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Title: Coordination between Trunk Muscles, Thoracolumbar Fascia, and Intra-abdominal Pressure towards Static Spine Stability

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ABSTRACT

Study Design: Numerical in-silico human spine stability finite element analysis. *Objective:* The purpose of this study was to investigate the contribution of major torso tissues towards static spine stability, mainly the thoracolumbar fascia (TLF), abdominal wall with its intra-abdominal pressure (IAP), and spinal muscles inclusive of their intramuscular pressure (IMP). *Summary of Background Data:* Given the numerous redundancies involved in the spine, current methodologies for assessing static spinal stability are limited to specific tissues and could lead to inconclusive results. A 3-dimensional finite element model of the spine, with structured analysis of major torso tissues, allows for objective investigation of static spine stability. *Methods:* A novel previously fully validated spine model was employed. Major torso tissues, mainly the muscles, TLF, and IAP were individually, and in combinations, activated under a 350N external spine perturbation. The stability contribution exerted by these tissues, or their ability to restore the spine to the unperturbed position, was assessed in different case-scenarios. *Results*: Individual activations recorded significantly different stability contributions, with the highest being the TLF at 75%. Combined or synergistic activations showed an increase of up to 93% stability contribution when all tissues were simultaneously activated with a corresponding decrease in the tensile load exerted by the tissues themselves. *Conclusion:* This investigation demonstrated torso tissues exhibiting different roles towards static spine stability. The TLF appeared able to dissipate and absorb excessive loads, the muscles acted as antagonistic to external perturbations, and the IAP played a role limiting movement. Furthermore, the different combinations explored suggested an optimized engagement and coordination between different tissues to achieve a specific task, while minimizing individual work.

KEY WORDS

Spine, finite element model, simulation, fascia, paraspinals, muscle activation, abdominal pressure, thoracolumbar fascia, spine stability, low-back pain.

KEY POINTS

- This study evaluated static spinal stability as provided by the synergistic activation of torso tissues.
- Thoracolumbar fascia, major torso muscles, and intra-abdominal pressure were major stabilizers, with a 93% overall stability contribution.
- The explored model suggested an optimized behavior provided by the surrounding active and passive tissues.

INTRODUCTION

Spine's stability is believed to be maintained through the coordination of adjacent tissues [1], [2]. In essence, paraspinal muscles are always thought of as first stabilizers, whereby spinal stability is supported by a combination of muscle effort. Paraspinal muscle co-activation increases spine's compressive forces, stiffening the spine in all potential instability modes [3], [4]. However, although muscle internal pressure, commonly referred to as intra-muscular pressure (IMP), has been proven to play an essential role in muscle contraction [5], [6], the harmony between IMP and the muscle structure in potentially providing spinal stability is usually disregarded [7], [8]. In addition, intra-abdominal pressure (IAP) is believed to stabilize the spine as illustrated via experimental [9] and analytical [8], [10], [11] studies. However, the coordination between the

abdominal wall and IAP towards stability is often overlooked. Recently, the thoracolumbar fascia (TLF) has enticed researchers to investigate its role in load transfer mechanism, providing a foundational support to contacting tissues [12], [13]. Although TLF has been mainly explored via mathematical modelling [14], [15] and rather simplified geometries [16], its role as a static spinal stabilizer is gaining in acceptance.

The clinical quantification of spinal stability, as provided by the coordination of spinal muscles with their IMP, IAP, and TLF, is an arduous task due to the high number of tissues involved [17]. This limits *in vivo* studies but highlights the use of modelling, mainly finite elements (FE), as an excellent experimental platform to explore complex biomechanical problems [18]. Consequently, a timely, accurate, and fully representative FE human spine model has been previously developed and extensively validated by the authors for the purpose of carrying out spinal stability analyses [19], which are otherwise not possible via *ex vivo* and *in vivo* platforms.

Therefore, the purpose of the present study was to objectively investigate, via a fully controlled research platform, the individual and collective contribution of major torso tissues, mainly TLF, abdominal wall with its IAP, and spinal muscles with their IMP, towards static spinal stability.

MATERIALS AND METHODS

Brief FE Model Description

The developed FE model was based on MRI-scans acquired from an anatomography; a database of 3D MRI-based human body parts, namely, “BodyParts3D/Anatomography”. It consisted of 17 vertebral bodies (12 thoracic and 5 lumbar) linked by 16 intervertebral discs, modelled as deformable volumetric bodies. The TLF was segmented from multiple scans and also modelled

as a deformable body. The longissimus, multifidus, psoas major, lateral intertransversarius, and latissimus dorsi muscles were included and modelled as fluid-filled pressurized tissues comprised of two-state, fluid and structure fields, as previously shown to be valid [20]. Lastly, IAP was modelled as a pressure build-up enclosed by an abdominal cavity, defined by the abdominal muscles. The full model is shown in figure 1. The model was then meshed via a novel technique, forcing adjacent surfaces to share the same nodes, whereby FE fixed contacts computations would be eliminated. This was achieved in steps using SpaceClaim (v.19.1, Concord, Massachusetts, United States) to define geometrical components and their associated visual meshes, then Blender (v.2.83.5, Netherlands) to align mesh nodes on contacting objects, and lastly ANSYS (v.19.1, Canonsburg, Pennsylvania, United States) to combine and transform visual into numerical meshes (Figure 2). The resulting element size was 3mm, with tetrahedral elements created for volumetric bodies while triangular elements for surfaces of fluid-filled tissues. Detailed mesh characteristics can be found in another complementary study [19]. Lastly, material properties were incorporated [19].

