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Lumbar Spine Postural Changes in Response to Operational Loads and Positions in Military Personnel

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Lumbar Spine Postural Changes in Response to Operational Loads and Positions in Military Personnel

A dissertation submitted in partial satisfaction of the requirements for the degree of Doctor of Philosophy

in

Bioengineering

by

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2015

The Dissertation of Ana Elvira Rodríguez Soto is approved, and it is acceptable in quality and form for publication on microfilm and electronically:

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2015

DEDICATION

A mi madre, por ser mi modelo a seguir, por tu fortaleza, esfuerzo, sacrificio y amor.

Celia, Arturo, Gisela, José y Daniel. Gracias por haberme inspirado a llegar hasta aquí y nunca ceder ante la adversidad. Los amo.

To Carlos, for always being by my side, for your unconditional love, cheering and support. Thank you for believing in me. I love you.

EPIGRAPH

`I didn't know that Cheshire cats always grinned;
in fact, I didn't know that cats *could* grin.'
`They all can,' said the Duchess; `and most of 'em do.'
`I don't know of any that do,' Alice said very politely,
feeling quite pleased to have got into a conversation.
`You don't know much,' said the Duchess; `and that's a fact.'

- Lewis Carroll, *Alice in Wonderland*

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LIST OF ABBREVIATIONS

BW	Body Weight
CoM	Center of Mass
CoP	Center of Pressure
CT	Computed Tomography
CV	Coefficient of Variation
ETL	Echo Train Length
FOV	Field of View
FSPGR	Fast-Recovery Fast Gradient Echo
FSU	Functional Spinal Unit
ILBE	Improved Load Bearing Equipment
IVD	Intervertebral Disc
LBP	Low Back Pain
LCS	Load Carriage System
LO2	Immediately after donning load, task
LO3	After 45 minutes of standing with load, task
LO4	After 45 minutes of walking with load, task
LS	Lumbar Spine
MRI	Magnetic Resonance Imaging
NEX	Number of Excitations
RMSE	Root Mean Square Error
ROI	Region of Interest
TE	Echo Time
TR	Repetition Time
UN1	Without load, task
UN5	After 45 minutes of side-lying recovery, task

LIST OF FIGURES AND TABLES

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Sato EJ, Killian ML, Choi AJ, Lin E, Choo AD, **Rodriguez-Soto AE**, Lim CT, Thomopoulos S, Galatz LM, Ward SR: Architectural and biochemical adaptations in skeletal muscle and bone following rotator cuff injury in a rat model. *J Bone Joint Surg Am* 2015, 97(7):565-573.

Rodriguez-Soto AE, Jaworski R, Jensen A, Niederberger B, Hargens AR, Frank LR, Kelly KR, Ward SR: Effect of load carriage on lumbar spine kinematics. *Spine* 2013, 38(13):E783-791.

Rodriguez AG, **Rodriguez-Soto AE**, Burghardt AJ, Berven S, Majumdar S, Lotz JC: Morphology of the human vertebral endplate. *J Orthop Res* 2012, 30(2):280-287.

Rodriguez AG, Slichter CK, Acosta FL, **Rodriguez-Soto AE**, Burghardt AJ, Majumdar S, Lotz JC: Human disc nucleus properties and vertebral endplate permeability. *Spine* 2011, 36(7):512-520.

Folkesson J, Goldenstein J, Carballido-Gamio J, Kazakia G, Burghardt AJ, **Rodriguez A**, Krug R, de Papp AE, Link TM, Majumdar S: Longitudinal evaluation of the effects of alendronate on MRI bone microarchitecture in postmenopausal osteopenic women. *Bone* 2011, 48(3):611-621.

Rodriguez-Soto AE, Fritscher KD, Schuler B, Issever AS, Roth T, Kamelger F, Kammerlander C, Blauth M, Schubert R, Link TM: Texture analysis, bone mineral density, and cortical thickness of the proximal femur: fracture risk prediction. *J Comput Assist Tomogr* 2010, 34(6):949-957.

Selected Abstracts

Rodriguez-Soto AE, Berry DB, Valaik E, Palombo L, Kelly K, Ward SR. The effect of load magnitude and distribution on lumbar spine posture of active-duty Marines. Annual Meeting of the American Society of Biomechanics (2015); Columbus, OH.

Rodriguez-Soto AE, Stambaugh J, Su J, Berry DB, Gombatto S, Palombo L, Kelly K, Ward SR. Spinal muscle quality changes in physically active individuals with disc degeneration. Annual Meeting of the Orthopaedic Research Society (2015); Las Vegas, NV.

Berry DB, **Rodriguez-Soto AE**, Tokunaga JR, Gombatto SP, Ward SR. An endplate-based joint coordinate system for measuring kinematics in normal and abnormally shaped lumbar vertebrae. Annual Meeting of the Orthopaedic Research Society (2015); Las Vegas, NV .

Berry DB, **Rodriguez-Soto AE**, Gombatto SP, Jaworski R, Kelly KR, Ward SR. Lumbar spine postures in Marines during simulated operational conditions. World Congress of Biomechanics (2014); Boston, MA.

Rodríguez-Soto, AE, Jensen A, Jaworski R, Frank LR, Kelly KR, Ward SR. Load carriage lumbar spine kinematics in Marines. Annual Meeting of the American Society for Pharmacology and Experimental Therapeutics (2012); San Diego, CA.

Rodríguez-Soto AE, Jaworski R, Jensen A, Niederberger B, Hargens AR, Frank LR, Kelly KR, Ward SR. Lumbar spine kinematics in Marines carrying heavy loads. Annual Meeting of the American College of Sports Medicine (2012); San Francisco, CA.

Rodríguez-Soto AE, Rodriguez AG, Burghardt AJ, Majumdar S, Lotz J. Automatic segmentation of the vertebral endplate VOI from μCT images. Annual Meeting of the American Society for Bone and Mineral Research (2009); Denver, CO.

Rodriguez AG, Slichter C, **Rodríguez-Soto AE**, Buser Z, Majumdar S, Lotz Jc. Human intervertebral disc endplate permeability and its relation to cell density and degeneration. Annual Meeting of the Orthopaedic Research Society (2009); Las Vegas, NV.

Rodriguez-Soto AE, Carballido-Gamio J, Link TM, Majumdar S. Gradient-field based MRI knee cartilage segmentation algorithm with self correction. Annual Meeting and Exhibition of the International Society for Magnetic Resonance in Medicine (2008); Toronto, ON, Canada.

ABSTRACT OF THE DISSERTATION

Lumbar Spine Postural Changes in Response to
Operational Loads and Positions in Military Personnel

by

Ana Elvira Rodriguez Soto

Doctor of Philosophy in Bioengineering

University of California, San Diego, 2015

Professor Samuel Ward, Chair
Professor Robert Sah, Co-Chair

Lumbar spine (LS) problems are the number one cause for medical encounters and lost work time among military personnel. This has been associated with the heavy loads carried during operational duties. Consequently, the effect of load carriage on military *performance* has been studied extensively. However, LS *postural adaptations* during load carriage have not been described. Thus, the overall goal of this dissertation is to understand *in vivo* LS postural response to load carriage in active duty Marines.

This dissertation begins with a feasibility study to demonstrate that LS postural differences between unloaded and loaded tasks are measurable *in vivo* using an upright magnetic resonance imaging (MRI) scanner. Increased lumbar flexion and reduced lordosis were measured during loaded tasks. In Chapter 3 we examined LS response to load carriage as Marines progressed through the School of Infantry (SOI) Training. Although no effect of training was observed, we found that Marines with disc degeneration carried loads slightly different.

These results suggested that load characteristics determined load carriage posture of the LS, regardless of Marines physical fitness. Therefore, in Chapter 4 we explored the interaction between operationally relevant load magnitudes and distributions on LS posture. Specifically, balancing loads in the anterior-posterior direction yielded lumbar spine postures that are nearly identical to standing. Contrastingly, when load was carried with a posterior bias, superior LS lordosis increased as a function of load magnitude, while inferior LS lordosis was distribution dependent.

These data represent the first *in vivo* measurements of LS posture under operationally relevant loads. The key findings are: 1) lumbar lordosis is reduced as the spine and hips are flexed forward, 2) improving strength and endurance through training does not appear to change LS behavior during load carriage, and 3) manipulating the load magnitude and distribution does yield significant changes in LS behavior.

Clinically and operationally, these findings are significant because they provide direct, quantitative data that can be used to make informed decisions about load carriage recommendations. Future work may include a longitudinal study to test whether balanced loads prevent injury or enhance operational efficiency in the military.

CHAPTER 1 : INTRODUCTION

1.1 Heavy Load Carriage and the Armed Forces

Spine problems are the number one cause for medical encounters and lost work time among the musculoskeletal diseases that affect the military population.[1] Approximately 67% and 28% of spine-related ambulatory visits and hospitalizations were attributed to non-specific low back pain (LBP), respectively.[1] Furthermore, during Operation Iraqi Freedom and Operation Enduring Freedom, LBP was the cause of approximately 7% of all medical evacuations, which has also been associated with a low probability of return to duty and impeding training completion.[2, 3]

Consequently, several efforts have been made to identify factors that may increase the risk of developing LBP in the deployed armed forces.[4-7] Interestingly, the estimated risk of LBP is different among the branches of the US Armed Forces, increasing in the following order: Marines, Army, Air Force and Navy.[6] Other factors that have been previously associated with the development of LBP are: older age, female gender, smoking, high BMI, low fitness level, poor aerobic endurance, carrying heavy loads, wearing body armor for long periods of time, and psychosocial distress.[4-7] To note, in a study by Roy *et al.*, 77% of a group of US Army soldiers reported at least one episode of LBP during a one-year deployment and ~40% of them reported carrying heavy loads as the activity that triggered the pain episode.[4]

There are three kinds of combat loads depending on the nature of a mission: fighting, approach march, and emergency approach march loads.[8] A fighting load includes essential items for environmental and threat protection in the case of hand-to-hand combat and operations requiring stealth. Fighting loads should be kept at a minimum to allow for soldiers to move and fight effectively. An approach march load is similar to a fighting load, however it is used in more prolonged operations and therefore, it involves extra gear to allow soldiers to fight and subsist until resupply. Finally, emergency approach march loads are those carried under circumstances when terrain is not accessible to any transportation resources, which results in soldiers carrying the necessary gear, food and water for longer periods of time. Currently, the US Armed Forces recommend load limits of 22kg, 33kg and 55kg for fighting, approach march, and emergency approach march loads, respectively.[8] Average loads carried by infantry soldiers during dismounted operations in Afghanistan were 29kg, 43kg and 58kg for fighting, approach march and emergency approach march loads, respectively.[9, 10] Although these loads are well in excess of the load carriage limit recommendations, members of the Armed Forces must keep the necessary equipment to fulfill their mission readily available.

As mentioned above, efforts to improve physical conditioning of the Armed Forces include the development of efficient physical training programs. Significant improvements in speed and endurance during high intensity marches, while carrying heavy loads, suggest that a combination of aerobic and resistance exercises are essential to military training.[11-14] As a result, an important element of basic military training consists of loaded marches, during which march distances and carried loads are simultaneously increased as training progresses.[15] This element of training is aimed specifically at improving the physical condition of the members of the Armed Forces to carry heavy loads.[15] However, the effect of military training on load carriage biomechanics remains unknown.

Multiple efforts have been made to maximize load carriage performance and reduce the risk of injury. These include the development of specialized load-carrying equipment and of physical training aimed to condition soldiers for load carriage.[16] In order to develop improved load-carrying equipment, the effect of both load magnitude and distribution on gait biomechanics has been previously studied. These data are commonly recorded using motion caption and photography analysis technologies, which allow measuring gross body posture. However, in the case of the LS, the accuracy of regional and lumbar level dependent measurements is limited as markers are commonly blocked by the load carriage system (LCS).[17] Known biomechanical adaptations in response to carrying load with a posterior bias (backpack) include increased trunk and hip flexion, as well as increased knee range of motion as load magnitude increases.[18] In terms of load distribution, it has been shown that the only postural difference between carrying load with a posterior bias and in a more evenly distribution anteriorly and posteriorly is the position of trunk. When load is carried more evenly distributed, trunk and hip flexion are reduced to values similar to those of standing without external load.[19] Therefore, one can hypothesize that the LS, pelvis and hip joints are the main contributors to the postural differences between load distribution configurations. However, data describing detailed orientation of these joints as a function of both load carriage and distribution are not available. Understanding the individual contribution of these joints to the overall body posture will allow identifying structures that may be under higher risk of injury.

Although the general biomechanical behavior of the human body during load carriage is somewhat understood, there are no data available describing how load carriage changes regional and level-dependent LS posture. These data would help elucidate potential mechanisms of injury and to improve our understanding of the association between heavy load carriage and the development of LBP. In addition, it would allow generating load magnitude, distribution and training recommendations for the benefit of both military and civilian populations.

1.2 Spine Anatomy

Vertebral column provides both stability and flexibility between the skull and the pelvis, and is also the origin for muscles of the spine, trunk and limbs.[20] It is comprised of five regions: cervical, thoracic, lumbar, sacral and coccygeal (Fig. 1−1). The natural curvature of the spine is comprised of two lordoses (anterior curves) in the cervical and lumbar spines, and two kyphoses (backward curves) in the thoracic and sacral spines. These curves distribute body weight and axial loads efficiently during movement.[21]

The spine has 33 individual vertebral bodies or vertebrae named according to their position in the vertebral column: 7 cervical, 12 thoracic, 5 lumbar, 5 sacral and 4 coccygeal (Fig. 1−1). Importantly, only the vertebrae of the cervical, thoracic and lumbar regions form articulating joints, while the remaining nine are fused.[20] The articulations of the vertebral column are: one fibrous joint, which is formed by two contiguous vertebral bodies and the interposed intervertebral disc (IVD), and two zygapophyseal joints (facet joints). This structural arrangement is referred to as functional spinal unit (FSU, Fig. 1−2).[22] Vertebrae of different regions of the spine also vary in shape, therefore limiting motions differently along its length. Regardless of the region in the spine, all articulated vertebrae consist of two different parts: the solid anterior body and the posterior arch, which encapsulates the spinal cord (Fig. 1−3).[20]

The vertebral bodies are the largest part of the vertebra and in healthy individuals are flattened superiorly and inferiorly. The central region of the superior and inferior surfaces is the vertebral endplate.[20] Cortical bone of the endplates is porous and covered by a thin layer of hyaline cartilage.[23] This structure rigidly joins the IVD and vertebral bodies.[24] Intervertebral discs are the largest avascular structure in the body; its nutrient supply is partially provided by a large number of capillaries located in the endplate. Nutrients move mainly by diffusion from the capillaries through the cartilage endplate and into the core of the IVD.[23] Each IVD is formed by two compartments: nucleus pulposus and annulus fibrosus (Fig. 1−4). The nucleus pulposus is the core of the IVD, which consists of a proteoglycan-rich hydrated gel sustained by a matrix of type II collagen. This environment confers a negative fixed charge in the nucleus pulposus, resulting in a large affinity for water and, therefore, increased osmotic pressure. This pressure enables the IVD to resist compressive loads.[23] The annulus fibrosus is formed by a series of 10-20 concentric sheets of type I collagen called lamellae. The orientation of the collagen fibers is approximately 60° from the horizontal, and the directionality is alternated between contiguous lamellae (Fig. 1−4).[25] Intervertebral discs act as shock absorbers during everyday activities, while allowing for the vertebral column to bend. In a healthy IVD, compressive forces increase the internal pressure of the nucleus pulposus, which causes stretching of the annulus fibrosus. This mechanism transforms compressive forces into tensile (shear) forces resisted by the lamellae of the annulus (Fig. 1−4).

