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FINITE ELEMENT SIMULATION OF DAILY ACTIVITIES HELD BY THE INTERVERTEBRAL DISC

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Abstract: *The spine is a complex system, capable of maintaining stability and simultaneously performing movements and the lumbar region is fundamental for this ability. The present work aims to validate a Finite Elements (FE) model of a lumbar motion segment, through numerical simulation of the required mobility on the intervertebral disc, accordingly to some daily activities. For this study, only the simplest motion on the spine was simulated: flexion, extension and lateral bending. The biomechanical response of the FE model has proved to be suitable for predictions on flexion, but on extension and lateral flexion unexpected extension angles were obtained. These results showed the need of improvements in the mesh geometry, along with the introduction of a model considering of the external ligaments restraining effect, in order to get a more reliable and closer to reality simulation of the all biomechanical system.*

KEYWORDS: Intervertebral Disc; Motion Segment; Flexion/Extension; Lateral Flexion

1 Introduction

The Human spine is a complex and unique system in the animal world, mainly due to its upright posture. However, this posture makes the Human Spine very prone to injuries and degenerative pathologies.

The functional element of the spine is the Motion Segment (MS). Each MS is composed by two vertebrae (VB) connected by one intervertebral disc (IVD) and two facet joints.

The IVDs play a paramount role in Human mobility and trunk flexibility. The IVDs are fibrocartilaginous structures formed by the annulus fibrosus (AF) and the nucleus pulposus (NP) [1, 2]. The NP has a large water content, which allows the radial transmission of forces to the AF. The AF is formed by a complex set of collagen fibers which resists the loads from NP [2].

The non-linear viscoelastic properties of the IVD contribute to loading's absorption, relative displacements and rotation between two adjacent vertebrae, trunk flexibility and the execution of movements as such flexion/extension, lateral bending and rotation [1].

The lower part of the spine (lumbar part) is the most heavily loaded. Besides, in order to maintain body stability, the motions in lumbar IVDs are within a restricted range: the rotation is always low, lateral flexion occurs mainly on L2/L3 IVDs whereas flexion/extension on lower IVDs [3].

However, throughout Human life, IVDs lose naturally and gradually the ability to reabsorb water, and thus their loading capacity and mobility are decreased [3]. Among others, an interesting question is: what is the mobility of our spine when submitted to any well-defined daily activity.

The present work aims to validate a Finite Element (FE) model of a MS accordingly to some daily activities. Such validation has been performed through numerical simulation of the required mobility on IVDs for those activities, with a special emphasis on flexion/extension and lateral flexion.

2 Methods

2.1. FE model

The FE model used replicates the full L3-L4 Human lumbar MS. This model was based on a set of a Human VB and two IVDs previously published by Smit (1996), as described on the work of Castro *et al.* [4,5].

The FE mesh, shown on figure 1, has an average height of 60.9 mm while the IVD has a height of 12.8 mm and an axial cross section of 1555.3 mm^2. Overall, The FE mesh is discretized with 2844 27-node quadratic hexahedra and 26531 nodes. Only vertex nodes are visualized.

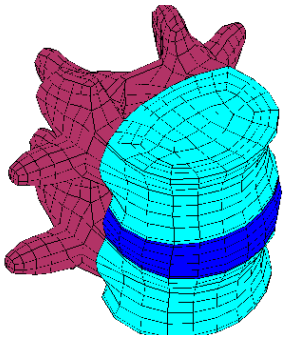

Fig 1. FE mesh of the full MS model

Regarding AF fibers, these are modeled in agreement with the work of Lopes and Alves [6], varying from ±23.2º at ventral position to ±46.6º at dorsal position with respect to the disc circumferential cross plan. Table 1 (see annex A) lists the parameters for the constitutive materials of MS mesh.

The FE simulations were performed with a home-developed open-source FE solver specifically developed to the Human spine.

2.2. Validation Process

The use of FE models offers an invaluable tool to study the biomechanical behavior of the spine. However, to have sufficient confidence in biomechanical response, models require validation by comparing data from *in-vivo* or *in-vitro* experiments.

Though providing important information, data from *in-vivo* experiments have many limitations. The loads applied on the spine are often unknown and the accuracy of the measurement system is usually poor [7].

In-vitro data are collected from experiments with lumbar cadaveric segments under specific conditions. Nevertheless, the artificial experimental conditions and the absence of muscle action can make these data slightly untrue [7].

The behavior of FE model was simulated by imposing both forces and moments. First only pure moments were applied to mimic the flexion/extension, lateral flexion and rotation. These moments were incremented up to 10Nm and during 2s. The results were compared with *in-vitro* studies of Panjabi *et al.* (1994) and Guan *et al.* (2007) [7, 8].

