Neuroendocrinology Letters *Volume 34 No. 7 2013*

ORIGINAL ARTICLE

Pre-activation and muscle activity during frontal impact in relation to whiplash associated disorders

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Submitted: 2013-09-13 Accepted: 2013-10-10 Published online: 2013-12-13

Key words: **head injury criterion; impact; injury; muscle activity; whiplash**

Neuroendocrinol Lett 2013; **34**(7):708–716 **PMID:** 24464011 NEL340713A06

Abstract

OBJECTIVE: For the evaluation of neck injury the relative distance was observed between a marker placed on the forehead and a marker placed on the shoulder and also by change of the angle. To compare the severity of head injury a value of maximum head acceleration was used, HIC and a 3 ms criterion. All criteria were related to the activity of musculus sternocleidomastoideus and musculus trapezius in a situation of expected or unexpected impact.
MATERIALS AND METHODS: The situation was recorded using a Qualisys system, head acceleration of probands in three axes was recorded using the accelerometer, activity of neck muscles was monitored by a mobile EMG.
RESULTS: Maximum head acceleration was 12.1 g for non-visual and 8.2 g for visual. HIC36 was 5.7 non visual and 4.0 for visual. 3-ms criterion was 11.5 g for non-visual and 7.8 g for visual. The average time of muscle activation of the observed group without visual perception is 0.027 s after hitting an obstacle, with visual perception 0.127 s before the crash.
CONCLUSIONS: Kinematic values indicate more favourable parameters for neck injuries for visual. Head injury criteria show an average decrease of about 30% for visual. We can conclude that the visual perception means a significant increase in pre-activation of the observed muscle group of almost 400% and lower activation in both following phases of approximately 40%.

INTRODUCTION

The cervical spine is principally more mobile than the other surrounding segments of the body and therefore it is flexible but also vulnerable. In addition, the head weight and rigidity, accent the vulnerability of the neck. The trunk-neck-head connections are secured by the spinal architecture and surrounding soft tissue. Not only do the shape and alignment of cervical vertebrae guarantee the neck-head stability, but also the ligaments and voluntarily controlled muscles, for example, the sternocleidomastoid muscle or the superior fibres of the trapezius muscle.

Head position in space is maintained by the cooperation of a supportive (musculo-skeletal) system and by a vestibular system. Information from the vestibular system and from the muscle and tendon receptors (muscle bodies, Golgi bodies) is processed by the cerebellum, which controls muscle tone and ensures an upright position and balance of the body.

According to the neck and head mechanical relationship, neck injuries and disorders are frequently linked to external impact. Although speed and mass (momentum) are directly proportional values that raise the probability of neck and/or head injuries. External impact, even at low speeds, can result in the head acceleration-deceleration mechanism that causes an energy transfer and a sudden distortion of the neck. The phenomenon is referred to as whiplash. According to the aforementioned statements, whiplash injuries are not only associated with high speed and head-on collisions but also with low speed and rear-end or side collisions.

This indicates that the voluntary muscle activity and awareness of the collision may prevent form whiplash associated disorders. Moreover, Kumar *et al.* (2004b) stated that the whiplash syndrome is not just an issue of acute injuries that is evident, which means that it often develops into chronic pain and latent symptoms that last for months or years. The finding supports the significance of voluntary muscle activity as a whiplash precursor. In spite of the fact, whiplash associated disorders are extremely complex and affect a wide range of tissues that a play crucial role in the cervical spine stability and functionality. Unfortunately the significance of each component and voluntary muscle activity in the hierarchy is still not fully understood and therefore a sufficient technique for finding solutions is not present (Chen *et al.* 2009). In the effort to better understand the whiplash phenomenon, researchers pay attention to the development of software packages and the methods of segments classification including findings from biomechanics, neuroendocrinology etc.

Due to increasingly sophisticated computer models and mannequins for the injury analysis presents an important aspect the knowledge of the response to impact with a concurrent human activity. Presented values can be an important input parameter for development of so-called active models that are supposed to respond in the same way as the human body. A major topic of current research in the field of automotive safety is the development of human models for the application in numerical crash simulation. In comparison to dummies or dummy models human models are expected to reproduce human kinematics in a more realistic and detailed way (Muggenthaler *et al.* 2008).

