IMPLEMENTATION OF HEAD INJURY CRITERIA IN CYCLIST-CAR ACCIDENTS

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ABSTRACT
Several head injury criteria, some of which are generally accepted whereas others are newly proposed, are implemented and applied to multibody accident simulations of typical bicycle-car accidents. The main goal is to establish the importance of car - and bicycle speed and relative car-bicycle trajectory for accident reconstruction studies. For this purpose, Madymo® cyclist and car models are used in a parametric sensitivity analysis study. Head injury criteria include the HIC, maximum change in rotational velocity of the head and peak rotational head acceleration. Injury parameters are investigated for 30 different cases of frontal and sideways collisions. For cyclist speeds ranging from 10 to 30 km/h and car speeds from 20 to 50 km/h, HIC values range from 200 up to 2500. Maximum change in head rotational velocity ranges from 20 to 60 rad/s and peak rotational acceleration from 4500 to 22000 rad/s². In general, sideways impacts lead to higher values for all head injury criteria. The influence of car speed on the head injury criteria values is larger in the sideways collision cases when compared to the frontal case. On the contrary, cyclist speed has a larger influence in the frontal case. It is concluded that car speed is the most critical factor and traffic expert input is needed to calculate car speed at time of impact based on objective information. Next to car speed, the relative car-bicycle position needs to be established iteratively by matching modelled impact positions with those reported in medical files, eyewitness reports, police- and traffic expert reports.

Key words: Head, head injury criteria, cyclist, Madymo, multibody model

INTRODUCTION

CYCLING IS A WORLDWIDE POPULAR means of transportation. In Belgium, cyclists account for approximately 8-10% of all traffic fatalities. Several studies indicate that cyclist-car collisions, as opposed to cyclist falls, lead to more severe injuries and a higher mortality rate (Collins, 1993; Kraus, 1987). These types of traffic accidents have therefore been studied for several decades using virtual modelling techniques. One of the first 3D mathematical cyclist models was the one from Janssen and Wismans (1985) which consisted of nine body segments. Cyclist arms were not included in this model. It showed that friction between the dummy on the one hand and the vehicle and ground on the other hand did not have a large influence on the mechanical results. Also the mass of the bicycle was found to have only a marginal effect. In 1988, Huijbers and Janssen (1988) reported on the effect of the car front shape on the kinematics of the cyclist during the accident. They conclude that the car front shape has a considerable influence on the relative head impact velocity in the case of an impact. More recently, Maki et al. (2002) and Serre et al. (2006) reported on the mathematical simulation of real life bicycle accidents and the validation of this simulation by physical accident reconstruction using either a post-mortem human subject (PMHS) and/or a pedestrian dummy model in combination with a bicycle. The former author reports relative head centre of gravity (CG) impact speeds with repect to the car up to 10 m/s at a car speed of 35 km/h for both frontal and lateral car-to cyclist collisions. The latter author reports on an oblique collision (angle of 30° with respect to frontal collision) at 40km/h and finds values of approximately 5 m/s for the head CG maximum velocity and 1400 m/s² for the peak linear acceleration of the head CG. The victim was a 13-year old.
The main goal of this study is to establish the importance of accident circumstances, more specifically car and bicycle speed and relative car-bicycle trajectory for accident reconstruction studies.

METHODS

MADYMO® MULTIBODY SOFTWARE (TNO, 2005) is used to perform a sensitivity analysis on two typical bicycle-car accident cases. The implemented cyclist model is a combination of the Madymo 50th percentile male pedestrian model and a bicycle model based on a real-life bicycle as shown in figures 1.a & 1.b. Passive elastic belts are applied between the hands and the bicycle handles to account for gripping onto the handlebars. To model the interaction between the pedals and the feet, elastic belts are also applied between the shoes and the pedals. Figure 1.c. shows the two initial conditions for the sensitivity analysis, which represent approximately 90% of real-life bicycle-car accidents (Maki, 2002). For these initial cases, the cyclist is positioned in the middle of the car, the car speed is 32km/h and the cyclist speed is 10 km/h. The car model is an adapted version of the car model as used by Yang (2000) and is also shown in figure 1.c. The cyclist model is constructed in a similar manner as the models that were used in validation tests performed by Maki (2002) and Serre (2006).

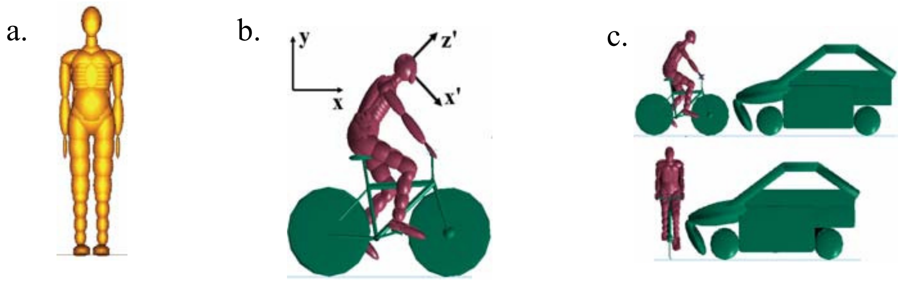

Fig. 1 - a. Madymo 50th percentile male consisting of 52 rigid bodies. b. Multibody cyclist modelled as a combination of the 50th percentile male and a bicycle consisting of 18 rigid bodies. c. Two initial cases for analysis. Top: Frontal case, Bottom: Sideways case. The car model is adapted from Yang et al. (2000).

