excursion in both groups compared to braced volunteers.

A different approach was taken in [12] where both drivers and co-drivers were subjected to different levels (0.3 g, 0.4 g, 0.5 g) of autonomous braking within a vehicle-based test framework. Kinematics were recorded using video cameras and optical targets. Four volunteer groups were tested ranging from 5th%ile female to 95th%ile male occupants. Several factors, including size, gender and position (driver, co-driver), were identified as main influences on the forward movement.

The present study is focused on specific muscle activation properties during lane change maneuver within a vehicle-based test framework on a test track and explicitly takes into account different awareness levels in the analysis. Twenty one male occupants in the co-driver position were exposed to a single lane change maneuver in a vehicle. The maneuver was performed three times at different awareness levels: unaware, anticipated, informed. Kinematics of vehicle and occupants were recorded as well as the EMG activation of seven trunk muscles bilaterally. The purpose of this study was to analyze the occupant's response on a muscular as well as a kinematic level during real-world driving maneuvers, as the importance of muscle activity was clearly demonstrated in previous studies. The focus of this paper is the measurement of muscle activity onset latency and amplitudes for the swerving maneuver and the identification of factors (e.g. alertness) that influence these quantities. The resulting information about muscle activation patterns can be utilized in the development of active human body models, where onset latencies and the order of activation of different muscle groups are important parameters.

II. METHODS

The vehicle tests were performed on 21 male subjects in a Mercedes-Benz S-500 (type W221; width: 1.78 m, length: 5.23 m, wheelbase: 3.17 m). Each subject gave written informed consent and the tests were reviewed and approved by the local ethics committee and the medical officer at the test track.

The subjects were selected such that they resembled the 50th%ile male occupant (mass: 77.8±6.7 kg, height: 179.4±4.4 cm, age: 32.8±9.4 y, sitting height: 91.0±2.0 cm; 50th%ile according to [13]: height: 175 cm, weight 77.1 kg) and all subjects held a valid driving license for operation of a passenger car.

On a closed test track a lane change situation at 50 km/h was simulated[1]. In the maneuver, which was performed to closely resemble the first part of the VDA lane change [14], the driver turned the steering wheel to the right around 200°, followed by a counter-movement to the left approximately 360° and a rotation to the right to reach the neutral position again. The driver operated at maximum effort and in a reproducible way.

Three awareness conditions were realized: an unaware (no information given to the subject), an anticipated (maneuver could be inferred from contextual information) and an informed condition (maneuver was

[1] Additionally an emergency braking maneuver at 12 km/h was simulated, which is not presented in this study. The maneuvers were carried out in alternating fashion.

announced with a count-down from three). The maneuvers were performed in fixed order and in arbitrary locations on the test track. In an informal questioning all subjects commented that the first maneuver surprised them.

The standard seat was replaced with a modified seat where the original seat frame was used but the cushions were replaced by wooden plates, covered in artificial leather. The seat surface (width: 440 mm, length: 490 mm) was inclined by 10° and the distance between the rear edge and footwell level was 180 mm. The angle between back rest (width: 550 mm, length: 512 mm) and seat surface was 104° and the original head rest was used. Additionally a lap belt only was used, where the original lower mounting points were utilized. The upper part of the belt was routed behind the seat and a clamp was placed close to the belt buckle to fix the length of the lap belt. The clamp was adjusted for each subject individually such that the belt was tight and that pelvis movement was restricted. These measures were taken to facilitate the transfer of testing data to numerical simulation models and to simplify modeling the interaction between seat, belt and the occupant.

In order to identify the vehicle driving state velocity, acceleration in lateral and frontal direction, steering wheel angle and angular velocity as well as the yaw rate were recorded from the CAN-Bus using a Dewetron Dewe5000. The steering wheel angle and angular velocity were sampled at 100 Hz, the velocity at 10 Hz, while the remaining channels were sampled at 50 Hz.

All data acquisition systems were synchronized using an event trigger. On activation the vehicle kinematics and the EMG system received a voltage signal, while the occupant kinematics system recorded an infrared light emitting diode signal in one of the cameras. Time instance t=0 is defined using a threshold method. The first time instance where the steering wheel angle exceeds 20° was identified. The steering wheel angular velocity at this point was then used to extrapolate to steering wheel angle 0° and the corresponding time instance was defined as t=0.

Both EMG and subject kinematics were recorded during the tests but the focus of this paper is on the former. Surface EMG electrodes (Ag/AgCl surface disc electrodes, 9 mm in diameter) were placed in a bipolar manner bilaterally on three neck muscles (m. sternocleidomastoideus/SCM, m. trapezius p. cervicalis/Neck ext. and p. descendens/Upper trap.) and four trunk muscles (m. rectus abdominis/Rect. Abd., m. obliquus externus abdominis/Ext. obliq, m. latissimus dorsi/Lat. dorsi, erector spinae (lumbar region)/ES). The skin in the areas of electrode application was shaved and prepared with an abrasive gel. Both application of the electrodes and verification were done according to [15] and [16] and all electrode pairs showed impedance values of less than 10 kΩ.

