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ORIGINAL ARTICLE

Kinematic and dynamic biomechanical values in relation to muscle activity during contact head impact

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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 contact 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 5.61 *g* for non-visual and 5.03 *g* for visual. HIC36 was 6.65 non visual and 5.97 for visual. 3-ms criterion was 5.37 *g* for non-visual and 4.89 *g* for visual and max. force was 291 N for non-visual and 314 N for visual. The average time of muscle activation of the observed group without visual perception is 0.355 s after hitting an obstacle, with visual perception 0.085 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 10% for visual. We can conclude that the visual perception means a significant increase in pre-activation of the observed muscle group of almost 745% and lower activation in following phase of approximately 90%.

INTRODUCTION

Head injury is the most frequent causes of death in productive age (Jennett 1996). Due to the fact that this trend has been rising, the research activity in simulation of origin and consequences of such an injury has increased accordingly (Finfer & Cohen 2001).

The basic method of contact impacts simulation is so called "fall test". It is a fall of a human head model on a platform from a defined height. Physical variables are being recorded during the test and correlated with mechanical damage. This test neglects a relation between a cervical spine and a head, furthermore there is no way to simulate expectations or readiness for the impact. In this study we have tested kinematic and dynamic aspect of contact head impacts in relation to the head – cervical spine connection and neck muscles.

Contact impacts utilizing a pendulum were analysed in detail by Verschueren *et al.* (2007). He was reaching maximum impact force of 15000N and 10% of total impact energy absorption in the head during testing of cranial fractures. Using double pendulum is appropriate according to his conclusions because it naturally simulates reaction of a head to an obstacle.

Verschueren's data (Verschueren *et al.* 2007) was used by Asgharpour (Asgharpour *et al.* 2013) to validate the SUFEHM (Strasbourg University Finite Element Head Model). Results of the SUFEHM simulation of contact impacts corresponded with experimental data. For adoption of the SUFEHM in forensic biomechanic validation is needed for further impact types. The results presented here may therefore serve to validate this model.

The work with living probands was performed by Fukushima *et al.* (2006). He recorded kinematics of the cervical spine, activity of *m. sternocleidomastoideus* and paravertebral muscles. The load was applied to a forehead and upper jaw using a weight and a system of pulleys. In the end the S-shaped flexion of cervical spine was proven.

The same machine as by Fukushima *et al.* (2006) was used by Ivancic (2013), but instead of living probands he used cadaver's head with cervical spine fixed to a mannequin. He aimed for comparison of results with measurements *in vivo*. The results showed that the rear-oriented force causes complex strain of head, neck and surrounding structures. Maximum force was 249 N with 3.6 kg weight and 504 N with 16.7 kg weight. Impact duration was 62 ms for lower weight and 96 ms for higher weight.

Contact impacts in sports were an area of interest for Wilcox *et al.* (2013). He was gathering data from one-axis accelerometers fixed to hockey players helmets. Maximum acceleration during impacts varied between 20 and 120 *g*.

Walilko *et al.* (2005) was dealing with head injuries in highly contact sport – box. The experiment was carried out on Hybrid III mannequin with frontal punch (direct). Average force value measured was 3427 (SD 811) N, hand velocity 9.14 (SD 2.06) m/s, head acceleration 58 (SD 13) *g*, duration 11.4 ms and HIC 71.

Considering variability level of observed variables and multiple test types during experiments Monea *et al.* (2013). attempted to theoretically systematize the whole area. He defined three groups of contact impact injuries: cranial fractures, brain concussions and acute subdural haematomas. Based on his research he determined basic relation between mechanical impact parameters and specific head injuries. He also described typical patterns of frontal, parietal and occipital cranial fractures.

Despite this attempt in theoretical systematization it is clear that currently there is plenty of impactors and approaches to simulation of impacts to a mannequin or its parts. On the other hand huge part of research aims at testing of protective materials and props. The issue of individual's reaction to an impact has been lessened. That is why a head area impactor has been built with possible fine tuning of momentum of intended impact while offering precise detection of real impact forces.

MATERIALS AND METHODS

The set for contact impact analysis consists of an impactor, gauges, accelerometers, a mobile EMG and a Qualisys system. The impactor for controlled impacts includes an inelastic hinge on a pivot. At the end of the hinge is a calibrated weight, deviation meter for measuring the angle of hinge deviation from the vertical and an impact board with detection and protection equipment. A pressure detector is mounted to the rear side of the impact board while on the impact side there is dampening foam or another type of material with properties appropriate for inelastic collision between the board and human head. An accelerometer is mounted on the contact side of the carrier. The technological essence of safe and precise recording of an impact to a human head is creation of a detection area where various detectors can be equipped or simply interchanged. The detection area exists between the impactor carrier and the impact board where there are guiding spikes planted in the impactor board and loosely set in the carrier guides. Prior to the measurement the sitting proband is equipped with an accelerometer, Qualisys markers and EMG detectors.

