USE OF MADYMO'S HUMAN FACET MODEL TO EVALUATE THE RISK OF HEAD INJURY IN IMPACT

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ABSTRACT

MADYMO® rigid-body models are widely used in the automotive industry for a range of occupant protection related applications. These models have been evaluated at various levels against a range of experimental conditions including blunt impacts. To date the greatest focus for head impacts has been the study of severe impacts. It appears beneficial to broaden the field of validation of these models, and to expand the knowledge of tolerance limits associated with lower severity injury. In this case, mild traumatic brain injury (MTBI).

A simulation protocol was developed using MADYMO's human facet models to reconstruct 27 real-life concussive head impacts from impact sports. The cases were selected from a set that had been studied previously using a video analysis protocol. The contact behaviour of the model was first evaluated against both experimental and numerical results available in the literature. The video impact cases were then reconstructed and simulated, allowing for the assessment of a range of global biomechanical parameters that have been shown to be correlated with injury risk. The reliability of these reconstructions was evaluated by means of a sensitivity analysis of the influence of several independent variables on these dynamic outputs.

The results showed that the use of MADYMO's human facet model was adequate to obtain a representative estimate of head dynamics associated with soft to medium impact severities. They also hinted at the model's limitations to accurately model short impact durations impacts. The following mean peak values for MTBI were obtained from the reconstruction of the real-life impacts: 103 g for the head centre of gravity linear acceleration, 8022 rad/s2 for the head angular acceleration and 359 for the HIC.

These values compare well with other studies and should contribute to the identification of the level at which injury first occurs.

INTRODUCTION

The head is exposed to the risk of impact and consequent injury in many areas, eg. transport, recreation, sport and assault.

In automotive accidentology, focus has been drawn so far on mitigating the risks of moderate to severe injuries, this mitigation being a priority in such impacts as pedestrian (Chidester and Isenberg 2001; Otte and Pohlemann 2001), rollover (Otte and Krettek 2005) or lateral impacts (Digges and Dalmotas 2001). Early approaches to understanding the mechanisms of head injury and the tolerance of the head to impact relied on human cadaver or animal experimentation and subsequent medical assessment of the injuries (Ommaya *et al.* 1967; Gennarelli *et al.* 1972; Ono *et al.* 1980). The advent of improved computing and numerical modelling techniques then provided additional methods of study (Ruan *et al.* 1993; Willinger *et al.* 1994; Zhou *et al.* 1995). In particular, mild traumatic brain injury or concussion had not lent itself well to cadaver or animal models, due to the functional nature of the injury and ethical issues; numerical techniques have proven a promising method to investigate this range of energy levels (Zhang *et al.* 2004). As MAIS injury levels have decreased in the last 20 years (Kullgren *et al.* 2002) thanks to improved passive measures, it appears that precise estimates of the risk of injury for contacts with softer parts of a vehicle (eg. dashboard) or other occupants may also benefit from both an improved modelization of the impact and a better knowledge of associated injury levels.

The availability of video of sports head injury events, specific medical information, and numerical methods has provided a new avenue for biomechanical analysis of the mechanisms of mild to moderate severity head injury and related tolerance limits (McIntosh *et al.* 2000; Pellman *et al.* 2003; Zhang *et al.* 2003). Sport provides the opportunity to study impacts that lead to concussion as these events are often filmed and the injured athlete is thoroughly assessed, especially in professional sport.

In this purpose, the MADYMO rigid-body modelling software package was used to simulate real-life concussive cases and to evaluate the dynamics associated with injurious levels. An evaluation of the ability of the model to describe impact dynamics was first performed in order to evaluate the reliability of these simulations and their implications for safety design strategies. Previously recorded real-life concussive head impacts between football players were then reconstructed using MADYMO.

This paper presents the design of this protocol, including an evaluation of the head contact properties, a parametric study of the main parameters of the impact, and the results of the simulations.

MADYMO's human facet model's behaviour had been validated previously (TNO 2005) for various blunt impact locations (thorax, shoulder and pelvis), this was not the case for it's head. Therefore it was decided to improve the contact properties based on recent experimental data available in the literature, and to evaluate the resulting behaviour.

MATERIALS AND METHODS

Background to the modelling

A set of a hundred videos of concussive impacts in both Rugby and Australian Rules Football was analysed and reported previously (McIntosh *et al.* 2000). For each player involved in an impact, anthropometric data (mass, height), conditions of the impact (location of the impact, head orientation, impacting segment) as well as medical assessment of the injury (definition and duration of the symptoms, concussion grade) had been collected. The kinematics of the players were then estimated, based on a 2D analysis of the videos. To refine these first calculations and precisely take into account their out-of-plane components, a 3D numerical analysis was chosen for the study presented here. Depending on the nature and duration of the impacts, an influence of the neck on the head dynamics could be expected (Beusenberg *et al.* 2001). Furthermore, a realistic simulation would depend on accurate modelling of the effective masses involved in the impact, and it was decided to model the whole players. The MADYMO human facet model's behaviour had been previously validated against several sled test as well as blunt test impact configurations (TNO 2005). Due to a more representative geometry, the contact behaviours were expected to be more precise than for the equivalent ellipsoid model. Moreover, its relative simplicity compared to an FE model allowed for easier parametrisation. For these reasons MADYMO's facet models were chosen for these simulations. The following flow-chart (figure 1) describes the methodology of the study, the grey blocks corresponding to the three stages associated to the numerical reconstruction and simulation process.

