DEVELOPMENT OF AN INJURY RISK FACTOR FOR DRIVERS

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ABSTRACT

The main objective of this research is to present a detailed method to numerically estimate the dynamic stresses and the risk of adverse health effects to which professional drivers are particularly prone. A parametric finite element model of the lumbar vertebrae has been generated to compute the dynamic stresses and the risk of fracture under harmonic excitations. An injury risk factor (IRF) has been developed. A design of experiments has been conducted by using the IRF as the dependent variable and by considering the posture, the body type, the apparent bone density, the damping rate, the body weight and the acceleration level as independent variables. The design of experiment is based on 972 numerical simulations. In this study, the effect of ageing is considered as representing a decrease of the apparent density of vertebral cancellous bones and of the damping rate of intervertebral disks. It was shown that the IRF increases with ageing and an IRF of 30% has been found a threshold for mechanical fatigue purpose. The results show that the drivers with a low density (0.1 g/cm^3), a low damping rate (10 %) and a heavy weight (98 kg) present a high probability of injury. These characteristics can be met among old drivers. An acceleration amplitude of 2 m/s^2 has been found has a threshold if we want to avoid any risk of injury whatever the driver, his weight, bone structure and age.

RÉSUMÉ

L'objectif de cette étude est de présenter une méthode numérique pour estimer les contraintes dynamiques et leurs effets sur la santé des conducteurs professionnels. Un modèle paramétrique de la colonne vertébrale a été généré afin de calculer les contraintes et le risque de fracture sous l'effet d'une excitation harmonique. Un facteur de risque au dommage (FRD) en a résulté. Fondé sur les résultats numériques, un plan d'expérience a été mené avec le FRD comme variable dépendante. La posture, la corpulence, la densité apparente de l'os spongieux, l'aire des vertèbres, le taux d'amortissement du disque intervertébral, le poids des individus ainsi que l'amplitude de l'accélération du siège à la résonance des vertèbres ont été considérés comme variables indépendantes. Le plan d'expériences a compris 972 simulations. L'effet de l'âge a été considéré comme représentant une diminution de la densité osseuse et du taux d'amortissement des disques intervertébraux. Une limite de 30% du FRD a été considérée comme critique afin de considérer l'effet de fatigue à long terme. L'effet de la densité osseuse, de l'amortissement et du poids a été mis en évidence. Les personnes ayant une faible densité (0.1 g/cm^3), un faible taux d'amortissement (10%) et un poids lourd (98 kg) présentent une forte probabilité de dommage. On retrouve ces caractéristiques parmi les personnes âgées. Une amplitude d'accélération de 2 m/s^2 a été trouvée comme une limite à ne pas dépasser pour éviter tout risque de dommage.

KEYWORDS Adverse health effect, Whole body vibration, Spinal loading, Ageing, Posture, Design of experiments.

1. INTRODUCTION

Long term whole body vibrations (WBV) may damage the lumbar spine, especially the three lower vertebrae (L3-L5) [1, 2]. Several bibliographical reviews published over the past 15 years show a higher occurrence of low back disorders among populations exposed to dynamic loading, such as heavy equipment drivers, than in the general population [3]. The vibratory response of structural components of the lumbar spine and consequently the risk of adverse health effects vary considerably based upon numerous factors [4]. Among the factors influencing the vibratory response are the posture [5], the amount and distribution of the bone mineral content [6, 7], the size of the vertebral bodies and discs [8], the degree of the disc degeneration [9] and the body weight. However, few numerical studies establish a relation between the dynamic stresses and the acceleration at the seat on which drivers are seated [10-11]. Most of these models are not designed to easily observe the effect of inter-individual variations (posture, bone structure, and body weight). On the other hand, the experimental study developed by Seidel et al. (1998) [12], allowed for the assessment of a relationship between the forces and the acceleration at the seat by considering posture and the bone structure. Thomas et al. (2004) [13] have studied the long term adverse health effect for drivers exposed to harmonic and random vibrations. However, these models of the lumbar spine were based on an analytical method. A more sophisticated numerical model is required to refine the results.

The paper is aimed at the elucidation of exposure response relationships in order to derive quantitative relations for the assessment of the injury risk of fracture due to WBV. A design of experiments was conducted in order to study the effects of the following parameters: driving posture, body weight, acceleration, bone structure (robust, intermediary or frail), apparent density and damping rate. In this study, ageing is represented by a decrease of the apparent density of the vertebral cancellous bones and damping rate of intervertebral disks.

2. MATERIALS AND METHODS

2.1. Dynamic stresses

The dynamic behaviour of the vertebral bone may depend on several variables. In this study, we considered the principal following parameters: posture (θ), body weight (M), acceleration level (A(f)) at the natural frequency f, bone structure (S), damping rate (ξ) and apparent density (ρ) of lumbar vertebrae. If all these variables are assumed to be independent and quantifiable, the dynamic stress σ_{dyn} may be expressed as:

$$\sigma_{dyn}=F(A(f),M, S, \xi, \theta, f) \quad (1)$$

By modeling the lumbar spine as a one degree of freedom (DOF), the compressive dynamic stress may be calculated from the transmitted force [13]. By applying a random signal that excites the natural frequency and by considering the bone structure and the posture angle, the compressive dynamic stress can be expressed in the following form:

$$\sigma_{dyn} = B\frac{\sqrt{1+(2\xi)^2}}{2\xi}\frac{A(f)*M*}{S}Cos(\theta) \quad (2)$$

where:

- M is the equivalent mass,
- A(f) the applied acceleration amplitude to the seat at the natural frequency of vertebrae (f),
- S the average cross-section of disc,
- θ the posture angle and
- ξ the damping rate.
- B is a statistical constant.

2.2. Injury risk factor IRF

In this study, a nonlinear model for predicting the adverse health risk has been developed. This model is based on the assumption that the vibratory response of lumbar spine may be represented as a one degree of freedom model (DOF) as suggested by Coermann (1962) [14] and Griffin (1990) [15]. These simplified models were derived in order to satisfy mechanical impedance (or apparent mass) data. A Dynamic Response Index (DRI) model was initially defined in order to express the relative severity of vibrations, more particularly to shocks [15].