A pressure detector is mounted to the rear side of the impact board while on the impact side there is dampening foam or another type of material with properties appropriate for inelastic collision between the board and human head.

There were six probands tested – men, age 24±3, weight 75±8 kg. The probands were in good health condition and never had had any cervical spine related issues. The biomechanical parameters were detected during the impactor impact to the frontal part of human head. Whole experiment was recorded by the Qualisys system (6 cameras) with recording frequency of 1 000 Hz and by a high speed digital camera. Kinematic analysis of head trajectory data during impact was carried out in Qualisys Track Manager interface. Furthermore the force acting towards the head via the impactor was recorded using the Dewetron technology with Kistler detectors with 1 000 Hz recording frequency as well as acceleration of the impactor and the proband's head in three axes using four accelerometers fixed on the top, rear and sphenoid areas of the head. Recorded data was processed in HyperGraph software, filtered according to Euro NCAP (European New Car Assessment Programme) methodology using CFC 1000 (EuroNCAP 2011) filter and rectified to resulting accelerations. An HIC_{36} (Head Injury Criterion) and 3 ms criterion were used for comparison of head injury severity. Muscular activity was measured by mobile EMG synchronized with accelerometer system and recording neck muscles activity, i.e. right and left *m. sternocleidomastoideus* (M. SCM) and right and left *m. trapezius* (M. T). Data was processed in the HyperGraph software. Observed course was divided in two phases: phase 1 – pre-activation: 0.5 s before impactor impact and phase 2 – post-impact: 0.5 s after impactor impact.

Normalization of the EMG signal for contact impacts was performed using an accepted method of maximum volitional contraction (MVC) (Zheng *et al.* 2013; Botelho *et al.* 2011). Ten percent of MVC was used as an activation time which corresponds to more than two times of standard deviation from base value. RMS (Root Mean Square) and Mean values were used for quantification of an EMG signal. Results for RMS is adequate to Mean values according to Fanta *et al.* (2013).

The measurement itself followed the protocol below. Each proband underwent an MVC test and then three probands together without visual perception, then with visual perception and then the remaining three probands together with visual perception and then without it. Standard statistics for normal distribution on significance level of 0.05 were used during the evaluation of the results. The experiment was approved by an ethic committee. Standard statistics for normal distribution on significance level of 0.05 were used during the evaluation of the results.

RESULTS

1. Kinematics

According to the records from Qualisys the average impact velocity of the impactor was 1.97±0.15 m/s, initial conditions were equal for all the measurements. Qualisys showed good results of recording marker trails in laboratory conditions. Controlled contact impact follows (Figure 1).

In the case of expected impact was the preparation move (leaning forward) 15.15±3.49 mm (7.80±2.23°) against moving impactor, in the case of unexpected impact it was 7.72±4.81 mm (2.87±2.17°). Maximum distance of a head from the base position was 80.33±10.69 mm and deviation 21.98±2.22° for unexpected impacts, 51.53±14.25 mm and 13.23±2.54° for expected impacts (see Figure 2).

2. Accelerometrics and dynamics

From the results from a dynamometer attached to the impactor we can state that the force of impact was similar in all measurements and reached an average value of 302.6 N, p-value for t-test was 0.196. All values of injury criteria differed in attempts with and without visual perception, p-value was lower than 0.05. In the same time the regression analysis did not prove that the dependency existed between amount of force measured on the impactor and maximum value of head acceleration in all measurements. Statistically processed data is in Table 1.

It has been statistically proven that maximum value of head acceleration is lower during the second measurement with visual perception than during the first without it. With visual perception is therefore an average decrease of maximum acceleration value equal to $0.6g$, which is a 10.3% decrease, HIC_{36} decreased by 0.7 (10%) and 3-ms criterion decreased by $0.5g$ (7%). Everything is clearly visible in Figure 3.

3. Electromyography

a) Time of activation

Activation of measured muscles was observed in a time line related to the impactor impact progress. The moment of maximum force measured on the impactor was considered a reference for further measurements – the beginning, the moment zero. Activation times of monitored muscles were determined in relation to this point. Activation before the reference point is stated as a negative number, activation after it is stated as a positive number.

In the following table (Table 2) is statistical evaluation of muscle activation times of *m. sternocleidomastoideus* and of *m. trapezius* in Table 3.

Fig. 1. Contact impact situation.

