NUMERICAL SIMULATION OF SHOULDER LATERAL IMPACTS FOR SHOULDER INJURY PREDICTION

Sonia DUPREY, Karine BRUYERE, Jean-Pierre VERRIEST
Laboratory of Biomechanics and Mechanics of Impact (UMR-T 9406 INRETS/UCBL), France

Key words: Human body; Models; Shoulder; Finite Element Method, Injuries

INTRODUCTION

IN CAR LATERAL impacts, the shoulder is directly exposed to injuries. These lesions are rarely fatal but produce long term impairments (Frampton, 1997). To develop protection devices, numerical tools such as finite element (FE) body models are used. If the first FE body models represented average males, geometrical personalization methods are now available (Besnault, 1998; Buhmann 2003) and allow taking into account the large inter-individual variability. Actually, an average model can not be used for predicting injuries for a subject whose anthropometry differs significantly from that of a 50th percentile model. It is not common to find validated shoulder FE models (Iwamoto, 2000; Thollon, 2002), and these last models represent 50th percentile male and did not allow the prediction of fractures. Thus, the purpose of this study is to assess an upgraded version of the shoulder part of the 50th percentile male HUMOS model and to assess the ability of geometrical personalization methods to improve shoulder injury risk prediction.

METHODOLOGIES

UDGRADED 50TH PERCENTILE MALE SHOULDER MODEL

THIS WORK IS BASED on the HUMOS1 model (Behr, 2003) whose shoulder has been upgraded. The main shoulder ligaments, e.g. sterno-clavicular, coraco-clavicular, acromio-clavicular, and gleno-humeral ligaments are modelled by means of membrane elements. The shoulder bones are modelled with shell elements representing their cortical part and with solid elements representing their spongious part. The clavicle model is composed of 90 shells of 2.6mm thickness and 42 solid elements; the humerus is made of 357 shells of 3mm thickness, solid elements are only present in its extremities; the scapula part is represented by its cortical part meshed with 599 shell elements of 5mm thickness. Material properties coming from the literature were integrated for shoulder ligaments and shoulder bones. Their mechanical behaviours are described by elastic-plastic laws; Table 1 lists their different parameters and their sources.

Shoulder part		Main parameters of the elastic-plastic law						Reference
		ρ (kg/l)	E (MPa)	ν	σ_y (MPa)	σ_{max} (MPa)	ε_{max} (%)	
Clavicle	cortical	1.8	11 000	0.3	80	80	3.0	Iwamoto, 2000
	spongious	1	500	0.1	4	4	1.5	HUMOS project
Scapula	cortical	2	16 000	0.3	80	80	3	Couteau, 2001
Humerus	cortical (tension)	2	18 000	0.3	90	90	1.5	Yamada, 1970 HUMOS project
	cortical (compression)	2	6 000	0.3	135	135	4	Yamada, 1970
	spongious	1.5	450	0.1	10	15	1.5	HUMOS project
Ligaments	AC	1	10.4	0.3	6.2	6.2	94	Koh, 2004
	CC	1	9.6	0.3	3.9	3.9	72	
	SC	1	11.7	0.3	3.4	12	55	
	GH	1	150	0.3	8	8.5	11.2	McMahon, 1998

Table 1 – Material properties of the upgraded shoulder model

ASSESSMENT METHOD OF THE 50TH PERCENTILE MALE SHOULDER MODEL

FIRST, MAIN SHOULDER bone models, clavicle and humerus, were assessed by comparing their responses under dynamic loading with results from experimental tests (Duprey, 2006, 2007). Then, a

general assessment of the upgraded shoulder model was realised. Simulations of the non-injurious tests carried out by Compigne et al. (2004) were performed. The impact force responses of the model were assessed by comparison with the non-injurious experimental data: at 1.5ms^{-1} (with three different directions: 0°, +15° and -15°) and at 3.5ms^{-1} (0°). The "ModEval" method (Jacob, 2000) was used in order to know the extent of the model validation. The experimental tests used for model assessments are listed in Table 2.

