Influence of the Body and Neck on Head Kinematics and Brain Injury Risk in Bicycle Accident Situations

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

Abstract Previous studies about the influence of the neck on head kinematics and brain injuries have shown different results. Today bicycle helmets are certified with only a headform in radial experiments but could be improved with oblique impacts. Then the question is how the helmet's performance will be affected by the neck and the rest of the body. Therefore, the objective of this study was to use finite element simulations to investigate the influence of the body on head kinematics and injury prediction in single bicycle accident situations with and without a helmet. The THUMS-KTH model was used to study the difference between head only and full body. In total, a simulation matrix of 120 simulations was compared by altering initial impact posture, head protection, and muscle activation. The results show that the body in impacts against a hard surface can change the amplitudes and curve shapes of the kinematics and brain tissue strain. The study found an average ratio between head only and full body for peak brain tissue strain to be 1.04 (SD 0.11), for peak linear acceleration 1.06 (SD 0.04), for peak angular acceleration 1.08 (SD 0.09) and for peak angular velocity 1.05 (SD 0.13).

Keywords bicycle, brain injuries, head, helmet, neck

I. INTRODUCTION

Every year, approximately 10 million people sustain traumatic brain injuries (TBIs) that result in mortality or hospital admissions [1]. Many of these victims live with a disability due to the TBI [2-3]. There are continuous efforts to understand and prevent TBIs throughout the world by developing and improving the design of surroundings, safety products, medical diagnostic tools and medical treatments. Experimental tests and computer models are two tools that are used to better understand and improve safety products for TBI. For example, bicycle helmets are certified by experimental tests in which the helmet is dropped vertically against a flat or curbstone impact anvils(EN1078) [4]. Whether or not these tests are realistic has been debated [5], since accident reconstructions show that oblique impacts are more common than purely vertical or radial impacts. A possible improvement is therefore to include oblique impacts in the current certification tests [6]. Many questions can be raised regarding how the oblique tests should be designed, including whether or not the body should be included. Today, helmets are tested using a free falling headform in Europe and a guided headform in the US. Both setups exclude the rest of the body. This approach of only using the head is also often applied when investigating head injuries with computer models, e.g. [7-8].

The influence of the body on the kinematics of the head has been investigated in several studies. Some studies [9–11] have used rigid body models to investigate the influence of the body during head impacts in bicycle, jockey and American football accidents. Meanwhile, [12] used FE models to study head impacts in motorcycle accidents; and a COST project [13] used helmeted, full-body Hybrid III dummies to study the influence of the body in motorcycle accidents. The previous publications [9–13] obtained differing results regarding how the body affects head kinematics. It was found in [9] that the body had little influence on linear acceleration, but had a significant effect on the angular acceleration of the head in American football impacts. It was concluded in a study [12] of motorcycle helmet impacts that the presence of the body had a significant influence on the linear and angular accelerations of the head, as well as the crush depth of the helmet liner. Also, [10] found in

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an investigation of two jockey accidents, that both the head linear and angular accelerations were affected by the rest of the body. Additionally, the results in [13] showed that the rotational acceleration differed in amplitude by about 20% when the effects of the body were considered. However, based on the reconstruction of bicycle accidents, [11] stated that if the head is the first body part to strike the impact surface, the head can be viewed as separate from the rest of the body for the first 2-3 milliseconds of an impact. These studies [9–13] have shown that the neck affects head kinematics, and that that the influence of the neck is dependent on the impact direction and the duration of the impact.

Given that a new helmet test method is under development, and that computer models are increasingly used as a tool to understand head injuries, there is a need to further understand the influence of the body in helmeted and unhelmeted head impacts. Therefore, the objective of this study was to use finite element simulations to investigate the influence of the body on head kinematics and injury prediction measures in single bicycle accident situations, with and without a helmet.

II. METHODS

In this study, the FE THUMS-KTH model [8][14-15] was used for full-body simulations, and the KTH head model [8] was used for head-only (HO) simulations. The cervical spine model consists of vertebrae, intervertebral discs, ligaments, facet joints and muscles. Discrete elements are used to model the muscles with an active and a passive part. The model lacks three-dimensional solid muscles, and flesh around the cervical spine. Therefore, only the first 15 ms of the simulations were analysed, before any contact between the chin and neck. More details about the neck model and its comparison against Post Mortem Human Subject (PMHS) experiments can be found in previous studies [14][18-19]. The KTH head model used in this study has been compared to both experimental tests of PMHS and accident reconstructions. More details about the head model can be found in previous studies [7][8][16-17].

Two different bicycle accident scenarios were simulated: one where the cyclist falls to the side, and one where the cyclist falls over the handlebars (Fig. 1). From these cases, various impact points were generated by rotating the human model around the local axes of the head, resulting in a total of 24 different impact situations. All impact situations can be visualised in Appendix 1. For the side falls, the body was rotated around the local x-axis (-30° and -60°), the y-axis (0°, 20° and 40°) and the z-axis (-30°, 0°, and 30°) (The *side fall* simulations named Xm30_* or Xm60_*). The combinations of rotations are presented in Fig. 1. The two different bicycle scenarios, each with 12 different impact situations. The left image shows side fall Xm60 (-60˚ rotation around the local head x-axis), and the right image shows 'fall over the handlebars' Y30 (30˚ rotation around the head local y-axis). . For the falls over the handlebars, the body was rotated around the local y-axis (30°, 45° and 60°), the local x-axis (0°, 30°, and 45°) and the local z-axis (0, -30 and -60°) (The 'fall over the handlebars' simulations named Y30_*, Y45_* or Y60_*). The combinations of rotations are presented in Fig. 1. The resultant velocity (6.5 m/s) with an impact angle of 45° was applied in the global negative y- and z-directions for the side falls, and in the global x-direction and negative z-direction for the falls over the handlebars.

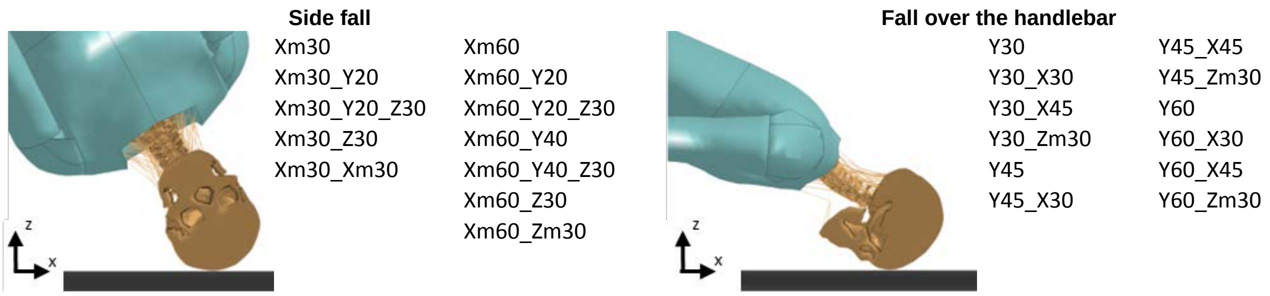

Fig. 1. The two different bicycle scenarios, each with 12 different impact situations. The left image shows side fall Xm60 (-60˚ rotation around the local head x-axis), and the right image shows 'fall over the handlebars' Y30 (30˚ rotation around the head local y-axis).

All impact scenarios were performed with no helmet, and with two different helmet designs (road and skate) (Fig. 2) for HO and FB. The road helmet is further described in a previous study [20]. The skate helmet was developed from the geometry of helmet existing on the market, and compared to experimental data from four different experimental tests for two different skate helmets. The comparison showed a correlation of 90% between the peak mean values of the experimental tests and the simulations. An 83% correlation was observed for timing of the peak values. A further description of the helmet model is presented in Appendix 2.

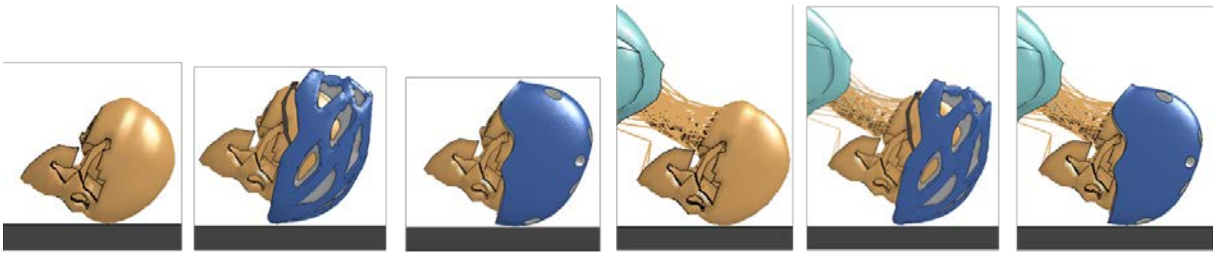

Fig. 2. HO and FB for the unhelmeted cases (first and fourth), road helmet (second and fifth) and skate helmet (third and sixth).

In all of the FB simulations presented above the neck musculature was not activated. However, a sensitivity study of the influence of muscle activation was done for all impact situations with the road helmet. The muscle activation scheme was taken from a previous study [15] that mimicked the startle reaction of a pedestrian in response to an oncoming danger, leading to a retraction of the neck with only small rotations, to a relatively stationery state. The activation was applied to the model for 120 ms, causing the cervical spine to become more compressed (Fig. 3). The new posture with the compressed neck was then used for the different impact situations presented in Fig. 1. These new impact situations with the compressed neck and simulated for HO, FB with muscle activation and FB without any muscle activation but with compressed neck.