The posterior vertebral arch neural arch is formed by the following bone structures: pedicles, laminae and a set of processes. The pedicles are two horizontal oval columns of bone that protrude posteriorly from each side of the vertebral body (Fig. 1−3).[20] The laminae are plates of bone that extend from each pedicle and fuse in the mid-sagittal line. These structures enclose to form the spinal foramen, which protects the spinal cord. The spinous process is a sheet of bone that protrudes posteriorly from the laminae junction. From the region where the pedicles join the laminae emerge three processes on each side: two articular processes, one that projects upward and one that projects downward, and one transverse that projects sideways. Spinous and transverse processes serve as attachment sites for trunk and back muscles and ligaments. The superior and inferior articular processes of contiguous vertebrae have a layer of cartilage and become in contact with each other, forming the zygapophyseal or facet joints.[20] These joints provide stability to the vertebral column by restricting axial rotation and forward translation of the vertebrae.[26] In a FSU, the inferior articular process of the superior vertebrae is medial to the superior articular process of the inferior vertebrae. The orientation of the lumbar facet joints is 25°-50° from the sagittal plane (Fig. 1−5), while its shape may be flat or curved.[21] Consequently, when the superior vertebra of the FSU attempts to rotate in the axial plane or translate forward the articular process of the inferior vertebrae resists these motions. Facet joints, however, are able to slide past each other in the superior-inferior direction, allowing for flexion/extension motion of the spine. Stabilization of spine joints is provided by active and passive structures such muscles, fascia, tendons and ligaments.[26]

The lumbar spine is under constant compressive and shear forces during everyday activities as it carries the trunk, arms and head weight. As mentioned above, contiguous vertebrae are joined together through the IVD and facet joints. The biomechanical consequence of this anatomical configuration is that both IVD and facet joints are load bearing structures. When a compressive load is applied, the fraction of the load that is resisted by each element varies as a function of the degrees of flexion/extension in the FSU. In extension, the facet joints resist most of the load, while during flexion the IVD and other passive structures bear the majority of the compressive forces.[27]

The balance of compression and shear forces between IVD and facet joints is relevant in the context of military load carriage as repetitive loading of these elements may result in LBP. For example, due to excessive compression loading of the IVD, vertebral endplates and adjacent trabeculae are sometimes damaged. Similarly, repetitive loading may cause facet joint cartilage breakdown and disruption of the joint capsule. These structural disruptions frequently result in an inflammation response, which may irritate innervated musculoskeletal components and cause acute LBP.[26] It has been previously hypothesized that repeated insults of this nature may result in development of chronic pain.[26]

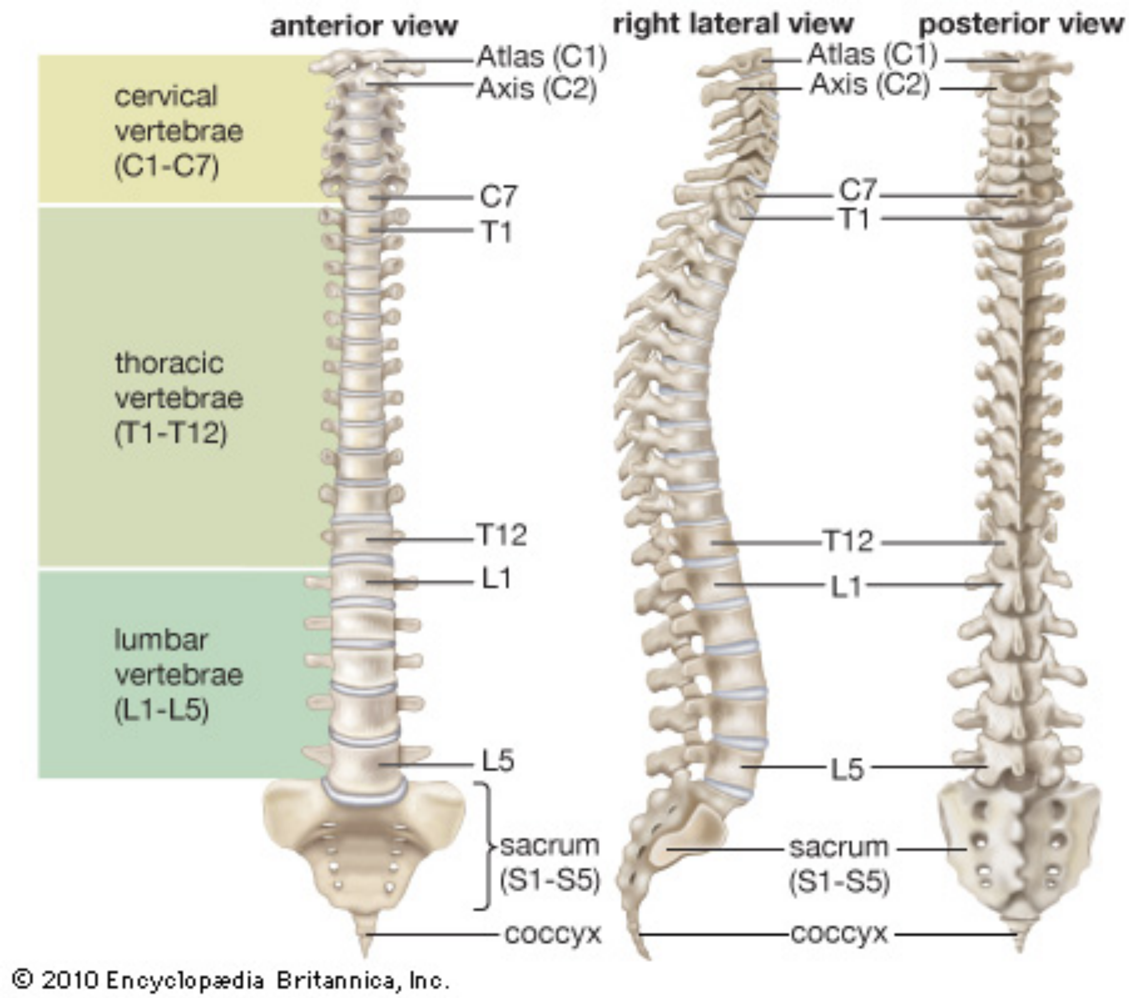

Figure 1−1 Anterior (left), lateral (center) and posterior (right) view of the vertebral column. It is divided in five regions: cervical, thoracic, lumbar, sacral and coccygeal. Only cervical, thoracic and lumbar regions form articulating joints. [28]

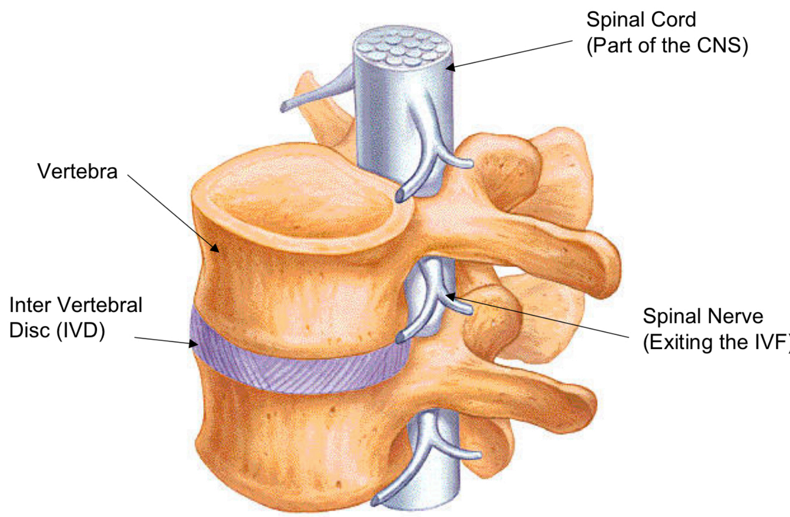

Figure 1−2 Functional spinal unit is the motion unit of the spine. It is composed of two contiguous vertebral bodies and the intervertebral disc between them. [29]

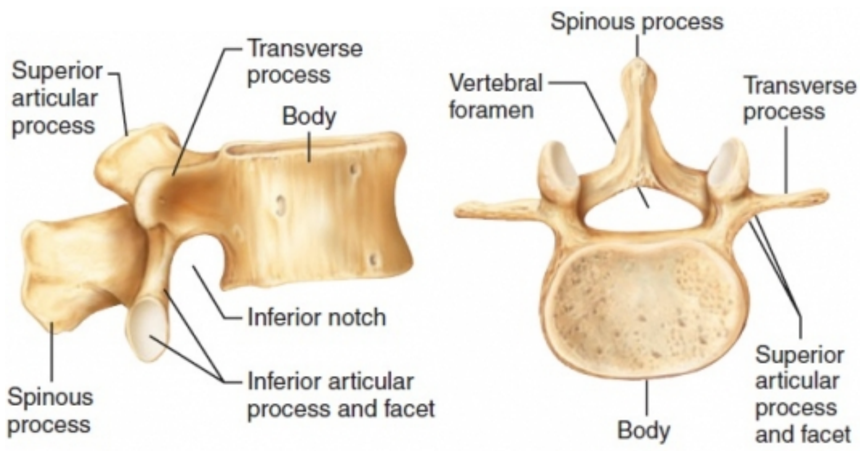

Figure 1−3 Anatomy of lumbar vertebrae.[30]

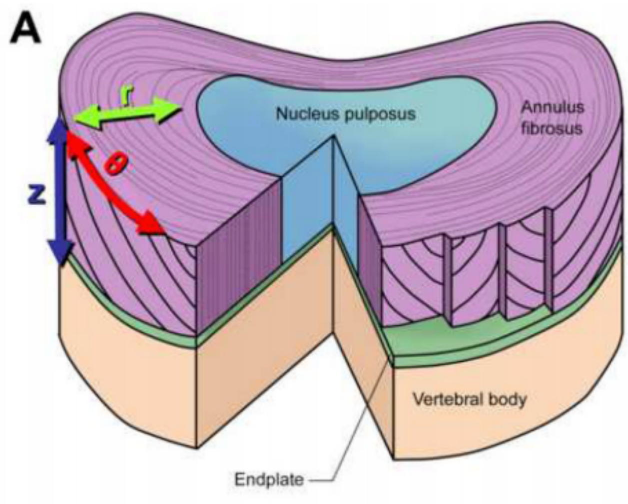

Figure 1−4 Schematic representation of the intervertebral disc structure: nucleus pulposus and annulus fibrosus.[25]

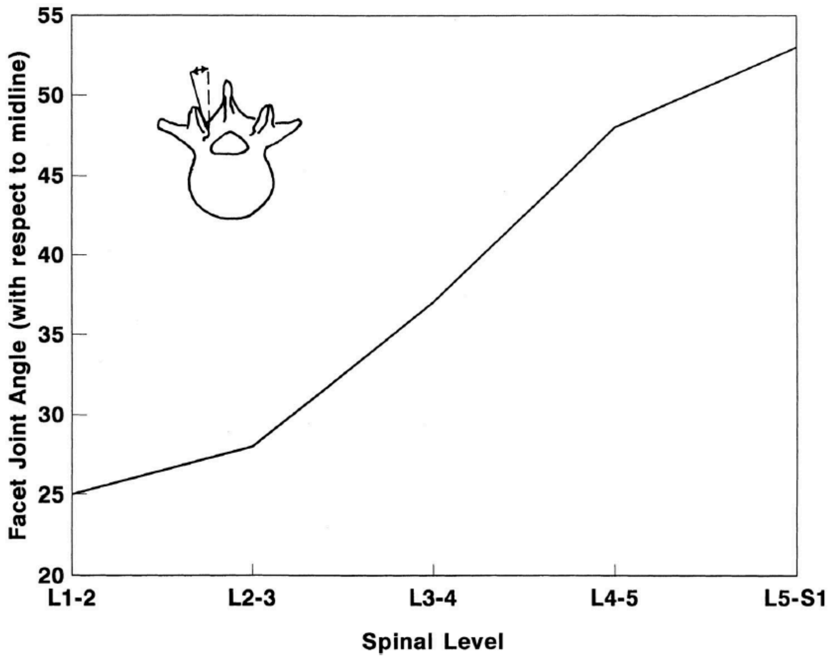

Figure 1−5 Facet joint orientation changes throughout lumbar levels. Data shown with respect to the midline.[21]

1.3 References

1. Army Medical Surveillance Activity. **Low Back Pain, Active Component, U.S. Armed Forces, 2000-2009**. MSMR. 2010;17(6):2-7.

2. Cohen SP, Brown C, Kurihara C, Plunkett A, Nguyen C, Strassels SA: **Diagnoses and factors associated with medical evacuation and return to duty among nonmilitary personnel participating in military operations in Iraq and Afghanistan**. *CMAJ : Canadian Medical Association journal = journal de l'Association medicale canadienne* 2011, **183**(5):E289-295.

3. O'Connor FG, Marlowe SS: **Low back pain in military basic trainees. A pilot study**. *Spine* 1993, **18**(10):1351-1354.

4. Roy TC, Lopez HP, Piva SR: **Loads worn by soldiers predict episodes of low back pain during deployment to Afghanistan**. *Spine* 2013, **38**(15):1310-1317.

5. Cohen SP, Gallagher RM, Davis SA, Griffith SR, Carragee EJ: **Spine-area pain in military personnel: a review of epidemiology, etiology, diagnosis, and treatment**. *The spine journal : official journal of the North American Spine Society* 2012, **12**(9):833-842.

6. Knox J, Orchowski J, Scher DL, Owens BD, Burks R, Belmont PJ: **The incidence of low back pain in active duty United States military service members**. *Spine* 2011, **36**(18):1492-1500.

7. Anderson MK, Grier T, Canham-Chervak M, Bushman TT, Jones BH: **Occupation and other risk factors for injury among enlisted U.S. Army Soldiers**. *Public health* 2015, **129**(5):531-538.

8. Field Manual 21-18, **Foot Marches**. Washington, DC: Department of the Army; 1990.

9. Friedl K, Santee WR: **Military quantitative physiology: problems and concepts in military operational medicine**. Fort Detrick, MD, USA; Borden Institute 2012.

10. Dean CD, FJ: **The Modern Warrior's Combat Load**. Fort Leavenworth, KS, USA: Center for Army Lessons Learned Report 2008.

11. Swain DP, Onate JA, Ringleb SI, Naik DN, DeMaio M: **Effects of training on physical performance wearing personal protective equipment**. *Military medicine* 2010, **175**(9):664-670.

12. Vickers Jr RB, AC;: **Effects of Physical Training in Military Pupolations: A Meta-Analytic Summary**. San Diego, CA, USA: Naval Health Research Center 2010.

13. Kraemer WV JA, Patton JF, Dziados JE, Reynolds KL: **The Effects of Various Physical Training Programs on Short Duration, High Intensity Load Bearing Performance and the Army Physical Fitness Test**. Natick, MA, USA: US Army Research Institute of Environmental Medicine 1987.

14. Mello RD AI, Reynolds KL, Witt CE, Vogel JA: **The Physiological Determinants of Load Bearing Performance at Different March Distances**. Natick, MA, USA: US Army Research Institute of Environmental Medicine 1988.

15. Field Manual 21-20, **Physical Fitness Training**. Washington, DC, USA: Department of the Army 1992.

16. US Army Development and Employment Agency (ADEA): **Report on the ADEA Soldier's Load Initiative**. Ft Lewis, WA, USA; 1987.

17. Smith B, Ashton KM, Bohl D, Clark RC, Metheny JB, Klassen S: **Influence of carrying a backpack on pelvic tilt, rotation, and obliquity in female college students**. *Gait & posture* 2006, **23**(3):263-267.

18. Harman EH, Han KH, Frykman P, Pandorf C: **The Effects of Backpack Weight on the Biomechanics of Load Carriage**. Natick, MA, USA: Military Performance Division of the US Army Research Institute of Environmental Medicine 2000.

19. Harman EF, Frykman P, Knapik JJ, Han KH: **Backpack vs. front-back: differential effects of load on walking posture**. *Med Sci Sports Exerc* 1994, **26**(S140).

20. Gray H, Clemente CD: **Anatomy of the human body**, 30th American Edn. Philadelphia: Lea & Febiger; 1985.

21. Benzel EC: **Biomechanics of spine stabilization**, Third edition. edn.

22. Mow VC, Huiskes R: **Basic orthopaedic biomechanics & mechano-biology**, 3rd edn. Philadelphia, PA: Lippincott Williams & Wilkins; 2005.

23. Huang YC, Urban JP, Luk KD: **Intervertebral disc regeneration: do nutrients lead the way?** *Nature reviews Rheumatology* 2014, **10**(9):561-566.

24. Wade KR, Robertson PA, Broom ND: **A fresh look at the nucleus-endplate region: new evidence for significant structural integration**. *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 2011, **20**(8):1225-1232.