There are several software packages that deal with the complexity of the neck and head region. They are usually based on theoretical background and experimental findings involving tests on cadavers and dummies. Basically, the two most common approaches, as the result of the simulations, are the finite element method (Pam-Crash, Radioss, etc.) and multi-body method. Nowadays, the approaches are frequently integrated to gain the advantages of both (specificity and speed), for example, MADYMO, ANSYS. Although the approaches reach a decent accuracy and allow one to investigate collisions at low and high speeds, they remain more or less passive and do not reflect voluntary or reflexive muscle response. There is no doubt that the field of whiplash injuries requires a more sensitive approach, for example, muscle and neuro-activity, viscoelastic behaviour of ligaments, etc.

The aim is to embrace the relationship between voluntary muscle activity and whiplash associated disorders. Furthermore, voluntary muscle activity is considered to be one of the essential factors in whiplash associated disorders and the probability of suffering from whiplash injuries is affected by the subject awareness before impact. Despite the awareness which has been included in some studies (Mazzini & Schieppati 1992) and the influence to the final muscle activity has been stated, authors have been discussing the significance of reflexive or voluntary muscle activation to whiplash associated disorders (Siegmund *et al.* 2003).

The most common method for monitoring muscle activity si surface electromyography (sEMG). sEMG has been used in both clinical and experimental fields and offers a wide range of variables that can be tracked (Rainoldi *et al.* 2004; Finsterer 2001; Casale *et al.* 2003). Nevertheless, the lack of sECM experiments that include sudden impact and voluntary or reflexive muscle activity limits the understanding of the whiplash associated disorder phenomenon. Of course, the monitored sECM variables vary in value, depending on the position of the electrodes on the skin, respectively on the muscle and measuring protocol (pass filters, triggering, etc.). Several studies have been published in an effort to standardize the methods. To minimize motion artefacts and noise, De Luca *et al.* (2010) suggests 20 Hz as the best compromise to avoid data loss. Frequency of 500 Hz is recommended by a low-pass filter for the neck muscles (Brault *et al.* 2000). According to van Boxtel (2001), optimal high-pass filter frequencies were determined for the mean power spectra based on visual estimation or comparison with a theoretical spectrum of the artefact-free EMG signal. The optimal frequencies for the different muscles varied between 15 and 25 Hz and were not influenced by stimulus or response modality. For all muscles, a low-pass filter frequency between 400 and 500 Hz was appropriate.

EMG signal normalization method was applied through the highest value achieved during the impact for a given muscle (so called Dynamic Peak Method – DPM). DPM normalized by expressing them as a percentage of the peak EMG from the same contraction. This method of normalization is used to compare activity between tasks (Lazaridis *et al.* 2010; Albertus-Kajee *et al.* 2011). DPM can be used to evaluate changes in load and speed of movement (Burden & Bartlett 1999). It cannot be used for comparison while increasing the external load. DPM reflects minor differences, thus producing greater homogeneity. Output sizes from isometric and isokinetic maximal voluntary contraction (MVC) are comparable with DPM (Burden & Bartlett 1999). The magnitude of the outputs from the Isometric and Isokinetic MVC methods are similar to DPM (Burden & Bartlett 1999). The Dynamic Peak method has successfully reduced inter-subject variability. This method would be expected to reflect changes in muscle activation levels between tasks (Burden & Bartlett 1999). In our study, we compare the particular muscle activity with and without visual perception for each muscle, which is not negatively affected by a DPM normalization.

This approach enables us to detect values from the sternocleidomastoid and the superior fibres of the trapezius among the volunteer participants and compare the response of voluntary and reflexive muscle activity with the head acceleration-deceleration kinematics measured by Qualisys (motion capture system).

MATERIALS AND METHODS

To simulate the deceleration, an impact simulator was utilized to mimic an impact of a car into a solid obstacle at a speed of 30 km/h. It is basically a cart with two automotive seats (just like the front seat of a car) with three-point safety belts, which descends down an inclined plane and crashes into a fixed barrier.