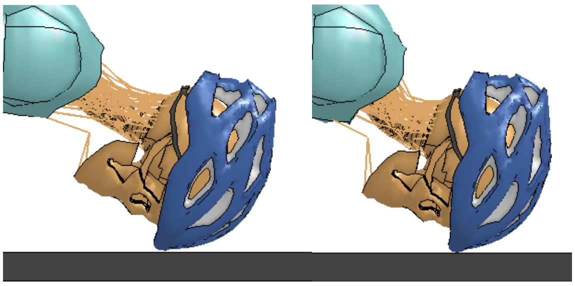

Fig. 3. The two different postures: original (left), and startle (right).

The impact surface was modelled as rigid. The coefficient of friction between the ground and helmet/scalp was set to 0.5. The same coefficient of friction was set between the helmet and scalp. All simulations were performed using the software LS Dyna (version 971 revision 5.1.1, LSTC, Livermore, CA, US). Analysis of the results was performed with LS PrePost (version 3.2, LSTC, Livermore, CA, US) and Matlab (version R2013a, The MathWorks Inc, Natrick, MA, US). The resultant head linear acceleration, angular acceleration, angular velocity and the first principal Green-Lagrange strain of the brain tissue in the FB and HO models were compared for each accident situation The results are presented primarily as ratios.. The ratio is defined as the ratio between the peak values for the different metrics of HO and FB (HO/FB). The kinematics were filtered with a SAE 300 filter before being analysed.

III. RESULTS

In Fig. 4, the kinematics and peak strain over time are shown for two examples from the 24 different impact situations (Y45_Zm30 and Xm60). Y45_Zm30 is an example with a relatively large ratio (HO/FB), and Xm60 an example with a ratio relatively close to 1. In Xm60, the shape of the curves and the peak values for HO and FB simulations are similar for both the kinematics and the first principal strain for the first 15 ms. The angular velocity plateaus after the peak values. This is representative of the majority of side falls in this study. Conversely, in Y45_Zm30 the FB curves change direction after the peak value and the resultant angular velocity decreases, whereas the HO curves plateau after the peak value. This is representative of the majority of the simulations with falls over the handlebars. The difference between the resultant angular velocities of the HO and FB simulations in falls over the handlebars is mainly due to the difference in the angular velocity about the y-axis. The shapes of the resultant linear acceleration curves for HO and FB are similar for all impact situations.

For all 24 impact locations and the three different head protections (no helmet, road helmet and skate helmet) the mean values of the ratio (HO/FB) of the peak values were: 1.06 (SD 0.04) for resultant linear acceleration, 1.08 (SD 0.09) for resultant angular acceleration, 1.05 (SD 0.13) for resultant angular velocity, and 1.04 (SD 0.11) for first principal strain.

The ratios between the peak values for HO and FB for the four metrics studied were close to one (0.98-1.07) for no helmet and road helmet, while the simulations with the skate helmet had slightly higher mean values for the different metrics (1.09-1.18). In Table I to Table IV a summary of the ratios between the peak values of HO and FB simulations for all four metrics is shown, as are the peak values. In Appendix 3 the ratio for every simulation and helmet type is shown.

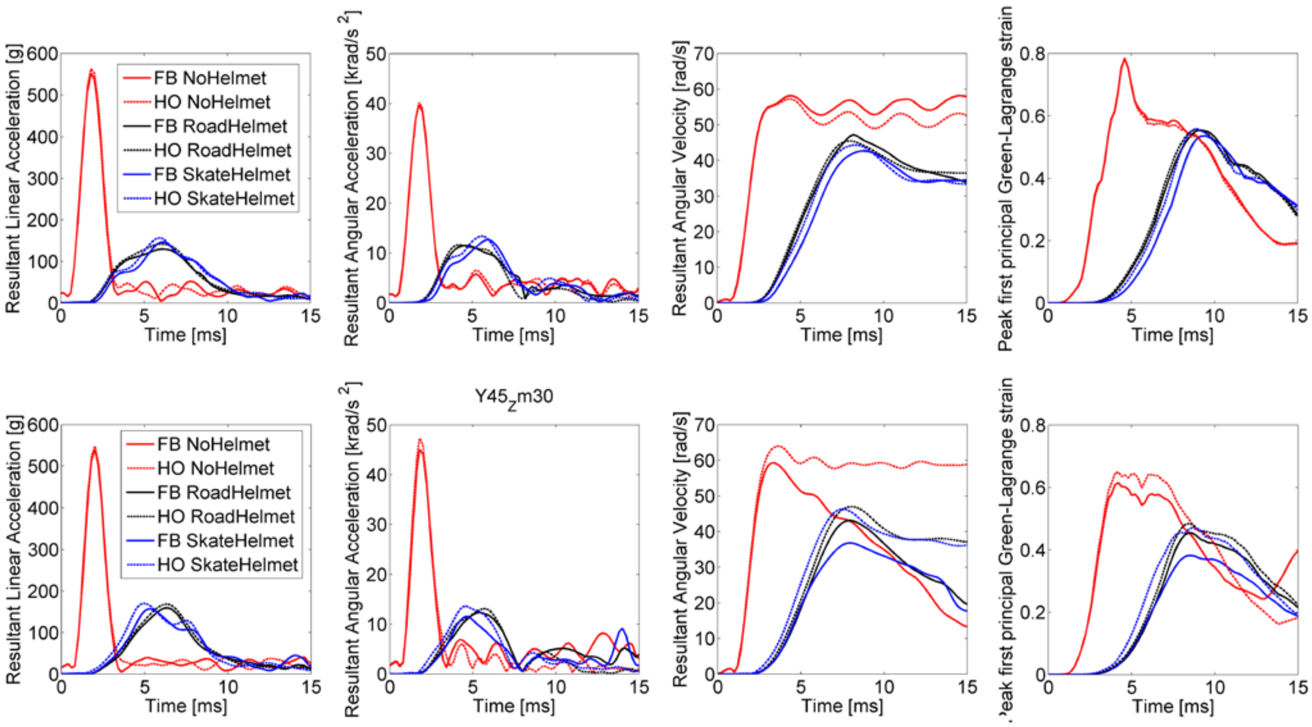

Fig. 4. Example of kinematics and strain from two impact situations with the three different head protections. The first row is for fall Xm60 (side fall) and the second row is for fall Y45_Zm30 (fall over the handlebar).

TABLE I.

Summary of the minimum, maximum, mean and standard deviation for the ratio (HO/FB) of the peak resultant linear acceleration and the magnitudes of the peak values.

	Ratio [-]			Peak value [g]					
	No helmet	Road helmet	Skate helmet	No helmet		Road helmet		Skate helmet	
				HO	FB	HO	FB	HO	FB
Min	0.99	1.03	1.04	535	526	102	94	128	115
Max	1.03	1.11	1.20	618	626	194	183	198	181
Mean	1.02	1.07	1.09	572	563	154	144	163	149
SD	0.01	0.02	0.04	26	27	20	19	16	15

TABLE II
Summary of the minimum, maximum, mean and standard deviation for the ratio (HO/FB) of the resultant angular acceleration and the magnitudes of the peak values.

	Ratio [-]			Peak value [krad/s^2]					
	No helmet	Road helmet	Skate helmet	No helmet		Road helmet		Skate helmet	
				HO	FB	HO	FB	HO	FB
Min	0.93	0.96	1.06	13.8	13.1	7.3	6.9	10.7	8.8
Max	1.07	1.16	1.33	52.0	51.7	17.9	17.0	17.2	15.2
Mean	1.02	1.05	1.18	39.0	38.4	12.5	11.8	13.6	11.6
SD	0.03	0.04	0.08	9.8	9.8	2.5	2.3	1.4	1.4

TABLE III
Summary of the minimum, maximum, mean and standard deviation for the ratio (HO/FB) of the resultant angular velocity and the magnitudes of the peak values.

	Ratio [-]			Peak value [rad/s]					
	No helmet	Road helmet	Skate helmet	No helmet		Road helmet		Skate helmet	
				HO	FB	HO	FB	HO	FB
Min	0.47	0.88	1.01	17.3	36.5	24.8	27.7	31.5	25.8
Max	1.12	1.11	1.42	64.0	60.8	54.3	53.0	51.1	48.3
Mean	0.98	1.03	1.14	53.0	53.2	46.3	44.7	45.7	40.3
SD	0.15	0.05	0.09	11.7	7.0	6.9	6.0	4.5	5.2

TABLE IV
Summary of the minimum, maximum, mean and standard deviation for the ratio (HO/FB) of the peak 1st principal Green-Lagrange strain and the magnitudes of the peak values.

	Ratio [-]			Peak value [-]					
	No helmet	Road helmet	Skate helmet	No helmet		Road helmet		Skate helmet	
				HO	FB	HO	FB	HO	FB
Min	0.52	0.93	1.01	0.16	0.32	0.28	0.30	0.35	0.29
Max	1.09	1.08	1.27	0.79	0.78	0.63	0.62	0.59	0.57
Mean	0.99	1.02	1.12	0.63	0.62	0.53	0.51	0.52	0.47
SD	0.13	0.03	0.08	0.15	0.12	0.08	0.08	0.06	0.08

The strain pattern for a sagittal section was evaluated for the two example cases with the road helmet (Xm60 and Y45_Zm30). The strain patterns had the peak values at the same location in both FB and HO, but the area with highest strain was slightly larger in the HO cases (Fig. 5).

In Fig. 6 the difference between activation and no activation of the neck musculature for two road helmet impact situations are shown. A small difference is seen for the Xm60 and a larger difference for Y45_Zm30. A summary of the difference between the ratio (HO/FB) with and without neck muscle activation is shown in Table V. The largest difference is seen for the falls over the handlebars (Appendix 4).

TABLE V

Summary of the minimum, maximum, mean and standard deviation value for the ratio (HO/FB) of the peak resultant linear acceleration, resultant angular acceleration, resultant angular velocity and 1st principal Green-Lagrange strain.

