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Influence of Anthropometry on the Kinematics of the Cervical Spine in Frontal Sled Tests

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Abstract

Objective: The objective of this study was to investigate the influence of anthropometric factors on the kinematics of the cervical spine during in-vivo frontal collisions.
Methods: Therefore a frontal collision with a mean velocity change (delta-V) of 10.2 km/h was simulated in a sled test with ten healthy volunteers (seven men and three women). A high-speed camera was used to document motion data. Acceleration data were recorded using accelerometers. The study analyzed the association of anthropometric factors with defined kinematical characteristics.
Results: A smaller neck circumference led to an earlier peak of the dorsal horizontal head acceleration (r=0.602), an earlier beginning of the ventral head translation (r=0.742) and a greater maximal head flexion (r=-0.717). A smaller body weight led to a later beginning of the head flexion acceleration (r=-0.713) and a greater maximal head flexion (r=-0.620). With a smaller thorax circumference the beginning of the dorsal horizontal head acceleration (r=0.623), the peak of the head flexion acceleration (r=0.756) and the peak of the head extension acceleration (r=0.679) were reached earlier.
Conclusions: The main findings of the present study consist in the identification of relevant anthropometric parameters (neck and thorax circumference and body weight) on the cervical spine kinematics. Specific anthropometric factors increasing the risk of injury could not be identified. The head movement is mainly associated with the neck circumference and the body weight. The onset and occurrence of the acceleration parameter is mainly associated with the thorax circumference.

Keywords
frontal impact, anthropometric factors, head-neck kinematics, volunteer sled tests, whiplash

Introduction

The neck kinematics has been investigated in detail in numerous sled tests [1-4] mainly for rear-end collisions. These studies have shown that the kinematics of the cervical spine is dependent on various external influence factors, like the crash impulse, the acceleration and velocity change [5-7] and the seat and head restraint properties [1,8].

Also the influence of anthropometric factors was increasingly recognized. Epidemiological studies have found that women suffer whiplash more frequently than men [8,9]. A comprehensive study by Siegmund et al. [10] investigated the influence of individual anthropometric factors on the kinematical reaction and risk of injury during simulated rear collisions. They could show in a kinematics study that some kinematic parameters concerning the peak amplitude and time-to-peak-amplitude of the head acceleration and head motion varied significantly with anthropometric parameters.

Although the accident mechanism operating during frontal collisions has been investigated in the last fifteen years in more detail [11-13], the influence of occupant anthropometry on the physical response of vehicle occupants in frontal collisions is mainly unknown. Links for the relevance of anthropometric influence parameters can be found in two former studies. Knox et al. [14] investigated indirectly the seat belt and air bag effect in high speed frontal crashes using mathematical dynamic modelling software. They found that that a properly timed air bag deployment reduced injury potential for all occupants of all sizes, but that the magnitude of this benefit was dependent on anthropometry. In another

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computer based model Armstrong et al. [15] found that the occupant posture was the most significant parameter affecting the overall risk of injury in frontal collisions.

As the first part of this study has investigated the muscle activity influence on the kinematics of the cervical spine [13], the second part of this study aimed to investigate the influence of anthropometric characteristics on the kinematics of the cervical spine during a frontal sled crash test. Ideally this information should serve as helpful hint in identifying possible anthropometric characteristics leading to an increased risk of injury during frontal collisions.

Methods

Subjects

The test procedure is analogous to the already for publication accepted article investigating the muscle activity influence on the kinematics of the cervical spine [13]. The work has been approved by the ethical committee of the University of Ulm. The subjects gave informed consent to participate in the study. Ten subjects (seven men, three women) aged 20 to 47 years (median: 35 years) without prior structural injuries to the spine participated in the study. Exclusion criteria were a history of whiplash injury of the cervical spine, neurological or psychiatric disease, functional impairments of the cervical spine or cervical spine pain.

Prior to the sled tests, subjects underwent clinical examination with determination of individual maximal cervical spine mobility. As anthropometric characteristics the head measurement (head circumference), the neck measurements (neck length and circumference) and the body measurements (thorax circumference, body height, seated height and body weight) were determined (see Table 1).