By assuming that the risk of adverse health is relative to the ratio of the applied stress to the ultimate stress, a new injury risk factor (IRF) has been developed:

$$IRF = 100*\frac{(\sigma_{dyn}+\sigma_{stat})}{\sigma_u}. \quad (3)$$

σ_{stat} is the compressive static stress as computed by our numerical model and σ_u ,estimated from literature as 41.668 $\rho^{1.9}$ [16-19], is the ultimate stress of the vertebrae. The ultimate stress of the cancellous bone has been considered because it represents 90% of the vertebrae volume.

By introducing equation (2) into (3), the following relationship has been obtained:

$$IRF(\%) = (B1 + B2*\frac{\sqrt{1+(2\xi)^2}}{2\xi}A*M\cos(\theta))\frac{1}{S*\rho^{1.9}} \quad (4)$$

where B1and B2 are constants to be extracted from the design of experiments.

2.3 Finite element model

A parametric model of the vertebrae (L1/L5) was developed in order to study the effects of posture (kyphosis, lordosis and neutral posture), bone structure (intermediate, frail and robust), body weight, degree of degeneration represented by the damping coefficient, acceleration and apparent density of cancellous bone on the injury risk factor. The parametric finite element model of the lumbar spine (L1-L5) was generated in a CAD software application by considering the parametric equations describing the shape of a vertebra and an intervertebral disc as established by Lavaste et al. (1992) [20] and the morphometric dimensions measured on various vertebral bodies by Berry et al. (1987) [21]. Figure 1 illustrates the main parameters of lumbar vertebrae.

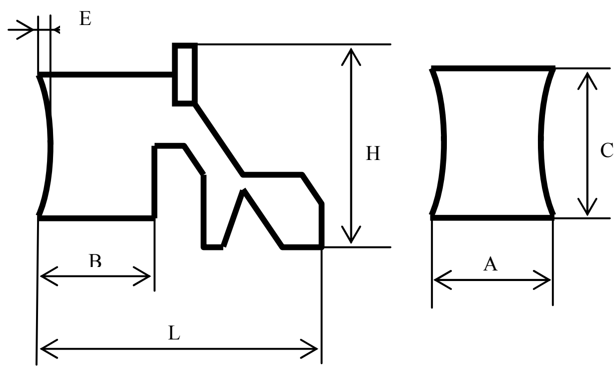

Figure 1: main parameters of lumbar vertebrae [20]

The main advantage for using a parametric model is the facility with which different human morphologies can be studied. This parametric dynamic model presents the advantage of simulating any condition of seated posture as well as the different geometrical variations. Three morphologies (frail, intermediate and robust body) have been studied by varying the cross section of the disk and vertebral body and keeping the same scale. This parametric model allows to study the effect of the sitting posture (kyphosis posture $\theta \leq 5°$, lordosis posture $\theta \geq 25°$ and average posture $\theta \approx 15°$), the bone structure and the body mass on the mechanical behaviour of spine exposed to whole body vibration. Figure 2 illustrates curves of the lumbar spine.

Each model of the vertebrae is composed of 33 structures (annulus, nucleus, endplate, cortical shell and cancellous bone) and 54 contact zones were inserted between the bodies. The contacts were modelized with contact elements (Ansys Target 170 with 8-node, and Conta 174 with 8-node). The volumes in each model were meshed separately with their meshing parameters. Accordingly with the geometrical complexity of the spine, the finite element mesh has to be fairly fine. The cortical shell, the posterior elements, the cancellous bone and the endplates were meshed by using 3D tetrahedral elements with 10 nodes (Ansys software: Solid 187). This type of element was selected because it allows a good interpolation of external geometry. The non homogeneous structure of the intervertebral disc was taken into account. The annulus fibrosus was modelized as a composite material. The nucleus pulposus was modelized by using volumic elements with a Poisson coefficient of 0.499.

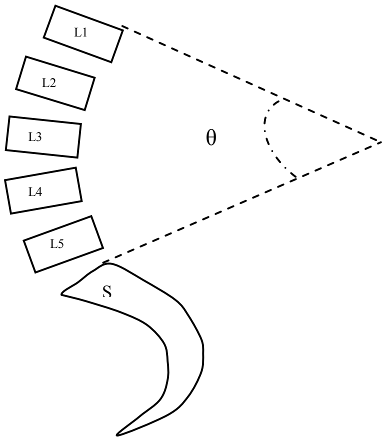

Fig.2: Curves of the lumbar spine [22]

In vivo, relative motion between posterior elements is assured by articular cartilages. A very low coefficient of friction has been applied to model the relative motion of cartilaginous structures. The contact element used for modelling the connection between the posterior elements has been chosen as 'frictionless' type. Springs of low rigidity were added to the contact element model in order to insure continuity. The total number of elements for the whole model is about 36500 and the number of nodes is 83808. The meshing of the total vertebrae and the contacts of "frictionless" type between apophyses are shown in Figure 3.