Cables were routed underneath a tight motion capturing suit (see Fig. 1), jointly emitted between pants and shirt and connected to the recording unit placed between the driver and co-driver seat. Neither tension nor bending of cables was observed during the tests. The EMG Signals were recorded using a Noraxon TeleMyo G2 wireless device with a sample rate of 1500 Hz.

(a) (b) (c)

Fig. 1. Setup of vehicle and subjects. The vehicle was equipped with eight Vicon infrared motion

capturing cameras. Note that the windshield and the passenger window are removed (a). Subjects wore a skin tight motion capturing suit and reflective markers are attached in the suit. EMG cables are collected and connected to a Noraxon recording unit (b) and (c).

In order to reduce artifacts from Electrocardiograms (ECG) in the signals a methodology based on [17] was used. In a baseline signal, where no significant muscle activity was recorded, one instance of an ECG artifact was identified. Using a pattern matching method several instances of ECG artifacts were recognized and an average contribution to each EMG signal channel was determined. For the EMG signal during activity the pattern matching algorithm utilized this baseline artifact to find instances of the heart beat concurrently in each channel and, at a recognition rate of 50%, a section was marked as potentially contaminated. In regions of very large muscle activity the performance of the pattern matching algorithm decreased, which led to a large number of false positives. Therefore, a selection algorithm was applied that computed the likelihood of various ECG-peak combinations based on the heart rate of the subject and returned the combination that was most probable. A complementary visual inspection of the patterns as well as an analysis of the wavelet-transform of the signal was performed as a cross check.

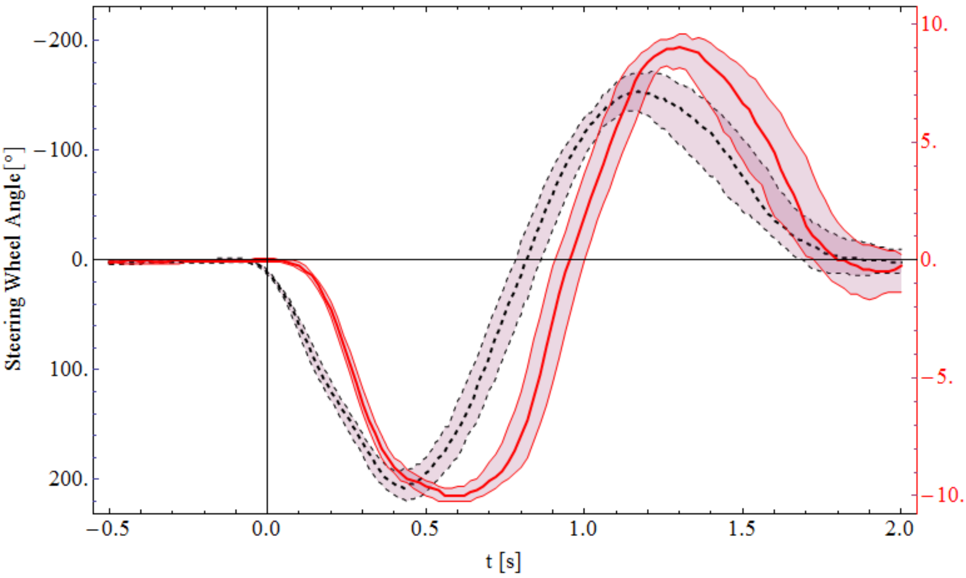
The resulting EMG signals were full-wave rectified and band pass filtered (30–400 Hz). The onset latency was determined separately for each trial relative to the time instance t=0. A 200 ms period before time 0 was taken as a baseline reference. Two onset times were defined: t1 was identified as the point at which 50 consecutive EMG samples (corresponding to approximately 30 ms) had exceeded 3 standard deviations from the mean baseline reference amplitude [18-19]. In addition each EMG onset was visually checked [20].

In the subsequent statistical analysis missing values (1.4% of the full set) were replaced with the mean value of the corresponding data set. Outliers were determined as values exceeding ±2.5 SD. Normality of the data sets was first tested using Kolmogorov-Smirnov tests. To evaluate the onset latency differences, three-factorial analysis of variance (ANOVA) with within-subject factors [*case* (unaware, anticipated, informed), *side* (left, right) and *muscle* (SCM, Rect. Abd., Ext. obliq., Neck ext., Upper trap., Lat. dorsi, ES)] was performed. In cases of significance, further two-factorial and one-factorial ANOVAs were carried out, followed by Bonferoni-corrected pairwise comparisons.