Tab. 1. Statistical values of injury criteria.

	Max. head acceleration (*g*)		HIC 36		3-ms criterion (*g*)		Max. force (N)	
	No vision	Vision	No vision	Vision	No vision	Vision	No vision	Vision
Mean value	5.613	5.035	6.655	5.973	5.373	4.898	291.1	314.1
Maximum	6.270	5.910	8.630	7.250	6.130	5.630	330.4	354.4
Minimum	4.820	4.280	4.930	4.870	4.420	3.980	252.9	262.7
Median	5.610	5.145	6.660	5.865	5.370	5.045	297.7	307.3
Standard deviation	0.504	0.543	1.160	0.857	0.602	0.519	28.5	31.5
Normality test	yes	yes	yes	yes	yes	yes	yes	yes
Diff. of mean values	−0.578		−0.682		−0.475		23.0	
Percentage diff.	−10		−10		−7		8	
Pair *t*-test	0.022		0.014		0.005		0.196	
Pearson's correlation coef.	0.722	0.105	0.960	0.002	0.966	0.002	0.345	0.504

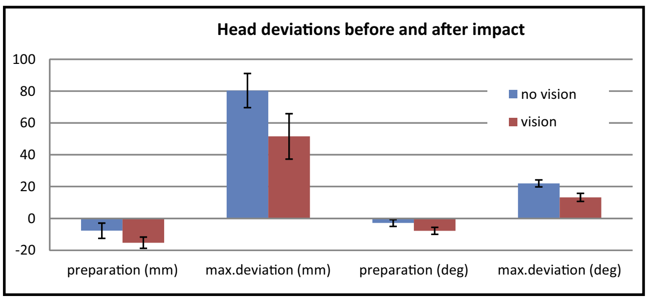

Fig. 2. Average head deviations before and after impact.

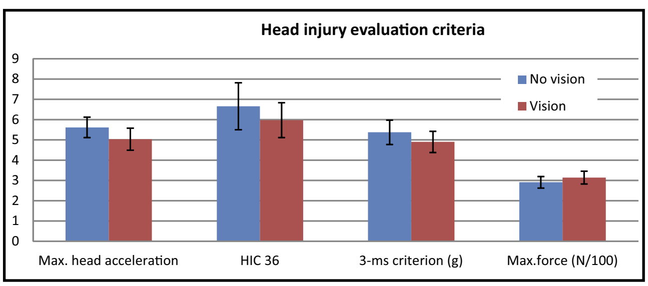

Fig. 3. The results summarized in the graph.

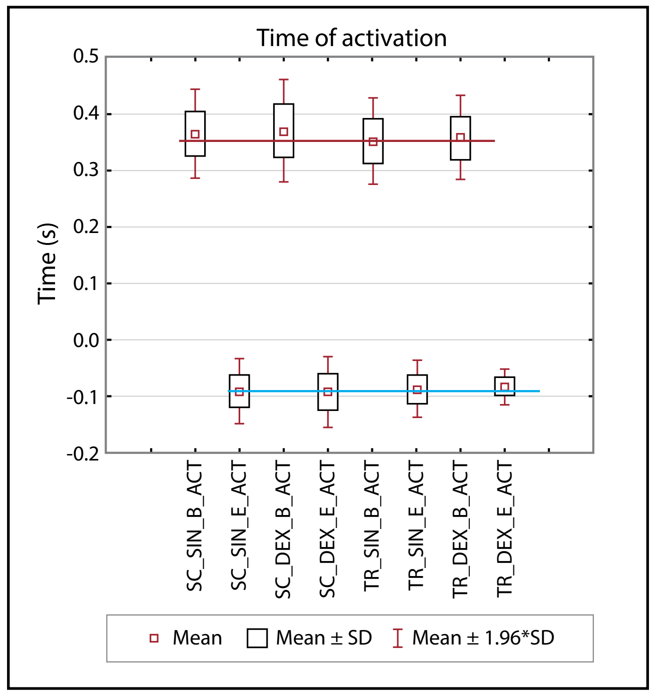

Fig. 4. Individual muscles activation.

Tab. 2. Timing of *musculus sternocleidomastoideus*.

	Left no vision	Left vision	Right no vision	Right vision
Mean value(s)	0.365	−0.091	0.370	−0.092
Maximum (s)	0.412	−0.048	0.429	−0.047
Minimum (s)	0.313	−0.133	0.303	−0.136
Median (s)	0.361	−0.089	0.370	−0.091
Standard deviation(s)	0.037	0.027	0.042	0.029
Normality test	yes	yes	yes	yes
Diff. of mean values (s)	−0.456		−0.462	
Percentage diff. (%)	−125		−124	

Tab. 3. Timing of *musculus trapezius*.