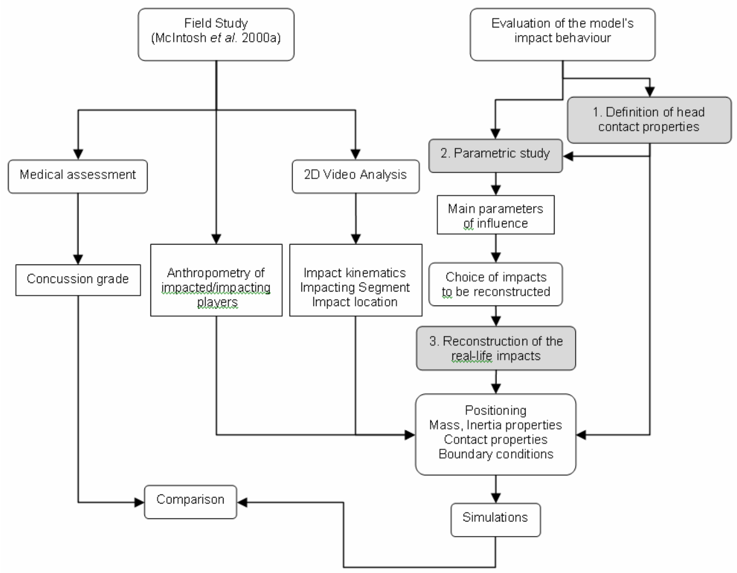

Figure 1. Flow-chart of the study's three stages.

Definition and evaluation of the model's head contact properties

Management of the contact in numerical impact models is of critical importance and after a preliminary run of the parametric study presented below it appeared that the contact characteristics had a significant influence on the model's head impact responses. Furthermore, and although

Earlier versions of MADYMO did not allow combining the contact properties of two impacting surfaces. Therefore, combined contact characteristics had to be defined by the user or one of the objects (eg. the head) had to be defined as a rigid body. Versions 6.2 and later now authorize a combined calculation based on the contact properties of each surface and on their respective penetration, which allows for a more accurate modelling of the contact behaviour, especially in the case of two surfaces with similar contact properties. As they included both quasi-static and dynamic test conditions, the results from (Yoganandan *et al.* 1995) were used to refine the contact properties of the head model. In this experimental study, twelve unembalmed cadaveric head segments were rigidly fixed and impacted by a hemispheric rigid anvil at various locations of the head (resp. frontal, occipital, parietal, temporal and vertex impacts) and the force-deflection characteristics were measured. The loading conditions included quasi-static tests at 0.002 m/s and dynamic loadings at 7.5 ± 0.35 m/s. These two conditions (geometry, positioning, loading velocity) were reproduced with MADYMO. A quasi-static stress/penetration characteristic was defined in the model so as to obtain a good fit, respectively between the average quasi-static experimental and simulation force-deflection curves. Damping amplification properties (see appendices) were then defined to fit the 7.5 m/s dynamic results, allowing for the definition of a complete contact characteristic. Experimental data from (McIntosh *et al.* 1993; Yoganandan *et al.* 2004) as well as FE simulation results from (Neale *et al.* 2004) were used to evaluate this behaviour.

(McIntosh *et al.* 1993) impacted seated human cadavers at head level, respectively in lateral and occipital impacts with a pneumatic impactor. Tests included both unpadded and padded (25.4 mm thick Ensolite®) impact conditions for three different velocities. Boundary conditions were clearly defined for these protocols and pulse durations of impact force and head accelerations were available. These conditions were reproduced with MADYMO and the results, in terms of impact force, head acceleration, and HIC were compared between experiment and simulation.

Published experimental results from (Yoganandan *et al.* 2004) were also used for the specific case of lateral impact. In these drop-test experiments, ten unembalmed cadaveric head specimens were dropped on a 50 mm thick, 40 Durometer material padded anvil, in order to obtain impacts in the temporo-parietal area. Impact velocities were up to 7.7 m/s and results included corridors of the measured force and acceleration responses. The boundary conditions of these tests were reproduced and simulated with MADYMO. The contact behaviour of the padded surface was defined based on Sorbothane® force/deflection characteristics. Force and acceleration results were compared with the experimental corridors.

As the simulation protocol and material properties were clearly described and included several impact conditions, results of Finite Element head drop-test simulations by (Neale *et al.* 2004) were finally used to evaluate the contact options for the head. In these drop-test simulations, a validated FE head model impacted an elastic block whose Young's modulus was chosen with values ranging from 0.63 to 25 MPa in order to control the impact durations (from 20 to 6 ms). The coefficient of friction was 0.3 and impact velocity 4.44 $m.s^{-1}$. To evaluate the MADYMO head model behaviour, frontal and parietal impacts to the head were modelled, reproducing the above described characteristics and boundary conditions for Young's moduli of respectively 3 MPa and 25 MPa. Results were compared in terms of acceleration of the head's CG, as well as contact forces.

Parametric study

Impacts between players were reconstructed and simulated using numerical rigid-body models in the present study. There are many degrees of freedom in such models, and assumptions regarding the model's geometry and mechanical properties influence the results. Errors may also come from the case reconstruction process, for example from the transfer of boundary conditions (eg. velocity) assessed on the videos, to the model. Therefore, before reconstructing the real-life impacts, a study of the influence of various independent parameters on the kinematics and the dynamics of the head impact was performed.