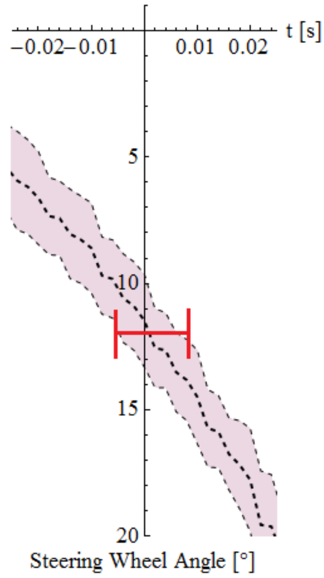
For the discussion of EMG amplitudes the signals were divided into 11 distinct intervals. Mean values of 100 ms intervals were computed and referred to as T1 through T10, where T1 contains $t \in [0\text{ ms}; 100\text{ ms})$. Additionally the interval (-100 ms; 0 ms) was taken as the baseline tonic activity and is denoted tonic. A two-factorial ANOVA with factors *side* (left, right) and *muscle* (SCM, Rect. Abd., Ext. obliq., Neck Ext., Upper trap., Lat. dorsi, ES) was then performed on the resulting data.

III. RESULTS

In contrast to braking tests, where the beginning of the maneuver can be defined exactly as the instance the brake pedal reaches a certain position or the brake pads touch the disk, in steering maneuvers this instance is not as clear. In Figure 2 the steering wheel angle and the lateral acceleration of the vehicle, averaged over all trials, are plotted. Due to the definition of the onset with a thresholding method there is already some steering wheel rotation before the defined onset of the maneuver. The lateral acceleration is delayed with respect to the steering wheel angle by 130 ms, which is mainly caused by inertia and activation of the suspension. Because of this delay the lateral acceleration shows no significant signal before t=0. Analyzing the steering wheel angle around t=0 reveals an uncertainty due to the onset definition of ±8 ms, which is displayed in Fig. 2. The peak chassis roll associated with the maneuver was below 5° in both the steering to the left and to the right.



(a)



(b)

Fig. 2. Median curves and a 68% corridor for steering wheel angle (dashed) and lateral acceleration (solid) are displayed over the course of a lateral lane change maneuver in (a). In (b) the steering wheel angle from (a) is displayed around the origin t=0. The time variation at the origin gives an estimate of ±8 ms for the uncertainty due to the onset definition.

Fig. 3 displays the kinematic response of a selected unaware trial of one single subject that showed average lateral excursion. The lateral displacements of representative points on head and torso of the same trial are displayed in Fig. 4 as well. The following distinct phases can be identified in all trials that were recorded.

1. During the first 200 ms no relevant lateral movement is observed (below 5% of maximum movement).
2. Thereafter the lateral movement to the left sets in and the maximum lateral excursion for the maneuver is typically reached after approximately 650 ms.
3. In the subsequent phase the rebound movement, i.e. the movement to the right, starts with another peak lateral excursion in the range of 1200-1400 ms. The initial, upright position is crossed after 1100-1300 ms.
4. In the final backward movement phase a large variation among the subjects is observed and no persistent pattern is found.

The remainder of the paper focuses on the first second of the maneuver, i.e. the initial response to the acceleration and the first part of the rebound movement, which is also most relevant for the development of restraint systems.

