PEDESTRIAN HEAD IMPACTS:
Development and Validation of a Mathematical Model

by

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ABSTRACT:

This paper describes the development of a mathematical crash victim simulation model (MADYMO based) used to predict the characteristics of the head impact in real pedestrian accidents. The model forms the connection between the accident circumstances and a detailed analysis of the head injuries in selected fatal pedestrian accidents. These accidents are being investigated as part of an on-going project into the mechanisms of head injury in road accidents.

The pedestrian model is based on a combination of measurements taken at autopsy and estimation from published data to describe the anthropometry of the victim. Available human and cadaver data are used for the contact stiffnesses and joint characteristics.

A comparison between the model predictions and the results of a series of pedestrian accident reconstructions using cadavers is made. The comparison demonstrates that this relatively simple two dimensional simulation can reliably predict the timing, position and velocity of the pedestrian head impact. Some limitations of the model are discussed.

1. INTRODUCTION

The NH&MRC Road Accident Research Unit is undertaking a major multidisciplinary study of head injury in road accidents (1)*. The objectives are to increase our understanding of the mechanisms of brain injury and human tolerance to head impact. This is being done by relating the nature and severity of the impact to the head to the nature and severity of the damage to the brain. The characteristics of the impact are determined from investigation of the crash followed by mathematical modelling of the collision sequence.

Thus far, the study has been concentrated on pedestrian accidents, because it was thought that a relatively simple two-dimensional mathematical model would adequately simulate this type of accident. Investigation of 85 pedestrian fatalities between June 1983 and September 1985 in Adelaide, South Australia (2), has shown that the majority (67%) of these cases involved a passenger car. Of those struck by a car, in 72% of cases the pedestrian was struck on the side by the front of the car, and the majority (58%) of vehicles involved were travelling at between 40 and 80 km/h when they struck the pedestrian.

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* refers to references at the end of this paper.

Most of the pedestrians struck by a passenger car had a primary head impact on the relatively soft front structure of the vehicle (65% of the cases). A secondary head impact often occurred with the road surface. These secondary impacts are usually equivalent to a fall from a height of about one metre (3). The superficial injuries to the head provide a reasonably reliable indication of the importance of any secondary impact. We have not attempted to simulate cases involving a significant secondary impact.

The mathematical model has therefore been designed to simulate the motion of a pedestrian who has been struck on the side by a car travelling at a speed of up to 80 km/h. The model used is based on the MADYMO crash victim simulation developed by TNO in the Netherlands (4).

The procedure used in the application of the mathematical model is as follows:

1. Examine the accident scene and the striking vehicle.
2. Attend the autopsy to obtain detailed information on injuries and dimensions of body segments.
3. Insert relevant vehicular and human parameters into the modified MADYMO model.
4. Iteratively apply the MADYMO model, varying the values of selected parameters, until the actual pedestrian/car contact points can be reproduced.
5. Record the velocity of the head on striking the car, together with the estimated acceleration assuming a value for the stiffness of the head/car contact.

2. PEDESTRIAN MODEL

2.1 Introduction

The application of the MADYMO crash victim simulation to pedestrian accidents has been described previously (5). It has been shown to be capable of accurately simulating accident reconstructions using anthropomorphic test devices (ATDs)(6).

The MADYMO pedestrian model was developed, and has most commonly been used, to simulate a fifty percentile Part 572 ATD. The characteristics inherent in this version of the mathematical model make it more suited to modelling ATD's than humans. However for this study it was felt that the model would have to resemble more closely the individual human victim in order to provide satisfactory results (2). The critical areas of the model were found to be the anthropometry and joint properties of the victim, and vehicle characteristics such as impact speed, overall frontal shape and the dynamic stiffness of individual components.

An extensive literature search was therefore carried out for information on human body parameters including link lengths, mass distributions, allowable joint motions and stiffnesses. It was found that these differ, in many cases significantly, from values specified for a Part 572 ATD. A procedure was developed to assist in the estimation of body parameters not amenable to direct measurement at autopsy or in the hospital. A description of this procedure is presented in Reference 2.