	Resultant linear acceleration		Resultant angular acceleration		Resultant angular velocity		Peak 1st principal strain	
	Activation	No activation	Activation	No activation	Activation	No activation	Activation	No activation
Min	1.11	1.04	0.89	0.99	0.62	0.67	0.77	0.81
Max	1.26	1.19	1.48	1.26	1.32	1.20	1.29	1.16
Mean	1.18	1.10	1.15	1.08	1.10	1.06	1.08	1.04
SD	0.04	0.03	0.13	0.07	0.15	0.11	0.11	0.08

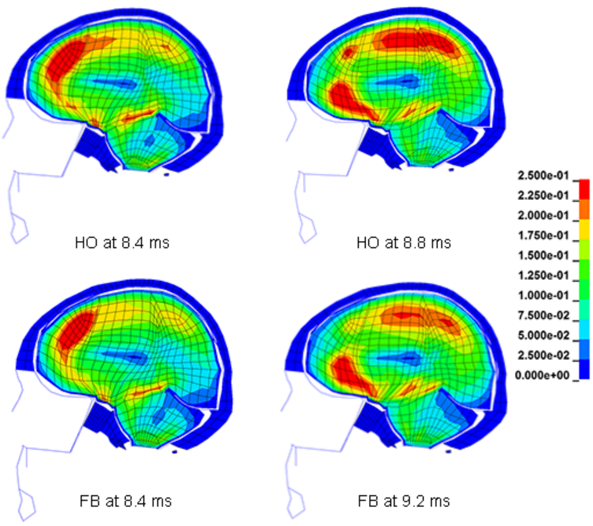

Fig. 5. The strain pattern for the road helmet simulations and impact situations Y45_Zm30 (left column) and Xm60 (right column). First row is HO and second row FB.

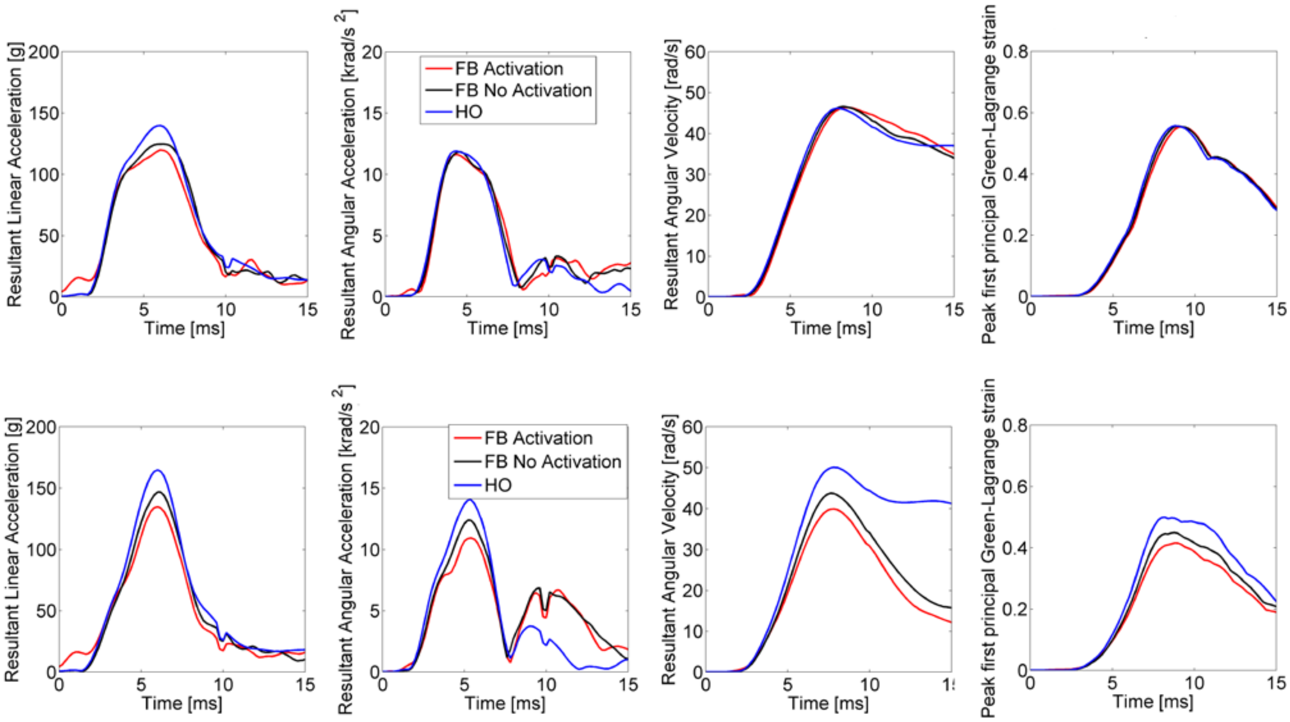

Fig. 6. Example of kinematics and strain from two impact situations for FB muscle activation, FB no muscle activation and HO (all with the road helmet). The first row is for fall Xm60 (side fall) and the second row is for fall Y45_Zm30 (fall over the handlebar).

IV. DISCUSSION

The objective of this study was to evaluate the influence of the body on head kinematics and injury predictions in single bicycle accidents using FE simulations. In total, a simulation matrix of 120 comparisons between HO and FB simulations was evaluated by altering impact situations, head protection and muscle activation. The results clearly show that the body in impacts against a hard surface can change the amplitudes and curve shapes of the head kinematics and brain tissue strain. The study also highlights the fact that many factors, such as helmet design, muscle activation and impact situation, can affect the results. The level of influence of different factors is dependent on the situation.

Influence of Helmet Design

For angular motion, in the majority of the impact situations the road helmet had a ratio (HO/FB) closer to one than the skate helmet. However, there were also impact situations where no difference in ratio (HO/FB) was seen. The skate helmets had, in most cases, a higher ratio than the road helmet. In these cases, the HO simulations showed only a minor difference in peak resultant angular velocity using the different helmet designs. In the FB simulations lower peak values were seen for the skate helmet compared to the road helmet.

The difference between the two helmet designs could depend on several factors. First, the geometry of the two helmets differ, which affects both the interaction between the helmet and the ground, as well as the interaction between the helmet and the head. The difference in geometry can influence the the direction of the load applied to the head. The geometry affects the lever arm between the impact point (where the normal force acts) and the centre of gravity (CoG) of the head, which in turn influences the rotation of the head. This also explains the low ratios for resultant angular velocity and peak strain for impacts Xm30_Y20_Z30 and Xm30_Zm30. The position of the head (Appendix 1) in these two cases orients the CoG such that rotation of the head is prevented by the tangential force. Therefore, in the HO simulations with no helmet, the head bounces primarily vertically without rotation. However, in the FB unhelmeted condition for these impact situations, the head initially bounces straight up, but then the neck affects the motion such that the rotational velocity increases with the peak value occurring at 15 ms. This explains the low peak strain values in Table IV, especially for the unhelmeted HO condition. These two cases show that in some loading conditions the body can have a large influence on kinematics and injury prediction.

A second factor that can explain the different effects of the two helmet designs is the initial position of the helmet on the head. The two helmets fit the head differently, which may affect the loading on the head. A limitation with both helmets is the lack of comfort padding, which could also affect the fit and the interaction between the helmet and the head. The interaction between the helmet and head is also affected by the coefficient of friction. Both the friction between the ground and the head/helmet, and between the head and the helmet were optimized to fit the experimental tests. The coefficient of friction between the head and the helmet could be considered low compared to the friction coefficient found in experimental tests of motorcycle helmets and artificial skin [21]. However, another study [22] concluded that the influence of a friction coefficient in the range of 0.3-0.6 was small compared to when the friction coefficient was set as low as 0.1. A sensitivity analysis in a previous study [20] showed that the peak strain varied according to the coefficient of friction, but the magnitude of variation was dependent on the impact situation. Also, a previous study [23] has shown that the difference between with or without the neck in helmets with low friction between the scalp and EPS liner was relatively small.

A third factor that can explain the different effects of the two helmet designs is that the EPS liners of the two helmets behaved differently. A larger deformation of the EPS liner was seen in the skate helmet, which increased the constraint on the head and changed the motion. For example, in the Xm30_Y20_Z30 impact with the skate helmet, the head was pushed deeper into the helmet liner in the FB simulation, and rotated less than in the HO simulation. The positioning of the head inside the helmet, combined with the loading of the neck in the FB simulation thus gave a lower peak value of the resultant angular velocity compared to HO simulation. This was not seen for the road helmet. A similar previous study with motorcycle helmets [12] concluded that the crushing distance of the liner increased with the presence of the body.

In the same study, the ratio between HO and FB for the resultant angular acceleration was between 1.13-1.21, depending on the impact velocity and direction, given a constant initial position (Xm30). In our study, for the same initial position the ratio was 1.02 with the road helmet and 1.21 with the skate helmet. The low value for the road helmet could be influenced by the difference in density of the liner, with a lower density found in the motorcycle helmet.

Another study [11] found a large variation of ratios (HO/FB) between different impact situations using a rigid body model. One case had a slightly higher peak resultant angular acceleration for FB simulation and the other case had a ratio around 1.4. In the current study, a variation is also seen depending on the impact situation, but mostly for the skate helmet. This could partly be due to the skate helmet structure being more elastic, resulting in larger deformations, which increase the time frame.

The two different helmet designs used in this study are typical designs available on the market. The helmet models were compared to four different experimental tests (three oblique impacts and one with a radial impact). The validation for the two helmets showed similar ratios between experimental and simulation results both for the peak values and the timing of peak values.