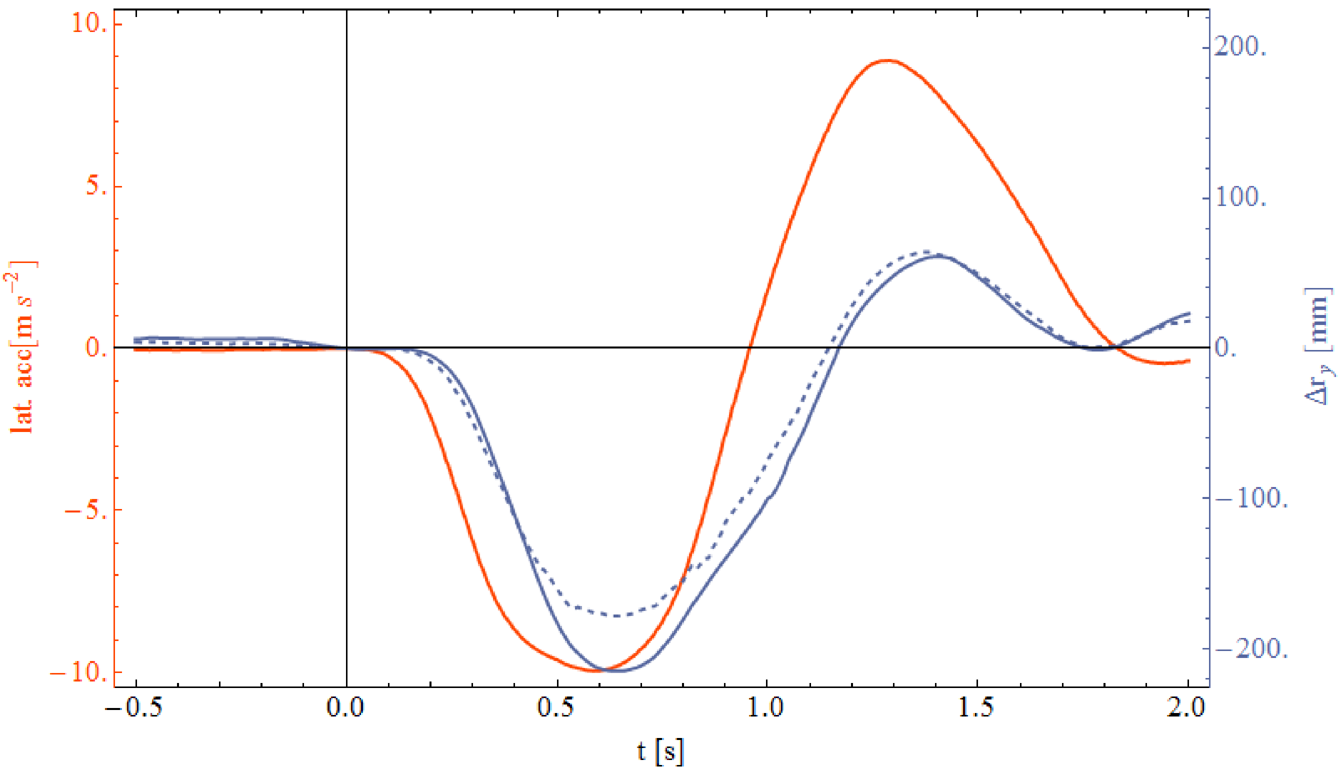


Fig. 3. Signals for lateral acceleration (solid orange line) and lateral displacement of representative points on torso (dashed blue line) and head (solid blue line) with respect to their positions at t=0 are displayed over the course of a lateral lane change maneuver for the same subject as displayed in Fig. 4.

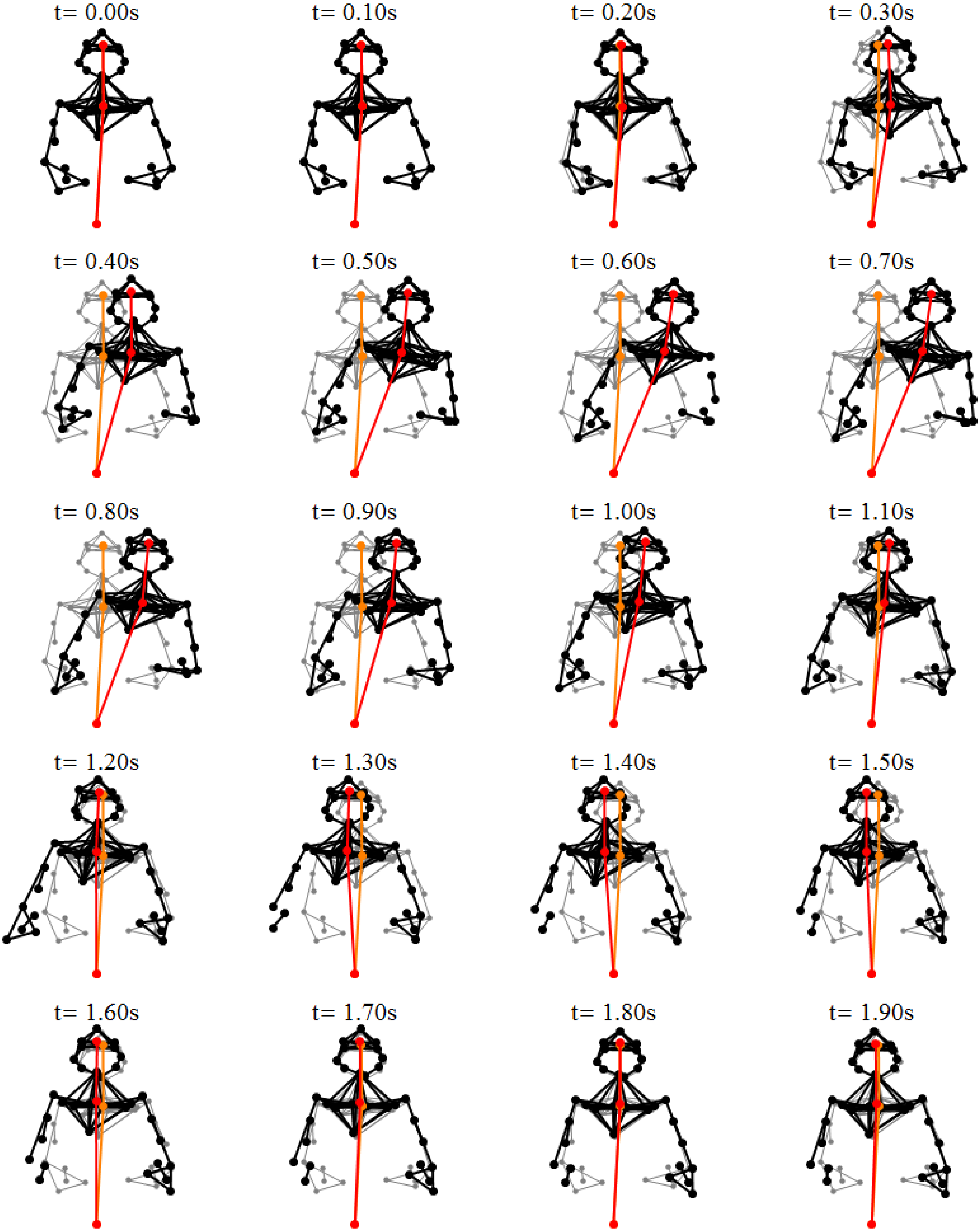


Fig. 4. Visualization of the occupant kinematics viewed from the front for a single subject and single trial that showed average lateral excursion. Head, torso and arm markers are displayed as dots, with lines indicating markers on the same body segment. The gray figure marks the subject's position at t=0 as a reference, while the black figure is the marker configuration at the denoted time instance. Additionally a representative point on head and torso is displayed and connected with the H-point of the seat. Again, red marks the current configuration and orange marks the configuration at t=0.

