OPTIMIZATION OF SINGLE SKIN SURFACES FOR HEAD INJURY PREVENTION – A COMPARISON OF OPTIMA CALCULATED FOR GLOBAL VERSUS LOCAL INJURY THRESHOLDS

Svein Kleiven[1], Rickard Juntikka[2]

[1]Division of Neuronic Engineering, CTV – Centre for Technology in Health Care, Royal Institute of Technology and Karolinska Institute, Stockholm, Sweden; [2]Division of Lightweight Structures, Dept. of Vehicle and Aeronautical Engineering, Royal Institute of Technology, Stockholm, Sweden.

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INTRODUCTION

THE HEAD INJURY CRITERION (HIC) has been used in crash testing for several years, as a measure of the likelihood of serious brain injury. HIC only treats the resultant translational acceleration and the duration of the impulse and no consideration is made for the direction of the impulse or rotational acceleration components (Bellora et al. 2001; Kleiven, 2003). The human head behaves in a more complex way during impact and since the validity of HIC is intensively debated there is reason to believe that the safety development could be made more efficient through use of more delicate tools, such as biomechanically representative FE models of the human head, together with local tissue strain thresholds. It is hypothesized that designs based on either of the two criteria would come out differently.

METHODS

A detailed and parameterized finite element (FE) model of the adult human head, including the scalp, skull, brain, meninges, cerebrospinal fluid (CSF), and neck with the extension of the spinal cord and the dura matter was used (Kleiven, 2003). This model has been experimentally validated in previous studies (*i.e.* Kleiven and Hardy, 2002). For comparison with the FE human head calculations, two head impact dummies were utilized; an FE-model of the Euro-NCAP pedestrian head dummy, modeled and verified according to the Euro-NCAP verification test; and an FE-model of the Hybrid III head developed by Fredriksson (1996), see Fig. 1. The FE head model had a mass of 4.64 kg, while the Euro-NCAP and Hybrid III dummies weighed 4.80 and 4.55 kg, respectively.

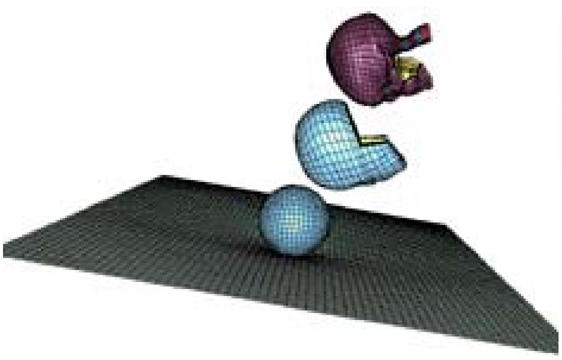

Fig. 1 - Impact simulations with (going upwards), Euro-NCAP dummy; Hybrid III dummy; FE-model of human head.

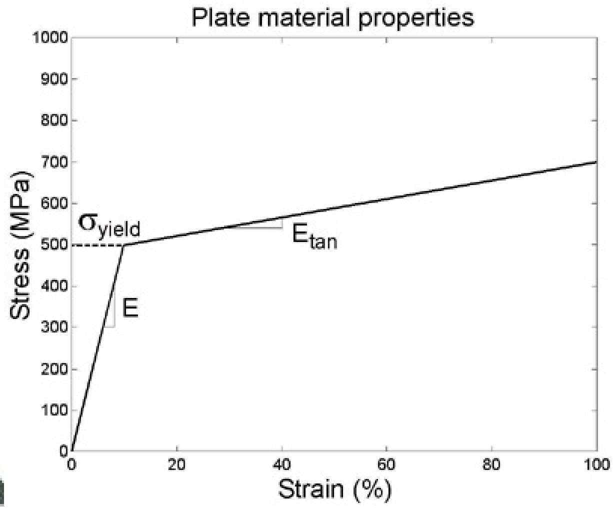

Fig. 2 - Material properties for the optimization of the plate.

The head models were impacted according to the Euro-NCAP pedestrian head impact test (40 km/h, horizontal angle 65°) towards a 900*900 mm quadratic plate with clamped boundary conditions at the edges with the plate rotated 10° to the horizontal to simulate the slope of an actual bonnet. The material properties of the plate were optimized for a minimum displacement of the plate in the vertical direction with constraints on HIC (HIC≤1000, Euro-NCAP, Hybrid III) and strains in the brain ($\varepsilon \leq 20$ %, Bain and Meaney, 2000). For the plate, a plastic-kinematic material model, see Fig. 2, was used and the optimization variables were density ρ, Young's modulus E, yield strength σ_{yield}, tangential modulus $E_{\tan}$ and thickness t. The calculations were performed using the FE-program LS-DYNA and the optimization program Xopt.

RESULTS

Substantially different results were obtained from the optimization with the rigid Euro-NCAP and the Hybrid III dummy heads when compared to the FE head model (Table 1).

Table 1 - Optimization results

	Euro-NCAP dummy		*Hybrid III dummy*		*FE human head*	
Property	*Start*	*Final*	*Start*	*Final*	*Start*	*Final*
Density ρ (kg/m^3)	5000	8000**	5000	7980	5000	8000**
Young's modulus E (Pa)	$1.00*10^{10}$	$9.48*10^{10}$	$1.00*10^{10}$	$7.20*10^{10}$	$1.00*10^{10}$	$1.43*10^{11}$
Yield strength σ_{yield} (Pa)	$1.00*10^{8}$	$5.00*10^{7}$*	$1.00*10^{8}$	$5.00*10^{7}$*	$1.00*10^{8}$	$5.00*10^{7}$*
Tangential modulus $E_{\tan}$ (Pa)	1000	0.00*	1000	6.80	1000	0.00*
Thickness t (mm)	0.60	0.76	0.60	0.67	0.60	0.40*
HIC	2000	997	2150	995	2870	944
Max strain in brain tissue (%)	-	-	-	-	49.4	38.8
Max vertical displacement (mm)	79.6	82.3	75.7	81.8	65.3	87.2
Max resultant load (kN)	9.92	5.92	9.48	5.57	7.51	3.90
Max resultant acceleration (g)	215	130	236	126	246	171
Plate max strain (%)	12.7	10.7	15.0	16.5	8.60	12.3

*Low limit of variable **High limit of variable

DISCUSSION / CONCLUSIONS

Due to the chosen limits of the optimization variables (bi-linear material properties ranging from values typical for polymers to high-strength steel), some variables reached their limit in all load cases. Consequently, the resulting optima are strongly dependent on the choice of the boundaries. However, the results show that even though HIC reached acceptable levels in all calculations, the strains in the FE human head model were still almost twice the recommended levels (Bain and Meaney, 2000). The strain in the brain varies however considerably with the stiffness of the brain tissue (Kleiven and Hardy, 2002). In addition, considerably higher initial HIC values were obtained for the FE head model compared with the rigid dummy heads. Large differences were also found in the resulting load and acceleration peaks throughout the optimizations, when comparing the rigid dummies with the FE head model. The results emphasize the importance of treating the human head as an elastic body. Hence, local tissue thresholds or more human-like dummies should be used to obtain more physically reliable optima in safety design. This result is conceptually obvious since a global criterion will never cover all the various injury mechanisms characterized by local tissue deformation.

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