# INJURY CRITERIA FOR ATHLETES SUBJECTED TO HEAD IMPACT

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## INTRODUCTION

Among the athletic actions leading to human body contact the ones involving head impact and spine impulsive loading are particularly serious, eventually leading to severe injuries, even disabilities. This is recognized in different sports, as American football or cycling, where the use of head protective equipment is mandatory. The prediction of injury measures and its relation to the injury mechanisms is decisive for a better design of the sports equipment as the body protective gear, pavillion and field pavements or furniture protections. The understanding of the injury mechanisms and their quantification can also be of utmost importance in setting the rules for new sports or for defining specific training practices.

Reliable computer models for the simulation of different activities involving human motion are fundamental tools to evaluate and improve safety (Wismans, 1994). Besides the biofidelity of these models, the correct treatment of the contact-impact conditions between the different segments of the biomechanical model and the surroundings must be taken into account. Many studies have been carried on in the area of body crash dynamics with three dimensional models (Bartz, 1971; Wismans, 1982; Bosio, 1986). The modeling of the contact impact forces has always been a fundamental part of these models. This problem was discussed by Lankarani (1995) who purposes a continuous force-impact model, based on Hertzian contact with energy dissipation, relating the geometric and material properties of the contact surfaces.

Due to tight safety requirements, the aerospace and automotive industries use measures of the human tolerance to impact and large accelerations (Steele, 1991). The criteria used to predict injury, *Head Injury Criteria* (HIC) or *Severity Index* (SI), are based on the integral of the head acceleration for a period of time (Lankarani, 1993). The potential for injury that they suggest are established as a result of extensive testing and can be used in impact cases other than car crashes. It is the objective of this paper to show that in sports activities these injury measures can be used. For this purpose, the total response biomechanical model is applied to an offside tackle of an athlete.

## METHODOLOGY

Based on a multibody methodology using natural coordinates (Jalon, 1994) a biomechanical model made of 12 rigid segments interconnected by 11 kinematic joints is used here, as shown in figure 1. The kinematic joints used in the model are spherical joints for the shoulders, neck-torso (between $12^{th}$ thoracic and $1^{st}$ lumbar vertebrae), hips and torso and revolute joints for the elbows, knees and head-neck (at occipital condyles). The hands and feet segments are not included in the model due to their reduced importance in crash prediction injuries. As a result of the 99 natural coordinates used to describe the rigid bodies and the 70 constraint equations the biomechanical model has 29 degrees of freedom. The masses, inertias and dimensions of the model are described in table 1.

Table 1 - Rigid body masses, inertias and centre of mass (Laananen, 1983).

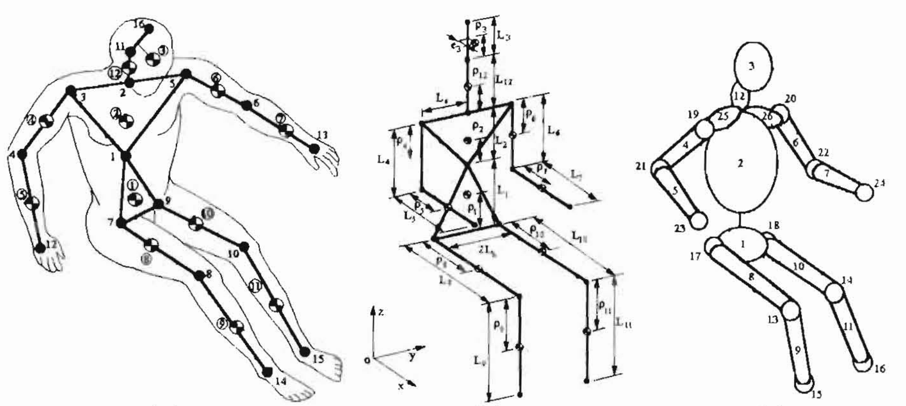
| Body | m [kg] / $I_{xx}$ / $I_{yy}$ / $I_{zz}$ [$10^{-4}$Kg m$^2$] | $\rho_i$[m] | $L_i$[m] |
|---|---|---|---|
| 1 | 14.20 / 22.41 / 5.56 / 12.92 | 0.064 | 0.240 |
| 2 | 24.95 / 22.82 / 9.63 / 16.37 | 0.193 | 0.333 |
| 3 | 4.24 / 0.09 / 0.57 / 0.13 | 0.141 | 0.216 |
| 4-6 | 1.99 / 0.05 / 0.35 / 0.01 | 0.153 | 0.295 |
| 5-7 | 1.84 / 0.08 / 0.74 / 0.14 | 0.180 | 0.376 |
| 8-10 | 9.84 / 0.87 / 4.13 / 3.60 | 0.215 | 0.434 |
| 9-11 | 4.81 / 4.38 / 3.42 / 1.73 | 0.230 | 0.467 |
| 12 | 1.06 / 0.00 / 0.06 / 0.00 | 0.049 | 0.130 |
| $L_s$ | - | 0.161 | - |
| $L_h$ | - | 0.094 | - |
| e | - | 0.051 | - |