The mean latency values for each awareness state and muscle are presented in Table 1. All identified onset values thereby fell into the first phase of the kinematic response, where no lateral displacement was observed.

The three-factorial ANOVA revealed significant main effect of *side* ($F(1,20)=56.135$, $P=0.00$), *muscle* ($F(6,120)=14.75$, $P=0.00$) and *case* ($F(2,40)=3.63$, $P=0.03$). Significant interaction effect *side*muscle* ($F(6,120)=4.87$, $P=0.00$) was also found. To test the effect of side for each muscle Bonferoni-corrected pairwise comparisons for each case were performed with factor *side* (left, right). For muscles SCM, Ext. obliq., Neck extensors, m. Upper Trap., Lat. dorsi and ES the right side responded with shorter latency then the left side. For m. Rect Abd no difference between left and right was observed

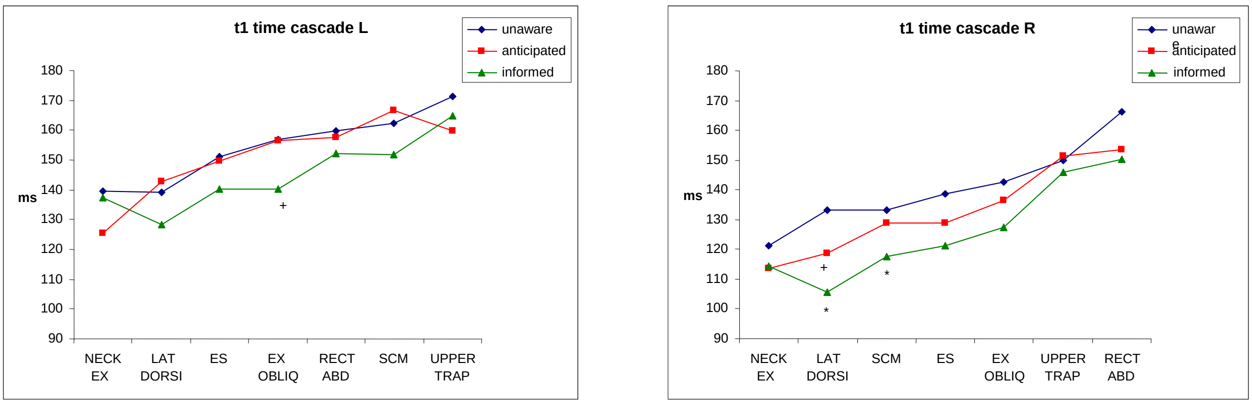
Furthermore, the latency differences between muscles were evaluated for every case separately for left and right side. At both sides shorter latency was found for Neck Ext. compared to the other muscles. At the right side abdominal muscles responded with longer latency compared to the other muscles while at the left side the latency differences between muscles was not so prominent.

To reveal any changes in latency between different cases, one way ANOVA with factor *case* for each muscle and each side were performed. The follow-up pairwise comparisons revealed significant effect between *unaware* and *informed state* ($P=0.03$). Results are presented in Fig 5.

TABLE 1

ONSET LATENCIES t1 FOR EACH MUSCLE AND EACH AWARENESS CONDITION. TIME INSTANCE t1 DENOTES THE POINT AT WHICH 50 CONSECUTIVE EMG SAMPLES HAD EXCEEDED 3 STANDARD DEVIATIONS FROM THE MEAN BASELINE REFERENCE AMPLITUDE.

	muscle	SCM		Rect. Abd.		Ext. Obliq.		Neck ext.		Upper Trap		Lat Dorsi		ES	
		right	left	right	left	right	left	right	left	right	left	right	left	right	left
unaware	mean [ms]	**133.12**	**162.42**	**166.14**	**159.75**	**142.54**	**156.82**	**121.29**	**139.47**	**149.91**	**171.40**	**133.14**	**139.27**	**138.52**	**151.14**
	SD [ms]	22.98	23.29	35.55	39.02	30.39	34.74	31.94	45.67	34.91	46.75	35.37	32.19	31.59	32.99
anticipated	mean [ms]	**128.81**	**166.48**	**153.55**	**157.63**	**136.47**	**156.49**	**113.55**	**125.37**	**151.33**	**159.61**	**118.62**	**142.90**	**128.92**	**149.66**
	SD [ms]	25.96	37.61	37.53	39.02	33.93	28.94	29.86	43.29	39.25	39.59	30.29	32.76	36.62	28.18
informed	mean [ms]	**117.42**	**151.93**	**150.25**	**152.31**	**127.42**	**140.26**	**114.34**	**137.29**	**145.75**	**164.87**	**105.72**	**128.14**	**121.10**	**140.28**
	SD [ms]	21.85	19.89	24.35	25.94	31.82	26.71	22.86	27.99	26.08	31.04	19.57	26.97	30.45	37.49