Experimental Design

For the frontal collision simulation we used a standard automobile seat (VW Passat, 1997 model, VW corporation, Wolfsburg, Germany) anchored to a target sled platform (see Figure 1). The seat sled was accelerated over a length of 20 meters towards the fixed iron barrier. Measurement of the sled acceleration was performed using a sensor (Endevco 2262, +/- 200g, uniaxial x-direction, CFC 60, Endevco Corporation, San Juan Capistrano, USA). The sled acceleration is characterized by a trapezoid impulse. The change of speed Δv was calculated by integration on the basis of the CFC180 filtered sled acceleration. The mean acceleration of the seat sled was 2.68 g (2.45-3.27 g), the mean duration of acceleration was 106 ms (100-112 ms) and the mean velocity change was 10.2 km/h (9.9-12.7 km/h).

An H-point dummy was used for seat adjustment and a 25° backrest angle was ensured before each test. After positioning the subjects on the test sled, head restraints were adjusted so that the upper edge of the head restraint was aligned with the vertex of the head of each subject (see Figure 1). A horizontal adjustment was not possible. The initial horizontal distance from the head to the head restraint ranged from 40 to 90 mm (median 60 mm). In addition, all subjects were secured with a three-point seat belt in passenger position. It is made up of a lap belt with buckle and a shoulder belt, which is adjusted to fit the subjects. The seat belt is equipped with a webbing-sensitive retractor that stops the belt from extending off the reel during severe deceleration. The subjects were then instructed to sit in the target sled without changing their initial position until the impact has been occurred and to stay in the target sled until the test has been fully completed.

Measurement Technique, Data Recording and Processing

The data mentioned below were recorded for all subjects from –800 ms to +800 ms, with “0” defining the time of the trigger signal, at the moment of the initial contact between the sled and the barrier. Data recording and processing according to SAE J211/1 (SAE 1995) was performed with Diadem® 8.0 (National Instruments Germany GmbH, Munich, Germany).

Motion Data

The experiments were recorded with a stationary LOCAM high speed camera (Visual Instrumentation Corporation, USA) and subsequent digitized with 100 images/s for the first 300 ms. The motion data were documented based on markings of the centre of gravity of the head, which is defined as the surface projection ca. 1.5 cm ventrally and cranially to the most cranial point of the external acoustic meatus [16,17], the Frankfort-plane (defined as the inferior margin of the osseous orbit and the upper margin of the external acoustic meatus) and the first thoracic vertebra. The target coordinates were all expressed in a sled-related coordinate system (x-axis (positive forwards), the y-axis (positive to the left) and the positive z-axis extends perpendicularly upwards). The data were further smoothed prior to further processing, using a third order spline which was least square error optimized. The relative horizontal head motion is calculated by the difference between the head and T1 motion in x-direction.

Accelerations

To measure the angular head acceleration a rotation rate sensor (Endevco 7302, up to 5000 rad/s², piezoresistive, CFC1000 (Endevco Corporation, San Juan Capistrano, USA)) was fixed in a position close to the projected centre of gravity of the head. The angular acceleration cannot be used in the form of raw data, as individual values of more than 1300 rad/s² appeared as spikes in the curve. Moving average smoothing was applied across 20 values.

In order to measure the T1 acceleration, a two-axial linear accelerometer (Endevco 7264, +/- 200 g, CFC180 (Endevco Corporation, San Juan Capistrano, USA)) was mounted on a pliable metal plate, which was padded with tape, adjusted to the contour of the subject's back and attached directly to the skin above the spinal process of the first thoracic vertebra. The starting position of the T1 sensor above the spinal process of the first thoracic vertebral body defined the sensor-related coordinate system. The positive x-axis pointed in ventral direction, perpendicular to the body surface, and the positive z-axis, which was perpendicular to the x-axis pointed in cranial direction. Zero compensation was performed when the measurement was recorded, which reduced the gravity component to 0.

