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# Rider–handlebar injury in two-wheel frontal collisions

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ABSTRACT

This work analyses blunt abdominal trauma produced by driver–handlebar collision, in low speed two-wheel accidents. A simplified dynamic model is introduced, whose parameters have been estimated on the basis of cadaver tests.

This model allows calculating the peak impact force and the abdominal penetration depth; therefore the likelihood of occurrence of serious injuries can be estimated for different masses of contacting bodies and different speeds. Results have been checked against literature data and true-accident reports.

Numerical simulations demonstrate that serious injuries (AIS > 3) can occur even at low speeds (< 20 km/h), therefore the design of protective clothing is recommendable.

The model can allow both the analysis of true accident data and the virtual testing of protective equipment in the conceptual design phase.



## 1. Introduction

Thoracic and abdominal injuries have been found to be common among motorcyclists accidents and to be associated with a higher risk of mortality (Kraus et al., 2002); in detail, the abdomen gains the third place among the most seriously injured anatomic regions, abdominal injury accounts for 20% of hospitalised motorcyclists injuries (Dischinger et al., 2006); lastly, head injury followed by chest and abdominal trauma were found to predict a reduced survival rate (Ankarath et al., 2002). Although many studies have documented the role of trunk injuries resulting from vehicular crashes (Morris et al., 2002, 2003), few have addressed these injuries among motorcyclists; in an analysis of motorcycle-related hospitalisations in the US, thoracic injuries are not even mentioned among the principal diagnoses as part of the 2001 Nationwide Inpatient Sample of the Healthcare Cost and Utilization Project (Dischinger et al., 2006).

More specifically, the incidence of abdominal injuries due to handlebar collision is likely to be underestimated due to insufficient documentation of the circumstances of injury events and a lack of applicable emergency codes specific for handlebar injury (Mezhir et al., 2007). On the other hand, data concerning impact with the handlebar in slow speed bicycling crashes demonstrate that this kind of injury can occur, even at low speed, it has a significant incidence, and it can lead to dramatic consequences: this phenomenon has been identified as a mechanism of life threatening injuries in children, and as the predominant cause of abdominal injury in children age 6–10 years (Bergqvist et al., 1985; Arkovitz et al., 1997).

In 1981, Hurt and his staff analysed 3600 motorcycle traffic accident reports in Los Angeles (Hurt et al., 1981); this study is now 30 years old, but it is still a reference for its completeness; one of its findings has been that the median pre-crash speed is 29.8 mph; analysing low-speed range is therefore a justifiable choice.

Considering secondary prevention issues for two-wheel drivers, the use of helmets has been widely proved to limit the incidence of head injury (Liu et al., 2008), while no true prevention has been really done in relation to chest and abdominal

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injury, apart from the recommendation of careful diagnosis and early treatment (Kraus et al., 2002). The above cited evidences suggest the importance of protecting these vital organs in addition to the head in motorcycle crashes and therefore the need for protective clothing in addition to helmets to reduce the mortality burden in this population. (ACEM, 2004; Haworth et al., 1997; Hurt et al., 1981). The design of protective systems can be aided by an understanding of injury mechanisms.

As said above, most abdominal injuries occur in automobile accidents (Stalnaker and Ulman, 1985); therefore injury criteria have been developed with reference to car accident with frontal or lateral impact. In frontal impact, driver–steering wheel collision can be assimilated to scooter driver–handlebar collision; the incidence of these events has been greatly reduced due to restraining systems and airbags; however, in the past, the study of this phenomenon has led to the description of abdominal injury mechanics, the execution of cadaver tests in order to identify injury predictors, and the development of simplified dynamical models which can be used in order to simulate real accidents and protective equipment efficacy.

From an anatomical point of view, the abdomen is even more vulnerable to injury than the chest because there is no bony structure like the ribcage to protect internal organs in front and lateral impacts. In detail, blunt impact of the upper abdomen is often reported where the liver and spleen have been injured: in the AIS$\geq$3 categories, the liver is the most frequently injured organ in frontal, right side and far side crashes; this is followed by spleen trauma (Yoganandan et al., 2001). In the liver, compression leads to the internal pressure rise; tensile or shear strains so generated eventually can lead to the laceration of the major hepatic vessels resulting in hemoperitoneum. Abdominal impacts can also cause the relative movement between lobes of the liver, stressing the vascular attachment at the hilar region (Viano and Andrzejak, 1993).