HEAD LESION PARAMETERS are calculated from the multibody simulations. Commonly accepted parameters include the HIC_{15}, peak head linear and angular acceleration, impact force and force impulse. Next to these parameters, two new proposed lesion parameters are studied.

A first new lesion parameter, generalized from diffuse axonal injury research by Margulies et al. (1989), is the *peak change in rotational velocity of the human head* $\Delta\dot{\theta}_p$, defined as:

$$\Delta\dot{\theta}_p = \max\left[\sqrt{\left(\int_{t_1}^{t_2}\ddot{\theta}_x(t)\cdot dt\right)^2 + \left(\int_{t_1}^{t_2}\ddot{\theta}_y(t)\cdot dt\right)^2 + \left(\int_{t_1}^{t_2}\ddot{\theta}_z(t)\cdot dt\right)^2}\right] \quad \forall t_1 < t_2$$

With xyz an inertial coordinate frame as in figure 1.b. $\Delta\dot{\theta}_p$ is hypothesized to correlate to brain injury. The change in rotational velocity is directly related to the moment of the impulses exerted by the tangential forces working upon the brain mass over the duration of the impact. These impulses are responsible for the compression of the brain mass in the tangential direction. Because the brain is a viscoelastic material and contains blood-filled veins, the time duration will be a relevant factor in the amount of brain compression. Hence the change in angular velocity, and the tangential impulses related to it, as they are a combination of forces and time duration, should be related to brain injury.

A second new head lesion parameter is the *peak angular acceleration of the head after triangulisation* α_p', which is defined, as explained in figure 2, as the peak Aac value of the triangulated pulse. It has been used in literature in acute subdural hematoma tests by Löwenhielm (1978) and later by Depreitere et al. (2006).

RESULTS

CAR SPEED, BICYCLE SPEED AND INITIAL CAR-BICYCLE POSITION were varied to study their influence on the head lesion predicting parameters. Figure 3 shows HIC, $\Delta\dot{\theta}_p$ and α_p' results as a function of car speed for both frontal and sideways impact conditions. Values for the 25 km/h frontal case were excluded from further analysis because of influence of the hand belts on the kinematics of the cyclist model.

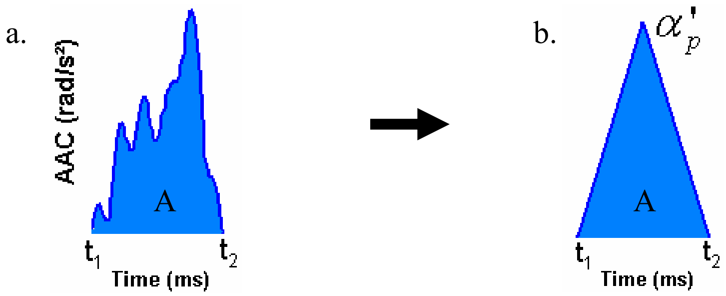

Fig. 2 - a. Example of angular acceleration (Aac) measurement of the head vs. time. b. Conversion of the signal to a triangular pulse with the same pulse duration and area A.