Tests	Number	Velocity	Impact	Reference
Clavicle	7	1.5 ms^{-1}	Axial compression	Duprey, 2006
Humerus	7	1.5 ms^{-1}	3-point Flexion	Duprey, 2007
Shoulder	7	1.5 ms^{-1}	0° impact	Compigne, 2004
	7	1.5 ms^{-1}	-15° impact	
	7	1.5 ms^{-1}	+15° impact	
	5	3.5 ms^{-1}	0° impact	

Table 2 – Experimental tests used for model assessments

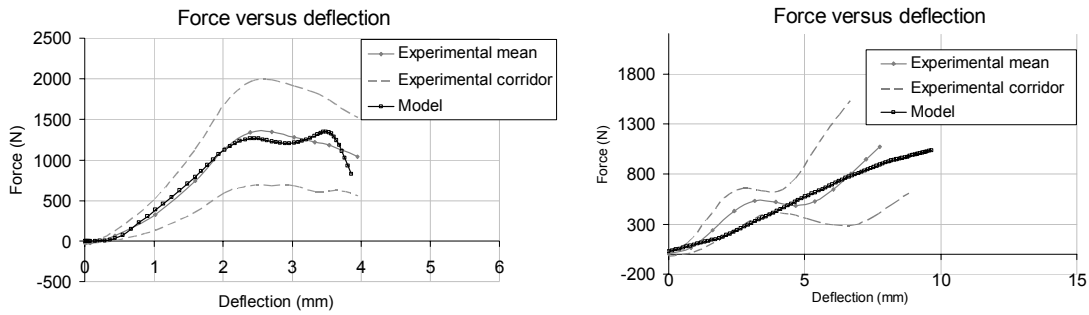
GEOMETRICAL PERSONALIZATION OF THE SHOULDER MODEL:

THE UPGRADED MODEL was geometrically personalized towards the anthropometries of the subjects tested in Compigne's study (2004) by means of the ScalingTool®. This software was developed in the HUMOS2 project and is based on the krieging method (Trochu, 1993), which is able to personalize the shape of bones or organs but is unable to modify the shell thickness. The fact that bone thickness can not be altered through such methods is a major limitation as cortical bone thinning is a frequent osteoporosis effect and a usual consequence of aging. Thus, the thickness of the clavicle model was adjusted according to the age of the tested subjects. Then, the simulated injurious consequences obtained with the geometrically personalized models were assessed by comparison with the experimental lesions. Fractures or ruptures of parts of the FE model occur when, in one element, both the maximum stress and the maximum strain are reached. If the element is a shell, then it is deleted; if it is a volume, its stress and strain tensors are definitely set to zero.

RESULTS

VALIDATION OF THE UDGRADED 50TH PERCENTILE MALE SHOULDER MODEL

Force-deflection curves obtained by simulation of tests on isolated clavicle and humerus are illustrated in Fig.2, as well as mean experimental curves and experimental corridor. The clavicle and the humerus models have simulated force-deflection curves close to the experimental mean curves. The force amplitude difference between the experimental mean curve and the model curves were calculated at each 0.5mm of deflection, from 0mm until 3.5mm for the clavicle model, and from 0mm until 7.5mm for the humerus model. The average differences are, for the clavicle and the humerus models, 4.6% and 8.6%, respectively. Furthermore, the simulated curves are inside the experimental corridors. Moreover, it appeared that the occurrence of fractures as a function of the impact velocity is in agreement with the experimental injurious thresholds which were 1.41ms^{-1} and 1.77ms^{-1} for the clavicle and the humerus, respectively.



a) Clavicle model b) Humerus model

Fig. 2 - Simulated force-deflection curves compared to experimental mean curves and corridors.

a) Clavicle compression tests at 1.5ms^{-1} - b) Humerus bending tests at 1.5ms^{-1}.