The measurement of the horizontal head acceleration was performed using a three-axial linear accelerometer MSC 123 sensor (+/- 100 g, CFC1000 (Micro-epsilon Messtechnik GmbH & Co. KG, Ortenburg, Germany)). The accelerometer was attached to the subject via a head harness with which the sensor could be positioned as close as possible to the projected centre of gravity of the head. Analogue to the motion data the anatomical head coordinate system was based on the Frankfort-plane. The relative acceleration between the head and T1 in x-direction was calculated from these accelerations. The maximal seat belt effect was defined by maximal negative peak of the thoracic acceleration.

Analysis

The analysis of the motion and acceleration curves was performed descriptively. Generally, the start of acceleration or motion was defined as the time at which 10% of the subsequent maximum/minimum was reached or the zero-crossing when a change of sign occurred. The following biomechanically relevant parameters of the motion and acceleration curves were ascertained chronologically and assigned to the kinematic phases (latency phase, translation phase, flexion phase, rebound phase) (see definition in Table 2 and 3).

The Spearman rank correlation coefficients (r) are used for the explorative analysis of the associations. If the coefficients are above +/-0.6, they are presented. Normally values under -0.5/-0.6 and above 0.5/0.6 are counted as statistical clearly recognizable correlation. Unfortunately standard guidelines in the literature don't exist. As the thresholds are floating and the number of cases investigated in this study is relatively small, we decided to choose the higher threshold with 0.6. No multivariate analysis is performed because of the small sample size. Thus, the calculated significance values without Bonferroni adjustment should only be seen as an indication of associations.

Results

Analysis of the motion and acceleration parameters

The curve progression of the motion and acceleration parameters was reproducible for each individual subject (see Table 4 and 5, Figure 2). In the latency phase (0ms-44ms) the head remained in its initial position. The head flexion acceleration started after 31ms (21-46ms). In the translation phase (44ms-68ms) the ventral head translation began without a rotational component 44ms (25ms-64ms). The dorsal horizontal head acceleration started after 49ms (26-75ms). In the flexion phase (68ms-196ms) the isolated translational motion was followed by the initiation of ventral flexion of the head after a median of 68ms (60ms-80ms). The maximal braking effect of the seat belt occurred after a median of 90ms (81ms-99ms). Furthermore the head flexion acceleration reached its maximum (median: 177rad/s², 110-377rad/s²) after a median of 96ms (73ms-138ms) and the maximal dorsal horizontal head acceleration (median: 5.0g, 4.1-6.2g) after a median of 106ms (97ms-134ms). This was followed by the events of maximal head extension acceleration (median: 260rad/s², 231-425rad/s²) after a median of 149ms (144ms-164ms) and the minimum of dorsal horizontal head acceleration (median: 1.2g, 0.4-1.7g) after a median of 156ms (143ms-168ms). After a median of 196ms (175-233ms) the maximal ventral head translation was reached (median: 112mm, 62-132mm) during the rebound phase (196ms-300ms). Furthermore the head reached the maximal head flexion (median: 36.2°, 21.2-48.2°) after a median of 223ms (192-239ms).

Association between anthropometry and kinematics

Angular head movement

A smaller body weight (r=-0,620) and a smaller neck circumference (r=-0,717) led to a greater maximal head flexion (see Table 6).

Relative horizontal head movement

A smaller neck circumference (r=0,742) was associated with an earlier beginning of the ventral head movement (see Table 6). Concerning the movement amplitudes no association to the anthropometric data could be found.

Angular head acceleration

The smaller the body weight, the later the head flexion acceleration began (r=-0,713) (see Table 6). With a smaller body height (r=0,622) and a smaller thorax circumference (r=0,756) the maximal head flexion acceleration was reached earlier. With a smaller thorax circumference also the maximal head extension acceleration (r=0,679) was reached earlier. Concerning the acceleration amplitudes no association to the anthropometric data could be found.

Relative horizontal head acceleration

A smaller thorax circumference led to an earlier beginning of the dorsal horizontal head acceleration (r=0,623) (see Table 6). With a smaller neck circumference the maximal dorsal horizontal head acceleration was reached earlier (r=0,602). Concerning the acceleration amplitudes no association to the anthropometric data could be found.