Future studies should complement this study with more helmet designs, as well as helmets designed for different activities, such as bicycle, equestrian and motorcycle. It would also be interesting to further evaluate how the boundary conditions with and without the body affects the ranking between different helmets.

Influence of Muscle Activation

The present study showed that a stiffer neck (with muscle activation) gave a higher ratio for the resultant linear acceleration. This could explain the difference found with a previous study [12]. In the previous study , that evaluated HO and FB for side falls (Xm30) with different impact velocities with a motorcycle helmet, found a ratio between 1.10 and 1.16 for the peak resultant linear acceleration, compared to 1.05 and 1.04 for the road and skate helmets, respectively, in the present study. . The results from the present study suggest that muscle activation is more important than helmet design for the ratio (HO/FB) of the peak resultant linear acceleration. The same conclusion could not be drawn for the angular motion.

The level of influence from muscle activation on the ratio of the peak resultant angular acceleration and peak resultant angular velocity was more sensitive to impact situation than the ratio for peak resultant linear acceleration was. Muscle activation tended to have a larger influence on the angular motion for falls over the handlebars.

The activation used in this study is one of many different possible activation schemes. The activation levels were based only on using the maximum levels possible with the model to compress the neck without inducing any flexion or extension motions. Other muscle activations schemes should also be evaluated in future studies.

Influence of Posture

Two different postures of the neck and head were evaluated in this study, a relaxed posture and the startle posture, which gave slightly different impact situations. The results suggest that the sensitivity to posture is dependent on the impact situation.

The alteration in posture was limited to the head and neck. The posture of the rest of the body was kept constant, which was the posture of a standing human. Another posture of the body could influence the effective mass.

The Simulation Matrix

A bicycle accident can occur in an infinite number of ways. In this study, variations of two types of bicycle accidents have been used: falling to the side, or over the handlebars. However, these two types of bicycle accidents account for many of the impact points at the front and side of the helmet that have been shown to be the dominant impact points on helmets [24–28]. Many of the studied cases have loads that result in a relatively high compression load on the neck. These scenarios could be seen as worst-case scenarios, where the head impacts the ground first. The accident scenarios are also simulated with only initial linear motion and no rotation. An initial rotation of the body before impact could affect the output. A next step could therefore be to include the initial rotation of the body.

All simulations were performed with impacts against a rigid surface, which is a simplification assuming that concrete or asphalt roads are close to rigid. The impact surface could also be softer, making the contact time between the helmet/head and impacting surface longer. It is believed that a increasing the impact duration increases the influence of the neck. In a study with jockey accidents [10], where the impact surface was turf, the conclusion was that the head acceleration pulses were too long (around 20 ms) to ignore the neck.

Since the impact surface was rigid, this study only included the first 15 ms. If longer time intervals should be studied, an improvement of the neck model is needed. Today the model lacks three dimensional elements of the cervical spine muscles. So if the model as it is modelled today was used for longer time intervals a unrealistic contact between the chin and cervical vertebrae could occur.

The impact velocity was kept constant for all simulations, with a resultant velocity of 6.5 m/s and an impact angle of 45°. These values are within the range of what previous studies have found when analysing bicycle accidents [11], [29]. The impact angle and the resultant velocity have, however, shown to affect the peak value of the resultant linear and angular accelerations [12]. In future studies, the simulation matrix should be complemented with other impact velocities and impact directions.

Global and Local Injury Criteria

The results in this study were analysed both for kinematics (resultant linear acceleration, resultant angular acceleration, and resultant angular velocity) and on the tissue level (first principal strain). The peak first principal Green-Lagrange strain was chosen as an injury metric since studies based on both experiments and simulations [8][21-22] have shown that the first principal Green-Lagrange strain correlates well with brain tissue damage.

The ratio for the peak strain followed the trend for the ratio of the peak resultant angular velocity, which is not surprising since a previous study [32] has shown that the change in angular velocity corresponds best to intracranial strains found in the FE model.

The present study is mainly limited to peak values of the resultant kinematics and first principal strain. Analysis on the component level, as well as analysis over time by comparing the shape and phase of the curves could add more information to the study.

Not only the peak strain of the brain tissue was evaluated but also the strain pattern, where two examples were evaluated (Fig. 5). The strain pattern can be used in e.g., accident reconstructions, to compare the prediction of location between the simulation and medical images. These two examples in Fig. 5 gave small differences in strain pattern for the specific time when maximum value occurred in the section. However, this needs to be further studied for the other cases, and over the whole impact duration, e.g. by comparing the overlap of high strain levels between the two models for the whole impact time [7].

The Neck Model

The neck model is a possible source of error, as it has not been validated for all of the impact situations studied. The neck model used in this study has been previously validated at functional unit levels (both upper and lower vertebrae levels) in flexion, extension, compression, lateral bending and torsion [14][29]. The loads used in these experiments were, however, relatively limited due to the laxity of the isolated vertebral units. It has also been validated in dynamic compression against isolated head-spine complexes (i.e. no muscles) [34] in terms of head and neck forces. Furthermore, it has been validated for head relative T1 motion in dynamic inertial loading against volunteer sled experiments, including muscle activation in flexion, lateral bending [16], and extension [35]. The validations previously performed have been limited to the availability of experiments found in the literature, and has focused, to some degree, on capturing head kinematics from inertial loading of the head. Due to the complexity of the loading experienced in the simulations in the current study, extended validations and model development may be needed. Specifically, more dynamic impact type validations on functional units, isolated spines, and complete cadaveric head-neck complexes should be included. The surrounding neck flesh should also be included to introduce natural limitations on the range of motion.

Helmet Test Standards

There is ongoing work of suggesting improvements of the current helmet test standard in Europe (European Committee for Standardization CEN/TC 158 working group 11) by adding oblique impacts[6]. One question within this work group is the inclusion or exclusion of the neck and the rest of the body in the helmet test standard. Previous studies [9–13] have found divergent results on how the body is influencing the head kinematics. The novelties with this study are, to the authors' knowledge, the evaluations of the influence of different parameters: different head protection, different level of neck muscle activations, and different posture of the head and neck. It also adds more impact situations and an analysis on local tissue level by including peak first principal strain. The study highlights the complexity of helmet test standards. The study has also shown that the neck and the rest of the body affect the kinematics to different levels depending on several factors such as helmet design, muscle activation, initial position and orientation of the body. However, no general recommendation for the work of designing a new helmet test standard can be given concerning the inclusion or exclusion of the neck and the rest of the body. The question regarding the inclusion of the body or not in helmet testing methods should perhaps be rephrased. While the test method should be as realistic as possible, it also has to be robust, repeatable and cost-efficient, with the main objective to promote helmets that can protect the head in as many accident types as possible.

Another question in connection to experimental helmet testing and inclusion of the neck or not is the option to test with a neck and body. The most common neck used as a substitute for the human neck today is the Hybrid III neck. It is however well known that the Hybrid III neck has its limitations and that it is only validated for frontal car collisions. Simulations were performed in [23] for three impact situations and with a Hybrid III neck, the KTH human neck and without a neck. The results showed that the simulations with human neck and no neck were closer in peak values of the kinematics compared to the Hybrid III neck. Therefore, the recommendation is either to design a test method without a neck and the rest of the body or to design a test using a biofidelic representation of the human neck.

V. CONCLUSIONS

The evaluation of the influence of the neck and the rest of the body on head kinematics and brain injury risk for unhelmeted and helmeted impacts against a hard road surface showed a variation among the impact situations and helmet designs used in this study. The studied cases showed for the first 15 ms an average ratio between HO and FB for peak brain tissue strain 1.04 (standard deviation (SD) 0.11), for peak linear acceleration 1.06 (SD 0.04), for peak angular acceleration 1.08 (SD 0.09) and for peak angular velocity 1.05 (SD 0.13).

The study also showed that the influence of neck and body is dependent on the degree of compression load, helmet design and neck muscle activation. Future studies should analyse more impact directions, speeds, impact materials, helmet designs etc. so a better understanding of how helmet test standards should be design to promote safer head protection for many types of accident situations.

VI. ACKNOWLEDGEMENT

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VIII. Appendix

Appendix 1 - The Impact Situations

All 19 impact situations are illustrated in Fig. A1 and Fig. A2.