A standardized protocol was chosen, where the full body model was positioned in a seated position and its head impacted horizontally by a spherical object. A parametric study was then performed, to assess the influence of the change in six independent variables (see table 1) on the results, when going from a low level (-1) to a high level (1) around a reference level (0) value.

Table 1. Parameters and their low/high levels

Variable	-1 / Low level	1 / High level
Velocity (m/s)	3.6	4.4
Position (cm)	Initial position -4 cm	Initial position + 4 cm
Orientation (degrees)	Perpendicular to sagittal plane	Perpendicular to sagittal plane + 20 deg
Neck stiffness (N.m)	No restraint moment	"aware" condition
Contact stiffness (N/m^2)	MADYMO limb contact stiffness – 20 %	MADYMO limb contact stiffness + 20 %
Friction coefficient	0.2	0.5

This protocol resulted in two matrixes of 64 simulations that were performed for two different changes (forward and backward) in horizontal position for the purpose of the sensitivity analysis. Simulations of the intermediate impact positions (level "0") were also performed as a check for consistency and results distribution.

In the video analysis (McIntosh *et al.* 2000), the minimal closing velocity for concussion was found to be 4.2 m/s. In the same study, the error in evaluating the velocity of the players was estimated to be less than 10 % on this set; as cases with potentially high parallax error had been excluded. Thus a mean velocity of 4 m/s, with a deviation of +/- 10 % was chosen to define the associated high and low level of this parameter in this study.

The reconstruction of the initial position of the players in the real-life impacts was performed by assessing the videos frame-by-frame. For many videos, these frames were blurry and they did not allow a precise assessment of both the location and orientation of the head impact. In order to take into account these potential sources of error, the parametric study included simulations where the centre of impact was varied with regard to an initial position (see figure 2). As this effect was thought to be maximal on the axial rotation of the head, the positions were chosen on a horizontal plane. For the same reason, the orientation of the blow was varied 20 degrees around an initial lateral impact direction (figure 2).

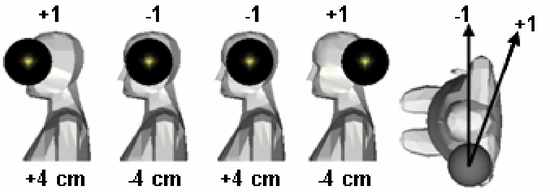

Figure 2. Positions and orientations used for the impactor, with associated low and high levels.

As the neck stiffness, in these cases representative of the player's level of muscular activation and strength, was expected to possibly influence the results, two levels were taken into account in the parametric study. The high level, corresponding to an 'aware' state, was modelled by adding restraint torques at neck level to model muscle contraction. The values, 52 N.m in extension, 30 N.m in flexion, 12 N.m in axial rotation and 31 N.m in lateral bending, were chosen at 80% of the range of maximal isometric neck torques defined in a review by (Portero and Genries 2003), the 80% threshold of maximal voluntary force having been proposed by (Mertz *et al.* 1997) for N_{ij} calculations with aware occupants in frontal impact. No restraint torques were added to the passive properties of each cervical level for the low level.
Finally, contact properties influence the results (Camacho *et al.* 1999); they may depend on the subject and on the location on the body. Therefore, the limb contact characteristics present in MADYMO and used for this analysis where varied within +/- 20% of their mean value to assess this influence. In a review by (Sivamani *et al.* 2003), dynamic friction coefficient for the skin was found to range from 0.2 to 0.7. As a value of 0.34 had been described for the forehead, and as high values induced noise in the calculations, the coefficient was varied between 0.2 and 0.5 in this study.
Simulations were performed on a 200 ms time frame with a time step of 10^{-3} ms, allowing for the description of both the impact and the kinematics shortly thereafter. The dependant variables chosen as output were the Head Impact Power (HIP)- (Newman *et al.* 2000b), Head Impact Criterion (HIC_{15}), 3ms and peak linear (at the head's CG) and angular acceleration of the head.

Reconstruction of the real-life impacts

Following the results of the parametric study, and in order to limit the effects of possible error in the assessment of the impact velocity due to the 2D analysis, 27 cases out of the 100 from the initial database were selected and reconstructed. These videos were chosen based on their clear description of the event; allowing for both the determination of accurate boundary conditions and the assessment of the reliability of the simulations. In particular, videos were chosen where the closing movement of the players occurred in the plane of the camera, to minimize errors made in the calculation of the initial velocity. In all cases head injuries had been well document and concussion graded according to the following criteria: Grade 1 – no loss of consciousness (LOC); Grade 2 – LOC < 1 min; and, Grade 3 – LOC > 1 min.
The simulations of an impact between two players were performed using the following protocol: first, the models were positioned using HyperMesh® to reproduce the relative position of each player just before the impact (figure 3).

Figure 3. Reconstruction of the relative positions before impact.

The masses and inertias of each model's body segments were calculated based on the known anthropometry of the players and GEBOD (Cheng *et al.* 1994) scaling equations. They were then inputed into the models. The initial velocities of each human model were the closing velocities previously assessed during the video analysis. The parametric study had concluded that neck stiffness had a low influence on the head behaviour compared to other variables. Furthermore, it was difficult to assess the awareness of the injured players on some of the videos. Therefore, a generic "unaware" state was modelled: joint restraint torques were input into the model so that it could just maintain the standing upright position in a pre-simulation. Finally, the initial position of the model, the initial velocity of each body segment and the stiffness of each joint were tuned in order to obtain a satisfactory match between the kinematic behaviour of the impacting bodies compared to the real event on video. The restraint torques used for the shoulder, elbow, hip, knee, and ankle, were chosen in the range of the values proposed by (Stobbe 1982). The simulation period was 200 ms which incorporated both the impact and immediate post impact kinematics. All simulations were run using HyperMesh v6.0 and MADYMO v6.2.2.