25. Nerurkar NL, Elliott DM, Mauck RL: **Mechanical design criteria for intervertebral disc tissue engineering**. *Journal of biomechanics* 2010, **43**(6):1017-1030.

26. Adams MA: **The biomechanics of back pain**, 3rd edn. Edinburgh; New York: Churchill Livingstone/Elsevier; 2013.

27. Adams MA, Dolan P, Hutton WC: **The lumbar spine in backward bending**. *Spine* 1988, **13**(9):1019-1026.

28. Encyclopaedia Britannica Inc.: **Encyclopædia Britannica online**. Chicago: Encyclopædia Britannica,; 2015.

29. SnipView. **Functional Spinal Unit.** Accessed: August 2015. http://www.snipview.com/q/Functional_spinal_unit. 2015.

30. StudyDroid. **Lumbar Vertebra Anatomy.** Accessed: August 2015. http://studydroid.com/imageCards/0l/pg/card-22856257-back.jpg. 2012.

CHAPTER 2: EFFECT OF LOAD CARRIAGE ON LUMBAR SPINE KINEMATICS

2.1 Abstract

This was a feasibility study on the acquisition of lumbar spine kinematic data from upright magnetic resonance images obtained under heavy load carrying conditions. The objective of this study was to characterize the effect of the load on spinal kinematics of active Marines under typical load carrying conditions from a macroscopic and lumbar-level approach in active-duty US Marines. Military personnel carry heavy loads of up to 68kg depending on duty position and nature of the mission or training; these loads are in excess of the recommended assault loads. Performance and injury associated with load carriage have been studied; however, knowledge of lumbar spine kinematic changes is still not incorporated into training. These data would provide guidance for setting load and duration limits and a tool to investigate the potential contribution of heavy load carrying on lumbar spine pathologies. Sagittal T2 magnetic resonance images of the lumbar spine were acquired on a 0.6T upright magnetic resonance imaging scanner for 10 active-duty Marines. Each Marine was scanned without load (UN1), immediately after donning load (LO2), after 45 min of standing (LO3) and walking (LO4) with load, and after 45 min of side-lying recovery (UN5). Custom-made software was used to measure whole spine angles, intervertebral angles, and regional disc heights (L1−S1). Repeated measurements analysis of variance and *post hoc* Sidak tests were used to identify significant differences between tasks ($\alpha=0.05$). The position of the spine was significantly ($p<0.0001$) more horizontal relative to the external reference frame and

lordosis was reduced during all tasks with load. Superior levels became more lordotic, whereas inferior levels became more kyphotic. Heavy load induced lumbar spine flexion and only anterior disc and posterior intervertebral disc height changes were observed. All kinematic variables returned to baseline levels after 45 minutes of side-lying recovery. Superior and inferior lumbar levels showed different kinematic behaviors under heavy load carrying conditions. These findings suggest a postural, lumbar flexion strategy aimed at centralizing a heavy posterior load over the base of support.

2.2 Introduction

Military personnel carry heavy loads of up to 68 kg (149.9lbs) depending on duty position and nature of the mission.[1] For example, fighting loads range between 24 and 37kg (52.9 and 81.5lbs), while approach march loads are carried in more prolonged operations and range from 50-67kg (110.2 and 147.7lbs).[2] These loads are well in excess from the recommendation of load limits of 22kg (or 30% of body weight) and 33kg (or 45% of body weight) for fighting and march loads respectively.[3] Load limits have been extensively studied in terms of optimum energy expenditure,[4, 5] situational awareness, and responsiveness[6] resulting in several load carriage system (LCS) configurations.[7] In general, the LCS backpack configuration is preferred among the military due to the proximity of the load to the center of mass of the system compared to other LCS configurations.[8, 9] Despite these efforts, heavy load carriage in the military population has been associated to low back pain (LBP).[10-12]

The kinematic behavior of the lumbar spine while carrying load using a backpack configuration has been previously studied in both in civilian and military populations.[7, 13-18] The majority of these studies have used optical markers[14-16, 18] and ground force plates[7, 18, 19] to measure body positioning and ground reaction forces. These methods approach the lumbar spine from a macroscopic perspective, that is, as a unit that joins the upper and lower bodies. In order to investigate the lumbar spine in greater detail, other studies have used noninvasive imaging methods such as magnetic resonance imaging (MRI)[20, 21], computed tomography (CT)[22] and myelography.[23] In these techniques the subject is lying in supine position as images are acquired, consequently, researchers have attempted to simulate upright and other functional positions by applying axial load to the subject with different devices[20, 21]. However, it has been shown that the kinematic data obtained in this setting do not reflect the state of the lumbar spine in the upright position due to alterations in bone-muscle interactions, spine length and curvature, spinal canal cross-sectional area and regional intervertebral disc (IVD) heights.[22, 24, 25] To overcome this situation and allow imaging of the body in realistic functional positions technologies such as upright MRI[25] and dual-plane fluoroscopy[26] were developed. Measurements generated from these technologies take into account gravitational and weight-bearing effects, producing a more accurate description of the kinematic state of the spine in real life situations.[27] Up until now, kinematic variables such as spine curvature, lumbar spine lordosis, and IVD compressibility have been measured in both young[17] and adult[13] populations using upright MRI. However, these data cannot be applied to the military population for three main reasons. First, the magnitude of the carried loads is small compared to those in a military context; in most studies load ranges between 15 and 30% of BW,[13, 17] while soldiers carry up to 80% of BW.[2] Second, the length of exposure to load in these studies is in the range of a few minutes, whereas military marches extend over several hours.[28] Finally, a fundamental aspect of a soldier's basic training consists of progressively increasing the amount of load and march distance.[3, 28] This training has been shown to have an impact on the endurance and performance of soldiers, therefore the kinematic behavior data cannot be compared to that of the civilian population.[16, 29] Demonstrating that kinematic data of the lumbar spine can be obtained under heavy load carrying conditions would provide guidance for setting load and duration limits and provide a research tool to investigate the potential contribution of heavy load carrying to LBP.

The purpose of this study was aimed to measure the kinematic behavior of the lumbar spine from both a regional and local (level-dependent) approach in active-duty US Marines while carrying heavy load. This study also investigated the length of exposure and activity that induced significant changes in the kinematic behavior of the overall lumbar spine and functional spinal units. We hypothesized that IVD compression and lumbar lordosis increased with load and time of exposure through the lumbar spine.

2.3 Materials and Methods

Subjects

Ten male Marines from the Marine Corps Base Camp Pendleton, with no history of lower back issues volunteered to participate in this study. The University of California, San Diego (UCSD) and Naval Health Research Center (NHRC) Institutional Review Boards approved this pilot study, and all volunteers gave oral and written informed consent.

Imaging

Marines were scanned using an 0.6T MRI scanner (Upright MRI, Fonar Corporation, Melville, NY, USA) and a planar RF coil. A brace was used to place the coil behind the volunteers' back at the lumbar spine (L1-S1) level while standing (Fig. 2-1A). The brace was tight enough to keep the coil in place and avoid as possible any alteration of the volunteer's natural standing position. A three-plane localizer and sagittal T2 weighted images (TR=610mS, TE=17mS, FOV=24cm, 210x210 acquisition matrix, 4mm slice thickness, no gap, scan duration 2min) were acquired.

Load carrying tasks

With the purpose of measuring kinematic changes in the lumbar spine under load-carrying conditions, Marines performed a series of tasks with load and were scanned at different time points in one session. Each Marine was scanned a total of five times in the following order: standing without load (UN1), immediately after donning load (LO2), after 45 min of standing with load (LO3), after 45 min of walking on a treadmill at 3mph with load (LO4) and (5) after a recovery period of 45 min side-lying (UN5), half the time on each side (Fig. 2-1). The total carrying load was 112lbs (50.8kg) including body armor and an Improved Load Bearing Equipment (ILBE) backpack loaded with tile. During standing tasks Marines were instructed not to lean on surfaces (i.e. walls and chairs), but moving around the scanner console room was permitted. It was not indicated to our volunteers how to stand during scans to avoid the alteration of their natural standing position.

Data analyses

A set of points was manually placed at the corners of the each vertebra $\mathcal{L}$ on the images acquired in the upright MR scanner using the region of interest (ROI) point tool available in Osirix (Fig. 2-2).[30]

The location of the seed points were used to fit planes to the inferior and superior endplates of each vertebra $\mathcal{L}$. In order to compare sagittal measurements between Marines, due to the differences in standing positions, Procrustes analysis was used to find the rotation matrix R between the inferior and superior vertebrae contiguous vertebrae with the inferior vertebra as reference. The rotation matrix R follows the x-y-z convention defined as:

$$R = \begin{bmatrix} R_{11} & R_{12} & R_{13} \\ R_{21} & R_{22} & R_{23} \\ R_{31} & R_{32} & R_{33} \end{bmatrix} = \begin{bmatrix} cos\theta cos\phi & sin\psi sin\theta cos\phi - cos\psi sin\phi & cos\psi sin\theta cos\phi + sin\psi sin\phi \\ cos\theta sin\phi & sin\psi sin\theta sin\phi + cos\psi cos\phi & cos\psi sin\theta sin\phi - sin\psi sin\phi \\ -sin\theta & sin\psi cos\theta & cos\psi cos\theta \end{bmatrix}$$

where ψ, θ and ϕ are the rotations in radians around the x, y and z axes respectively. These are the Euler angles that describe the orientation of a vertebra (L1-L5) with respect to the inferior contiguous vertebra (L2-S1). We solved for the Euler angles θ, ψ and ϕ from the rotation matrix R and obtained:

$$\theta = \sin^{-1} R_{31} \qquad \psi = \operatorname{atan2}\left(\frac{R_{32}}{\cos\theta}, \frac{R_{33}}{\cos\theta}\right) \qquad \phi = \operatorname{atan2}\left(\frac{R_{21}}{\cos\theta}, \frac{R_{11}}{\cos\theta}\right)$$

The rotations around the x and z axes were removed using the rotation matrix R with Euler angles $-\psi$, $-\phi$ and $\theta = 0$, maintaining the information in the sagittal plane.

Measurements

All variables were measured from the 3D geometric representation of the lumbar spine aligned on the sagittal plane. The geometric centroid $C_{\mathcal{L}}(x, y, z)$ of a set of points $r_{\mathcal{L}}(x, y, z)$ that belong to the vertebra $\mathcal{L}$ is defined by $C_{\mathcal{L}} = \frac{1}{n}\sum_{k=1}^{n} r_k$. The angle with respect to the horizontal is that formed by the line traced between the geometric centroids of L1 and S1, and the horizontal (Fig. 2-3A). It indicates the overall position of the lumbar spine with respect to the ground however, it does not convey information about the lumbar spine kinematics.

The Cobb angle is extensively used to measure the curvature of the spine[31] from images acquired through different methods and in different anatomical planes[32-34]. Here we have defined it as the angle formed by the planes that correspond to the superior endplates of L1 and S1 (Fig. 2-3B). An increment of the Cobb angle indicates an increase of the overall lumbar spine lordosis. Similarly, intervertebral angles (Fig. 2-3C) and regional disc heights were measured between the planes of the inferior and superior endplates of contiguous vertebrae that are in contact with a single IVD. These heights were measured as the Euclidean distances anteriorly, centrally and posteriorly along the midsagittal line. The analysis to generate all kinematic measurements was implemented in Matlab 2010b (Mathworks Inc., Natick, MA, USA).

Resolution

In order to understand errors associated with resolution we acquired data at multiple in-plane resolutions and slice thicknesses. A volunteer was scanned using a 3.0T MRI supine scanner (GE Discovery MR750) and GE 8ch CTL Spine Array Coil (Waukesha, WI). A three-plane localizer and sagittal T2 Fast-recovery fast spin-echo (FRFSE, TR=5000ms, TE=17.2ms, NEX=2, ETL=16, FOV=25.6cm, 1mm thickness, no gap) were acquired at 8 different in-plane resolutions (0.5x0.5, 0.6x0.6, 0.7x0.7, 0.8x0.8, 0.9x0.9, 1.0x1.0, 1.1x1.1, and 1.2x1.2 mm^2). Additionally, these data sets were averaged in the slice direction to generate 2mm, 3mm and 4mm slice thickness data. The

$0.5x0.5x1.0mm^3$ image set was analyzed five times by a single user; all kinematic data were averaged and used as a reference data set. The kinematic variables were measured from all other data sets and the root mean square errors (RMSE) between these and the reference values were calculated to investigate the effect of the in-plane and slice thickness resolution on the precision of the kinematic measurements. The coefficient of variation (CV) was computed to assess the precision of the technique within and between users. For this, five data sets from the upright MRI (standing without load) were analyzed three times by two different users. The RMSE was calculated between the reference data set and a total of 31 data sets of varying in-plane resolution (0.5x0.5, 0.6x0.6, 0.7x0.7, 0.8x0.8, 0.9x0.9, 1.0x1.0, 1.1x1.1, and 1.2x1.2 mm^2) and slice thickness (1mm, 2mm, 3mm and 4mm). The resolution of the images acquired in the upright MRI scanner was $1.14x1.14x4.0mm^3$ therefore, we report the results at $1.1x1.1mm^2$ and $1.2x1.2mm^2$ in-plane resolutions and 4.0mm slice thickness. The RMSE values for the angle with respect with the horizontal were 0.16° and 0.28°, while for the Cobb angle the values were 0.60° and 0.99°. The lumbar level dependent kinematic variables were averaged to report a single value per variable. The calculated RMSE values for the IVD angles and heights were 0.39° and 0.54°, and 0.79mm and 0.83mm respectively.

The within and between user CVs were calculated for two user and all kinematic variables, lumbar level dependent variables were averaged to obtain a single value. The within-user CVs for angle with respect to the horizontal, Cobb angle, IVD angles and heights were on average 0.2% (0.02°), 1.4% (0.44°), 3.9% (0.58°) and 3.3% (0.38mm), respectively. The between-user CVs for angle with respect to the horizontal, Cobb angle, IVD angles and heights were 0.4% (0.04°), 2.7% (0.90°), 4.9% (0.72°) and 4.5% (0.55mm).

Statistical Analysis

Angle with respect to the horizontal and Cobb angle were compared using one-way repeated measurements analysis of variance and *post-hoc* Sidak tests to identify significant differences between tasks ($\alpha = 0.05$). Additionally, IVD angle and distance measurements were compared by two-way repeated measurements analysis of variance and *post-hoc* Sidak tests to identify significant differences between lumbar levels through tasks. All data in plots are reported as means ± standard deviation unless otherwise stated.

2.4 Results

Volunteers' characteristics

Complete and usable images were obtained from eight Marines (average age=20.50±1.17 years, age range 19-2, average height=179.3±10.5 cm, average weight=73.5±8.0kg, body mass index=22.8±1.6). The images from one Marine contained motion artifacts severe enough to make measurements impossible and were therefore not included in the reported results. Another Marine had a large body habitus and did not fit in the scanner; therefore, data were not collected.

Measurement of lumbar spine kinematics

Regional kinematic measurements reflect that while carrying load (LO2, LO3 and LO4) the overall position of the spine was significantly more horizontal (25°-34°, $p<0.001$) than at baseline and recovery; indicating forward flexion of the trunk (Fig. 2-

4A). Simultaneously, lumbar lordosis was also reduced (10°-13°, $p<0.05$) when compared to unloaded tasks (Fig. 2-4B).

Local IV angles and regional heights were measured to investigate their individual contribution to the observed regional changes. Lordosis was significantly ($p<0.05$) increased from baseline (UN1) to after 45 min of standing with load (LO3) at L1L2 and L2L3 levels, 5±2° and 3±2°, respectively (Fig. 2-5). Posteriorly, L1L2 was significantly distracted 2.5±1.1mm after 45 min of walking on the treadmill (LO4) when compared to the IV height after 45 min of standing with load (LO3).

This is in agreement with the increase in kyphosis observed from between these two tasks. Additionally, L2L3 was significantly compressed centrally and posteriorly after standing for 45min (LO3) with load in comparison to values at baseline (UN1) and immediately after donning load (LO2). No significant changes in anterior heights were found at either L1L2 or L2L3 (Fig. 2-6). In contrast, at L3L4 level local lordosis was significantly decreased 4±3° after 45 min of standing and walking with load (LO3 and LO4). The magnitude of the kinematic changes was the largest at both L4L5 and L5S1 where local lordosis was significantly decreased 7±2° immediately after donning load (LO2), 11±3° after 45 min of standing with load (LO3), and 9±4° after 45 min walking on the treadmill (LO4, Fig. 2-5). In summary, we measured significant anterior and central compression, and posterior distraction during loaded tasks at these levels (Fig. 2-6).