2.2 Pedestrian Geometry and Mass Distribution

Figure 1 shows the reference axes used and the manner in which the body is divided into 9 segments. The arms are included as part of the thoracic segment.

For link lengths not directly measured, estimates are made in terms of the percentage of the total body length using a procedure developed from data in references (7) and (8)(see Table 1). For the neck an effective length of 125mm has been selected from volunteer data (9). Data from (7) were also used to estimate body segment diameters and centres of mass. Data from references (7) and (8) were combined to develop a way to estimate the mass of body segments as a percentage of total body weight, as shown in Table 1.

The moment of inertia (I) around the X axis for a body segment is expressed as a function of the segment mass (M), the segment radius of gyration about the X axis (k) and the segment link length (L).

$$I = [(k/L) \times L]^2 \times M$$

Coefficients defining the radius of gyration for most of the segments listed in Table 1 were derived from reference (8). Those not available from this source are calculated from the geometry of the body segment.

2.3 Joint Motion

The model permits definition of the following joint characteristics:

- the resistive elastic torque
- the resistive friction torque
- the resistive damping torque

In the extensive literature available on the subject there is only partial agreement about the means of measuring joint motion. This has led to differing definitions of, and limits for normal joint flexibility and of what constitutes hypo- or hyper-flexible joint motion. With these constraints a set of joint functions was derived from the literature. It is shown in Figure 2. The references on which these functions were based are listed in (2). The characteristics of the neck joint, which is probably the most important in this context, were taken from reference (9).

For the model, the friction in each joint is assumed to be zero. Data for joint damping are not readily available in the literature. However, reference (10) gives damping coefficients of 3.75 Nm sec/rad for the hip and 1.05 Nm sec/rad for the knee. In the absence of better data, 3.75 Nm sec/rad is used for all other joints.

2.4 Stiffness Data

For MADYMO each contact has to be defined in terms of the following characteristics of the two contacting surfaces: the combined dynamic stiffness and the coefficients of restitution and damping. We have used published results from volunteer and cadaver tests for the human stiffness data and combined this with the results of dynamic tests on vehicles using rigid head and leg forms.

TABLE 1: BODY GEOMETRY AND MASS ESTIMATION SYSTEM

LINK	REFERENCE JOINT	END OF LINK	LINK LENGTH (% of BODY HEIGHT)	SEGMENT DIAMETER (% of HEIGHT)	CENTRE OF MASS: DISTANCE FROM REFERENCE JOINT ON Z AXIS (% OF LINK LENGTH)	SEGMENT MASS (% of TOTAL BODY MASS)		RADIUS OF GYRATION ABOUT X AXIS (% OF SEGMENT LENGTH)
						MALE	FEMALE	
Head	Occipital Condyle/ C1	Vertex	10.6	9.3	50.0	7.1	5.8	31.6
Neck	C7/T1	Occipital Condyle/ C1	7.2	6.8	50.0	2.5	2.0	43.2
Thorax	T12/L1	C7/T1	13.9	19.5	50.0	31.8	30.4	42.2
Abdomen	L5/S1	T12/L1	10.5	14.5	50.0	13.5	13.6	46.9
Pelvis	Mid point between right and left hip joints	L5/S1	5.2	19.8	50.0	12.3	12.4	97.9
Thigh	Hip joint	Knee joint	24.1	8.5	43.0	10.4	11.4	27.9
Lower leg and foot	Knee joint	Base of heel	28.5	6.7	50.0	6.0	6.5	32.8
			100.0	100.0		100.0	100.0	