	Left no vision	Left vision	Right no vision	Right vision
Mean value(s)	0.352	−0.087	0.358	−0.082
Maximum (s)	0.405	−0.043	0.415	−0.053
Minimum (s)	0.315	−0.117	0.324	−0.098
Median (s)	0.342	−0.093	0.352	−0.086
Standard deviation(s)	0.036	0.023	0.035	0.015
Normality test	yes	yes	yes	yes
Diff. of mean values (s)	−0.440		−0.440	
Percentage diff. (%)	−125		−123	

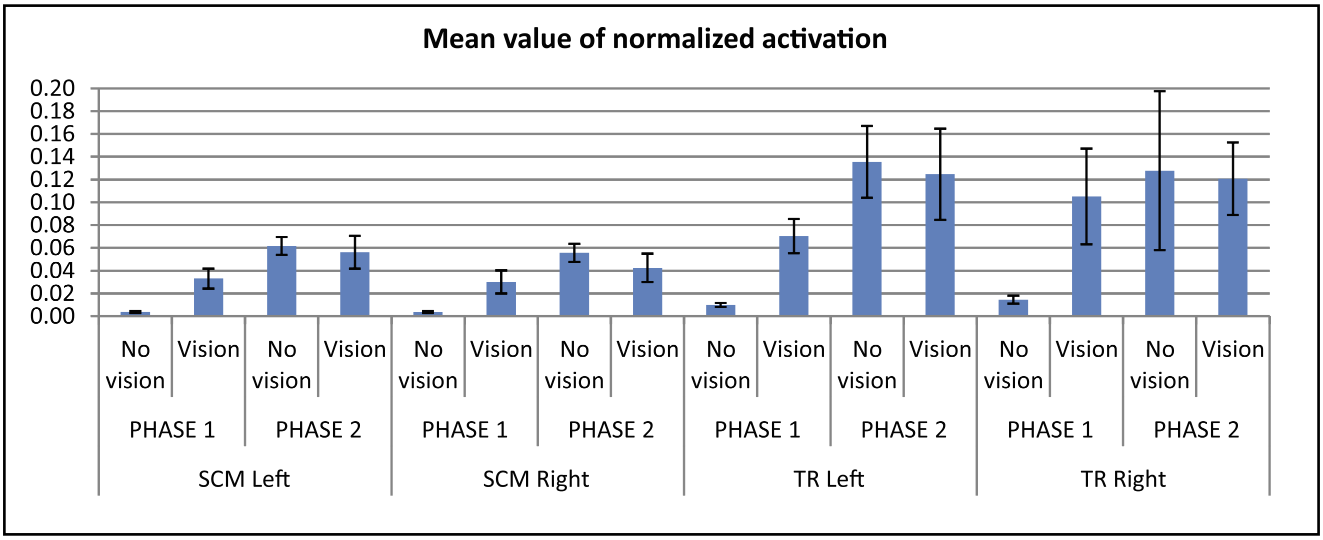

Fig. 5. Mean values of normalized activation for individual muscles and segment.

After summing up the results we can state that without vision the activation of muscles occurs 0.355 s after the impactor impact, with vision it is 0.0845 s before the impact. Activation times of individual muscles are displayed in the box graph (Figure 4.) (SC = Sternocleidomastoideus, TR = Trapezius, SIN = Left, DEX = Right, B = No vision, E = Vision) along with their average value. No vision impacts – red line, with vision impacts – blue line.

b) Quantification of activation

There have been mean values calculated from the measured courses of activation of monitored muscles in defined segments, which have been further averaged for the final comparison and the differences have been determined between impacts with and without vision. Percentage differences have been calculated in relation to values without vision.

From the differences of mean values there are significant differences apparent in activation of monitored muscles in individual evaluated segments between allowed and forbidden visual perception, when positive values mean higher activation with vision and negative mean higher activation without vision. Similarly to previous phase of the evaluation the stated conclusion is well apparent in graphical representation (Figure 5), where there is average of mean values of activation of all probands in given segment on the vertical axis.

It is therefore apparent that in phase 1, i.e. before impact in preactivation phase, is the activation with visual perception statistically more significant (p-value <0.05) in all muscles. Conversely in phase 2, after impact, is the activation lower with visual perception. Statistically significant (p-value <0.05) value has been reached only for the right M. SCM. There was fairly large standard deviation in other muscles in this regard and thus even though the values decreased, it cannot be verified statistically. In comparison of muscles by size the M. T. acted more than M. SCM. in relation to MVC and in both muscles the left sides were slightly more active. If we therefore consider a response of the whole monitored group, the following table can be assembled (Table 4.) with average values of percentage difference between allowed and forbidden visual perception.