The simulated response of the upgraded 50th percentile male shoulder model under lateral impacts is compared to experimental corridors on Fig. 3. According to the "ModEval" method, scores were calculated for the impact force curves. The mean score obtained was between 80% and 90%, which is considered as an acceptable result on the satisfaction scale defined in the method. At this point, the model did not appear to be efficient for injury prediction as it did not reproduce most of the experimental injuries when tested at higher velocities (see Table 3).

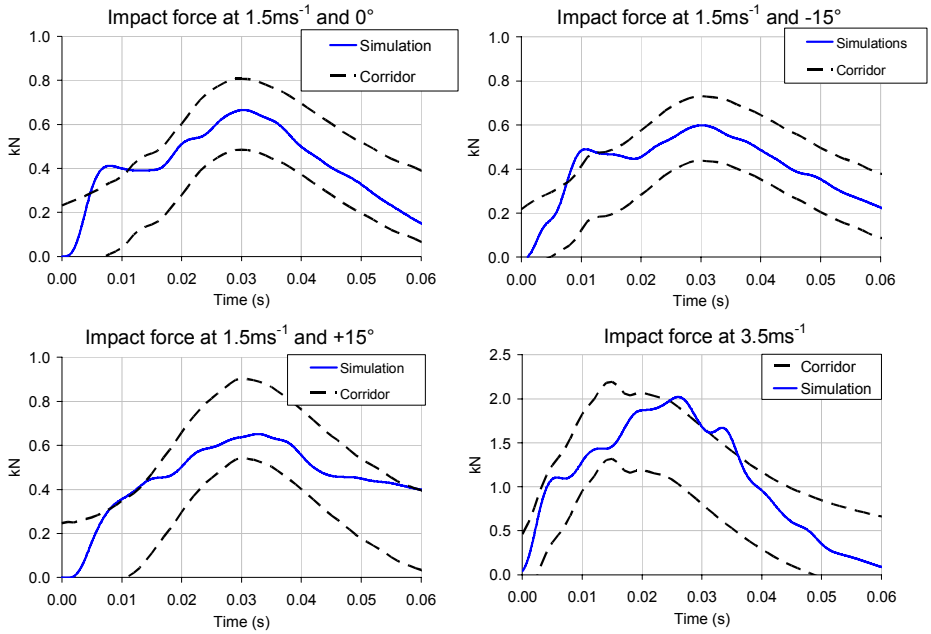


Fig. 3 – Simulated impact force versus time curves of the upgraded shoulder model compared to experimental corridors at 1.5ms^{-1} (0°, -15°, +15°) and at 3.5ms^{-1} (0°).

ASSESSMENT OF PERSONNALIZED SHOULDER MODELS

TABLE 3 LISTS THE INJURIOUS consequences obtained for the highest speed shoulder impacts. The simulated fractures obtained with the personalized models do not exactly match the experimental ones: the upgraded personalized model does not reproduce scapula fracture and is conservative, e.g. it predicts fractures even in some experimental cases in which no fracture occurred (see PMHS7). The geometrical personalization appears efficient as the injurious consequences are more realistic with the personalized upgraded model than with the base model. Furthermore, a sensitivity study performed on the clavicle thickness showed that this personalization parameter could largely affect the injurious consequences produced during a shoulder lateral impact.

		Fractures		
Subjects	Impact speed	Experimental	Upgraded model	Upgraded and personalized model
PMHS1	5.87 ms^{-1}	Clavicle fracture	Clavicle fracture	Clavicle fracture
PMHS3	4.24 ms^{-1}	Clavicle and scapula fracture	None	Clavicle fracture
PMHS4	4.27 ms^{-1}	Scapula fracture	None	Clavicle fracture
PMHS5	2.94 ms^{-1}	None	None	None
PMHS6	2.95 ms^{-1}	None	None	None
PMHS7	4.09 ms^{-1}	None	None	Clavicle fracture

Table 3 - Injurious consequences for the upgraded models with and without geometrical personalization, compared to experimental shoulder injuries