Measurements took place on 8 subjects (6 men, 2 women), 24–30 years of age, weight 79±6 kg. The subjects were healthy and never had any cervical spine problems. The whole situation was recorded using a Qualisys system (3 cameras) and adhesive passive markers at a scanning frequency of 1,000 Hz and also using a digital camera capable of recording in slow motion. At the same time the head acceleration of the subjects in three axes was recorded using a forehead mounted accelerometer and the cart acceleration in the direction of travel. Activity of the neck muscles was monitored by a mobile EMG synchronously to the acceleration recording using surface electrodes on the right and left musculus sternocleidomastoideus (M. SCM) and the left and right musculus trapezius (M. T).

Kinetic analysis was performed with Qualisys Track Manager software for the relative distance between the marker placed at the forehead and the marker placed on the shoulder of the subject during the impact and also the changing angle between the line connecting these two points and the horizontal axis. Distances in 2D were compared, side movements were not considered.

To record the acceleration, Dewetron technology was used with Kistler sensors set at a frame rate of 10000 Hz. Recorded data was imported into HyperGraph software and filtered according to the Euro NCAP methodology by CFC 1000 filter (3 dB limit frequency, 1650 Hz and stop damping 40 dB). Before the final evaluation a resulting acceleration curve was calculated from rectified data from the each individual axes.

EMG of the neck muscles was monitored by a multichannel Biomonitor ME 6000 device, which provides RAW data filtered in the highpass-lowpass range 8–500 Hz with frame rate of 1000 Hz.

Data was imported into the HyperGraph, rectified, normalized (DPM) and divided into three parts – preactivation before impact, the initial movement phase and the following movement phase according to Ekblom & Eriksson (2012).

Preactivation:
- 0.5 s before maximum simulator deceleration to maximum simulator deceleration
- detection of presence, respectively importance of muscle preactivation

Initial movement phase:
- maximum simulator deceleration to maximum head acceleration
- assessment of muscle activation during impact

Following movement phase:
- maximum simulator deceleration to 0.5 s after maximum simulator deceleration
- assessment of muscle activation after impact

The first parameter investigated was the time when muscle activation occurs. As the value for determining the activation time a value of 10% of the DPM value was used corresponding to more than twice the standard deviation of the baseline values. And then the quantification of the EMG signal that was designed as a mean size at any given time.

The final evaluation of the results of the examination was performed according to current statistics for a normal distribution with a significance level of 5%.

RESULTS

<u>Qualisys results</u>

According to the Qualisys records the impact speed of the cart was 2.96±0.02 m/s, input conditions were therefore identical for all measurements.

The Qualisys system monitored the distance between the forehead marker and the cart marker. The average distance between the resting position and maximum head displacement of all the measurements is shown in Figure 1. It is clear that in the case of unexpected impact (no visual perception) the head reaches a larger displacement and the distance to the extreme head position is therefore longer. This is confirmed by another observed parameter – the change in angle between the line connecting the head and the cart related to the horizontal plane during movement – Figure 2.

One can see the impact course of the average values from all measurements of the distance between the frontal bone and the cart (Figure 3). Before time 0,

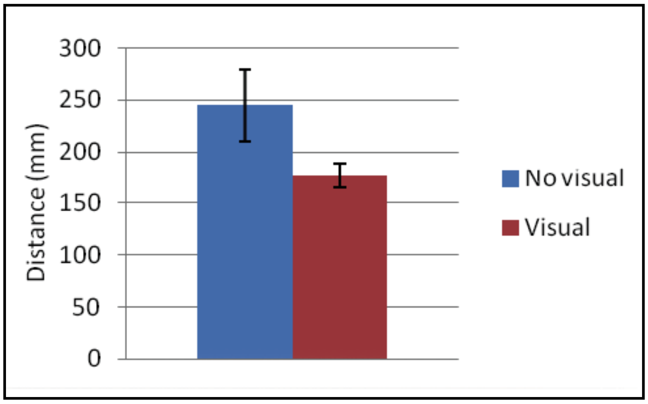

Fig. 1. The average difference in distance between resting position and the maximum displacement.