Tab. 4. Activation differences with and without visual perception.

	Section 1	Section 2
Mean values difference	0.052	−0.009
Percentage difference	745	−90

DISCUSSION

Results of the head movement process can be described as deliberate lean forward during foreseen impact which can be 100% further than during unforeseen impact. This activity can be called preparation for impact.

When impacting by means of solid object to the skull, there is a spreading pressure inside a skull. Over a time the pressure wave reflected from the back wall of the skull. Cranial parenchyma is easily damageable by a tension compared compression. Region on the opposite site of the skull against a stress point has the character more damage than the lesion site. The tolerance of the organism to strike effect is dependent on the size of the intracranial pressure. The publications (Shardlow *et al.* 2011; Shardlow & Jackson 2011; Sahoo & Agrawal 2013) it is clear that the critical threshold for intracranial pressure >60 mmHg.

Another control mechanism, which can affect the kinematics and dynamics of the involvement of the cervical muscles after impact by a co-contraction of antagonists. The importance of this reaction is seen in the protective function against overload the musculoskeletal system due to intense muscle contraction with agonists (Choi 2003).

According to conclusions of Muggenthaler *et al.* (2008) is the head deviation from the initial position smaller for preactivated muscles and the initial head movement was not observed during tests with preactivated muscles. Main head and C_7 deviations were almost identical for preactivated muscles whereas for relaxed muscles were the head deviations twice as big as vertebral deviations (Muggenthaler *et al.* 2008). This has been confirmed by our results as well because maximum distance of a head from an initial position is 80.33±10.69 mm and deviation is 21.98±2.22 ° for unforeseen impacts. Foreseen impacts have 51.53±14.25 mm distance and 13.23±2.54 ° deviation. Tests based on Muggenthaler *et al.* (2008) further showed more pronounced differences typical for kinematics with dependency on initial state of a muscle and the most significant differences were between tests with relaxed and preactivated muscles.

From graphic interpretation of our acceleration progress can be stated that a head reaches it maximum acceleration before reaching maximum impact force. From the perspective of head/brain damage possibility is the first phase of the movement, when a head has the best conditions for acceleration, the most critical. Then the acceleration decreases but a risk of cervical spine injury increases, physiological head deviation is reached and the biggest resistance comes from neck vertebrae and the head-affecting force culminates. These conclusions are in disagreement with statements by Verschueren *et al.* (2007) that there is no connection between maximum force and head movement to the back, because the maximum value is reached before the head moves back. After the calculation and comparison with the head injury criteria is it clear that expecting the impact has positive influence on these criteria.

It is also apparent that if the human knows exactly the moment of impact he is capable of preparing for it and his muscles are certain level of activation (contraction) in the moment of impact. It is necessary to realize further rules like the 0.2 s human reaction time to an action. Preparation for impact with visual perception thus begins approximately 0.5 s before the impact itself. Another interesting aspect is comparison of muscle activation time and reaching maximum head acceleration and force values in relation to head‘s back motion. The results conclude that due to a timely activation during an impact with visual perception a head reaches lower deviation value (28.8 mm in average) which is positive according to possible cervical spine injuries, possible whiplash injury and lower head trauma values (0.68 in HIC in average).

Considering the graph Figure 5 and Table 4 it can be stated that visual perception results in very significant preactivation increase in monitored muscle group by almost 750% and conversely lower activation by 90% in following post-impact phase which makes an early warning critical for decreasing the injury severity.

CONCLUSION

It can be stated that a body reaction significantly influences head movement kinematics for relatively weak impacts. Due to timely preactivation of neck muscles during expected impacts the deviation can be minimized and head injury criteria decreased. Proband reacts to incoming impact with neck muscles preactivation and during compact impacts moves the head towards the impactor which positively affects the monitored parameters. Body reactions influence motion kinematics before impact, after impact, it affects injury evaluation criteria and also influences time and amount of muscle activation.

Results from measurements on living probands create a good base for implementing an actual human reaction in active human models in simulation software so that it could be possible to determine internal biomechanical response of a head and whiplash injury probability thanks to the validated results. It is nowadays an essential question for forensic biomechanics and judge advocates in medicine area especially in relation to insurance frauds and made-up consequences of traffic accidents. There are however other areas that can benefit from improvement of whiplash qualification factors, especially in research of its effect on brain, CNS or endocrine glands. In these areas the information about muscle preactivation and activation in certain conditions together with different head acceleration values can serve as a parameter that would allow more precise qualification of research groups.

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