It was observed that superior lumbar levels (L1L2, L2L3 and L3L4) show significant changes only after 45 min of standing with load, while L4 and L5 decrease immediately after donning load and during all loaded tasks (Fig. 2-5). Local lordosis at all lumbar levels recovered to baseline values after the recovery period. Overall, through different tasks, the kinematic changes of superior lumbar levels (L1L2 and L2L3) are different from inferior levels (L4L5 and L5S1). Interestingly, L3L4 showed significantly different ($p<0.05$) kinematic behavior from L1L2 and L2L3 at baseline and after the recovery period, and from L4L5 and L5S1 after standing with load for 45min. No significant differences were found between L3L4 and other lumbar levels immediately after donning load or after walking for 45 min with load. In summary, the kinematic behavior of L3L4 is similar to inferior levels during tasks without load and similar to superior levels under load-carrying conditions.

2.5 Discussion

The objective of this study was to investigate the kinematic behavior of the lumbar spine of active-duty U.S. Marines while carrying heavy load. To our knowledge, this is the first study to measure level dependent lumbar spine kinematics in active-duty U.S. Marines under load-carrying conditions using an upright MRI scanner. It was possible to scan Marines with backpack LCS loaded with 50.8kg, an amount of weight that is typical of that carried in training and combat situations. However, a constraint of the technique is that the shoulder width of the subjects is limited to 31 inches. Although it is possible to scan a person with wider shoulders, the acquired images would not represent the true state of the spine when carrying load because they are supported. For this reason, it was verified that Marines were as comfortable as possible and standing on their own through the duration of the scans. Since Marines had to stand still in the scanner with the donned load, motion artifacts were found in some images, however we considered these images to still be measurable. In one case, severe motion artifacts were present and the data set was removed from the analysis. We measured the following lumbar spine kinematic variables: overall angle with respect to the horizontal, sagittal Cobb angle, IVD sagittal angles and regional IVD heights. From these, the angle with respect to the horizontal has been previously reported to be progressively reduced in proportion to the amount of load been carried by soldiers.[16, 35] Our results indicate a change in magnitude of angle with respect to the horizontal between unloaded and loaded conditions of 25-34° depending on the loaded task, which is larger than that reported by Atwells et al of ~18°.[16] The maximum load in both studies is about 50kg. The variation in reported magnitude of this angle may be due to the different LCSs, measuring techniques and the location of body markers used by Atwells et al. It has been suggested that this motion is aimed to reorient the center of mass of the system over the feet to keep balance. [9, 36] Similarly, our results indicating a reduced lordosis when carrying load are in agreement with those in the literature, however, the magnitude of the results cannot be compared due to the differences in techniques and load weights.[37] The overall spine reduction in lordosis appears to be driven by the kinematic changes occurred at the L4L5 and L5S1 levels. The magnitude of the changes in these levels is significantly larger than that of L1L2, L2L3 and L3L4 during load carrying tasks (data not shown).

Lumbar level-dependent lordosis data indicate that the superior and inferior lumbar spine has different behavior under load-carrying conditions. Superior lumbar spine levels present increased lordosis while simultaneously, inferior levels become straighter when carrying load. However, the kinematics of L3L4 seem to indicate a transition level between superior and inferior lumbar spine, whose behavior depends on the presence of load. Correspondingly, IVD height is anteriorly decreased and posteriorly increased in inferior lumbar levels. The fact that most significant changes occur in the lower lumbar spine might be related to the greater forces acting on inferior levels through the lumbar spine[38] and that IVDs of inferior levels undergo greatest posterior migration.[39]

Given that the kinematics changes in the spine may be driven by a need to realign the center of mass, it is tempting to think about a new pack design, where load is distributed differently between the front and back. However, previous work in this field has demonstrated that distributing the weight towards the front of the trunk is uncomfortable and interferes breathing and operational use of the arms.[8]

2.6 Limitations

Upright MRI has the advantage of acquiring images while subjects are in functional and relevant positions. However, this system also has characteristics that impose limitations on this study. Although this technique does not allow measuring the kinematic changes during gait, it permits evaluation of the kinematic changes over time of exposure to load and the response to tasks with load in natural posture.

The low strength magnetic field of this system directly limits the in-plane resolution, slice thickness and scan time. In this study we have shown that there are no significant differences between the kinematic variables measured from high-resolution ($0.5x0.5x1.0mm^3$) images acquired using a 3T supine MRI scanner and those from images collected with the 0.6T upright MRI scanner ($1.14x1.14x4.0mm^3$). Scan time was the main constraint of in-plane resolution and slice thickness in order to reduce the period of time that Marines had to stand still with donned load of 50.8kg. The effect of acquiring thinner slices would be an increase the acquisition time, which should remain as short as possible in order to reduce motion artifacts and assure the safety of Marines when carrying load. A balance between voxel size and scan time was then established to acquired images with tolerable motion artifact in standing position. Unfortunately, the selected slice thickness does not allow the IVD movement (i.e. bulging, herniation) to be quantitatively assessed. Another disadvantage of this system is that the biochemical state of the IVDs cannot be described using techniques readily available on a high-resolution supine scanner (i.e. T2 mapping, T1rho, spectroscopy).

The FOV was limited by the size of the available planar RF coil, which did not permit the acquisition of other bony anatomical references and the lumbar spine in a single image set. Therefore, the rotational corrections applied in the axial and coronal planes were made with reference to the position of the superior plane of S1. This might result in intervertebral angles measured in the sagittal oblique plane; however, these angles still reflect the relative position between vertebrae of the lumbar spine.

The present study was performed in non-obese male Marines, which makes it difficult to extrapolate our results to populations of different age, gender and body habitus. Additionally, Marines in the present study wore a body armor, which may reduce the range of motion (ROM) of the lumbar spine. Assuming that the effect of the body armor on lumbar spine kinematics is negligible, it is possible that in subjects with increased ROM such as children[17] and females[40] the magnitude of the changes observed are greater than those observed here. Inversely, in an older population with known decreased ROM[41] the magnitude of level-dependent kinematics is expected to decrease therefore, compromising the capacity of the lumbar spine to accommodate kinematic changes.

2.7 Conclusions

In conclusion, we measured the kinematic behavior of the lumbar spine of active-duty US Marines while carrying heavy loads. Our results suggest that when Marines carry load and lean forward the superior functional units of the lumbar spine act differently from the inferior units. Locally, the superior levels go into lordosis while inferior levels become more kyphotic. The contribution of each intervertebral level is reflected in lumbar spine flexion and reduced lordosis during load-carrying tasks. Moreover, the anterior disc region of inferior lumbar levels is compressed while the posterior disc region is distracted leading to immediate kinematic changes after donning load. This is in contrast to superior lumbar levels, which undergo changes in their kinematic behavior over a longer load duration. Future research is needed to investigate how this behavior over time affects health outcomes related to LBP and degeneration in military and civilian populations.

2.8 Acknowledgements

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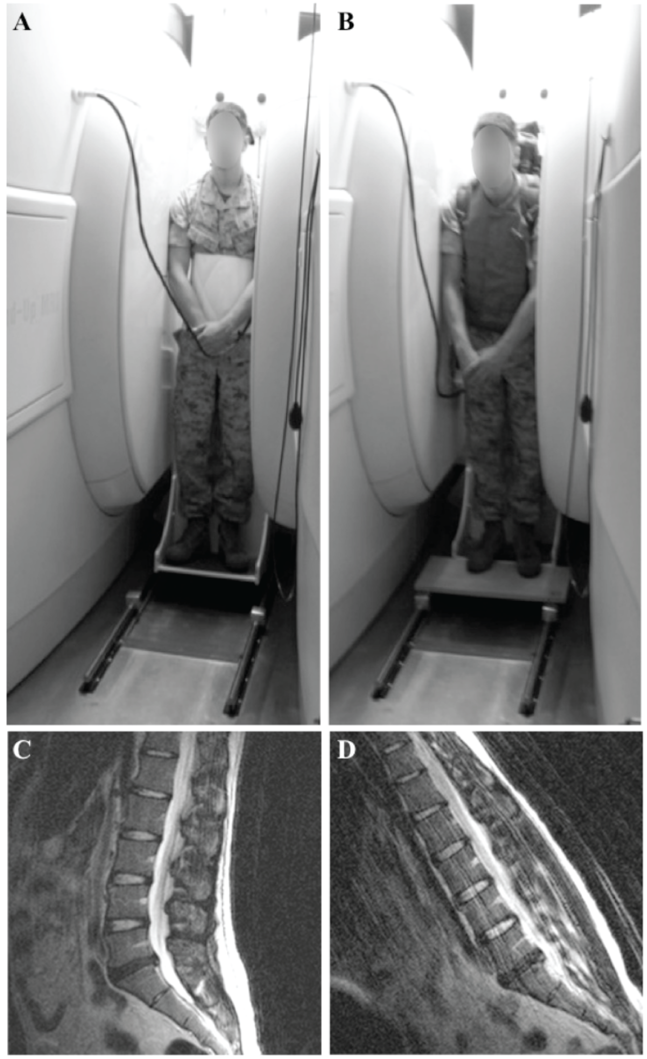

Figure 2−1 Photograph of a Marine standing in the MRI A) without load and B) with load. Representative midsagittal magnetic resonance image of the lumbar spine C) without load and D) with load.

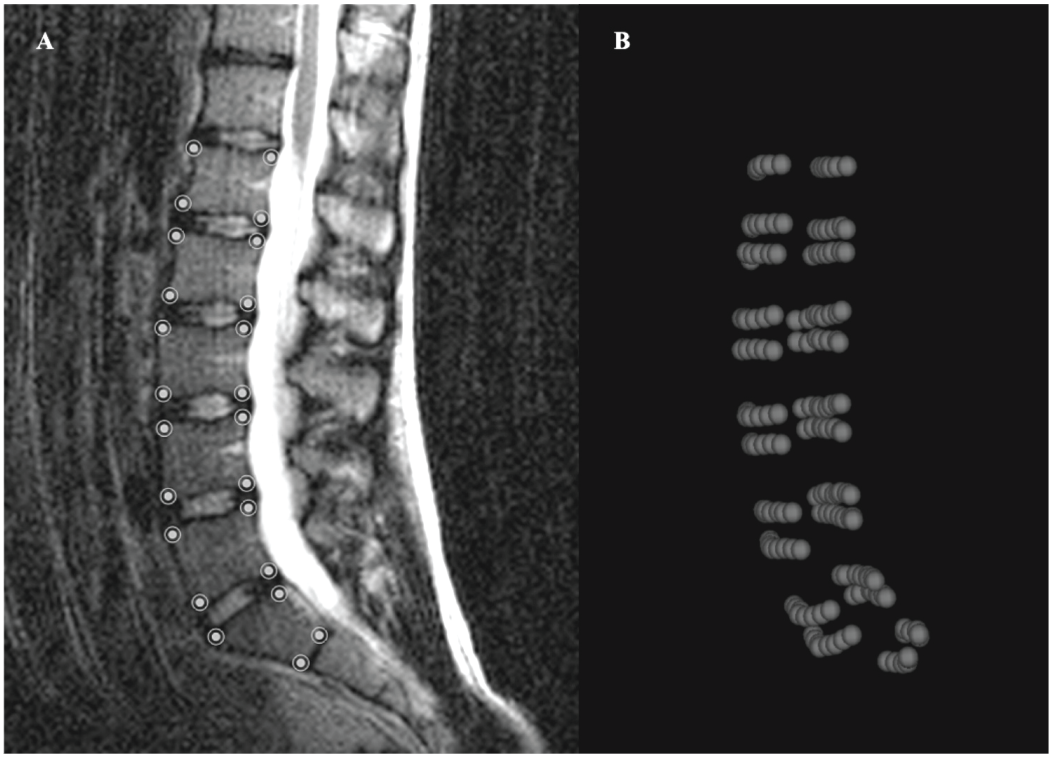

Figure 2−2 Spinal MR images in upright posture. A) Example of upright MRI T2-weighted image of the lumbar spine with ROI points at the corners of each vertebra (L1−S1), B) 3D representation of the lumbar spine vertebrae.

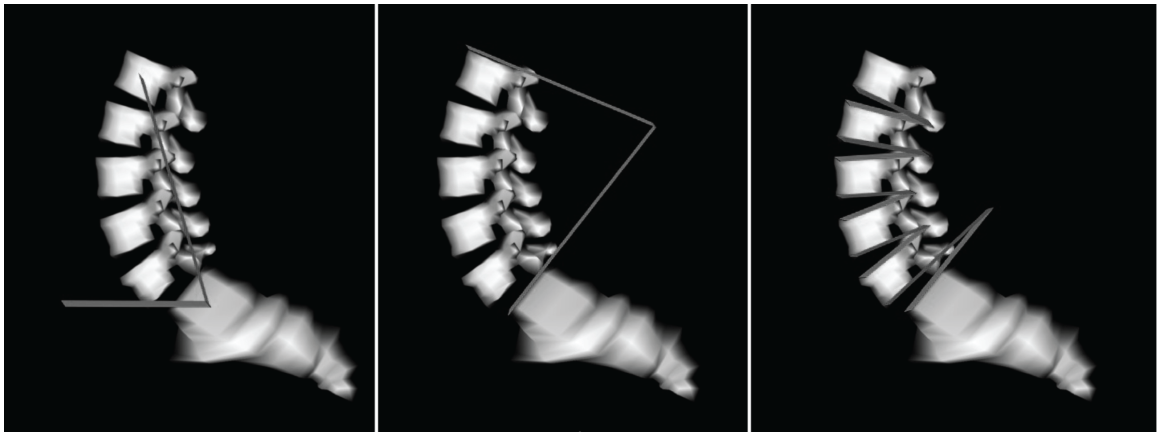

Figure 2−3 Lumbar spine kinematic measurements on a graphical representation of the lumbar spine. Angle with respect with the horizontal (left), sagittal Cobb angle (center), and intervertebral sagittal angles (right). These images were generated using OpenSim model of lumbar spine.[40, 41]

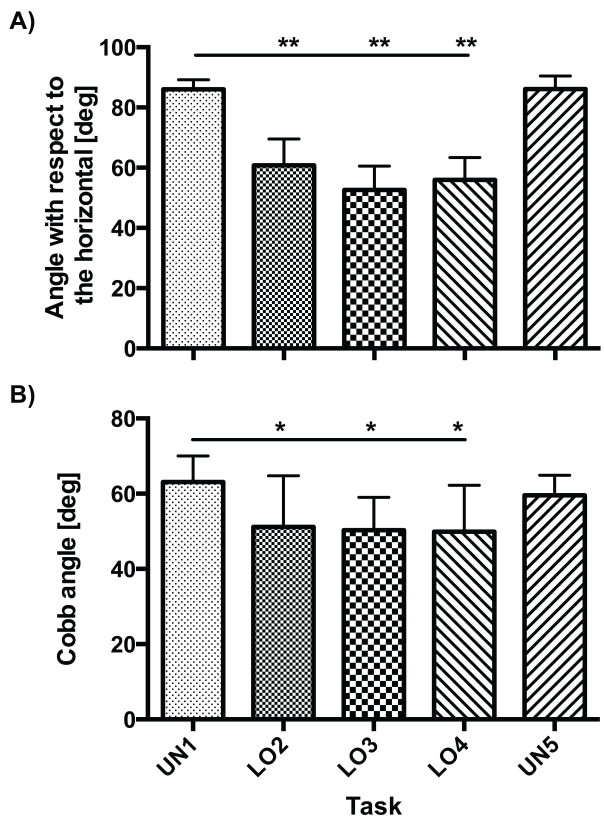

Figure 2−4 Whole lumbar spine kinematic measurements. A) angle with respect to the horizontal, B) sagittal Cobb angle, per task (UN1: unloaded, LO2: immediately after donning load, LO3: after 45min of standing with load on, LO4: after walking for 45min with load on, UN5: after side-lying for 45min). Significant differences were found between loaded and unloaded tasks ($p<0.0001$) for both angle with respect to the horizontal and Cobb angle.

