**TITLE:**

**The use of a robotic spine in the preliminary study to evaluate the correlation between intra-abdominal pressure and spinal stability**

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**Abstract:**

Lower back pain (LBP) places a large economic burden on health and welfare systems and is one of the leading causes of inability to work, affecting 70-85% of individuals at some point in their lifetime. However, the etiology and pathomechanism of back pain are often unknown. This technical brief evaluated the correlation between intra-abdominal pressure (IAP) and spinal stability. Furthermore, it studied the effect that intramuscular pressure (IMP) in the paraspinal muscles could have on spinal stability, and, indirectly, if IMP could be related to LBP. For this preliminary study, a robotic spine model was designed with analog bones, pneumatic muscles, and thoracic and abdominal-pelvic cavities. Twelve pneumatic muscles were attached to the spine, representing the multifidus, the erector spinae, the psoas major, and the rectus abdominis. Then, a load was applied vertically on the spine. The IAP was varied and the spinal stability was evaluated. Next, by varying the pressure in the muscles required to maintain spinal stability, the required variation in IMP was analyzed. It was determined that there is a correlation between IAP and spinal stability. Moreover, the increase in IMP, subsequent to lower IAP, could be a potential cause for LBP.

*Keywords: lower back pain, spinal stability, intra-abdominal pressure, intramuscular pressure*

**Introduction:**

Lower back pain (LBP) is one of the leading causes of inability to work and consultation of healthcare professionals [1, 2], placing massive economic burdens on health and welfare systems [2-5]. Numerous studies have observed that 70-85% of individuals will be affected by LBP and spinal disorders at least once in their lifetime [5-8]. However, the etiology of many episodes of back pain is not always apparent [1, 3, 4, 6, 9-13]. Due to the significant impact that spinal disorders and dysfunctions place on individuals and on society, it is crucial to find solutions to improve our understanding of the underlying biomechanical principles and correspondingly improve or develop novel treatments – hopefully helping reduce the need for work absences and medical consultations.

There are numerous possible causes of back pain, including heavy or improper lifting [1, 2, 10, 14-16], prolonged walking [15], sitting [1, 2, 7], or standing [7, 15], high body mass index [1, 17-20], smoking or chronic coughing [10, 16, 19], or lower levels of activity [2, 14, 17, 21], among others. Weakened trunk muscles and imbalanced muscle strength are also correlated with back pain [22-25]. Furthermore, muscle fatigue is a potential cause of LBP since long lasting contractions and high intramuscular pressure (IMP) cause restriction of blood flow to the muscles, contributing to pain, risk of ischemia, and degenerative change [13, 25-29]. Long lasting contractions can also lead to increased spinal stiffness, which could cause higher spinal displacement in certain activities, such as balancing tasks, leading to an elevated risk of injury [25]. Numerous studies have attempted to find the cause of LBP, but in order to understand LBP, there must be an understanding of spinal stability and the mechanics of the spine.

There are several definitions of spinal stability and interpretations vary depending on the context. For the purpose of this study, spinal stability was defined similarly to the definition from Reeves et al. as the ability of the spine to remain within a given range of motion [25].

To the author’s knowledge, spine loaded bench side experiments have not been performed on a full spine model, inclusive of pressurized muscles and cavities, and have often only considered the lumbar spine. Past studies by Patwardhan et al. observed that the vertical compressive load sustainable by the lumbar spine was 80 to 120 N [30, 31]. The spines in these experiments buckled under a load much lower than what the human spine is capable of sustaining *in vivo*, valued at 1000 to 1200 N [30, 31]. When analyzed *in vivo*, the spine is subjected to what likely resembles a follower load: a load under pure segmental compression. Multiple studies have concluded that *ex vivo* experiments excluding back muscle forces will be unable to withstand equivalent loads of *in vivo* magnitude [30, 32, 33]. Previous studies on spinal models or cadavers often do not consider the effects that pressure in the paraspinal muscles or internal cavities has on stability. Therefore, the spinal models or cadavers are unable to withstand the compressive loads that are experienced by a human spine. Literature suggests that variation in intra-abdominal pressure (IAP) has a significant impact on spinal stability [34-36]. Therefore, both IMP and IAP are key factors to consider when studying back pain.