Later, simple motions were simulated using forces and moments from Orthoload database. The analysis of the results was done by comparing both the linear and angular relative displacements under flexion/extension and lateral flexion with the maximum angles for each lumbar IVD according S. S. Tanz [3]. These angles are listed on table 2 (for flexion) and table 3 (for lateral flexion).

Table 2 Flexion/extension angles depending on age according S.S. Tanz [3]

Age	2-13	35-49	50-64	65-77
L1	-	6º	4º	2º
L2	10º	8º	5º	5º
L3	13º	8º	5º	5º
L4	17º	12º	8º	7º
L5	25º	8º	8º	7º

Table 3 Lateral flexion angles depending on age according S.S. Tanz [3]

Age	2-13	35-49	50-64	65-77
L1	12º	5º	6º	4º
L2	12º	8º	7º	7º
L3	16º	8º	8º	6º
L4	15º	8º	7º	5º
L5	7º	2º	1º	0º

2.3 Orthoload database

Orthoload database presents the plotting of both forces and moments measured experimentally by an instrumented Vertebral Body Replacement (VBR) for several activities [9, 10].

To sense data, VBR uses 6 load sensors (2 for each axis) and one telemetry unit.

It is important to highlight that forces and moments are quite different, depending on the patient, and even for the same patient at a different moment. However, the resultant force is close to the axial direction, varying only slightly during exercises.

The resultant moments vary substantially for almost activities [9] although always keep under 2Nm. For example, flexion/extension causes higher moments on sagittal plane whereas lateral flexion origins higher moments on coronal plane. Torsion moments show to be less sensitive [9,10].

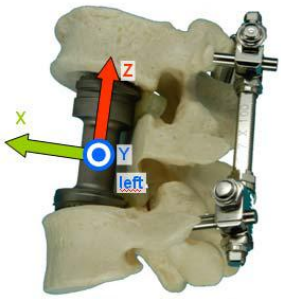

Fig 2. Instrumented VBR with the axis system used for plotting data [9].

3. Results

3.1. Validation with *in-vitro* data

The simulation of pure moments on FE model revealed a nonlinear behavior for all situations. In particular, this behavior is more noticeable during lateral flexion.

The results indicate that the extension is higher than flexion, therefore an unexpected result [8, 9]. Moreover, a pure moment for lateral flexion causes a substantial extension angle, as showed on figure 3.

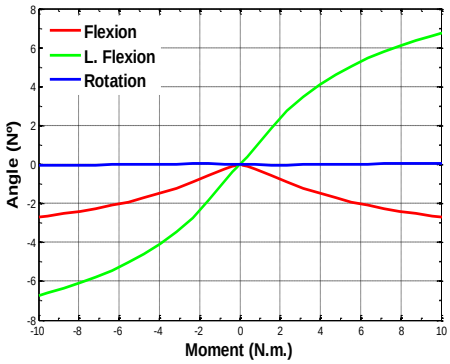

Fig 3 Angles caused by lateral flexion moment

Except for the extension, the trend of computational and experimental curves of Guan *et al.* (2007) [8] presented a good agreement, as it can be seen on figure 4. Besides, the numerical curve falls within the range of experimental standard deviation. Although somewhat different, the same conclusions are also observed when compared with the curve of Panjabi *et al.* (1994) [7].

3.2. Validation with *in-vivo* data

A second step for the validation involved the FE simulation through the imposition of both forces and moments from Orthoload database.

In all situations, in spite of the strong amplitude of axial compressive forces, axial displacement remains relatively low, usually below 1.0mm. On the other hand, the combination of lateral forces (Fx and Fy) and moments caused higher displacements on sagittal and coronal planes.

In what concerns to the angles between the two VBs, the rotation was found to be very low, as expected. In typical flexion motion, the output angles showed to be consistent with the motion performed and the subject's age. However, a typical extension motion caused an extension angle too high. Finally, a lateral flexion motion caused an unexpected high extension angle.