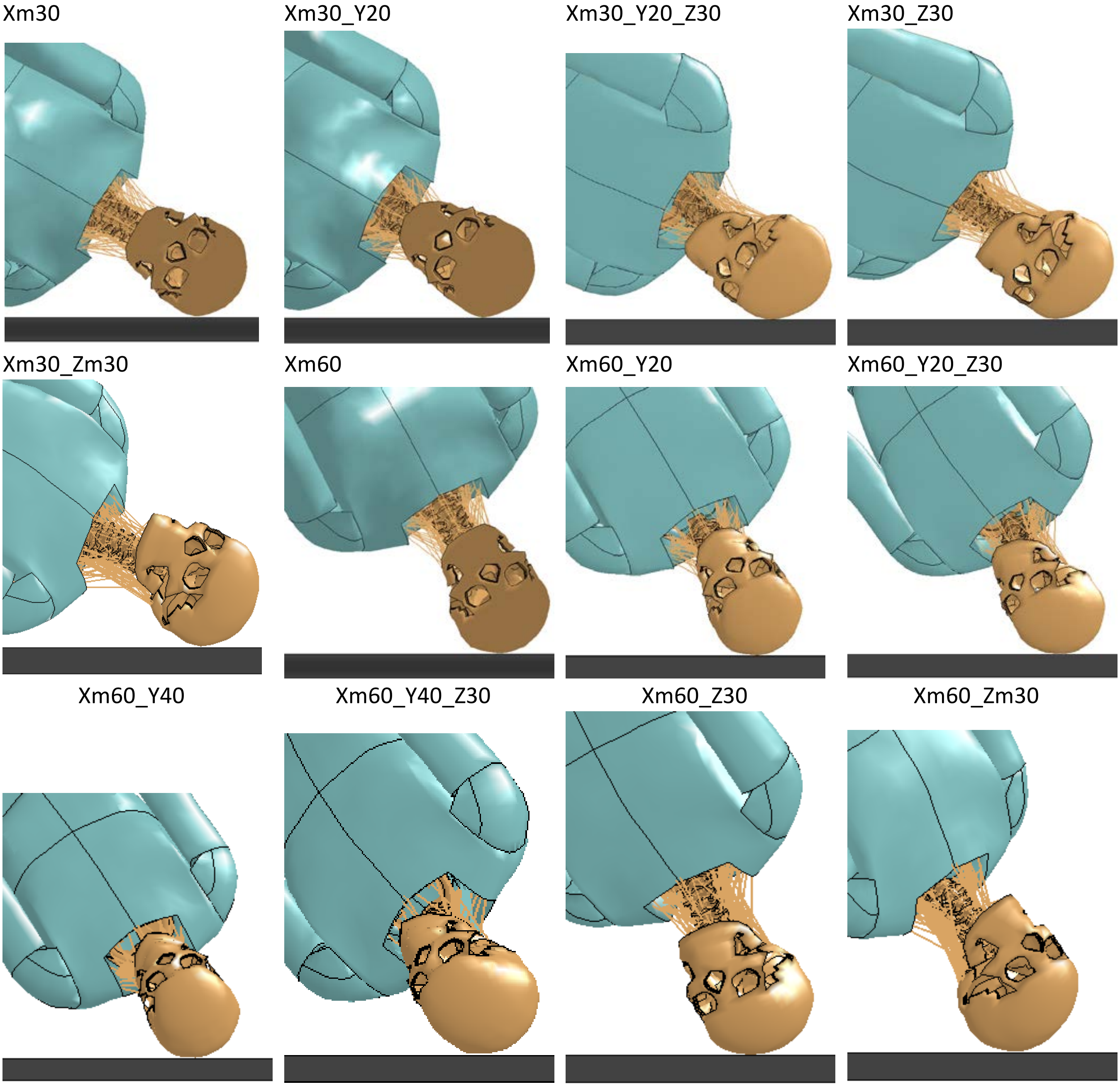

Fig. A1. Initial impact situations for side fall.

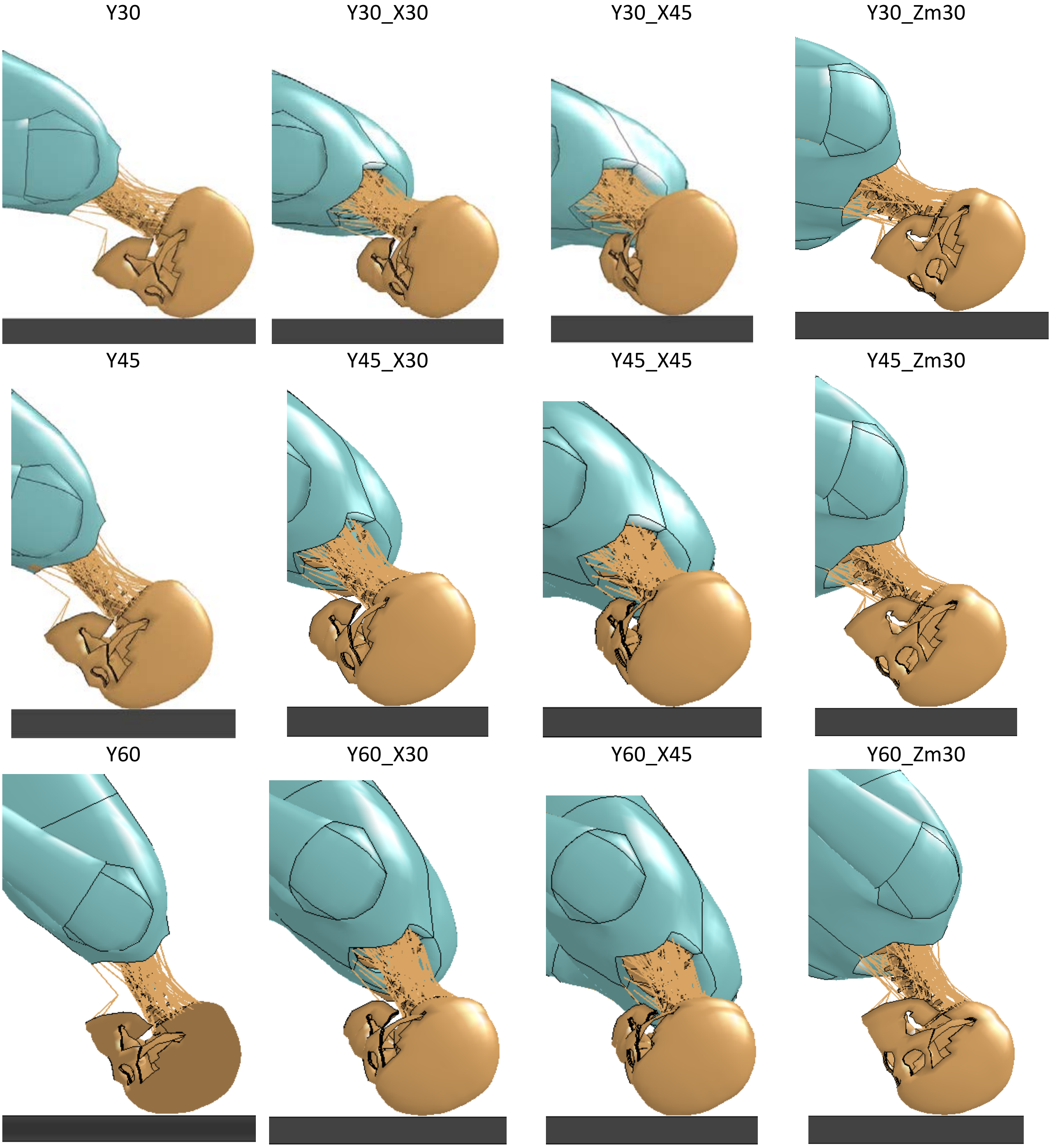

Fig. A2. Initial impact situations for fall over handlebar.

Appendix 2 - The Skate Helmet Model

The skate helmet was constructed from Computer Aided Design (CAD) surfaces of a helmet available on the market. The surfaces were used to mesh the model in Hypermesh (version 13). The expanded polystyrene (EPS) liner was meshed with tetrahedral elements and the outer shell with shell elements (Belytschko-Tsay) with a thickness of 2.5 mm. The helmet model consist of 99 678 tetrahedral elements for the liner and 11 154 triangle shell elements for the outer shell.

The EPS liner was modelled with an isotropic crushable foam material model (Material model number 126 (*MAT_BILKHU/DUBOIS_FOAM) in LS DYNA [36]). The outer shell was modelled with an elastic material model. All material constants are presented in Table AI.

TABLE AI
MATERIAL CONSTANTS OF THE HELMET

	Density [kg/m^3]	Young's modulus [MPa]	Poisson's ratio
EPS liner	70	37	0.01
Outer shell	1162	1640	0.45

The helmet model was compared to different experimental tests (one radial and three oblique impacts) as shown in Fig. A2. For the oblique impacts, two different helmet designs were tested (two helmets of one of the helmet designs and one helmet for the other helmet design). The resultant angular velocity and both resultants angular and linear acceleration of the Hybrid III head was compared between the experiments and the simulations. The data was filtered with a SAE 180 filter before it was analysed. The results from the three helmets and the results from the simulation are shown in Fig. A3. The ratio of the peak value and the ratio of timing of peak value between the experiment and simulations are presented in Table AII. The experimental value was taken as a mean curve of the three helmets. The average ratio of peak value for all four tests was 90% and the average ratio of timing of peak value was 83%.

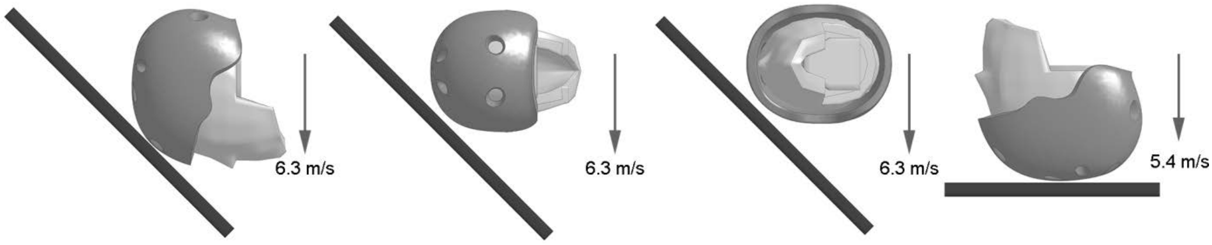

Fig. A1. Test set up for the helmet tests (from left to right front, lateral, pitched and radial).