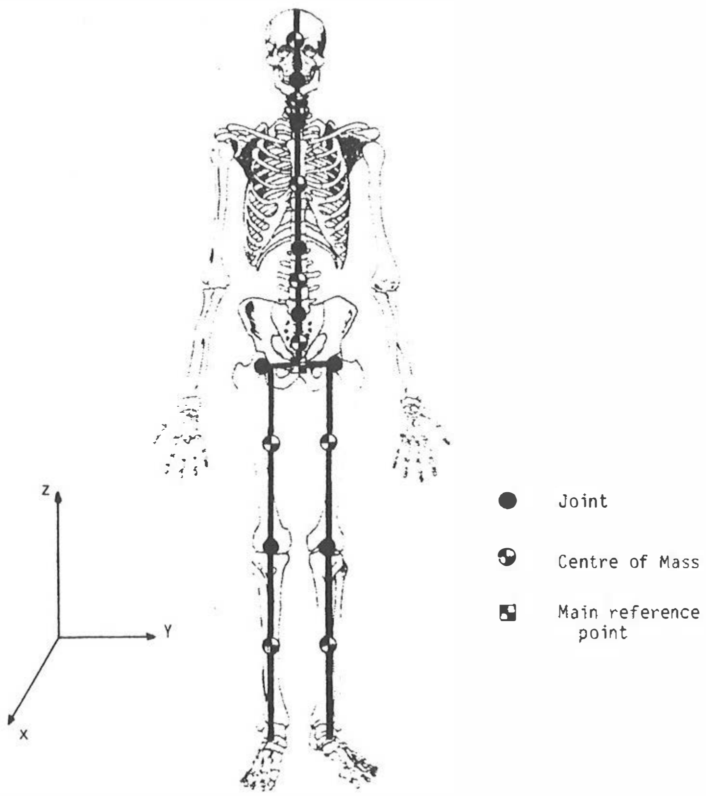

FIGURE 1. System of Body Segmentation.

2.4.1 Head

For the head an initial stiffness of the skull of 2.1 kN/mm was used from the experiments reported in reference (11) and this was combined with a final stiffness of 0.8 kN/mm up to skull fracture, as shown in Figure 3. When combined with the effect of the soft tissues of the scalp these results compare well with those presented in reference (12). Ideally the skull stiffness should be varied to take into account the characteristics of the surface impacted, but there is only a limited amount of information of this type available.

2.4.2 Thorax

The response of the torso to blunt impact is velocity dependent (13). Cadaver reconstructions of pedestrian collisions with a vehicle travelling at a speed of 40 km/h have resulted in a thorax impact velocity of 20 km/h. Data from cadaver chest impacts at a velocity of about 20 km/h are given in reference (13). For the model a stiffness of 0.5 kN/mm was assumed with a plateau at 5.0 kN (see Figure 3).

2.4.3 Pelvis

Very little stiffness data is available for the pelvis. In reference (14) a lumped parameter model of the pelvis and thigh is presented. This gives the values used in the model: a stiffness of 0.5 kN/mm for the pelvis and damping of 3.5 N sec/mm.

2.4.4 Thigh and lower leg

A considerable amount of consistent information is available for the stiffness of the thigh and lower leg (2). For the model it is assumed that the impact tolerance to fracture for the lower leg is 4.0 kN with a stiffness of 0.25 kN/mm and for the thigh 7.5 kN and 0.35 kN/mm.

3. VALIDATION

3.1 The Experiment

Some form of validation of the simulation was necessary. The main output required from the simulation is a reasonable estimate of head impact velocity, given the type of information available from a detailed accident investigation. This normally includes the head impact location on the vehicle and an estimate of the vehicle speed at impact.

A series of seven staged car/pedestrian accident reconstructions using cadavers has been published by ONSER (3, 15). These provide sufficient information to enable us to compare the cadaver reconstructions with the mathematical simulation in terms of trajectory (Figures 4 and 5), head impact time, head impact location and velocity (Table 2) given known initial conditions. Vehicle stiffnesses given in reference (15) were combined with assumed body stiffnesses to give the values used in the model (Figure 6). A summary of the cadaver test data is given in Table 3.