In this preliminary study, a compressive load was applied vertically on a bench side robotic spine. The purpose of this technical brief was to evaluate the correlation between spinal stability and IAP. By varying the IAP, the required change in IMP, which then conveys a force to the spine serving to restore its original position, in the paraspinal muscles to maintain spinal stability was determined. It was hypothesized that variation in IAP would influence IMP and spinal stability, and therefore provide further support towards it being a potential cause for LBP.

**Methods:**

Definition of spinal stability:

For this experiment, the range of stability used to determine normal sagittal balance was defined by the Scoliosis Research Society in which a plumb line was drawn vertically downwards from the center of the C7 vertebra. The plumb line should be ±2 cm (0.787 in) from the posterior superior aspect of the S1 vertebra to be considered stable [37].

Model Setup:

A bench side robotic spine model was designed, complete with analog bones, pneumatic muscles, and cavities, as depicted in Fig. 1. The analogue spine consisted of intravertebral discs, vertebrae, pelvic bones, and a rib cage. The cervical vertebrae were removed from the spine model, with the exception of C7, to allow for accurate sagittal spinal balance measurements as defined by the Scoliosis Research Society. The custom-made pneumatic muscles, modeled after McKibben air muscles, were assembled from machined end caps and valves, expandable sleeving, and latex tubing, allowing the muscles to contract when inflated to simulate human muscle contractions. The thoracic and abdominal-pelvic cavities were also custom designed from PVC material and inflated to resemble human internal cavities.



**ITP=52mmHg**

**IAP**

**[0,52]**

**mmHg**

**IMP**

**[1551,2586]**

**mmHg**

**W=50N**

***Plumb***

***line***

**UNSTABLE STABLE UNSTABLE**

***Range of motion***

**C7**

**S1**

Fig. 1: Experimental set up of robotic spine model. The given model is within the stable range, with the plumb line 2cm from the center line.

The pneumatic muscles were attached vertically surrounding the posterior and lateral sides of the spinal column. The muscles were grouped to represent a simplification of the major muscle groups that are central to maintaining spinal stability. In total, the pneumatic muscles represented on the spine were four erector spinae, four multifidi, two psoas major, and two rectus abdominis. The model was simplified by assuming that the other abdominal muscles were incorporated in the abdominal-pelvic cavity and that these muscles contributed to the model’s IAP.

Bleed Rate:

It should be noted that the bleed rates of the artificial muscles and cavities were re-evaluated after each test to ensure that the pressure did not differ between trials.

Muscle Strength:

The strength of the pneumatic muscles was evaluated. An artificial muscle was hung from one end and the muscle was inflated by increments of 517 mmHg (10psi) from 0 mmHg to 4137 mmHg (80 psi). Then, a 22 N (5 lb) load was hung from the bottom end of the muscle and elongation was evaluated. The experiment was repeated with muscles of varying lengths, from 4.5 inches to 11.5 inches, in order to analyze deformation under varying IMP.

Spinal Stability:

The thoracic and abdominal-pelvic cavities were inflated to 52 mmHg (1 psi) as the control value. Intrathoracic pressure (ITP) was kept constant throughout the experiments at 52 mmHg. The initial position of the spine model was observed. A 50N force was applied in vertical compression on the spine and the displacement was observed and recorded. The spine was determined to be stable, at the baseline, which required each muscle was inflated to an IMP of 1551 mmHg (30 psi).

The experiment was repeated with a decrease in IAP by increments of 13 mmHg (0.25 psi), resulting in a minimum and maximum IAP of 0 mmHg and 52 mmHg, respectively. The spinal displacement at each increment was recorded. Next, the IMP in the paraspinal and abdominal muscles was adjusted accordingly to maintain spinal stability.