Boundary and Loading Conditions

The model was previously validated against a numerical model, amongst many, constructed in LifeMOD [21], whereby a flexion force ranging from 0 to 350 N was applied on the first thoracic vertebra (T_1), and displacements of vertebral bodies T_{10} to L_5 were recorded. The pressure in the intervertebral discs (IVDs) was recorded and validated against normal physiological ranges.

Static spinal stability is defined as the spine's ability to retrieve its initial position following an applied external perturbation [22]. As such, for the current study, the validated maximum T_1

force of 350N, imposing a flexion, was used as the external perturbation (Figure 1). Thereafter, a specific set of soft tissues were activated to investigate their ability to restore the spine to its initial, unperturbed, position. Those activated tissues were based on a series of case-studies to be detailed. During each test, the sacrum's position was fixed. Furthermore, in tests where the TLF was included, the extremities of the tendons attached to the latissimus dorsi, the back muscle in contact with the TLF, were also fixed. All other tissues were free to deform, translate, and rotate in all degrees of freedom.

Tests

- Case 0 (Baseline): In this test, none of the tissues were activated. This served as the comparator, whereby under a forward flexion of 350N, L_1 to L_5 vertebral forward displacements, and IVD_1 to IVD_5 pressures were recorded.

The results of all subsequent tests were compared to case 0 in order to find the involvement of the tissue of interest to spinal stability, thus finding a percentage stability contribution computed by:

$$\%_{Stability\ Contribution} = \frac{|(Average\ Vertebral\ Displacements)_i - (Average\ Vertebral\ Displacements)_0|}{(Average\ Vertebral\ Displacements)_0} \times 100$$

where, *i is case 1, 2, …, or 8; while 0 is the baseline (case 0).*

- Case 1: Muscles were included as passive tissues only to investigate their individual stability contribution under passive conditions. With IMP being coded as active pressurized fluidic

component inside each muscle, all such inputs were disabled in this case. Furthermore, all muscular force inputs, modelled as actin-myosin active components, were deactivated. This limits muscles' capabilities to only passive contraction generated by inherent material behavior and properties.

- Case 2: Muscles were included and activated for tensional force and corresponding IMP. In this test, along with subsequent ones inclusive of muscles, recorded EMG muscle forces [23] were introduced in each corresponding muscle as an antagonistic effect to the applied perturbation. These forces were also previously utilized to validate the model.
- Case 3: The thoracolumbar fascia (TLF) was solely activated to investigate its individual stability contribution.
- Case 4: The abdominal wall, via its IAP, was solely activated to investigate its individual stability contribution. In this particular case, along with all subsequent ones where IAP was included, a 30 mmHg abdominal pressure was introduced [19], [24].
- Case 5: Muscles were activated and the TLF was included.
- Case 6: Muscles and IAP were both activated.
- Case 7: IAP was activated and the TLF was included.
- Case 8: All tissues were included. With the muscles and IAP activated, as well as the inclusion of TLF, the overall stability contribution of the major torso tissues was investigated.

RESULTS

In accordance with previously validated results [19], for the baseline (Case 0), forward vertebral body displacements were between 6.1 to 1 cm, in the anterior direction, from L_1 to L_5

respectively and in a decreasing trend. IVD pressure increased from 0.50 to 0.54 MPa between IVD_1 to IVD_5, also mimicking physiological documented values [19], [25].

Under the identical conditions, case 1 recorded slightly smaller vertebral displacements (8% stability contribution) and IVD pressures (2%) when muscles were included as passive tissues only (Figures 3 and 4). However, when muscles were activated, as per test 2, displacement results decreased to almost half of the baseline (53% stability contribution), with measured vertebral displacements between 2.9 and 0.4 cm. Whereas an increasing IVDs pressure from 0.28 to 0.34 MPa in the lumbar region was measured (Figures 3 and 4), accounting for a 40% average difference from the baseline. Muscles activations were accompanied with an intramuscular pressure (IMP) varying from 258 mmHg for the Psoas Major (P), 372 mmHg for the Longissimus (L), 94 mmHg for the Multifidus (M), to barely 12 mmHg for all Intertransversarius (I) muscles (Figure 7).

On the other hand, disregarding the muscular system, activating only the TLF, as per test 3, vertebral displacements decreased to a range of 1.4 to 0.3 cm (75% stability contribution). The measured IVD pressures were between 0.24 to 0.29 MPa, an average 49% decrease from the baseline (Figures 3 and 4). As for test 4, solely activating the abdomen with a 30 mmHg IAP showed the least individual tissue stability contribution (25%), recording a range of 4.6 to 0.7 cm vertebral body displacements and 0.32 to 0.42 MPa IVDs pressure with a 29% change from the baseline (Figures 3 and 4).