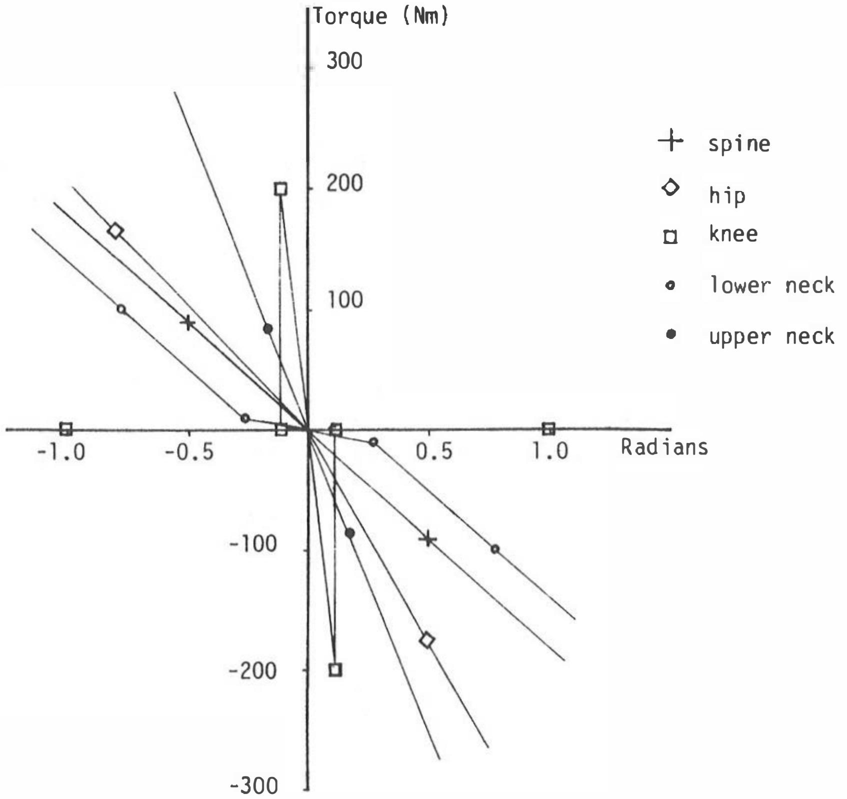

Figure 2. Joint Functions used in the 2D 9 Segment Pedestrian Model.

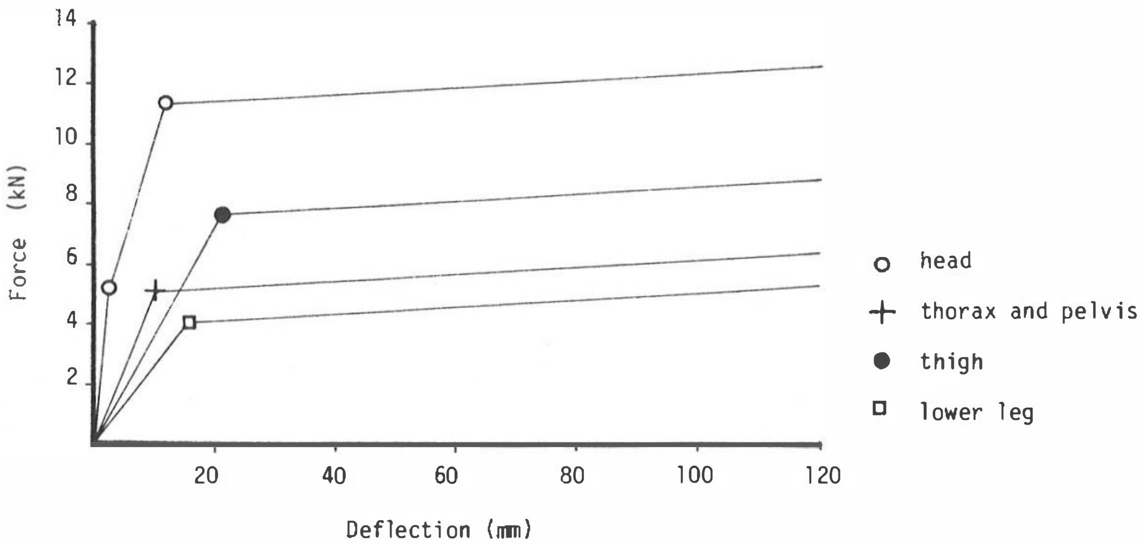

Figure 3. Body Segment Stiffness Characteristics Assumed for the Pedestrian Model.