TABLE AII
THE RATIO OF THE DIFFERENCE BETWEEN THE EXPERIMENTAL AND SIMULATION RESULTS FOR THE PEAK VALUE AND TIMING OF PEAK VALUE FOR ALL FOUR TESTS

	Ratio of Peak Value			Ratio of Timing of Peak Value		
	Linear Accleration	Angular Accleration	Angular Velocity	Linear Accleration	Angular Accleration	Angular Velocity
Front	95%	82%	99%	95%	80%	62%
Lateral	89%	91%	95%	95%	99%	98%
Pitched	80%	99%	81%	99%	68%	60%
Radial	81%			87%		
Average	86%	91%	92%	94%	82%	73%

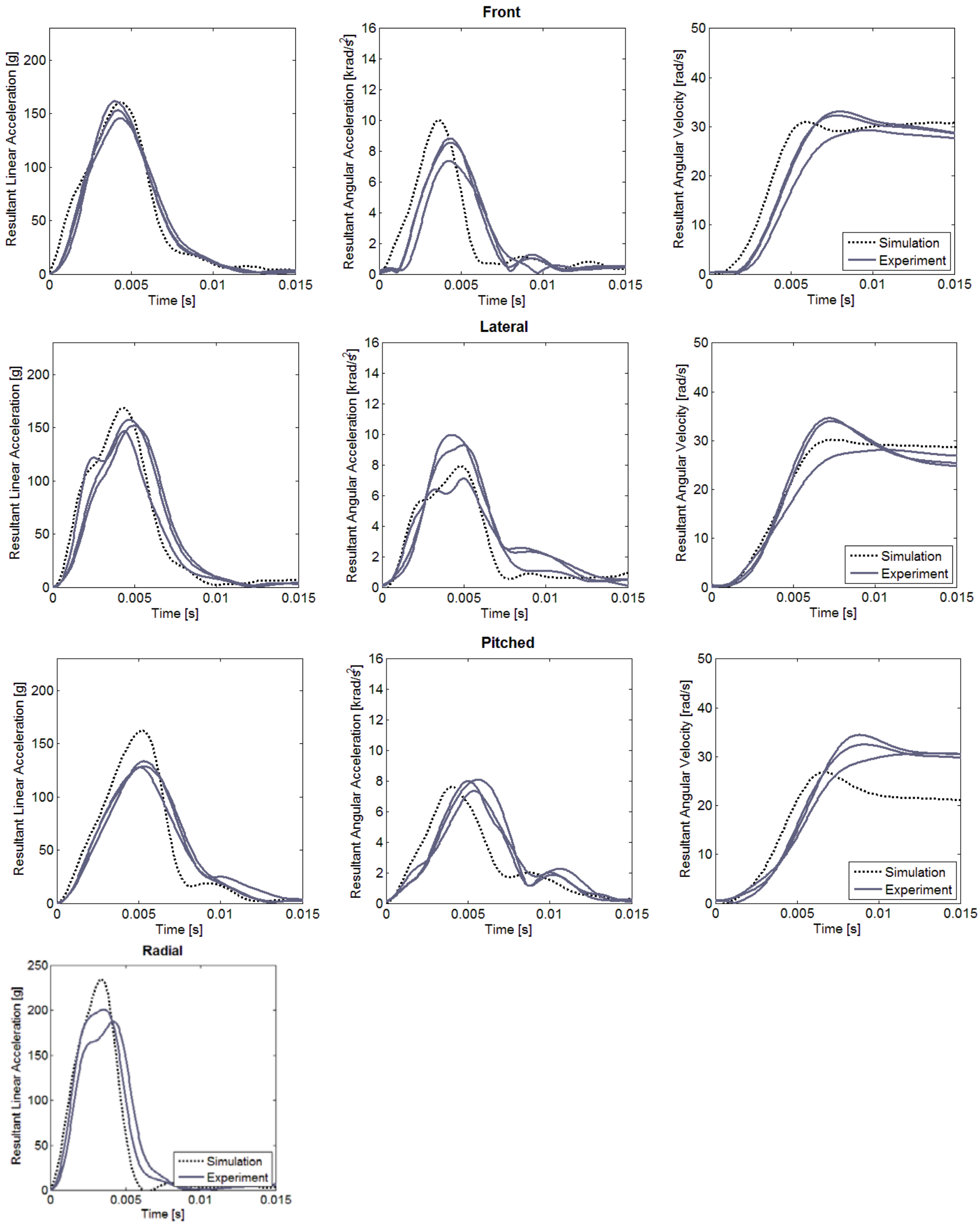

Fig. A3. The comparison between the experimental tests of two different skate helmets and the simulations for four different impacts (from top to bottom Front, Lateral, Pitched, Radial). The first column show linear acceleration, second column angular acceleration and third column angular velocity.

Appendix 3 – Ratio between HO and FB for No Helmet, Road Helmet and Skate Helmet

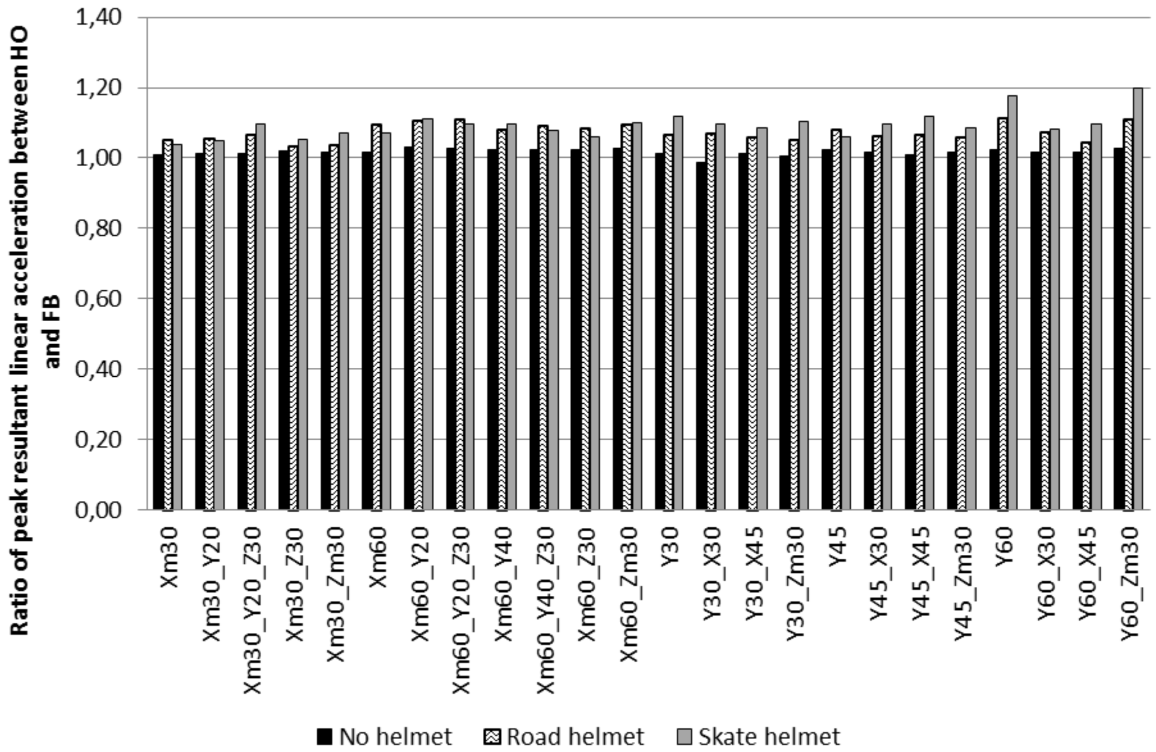

Fig. A4. Ratio between head only (HO) and full body (FO) for the peak resultant linear acceleration for different head protections.

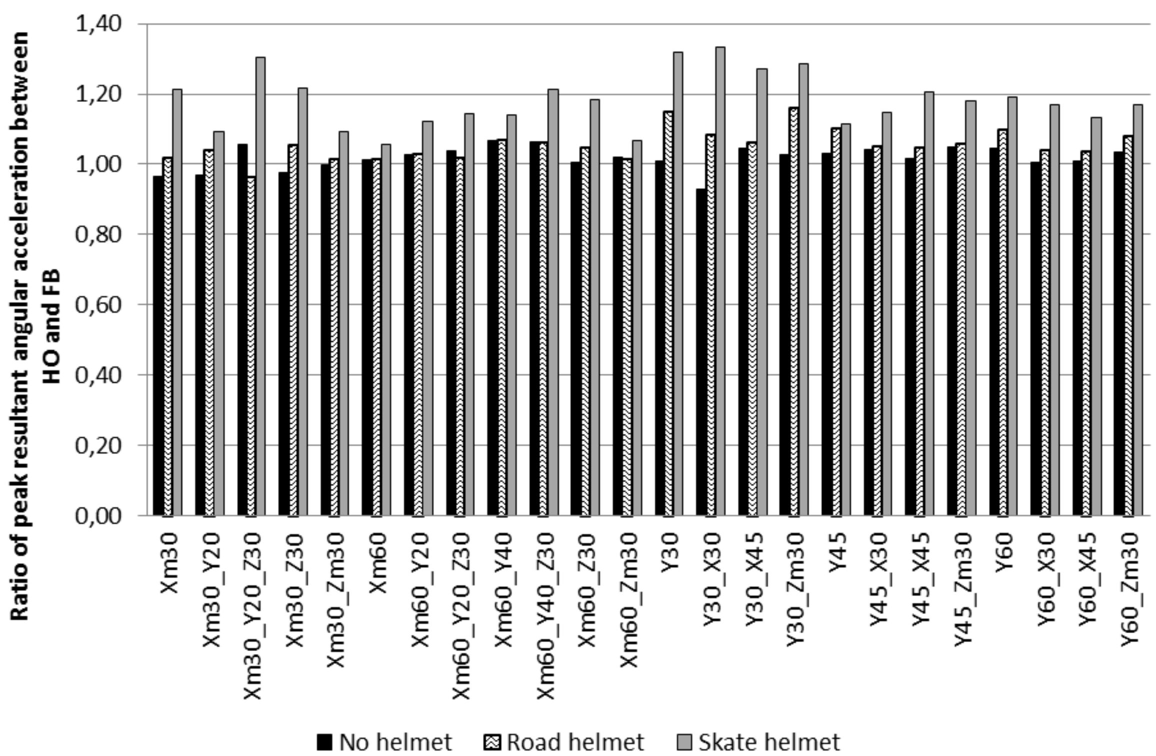

Fig. A5. Ratio between head only (HO) and full body (FO) for the peak resultant angular acceleration for different head protections.