Biomechanical Output Data

In order to compare the results with the existing video analysis data, the Peak Velocity Change (PVC), impulse and impact energy of the head were calculated. For means of comparison with the literature, the impact energy was calculated as the energy needed to allow the head's peak change in velocity, allowing the definition of an equivalent drop-test impact energy. The impulse was calculated at the same time of peak change. The head's CG linear acceleration, head angular acceleration, HIC_{15} and HIP were also calculated in order to study the biomechanics of concussion and for comparison with the literature.

RESULTS

Definition and evaluation of the model's head impact properties

An example of simulations of an impact following each protocol (McIntosh *et al.* 1993; Neale *et al.* 2004; Yoganandan *et al.* 2004) is presented in figure 4.

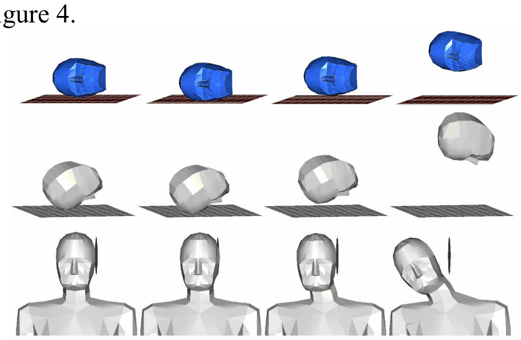

Figure 4. Simulations of Yoganandan et al., Neale et al. and McIntosh et al. (from top to bottom) impact conditions.

<u>Simulation of Yoganandan's experiments :</u> Figure 5 presents a comparison between the simulation and experimental results for this series of drop-tests. Experimental results present the average and standard deviation of the 10 tests. Simulations results present outputs for simulations with the min/average/max head weight from the experiment. Both the peak force and acceleration compare well with the experiment for this relatively soft impact.

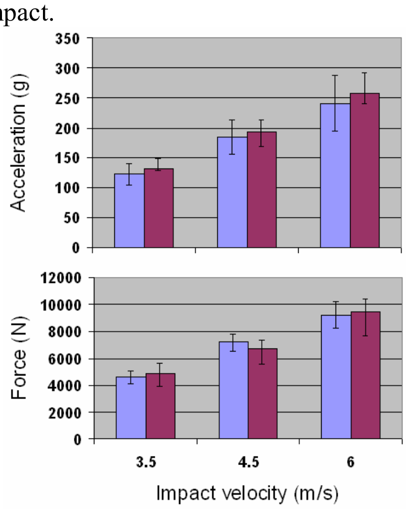

Figure 5. Comparison of experimental (in blue) and simulation (in red) peak force and acceleration.

<u>Simulation of Neale's simulation protocol :</u> The results are presented in table 2 and table 3. In the 3 MPa frontal and parietal impacts, differences between simulation and experimental results are within 11%. In the 25 MPa impacts, the same trend is observed for the peak force and linear acceleration. However, there are large differences (up to 50 %) between the simulated and experimental peak angular accelerations for both impacts. For the 25 MPa impacts, force pulse durations were significantly higher (up to 25%) in the MADYMO simulation.

<u>Simulation of McIntosh's experiment :</u>

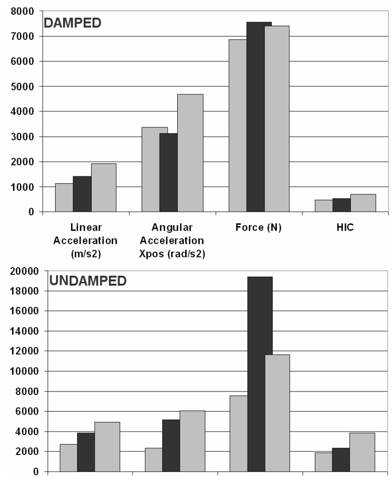

Figure 6. Comparison between simulation and experimental results from (McIntosh *et al.* 1993) for the Aluminium (undamped) and Ensolite (damped) 5.9 m/s parietal impacts.

Figure 6 presents peak values in linear acceleration of the head's CG, Force and HIC for the Aluminium (5 ms) and Ensolite (11 ms) parietal impacts, compared between simulations and experiment. Results are within, or close to the experimental range of values for both impacts. However, for the undamped impact, the peak force output is significantly out of the experimental corridor by 66%.

Parametric study

Table 4 gives a statistical description of the 192 simulations set (including intermediate positions), and Figure 7 shows an example of the distribution of the mean HIC values and peak angular acceleration of the head for the various positions. As none of the outputs were found to be normally distributed, the relative influence of each of the parameter was assessed by representing it as a scatter plot with the means (see figure 7 for an example of the effects). In order to compare the influence of each parameter, changes in the variables between low and high level were normalized by expressing them as a percentage of

Table 2. Comparison between simulations and (Neale *et al.* 2004) for the 12ms and 6ms frontal impacts

FRONTAL IMPACT					
	Acceleration		Force	Duration	HIC
	Linear (g)	Rotational (rad/s^2)	(N)	(ms)	
E = 3 MPa					
Mean	144	1839	6700	12.5	906
(Neale 2004)	132	1727	6700	12.0	-
E = 25 MPa					
Mean	248	4293	11550	7.5	1536
(Neale 2004)	231	8510	11900	6.0	-