As explained, tests 5 through 7 investigated a combination of the tissues of interest while test 8 included all tissues and activation together. Figures 5 and 6 present the results of these scenarios, with figure 7 reporting the developed IMP for cases where muscles were activated. All aforementioned results were also numerically quantified in Table 1. As introduced earlier,

stability contribution is the ability of a specific tissue to participate towards spine static stability, calculated in Table 1 as a percentage stability contribution relative to the baseline (Case 0) using the average vertebral body displacements of each case as explained in the materials and methods section.

DISCUSSION

Computational biomechanics by means of finite element models offers an objective and controlled platform to accurately represent and study the behavior of the human torso. When such models are appropriately validated and credible, they offer a complementary experimental platform to other *ex vivo* or *in vivo* studies. A number of prior research studies have analyzed spine stability, laying the foundations for this perplexing problem. Given the high number of tissues coordinating to achieve stability, simplifications are often required in experimental studies. These often include neglecting IMP and the TLF, or focusing only on active tissues, which may hinder results. Consequently, in the present study, a previously constructed and extensively validated novel three-dimensional full-scale FE model of the spine was leveraged towards analyzing individual and synergistic tissue contribution to static spinal stability. The model not only extends beyond the lumbar region and includes full thoracic spine, but also accurately represents IMP involved in active muscle contraction, IAP buildup inside the abdomen, and the full TLF tissue [19]. In the present study, each of the aforementioned tissue inclusions were considered individually and in combinations to explore their contribution to stability of the spine under external perturbations.

Under a 350N forward flexion force causing a perturbation, case 0 was performed to provide a reference or a validated comparator for all other cases, both in terms of vertebral displacements [21] and IVD pressures [25]. When investigating passive conditions (case 1), passive muscles contributed to about 8% of the simulated spine stability (Table 1), supporting previous claims made regarding the role of muscles coordination as active tissues towards stability and locomotion [26]. This was further emphasized by the results of case 2 when muscles were actively engaged, showing an individual contribution of 53% towards spine stability. This agrees with the potential antagonistic role of muscles to counter excessive loads faced by the spine under external perturbations. Such results, as well as previous studies [4], [27], show the importance of increased muscle endurance to maintain spinal stability and as a protective measure against spinal deformities and conditions, such as low back pain (LBP). In other words, rehabilitation and clinical strengthening procedures of back muscles would increase muscular endurance, providing higher spinal stability, which would potentially help LBP patients. Besides, since a scoliotic spine is characterized by intrinsic instability [28], increasing spinal stability via muscle activation exercises can be a therapeutic strategy. However, spinal deformities are attributed to different causes, and are characterized by different bony alignments, to which physiotherapists should be careful which muscle groups, if any, to activate.

The results of individually including the TLF (case 3), showed a significant 75% contribution to stabilizing the spine as defined by static equilibrium. This contribution agrees with others who have researched or alluded the implications of the TLF in support of the spine [12]–[14]. To better understand its role, the spine-TLF junctions were analyzed, showing elevated levels of tension developing at the contact points, which also agree with prior TLF studies [16]. As such, disrupting the fascial anatomy at TLF joints could reflect much less tensile forces, prohibiting

the TLF from performing its stability role. Surgeons are thus advised to conduct minimally invasive spinal surgeries, such as IVD discectomy, without much TLF disruption, especially at its attachment points, to maintain higher spinal stability. Besides, stability provided by the TLF could have important clinical rehabilitation implications for patients with LBP. Studies have shown that fascia is the most sensitive deep tissue to pain in the lower back [29], and its instability may hence play a major role in acute localized LBP. Physiotherapeutic exercises to strengthen and stabilize the TLF, following spinal surgeries, could thus be of great help to limit the possibility of developing LBP episodes.

When including the abdomen with its IAP, results also support its growing understanding towards its potential role in spinal stability [30]–[32]. This measured 25% stability contribution is significant for a relatively small 30 mmHg pressure build-up. Taking a closer look at the frontal side of the spine, the IAP vectors, by which it provided force interaction to the spine, seemed to have developed and converged to the anterior faces of the vertebrae bodies. As such, this suggests the resistive role of the IAP, exerting a counter effort to the spine during external perturbations imparting flexion. Thus, breathing and abdominal muscles activation exercises could be of great help to impose elevated IAP levels, within normal physiological ranges [33], to increase spinal stability.

When tissues were iteratively activated together, the spine stability increased but to different proportions when compared to those observed with individual tissues. This selective approach lends insight on the inter-coordination between these tissues in concert towards providing spinal stability and what influence they have on each other. That is, the TLF-muscle (case 5) combination led to a significant increase in stability but observed decrease in the muscle IMP. This may support the role of the TLF in support of the muscles as well as the spine, whereby the

presence of the TLF may lead to less stress being imparted on the muscles. This is otherwise interpretable as requiring less muscle force to maintain stability as others have implied in the past [14], [15]. The same appears to apply for cases 6 and 7 where the activation of IAP and muscles, as well as the TLF with IAP, may achieve a point where both tissues coordinate to stabilize the spine while their prior measure individual efforts are reduced consequently. Lastly, the activation of all tissues together, as explored with case 8, supports this coordinative load sharing notion provided the individual effort of each tissue drops compared to all other cases explored. Specifically, an example is the drop in IMP for case 8, compared to cases 2, 5, and 6 (Table 1), while maintaining a 93% stability contribution. Case 8 does not only shine light on a potential optimized tissue activation, but further supports the notion that the tissues under consideration in the present study can be considered major players in static spinal stability.