Discussion

The present study was conducted in order to elucidate the associations between anthropometric factors and kinematical characteristics of frontal collisions. Sled tests with human subjects have not, to date, been adequately employed in describing the kinematical processes occurring during frontal collisions. Previous studies have in most cases failed to provide an exact characterization of the biomechanical mechanisms [18,19]. It has simplifying been assumed that a hyperflexion movement of cervical spine occurs during crashes of this kind. Evidence that the kinematical reaction of the cervical spine is more complex than initially assumed has been provided by the study findings which are already accepted for publication [13].

In summary the kinematical analysis shows that a ventral translation movement begins at about 40ms, followed at about 70ms by a flexion movement. No significant dorsal translation or extension movement was observed. The study by Kumar et al. [12], which did not report movement data, documented the onset of head acceleration after 35.6ms with a mean sled acceleration of 1.4g. These data approximate findings of the present study regarding the onset of ventral head angle acceleration (31ms) and of dorsal horizontal acceleration of the head relative to T1 (49ms).

Unlike rear collisions the head-to-headrest distance is not associated with any effect on cervical spine kinematics [20,21]. Beside sled acceleration and change in velocity of the test sled, the restraint provided by the seat belt represents an important external parameter. For example, Siegmund et al. [22] found that maximum motion and acceleration values differed widely in relation to anchoring and tension of the seat belt. In their study, lax belt slack was associated with an increase in both ventral thorax movement and ventrally directed acceleration amplitudes.

Most correlations between the kinematics and the anthropometric factors have been found for the neck and thorax circumference and the body weight. Head circumference and neck length, however, were not found to affect cervical spine kinematics. By contrast, it was these characteristics that were shown to be the crucial parameters affecting cervical spine kinematics in rear-end collisions [10]. This difference could be explained by the different kinematic characteristics in frontal and rear-end collisions. As the kinematics in rear-end collisions are characterized by an initial extension and subsequent flexion of the head relative to the cervical spine the existing rotational movement plays an important role. Probably this rotational movement of the head relative to the cervical spine is getting greater, with a greater head circumference and neck length. In contrast to that these influence factors are less important in frontal collisions with an isolated flexion movement of the head.

Neck circumference correlates primarily with head movement. Smaller neck circumferences are associated with an earlier onset of ventral head translation relative to T1. This could be explained by the fact that a smaller neck circumference could be associated with less tissue resistance and therefore results in an earlier beginning of the relative movement and acceleration between the head and T1. Another possible explanation could be the fact that a smaller neck circumference leads to a better measurable differentiation between the head and T1 position in horizontal direction and therefore is simply a metrological phenomenon. Smaller neck circumferences are further associated with the development of a greater flexion amplitude. This may be explained by the fact that a smaller neck circumference could be associated with less muscle tissue, which in turn could lead to less active muscle resistance to the flexion movement. Furthermore a smaller body weight was associated with greater flexion amplitude but a later beginning of the head flexion acceleration. It is conceivable, that greater flexion amplitude could lead to a higher risk of injury, but this assumption couldn’t be proved in this investigation.

Smaller thorax circumference is associated with the earlier onset of the dorsal horizontal acceleration and the occurrence of the maximal head flexion and extension acceleration. This means that, with smaller thorax circumferences, there is a more rapid change in the forces of acceleration operating on the body, resulting in a larger impulse. This may be explained by the fact that, with increasing thorax circumference – and probably an increasing body weight –, there is increasing delay in the onset of the maximal seat belt effect. Although this assumption is the most reasonable, it couldn’t be directly proved in this investigation.

One limitation of the study is the small number of subjects, which makes it difficult to achieve statistically strong data. Up to now comparison studies of the influence of anthropometric parameters on cervical spine kinematics in frontal collisions have not been published in the literature. Hence, the findings of the present study should be considered pilot data requiring targeted evaluation in further kinematical studies. As further limitation the test setting represents only approximately the real crash situation. As the subjects are prepared for the upcoming crash event, cervical spine kinematics may be altered by pre-activated muscular influence. Concerning the risk of injury it is not possible to definitively determine which anthropometric factor is associated with an increased risk of injury, as multiple factors are involved in the kinematical response of the subjects.