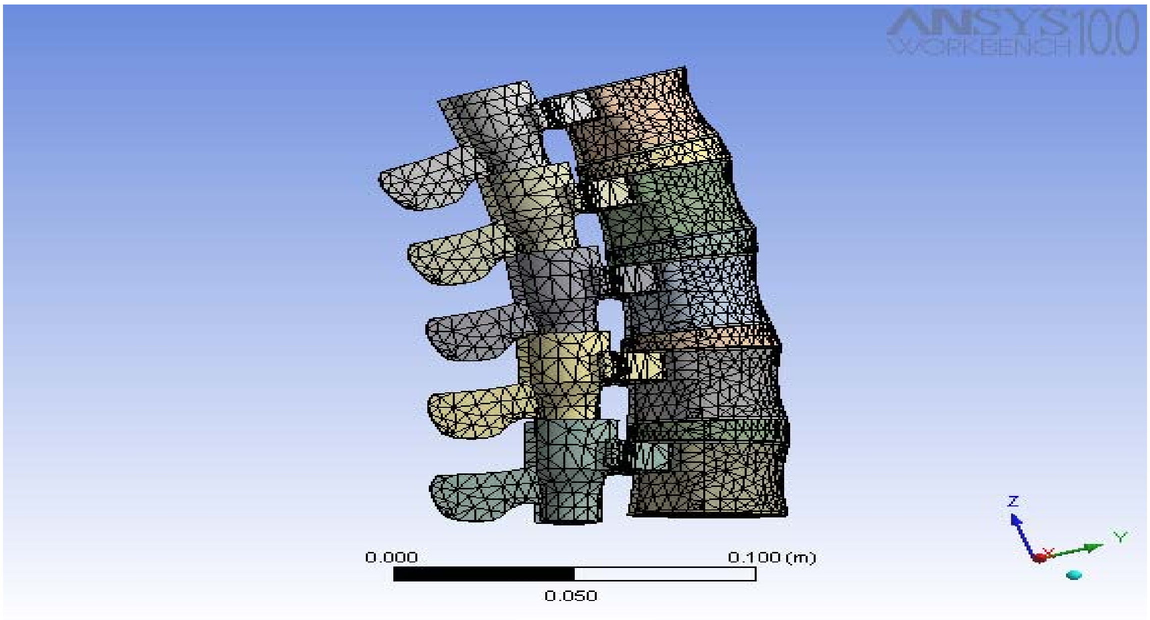

Figure 3: FE models of the Lumbar spine

Once the vertebrae model has been generated, the dynamic analyses may be carried out on the model through the finite element method. The mechanical properties of the various elements (cortical shell, cancellous bone, posterior elements and cartilaginous endplates, intervertebral disc) forming the vertebral body were deduced from literature [20, 23, 24, 25]. The material properties are given in Table 1.

Table 1: Material properties

Material	*Element type*	*Elastic Modulus (Mpa)*	*Poisson's Ratio*
Cortical bone	volumic	12000	0.3
Posterior elements	volumic	1000	0.25
Cancellous bones	volumic	100	0.2
Cartilaginous end plates	volumic	24	0.4
Annulus fibres	volumic	500	0.3
Annulus matrix	volumic	4.2	0.45
Nucleus	volumic	1.3	0.499

For the dynamic analysis of the lumbar spine, a distributed mass of about 57% the body weight was applied to the upper face of the vertebrae [13, 24, 25] to simulate the upper body. The base was subjected to various vertical accelerations ranging from 1 m/s^2 to 4 m/s^2. These acceleration amplitudes were extracted from the ISO 2631-1 [26] curves that define the limits of exposure in vertical acceleration according to the frequency and duration of exposure. The viscous damping rate depends on the level of degeneration of the intervertebral disc (grade of the disc) and muscle activity [23]. The stress responses were evaluated by considering a viscous damping rate of 10%, 20% and 30% as determined from experimentation by Kasra et al. (1992) [24, 25].

2.4. Limits of Injury Risk Factor (IRF)

Based on experimental data, Seidel et al [12] estimated that a stress ratio of 20% normalised with the ultimate stress could be regarded as an endurance limit to avoid any risk of damage by fatigue. However, Brinckmann et al. (1988) [27] argued that a stress ratio of 30% normalised with the ultimate stress could be regarded as an endurance limit for in vivo exposure. Consequently, the threshold of 30% has been assumed as a limit for avoiding any risk of fatigue after long duration of exposure to dynamic excitations as it is usual in mechanical fatigue problems. This IRF level of 30% could be considered as representing a moderate probability of injury.

Therefore, we have considered in this study the following criteria for IRF:

- If IRF< 30%, a driver will have a low probability of injury;
- If 30% < IRF < 50% , a driver will have a moderate probability of injury;

- If IRF > 50%, a driver will have a high probability of injury.

2.5. Numerical simulations protocol

In order to study the effects of parameters appearing in equation (4) on the injury risk factor IRF, a numerical design of experiments was conducted. A full factorial design was selected to allow all interactions between the independent variables to be effectively investigated.

Nine geometrical models of the vertebrae were generated (3*3) for the design of experiments: three postures (lordosis, flexed and intermediate) and three bone structures (frail, intermediate and robust body), by varying the cross section of the disk and vertebral body and keeping the same scale. The levels for each factor were extracted from literature:

- The acceleration amplitudes were extracted from the ISO 2631-1 curves [26]. In this study, four levels were considered. Levels of the applied accelerations were 1, 2, 3.15 and 4 m/s^2.
- Since the density is depending from the ageing, three levels of the vertebral apparent density, ranging from $0.1 g/cm^3$ to a maximum value of $0.3 g/cm^3$ [16-19], with an intermediate value of $0.2 g/cm^3$ were considered.
- Three levels of body weight were chosen (heavy: 98 kg), intermediate: 75 Kg and light: 55 kg).
- In sitting posture, three critical levels of posture angle were chosen as determined by Dolan and Adams (2001) [22]: the kyphosis posture, the lordosis posture and the intermediate posture.
- Three levels of the cross sectional areas (S) were considered. The cross sectional areas (S) at the L3-L4 represent the bone structure (frail, intermediate or robust body). The same parameter (S) was used by Seidel et al, (1998) [12].
- The damping rate depends on the degree of degeneration of the intervertebral disc and consequently from the ageing. Three limit levels of the damping rate were considered [23, 24, 25]. Levels of the damping rate were 10%, 20% ant 30%.

Table 2 establishes the values taken for each factor. The total number of numerical simulations is 972.