DISCUSSION & CONCLUSION

CONCERNING THE ISOLATED shoulder bone models, i.e. clavicle and humerus models, their responses under dynamic loading were close to the experimental ones. The upgraded shoulder model, when submitted to lateral impacts at different speeds and with different directions, produced high

validation scores allowing us to consider the quality of its validation as acceptable. However, at higher velocity, the 50th upgraded shoulder model did not produce realistic injurious consequences. The geometrical personalization process appears to be efficient as upgraded personalized model responses were closer to experimental results than upgraded model responses. The clavicle thickness appeared as one of the geometrical parameters that has the most importance in the geometrical personalization. Actually, bone thickness is linked with age and adjusting the geometrical parameters of cortical bone, allows for the effects of aging to be taken into account. Thus, if the upgraded model representing an average model is validated against non-injurious tests, only models seem able to produce realistic injuries when submitted to shoulder lateral impacts.

THIS FIRT STEP towards shoulder injury prediction by means of a FE model is promising since the upgraded personalized shoulder model can simulate realistic shoulder injuries. Furthermore, this study puts forward the necessity of geometrical personalization.

REFERENCES

- Behr M. et al., A Human model for road safety: from geometrical acquisition to model validation with Radioss. J of Comp Meth in Biomech and Biomedical Eng, 2003, vol. 6, pp.263–273.
- Besnault B. et al., Morphometric study of the human pelvis. Proc of the 16th ESV conf, 1998, Windsor, Canada, NHTSA Editor, pp.2065-2072.
- Buhmann M. D. Radial basis functions: theory and implementations. Applied mathematical sciences, Cambridge University Press, 2003, 259p. Cambridge Monographs on Applied and Computational Mathematics N°. 12.
- Couteau B. et al., Finite element analysis of the mechanical behaviour of a scapula implanted with a glenoid prosthesis. J of Clin Biomech, 2001, vol.16, pp.566-575.
- Compigne S. et al., Non-injurious and injurious impact response of the Human Shoulder – Three dimensional analysis of kinematics and determination of injury threshold. STAPP Car Crash Journal, 2004, vol.48, pp.89-125.
- Duprey S. et al., Experimental and simulated flexion tests of humerus. 2007, IJCrash, in Press.
- Duprey S, Development of a shoulder finite element model and simulation of its response under lateral loading, PhD report, Feb 2007, Insa de Lyon, France.
- Duprey S. et al., Evaluation of clavicle geometry influence during dynamic compression. 2006, Proc of the 5th World Cong of Biomech, J of Biomech, vol.39, suppl 1.
- Frampton R.J., Morris A.P., Thomas P., et al. An overview of upper extremity injuries to car occupants in UK vehicles crashes. Proc of the IRCOBI conf, 1997, Hannover, Germany, pp. 37-52.
- Iwamoto M. et al., Development of a finite element model of the human shoulder. STAPP Car Crash Journal, 2000, vol.40, SAE paper n°200001SC19, pp.281-297.
- Koh S.W. et al., Mechanical properties of the shoulder ligaments under dynamic loading. STAPP Car Crash Journal, 2004, vol.48, pp.125-153.
- McMahon P.J. et al., The anterior band of the inferior glenohumeral ligament: Biomechanical properties from tensile testing in the position of apprehension. J of Should and Elb Surg, 1998, vol.7, pp.467-471.
- Serre T., Brunet C., Bruyere K., et al., HUMOS (Human Model for Safety) Geometry: From one specimen to the 5th and 95th percentile, SAE 2006 Transactions Journal of Passenger Cars Mechanical Systems, Paper n° 2006-01-2324, 2006.
- Thollon L. et al., Finite element modelling and simulation of upper limb with radioss, IJCrash, vol.7 no.3, 2002.
- Trochu F., A contouring program based on dual kriging interpolation, Engineering and computing, 1993, Vol. 9, pp.160-177.
- Yamada H. Strength of biological Materials. Baltimore, MD, USA: edited by Evans, Williams and Wilkins, 1970.