Given the impact on stability of the tissues explored herein, accurate inclusion and modelling of such spinal tissues would provide great insights regarding designing and optimizing spinal instrumentations. Such efforts are in sync with growing trends that leverage numerical models in medical device design [34]. That is, validated models with representative spinal tissues can be used to compliment bench and clinical data towards the improvement and reliability of spinal instrumentation. Many research groups utilize modeling within their device design framework or assessment strategies [35]–[43]. The authors thus opine that as modeling and computing power progresses, it presents an opportunity to improve numerical models to be as physiologically accurate conditions as possible, to clinical initial and boundary conditions, perhaps leading to improve biomechanical understandings and corresponding device design.

Limitations

Similar to any *in silico* or finite element model, limitations are always present due to the model's numerical approach. However, with assumptions kept to a minimum, this does not limit model's ability to approximate and critically explore the overall biomechanical behavior of the spine. Some common modelling approximations, such as material properties, mesh, and model application, were justified and supported via previous successful validation efforts [19]. Another potential limitation is neglecting the contribution of the other torso tissues not explored herein such as ligaments and the rib cage. With the ligaments' role mostly coming into effect under high deformations [44], their elimination is a reasonable assumption as the present study only considered small static physiological loadings. In addition, the stabilizing action of the rib cage is mostly described by the dynamic respiration process [45], where it is believed to apply supportive spinal load as the lungs inflate. This effect is beyond the scope of this research, with static conditions being the primary focus. Under the above limitations, the model proved robust and underwent extensive validation thus lending credibility and confidence to the relative tissue contribution towards static spine stability discussed herein.

In conclusion, this study leveraged a novel validated 3-dimensional finite element model of the spine inclusive of vertebral bodies, intervertebral discs, thoracolumbar fascia, accurate modelling of the abdominal pressure, and a fully representative model of the torso muscles to investigate their individual and combined contribution to static spinal stability. Several on-off case tests of these tissues were conducted and each revealed their respective and combined stability contribution. These novel analyses may provide insight towards how static spinal stability, as perceived in the present study, can be in part achieved or attempted via individual and/or combined torso tissue engagements.

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Conflict of Interest and Source of Funding

To the best of our knowledge, this research has no conflict of interest for any of the authors.

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Device Status/Drug Statement

The manuscript submitted does not contain information about medical device(s)/drug(s).

Tables:

Cases	VBs Displacements (cm)					IVDs Pressure (MPa)					IMP (mmHg)				Stability Contribution
	L_1	L_2	L_3	L_4	L_5	IVD_1	IVD_2	IVD_3	IVD_4	IVD_5	P	L	M	I	
0	6.1	4.9	3.4	2	1	0.497	0.502	0.514	0.523	0.538	-	-	-	-	-
1	5.8	4.5	3.1	1.8	0.8	0.49	0.496	0.506	0.513	0.528	-	-	-	-	8%
2	2.9	2.3	1.6	0.9	0.4	0.28	0.29	0.304	0.318	0.337	258	372	94	12	53%
3	1.4	1.13	0.88	0.58	0.3	0.238	0.244	0.258	0.272	0.29	-	-	-	-	75%
4	4.6	3.7	2.5	1.5	0.7	0.321	0.348	0.353	0.376	0.418	-	-	-	-	25%
5	0.6	0.48	0.34	0.22	0.11	0.145	0.158	0.17	0.192	0.223	37	146	32	7	89%
6	1.9	1.55	1.1	0.65	0.25	0.2	0.22	0.258	0.268	0.279	132	287	31	10	69%
7	1.24	1.03	0.76	0.5	0.27	0.206	0.22	0.237	0.246	0.26	-	-	-	-	78%
8	0.47	0.37	0.26	0.16	0.08	0.084	0.103	0.12	0.137	0.158	21	128	26	6	93%

Table 1: Results summary for all cases along with their stability contribution.

L_1, L_2, L_3, L_4, and L_5 represent the results for the first, second, third, fourth, and fifth lumbar vertebral bodies respectively. IVD_1, IVD_2, IVD_3, IVD_4, and IVD_5 represent the first, second, third, fourth, and fifth lumbar intervertebral disc respectively. P, L, M, and I represent the psoas major, longissimus, multifidus, and intertransversarius muscles respectively.

FIGURE LEGENDS

Figure 1: Finite element depiction of the utilized spine model.

Figure 2: Depiction and steps realized to generate the finite element mesh.

Figure 3: Vertebral forward displacements results for both the passive (case 1) and active (case 2) muscles conditions, thoracolumbar fascia 'TLF' inclusion (case 3), and intra-abdominal pressure 'IAP' activation (case 4), as compared to the baseline (Case 0). L_1, L_2, L_3, L_4, and L_5 represent the first, second, third, fourth, and fifth lumbar vertebral bodies respectively.