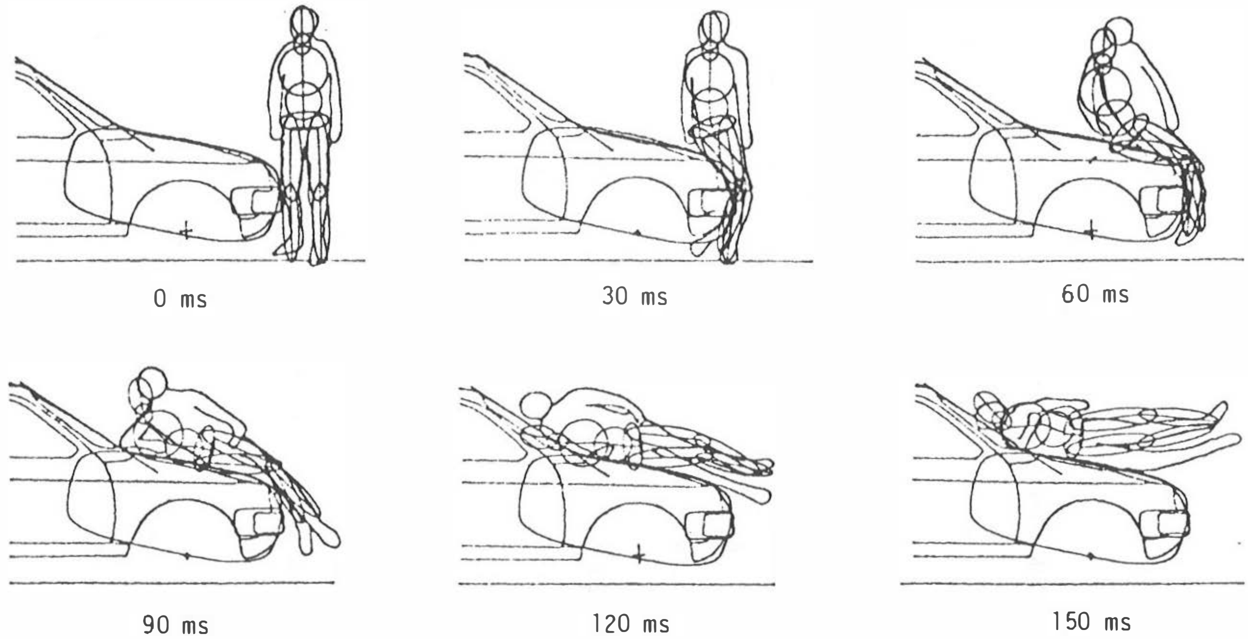

Figure 4. Comparison of Pedestrian Trajectory Sequence for the Model and a Cadaver Test (Citroen VISA - FOC 45).