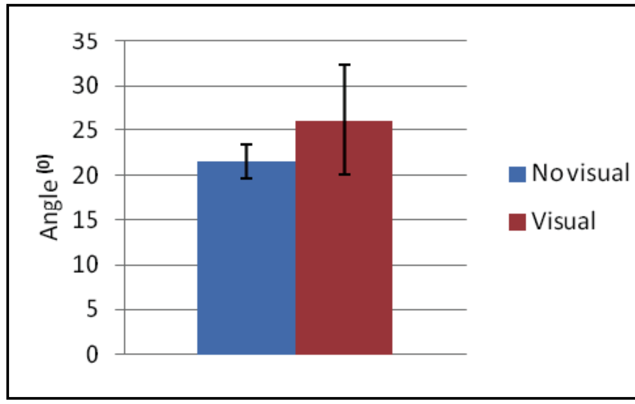

Fig. 2. Average difference in angle between the line connecting the head and cart during movement.

which is the moment of impact, the head of subjects with visual perception purposefully leans back (closer to the headrest). This is followed by impact and among the subjects without visual perception we can see a steeper increase in distance in time mainly due to the unpreparedness of the neck muscles for impact. Subjects expecting the impact activated the muscles and displacement increase is gradual. Changes in the curve flow at the time of approximately 0.05 s after impact are caused by the safety belt holding the subjects. After reaching the maximum displacement which is lower by about 50 mm among subjects with visual perception the distance is closing again (moving back towards headrest). Among subjects expecting the impact we can see faster distance reduction during the impact in the aftermath of overload applied to the head in the direction of movement. The head therefore actively returns back to the original position on the headrest.

The results of monitoring the head speed course after the impact show that the maximum speed of the head achieved without visual perception was 4.94±1.09 m/s, with the visual inspection then 4.27±0.67 m/s.

Acceleration results

Average deceleration of the simulator calculated from all measurements was 27.28±1.19 m/s^2. All initiation forces acting on the subjects can therefore be considered identical.

Table 1 shows the basic results of statistical processing of this data, which is a base of the final evaluation of the measurement.

From Table 1 it can be seen that the measured data set confirms the normality of the distribution, and therefore a pair test was used for the resulting comparison to find differences between sets of measurements with and without exclusion of visual perception (Table 2).

For better clarity, the data has been organized in Figure 4, including the display of standard deviations with error bars.

From the graph a slight decrease in all three monitored variables can be seen, due to the visual perception. Based on the values of Tables 1, 2 we can add that for the chosen significance level, this change is statistically significant at the maximum acceleration values and 3ms criterion. If we consider the values without visual perception a 100, then we can formulate the following partial results:

- Maximum head acceleration with visual reduction of 3.9 g (–32%)
- HIC 36 with visual reduction of 1.7 (–29%)
- 3 ms criterion with visual reduction of 3.7 g (–32%)

Tab. 1. Statistical values of measured accelerations.

	Max. head acceleration (g)		HIC 36		3-ms criterion (g)	
	No visual	Visual	No visual	Visual	No visual	Visual
Mean	12.096	8.163	5.702	4.017	11.466	7.812
Standard deviation	2.414	2.760	1.809	2.907	2.192	2.333
Min	8.235	5.769	2.462	2.028	7.717	5.689
Max	15.978	14.698	8.380	11.356	14.899	13.179
Median	12.436	7.299	5.979	2.846	12.021	7.054
Norm. test Kolmogorov (prob)	1.000	0.359	1.000	0.335	1.000	0.474
Norm. test Shapiro-Wilk (prob)	0.953	0.019	0.898	0.001	0.930	0.041

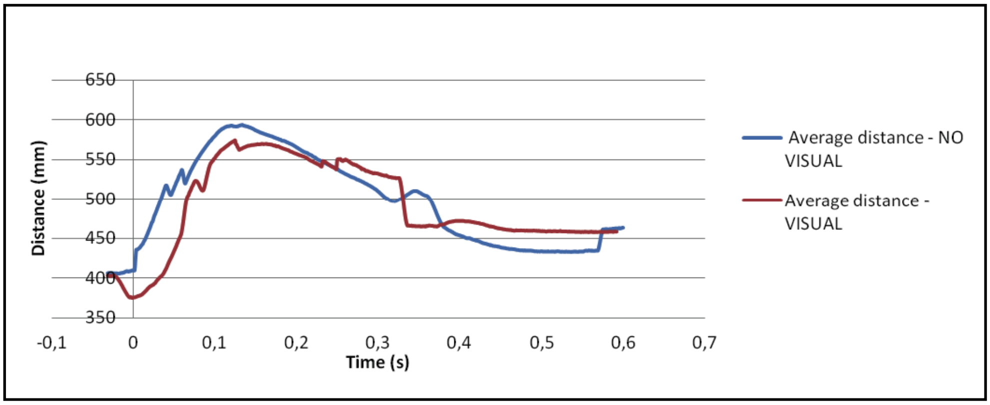

Fig. 3. Average distance between the frontal bone and cart.