Table 3. Comparison between simulations and (Neale *et al.* 2004) for the 12ms and 6ms parietal impacts

PARIETAL IMPACT					
	Acceleration		Force	Duration	HIC
	Linear (g)	Rotational (rad/s^2)	(N)	(ms)	
E = 3 MPa					
Mean	151	3372	7026	12.0	1019
(Neale 2004)	140	3774	6800	12.0	-
E = 25 MPa					
Mean	222	5313	10283	7.4	1450
(Neale 2004)	210	7773	11800	6.0	-

Table 4. Descriptive statistics for each parameter, for 192 simulations, including intermediate positions

	HIP (W)	HIC	Linear acceleration (m/s^2)		Angular acceleration (rad/s^2)	
			3ms	Max	3ms	Max
Mean	9081	287	504	788	2981	4995
Stdev[1]	2624	124	125	183	711	1994
CV[2]	0.29	0.43	0.25	0.23	0.24	0.4
CL Sup.[3]	9393	302	519	810	3065	5232
CL Inf.[4]	8770	272	489	767	2896	4759

(1) Standard deviation

(2) Coefficient of variance

(3),(4) 95% Confidence Intervals of the mean, upper and lower limit.

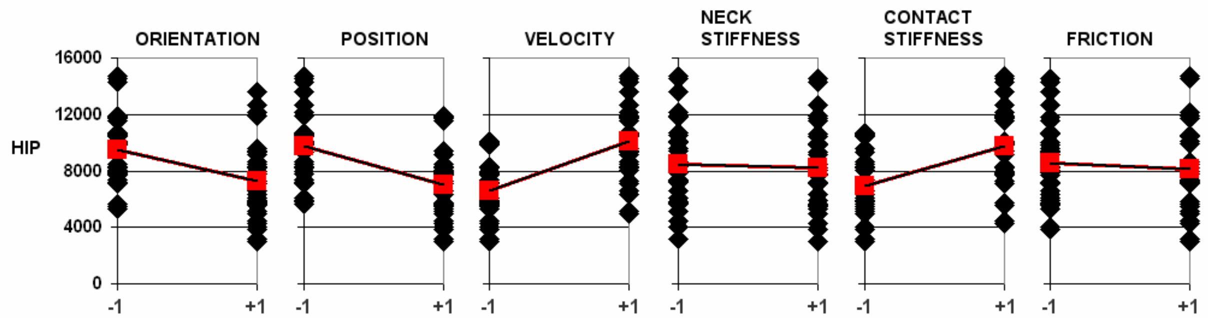

Figure 7. Scatter and mean plot of the influence of each of the six parameters on the HIP values, in the case of the rearward change in position.

the initial (low level) value. These percentages of change were then averaged between the two sets (forward and rearward direction of change), allowing a ranking of each parameter's influence on the output variable (see table 5). From these results, peak angular acceleration of the head is influenced dramatically by changes in position, while changes in velocity affect primarily HIC and HIP values. Changes in contact stiffness of the model also have a significant influence on each variable, although this is not true for the 3ms values (linear and angular acceleration). Friction coefficient, orientation of the impact and neck stiffness have relatively no significant influence. Overall, HIC value and peak angular accelerations of the head are the most influenced by change in the parameters, and 3ms values are the least. Finally, object contact stiffness shows the most influence on variables which depend on durations (HIC and HIP).
In the same way, influence of a combination of two variables, or cross-effects, were then evaluated for combinations of the major influencing parameters. Assuming that cross-effects between two variables would account for the main changes in the model's behaviour, the result show that a cross-effect between velocity and contact stiffness has an important influence both on the HIP and HIC values, while a combination of each of this variable with position influences mainly peak angular acceleration. These effects reached respectively 110%, 141% and 95% of the low level value. The 3ms accelerations, linear and angular are the least influenced, percentages being respectively 40 and 50%.

Real life impacts reconstruction

Figure 8 presents a visual comparison between one of the impacts and its simulation.

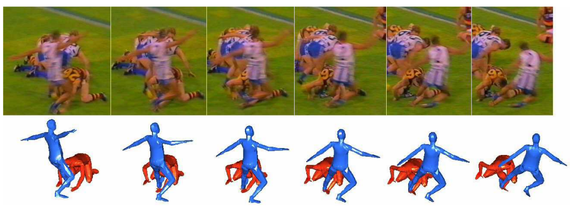

Figure 8. Simulation of an Australian Rules Football impact.

In this case (nb. 4) the player was hit by an opponent's knee while crouching to catch the ball. He was not aware of the incoming impact and suffered no LOC from it, resulting in a Grade 1 concussion classification. Table 6 presents a summary of the mean peak values and range for each biomechanical variable stratified according to each concussion grade and table 7 presents the results of each simulation. There were nine simulations for each grade of concussion. HIC values for concussion ranged from 87 to 994. The latter HIC was reached for one of the most severe impacts, where peak values of 200 J in impact energy and of 43 kg.m/s impulse were reached. The overall mean values for HIC, peak linear and angular acceleration were 359, 103g, and 8022 rad/s^2, respectively. Although some of the results demonstrated high standard deviations (respectively 68% and 69% of the mean value for HIC and HIP), a common trend between injury severity and some of the biomechanical parameters can be observed.