A set of contact surfaces is defined for the calculation of the external forces exerted on the model by the surrounding objects, surfaces and other body segments. These surfaces are ellipsoids and cylinders as depicted by figure 1(c). The contact forces involving the impact of the body segments are modeled using an Hertzian contact theory which accounts for energy dissipation (Lankarani, 1995). The body segments are constrained from achieving unfeasible relative orientations by applying joint penalization moments between them any time that their relative angle exceeds prescribed values (Silva, 1996).

There are several head injury prediction criteria used for car crash events that have been established using a wide variety of tests. The Severity Index (SI) is a weighted impulse criterion for head impact evaluated as

$$SI = \int_0^T a(t)^{2.5} dt \qquad (1)$$

where a(t) is the acceleration in g's, T is the pulse duration and t is the time in seconds. The tolerance level of concussion for frontal impact is 1000 while for non-contact impact the threshold is 1500.



(a) (b) (c)

Figure 1 - Three-dimensional biomechanical model: (a) Human body; (b) Local coordinate frames; (c) Contact surfaces.

Another criterion that is generally used as a measure of the likelihood for head injury is the Head Injury Criterion (HIC) calculated as:

$$HIC = \left\{ (t_2 - t_1) \left[ \frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t)dt \right]^{2.5} \right\}_{max} \tag{2}$$

where a(t) is the resultant head acceleration in g's (measured at the head center of gravity), and $t_1$-$t_2$ is the time interval during which the HIC reaches its maximum value. Notice that in the initial and final time of the interval must be searched in order to obtain the maximum HIC.

## RESULTS

The biomechanical model is applied in to the simulation of a player experiencing an offset tackle by another player. The athlete suffering the tackle is a $50^{th}$ percentile human male that is standing while the incoming player with a total mass of 75 Kg is moving forward with a velocity of 3 m/s, as represented by the first frame of figure 2. The contact between body segments and surface are evaluated as the integration of the equations of motion of the system proceeds. The representation of the animation of the athlete motion is displayed in figure 2.

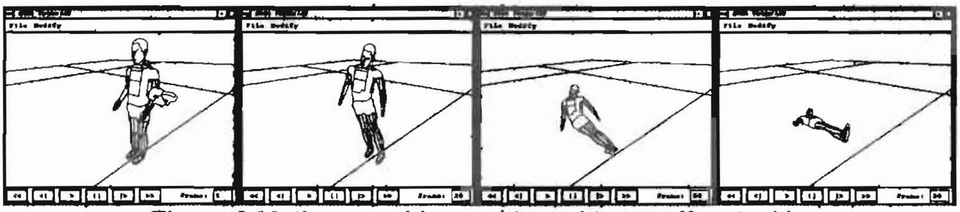


Figure 2 Motion an athlete subjected to an offset tackle

In the simulation of the impact the displacements, velocities, accelerations and forces acting upon the body segments are calculated. In particular, the forward acceleration of the athlete's head is represented in figure 3.

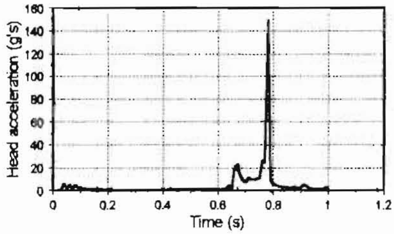



Figure 3 Head acceleration of the player subjected to impact

Based on the results of the simulation a SI value of 2170 and a HIC value of 873 for the injury criteria are calculated. These results suggest that the action simulated is within the head injury threshold and consequently head protective equipment is highly recommended to preserve the player's physical integrity.

## CONCLUSIONS

The biomechanical computer model of the athlete is applied in the simulation of an offset tackle and measures for the injury criteria is obtained. The interaction of the biomechanical model with the surrounding objects is effectively modeled and related with their material and geometric characteristics. The procedure shows that the injury criteria for head impact used in the aerospace and automotive industries is still suitable for application to sports activities. However, further investigation must be carried in terms of the quantification of the threshold of various levels for different injuries. The measures are also influenced by the neck and spine models. More precise models of the head-neck complex and of the spine must also be considered for a better measure of injury potential of athletes, specially if these are used to evaluate a wider range of actions.

## AKNOWLEDGEMENTS

This research is sponsored by Junta Nacional de Investigação Científica e Tecnológica (JNICT) with the project PRAXIS 3/3.1/CTM/10/94.

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