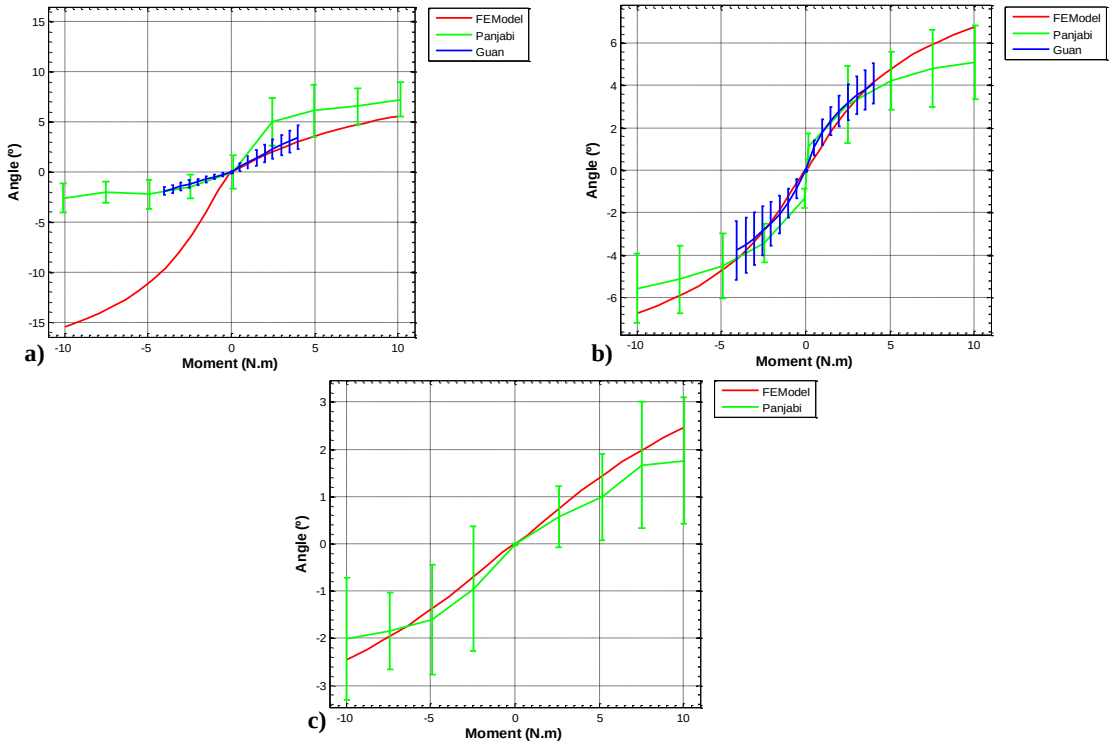

Fig. 4 Comparison between the relative angle predicted by the FE model and that measured by Guan *et al.* (2007) and Panjabi *et al.* (1994) under **a)** flexion-extension **b)** lateral flexion **c)** torsion pure moments [7,8]

4. Discussion

The aim of this study was to validate the FE model accordingly to some daily activities. The behavior of FE model was compared with both *in-vitro* and *in-vivo* studies.

The first evidence from the results is the non-linear behavior of FE model showed by the numeric curve of the model. This behavior was expected because of the viscoelastic nature of both NP and AF.

The simulations also exhibited a much higher extension angle than flexion angle. These results were completely contrary to the experimental results. There may be several explanations for it, but the most likely are the inadequate geometry of the MS mesh (vertebrae are parallel each other) and, mainly, the absence of ligaments on the FEM model.

Except for the extension motion, the trend of computational and experimental curves showed a good correlation. All points of the numeric curve fall within this experimental standard deviation interval. This proof is clearly more visible for Guan *et al.* (2007) experiments [8].

Thus, to make the FE model fully reliable and assertive to reproduce the biomechanics of IVD it will be necessary both to improve the MS mesh and adding the ligament effect to limit the motions, namely on extension.

Simulating the conditions described by Orthoload database, the output angles didn't show coherent with all types of motion performed by the subject. In particular, the angle for extension was clearly superior to the real extension angle. This effect was also seen when a pure extension moment was simulated. Moreover, a lateral flexion exhibited a wide extension angle, as already referred above.