Table 2: Independent variables

Variables	Levels			
	1	**2**	**3**	**4**
Acceleration A (m/s^2)	1	2	3.15	4
Apparent density ρ (g/cm^3)	0.1	0.2	0.3	
Body weight M (Kg)	55	75	98	
Posture angle θ°	5	15	25	
Cross sectional area L3-4, S (mm^2)	1200	1500	1800	
Damping rate ξ (%)	10	20	30	

3. RESULTS

3.1. ANOVA analysis

Analysis of variance (ANOVA) was applied to investigate the main effects of the independent variables, together with their two-level interaction effects on IRF. The software Statgraphics was used to make the statistical analysis and to carry out the variance analysis (method ANOVA). For the Anova analysis, we selected the p-value at 0.01 level that indicates a 99% confidence level and the calculated Fisher-ratio for testing the significance of the main effects and two-level interaction effects. The computer ANOVA output and the calculated F ratios with their p-value are shown in Table 3 for each significant effect. The ANOVA table decomposes the variability of IRF into contributions due to various factors (Table3).

Table 3: Anova analysis

Source	F-Ratio	P-Value
Main effects		
apparent density : ρ	11982,02	0,0000
Body weight :M	1207,69	0,0000
Damping rate : ξ	822,13	0,0000
Acceleration : A	513,18	0,0000
Cross sectional area S	435,92	0,0000
Posture angle : θ	46,25	0,0000
Interactions		
M* ρ	409,86	0,0000
ξ *ρ	270,18	0,0000
A *ρ	173,98	0,0000
S* ρ	148,11	0,0000
M* ξ	64,60	0,0000
A* ξ	61,33	0,0000
A*M	22,86	0,0000
M*S	21,95	0,0000
Θ* ρ	15,69	0,0000

The contribution of each factor is measured having removed the effects of all other factors. Since 12 P-values are less than 0.01, these factors have a statistically significant effect on IRF at the 99.0% confidence level.

Only the significant effects are shown and they are classified from the most significant to the lesser one. In this study, we have chosen a Fisher ratio of 15 as a threshold and we will describe the 9 most important interactions on IRF. The analysis showed that the osseous density has an effect definitely higher than all the other factors on the injury risk of fracture.

3.2 Interaction analysis on damage injury risk factor IRF

The effects of interactions implying the density and body weight are definitely higher than the others on the injury risk factor. The risk of adverse health increases with body weight and when density decreases. This effect is critical for the adverse health when density is very low, whatever the mass. Figure 4 shows the interactions between these two variables. This probability of injury becomes moderate for drivers with a high weight and a density of 0.2 g/cm^3.

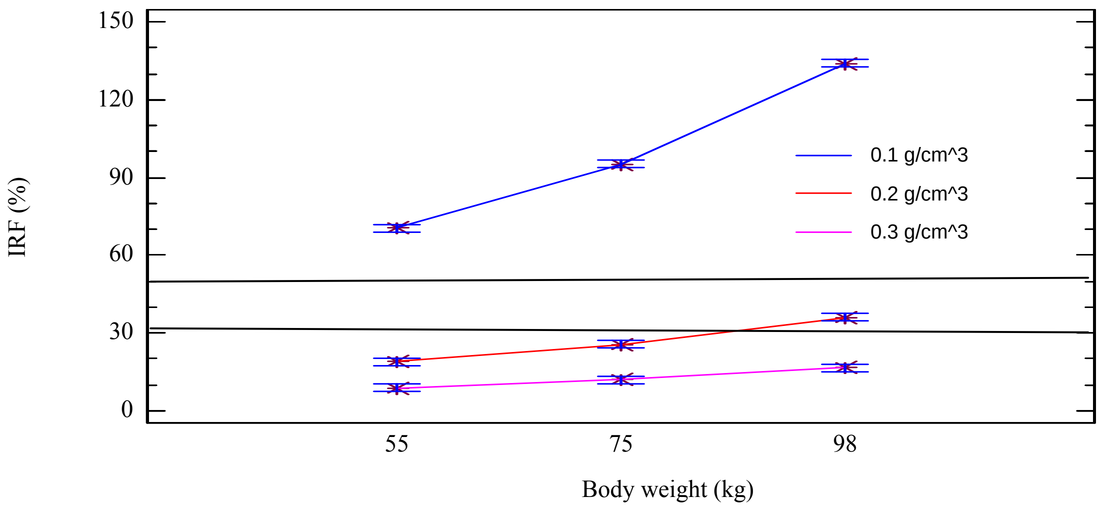

Figure 4: Interaction of mass and density on IRF

The damping rate and density is the second significant interaction in importance (Figure 5). It is shown that the risk of adverse health effect decreases with the increase of density and damping rate.

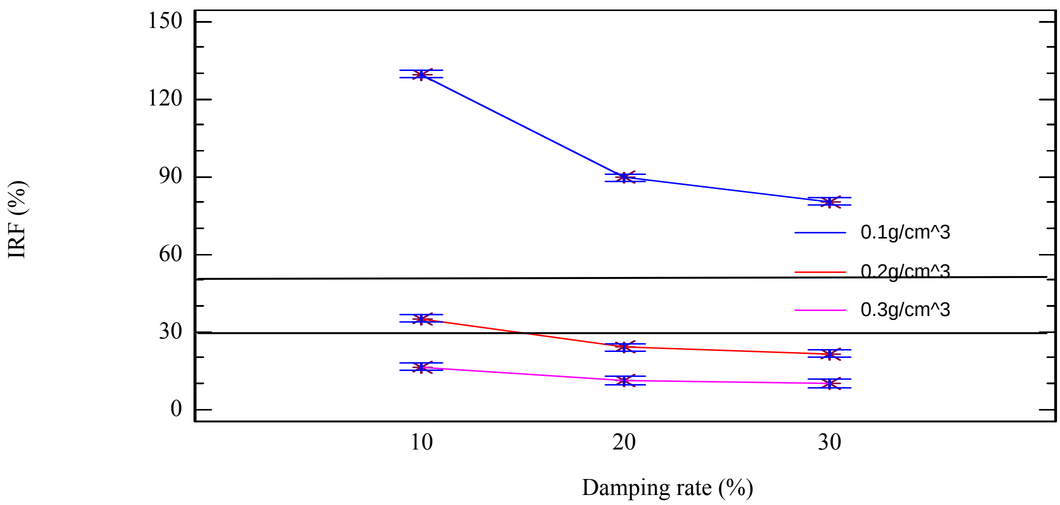

Figure 5 Interaction of density and damping on IRF

If the osseous density is low 0.1 g/cm^3, we observed a high probability of fracture whatever the damping. If it is high (0.3 g/cm^3), there is a low probability of fracture. The probability of fracture becomes moderate for drivers with a density of 0.2 g/cm^3 and a very low damping (10%).