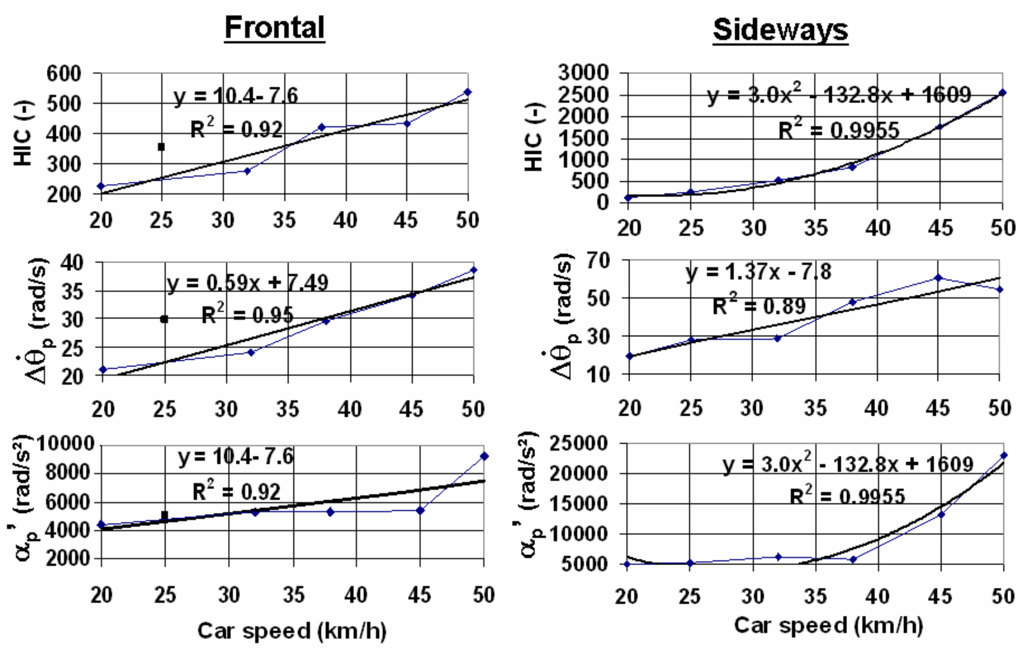

Fig. 3 – HIC, $\Delta\dot{\theta}_p$ and α_p' results as a function of car speed for frontal and sideways impacts. y = analytical function of trendline, R^2 = correlation coefficient.

Table 1 compares the relative influence of car speed, bicycle speed, trajectory angle and distance to the centre line of the car. Trajectory angle influence is compared by rotating the cyclist around a vertical axis Distance to the centre line of the car was studied by moving the cyclist left and right with respect to the car over a distance of 0.3 and 0.6m. Head CG impact speed varied between 1.87 and 11.09m/s while the peak linear acceleration varied between 143 and 1838 rad/s².

Table 1 – Maximum changes (difference between minimum and maximum value) for HIC, $\Delta\dot{\theta}_p$ and α_p' for changes in bicycle speed from 10 to 30 km/h, car speed from 32 to 50 km/h, trajectory angle from -10° to +10° and distance to car centre from -0.6 to +0.6m.

	Frontal		Sideways		Frontal		Sideways	
	Bicycle speed	Car Speed	Bicycle speed	Car Speed	Trajectory angle	Distance to car centre	Trajectory angle	Distance to car centre
Parameter	Max. change (%)		Max. change (%)		Max. change (%)		Max. change (%)	
HIC	26	94	19	390	29	31	25	8
$\Delta\dot{\theta}_p$	41	60	20	110	183	95	68	44
α_p'	32	76	25	259	250	309	118	11

DISCUSSION AND CONCLUSION

LESION PARAMETER COMPARISON for car speed (figure 3) shows linear trend lines for all frontal cases and for $\Delta\dot{\theta}_p$ of the sideways case. For the two other sideways lesion parameters, a second degree polynomial correlates best to the data. When comparing the sideways to the frontal case, this results in comparable values for both cases for HIC (±230) and α_p' (±5000) at low car speed and significantly higher values for the sideways case at high car speed. This trend, although less pronounced, is also observed for $\Delta\dot{\theta}_p$. At low car speed sideways values are comparable or even smaller than in the frontal case, while at high speeds sideways values are significantly higher. The same can be seen in table 1.

Also from table 1, it is found that bicycle speed influence on head lesion parameters is significantly smaller than car speed, in particular for the sideways lesion case. The largest influence of bicycle speed is observed in the frontal case. Concerning the trajectory angle and distance to car centre influence on the head lesion parameters, it is found that the frontal case is more critical than the sideways case, especially for the rotational parameters $\Delta\dot{\theta}_p$ and α_p'.

To study the sensitivity of the head lesion parameters to changes in the car shape, the hood was prolonged by a value of 20 cm and the same series of analysis was performed. The head impact parameters results were slightly lower when compared to the same cases with the shorter car hood. When comparing the maximum changes for HIC, $\Delta\dot{\theta}_p$ and α_p' for the longer car, in the same way as is done for the shorter car in table 1, results are very comparable to the ones given in that table.

As stated in the introduction, the model was constructed in the same way as the models of Maki (2002) and Serre (2006), which were validated through physical reconstruction of real-life accidents. Although a more thorough validation of the model might be beneficial, comparison of Serre's results to the ones from this study shows a high degree of agreement. Serre reported on values of 5 m/s for peak head CG speed and a peak linear acceleration of 1400 m/s² for a case as described above. Interpolated values from this study for similar circumstances show a peak head CG speed of 4.2 m/s and a peak linear acceleration of 1050 m/s². Both values are approximately 20% smaller than the values reported by Serre. This difference is caused by the inertial differences between the 50th percentile male (our model) and the 13-year old model used by Serre.

CONCLUDING, the results show several relatively large changes in lesion parameters for only small changes in car - or bicycle speeds and trajectories. For the variation in speed, it is found that the car speed is the most important factor. For real-life accidents, it is critical to have an accurate estimation of car speed at the moment of impact. This speed needs to be calculated by traffic experts based on objective information like braking distance, weather conditions, road surface, etc. For the pre-accident position parameters (trajectory angle and distance to car), the high lesion parameter sensitivity is caused by the large change in cyclist kinematics. A small change in relative position causes the impact positions on the body, and thus also on the head, to differ strongly throughout the study. In real-life accident reconstruction, this problem needs be solved by iteratively matching modelled impact positions with those reported in medical files, eyewitness reports and police- and traffic expert reports.

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