It can therefore be concluded that the visual perception will enable faster reaction of muscles and reduce the risk of injury.

<u>*EMG results*</u>

a) Time of activation

Activation of muscles measured, were in the first step of data processing, monitored on the time axis in relation to the simulator acceleration course. The moment of maximum simulator acceleration was considered in three factors: a reference, a default and zero time for further evaluation. Relative to this point, the times of maximal activation of monitored muscles were determined. Activation before the reference point is shown as a negative number, activation after, as a positive sign. A summary of results is in Table 3 which provides the mean values, differences in mean values of activation times without visual perception versus with visual perception. Based on this data we can say that muscle activation without visual perception commences from 0.025 s (m. trapezius) to 0.029 s (m. sternocleidomastoideus) after impact. With visual perception the monitored value changes from –0.136 s (m. sternocleidomastoideus) to –0.117 s (m. trapezius), ergo before the impact.

If we compare the average time of muscle activation of the observed group, we can say that without visual perception it comes at 0.027 s after hitting an obstacle, with visual perception 0.127 s before the crash.

These conclusions are clearly evident in the Figure 5.

b) Quantification of activation

The measured data was further assessed in terms of the degree of involvement of muscle groups monitored during the collision. For further work a previously defined reference point on the time axis was used (the moment of minimum simulator acceleration) which is referred to as impact in the following text.

Tab. 2. Statistical comparison of measured accelerations.

	Max. head acceleration (g)		HIC 36		3-ms criterion (g)	
	No visual	Visual	No visual	Visual	No visual	Visual
Mean difference	–3.934		–1.684		–3.654	
Percentual difference	–32.518		–29.543		–31.864	
Pair t-test (prob)	0.004		0.130		0.004	

Tab. 3. Timing differences.

	Sternocleidomastoideus				Trapezius			
	Left		Right		Left		Right	
	No visual	Visual	No visual	Visual	No visual	Visual	No visual	Visual
Mean	0.028	–0.165	0.030	–0.107	0.019	–0.122	0.031	–0.112
Std. Deviation	0.016	0.047	0.017	0.052	0.011	0.027	0.017	0,035
Difference of Std. Deviation	–0.194		–0.138		–0.141		–0.144	
Percentage difference	–682		–455		–726		–460	

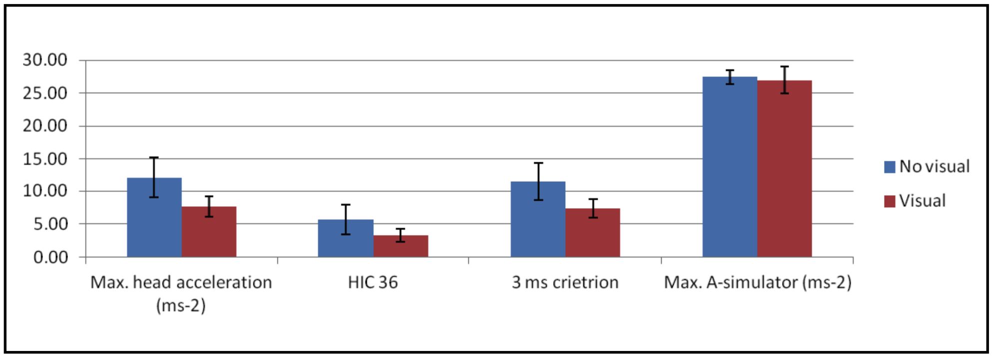

Fig. 4. The results summarized in the graph.

Tab. 4. Activation evaluation on m. sternocleidomastoideus, left.