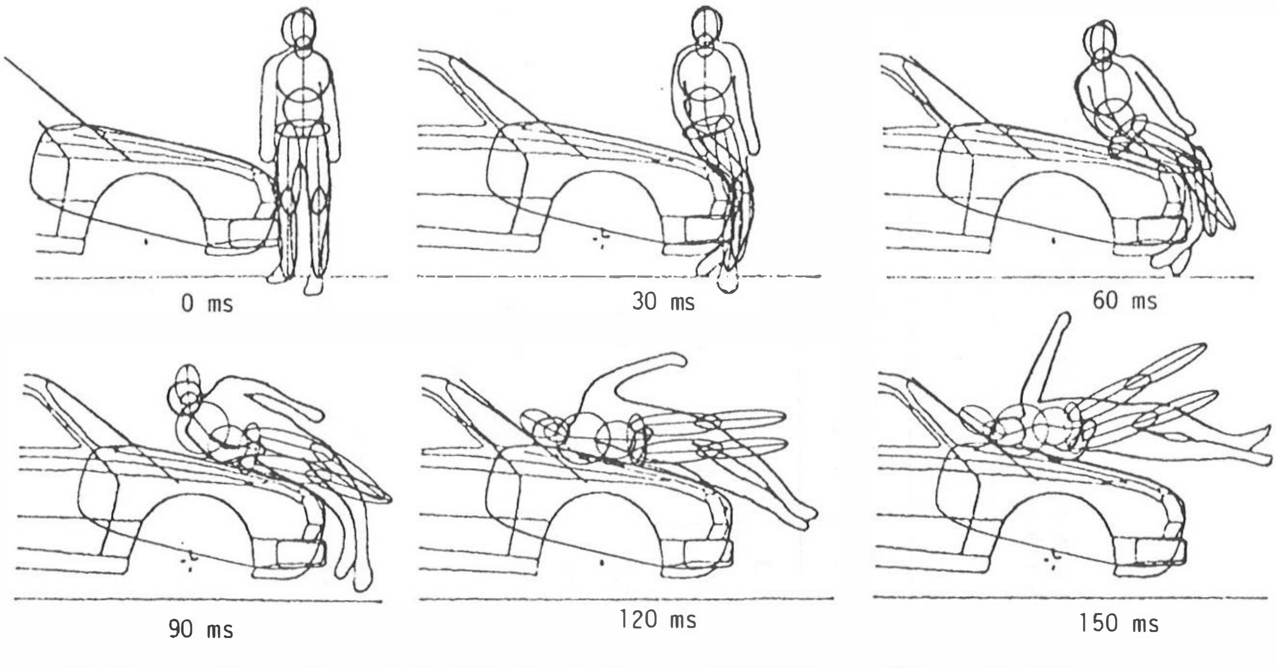

Figure 5. Comparison of Pedestrian Trajectory Sequence for the Model and a Cadaver Test (Citroen GS - FOC 52).

TABLE 2: RESULTS FOR HEAD IMPACT TIME, LOCATION AND VELOCITY FOR SIMULATION COMPARED WITH CADAVER RESULTS (3) AND (15). ATD RESULTS ALSO SHOWN

	CITROEN GS						CITROEN VISA					
	TIME (ms)		DISTANCE (m)		VELOCITY (m/s)		TIME (ms)		DISTANCE (m)		VELOCITY (m/s)	
	Sim.	Cad.	Sim.	Cad.	Sim.	Cad.	Sim.	Cad.	Sim.	Cad.	Sim.	Cad.
Number of tests	3	3	3	3	3	3	4	4	4	4	4	4
Mean	127	133	1.34	1.32	11.0	11.8	128	134	1.36	1.31	8.7	8.7
Standard Deviation	7.8	23.1	0.14	0.26	1.6	2.2	12.3	20.7	0.14	0.17	1.3	0.7
Maximum	136	160	1.50	1.62	12.1	14.3	138	162	1.46	1.49	10.5	9.4
Minimum	122	120	1.24	1.13	9.2	10.2	110	112	1.16	1.07	7.4	7.7
p*	0.548		0.805		0.456		0.564		0.646		0.942	
ATD(n = 1)	110		1.06		15.4		126		1.28		11.6	

* p value for test of no difference between the mean values for the simulation and the cadaver test.