The third effect is the interaction between the acceleration amplitude and the density (Figure 6). The risk of adverse effect increases with acceleration amplitude and decrease of density. The drivers with a very low density (0.3 g/cm^3) present a high probability of fracture. If it is high (0.3 g/cm^3), there is a low probability of fracture. The probability of fracture becomes moderate for drivers with a density of 0.2 g/cm^3 exposed to vibration amplitude greater than 3 m/s^2.

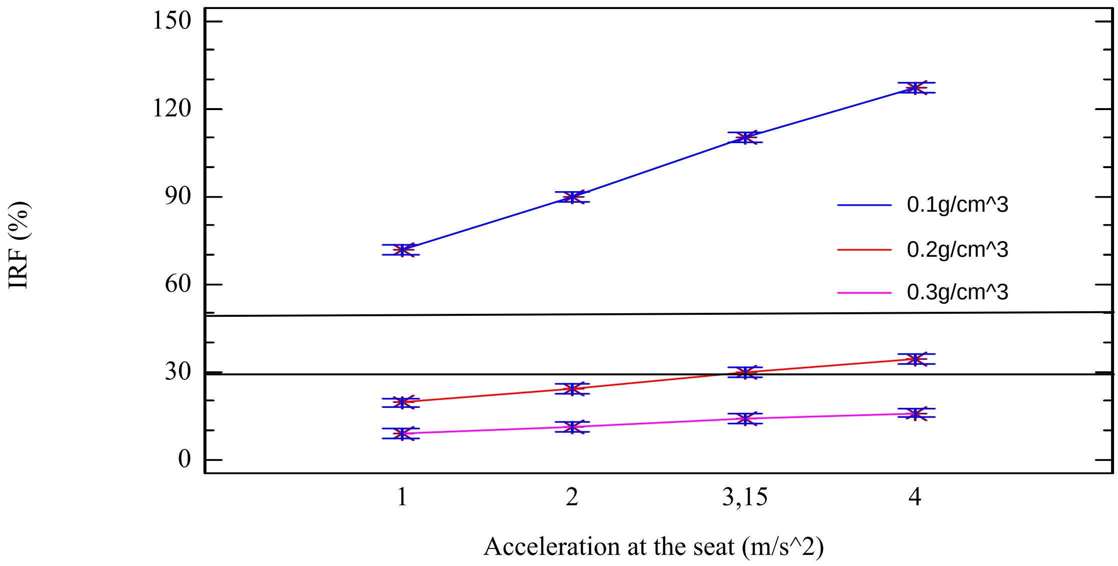

Figure 6 Interaction of acceleration and density on IRF

The fourth interaction in importance is the effect of area (bone structure) and density on IRF (Figure 7). The risk of adverse health effect increases with a decrease of the lumbar area and with density. The probability of injury is high for drivers with a very low density.

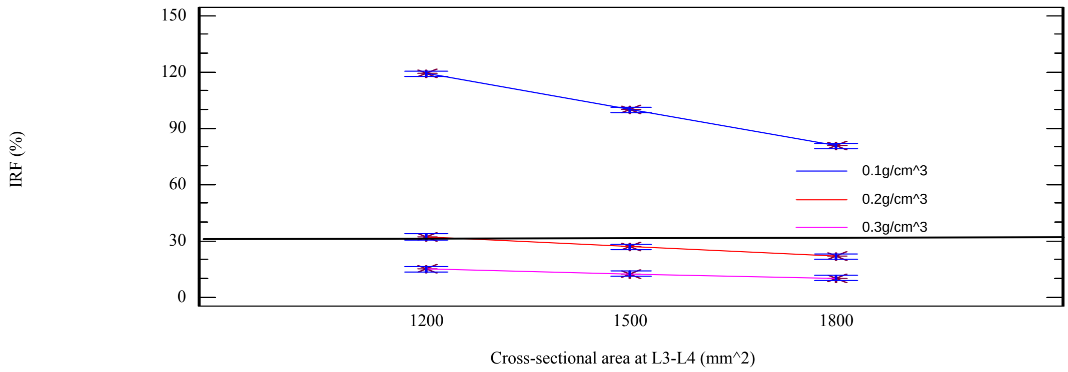

Figure 7: Interaction of area and density on IRF

If we consider a level of IRF of 50 % as critical and 30% as a moderate probability of injury, it can be concluded from figures 4 to 7 that a density of 0.2 g/cm^3 can be considered as threshold. A lower density is critical for the adverse health, whatever the mass, the lumbar area, the damping rate or the acceleration level. The 0.1 g/cm^3, 0.2 g/cm^3, 0.3g/cm^3 levels of density correspond to persons of 65, 45 and 25 years old respectively [16 - 19]. This study puts in evidence that the risk of fracture increases significantly for older persons.

The effects of the interactions implicating the damping rate and acceleration at the seat or body weight have also an effect significant on the injury risk factor. Figure 8 describes the interaction between the body weight and damping on IRF. The risk of adverse health increases with mass and with a decrease of damping. The probability of injury is moderate for drivers with damping rates greater than 10%. The probability of injury becomes high fir drivers with weight greater than 75 kg and a low damping rate (10%).