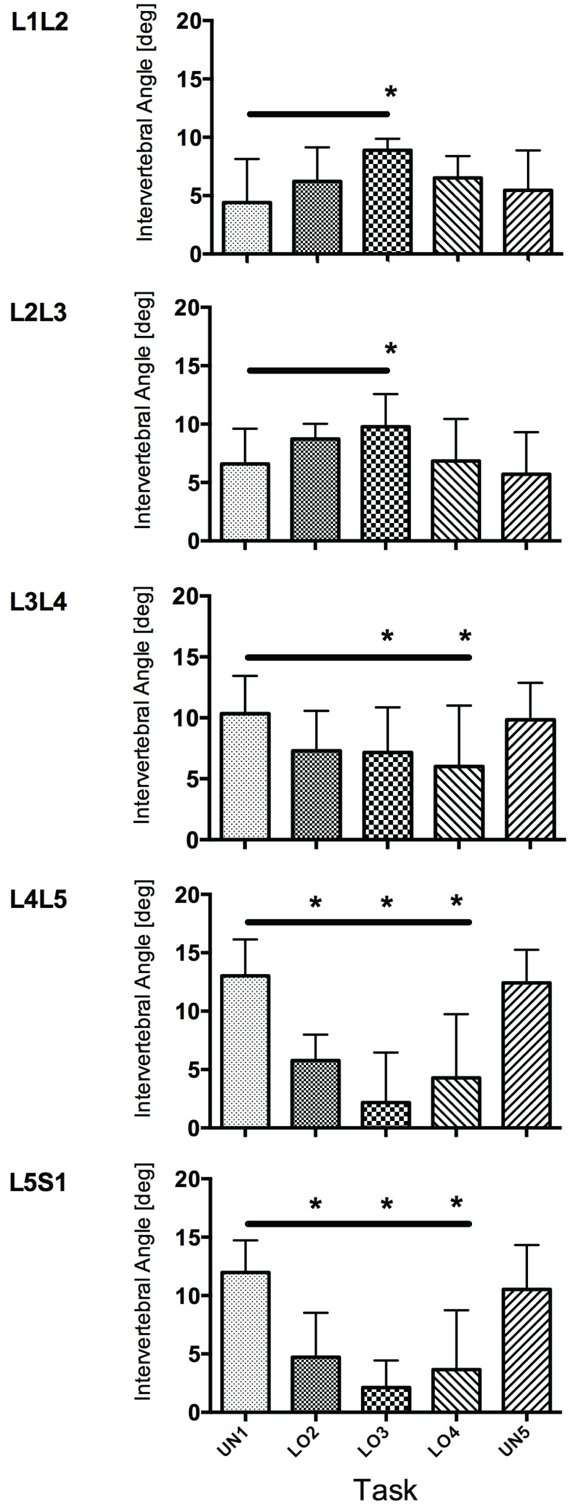

Figure 2−5 IVD angle per lumbar level (L1-L5) and task. Most significant differences were found after LO3 ($p<0.05$) through all lumbar levels (L1-L5). Additionally, L4L5 and L5S1 became significantly more kyphotic during all tasks with load ($p<0.05$).

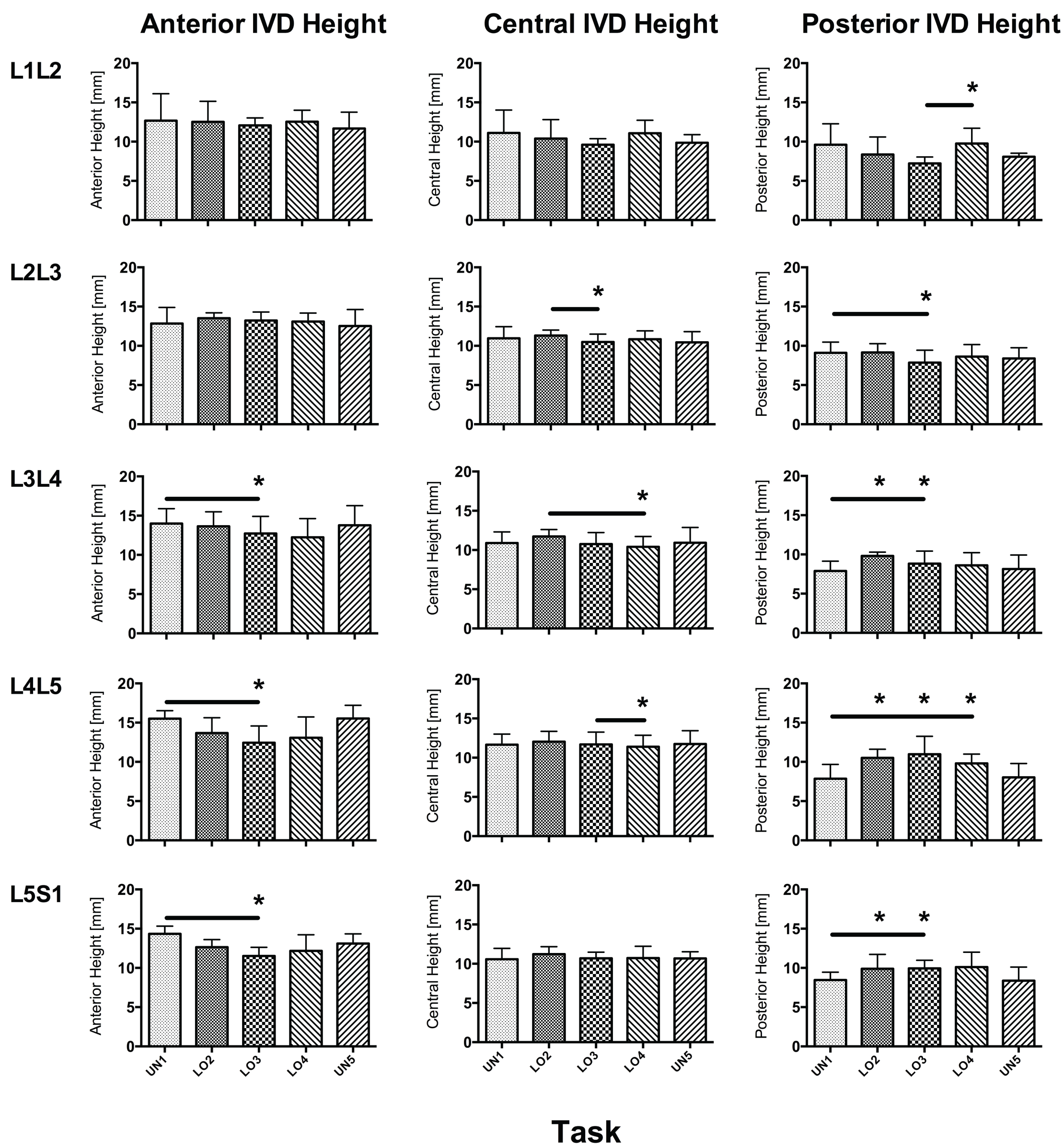

Figure 2‒6 Anterior (left), central (center) and posterior (right) IVD heights at L1L2, L2L3, L3L4, L4L5 and L5S1, per task (UN1: unloaded, LO2: immediately after donning load, LO3: after 45min of standing with load on, LO4: after walking for 45min with load on, UN5: after side-lying for 45min). The IVDs of lumbar levels L3L4, L4L5 and L5S1 were anteriorly compressed after LO3 and posteriorly distracted in most tasks with load ($p<0.05$). Central IVD heights were significantly reduced between load-carrying tasks with ($p<0.05$).

2.7 References

1. Knapik JJ, Reynolds KL, Harman E: **Soldier load carriage: historical, physiological, biomechanical, and medical aspects.** *Mil Med* 2004, 169(1):45-56.

2. Dean C: **The Modern Warrior's Combat Load. Dismounted Operations in Afghanistan**. In. Ft Leavenworth, KS: U.S. Army Center for Army Lessons Learned 2004.

3. Field Manual 21-18, **Foot Marches**. Washington, DC: Department of the Army; 1990..

4. Datta SR, Ramanathan NL: **Ergonomic comparison of seven modes of carrying loads on the horizontal plane.** *Ergonomics* 1971, 14(2):269-278.

5. Bhambhani Y, Buckley S, Maikala R: **Physiological and biomechanical responses during treadmill walking with graded loads**. *Eur J Appl Physiol Occup Physiol* 1997, **76**(6):544-551.

6. May B, Tomporowski PD, Ferrara M: **Effects of backpack load on balance and decisional processes**. *Military medicine* 2009, **174**(12):1308-1312.

7. Polcyn AB, CK. ; Harman, EA. ; Obusek, JP.: **The Effects of Load Weight: A Summary Analysis of Maximal Performance, Physiological, and Biomechanical Results from Four Studies of Load-Carriage Systems.** Natick, MA: U.S. Army Soldier Systems Center 2007.

8. Johnson RF, Knapik JJ, Merullo DJ: **Symptoms during load carrying: effects of mass and load distribution during a 20-km road march.** *Percept Mot Skills* 1995, 81(1):331-338.

9. Knapik J, Harman E, Reynolds K: **Load carriage using packs: a review of physiological, biomechanical and medical aspects.** *Appl Ergon* 1996, 27(3):207-16.

10. Cohen SP, Gallagher RM, Davis SA, Griffith SR, Carragee EJ: **Spine-area pain in military personnel: a review of epidemiology, etiology, diagnosis, and treatment.** *Spine J* 2012, 12(9):833-42.

11. Ulaska J, Visuri T, Pulkkinen P, Pekkarinen H: **Impact of chronic low back pain on military service**. *Military medicine* 2001, **166**(7):607-611.

12. Toblin RL, Riviere LA, Thomas JL, Adler AB, Kok BC, Hoge CW: **Grief and physical health outcomes in U.S. soldiers returning from combat**. *J Affect Disord* 2012, **136**(3):469-475.

13. Meakin JR, Smith FW, Gilbert FJ, Aspden RM: **The effect of axial load on the sagittal plane curvature of the upright human spine in vivo.** *J Biomech* 2008, 41(13):2850-2854.

14. Yen SC, Gutierrez GM, Ling W, Magill R, McDonough A: **Coordination variability during load carriage walking: Can it contribute to low back pain?** *Hum Mov Sci* 2012, 31(5):1286-301.

15. Bettany-Saltikov J, Cole L: **The effect of frontpacks, shoulder bags and handheld bags on 3D back shape and posture in young university students: an ISIS2 study.** *Stud Health Technol Inform* 2012, 176:117-121.

16. Attwells RL, Birrell SA, Hooper RH, Mansfield NJ: **Influence of carrying heavy loads on soldiers' posture, movements and gait**. *Ergonomics* 2006, **49**(14):1527-1537.

17. Neuschwander TB, Cutrone J, Macias BR, Cutrone S, Murthy G, Chambers H, Hargens AR: **The effect of backpacks on the lumbar spine in children: a standing magnetic resonance imaging study**. *Spine (Phila Pa 1976)* 2010, **35**(1):83-88.

18. Schiffman JM, Bensel CK,Hasselquist L, Norton K, Piscitelle L: **The Effects Of Soldiers' Loads on Postural Sway**. Natick, MA: U.S. Army Soldier Systems Center 2004.

19. Filaire M, Vacheron JJ, Vanneuville G, Poumarat G, Garcier JM, Harouna Y, Guillot M, Terver S, Toumi H, Thierry C: **Influence of the mode of load carriage on the static posture of the pelvic girdle and the thoracic and lumbar spine in vivo**. *Surg Radiol Anat* 2001, **23**(1):27-31.

20. Fujii R, Sakaura H, Mukai Y, Hosono N, Ishii T, Iwasaki M, Yoshikawa H, Sugamoto K: **Kinematics of the lumbar spine in trunk rotation: in vivo three-dimensional analysis using magnetic resonance imaging**. *Eur Spine J* 2007, **16**(11):1867-1874.

21. Kimura S, Steinbach GC, Watenpaugh DE, Hargens AR: **Lumbar spine disc height and curvature responses to an axial load generated by a compression device compatible with magnetic resonance imaging.** *Spine (Phila Pa 1976)* 2001, 26(23):2596-2600.

22. Hioki A, Miyamoto K, Sakai H, Shimizu K: **Lumbar axial loading device alters lumbar sagittal alignment differently from upright standing position: a computed tomography study.** *Spine (Phila Pa 1976)* 2010, 35(9):995-1001.

23. Willen J, Danielson B, Gaulitz A, Niklason T, Schonstrom N, Hansson T: **Dynamic effects on the lumbar spinal canal: axially loaded CT-myelography and MRI in patients with sciatica and/or neurogenic claudication.** *Spine (Phila Pa 1976)* 1997, 22(24):2968-2976.

24. Alyas F, Connell D, Saifuddin A: **Upright positional MRI of the lumbar spine**. *Clin Radiol* 2008, **63**(9):1035-1048.

25. Jinkins JR, Dworkin J: Proceedings of the State-of-the-Art Symposium on Diagnostic and Interventional Radiology of the Spine, Antwerp, September 7, 2002 (Part two).

Upright, weight-bearing, dynamic-kinetic MRI of the spine: pMRI/kMRI. *JBR-BTR* 2003, 86(5):286-293.

26. Wang S, Passias P, Li G, Wood K: **Measurement of vertebral kinematics using noninvasive image matching method-validation and application.** *Spine (Phila Pa 1976)* 2008, 33(11):E355-361.

27. Janssen M, Nabih A, Moussa W, Kawchuk GN, Carey JP: **Evaluation of diagnosis techniques used for spinal injury related back pain**. *Pain Res Treat* 2011, **2011**:478798.

28. Knapik JJ, Harman EA, Steelman RA, Graham BS: **A systematic review of the effects of physical training on load carriage performance**. *J Strength Cond Res* 2012, **26**(2):585-597.

29. Williams AG, Rayson MP, Jones DA: **Training diagnosis for a load carriage task**. *J Strength Cond Res* 2004, **18**(1):30-34.

30. Rosset A, Spadola L, Ratib O: **OsiriX: an open-source software for navigating in multidimensional DICOM images.** *J Digit Imaging* 2004, 17(3):205-216.

31. Cobb JR: **Outline for the study of scoliosis**. *AAOS, Instructional Course Lectures* 1948, **5**(Journal Article):261-275.

32. Schmitz A, Kandyba J, Koenig R, Jaeger UE, Gieseke J, Schmitt O: **A new method of MR total spine imaging for showing the brace effect in scoliosis**. *J Orthop Sci* 2001, **6**(4):316-319.

33. Geijer H, Beckman K, Jonsson B, Andersson T, Persliden J: **Digital radiography of scoliosis with a scanning method: initial evaluation**. *Radiology* 2001, **218**(2):402-410.

34. Asghar J, Samdani AF, Pahys JM, D'Andrea L P, Guille JT, Clements DH, Betz RR: **Computed tomography evaluation of rotation correction in adolescent idiopathic scoliosis: a comparison of an all pedicle screw construct versus a hook-rod system.** *Spine (Phila Pa 1976)* 2009, 34(8):804-807.

35. Bust PD, McCabe PT: Contemporary Ergonomics 2005: Proceedings of the International Conference on Contemporary Ergonomics (CE2005), 5-7 April 2005, Hatfield, UK: Taylor & Francis; 2005.

36. Bloon DW & Woodhull-McNeal AP: **Postural adjustments while standing with two types of loaded backpacks.** *Ergonomics* 1987, 30.

37. Chow DH, Leung KT, Holmes AD: **Changes in spinal curvature and proprioception of schoolboys carrying different weights of backpack.** *Ergonomics* 2007, 50(12):2148-2156.

38. Pal GP, Routal RV: **Transmission of weight through the lower thoracic and lumbar regions of the vertebral column in man.** *J Anat* 1987, 152:93-105.

39. Alexander LA, Hancock E, Agouris I, Smith FW, MacSween A: **The response of the nucleus pulposus of the lumbar intervertebral discs to functionally loaded positions**. *Spine (Phila Pa 1976)* 2007, **32**(14):1508-1512.

40. Van Herp G, Rowe P, Salter P, Paul JP: **Three-dimensional lumbar spinal kinematics: a study of range of movement in 100 healthy subjects aged 20 to 60+ years.** *Rheumatology (Oxford)* 2000, 39(12):1337-1340.