Figure 4: Intervertebral discs pressure results for both the passive (case 1) and active (case 2) muscles conditions, thoracolumbar fascia 'TLF' inclusion (case 3), and intra-abdominal pressure 'IAP' activation (case 4), as compared to the baseline (Case 0). IVD_1, IVD_2, IVD_3, IVD_4, and IVD_5 represent the first, second, third, fourth, and fifth lumbar intervertebral discs respectively.

Figure 5: Vertebral forward displacements results for the different tissue combinations, cases 5 through 8, as compared to the baseline (Case 0).

Figure 6: Intervertebral discs pressure results for the different tissue combinations, cases 5 through 8, as compared to the baseline (Case 0).

Figure 7: Intramuscular pressure 'IMP' results for the different cases in which muscles were activated. P, L, M, and I represent the psoas major, longissimus, multifidus, and intertransversarius muscles respectively.

Figures

Figure 1:

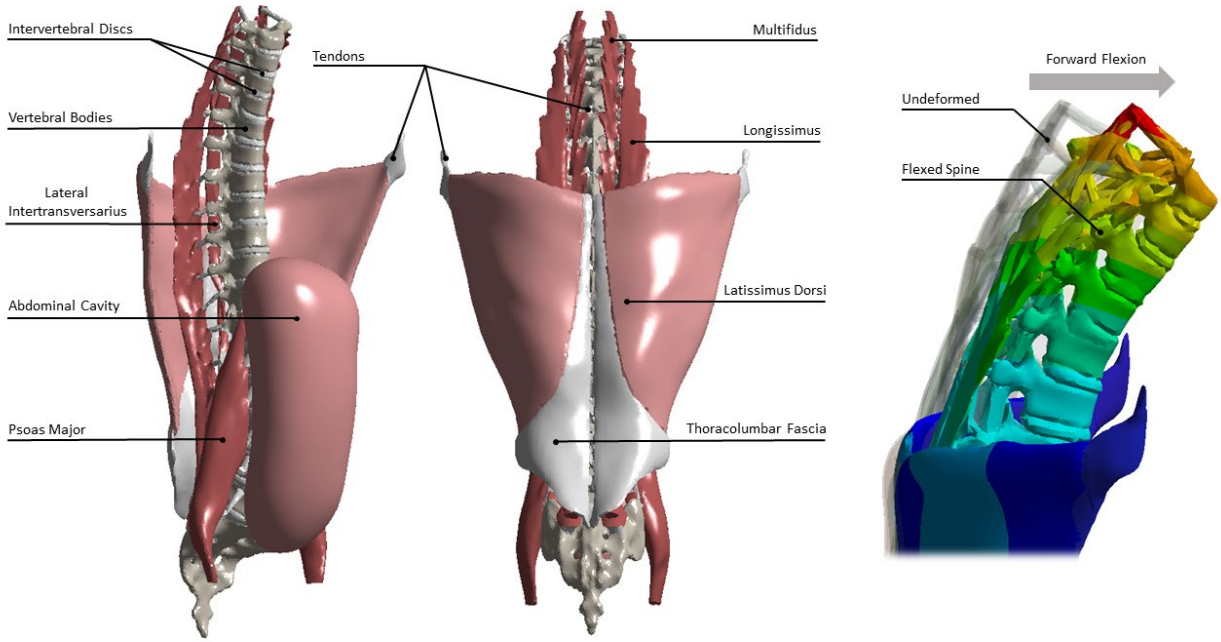


Figure 2:

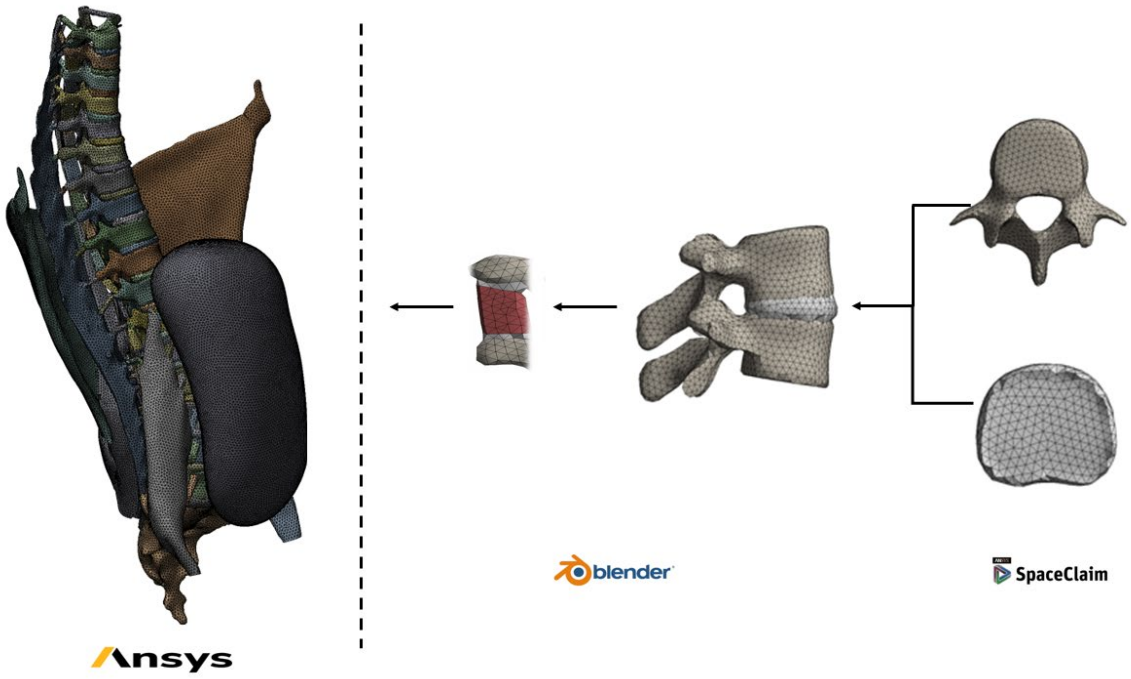


Figure 3:

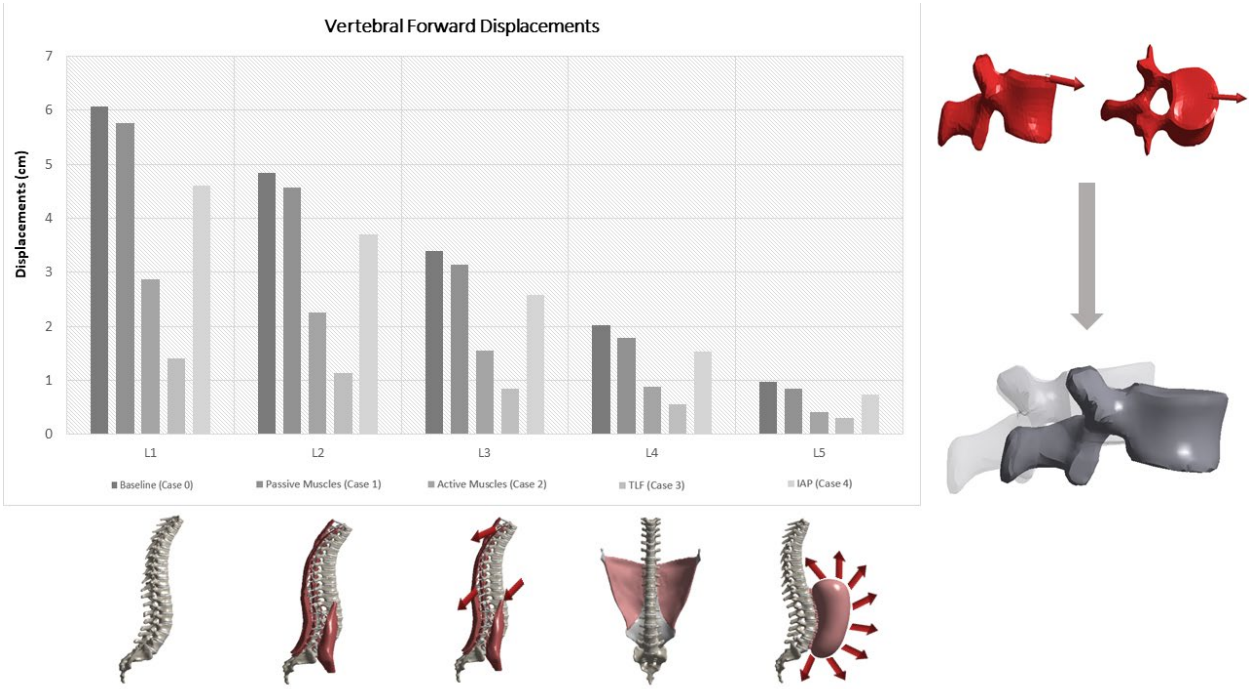


Figure 4:

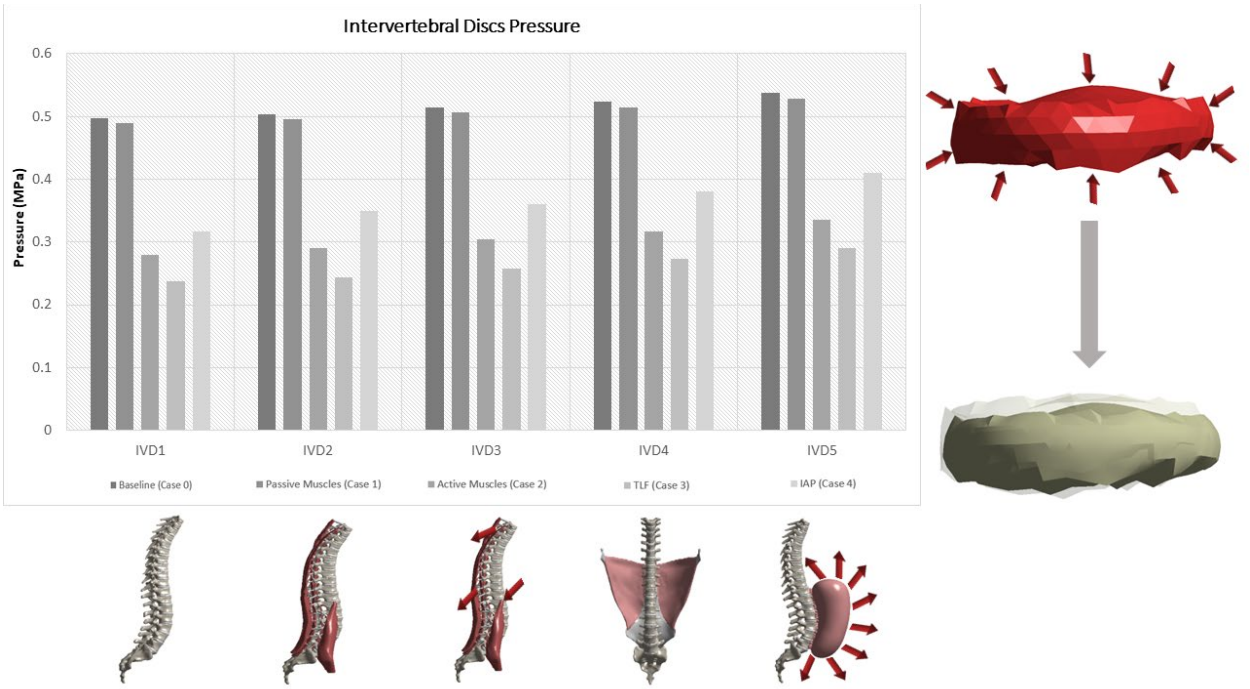


Figure 5:

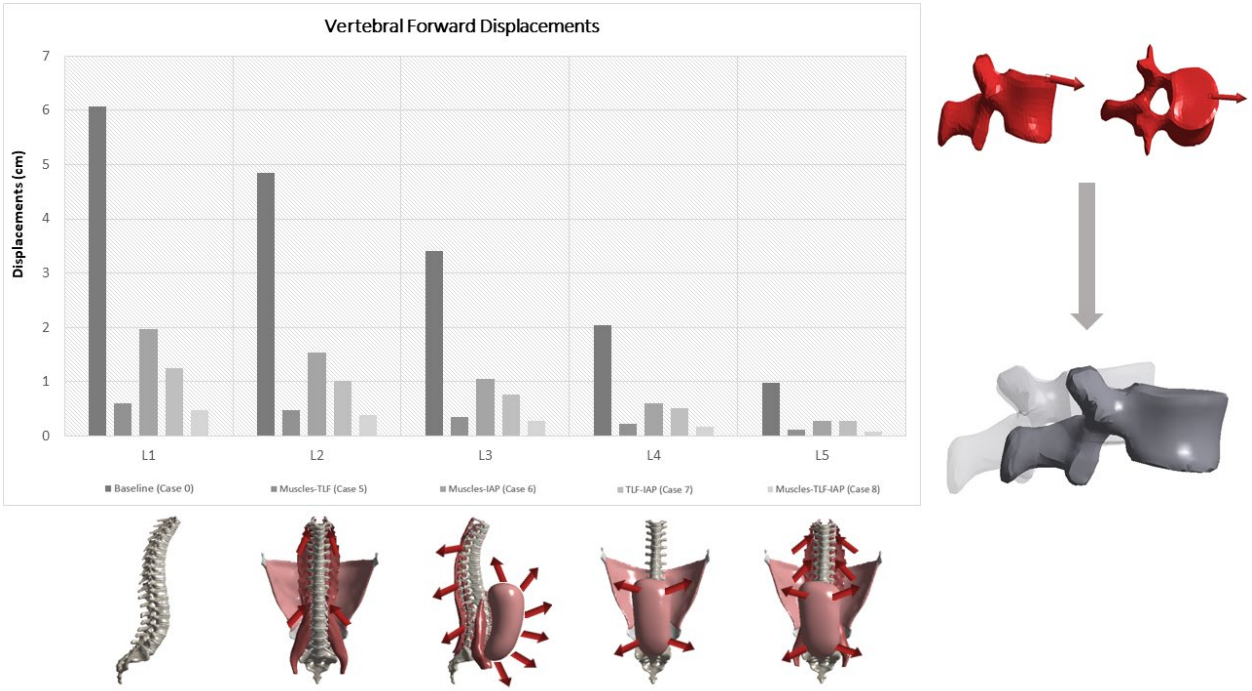


Figure 6:

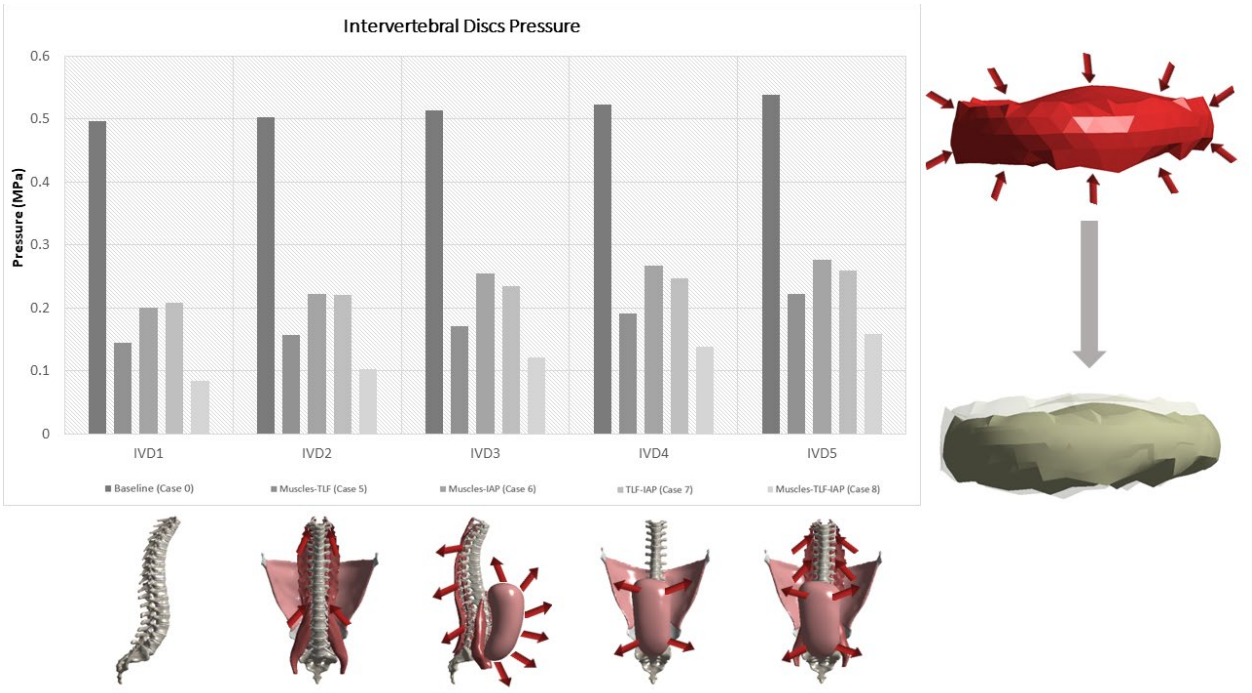
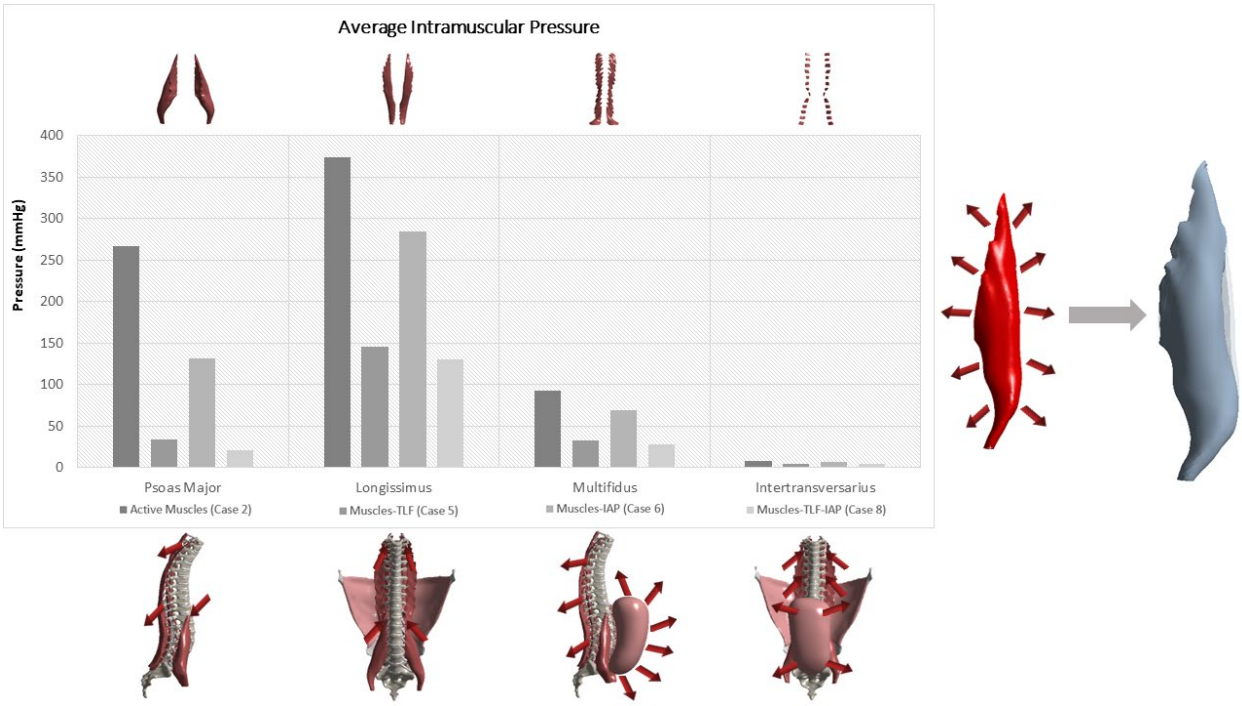


Figure 7:


Average Intramuscular Pressure
Pressure (mmHg)
400
350
300
250
200
150
100
50
0
Psoas Major
Longissimus
Multifidus
Intertransversarius
Active Muscles (Case 2)
Muscles-TLF (Case 5)
Muscles-IAP (Case 6)
Muscles-TLF-IAP (Case 8)