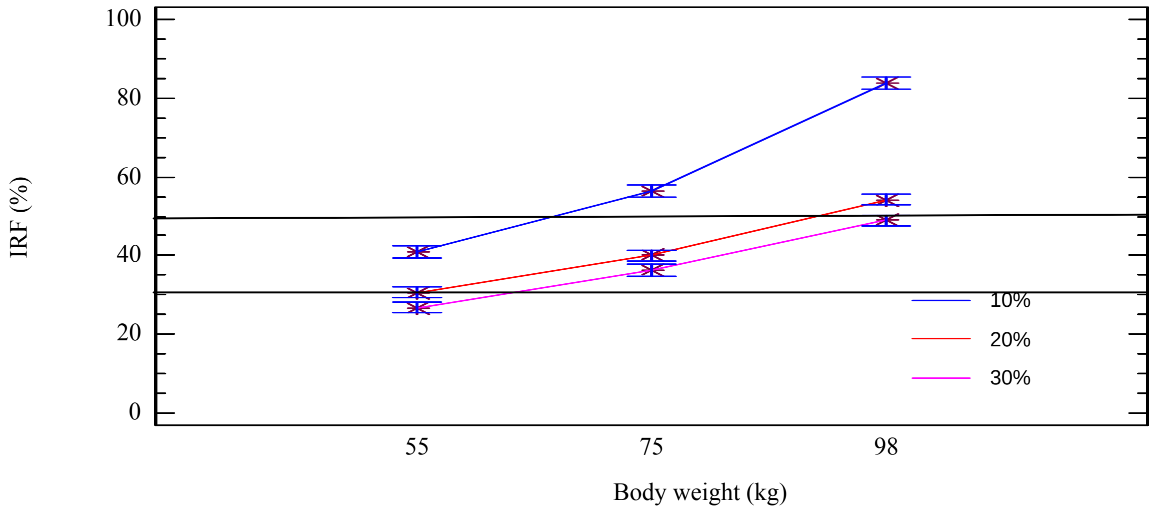

Figure 8 Interaction between mass and damping on IRF

Figure 9 describes the interaction between the acceleration level and the damping. As expected, it is shown that the IRF increases with acceleration and decrease of damping. This effect is more pronounced for a low damping.

If we consider an IRF of 30% as a threshold for a moderate probability of injury and 50% for a high probability, this result shows that old drivers with a low damping present a risk from moderate to high and that the others present a moderate risk of Injury. Te acceleration amplitude must be maintained to levels lower than 2 m/s^2 for old drivers.

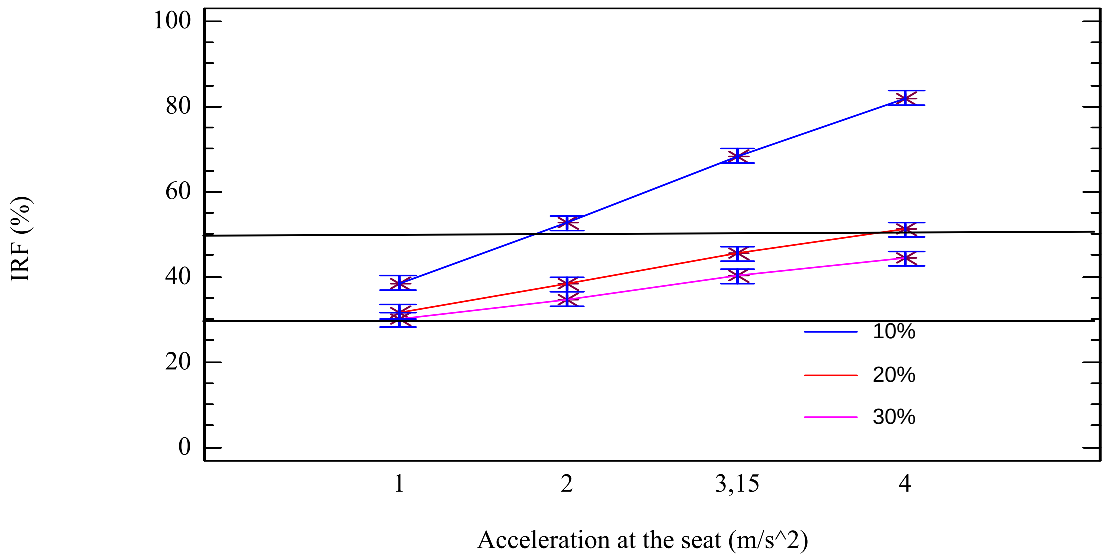

Figure 9 Interaction of acceleration and damping on IRF

As seen previously, the risk of adverse health increases with the weight (Fig. 10).

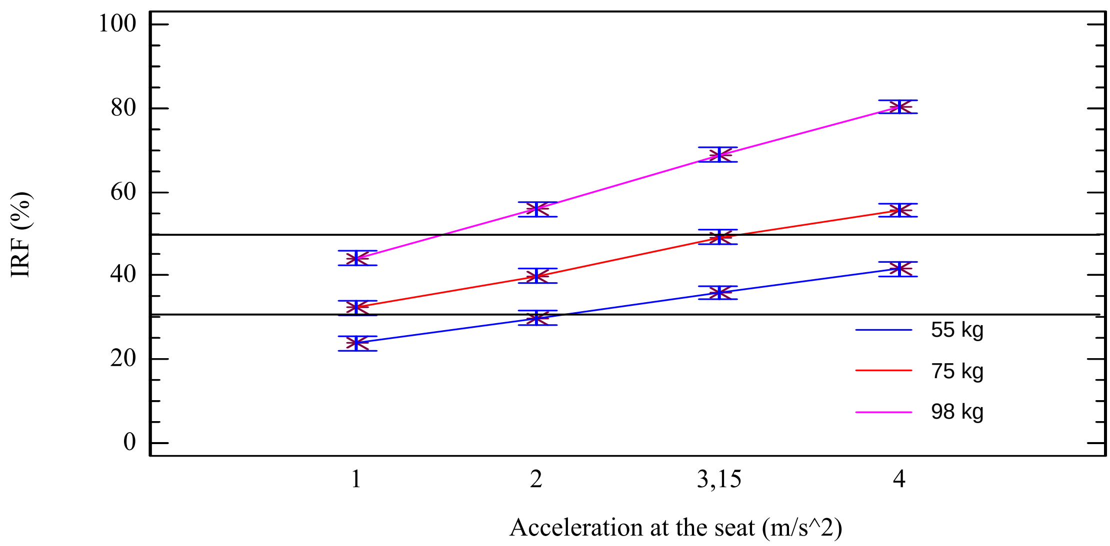

Figure 10: Interactions between acceleration and weight

Very heavy drivers are subject to a high probability of injury if the acceleration amplitude is greater than 1.5 m/s^2. The drivers with a medium weight have a moderate probability of injury and the drivers with a light body weight, a low probability of injury for acceleration amplitudes lower than 2 m/s^2.