DISCUSSION

The initial aim of this study was first to improve the reliability of the case study video analysis performed previously, and to evaluate the dynamics associated with concussive head impacts.
Secondly we also aimed at evaluating the reliability of using a rigid-body software such as MADYMO to estimate these parameters. Although such models are used both in research and in the automotive industry to model pedestrian impacts, an evaluation of the facet model's head behaviour had not been documented before.
A parametric analysis was undertaken, which showed that the rigid-body model's head contact properties influenced the biomechanical parameters used as estimators of the risk of concussion. New contact characteristics were proposed, that allow taking into account the combined behaviours of the two impacting objects, and refine the existing ones by taking into account damping effects. As these contact properties proved to be adequate to model the relatively soft impacts between players, they were used to reconstruct 27 real life concussive impacts in order to obtain an estimate of the biomechanical parameters associated with this first level of injury.

Definition of the model's head contact properties

Results in terms of peak accelerations, peak forces and HIC values compare well with the experiments for the three sets of evaluations. However, significant differences are found for the peak angular accelerations and forces for the short-duration impacts (≤ 6 ms). In the simulation of Neale et al.'s protocol, the differences in angular rotations may be explained by a relative coarseness of the mesh in this version of the model, meaning that small differences in initial positioning between the head models may yield significant differences in their rotational behaviour.
In the same protocol, the peak force results of the simulation for the undamped impact show an important difference (66%) which is accounted mainly by high damping forces. At this stage, it is unclear if this short duration peak is an artefact of the simulation, due for example to the rough mesh, or if the head damping properties have to be

adapted for this kind of very short duration impact. As skull fractures were not modelled, but occurred in each of the undamped experimental impacts, simulations may yield these unrealistic results. Part of the above described inconsistencies may also relate with previous observations by (Neale *et al.* 2004), and may reflect the fact that a rigid-body model may not be accurate enough to model short-duration impacts because of the coupling/decoupling process involved between the brain and skull.

The impact conditions assessed in Yoganandan and al.'s experiments correspond to an impact into a polyurethane material, similar to a dashboard. The combined contact definition yields satisfying results for these impact conditions. Several limitations may be associated with the evaluation process presented here. First, the experimental force-deflection curve used to define the contact properties is an average of results obtained for several impact locations on the head. It is acknowledged that differences in bone properties and skull thickness will influence the local behaviour of the head, however at this stage our aim was to improve the existing modelling, and to obtain a reasonable estimate of the impact dynamics. The parametric study was intended to allow the definition of possible uncertainties. It is also acknowledged that the impact behaviour will depend from the modelling of the second impact surface (in our case, limb, thorax, abdomen or head of the impacting player). The associated MADYMO bi-linear contact properties had been evaluated previously by means of blunt test simulations based on PMHS experiments, and were used as such. They would however benefit from a refined definition for the purpose of improving the combined contacts approach.

Table 5.
Influence of each parameter (expressed in percentage of change from low to high level), ranked from highest to lowest

(%)	HIP	HIC	Linear acceleration		Angular acceleration		Average effect
			3ms	Max	3ms	Max	
Position	-21.1	-37.9	-37.8	-22	26.5	96.9	40.4
Velocity	51.8	76.9	13.8	25.6	19.8	23.7	35.3
Contact Stiffness	38.6	35.8	1.5	34.8	13.3	21.9	24.3
Orientation	-12.8	-10.5	-5.6	-7.1	-4.4	-16.6	9.5
Friction	-6.3	8.8	8.4	-0.7	-0.2	-5.5	5
Neck Stiffness	-2.4	-1.3	-1.3	-0.7	1.5	-0.90	1.4
Average Effect	22.2	28.5	11.4	15.1	11	27.60	

Table 6.
Mean peak values reached by the biomechanical parameters during the simulations

		Impact energy (J)	Impulse (kg.m/s)	HIP (W)	HIC	Acceleration Linear (g)		Angular (rad/s^2)		PVC[1] (m/s)	Duration (ms)
						3ms	Max	3ms	Max		
Grade1[2]	Mean	63	24	8830	231	64	86	4380	7240	5.0	21
	Min	25	15	4600	87	47	60	2010	3470	3.2	8
	Max	103	32	14990	471	81	100	9500	14720	6.5	46
Grade2[2]	Mean	82	27	11030	333	72	101	4760	7350	5.7	24
	Min	28	17	5080	111	50	60	2700	3880	3.4	7
	Max	164	40	15550	976	109	183	7100	15130	8.3	54
Grade3[2]	Mean	105	31	21280	513	93	123	5650	9470	6.5	12
	Min	51	22	6800	232	70	84	2950	5100	4.7	7
	Max	200	43	53990	994	122	152	10900	16450	9.3	16
All	Mean	83	27	13715	359	76	103	4930	8020	5.8	19
	Min	25	15	4600	87	47	60	2010	3470	3.2	7
	Max	200	43	53990	994	122	183	10900	16450	9.3	54