	SECTION 1		SECTION 2		SECTION 3	
	No visual	**Visual**	**No visual**	**Visual**	**No visual**	**Visual**
Mean	0.008	0.022	0.216	0.151	0.122	0.068
Standard deviation	0.006	0.011	0.016	0.014	0.042	0.007
Means difference	0.013		−0.065		−0.053	
Percentual difference	161		−30		−44	
Pair t-test (prob)	0.00058		0.00006		0.01248	

Tab. 5. Activation evaluation on m. sternocleidomastoideus, right.

	SECTION 1		SECTION 2		SECTION 3	
	No visual	**Visual**	**No visual**	**No visual**	**Visual**	**No visual**
Mean	0.004	0.015	0.256	0.110	0.082	0.046
Standard deviation	0.006	0.018	0.030	0.044	0.022	0.015
Means difference	0.011		−0.146		−0.036	
Percentual difference	248		−57		−44	
Pair t-test (prob)	0.04417		0.00004		0.00315	

Tab. 6. Activation evaluation on m. trapezius, left.

	SECTION 1		SECTION 2		SECTION 3	
	No visual	**Visual**	**No visual**	**No visual**	**Visual**	**No visual**
Mean	0.010	0.029	0.265	0.145	0.122	0.079
Standard deviation	0.005	0.007	0.136	0.044	0.027	0.022
Means difference	0.020		−0.120		−0.043	
Percentual difference	207		−45		−35	
Pair t-test (prob)	0.00000		0.01195		0.01031	

Tab. 7. Activation evaluation on m.trapezius, right.

	SECTION 1		SECTION 2		SECTION 3	
	No visual	**Visual**	**No visual**	**No visual**	**Visual**	**No visual**
Mean	0.003	0.017	0.139	0.098	0.103	0.060
Standard deviation	0.001	0.006	0.062	0.053	0.047	0.019
Means difference	0.015		−0.041		−0.043	
Percentual difference	570		−29		−42	
Pair t-test (prob)	0.00026		0.00026		0.00756	

For the purposes of further evaluation from the above mentioned point of view, three sections were introduced in the time line, which could be with respect to literature data (Ekblom & Eriksson 2012) considered typical:

SECTION 1: "Pre-activation"
– 0.5 seconds before impact to the impact
SECTION 2: "The initiation phase of movement"
– from the impact to maximum head acceleration
SECTION 3: "The following movement phase"
– from the impact to 0.5s after the impact

From the measured activation courses of observed muscles, their mean values were calculated in defined sections. In order to perform a final comparison they were further averaged and the differences were determined between the possible, respectively impossible visual perception. Percentage differences were calculated relative to the values without visual perception (Tables 4–7).

From the above stated, the differences are clearly significant in observed muscle activation in individually evaluated sections between the possible and impossible visual perception, when positive values indicate higher activation with visual and negative indicate higher activation without visual.

Similar to the previous phase of the evaluation, the conclusion is clearly visible in the illustrative graphical display (Figure 6), where the average value of activation in respective section is on the vertical axis.

If we again consider the response of the entire observed muscle group, we can construct the following Table 8, with average values of the percentage difference between possible and impossible visual perception.

Regarding the above stated we can conclude that the visual perception means a significant increase in pre-activation of the observed muscle group of almost 400% and lower activation in both following phases of approx. 40%.

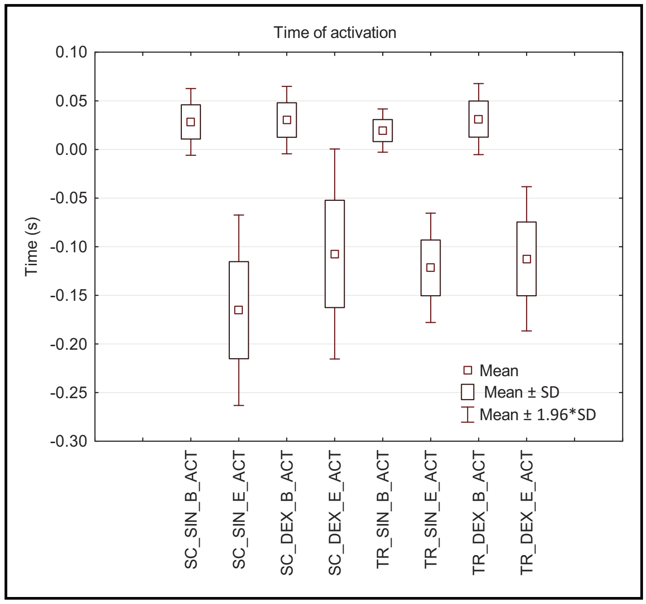

Fig. 5. Activation of individual muscles.
SC=sternocleidomastoideus, TR=trapezius, SIN=left, DEX=right, B=no visual perception, E=visual perception.