The main findings of the present study consist in the identification of relevant anthropometric parameters (neck and thorax circumference and body weight). A smaller neck circumference and a smaller body weight are associated with greater flexion amplitude. A smaller thorax circumference is associated with an earlier onset of the dorsal horizontal acceleration, the head flexion acceleration and the head extension acceleration, which could be summarized in resulting in a larger impulse to the torso during the frontal impact. These parameters are distinct from those relevant in rear-end collisions (head circumference and neck length), which could be explained by the different kinematic characteristics in frontal and rear-end collisions.

Acknowledgements

The experiments comply with the current laws of Germany and were performed with approval of the local ethics board.

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Figures

Fig. 1: Side view of test sled

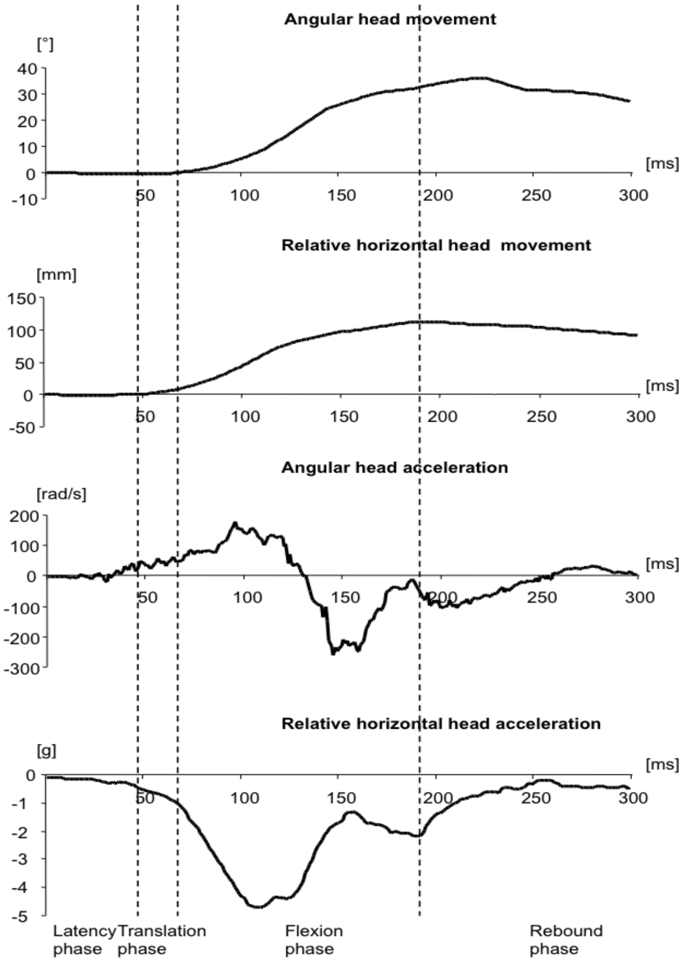

Fig. 2: Mean curves of the motion and acceleration parameters

Table 1: Subjects' anthropometric characteristics

Parameter	Definition	Median	Min	Max
Head measurement				
Head circumference [cm]	Head circumference 1cm parallel cranial to the Frankfurt plane *	**58**	53	61
Neck measurements				
Neck length [cm]	Distance between the Protuberantia occipitalis externa and C7 in 0° position	**16**	13	19
Neck circumference [cm]	Average neck circumference on the level of the Protuberantia occipitalis externa and C7	**40**	26	49
Body measurements				
Thorax circumference [cm]	Thorax circumference on the level of the inferior margin of the sternum	**102**	80	112
Body height [m]		**1.80**	1.70	1.91
Seated height [m]		**0.92**	0.82	1.05
Body weight [kg]		**83.5**	61.0	110.0

* Frankfurt plane (auriculo-orbital plane): a plane passing through the inferior margin of the left orbit and the upper margin of each ear canal)

Table 2: Definition of the motion phases

Motion phase	Parameter	Definition
Latency phase		
Start	Start of sled acceleration	Trigger signal at sled impact
End	Start of ventral translation	10% of the maximal head translation
Translation phase		
Start	Start of ventral translation	10% of the maximal head translation
End	Start of flexion	10% of the maximal head angle
Flexion phase		
Start	Start of flexion	10% of the maximal of head angle
End	Time of maximal ventral translation	Maximal head translation
Rebound phase		
Start	Time of maximal ventral translation	Maximal head translation
End	Time of maximal flexion	Maximal head angle