41. Keorochana G, Taghavi CE, Lee KB, Yoo JH, Liao JC, Fei Z, Wang JC: **Effect of sagittal alignment on kinematic changes and degree of disc degeneration in the lumbar spine: an analysis using positional MRI.** *Spine (Phila Pa 1976)* 2011, 36(11):893-898.

CHAPTER 3 : THE EFFECT OF TRAINING ON LUMBAR SPINE KINEMATICS AND INTERVERTEBRAL DISC DEGENERATION IN ACTIVE-DUTY MARINES

3.1 Abstract

Low back pain (LBP) in the military population has been associated with heavy load carriage. Military training consists of a combination of exercises to improve high intensity load bearing performance. There are limited data available that address the changes in spinal musculoskeletal elements as related to load carriage tasks throughout military training progression. Therefore, the purpose of this study was to understand how training influences the postural behavior of the loaded lumbar spine (LS). We hypothesized that training would reduce lumbar lordosis and trunk flexion, and that intervertebral disc (IVD) degeneration would alter LS kinematics. Active-duty Marines (n=27) were scanned with and without load (22.6 kg) at the beginning, middle and end of basic School of Infantry (SOI) training using an upright magnetic resonance imaging (MRI) scanner. Images were used to grade IVD degeneration at the L5−S1 lumbar level and post-processed to measure global and level specific LS posture. Two-way repeated-measures ANOVA and Sidak *post-hoc* tests were used to compare LS posture as a function of training time and task, and disc health and magnitude of change in posture between tasks (α=0.05). No changes in posture and IVD degeneration were found throughout training. The LS was less lordotic and the sacrum became more horizontal when carrying load. The origin of the change in lordosis appears to be the inferior LS (L3-S1), as superior and inferior lumbar levels showed different behaviors during load

carriage. Marines with degeneration at the L5-S1 level had larger sacral postural perturbations in response to load. Although the posture of the LS changes in response to loading, load-induced postural changes did not change throughout the training period. This finding suggests that the postural response to load is defined more by the task needs than by the physical condition of the Marine. Given the general pattern of sacral and LS flexion in response to load, it is possible that the postural strategy is simply aimed at centralizing a heavy posterior load over the base of support.

3. 2 Introduction

Low back pain (LBP) in the military population has been associated with carrying heavy loads during training and operational tasks [1-4]. In an effort to reduce these adverse effects, the optimum balance between load carriage training, physical fitness, and performance in the military population has been studied in terms of energy cost, distance, and speed [5-7]. It is widely accepted that to improve high intensity load carriage performance, military training should consist of a combination of aerobic and resistance exercises [6].

The United States Marine Corps (USMC) School of Infantry (SOI) West at Camp Pendleton, California, follows this training paradigm. The School of Infantry is the second stage of Marine Corps training for infantrymen immediately following 10 weeks of boot camp. Prior to this stage, Marines are naïve to heavy load carriage. The duration of SOI is 41 days, during which march distances are progressively increased—5 km, 10 km, 15 km, and 20 km performed around days 12, 16, 28, and 40—under load. All marches are conducted with a standard fighting load, which is approximately 33.6 kg.

During the 15km and 20km training marches, Marines are also required to carry their designated weapon system during training.

Despite the association between LBP, load carriage and the structured SOI training paradigm, which progressively increases intensity of load carriage via increased hike duration, there are no data documenting the behavior of spinal structures as Marines progress through SOI. To date, one study by Aharony et al. measured the impact of Israeli Navy Special Forces training on lumbar spine (LS) pathology through physical examination and radiological evaluation; however, no overuse changes or new injuries in the LS were noted [8]. In a previous study, we measured whole LS and lumbar level-dependent postural changes in active-duty Marines while carrying a load of 50.8 kg [9]. We found that these changes appeared to be responses to center of mass realignment (subject and backpack). More locally, these observed changes originated from the disparate postural behavior of the superior and inferior LS. However, the Marines who were evaluated in our previous study had already been in operation for 8–48 months and were conditioned to carry heavy loads while marching. Importantly, Marines participating in both studies (Aharony *et al* and Rodriguez-Soto *et al*) had measureable, pre-existing structural changes in muscles, bones, and IVDs that may have affected load-carrying posture [8, 9].

The interaction between pelvic and LS posture has been previously investigated in the standing position.[10-12] The strongest association found exists between sacral inclination and LS lordosis, which reveals that these two variables are proportional to each other.[11, 12] Meaning that in people with a more horizontal sacrum, the LS is more lordotic, and vice versa. Furthermore, in the presence of IVD degeneration (herniations and general LBP) reduced sacral inclination and LS lordosis have been reported.[13, 14] In the context of load carriage, pelvic and LS orientation have been previously estimated using motion capture technology, but never measured directly.

Given the lack of data documenting LS posture, and structural changes for Marines exposed to load carriage, the purposes of this study were to: (1) compare LS postural adaptations to load over the course of SOI training, (2) understand the effect of training on IVD degeneration, and (3) understand the effect of IVD degeneration on LS postural adaptations during training. We hypothesized that the magnitude of load carriage-induced LS postural changes will be reduced with training, and that Marines with IVD degeneration will manage loads and adapt to training differently than Marines without IVD degeneration.

3.3 Materials and Methods

Subjects

Forty-one male Marines from three different companies enrolled at SOI West Marine Corps Base at Camp Pendleton, and with no recent history of LBP volunteered to participate in this study. The University of California, San Diego and Naval Health Research Center institutional review boards approved this study, and all volunteers provided oral and written informed consent.

Imaging

Marines were scanned using an upright 0.6T magnetic resonance imaging (MRI) scanner (UPRIGHT® Multi-Position MRI, Fonar Corporation, Melville, NY, USA) and a planar coil. A soft sleeve was used to retain the coil behind the volunteer’s back at the lumbar spine (L1–S1) level while standing. The sleeve was tight enough to keep the coil in place yet loose enough not to alter the volunteer's natural standing position. A three-plane localizer and sagittal T2-weighted images (repetition time 1974 msec; echo time 160 msec; field of view 32 cm; 224×224 acquisition matrix; 1.43×1.43 mm^2 pixel size; 4.5mm slice thickness; 0.5 mm gap; number of averages 1, scan duration 2 minutes 30 seconds) were acquired.

Load-Carrying Tasks

Marines were transferred from Camp Pendleton to MRI facilities at three time points: around day 1, day 20, and day 40 of SOI training. At each visit, Marines were first scanned standing without external load (unloaded) and after standing with a total load of 22.6 kg in an Improved Load Bearing Equipment (ILBE) backpack for 45 minutes (Fig. 3−1). This load mass (22.6kg) was selected because it is operationally relevant and to avoid injuries induced by early overloading during SOI training. The magnitude of the load was kept constant during the experimental period to determine if training improved the ability to manage a constant load. During the standing period, Marines were allowed to move around the waiting room (~100 sq ft), but were instructed not to lean on surfaces or against the wall. After the 45 minute load-carriage period, Marines were scanned a second time while carrying the same load. All ILBE backpacks were previously screened for ferromagnetic components; no metal components were found; therefore, no alterations were needed to make the backpack MRI-safe. For this second scan, the coil was placed between the backpack and the Marine's spine. In addition, Marines were purposefully not given instructions on how to stand in the scanner, but they were instructed to remain still during the entire MRI acquisition.

Data Analysis

Each image set was analyzed as previously described [9]. Briefly, a set of markers was manually placed at the corners of each vertebra (L1–S1) and posterior elements to model vertebral position and orientation. Relative rotations in the axial and coronal planes between contiguous vertebrae were removed, and the resulting vertebral endplate representations were used to generate postural measurements in the sagittal plane.[15]

Measurements

The degeneration level of the IVDs was determined for all data sets by an experienced radiologist (C.B.C.) using the Pfirrmann scoring system. This grading scale has five levels (I–V), where I corresponds to normal, II to mild degeneration, III to moderate degeneration, IV to severe degeneration, and V to advanced degeneration [16]. Marines were grouped based upon the degeneration of the L5–S1 level IVD; those graded with Pfirrmann scores of I and II were assigned to the 'non-degenerated' group, and those with scores of III, IV, and V were in the 'degenerated' group [17].

Postural measurements of the LS and pelvis in the sagittal plane were generated from vertebral endplates as previously described in the *Data Analysis* section. These variables were:

- *Angle with respect to the horizontal:* quantifies the overall position of the LS (L1 to S1) with respect to the ground (i.e., flexion, extension); however, it does not convey relative postural information among LS levels.
- *Sacral slope (SS*): defined as the angle between the superior endplate of S1 and the horizontal. Sacral slope was considered a surrogate measurement of pelvic tilt, assuming that the motion between sacrum and pelvis is negligible. A small SS

value indicates that the orientation of the sacrum is close to the vertical, while a larger value describes a more horizontal sacrum.

- *Cobb angle*: has been extensively used to measure the curvature of the spine [18-20]. Here, we have defined it as the angle formed by the planes corresponding to the superior endplates of L1 and S1 in the sagittal plane. As such, an increase in LS lordosis will be reflected by an increase in the Cobb angle, and vice versa. It has been previously reported that superior and inferior regions of the LS have different postural adaptations to load; therefore, the *superior sagittal Cobb angle* was defined as the angle formed by the superior endplate of L1 and the inferior endplate of L3, and the *inferior sagittal Cobb angle* as the angle between the superior endplate of L4 and inferior endplate of S1 [9].
- *Segmental intervertebral angles and regional disc heights* were measured between the planes of the inferior and superior endplates of adjacent vertebrae. Intervertebral heights were measured as the shortest distance between inferior and superior endplates anteriorly, centrally, and posteriorly in the midsagittal plane.

Statistical Analysis

All data distributions were tested for normality using Shapiro-Wilk tests. The absolute values of all variables were compared across training time using two-way repeated-measures analyses of variance (ANOVA) with Sidak *post-hoc* tests to identify significant differences as a function of task and time. Additionally, the effect of IVD degeneration on the magnitude of change of each postural measurement throughout training was investigated using two-way repeated measures ANOVAs (IVD degeneration x time). Again, Sidak *post hoc* tests were used to identify significant differences between

IVD degeneration and training time. The threshold for significance (α) was set at 0.05 for all analyses. Statistical analyses were performed using SPSS Statistics software (version 20.0, IBM, Armonk, NY), and all data are reported as mean ± standard deviation (SD) values.

3.4 Results

Volunteer Characteristics

Complete image data sets for each time point were obtained from 27 Marines (mean ±SD age, 19.5±1.8 years; age range 17–25 years; height, 178.4±5.6 cm; weight, 82.3±8.4 kg; body mass index, 25.8±1.8 kg/m^2). Of the 41 Marines enrolled in the study, 14 (34%) missed at least one visit and those cases were omitted from analysis.

Measurement of IVD Degeneration

The distribution of the Pfirrmann grades by lumbar level is shown in 3−1. The incidence of degenerated IVDs progressively increased from superior to inferior lumbar levels, but no progression in degeneration was observed during the training period. There were 16 volunteers in the non-degenerated group and 11 in the degenerated group, based on the Pfirrmann grades of the L5–S1 IVD.

Measurement of Lumbar Spine Load-Carriage Postural Changes

No changes were observed in any of the measured variables between loading tasks throughout training. The overall position of the spine was significantly ($p<0.05$) more horizontal when carrying load compared with those without load, at all time points (Fig. 3−2, Table 3−2). Simultaneously, the sacrum orientation became significantly more horizontal ($p<0.05$) when carrying load, compared to its orientation when standing unloaded (Fig. 3−3A). Marines with L5-S1 IVD degeneration had a larger ($p<0.05$) change in sacrum orientation (7.94°±4.17°) between unloaded and loaded tasks, compared to Marines without degeneration at the same lumbar level (4.13°±4.18°; Fig. 3−3B). Absolute SS values suggest that this difference is attributed to the orientation of the sacrum when loaded (degenerated 43.39°±4.01°, non-degenerated 38.38°±7.56°).

Additionally, during load carriage, simultaneous to LS flexion and sacrum orientation changes there was a reduction on whole LS lordosis ($p<0.05$, Fig. 3−4A). No significant differences in the magnitude of postural response to load were found between L5–S1 IVD degeneration groups (Fig. 3−4B). However, we observed a trend ($p=0.07$) towards reduced change in lordosis in the group with degeneration (Fig. 3−4B). This suggests that individuals with L5S1 degeneration may have a reduced postural adaptation in response to loading compared to individuals without degeneration.

In order to investigate which LS regions contributed to the overall lordosis changes induced by load exposure, we measured the curvature of both superior and inferior LS. The exposure to load did not cause any detectable changes in the curvature of the superior LS; however, the inferior LS became less lordotic in response to load (Fig. 3−5A). The magnitude of change between tasks was not different between Marines regardless of the presence of degeneration at the L5-S1 IVD (Fig. 3−5B).

In terms of the intervertebral angles across lumbar levels, we observed that overall the magnitude of the response to load is larger at inferior lumbar levels than at superior levels (Fig. 3−6). Specifically, the L1–L2 level became more lordotic (unloaded 5.05°±1.63°, loaded 6.01°±1.60°) in response to load—in contrast to inferior levels L3-L4 (unloaded 9.28°±1.80°, loaded 8.3°±2.45°), L4–L5 (unloaded 10.83°±2.23°, loaded

7.48°±3.56°) and L5–S1 (unloaded 10.83°±4.04°, loaded 6.82°±2.58°), which became less lordotic ($p<0.05$, Fig. 3−6 A–E). No postural changes were detected in response to load at the L2–L3, suggesting that it acts as a "transition" level.

Anterior and posterior intervertebral distances at the L1-L2 and L2-L3 levels were not different by task. Overall, changes in regional IVD distances reflect postural kinematics throughout lumbar levels (Fig. 3−7). Interestingly, the L3-L4 level became anteriorly and centrally compressed ($p<0.05$), but no changes were observed posteriorly. Similarly, at the L4-L5 level the IVD was anteriorly and centrally compressed ($p<0.05$), and posteriorly distracted during load carriage. Finally, at the L5-S1 level only anterior compression and posterior distraction were observed. These data suggest that the center of rotation of the IVD may shift from a posterior position from L3-L4 to a more central location by L5-S1.

3.5 Discussion

The main objective of this study was to measure the postural changes of the LS with and without load throughout USMC SOI training. School of Infantry training includes both aerobic and resistance exercise (e.g., long training marches and heavy load carriage), presumably improving Marines' endurance and strength while progressively exposing them to load carriage. Based on this paradigm, we hypothesized that it would become progressively easier to carry a fixed-load magnitude over the training period because of improvements in endurance, strength, and motor learning. Additionally, we hypothesized that the presence of IVD degeneration would alter LS postural adaptations to load carriage. Other authors have evaluated the outcomes of military training in terms of physical condition testing and radiological evaluation of the IVDs. However, biomechanical data on the adaptation of the LS to load carriage as a function of SOI training progression was lacking. In this study, we applied novel and valid tools, which allow postural changes in response to load and training to be quantified. This strategy allowed us to document the changes in LS load carriage kinematics between a group of active-duty Marines with and without degeneration of the L5-S1 IVD.

Across all subjects, no differences were found in LS posture in response to load during the training period. However, differences between subjects with IVD degeneration and those without were observed. Specifically, subjects with IVD degeneration demonstrated larger sacral and trended towards smaller perturbations in LS lordosis in response to load.

To quantify global LS posture, we measured LS flexion, whole LS and regional lordosis, and sacral inclination. Intervertebral disc angles and heights were used to assess local lumbar postural changes. These data suggest that when external load is applied the LS becomes more flexed, which is in agreement with previous reports [1, 21, 22]. This increase in lumbar flexion may be a compensatory response used to reorient the center of mass of the system (body + loaded pack) over the feet [3, 23] however, this idea needs to be tested explicitly. In this study, LS flexion was on average 72.74±5.04° (or 17.26° anterior to vertical) when carrying a load of approximately 25% BW. In a previous study [9], LS flexion was roughly 52° (or 38° anterior to vertical) when carrying a load of 50kg (~68% BW). These findings suggest that there is a proportional increase in trunk flexion with increasing load, which is again consistent with previous literature [3, 6]. For example, when using different methods (motion capture), Atwells *et al*. reported trunk flexion between 77° and 80° (with respect to the horizontal) when walking with loads of 15.95kg (22% BW) on a waist belt and 20kg (27% BW) in a backpack, respectively [1]. We attribute the variation in magnitude to the differences in measurement tools and experimental setups between these three studies. Of note, we have presented direct measurements of spinal elements versus LS surface measurements.