The effects of body weight and bone structure (area) on IRF are shown in Figure 11. The risk of adverse health effect increases with the body weight and with a decrease of the lumbar area. The drivers with a high weight (98 kg) present a high probability of injury.

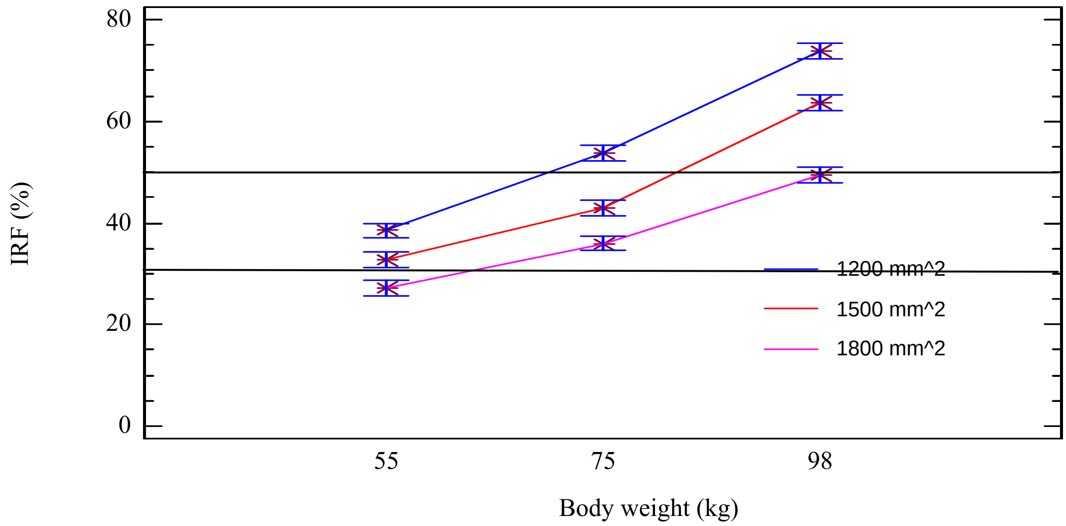

Figure 11 Interaction between weight and area

At a lower level of significance, Figure 12 demonstrates the effect of posture. For a high density, posture has no effect on IRF.

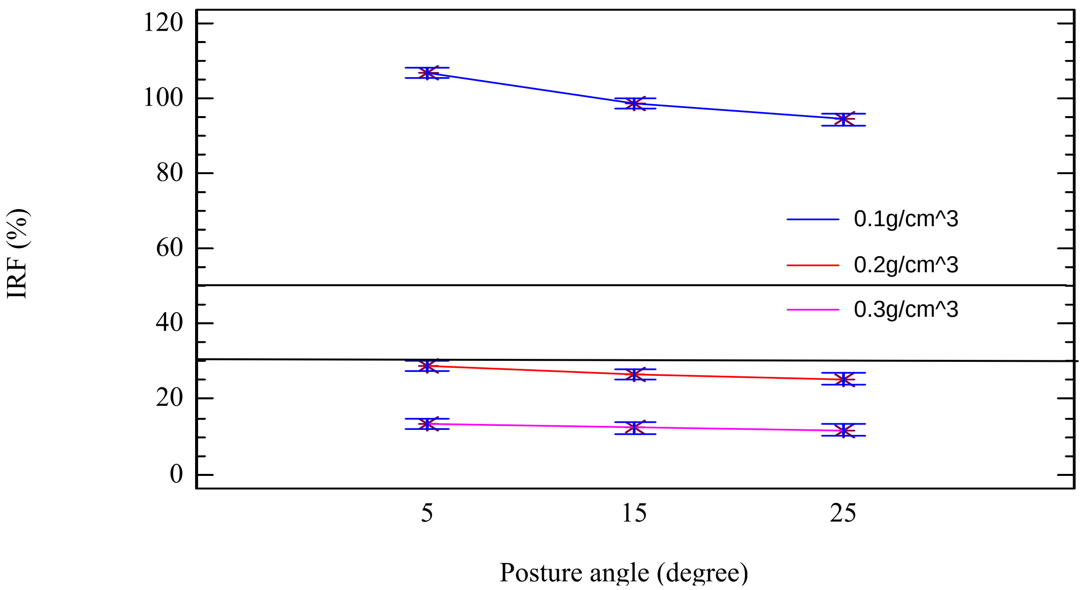

Figure 12 Interaction of posture and damping on IRF

4. DISCUSSION

By performing a non-linear regression analysis on parameters, a model describing the relationship between IRF and the independent variables has been obtained. The equation (4) of the fitted model becomes:

$$IRF(\%) = \frac{1}{S*\rho^{1.9}}\left(715.106 + 1.75\frac{\sqrt{1+(2\xi)^2}}{2\xi}A*M*\cos(\theta)\right) \text{ with } R^2{=}95.7\% \quad (5)$$

The adjusted R-Squared statistic, which is more suitable for comparing models with different numbers of independent variables, indicates that this model explains 95 % of the variability in IRF.

