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Biomechanical analysis and modeling of lumbar belt: parametric study.

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1. Introduction

Low back pain is a major public health problem in European Countries. In France, about 50% of population is suffering of this pathology every year [Fassier, 2011]. Because of health care cost and sick leave [Fassier, 2011], [Leclerc, 2009], low back pain as both societal and economic adverse consequences. Concerning medical devices, such as lumbar belts, several clinical trials have shown their efficiency [Calmels, 1999]. It might be essential to know which parameters are important so as to increase the efficiency of their medical effects on patients. Nevertheless, both mechanical and physiological effects of lumbar belts remain unclear. In this study, a parametrical method is used in order to identify which parameters are more important than others through numerical simulation. This study is indispensable to design lumbar belts. For this purpose, we selected 11 variables according to several clinical tests:

- on the patient: size (fat or slim), height (tall or small), hollow-back (hyper or hypo),
- on the belt: height (2 values), type by varying its linear tension,
- on the mechanical properties: abdomens, other soft tissues of the trunk, annulus, nucleus, bone and skins.

2. Methods

The 3D geometry of the trunk was acquired by parameters measurement in lateral radiography (vertebral length, width and endplate slope) and in patients (bust, waist, hips and stature measurement). Thanks to these parameters, a generic model with six components (vertebras, annulus, nucléus, iliac crests, abdomen and other soft tissues) has been built. All components are represented by tetrahedral elements.

Mechanical properties of all the components of the model were taken from published data [Goel, 1993], [Sylvestre, 2007], [Clin, 2011]. Its mechanical behaviour was considered as linear elastic.

A pressure was applied onto the trunk to simulate lumbar belt wearing. The surface of applied pressure in the model was the same than the area applied on the patient by the belt. Pressure was calculated by the Laplace's law [Dubuis, 2012]:

$$P = T/R \quad (1)$$

Variables	Upper and lower bounds	Type of variable
Morphology	fat or slim	binary
Height	tall or small	binary
Hollow-back	hyper or hypo	binary
Height of the belt	21 or 26 cm	binary
Type of the belt	Lumbaskin® or Lombacross Activity®	binary
Abdomen	0.01 to 1 MPa	continuous
Soft tissue of the trunk	0.55 to 1 MPa	continuous
Annulus	4 to 14.9 MPa	continuous
Nucleus	3 to 8 MPa	continuous
Bone	1000 to 12000 MPa	continuous
Skin	1 to 5 MPa	continuous

Table 1 Chosen variable for parametrical study.

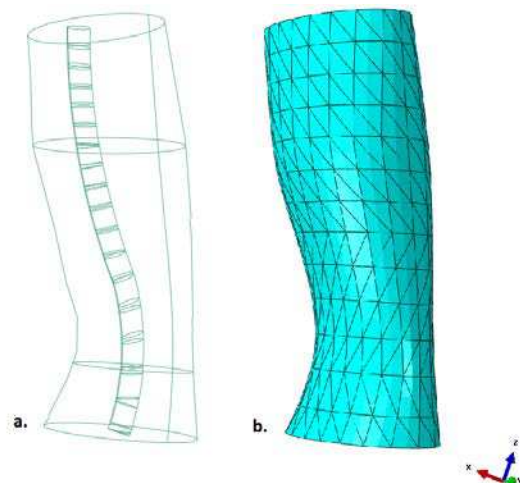


Figure 1 Finite element model of the trunk, a. Entire geometric model, b. Meshing model.

with P the pressure, T the line tension and R the radius of curvature. Linear tension varies with lumbar belt type. As a consequence, mean pressure was between 4 and 13 kPa.

Different boundary conditions were also studied to make sure the validity of the numerical model, because they might give a great influence on the results. Finally, we took the followings: upper surface of the trunk was blocked to only allow translation in longitudinal direction and in lower surface, axis in the frontal plane passing through the iliac crests was blocked.

Parameters were studied by a design of experiments, built by a stratified Latin hypercube methods with 100 experiments. The output parameters are the abdominal pressure, the interdiscal pressure and the lordosis.

Lumbar belt simulation was evaluated thanks to a clinical study by the comparison of the volume variation of the trunk with and without wearing a lumbar belt. Measured pressure during clinical study and applied pressure in the numerical model were compared.

3. Results and Discussion

Figure 1 shows the finite elements model of the trunk. According to the convergence test, optimal model contains more than 1 200 000 elements with more than 1 600 000 nodes.

Modification of the spine posture is characterized by the existence of a displacement gradient (up to 3mm for the trunk and 2.5mm for the spine). The mean abdominal pressure variation is 10 kPa when the stiffer lumbar belt is used (mean applied pressure of 13 kPa).

Interface pressure applied to the model according to the Laplace's law is equivalent to pressure applied by the lumbar belt to a patient according to the clinical study. Volume variation of the trunk when wearing a lumbar belt are equivalent between the numerical and clinical studies.

Parameters having the greatest influence on the abdominal and intradiscal pressure is the mechanical properties of abdomen, annulus, nucleus and other soft tissues. Height and type of lumbar belts have influence on lordosis. Morphology of patients seems to have no significant influence on the three output parameters.

4. Conclusion and future works

This project was a first parametrical study on the mechanical and physiological impact of lumbar

belts. It is a first step to a global numerical and experimental study. It permits to understand which parameters have significant influence on the abdominal, intradiscal pressures and the lordosis when wearing a lumbar belt.

Next steps of this study are the comparison of numerical results to clinical ones (interface pressure and displacement measurement) and the numerical simulation of patient specific models.

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