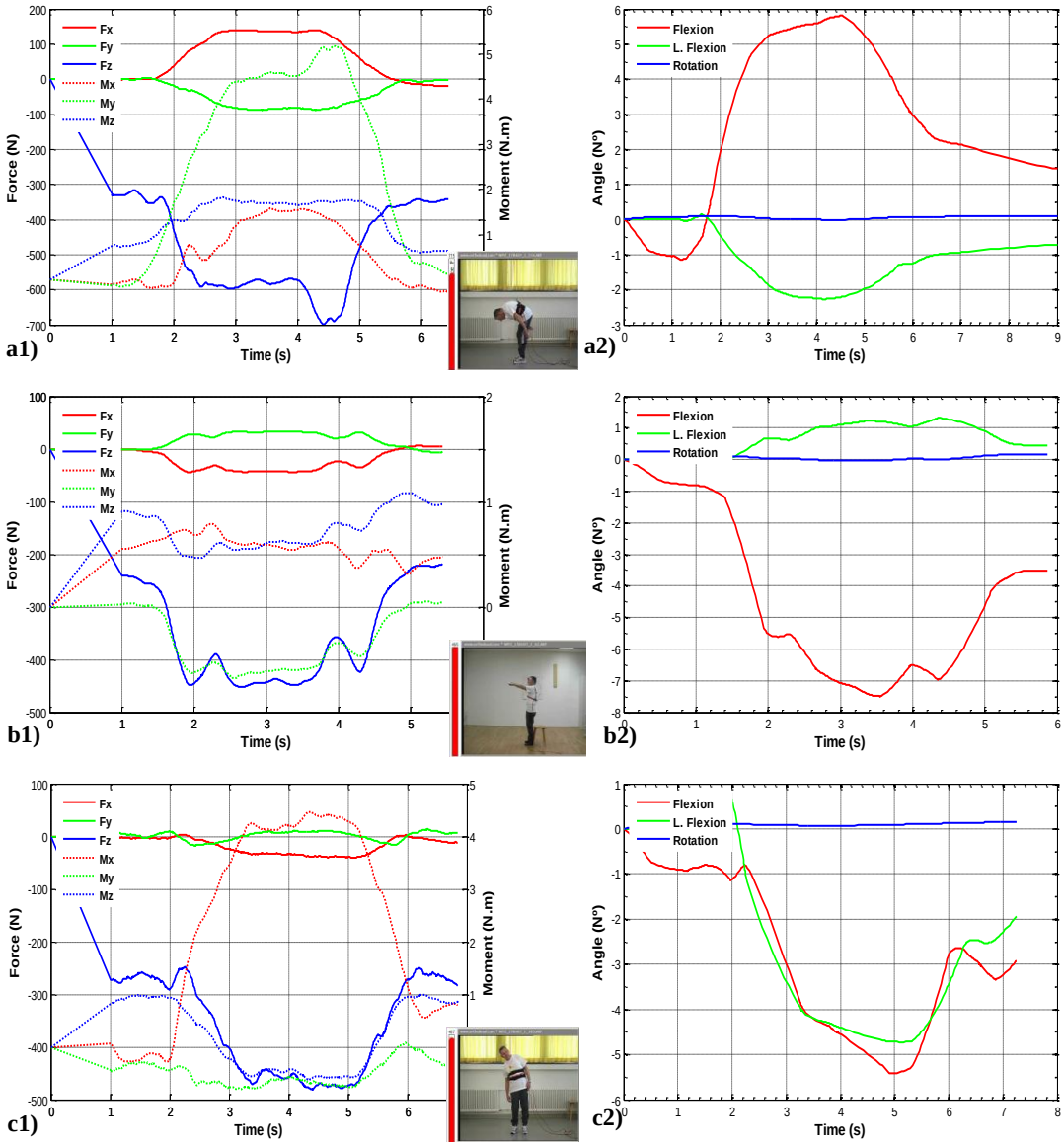

Fig 5 Angles (º) and axial displacement (mm) obtained by computational simulation using Orthoload data
a) Flexion motion: **1**-Forces and moments applied; **2**-Angles obtained
b) Extension motion: **1**-Forces and moments applied; **2**-Angles obtained
c) Lateral flexion motion: **1**-Forces and moments applied; **2**-Angles obtained.

5. Conclusion

As mentioned, IVDs absorb loads and allow spine flexibility which, in turn, presupposes a combination of forces and motions applied to the IVDs.

Through the MS model and the home-developed open-source FE solver was possible to simulate the biomechanical behavior of a MS performing several daily activities.

Excluding the extension angle, the angles obtained by simulations were consistent with experimental studies of Guan *et al.* and Panjabi *et al.* Besides, the angles were consistent with both the age and the type of activity being performed by an individual from Orthoload database.

In a near future, the improvement of the geometry of the MS mesh is been faced, in order to investigate its influence on the overall performance prediction of the IVD.

Simultaneously, adding the role of the ligaments looks like an essential step in the development of the simulation, to pursue a fully reliable model.

6. Acknowledgements

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Annex A

Table 1 Material properties of the MS components: NP, AF, CEP (Cartilaginous endplate), TB (trabecular bone), CB (cortical bone) and FJ (facet joints). These properties were based on literature data.

		NP	AF	CEP	TB	CB	FJ
Isotropy [12]	C_{10} [MPa]	0.003	0.05	1.00	1300	1300	1300
	C_{01} [MPa]	0.0	0.045	0.00	300	300	300
Anisotropy [13]	$\bar{k}$	-	300.0		-		
	$k_4 = k_6$ [MPa]	-	12.0		-		
Viscoelasticity [14,15]	a_1	1.7	1.7		-		
	τ_1 [s]	11.765	11.765		-		
	a_2	1.2	1.2		-		
	τ_2 [s]	1.100	1.100		-		
	a_3	2.0	2.0		-		
	τ_3 [s]	0.132	0.132		-		
	a_4	6.0	6.0		-		
	τ_4 [s]	0.01	0.01		-		