It should be noted that numerous iterations were performed for each experiment to account for variation. These experiments were then repeated with five additional pneumatic muscles, three erector spinae and two psoas major, to determine if the results from the initial 12 pneumatic muscles were sufficient and to evaluate the sensitivity of the prior selected and explored muscles in the experimental model.

**Results: Serves to describe your results (not interpret the results). Just list them as if you were a thoughtless robot.**

Muscle strength:

Fig. 2: Pressure development in the pneumatic muscles as a function of deformation, with and without a 22 N load applied

Displayed in Fig. 2, the pneumatic muscles could not withstand an IAP greater than 3620 mmHg (70 psi) to 4137 mmHg, due to material strength. It was observed that longer muscles underwent larger change in deformation. Furthermore, when no load was applied, the muscles were observed to shorten in length through contraction. However, when the 22 N (5 lb) load was applied, the muscles had to compensate for the downward force, therefore contracting less.

Spinal stability:

Fig. 3: Comparison of the mean spinal displacement of 12 to 17 pneumatic muscles under a constant applied load of 50 N, constant IMP of 1551 mmHg, and varying IAP.

Figure 3 represents the change in spinal displacement when the IAP was varied from the stable baseline. As IAP was decreased from 52 mmHg to 0 mmHg, the displacement from the center line was increased until the spine was not able to remain in an upright position at an IAP of 0 mmHg, portrayed by the 10 cm displacement in Fig. 3. The pressure in the pneumatic muscles was kept constant at 1551 mmHg.

In addition, Fig. 3 represents a comparison between the initial 12 pneumatic muscles and 17 muscles. It can be observed that there is no significant difference between the number of muscles used.

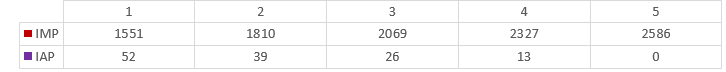


Fig. 4: Comparison of variation of IAP and required IMP to remain within the stable range (±2 cm from the center line) with 12 pneumatic muscles

Figure 4 displays the required IMP to maintain spinal stability under varying IAP. Test #1 in Fig. 4 represents the control values for the experiment, with IAP at 52 mmHg and IMP at 1551 mmHg. As IAP was decreased from the control value, IMP was increased to maintain stability, resulting in a maximum IMP of 2586 mmHg (50 psi) when IAP was 0 mmHg.

**Discussion:**

A novel analog spine model was successfully created that included muscles which provide contraction force and internal expansion pressure. Furthermore, the model included abdominal-pelvic and thoracic cavities. Each analog physiological element had their internal pressures controlled while different intra-abdominal pressures were imposed while muscle pressures, required to maintain spinal stability under a compressive load, was monitored. Globally, results from this preliminary study support the notion that IAP plays a role in spine stability.

In regard to the selection of which muscle groups to simulate, published research has indicated the significance of select muscle groups in spinal stability. Thus, the simplified muscle groups included in the bench side robotic spine model were the following: multifidus [12, 13, 23, 24, 29, 32, 38, 39], erector spinae [12, 13, 23, 32, 38, 40], rectus abdominis [12], and psoas major [12, 32]. Numerous iterations were performed on the robotic spine with an additional 5 pneumatic muscles. Results for spinal displacement did not vary significantly when comparing 12 or 17 pneumatic muscles activated on the spine, as seen in Fig. 3. Therefore, it was determined that the 12 initial muscles were sufficient for accurate results. Thus, this assumption was deemed acceptable.