In table 4, we have regrouped the variables for representing the effect of ageing (density and damping) and the effect of morphology (weight and bone area) accordingly to the only parameter that can be controlled: the level of acceleration in the 4-8 Hz frequency range.

The injury risk of fracture (IRF) developed in this research allows for predicting the instantaneous risk of rupture. Table 4 allowed to defined three classes of risk:

- $0 < IRF < 30$ % No risk;
- $30 < IRF < 50$ % Alarm;
- $IRF > 50$ % High risk;

As expected, the results show that the risk of injury increases with ageing because the osseous density and the damping rate decreases. By considering a healthy person, acceleration amplitudes of 3 m/s^2 applied at the seat exciting the lumbar spine at its natural frequency can produce a risk of damage between 10-12 % for a young driver (25 years) by considering an apparent density of 0.3 g/cm^3 and a damping rate about 30%, while, for the same level of excitation, the Injury Risk Factor increases to levels between 16-19 % for a 45 years old driver (apparent density of 0.24 g/cm^3, damping rate of 25%) and between 30-36 % when the driver is 65 years old (apparent density of 0.18 g/cm^3 and damping rate of 20%). By considering an IRF of 30%, Table 4 shows that the acceleration recorded at the seat has no significant effect for young and intermediate drivers (25 and 45 years old), but the risk is moderate for old drivers exposed to acceleration amplitudes greater than 3 m/s^2 and becomes high with its weight (95 Kg) if the acceleration is 6 m/s^2.

In many cases, such as machinery travelling over rough surfaces, dynamic loading should be considered as vibration containing multiple shocks. A method for estimating vibration containing multiple shocks has been produced by ISO 2631-5 [30], which is concerned with the lumbar spine response. The ISO 2631–5 makes it possible to predict the risk of fatigue per endurance. For impact excitation, the ISO 2631-5 standard proposes a procedure for estimating the daily equivalent static compression dose S_{ed} (Mpa).

- If $S_{ed} < 0.5$ Mpa, a 20 years old man working 240 days a year will have a low probability of injury (in green in Table 4);
- If $0.5 < S_{ed} < 0.8$ (Mpa), a 20 years old man working 240 days a year will have a moderate probability of injury (in orange in Table 4);

- If S_{ed} > 0.8 (Mpa), a 20 years old man working 240 days a year will have a high probability of injury (in red in Table 4);.

Table 4: Risk of injury according to morphology and ageing ($\sigma_{tot} = \sigma_{stat} + \sigma_{dyn}$)

			Effect of morphology								
			Light weight			**Medium weight**			**Heavy weight**		
Effect of ageing	**25 years old Density 0.3g/cm^3 damping 30%**	Accel. (m/s^2)	1	3	6	1	3	6	1	3	6
		IRF (%)	7	10	14	8	11	15	9	12	16
		σ_{tot} (Mpa)	0.31	0.42	0.58	0.34	0.46	0.63	0.37	0.5	0.7
	45 years old density 0.24g/cm^3 damping 25%	Accel. (m/s^2)	1	3	6	1	3	6	1	3	6
		IRF (%)	12	16	23	13	17.5	25	14	19	27
		σ_{tot} (Mpa)	0.32	0.44	0.62	0.35	0.48	0.70	0.38	0.53	0.75
	65 years old density 0.18g/cm^3 damping 20%	Accel. (m/s^2)	1	3	6	1	3	6	1	3	6
		IRF (%)	21	30	44	23	33	48	25	36	52
		σ_{tot} (Mpa)	0.33	0.48	0.7	0.36	0.53	0.77	0.4	0.57	0.84

We can notice in table 4, some differences between the IRF and ISO 2631–5. In fact, the IRF is not only based on the applied stress (as it is in ISO2631-5) but also on the ultimate stress. The difference is due to the fact that IRF takes into account the apparent density of the vertebral cancellous bone and damping rate of intervertebral disks. The variations of these values that are affected by the age are perturbing the ultimate stress of bones. By considering an intermediate sized person (intermediate body weight, intermediate angle posture and intermediate bone structure), an acceleration of about 3 m/s^2 can produce a total compressive stress ($\sigma_{tot} = \sigma_{dyn} + \sigma_{stat}$) between 0.46 Mpa to 0.53 Mpa. The risk of fracture for this acceleration level is very about 11% for a 25 years old person, about 17 % for a 45 years old person and 33% for a 65 years old person.

5. CONCLUSION

The design of experiments method allows for statistically analyzing results having a great variability, such as weight, size, sex, age, etc. for a human body. A finite element model of lumbar spine was developed by using a parametric model because this strategy allows for a very quick simulation of various anatomies (robust, average or frail body-types). The dynamic model that has been developed is aimed at computing the dynamic mechanical stresses of the lumbar spine produced by whole-body vibrations of a person in a seated position in order to evaluate the risk of adverse health effect which could occur for professional drivers by considering the effect of the posture, the damping rate, the body weight, the bone structure and the apparent density. This parametric finite element model was validated by using published results of forces calculated at the L3-L4 and the experimental dynamic stresses estimated at the L5-S1. An Injury Risk Factor (IRF) of damage in vertebral body was developed for any given vibratory amplitude coming from the seat. From 972 computations, an ANOVA analysis revealed that the osseous density (and consequently the ageing) is definitely the most important factor affecting the risk of fracture. The effects of the interactions combining density and damping rate were found more significant than all the other combinations of variables on the injury risk factor. It was found that the risk of adverse health is more significant if the apparent densities is lower than 0.2 g/cm3 and if the damping rate is lower than 20%. These effects are related to the age of drivers and this study confirmed that older drivers present a more significant probability of injury than the younger's one. If an Injury Risk Factor of about 30% is defined as the limit of endurance in order to avoid fatigue problems, the results show that drivers older than 45 years old are susceptible to long term injury and that the weight has a significant effect. The level of acceleration must be controlled to amplitudes less than 2 m/s^2 if we want to avoid any risk of injury whatever the driver, his weight, bone surface and age.

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