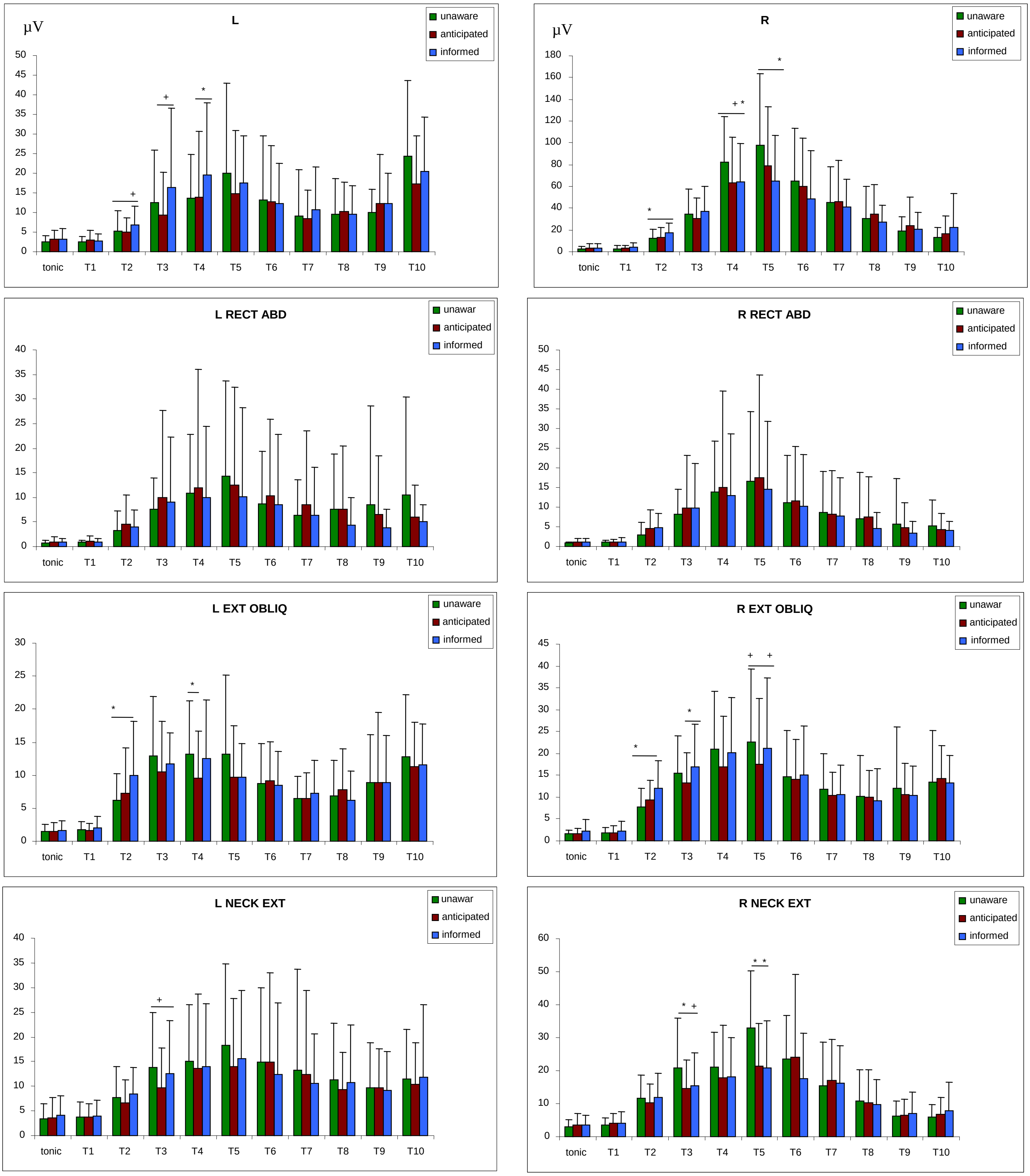


(a) (b)

Fig. 5. The onset latency cascade t1 for each muscle on the left (a) and right side (b). The different awareness states are displayed in each figure. Note that the order of the muscles on the abscissa is such that lower latencies are closer to the origin. Significant differences of the one-way ANOVA are marked by *($P<0.05$) and +($0.05>P<0.1$).

In the two-factorial ANOVA (factors side and muscle) to address the EMG amplitudes, a significant main effect of the factor *side* was found for all cases from the start of the motor response (T2) to its end (T7-T8). The main effect of the factor *muscle* was found to be significant in all time intervals. Furthermore the effect of *side* was assessed for each muscle separately. Significant levels are presented in Fig 6. The right m. SCM, m. Ext. obliq., m. Neck ext. and m. ES responded with significantly higher amplitude then the left ones in all cases. For m. Rect. abd. and m. Lat. dorsi this effect was less prominent. In contrast higher amplitudes at the left side were observed in m. Upper Trap during time interval T4-T10.