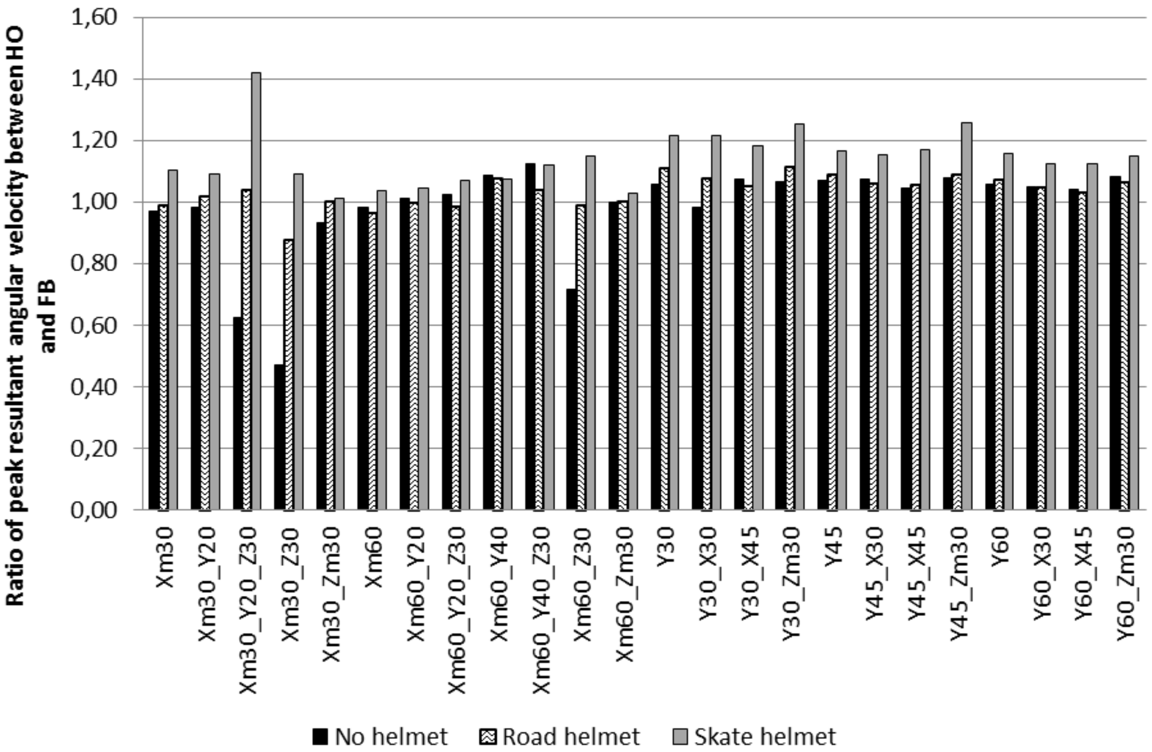

Fig. A6. Ratio between head only (HO) and full body (FO) for the peak resultant angular velocity for different head protections.

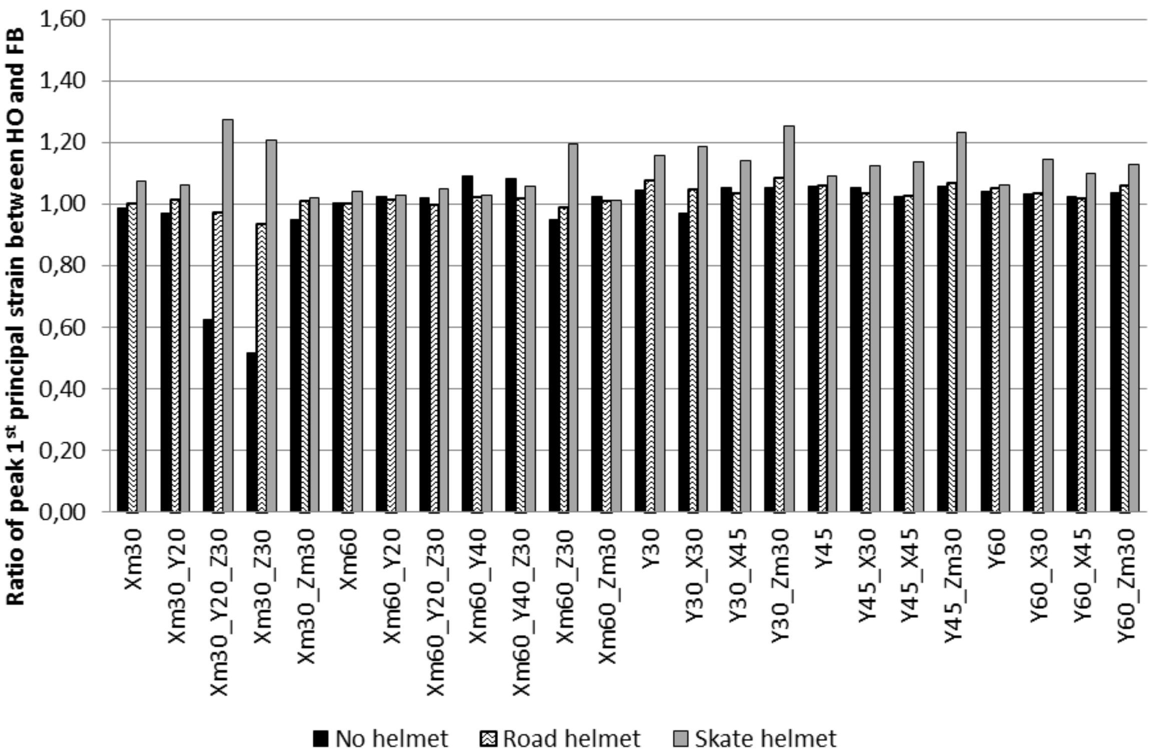

Fig. A7. Ratio between head only (HO) and full body (FO) for the peak first principal strain of the brain tissue for different head protections.

Appendix 4 – Ratio between HO and FB for Road Helmet with and without Muscle Activation

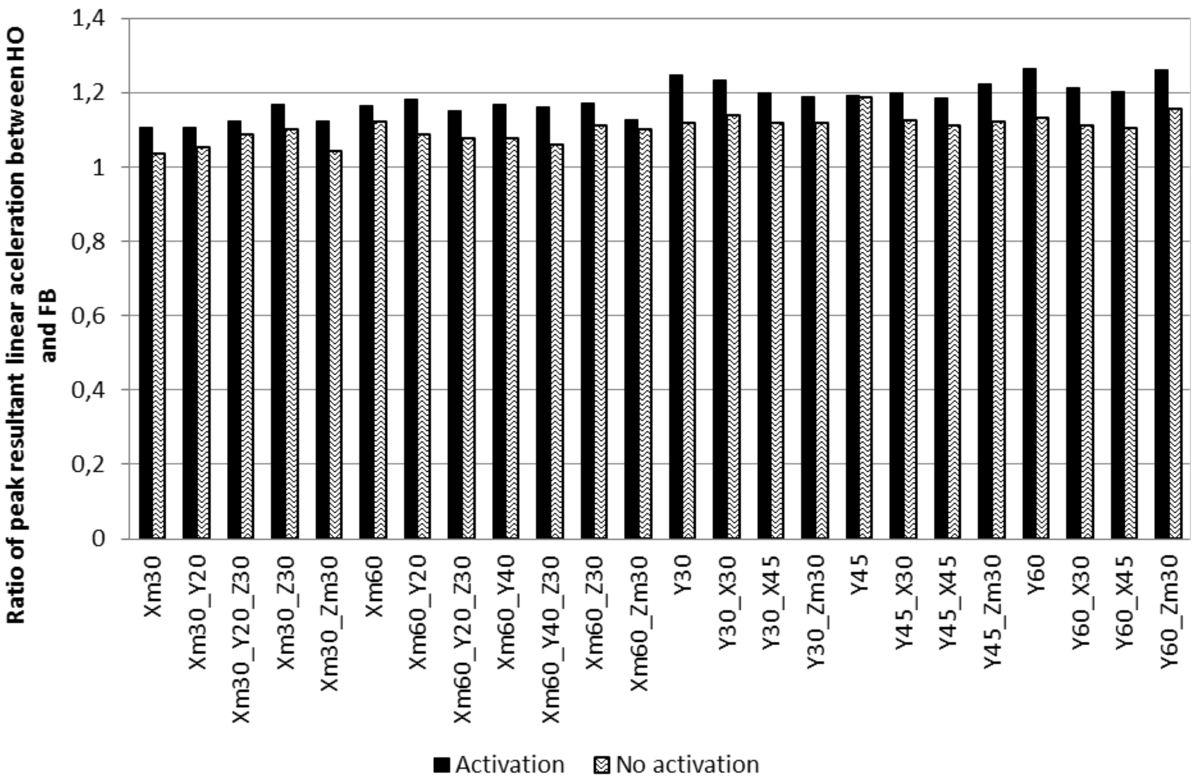

Fig. A8. Ratio between head only (HO) and full body (FO) for the peak resultant linear acceleration with and without muscle activation.