Some limitations of the study include the compressive load applied, the strength of the cavities, and the pneumatic muscles. A 50 N load was applied in vertical compression on the spine. The spine model was not able to maintain stability without interference if loads greater than 50 N were applied. For the artificial cavities, the abdominal-pelvic and thoracic cavities could withstand a maximum pressure slightly above 103 mmHg (2 psi). The maximum contractile force in the human thoracic and abdominal-pelvic cavities is approximately 120 mmHg (2.3 psi) [41]. Due to the strength of the artificial cavities in the experiment, 52 mmHg IAP was defined as the control. This permitted variation in IAP from the control value that would avoid damage to the cavities. The muscles could withstand a maximum internal pressure of 3620 mmHg before failure. It was shown that the maximum contractile pressure found in human back muscles was in the multifidus muscles, valued at 375 mmHg (7.2 psi) [29]. The values for IMP on the spine model in the experiment were significantly higher than the values found from *in vivo* experiments. Potential causes for this increased muscle pressure include the weight difference between analog bones and human bones, the lower IAP in the model compared to a human abdomen, perhaps a disconnect between pressure and contraction force profiles, and/or the simplification of the pneumatic muscle groups. Future studies should consider the effect of other paraspinal and abdominal muscles, as well as the diaphragm, to determine stability with greater accuracy.

As seen in Fig. 3, spinal displacement is shown to be dependent on IAP. As the pressure in the abdominal cavity was decreased from the control, the spine became unstable as it collapsed forward. These results agree with Hodges et al.’s and O’Sullivan et al.’s studies showing that a variation in IAP can cause spinal instability [42-45]. Clinical studies by Hodges and Richardson, and Hodges et al. portrayed evidence that the reduction of the ability to contract and the delayed contraction of abdominal muscles is a cause of inefficient spinal stabilization from the abdomen [42-44]. Similarly, a clinical study by O’Sullivan et al. observed that subjects with LBP tend to have weaker abdominal muscles, resulting in the inability to maintain stability through abdominal contraction [45]. Common medical advice from practitioners recommends that patients perform exercise tasks to strengthen their abdominal region, therefore increasing stability and relieving back pain [21]. One’s ability to control their abdominal region, by limiting expansion, would in theory augment their IAP while a compression is being imposed.

Considering Fig. 4, it was observed that by keeping the spine in the stable range, IMP is dependent on IAP, among other variables. These results demonstrated that an increase in IMP was required to maintain spinal stability if there was a variation in IAP. The higher the IMP value the higher the simulated muscle contractile force is imposed on the spine which serves to restore it to what was pre-determined to be considered a stable configuration. In a clinical study, Konno et al. observed that IMP in the lumbar muscles was significantly higher in patients who were affected by spinal disorders or dysfunctions compared to individuals who had not experienced LBP [13]. Therefore, this increase in IMP could be a potential cause for LBP [13]. As discussed previously, in reality, increased IMP can cause higher or longer lasting muscle contractions. This can lead to a restriction of blood flow to the muscles or muscle fatigue and stiffness. These negative effects could be potential causes for LBP.

In conclusion, this preliminary study performed on a full thoracolumbar spine model established the importance of including paraspinal muscles and internal cavities when evaluating the effect of a pure compressive load applied on a spine. In addition, via the analogue spine model, it was demonstrated that there is a correlation between IAP and spinal stability. If an individual has lower IAP, then the paraspinal muscles may need to compensate by over contracting, therefore increasing the IMP required to maintain spinal stability. Consequently, increased IMP may result in LBP, further supporting the conclusion that variation in IAP may influence LBP.

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**Figures and Tables**



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**UNSTABLE STABLE UNSTABLE**

***Range of motion***

**C7**

**S1**

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Fig. 3: Comparison of the mean spinal displacement of 12 to 17 pneumatic muscles under a constant applied load of 50 N, constant IMP of 1551 mmHg, and varying IAP.

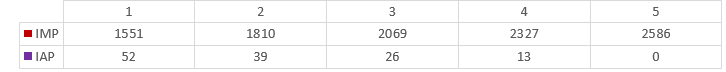


Fig. 4: Comparison of variation of IAP and required IMP to remain within the stable range (±2 cm from the center line) with 12 pneumatic muscles