To evaluate the effect of the awareness condition, denoted as the factor *case* (unaware, anticipated, informed), one-factorial ANOVA with the factor *case* was performed for each time interval, each muscle and each side. For time intervals tonic and T1 no significant differences between cases were found. At the right side increased amplitude in the unaware condition compared to other cases were found for m. SCM and for Neck ext. In Fig. 6 the between-cases differences are displayed for each muscle.



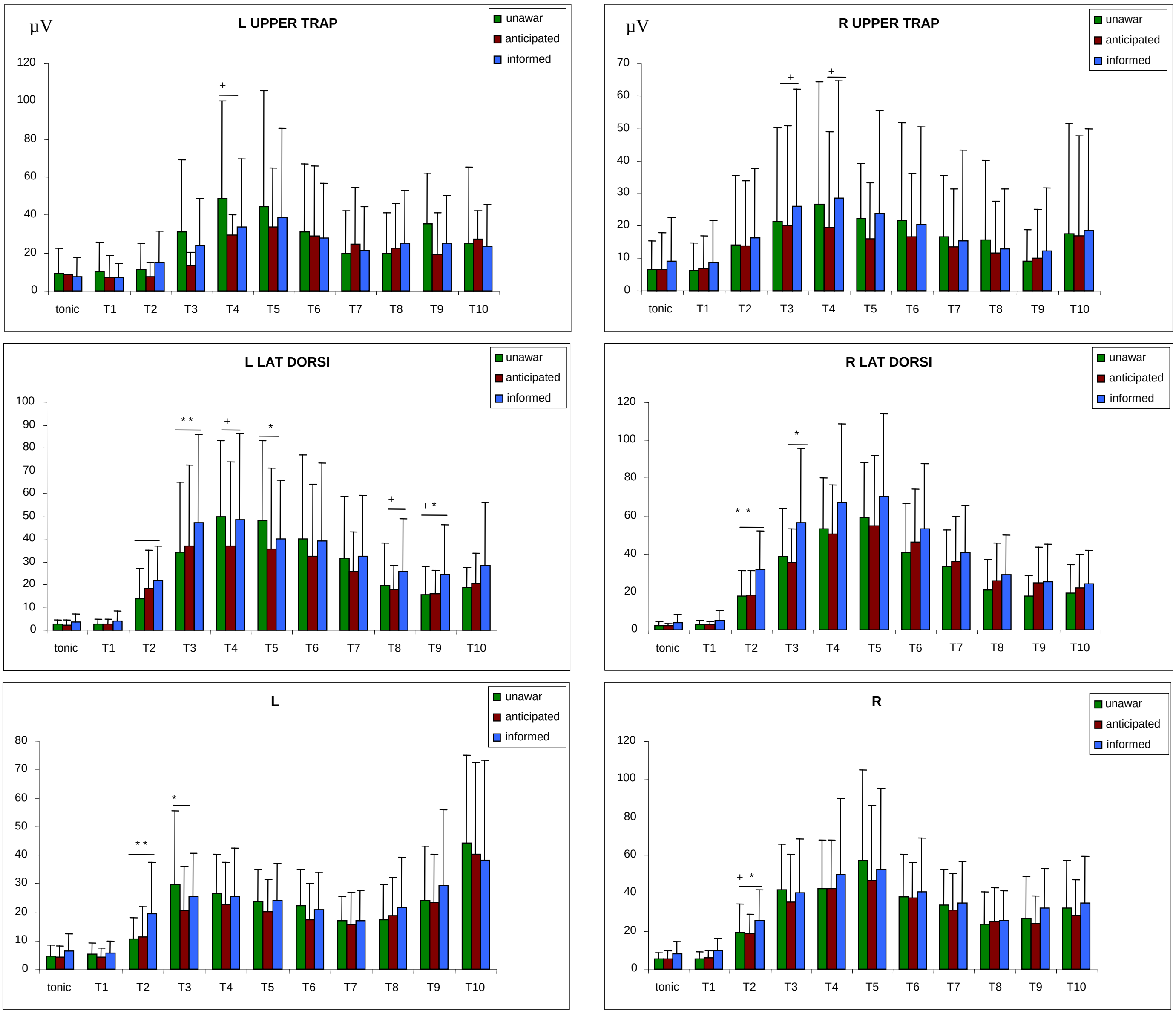


Fig. 6. The average muscle activity (μV) per 100 ms interval displayed for each muscle and each case. Significant differences of the one-way ANOVA are marked by *(P<0.05) and +(0.05>P<0.1). Note that the scaling of the ordinate axis is adjusted such that between-case differences are clearly visible when left and right muscles are compared.

IV. DISCUSSION

Onset latency

As can be inferred from the setup of the lane change maneuver, the right side muscles reacted faster than the muscles on the left side. This finding is consistent with results from [6], although the order is reversed there due to the opposite direction of the acceleration. The observed absolute values between 106 ms and 171 ms are also in agreement with the figures in said paper, although no explicit numbers are given. Compared to literature on frontal loading [5], where onsets of the SCM and paravertebral muscles of around 100 ms were reported, the presented values in this work are rather large. This can, however, be explained by the smooth onset of the lateral acceleration in the lane change maneuver of the vehicle-based test setup, which is in contrast to a sled setup, where higher jerks are realized.