TABLE 3: PEDESTRIAN ACCIDENT RECONSTRUCTION DATA (15)

TEST NUMBER	HEIGHT (cm)	WEIGHT (kg)	VEHICLE	IMPACT SPEED m/s	HEAD INJURY SEVERITY (AIS)	CAUSE
FOC 10	167	55	GS	10.92	5	Plenum & Ground
FOC 12	185	77	GS	10.93	5	Plenum & Ground
FOC 52	166	62	GS	10.61	3	Plenum
FOC 06	176	50	VISA	10.87	1	Windshield
FOC 08	180	63	VISA	10.79	1	Windshield
FOC 41	173	63	VISA	10.78	1	Windshield
FOC 45	156	70	VISA	10.78	1	Windshield

3.2 Discussion of Results

The mathematical simulations which we have carried out, using an ATD data set, (Table 2) of a pedestrian impact with the Citroen GS and VISA yield head impacts located nearer the bonnet leading edge, of higher impact velocity, and sooner after the initial impact than were recorded in cadaver tests. Similar findings are reported from comparisons of actual cadaver and dummy impacts (3, 15, 16).

Changes in model parameters to more closely resemble human characteristics have led to results much closer to cadaver tests. While the predicted head impact occurs slightly early (5% on average) and slightly further rearward (3%) the simulation results all fall within the variability of the test results (as seen in Table 2). The average head impact velocity predicted by the model is 3% lower than measured in the reconstructions. Two of the reconstructions contained apparent anomalies which have led to more substantial discrepancies when compared with the simulation. In FOC 41 the legs of the cadaver appear to have been trapped by the bumper to a greater extent than is usual as evidenced by a large number of fractures in the lower legs. This may have been due to weakened bones in the cadaver, but the result was an early head impact close to the front of the vehicle. In FOC 12 a particularly tall cadaver appears to have slid more over the vehicle bonnet than usual. This could have been due to the pelvis of this tall cadaver contacting the less compliant area over the engine of the Citroen GS. The lower deformation would have absorbed less energy, so allowing a higher sliding velocity over the vehicle bonnet.

Some inadequacies continue to be inherent in the model. It is not possible to match the motion and energy absorbing characteristics of the human frame with such a simple model. On impact the cadaver legs absorb considerable energy through wrapping around the vehicle. When a cadaver impact is viewed on high speed film the joints no longer exist as discrete discontinuities in the structure. There is a gradual flowing of the body around the front of the vehicle. This tendency is increased by the fractures to the lower legs and pelvis, such as were recorded for the cadavers involved in the testing. The inadequate modelling of the energy absorbed in the lower legs imparts in some cases too high a rotational velocity to the body of the pedestrian leading to the head impact being short, early and having too high a velocity.

The energy absorbed in the joint motion as a result of the torso flexing in the simulation also seems to be insufficient. This can be seen in the higher joint rotations for the simulations in Figures 4 and 5. A similar problem occurs with the neck where the head in the simulation remains too upright throughout the impact. The response of the simulation could be improved in these areas by increasing the joint damping and possibly adding some friction, but there is no physical evidence available to justify this.

By using a two dimensional model no energy is transferred into axial rotation of the pedestrian, this also tends to increase the velocity of the head impact in the simulation. An attempt has been made to minimise this effect in this set of tests by the initial conditions chosen for the reconstructions; an almost straight lateral impact on the cadaver.

4. CONCLUSIONS AND FUTURE DEVELOPMENTS

The validation exercise indicates that the pedestrian model is able to provide results of sufficient accuracy to fulfill the requirements of the study. A reasonable agreement with cadaver reconstructions (which for the purposes of the study are accepted as being the most satisfactory human surrogate) can be expected, given input data of sufficient quality.

We hope to be able to improve the model in a number of areas:

- More directly applicable biomechanical data on the human body will be used as available.
- The lower leg model needs to be refined to better simulate the kinematics of a cadaver and real pedestrian. The new version of MADYMO will facilitate this; by allowing flexible mounting of the pivot points within a segment it will no longer be necessary to use inextensible links. If the knee and hip joints are flexibly mounted with appropriate damping considerable improvements are expected.
- There is a need to move to a three dimensional model both to remove the constraint on axial rotation and to allow three dimensional predictions of rotational acceleration.

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