In order to understand the contribution of both pelvic and LS components to the overall LS posture, we measured SS. In the present study, the SS when standing without external load was 34.43°±8.3°, whereas most of the values previously reported in the literature range between 39° and 42°.[11, 12] The discrepancy between these data might be caused by the difference in measuring tools. All SS values reported in the literature while standing were performed using X-rays, while we have used an MRI based three-dimensional tool to measure posture. Another possible explanation might related to high variation in postural characteristics of the population; the range of individual SS values reported in the literature varies around 20°-65°. Furthermore, we directly measured SS during load carriage in a group of young active-duty Marines− data that was lacking in the literature. Other authors have previously studied the effect of load carriage on pelvic tilt during gait in a group of soldiers, female students and children.[24-26] In all cases, the authors used motion capture to perform measurements of the hip joint range of motion and did not report absolute values of pelvic tilt, making comparison to our data impossible.

In addition to increased lumbar flexion and sacral inclination, LS lordosis was reduced when carrying a load, which is also consistent with previous observations. Neuschwander et al. measured lumbar lordosis in children carrying backpacks of 10%, 20%, and 30% BW from images acquired using an upright MRI scanner. These authors reported ~60° of lordosis when standing without load and ~55° of lordosis when carrying 30% BW [27]. These values were obtained using a similar definition of the Cobb angle used in this study, but they were measured two dimensionally. In this study, we have found 50° of lordosis when standing without load and 40° after 45 minutes of standing with ~25% BW. Such findings are also in agreement with our previous study, for which LS lordosis was 52° when standing without load and 40° after standing for 45 minutes with 50kg of load [9]. Interestingly, whole LS lordosis values reported by Neuschwander et al. [27], Rodríguez-Soto *et al*. [9], and in the present study are very similar despite the differences in the magnitude of the load carried. However, when comparing the local lordosis at each lumbar level previously reported by our group[9] and those of the present study, we identified that the superior LS had a larger increase in lordosis when carrying 50kg of load than when carrying 26kg. Similarly, the reduction in inferior LS lordosis was larger when carrying the heavier load; resulting in a similar value in whole LS lordosis, but with different contributions from each lumbar level.

Local LS posture measurements indicate that the overall reduction in LS lordosis is primarily driven by the changes that occurred at the L4–L5 and L5–S1 levels. These data also suggest that the LS experiences two opposing motions under load-carrying conditions; at L1–L2 lordosis increases, while inferior levels (L3–L4, L4–L5, and L5–S1) become straighter. The lack of postural changes at the L2–L3 level suggests it serves as transition level between superior and inferior LS. Interestingly, the location of these transition levels appears to depend on the presence and magnitude of load. In our previous study, we reported that the transition level was L3–L4 when carrying 50kg of load. However, in that evaluation, Marines wore body armor as part of their total load, while in the present study they did not. This is a limitation to comparing relative changes of the LS since the body armor may (or may not) have affected how the LS changes with load. Future work is being conducted to elucidate the effects of body armor on LS posture both with and without load. Additionally, the location of transition levels during load carriage might be associated with the location of the lumbar lordosis apex of each person when standing unloaded. The variation of the apex location ranges from the base of L3 to the middle region of L5 depending on the pelvic and lumbar sagittal alignment of each person.[28]

Another aim of this study was to evaluate the effect of SOI training on the degenerative state of IVDs and its relation to LS posture. All IVDs of Marines with complete and useful data sets were graded using the Pfirrmann scoring system for IVD degeneration. The incidence of IVD degeneration (at least one degenerated IVD) among these Marines was 47.5%, while the incidence of degeneration at the L5–S1 level was 40.7%. Analysis of a larger data set is needed to examine how the combination of multiple degeneration scores through lumbar levels in a single individual can predict LS postural load carriage behavior. Additionally, the fact that most significant postural changes and higher incidence of IVD degeneration occurred at the inferior LS may be related to the greater forces acting on these levels through the LS [29, 30]. It has been previously suggested that in the presence of IVD degeneration at inferior levels a compensatory mechanism of increased lordosis occurs at superior lumbar levels [9, 31]. However, in the present study, we did not find evidence of this phenomenon.

We found that Marines with degeneration at the L5-S1 level demonstrated larger sacral postural perturbations in response to load as well as a trend of reduced change in lumbar lordosis. Absolute values of SS and LS lordosis of the L5-S1 degeneration group suggest that during load carriage, two postural differences exist compared to the non-degenerated group: 1) sacral inclination is greater, and 2) LS lordosis is retained. Together, these data suggest that overall LS posture (with respect to the ground) is similar in these two groups, but individuals with degeneration achieve that position with more pelvic movement and less lumbar spine deformation. This interpretation of the data during load carriage is counterintuitive to that previously reported when standing without external load in the presence of degeneration. In that case, a more vertical sacrum and reduced LS lordosis were reported.[32] However, in the present study we did not find any indication of these differences while standing without load.

There are a number of limitations to this study. An inherent limitation of *in vivo* MRI studies is the trade-off between voxel dimensions and scan duration. It was imperative to maintain the short scan duration because Marines had to stand still in the scanner while donning load. We have previously demonstrated that the LS posture measured from high-resolution images is not significantly different from those measured from images at the voxel dimensions ($1.43 \times 1.43 \times 4.5$ mm^3) used in this study.[9] However, this resolution does not allow for proper measurement of IVD bulging or protrusion, which would complement our IVD distance measurements. Another constraint of this study was the attrition rate (~35%), which limited the number of complete data sets available for analysis and reduced the power of some of our non-significant findings. A final limitation is that all of our subjects were pain-free at the time of enrollment and graduated from SOI. It is possible that the presence of pain would profoundly alter LS posture in the presence of load—a topic of ongoing research in the laboratory.

3.6 Conclusions

In conclusion, when Marines carry a 22.6kg load in a standard military load carriage system without wearing body armor, there is an observable compensatory forward lean and an overall reduction in the LS lordosis. Locally, L1–L2 becomes more lordotic, L2–L3 does not change, and L3–L4, L4–L5, and L5–S1 become more kyphotic. Moreover, the anterior and central IVD regions of inferior lumbar levels experience compression, while the posterior disc region becomes distracted, leading to postural changes after standing for 45 min with load. The contribution of each intervertebral level is reflected in lumbar spine flexion and reduced lordosis during load-carrying tasks. Additionally, training did not induce further progression of IVD degeneration in any participant of this study. However, Marines with degenerated IVDs at L5–S1 exhibited a larger sacral postural perturbations and smaller lumbar lordosis changes in response to load. These data suggest that LS postural adaptations to load may not be regulated by physical conditioning as much as they are inherent strategies to manage the overall load over the base of support. However, this concept needs to be tested explicitly.

3.7 Acknowledgements

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Table 3−1 Distribution of intervertebral disc degeneration as scored by Pfirrmann grading, by lumbar level.

Level/ Pfirrmann Grade	**I**	**II**	**III**	**IV**	**V**	**Total**
L1-L2	3	23	1	0	0	27
L2-L3	3	22	2	0	0	27
L3-L4	4	19	3	1	0	27
L4-L5	1	21	1	4	0	27
L5-S1	1	15	4	6	1	27
Total	12	100	11	11	1	

Table 3−2 Values for all angular posture measurements in the unloaded and loaded conditions by visit.

Variable/Task	Visit 1		Visit 2		Visit 3	
	Unloaded	**Loaded**	**Unloaded**	**Loaded**	**Unloaded**	**Loaded**
Angle w.r.t. Horizontal	83.27°±5.41°	72.81°±6.80°	82.04°±4.67°	73.23°±7.34°	82.55°±5.54°	72.18°±6.14°
Sacral Slope	33.84°±9.35°	40.73°±9.69°	34.64°±9.41°	39.24°±9.89°	34.81°±7.83°	40.33°±7.23°
Cobb Angle	50.73°±11.12°	43.02°±14.92°	49.12°±9.93°	39.77°±15.90°	52.36°±11.75°	41.22°±13.23°
Superior LS Cobb Angle	7.19°±7.54°	10.18°±12.42°	6.52°±8.01°	6.52°±8.18°	6.90°±6.21°	7.41°±8.013°
Inferior LS Cobb Angle	23.68 °±5.60°	18.22°±7.40°	23.44°±5.33°	17.44°±8.37°	23.83°±5.71°	17.83°±8.38°
IVD L1L2 Angle	5.38°±2.23°	6.17°±2.40°	4.94°±2.38°	5.98°±2.50°	4.82°±1.85°	5.88°±2.78°
IVD L2L3 Angle	7.70°±2.44°	7.78°±2.58°	7.64°±2.46°	7.56°±2.78°	7.26°±2.05°	7.17°±2.94°
IVD L3L4 Angle	9.06°±2.29°	8.22°±2.56°	9.04°±2.12°	8.58°±3.24°	9.74°±1.81°	8.09°±2.92°
IVD L4L5 Angle	10.69°±2.73°	7.04°±4.58°	10.40°±2.59°	7.56°±3.87°	11.41°±2.77°	7.84°±3.98°
IVD L5S1 Angle	11.08°±4.62°	7.10°±3.09°	10.46°±4.00°	6.45°±2.95°	10.96°±4.54°	6.92°±3.90°

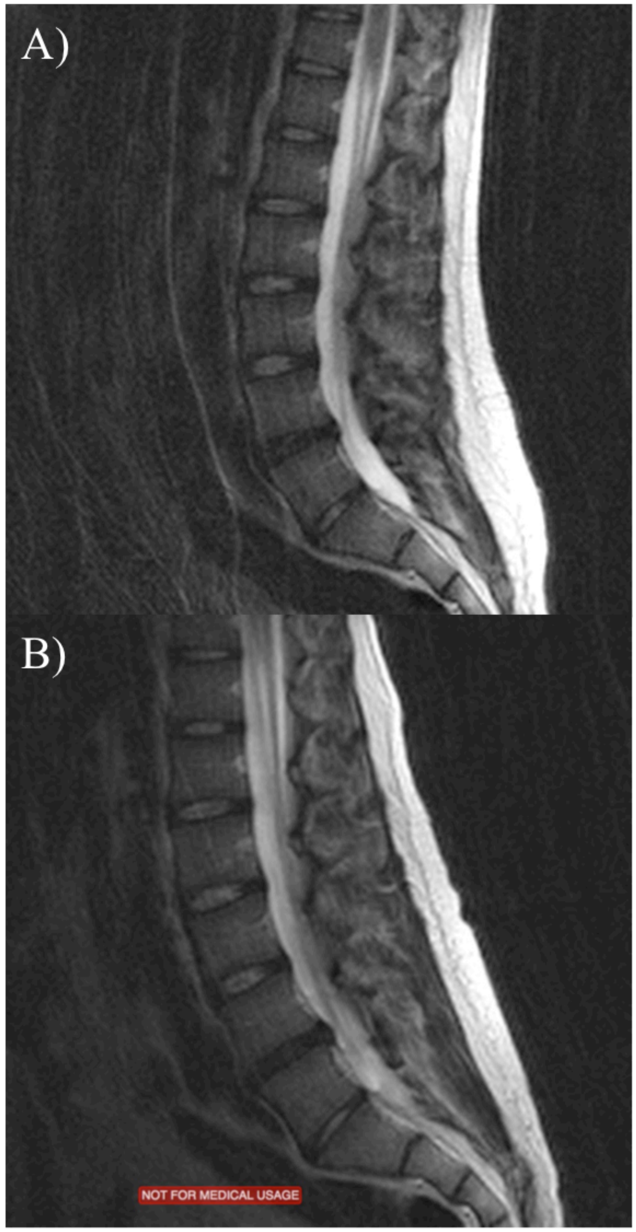

Figure 3−1 Representative sagittal magnetic resonance images of the lumbar spine without load (A) and with load (B).

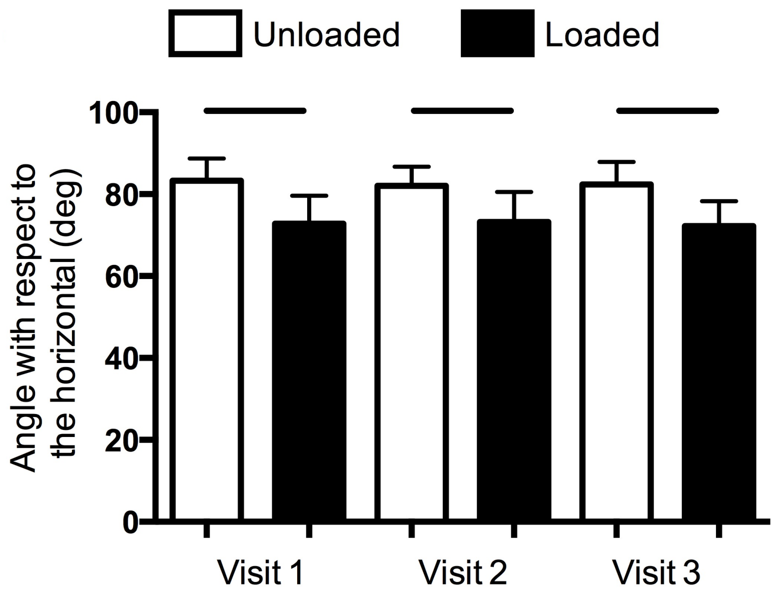

Figure 3−2 Trunk flexion measurements per task and visit. Significant differences (p<0.001) were found between unloaded (white) and loaded (loaded) tasks but not throughout training. Horizontal bars represent statistical difference (*p*<0.05).

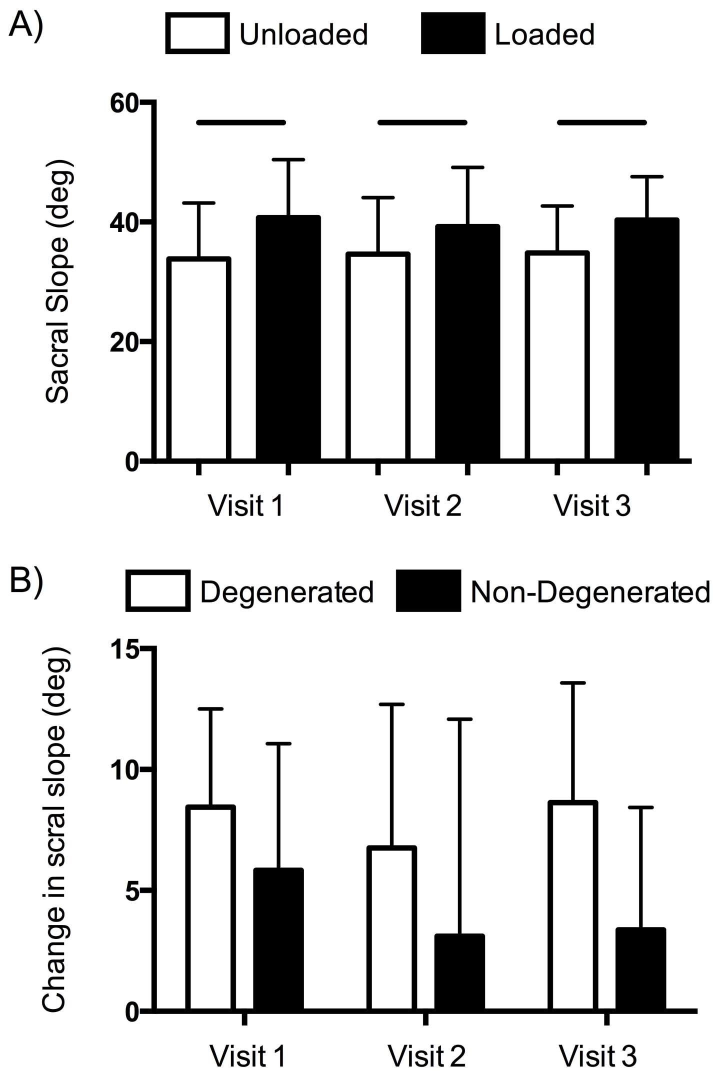

Figure 3‒3 A) Sacral slope (SS) per task and visit. Significant differences (p<0.001) were found between unloaded (white) and loaded (loaded) tasks but not throughout training. B) Change in sacral slope between tasks by L5-S1 IVD degeneration, throughout training. A significant (p<0.05) main effect of degeneration was found: Marines with degeneration had a larger change in SS between tasks. Horizontal bars represent statistical difference (p<0.05).