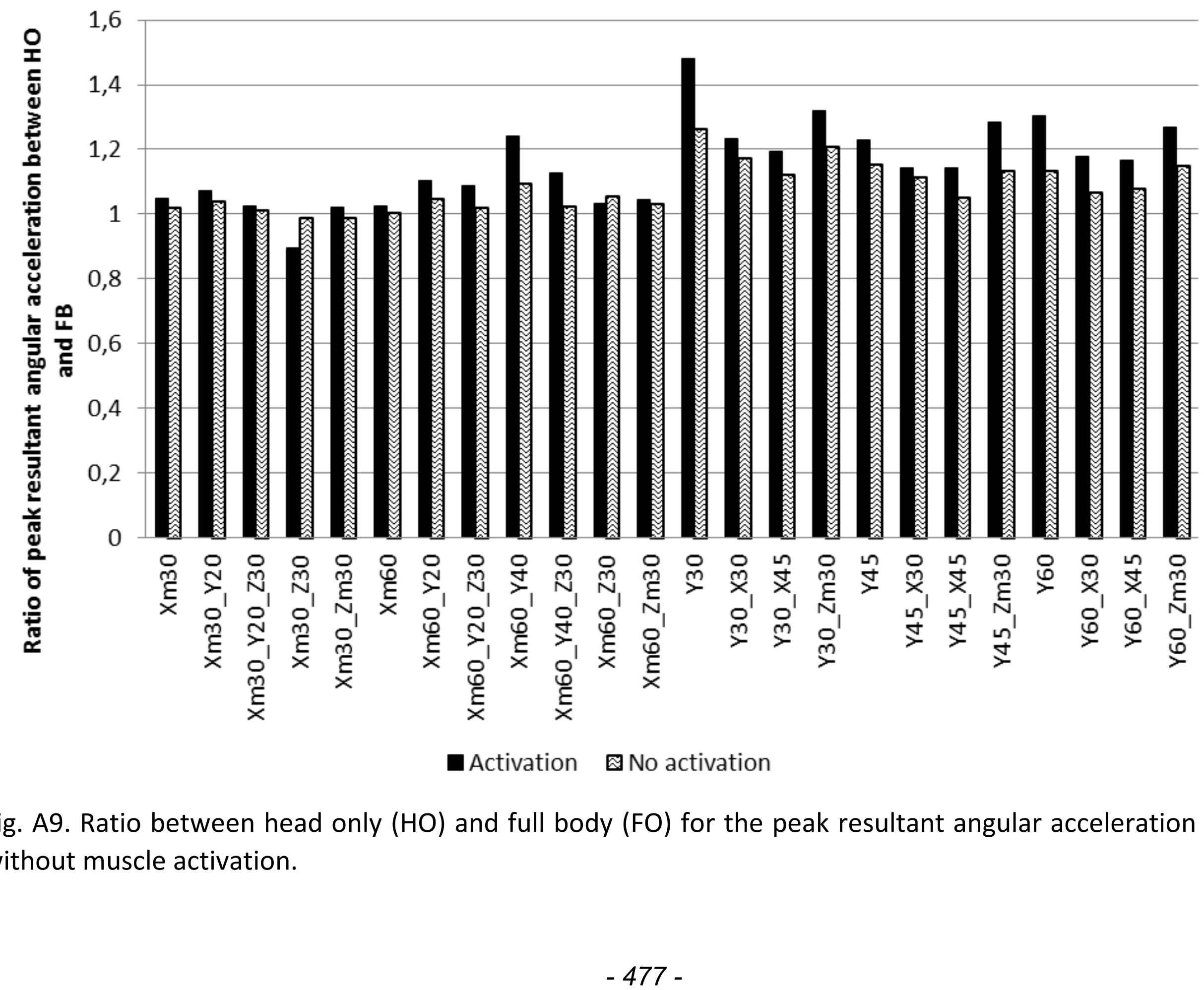

Fig. A9. Ratio between head only (HO) and full body (FO) for the peak resultant angular acceleration with and without muscle activation.

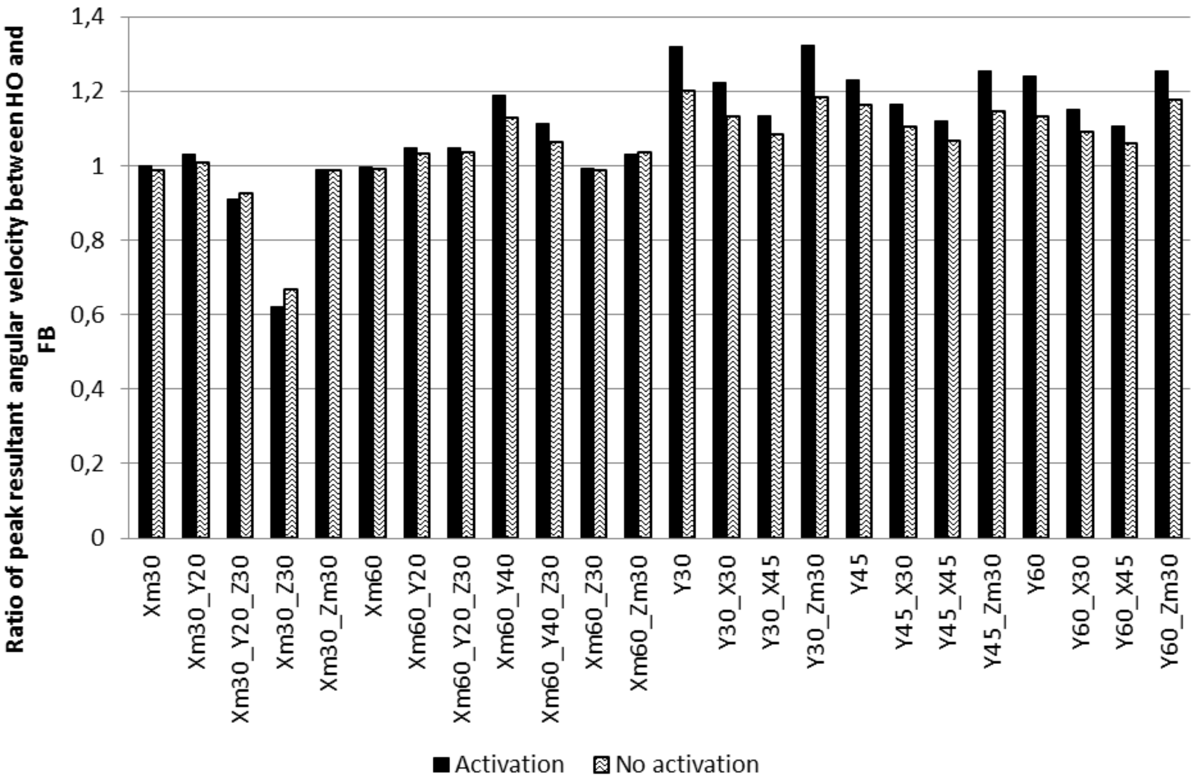

Fig. A10. Ratio between head only (HO) and full body (FO) for the peak resultant angular velocity with and without muscle activation.

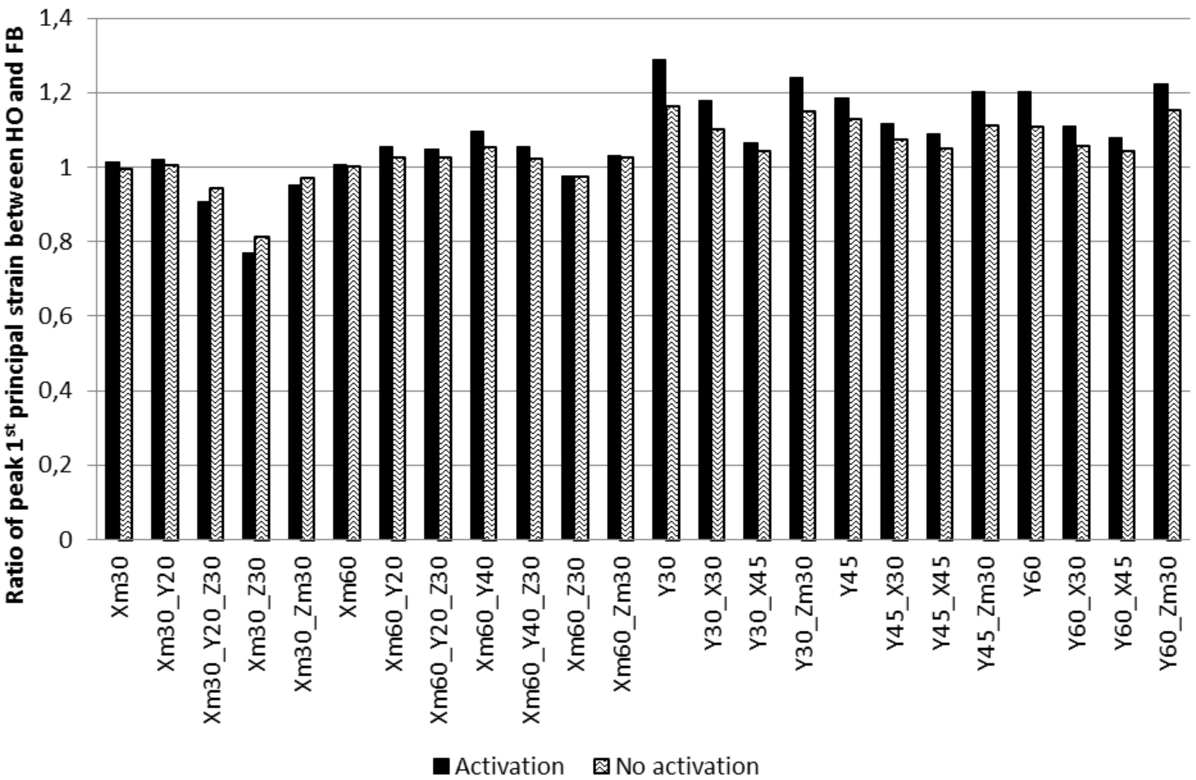

Fig. A11. Ratio between head only (HO) and full body (FO) for the peak first principal Green-Lagrange strain with and without muscle activation.