Table 3: Definition of the kinematic data and the seat belt effect and the time of occurrence

Parameter	Definition	Phase
Angular head movement		
Beginning of head flexion	10% of maximum	Flexion
Maximal head flexion	Maximum	Rebound
Relative horizontal head movement		
Beginning of ventral head translation	10% of maximum	Translation
Maximal ventral head translation	Maximum	Rebound
Angular head acceleration		
Beginning of head flexion acceleration	10% of maximum	Translation
Maximal head flexion acceleration	Maximal positive peak	Flexion
Beginning of head extension acceleration	Zero crossing	Flexion
Maximal head extension acceleration	Maximal negative peak	Flexion
Relative horizontal head acceleration		
Beginning of dorsal horizontal head acceleration	10% of maximum	Latency
Maximal dorsal horizontal head acceleration	Maximal negative peak	Flexion
Minimal dorsal horizontal head acceleration	Relative minimal negative peak	Flexion
Braking effect of the seat belt		
Maximal thoracic deceleration	Maximal negative peak	Flexion

Table 4: Time of occurrence of the motion and acceleration parameters in [ms]

Parameter	Median	Min	Max	Mean	SD
Angular head movement					
Beginning of head flexion	68	60	80	69	6
Time of maximal head flexion	223	192	239	212	19
Relative horizontal head movement					
Beginning of ventral head translation	44	25	64	43	12
Time of maximal ventral head translation	196	175	233	203	21
Angular head acceleration					
Beginning of head flexion acceleration	31	21	46	32	8
Time of maximal head flexion acceleration	96	73	138	102	21
Beginning of head extension acceleration	131	120	144	131	7
Time of maximal head extension acceleration	149	144	164	152	6
Relative horizontal head acceleration					
Beginning of dorsal horizontal head acceleration	49	26	75	53	16
Time of maximal dorsal horizontal head acceleration	106	97	134	108	10
Time of minimal dorsal horizontal head acceleration	156	143	168	158	7
Seat belt effect					
Time of maximal thoracic deceleration	90	81	99	90	10

The median value, minimum value (min), maximum value (max), mean value (mean) and the standard deviation (SD) are for each parameter presented.

Table 5: Amplitudes of the motion and acceleration parameters

Parameter	Median	Min	Max	Mean	SD
Angular head movement (°)					
Maximal head flexion	36.2	21.2	48.2	35.0	7.1
Relative horizontal head movement (mm)					
Maximal ventral head translation	112	62	132	110	20
Angular head acceleration (rad/s²)					
Maximal head flexion acceleration	177	110	367	207	69
Maximal head extension acceleration	260	231	425	308	82
Relative horizontal head acceleration (rad/s²)					
Maximal dorsal horizontal head acceleration	5.0	4.1	6.2	5.1	0.6
Minimal dorsal horizontal head acceleration	1.2	0.4	1.7	1.1	0.4

The median value, minimum value (min), maximum value (max), mean value (mean) and the standard deviation (SD) are for each parameter presented.

Table 6: Associations of anthropometry with the kinematic data

Parameter	Head circumference	Neck length	Neck circumference	Body height	Seated height	Body weight	Thorax circumference
Angular head movement (°)							
Maximal head flexion	o	o	-0,717	o	o	-0,620	o
Relative horizontal head movement (mm)							
Beginning of ventral head translation	o	o	0,742	o	o	o	o
Angular head acceleration (rad/s²)							
Beginning of head flexion acceleration	o	o	o	o	o	-0,713	o
Time of maximal head flexion acceleration	o	o	o	0,622	o	o	0,756
Time of maximal head extension acceleration	o	o	o	o	o	o	0,679
Relative horizontal head acceleration (rad/s²)							
Beginning of dorsal horizontal head acceleration	o	o	o	o	o	o	0,623
Time of maximal dorsal horizontal head acceleration	o	o	0,602	o	o	o	o

Spearman correlation coefficients |r|>0.6 are presented.