[1] Peak Velocity Change

[2] Grade 1: no LOC
Grade 2: LOC < 1 min
Grade 3: LOC > 1 min

Table 7. Peak values reached by each biomechanical parameter for each case

Case	Grade	Impact energy (J)	Impulse (kg.m/s)	HIP (W)	HIC	Linear acc. (m/s^2) (3ms)	(Max)	Angular acc. (rad/s^2) (3ms)	(Max)	PVC (m/s)	Duration (ms)
1	1	78	27	4604	471	799	934	5780	7659	5.8	9.1
2	1	47	21	7788	142	523	710	2610	4070	4.5	14.1
3	1	46	20	8614	87	464	586	7050	9806	4.5	42.0
4	1	59	24	12437	218	653	795	2410	5957	5.0	11.8
5	1	39	20	5536	208	596	980	3085	8526	4.0	8.3
6	1	86	29	6614	294	729	980	2950	3466	5.9	12.4
7	1	25	15	12784	127	520	796	9500	14718	3.2	7.7
8	1	82	29	6101	229	640	906	2010	4379	5.7	42.0
9	1	103	32	14990	301	723	930	4020	6594	6.5	46.0
10	2	78	27	5078	178	542	596	2700	3881	5.8	21.7
11	2	28	17	12963	203	571	963	6345	9301	3.4	7.4
12	2	164	40	14831	250	721	921	4020	5670	8.3	42.0
13	2	93	29	12867	585	983	1338	5150	8285	6.4	9.5
14	2	47	21	6955	111	492	588	5410	6550	4.6	20.7
15	2	106	32	8091	238	674	905	4050	6560	6.6	40.0
16	2	53	23	15549	241	641	1001	3460	4697	4.7	54.0
17	2	62	24	9724	214	651	799	4630	6087	5.2	14.1
18	2	110	33	13233	976	1065	1793	7100	15133	6.7	7.8
19	3	78	28	14074	501	878	1432	6200	13160	5.6	7.0
20	3	120	34	21145	641	1031	1266	4541	7282	7.1	11.3
21	3	99	30	6799	405	900	1136	2950	6366	6.5	14.8
22	3	136	36	30256	684	1078	1300	3400	7738	7.5	12.2
23	3	103	31	19668	422	867	1077	4000	5100	6.5	12.0
24	3	51	22	53993	293	725	1108	10900	16450	4.7	9.8
25	3	77	27	15000	232	687	827	5240	9079	5.8	15.5
26	3	200	43	16258	994	1200	1490	8900	10935	9.3	13.2
27	3	79	27	14358	443	888	1209	4700	9132	5.8	10.7

Parametric study

The results of this study show that errors in evaluating both the exact location and the velocity of the impacting objets may have the strongest influence on the biomechanical parameters used as estimators of the risk of concussion (see table 5). These parameters (HIC, HIP, head linear acceleration and angular acceleration) had been previously shown (Newman *et al.* 2000a; Zhang *et al.* 2004) to be acceptable estimators of the injury risk.

Beusenberg et al. (2001) emphasised that the head's behaviour during an impact simulation was dependant on the modelling of the neck with consideration for the rest of the body. This is not the case in this study, and may be explained by the importance of effective masses and compressive loading mechanism of the neck in Beusenberg's simulation protocol; no such loading direction was performed in our real-life scenarios. Also, the range chosen for the low and high level in our parametric study have an influence on the results. They were however chosen carefully to be representative, either of possible errors in the reconstruction of the boundary conditions, or of possible fluctuations in the model's degrees of freedom. Results may also have varied depending on the initial location of the centre of impact. For this reason, the changes in variables were averaged for changes both in the forward and in the rearward direction. Intermediate positions were also simulated in order to check the consistency of the evolution in behaviour.

These results showed that estimating precisely the position and velocity were important for the real-life case study simulations. This suggested restricting the ongoing reconstructions to impacts that were in the plane of the videos and where the location of the impact could be estimated precisely. Therefore, out of the initial 100 videos available, a set of 27 cases was chosen, where these constraints were met.

Results showed that the model's contact stiffness properties also had an influence on the results, and for this reason the evaluation of the model's head behaviour was undertaken. Following this study, the impact durations of the real-life reconstructions (7-54 ms) were deemed long enough to ensure that the results were not influenced by errors due to short duration impacts reconstructions with MADYMO.

Finally, the low and high levels were chosen to allow a full range of deviation (assuming for example an uncertainty of 20% in estimating the velocity, or of 8 cm in positioning). If we assumed a worst-case scenario with a cross-effect of the two main influencing effects, the boundary of the associated uncertainties would range from +/- 20 % for the 3ms linear and angular acceleration, up to +/- 70 % for the HIC value.

Reconstruction of real-life impacts

Table 8 presents a comparison of the mean peak values reached with the MADYMO simulations with the ones calculated from the video analysis, for the same set of 27 cases and for the whole set of 100 cases. The results show a similar trend between the simulation and the video analysis although mean peak values for the simulations were slightly higher. This difference may be explained by the fact that numerical reconstructions allowed consideration for velocities after impact that were out-of-plane, which could not be evaluated in the previous 2D video analysis.

Table 8.
Compared results for the simulations and previously performed video analysis

SIMULATION			VIDEO ANALYSIS (same 27 cases)			VIDEO ANALYSIS (all 100 cases)		
Impact energy (J)	Impulse (kg.m/s)	PVC (m/s)	Impact energy (J)	Impulse (kg.m/s)	PVC (m/s)	Impact energy (J)	Impulse (kg.m/s)	PVC (m/s)
83	27	5.8	73	24	5.2	67	23	4.8

Numerical studies on head injury biomechanics have been performed previously, and a large number were aimed at describing the mechanisms of injury and therefore used more precise FE models (Ruan *et al.* 1993; Miller *et al.* 1998; Kleiven and Von Holst 2002; Zhang *et al.* 2004). Only a few studies reported on real life accident reconstruction, either using dummy human surrogates (Newman *et al.* 1999; Pellman *et al.* 2003) or numerical models (Baumgartner *et al.* 2001; O'Riordain *et al.* 2003; Zhang *et al.* 2004). Some of these results are presented and compared with our results in table 11. The results of the present work show that concussion occurs for similar values of HIC, HIP, and accelerations as in similar studies.