Numerical models can allow the virtual evaluation of accident mechanics and of the efficacy of protective equipment; very complex models have been developed for the prediction of trauma and human body response in auto crash (Forbes et al., 2005): these models can allow a predictive evaluation of injury under a variety of loading conditions, once the mechanical properties of simulated organs have been identified (Miller, 1991), and the models themselves have been evaluated on the basis of experimental data.

A simplified approach has been here employed since the objective of the numerical simulation was the only assessment of peak impact force, instead of stress and strain distributions. Besides, the variety of possible loading conditions has been strictly limited to the analysis of the frontal impact of a rigid body, at low speed, and the enquiry has been focused on the lower abdomen; the lumped-mass model of the chest developed by Lobdell et al. (1973) was therefore taken as a reference; it was simplified and characterised on the basis of experimental data obtained by Nusholtz et al. (1988, 1985) and the respective results provisions have been verified on the basis of literature data and real accidents data. This model can be numerically solved in few seconds, providing information about critical speeds in motorcycle–car frontal accidents, and giving an indication of the probability of injury occurrence. Model predictions have been verified on the basis of literature data and real accidents data.

## 2. Materials and methods

A single mass, one and a half-degree-of-freedom dynamic model has been developed in Simulink (by Matlab™); as illustrated in Fig. 1a, it was made of a spring, in series with a Voigt spring–dashpot system. Considering Nusholtz tests on unembalmed cadavers (Nusholtz et al., 1988), the input displacement is the impactor displacement; considering true accident data, the model input is the displacement of the vehicle and the scooter moving together ($x_{C,f}$) after the collision. The hypothesis behind this model is that the driver keeps moving horizontally at the same speed he had before the impact between the scooter and the car, while the scooter and its handlebar move together with the car at a speed which can be estimated considering an inelastic impact; this hypothesis holds as far as the driver does not hit any other body (the handlebar is considered to be the first striking object), and the force exchange between the driver and his seat can be overlooked (driver sliding with no friction or loosing contact with its seat); it is a conservative assumption since it leads to the most pessimistic estimate of driver speed. Besides, it should be emphasised that the hypothesis of forward trunk projection holds for low-speed accidents; otherwise more complex multi-body models should be used to forecast trunk trajectory (Serre and Llari, 2010). Momentum conservation equation is:

$$(M_D + M_S)v_{S,i} + M_C v_{C,i} = (M_S + M_C)v_{C,f} + M_D v_{S,i} \quad (1)$$

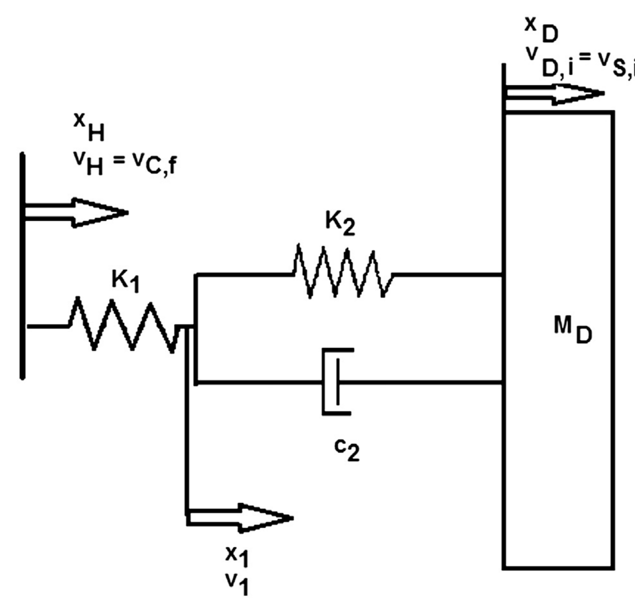


**Fig. 1 – Dynamic model employed in this work: a spring ($k_1$) is put in series with the parallel of a spring ($k_2$) and a damper ($c_2$); the left constrain represents the handlebar (*H*) with its displacement and its speed ($x_H$, $v_H$, respectively), the right mass represents the driver (*D*) with its displacement and its speed ($x_D$, $v_D$); the handlebar is forced to move at a constant speed, equal to the final car speed $v_{C,f}$; as a consequence, the driver mass *MD* oscillates around its equilibrium position, having an initial speed $v_{D,I}$ equal to the initial scooter speed $v_{S,i}$.**