Tab. 8. Activation evaluation on m.trapezius, right.

	SECTION 1	SECTION 2	SECTION 3
Means difference	0.015	−0.093	−0.044
Percentual difference	398.9	−40.4	−41.3

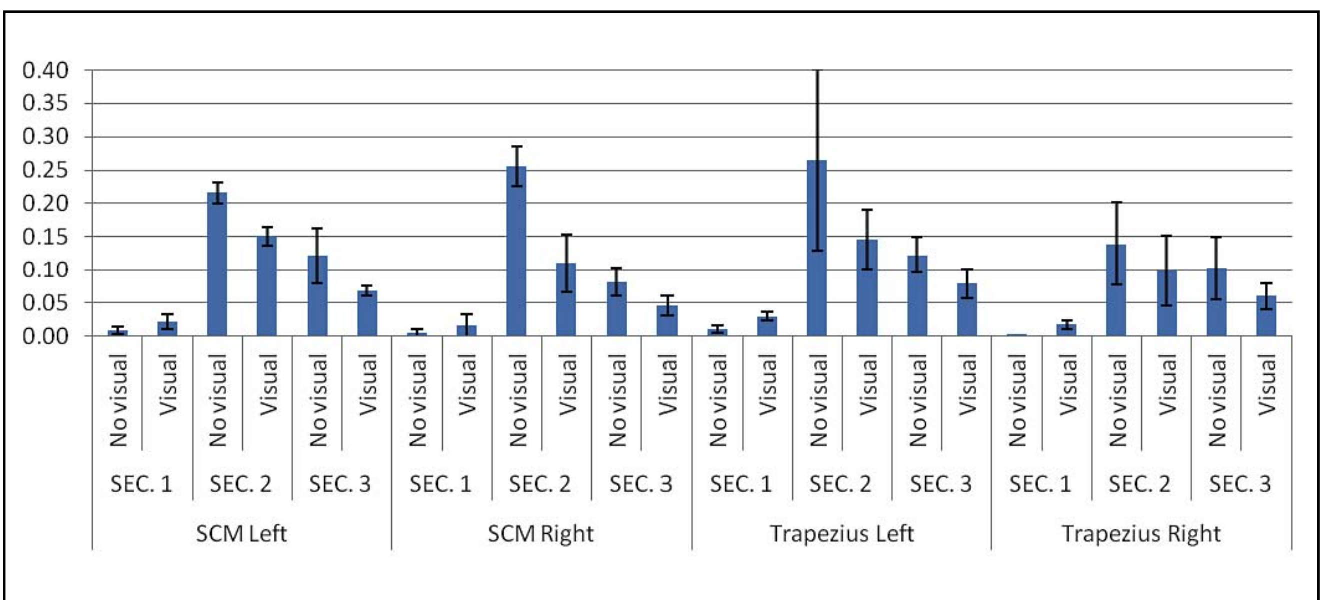

Fig. 6. Mean of normalized activity

DISCUSSION

Kumar published, between 2001–2006, more than 7 works dealing with EMG during whiplash (Kumar *et al.* 2005a;b; Kumar *et al.* 2006; Kumar *et al.* 2004a;b; Kumar *et al.* 2001; Kumar *et al.* 2002; Kumar *et al.* 2003). Kumar performed the experiment with 10 subjects and used a plastic chair. The acceleration was provided by the propulsion of a pneumatic cylinder. In our research 8 subjects were tested while seated in an actual car seat which collided with a solid barrier. The acceleration values reached for sledges by Kumar were 14 m/s^2 for unexpected impacts and 12 m/s^2 for expected. The maximum head accelerations reached 8.4 m/s^2 for unexpected impacts and 7.8 m/s^2 for expected. The acceleration of the simulator that was adopted for this experiment was 27.3±1.2 m/s^2 for all performed measurements. The head acceleration was 12.1±2.4 m/s^2 for unexpected impacts and 8.2±2.8 m/s^2 for expected. With increase in acceleration the muscle activity increased. Kumar further discovered that EMG activity decreases during expected impact two times more than during unexpected impact. For comparison of percentage activation of individual muscles, Kumar began with a reference value of 100%, which represented maximum voluntary activity in flexion, extension and rotation for following muscles in this order: m. sternocleidomastoideus, m. trapezius, m. splenius capitis. The highest activity was recorded in musculus trapezius and it was between 38–79%. This activity increased proportionally to increasing acceleration during unexpected impacts. During expected impacts the activity also increased with increasing acceleration, however with lower total values (32–53%) (Kumar *et al.* 2003). According to Kumar *et al.* (2003) it turned out that there is no gender dependency in the muscle activity.