Table 9.
Comparison with similar studies

	Impact energy (J)	HIC	HIP (W)	Acceleration Linear (g)	Angular (rad/s^2)	PVC (m/s)	Risk of Concussion
(Newman 2000)		240 485	12790 20880	78 115	6322 9267		50% 95%
(Pellman 2003)	118	381	-	98	6432	7.2	Mean value
(Zhang 2004)		240 369	- -	82 106	5900 7900		50% 95%
This study	83	359	13715	103	8020	5.8	Mean value

These results are encouraging, as they show realistic head dynamics. It is however difficult to assess both the presence and severity of all injury types. Reasons are numerous and relate to the difficulty in taking into account variability both in the injured human and in the impact situations and the range of experimental data acquired. Indeed, the validity, from content to external, of global mechanical parameters and injury criteria to assess injury risk remains a point of argument and controversy (King *et al.* 2003). For example, the HIC is based on the experimental assessment of the presence or absence of fractures and it is a significant extrapolation to use it as a general predictor of concussion or MTBI, as it is now characterised in sport. Nevertheless, some studies have found significant correlations between global criteria and the risk of concussion (Ruan *et al.* 1993; Newman *et al.* 2000a; Zhang *et al.* 2004). They are also simple to calculate, generally highly reliable in impact testing, and may be used effectively for means of comparisons. Some, like the impact energy are a simple approach by which an equivalent impact energy for testing can be determined.

Although no control (no-injury) cases were included in this study, our results are in close agreement with previously (Zhang *et al.* 2004) suggested values for a tolerable reversible brain injury. As Grade 1 MTBI's were associated with mean HIC values of 230, HIP values of 8830 and combined linear and angular acceleration of respectively 86 g and 7240 rad/s^2, these could be added to the pool of existing tolerance values proposed for this specific injury.

CONCLUSION

Twenty-seven cases of medically verified concussion from rugby union and Australian football were reconstructed using numerical simulation. The simulations were able to refine and add to data obtained from a previously performed video analysis. By modelling real-life concussive head impacts, the results of this study allow us to precise the knowledge of biomechanical tolerance levels associated to the presence of an injury.

Results from the sensitivity study show that HIC values and peak angular accelerations of the head are significantly influenced by both the degrees of freedom in the model and the boundary conditions of the impact. These conclusions oriented us to restrict drastically the number of reconstructions in the real-life study that followed, and to evaluate the behaviour of the model's head during impact.

In particular, these real-life reconstructions allow presenting the following findings:

- Grade1 concussions occurred for impacts involving mean energies of 60 J, and impulses of 24 $kg.m.s^{-1}$. These values confirm previous findings and may contribute to the design of experimental testing procedures.
- Based on our results and similar studies (Pellman *et al.* 2003; Zhang *et al.* 2004), suggested tolerance values for concussion are as follows: 230 for HIC_{15}, 8830 W for HIP, 85 g and 6000 rad/s^2 for combined peak linear and angular acceleration of the head.

Finally, the evaluation performed in this study is a contribution towards an improvement in the use of head impact models in rigid body simulations. Results from this evaluation suggest that, although the behaviour has to be improved for short impact durations, HIC values, forces and peak linear acceleration of the head's CG obtained by using a rigid-body model with adequate contact characteristics are representative of real-life impacts. As they were not evaluated against experimental results, or presented significant differences with previously published experimental or simulation data, respectively HIP values and angular acceleration of the head should be assessed with caution.

Despite these restrictions, using human rigid-body models in impact present several advantages when the aim is to study the risk of injury associated with real-life accident reconstruction. It is also believed that this approach may be beneficial as an input to more refined simulations more focused at assessing the associated injury mechanisms.

APPENDICES

$$\sigma = \sigma_e + \left[C_d \frac{\dot{\lambda}}{t} \right] . f_d \qquad (1)$$

$$\sigma_e = f\left(\frac{\lambda}{t}\right), f_d = f(\sigma_e)$$

σ_e = elastic (quasi-static) stress function

C_d = damping coefficient (0.1333)

f_d = damping amplification function (difference between quasi-static and dynamic functions)

λ = penetration

t = surface thickness (normalized to 1.0 in our case)

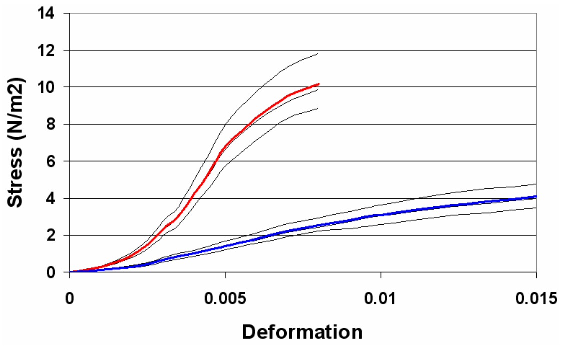

Figure 9. Averaged quasi-static and dynamic (resp. blue and red) stress functions used for MADYMO's head contact properties, based on the response to the occipital, parietal and frontal impacts (corridors in black) described in (Yoganandan *et al.* 1995).

Due to the choice of an averaged normalizing thickness, these characteristics do not represent the stress/penetration characteristics for a human head. However, their use in the previous equations (1) allow for a good fit of the resulting quasi-static and dynamic force-deflection curves obtained with this facet model.

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