In the sled study [21], where rearward impacts with load levels of 4.5 $m \cdot s^{-2}$ and 10.1 $m \cdot s^{-2}$ were performed, onsets of the SCM muscle of between 86 ms and 131 ms were found. Also in [22] and [23] rearward, whiplash like impacts at 1.1 g and 1.5 g peak sled acceleration respectively were studied. In [22] muscular activation onsets of between 55 ms and 72 ms were found for scalenus, SCM, paraspinalis and trapecius muscles. In [23] onset latencies of 67-75 ms for SCM and 75-81 ms for the paraspinal muscles were observed and were constant over repeated trials, while EMG amplitudes were significantly attenuated. The latencies found in the current study are larger, which could again be due to the smooth onset of the lateral acceleration in the lane change maneuver.

The smooth lateral acceleration onset and the comparably low jerk in this study are significant differences to sled studies and vehicle braking studies. While in the later type of studies the maximum acceleration is typically reached before the first active muscle responses, the lateral acceleration only reaches between 0.3 $m \cdot s^{-2}$ and 1.3 $m \cdot s^{-2}$ before muscle onsets are detected.

The neck extensors responded with the shortest latency on both sides probably because of the contribution of the fast vestibulocollic and cervicocollic reflexes [24]. Furthermore, no significant difference between the three cases was observed, although a trend for reduced latency in the informed case was present.

EMG amplitudes

The swerving maneuver was performed initially to the right, such that the occupants initially moved to the left. For this phase we observed a strongly asymmetric muscle activation effect, where right side muscles were activated significantly stronger than left side muscles. Similar findings were shown in [6], where the left side muscles were activated more strongly due to the opposite direction in which the maneuver was performed.

The main outlier in this observation was m. upper trapezius. During the time intervals T4-T10 the left side muscle was activated more strongly than the right side. One of the main functions of the upper part of the trapezius is elevation and stabilization of the shoulder. Visual checks of the kinematic data confirmed that no significant shoulder elevation took place at the left side during this period. A possible explanation for this behavior is that the trapezius was activated mainly in stabilization of the left arm. The time intervals T4-T10 contain the maximum excursion to the left and due to the lack of lateral stabilization of the modified seat, bracing at the center console could be a reaction. Nevertheless actual bracing was not observed and co-contraction patterns cannot be checked, because no signal of trapezius' antagonist was recorded.

Interestingly, no significant difference in the signal's amplitude could be observed between the different awareness states, although a tendency for lower latencies in the informed case was observed.

Limitations and improvements

With vehicle-based testing performed on a test track one step towards a naturalistic response of the occupant is taken. Nevertheless, the tests still had signs of a testing situation, i.e. application of the EMG-electrodes and EMG-cables, subjects had to wear a motion tracking suit with markers attached, a non-standard seat was used, the windshield was taken out, cameras were present and driving took place on a closed test track.

This study concentrated on male subjects only and the tested population was on average 4 cm taller than the 50^{th}%ile male. Furthermore only the co-driver position was tested, hence effects of bracing could not be evaluated. In follow-up studies the influences of gender and anthropometry as well as the difference between co-driver and driver need to be addressed.

The study design with a constant sequence of unaware, anticipated and informed conditions allows to distinguish between the unaware condition and repetitions. A distinction between the two remaining aware conditions cannot clearly be made, as influences from habituation and information cannot be disentangled [22-23].

Another limitation of the study is that no maximum voluntary contractions were preliminarily obtained from the target muscles in order to normalize the amplitude data samples [25]. Therefore, for discussion of amplitudes an individual measure that relates the voltage signal to the activation level at which the muscle operates is missing. Nevertheless, interpretation about relative differences between left and right side and between the different awareness states are still valid, assuming that the quality of the EMG assessments was very good and that distortion effects by the tissue between muscle and electrodes were comparable on both sides.

Of course the general limitations of studying muscle activity with surface EMG also apply in this study, e.g the activation of deep muscles that play a role in the stabilization of the trunk cannot be measured on the surface and especially in areas like the neck, where there are various muscles close to each other, activation from adjacent muscles will be present in the signal.

Finally, the presented values only cover the average muscular activation over all subjects. As a next step correlation of individual kinematic responses and the respective muscle response need to be analyzed.

V. CONCLUSIONS

In this paper results from full vehicle tests with 21 subjects are presented. Peak lateral accelerations of approximately 1 g were achieved in the lane change tests with lap belt only. The resulting analysis of the onset latency of 2x7 muscles allows determining an onset cascade of the stabilizing muscles in the maneuver. This information can in turn be used for designing and refining numerical human body models that take the reactive muscle contribution into account.

A tendency for lower latencies in the informed cases is observed, however significant only for particular muscles. In addition no differences in the EMG amplitudes between different awareness states were found.

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