Contact-less impacts were further analyzed by Gong *et al.* (2008). Using a finite elements method, he created a model of the head and neck, which he integrated with ATB (Articulated Total Body (McHenry 2004)). This system was placed on a car seat model with a seat belt and applied deceleration of 13.3, 23.5 and 33.7 g. Gong *et al.* (2008) particularly monitored intra-cranial pressure and shear stress. His acceleration values measured on the head were three times higher than the initial impact phase. During our tests on the impact simulator, we measured accelerations lower by a third on the head compared to the simulator cart.

Available studies further state that reflexive muscle response to external stress was detected by EMG in range 30–150 ms (Larivière *et al.* 2010). Murakami (2010) presented that the reaction time between noticing the object and detecting EMG is 0.2 s and that the delay between EMG activation and the start of actual motion is 0.05 s. Both cited sources thus confirm our measured activation values were within 130 ms time. We further discovered in our measurement that without the knowledge of exact impact time, the neck muscles activate 27 ms after impact.

It is assumed that head kinematics is influenced by muscle activation only to certain acceleration value and impact force. Nevertheless it has been demonstrated that whiplash injury is very frequent in low speeds and relatively weak impacts. For instance, during a rear impact at 8 mph the acceleration of a car can reach 2 g and head acceleration can reach 5g with effect time of 300 ms (Teo *et al.* 2007).

The diagnostics of neck and whiplash injury is medical and traumatological problem (Rodriquez *et al.* 2004; Geiger & Aliyev 2012; Malatova *et al.* 2007). There are, however, more areas which could make use of better factors of Whiplash Associated Disorders (WAD) qualification, for example, the fields that investigate the influence of whiplash on brain, CNS and endocrine glands respectively. In these areas may the information about muscle pre-activation in various conditions, may, along with various values of head acceleration serve as a parameter, allowing more precise classification of investigated groups. The methodological improvement is called for by e.g. Gaab *et al.* (2005), who in his preliminary study deals with regulatory dysfunction of hypothalamus–pituitary–adrenal (HPA) axis connected with WAD. Among 20 patients with chronic WAD, a cortisol level was examined during the day and the effect of administering dexamethasone was evaluated. Despite differences found between the groups with and without WAD, there was a lack of qualitative criterion for detecting WAD among the individuals with WAD.

CONCLUSION

The results unequivocally conclude that impact awareness and an active preparation in advance through increasing motor neuron potential helps to reduce head deceleration and therefore it is possible to expect less severe injury. It was proven that expecting the impact increases neck muscle activation by 400%. This activation culminates approx. 130 ms before impact in the case the upcoming impact is recognized. This results in lower head deceleration course and head displacement angle and consequently in decrease of maximum head deceleration and head injury criterion values HIC36 and 3-ms criterion.

Eventhough the conducted data on live subjects offer a wide range of information that is necessary for improvement of active human models in simulation software, further experiments in related fields, for example, neurology, psychology and endocrinology are necessary. In addition, after that it would be possible to increase the success of predicting WAD using the software. Furthermore, a better WAD evaluation instrument has recently been sought by those in forensic biomechanics, judges and solicitors and even by leaders in the automotive industry, mainly in relation to warn passengers in time and reduce the severity of traffic collisions.

ACKNOWLEDGEMENTS

This project was supported by grants GAUK 111310, GAUK 956213, GACR 407/10/1624, and SVV 2013-267603.

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