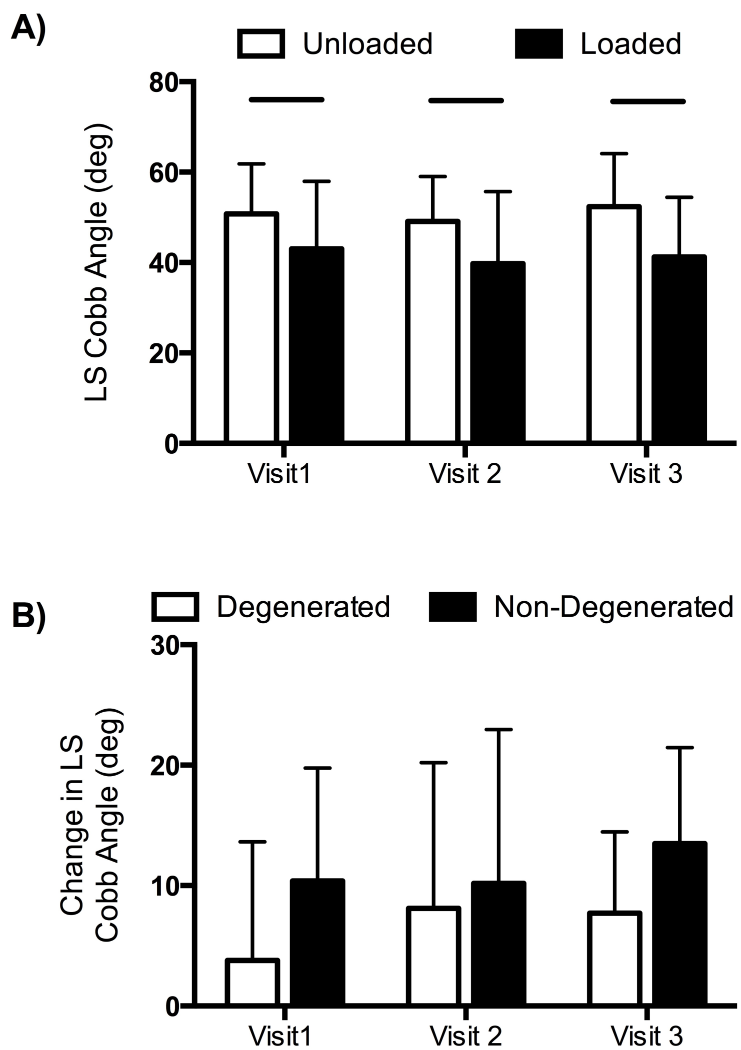

Figure 3−4 A) Results for whole lumbar spine (LS) lordosis per task and visit. Overall LS became straighter during load carriage. B) Change in LS lordosis between tasks by L5-S1 IVD degeneration groups. Horizontal bars represent statistical difference ($p<0.05$).

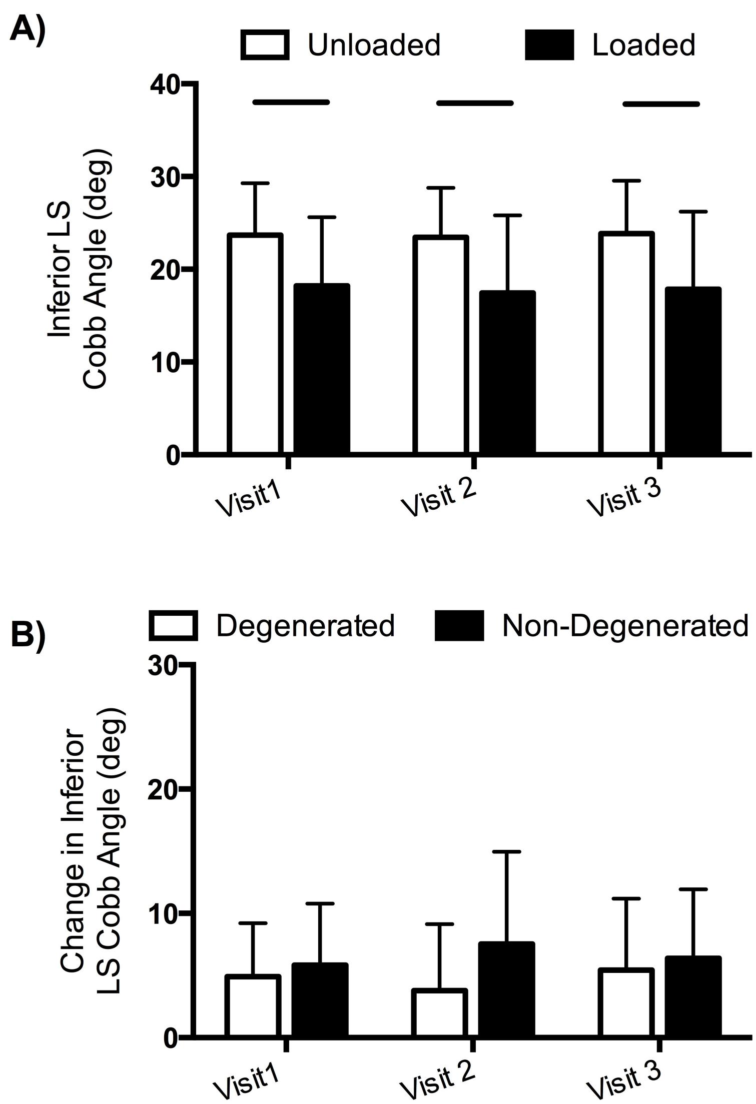

Figure 3‒5 A) Results for inferior lumbar spine (LS) lordosis per task and visit. Inferior LS became straighter during load carriage. B) Change in inferior LS lordosis between tasks by L5-S1 IVD degeneration groups. These data show that postural response to load is driven by changes in the inferior LS. Horizontal bars represent statistical difference ($p<0.05$).

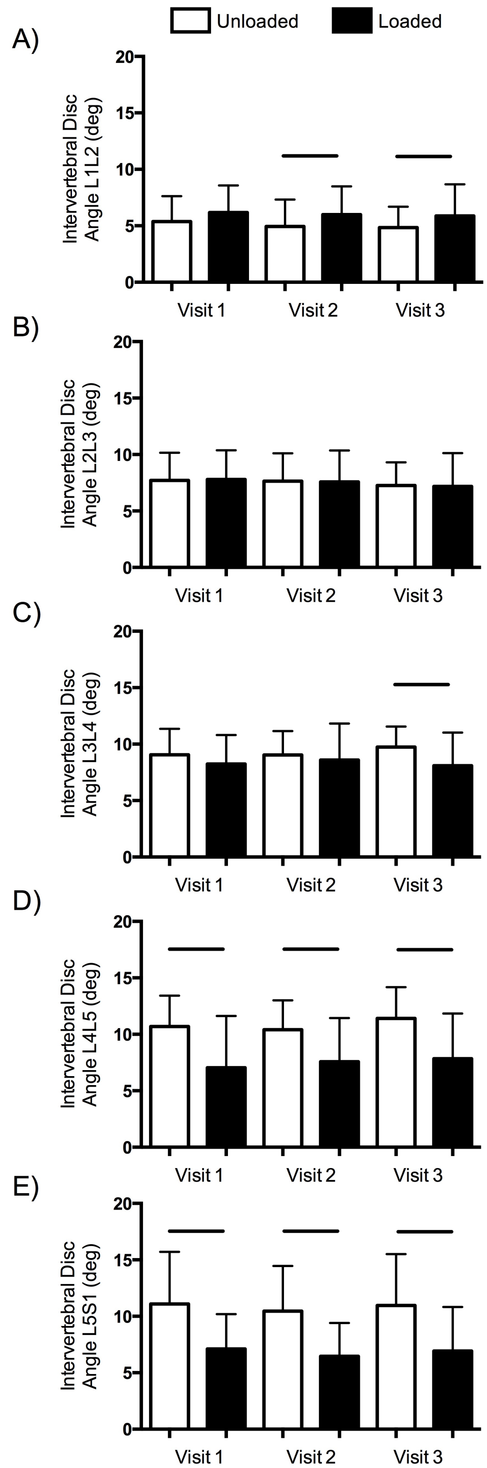

Figure 3−6 Lumbar-level dependent lordosis measurements. Intervertebral disc (IVD) angles (A–E), change in IVD angle in response to load (F–J), and change in IVD angle in response to load per visit for subjects with and without degeneration at the L5–S1 level (K–O) per lumbar level. Horizontal bars represent statistical difference (p<0.05).

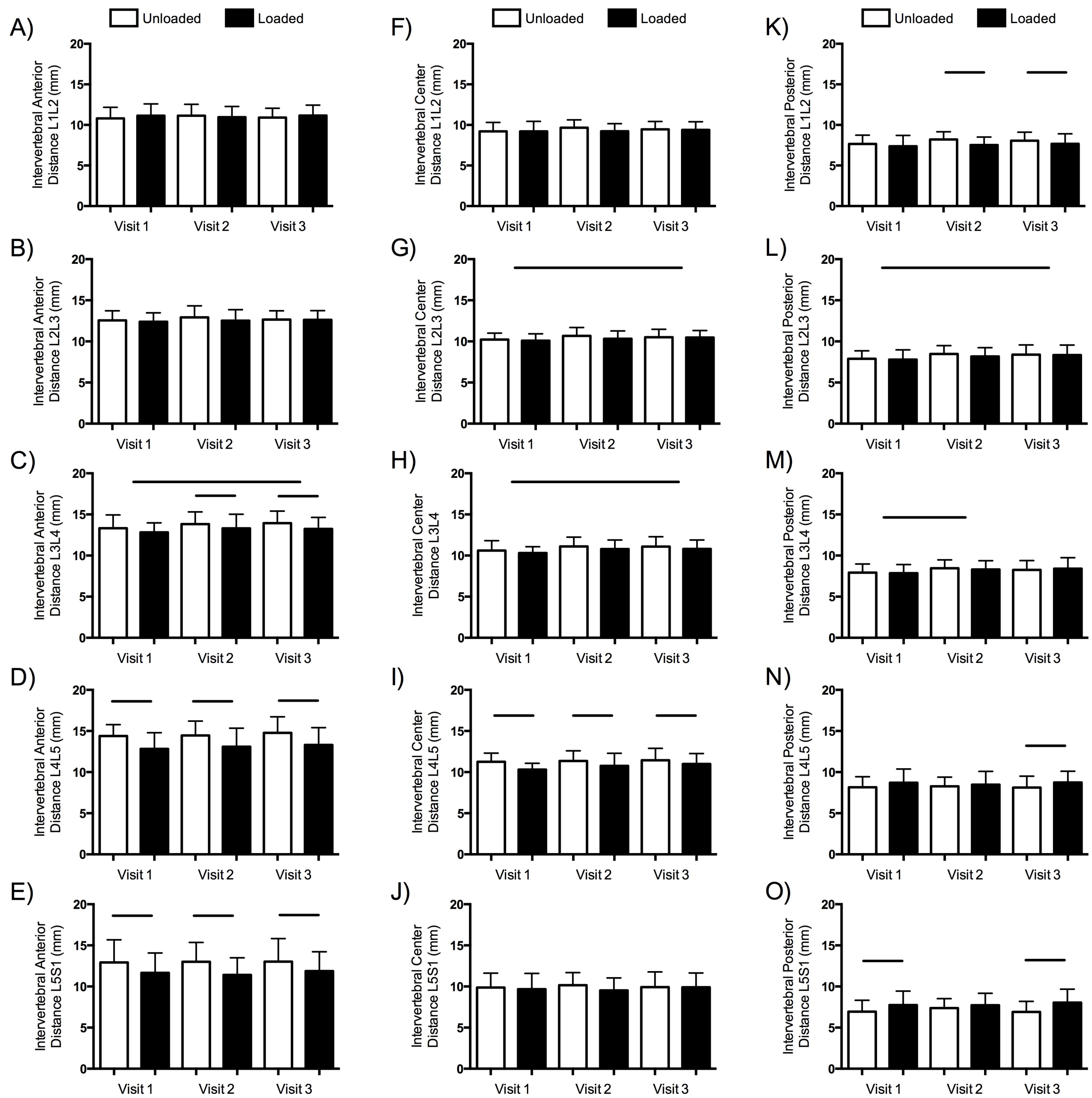

Figure 3−7 Lumbar-level dependent regional intervertebral distances. Anterior (A–E), central (F–J), and (K–O) intervertebral disc (IVD) distances per task and visit. Horizontal bars represent statistical difference ($p<0.05$).

3.8 References

1. Attwells RL, Birrell SA, Hooper RH, Mansfield NJ: **Influence of carrying heavy loads on soldiers' posture, movements and gait**. *Ergonomics* 2006, **49**(14):1527-1537.

2. Heir T, Glomsaker P: **Epidemiology of musculoskeletal injuries among Norwegian conscripts undergoing basic military training**. *Scandinavian journal of medicine & science in sports* 1996, **6**(3):186-191.

3. Knapik J, Harman E, Reynolds K: **Load carriage using packs: a review of physiological, biomechanical and medical aspects**. *Applied ergonomics* 1996, **27**(3):207-216.

4. Taanila H, Suni J, Pihlajamaki H, Mattila VM, Ohrankammen O, Vuorinen P, Parkkari J: **Musculoskeletal disorders in physically active conscripts: a one-year follow-up study in the Finnish Defence Forces**. *BMC musculoskeletal disorders* 2009, **10**:89.

5. Harman EA, Gutekunst DJ, Frykman PN, Nindl BC, Alemany JA, Mello RP, Sharp MA: **Effects of two different eight-week training programs on military physical performance**. *Journal of strength and conditioning research / National Strength & Conditioning Association* 2008, **22**(2):524-534.

6. Knapik JJ, Bahrke M, Staab J, Reynolds K, Vogel J, O'Connor J: **Frequency of Loaded March Training and Performance on a Loaded Road March**. In: *US Army Research Institute of Environmental Medicine.* vol. T13-90. Natick, Mass; 1990.

7. Swain DP, Onate JA, Ringleb SI, Naik DN, DeMaio M: **Effects of training on physical performance wearing personal protective equipment**. *Military medicine* 2010, **175**(9):664-670.

8. Aharony S, Milgrom C, Wolf T, Barzilay Y, Applbaum YH, Schindel Y, Finestone A, Liram N: **Magnetic resonance imaging showed no signs of overuse or permanent injury to the lumbar sacral spine during a Special Forces training course**. *The spine journal : official journal of the North American Spine Society* 2008, **8**(4):578-583.

9. Rodriguez-Soto AE, Jaworski R, Jensen A, Niederberger B, Hargens AR, Frank LR, Kelly KR, Ward SR: **Effect of load carriage on lumbar spine kinematics**. *Spine* 2013, **38**(13):E783-791.

10. Legaye J, Duval-Beaupere G, Hecquet J, Marty C: **Pelvic incidence: a fundamental pelvic parameter for three-dimensional regulation of spinal sagittal curves**. *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 1998, **7**(2):99-103.

11. Vaz G, Roussouly P, Berthonnaud E, Dimnet J: **Sagittal morphology and equilibrium of pelvis and spine**. *European spine journal : official publication of the*

European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society 2002, **11**(1):80-87.

12. Jackson RP, Kanemura T, Kawakami N, Hales C: **Lumbopelvic lordosis and pelvic balance on repeated standing lateral radiographs of adult volunteers and untreated patients with constant low back pain**. *Spine* 2000, **25**(5):575-586.

13. Berthonnaud E, Dimnet J, Roussouly P, Labelle H: **Analysis of the sagittal balance of the spine and pelvis using shape and orientation parameters**. *Journal of spinal disorders & techniques* 2005, **18**(1):40-47.

14. Schwab F, Lafage V, Patel A, Farcy JP: **Sagittal plane considerations and the pelvis in the adult patient**. *Spine* 2009, **34**(17):1828-1833.

15. Berry DB, Rodriguez-Soto AE, Tokunaga JR, Gombatto SP, Ward SR: **An Endplate-Based Joint Coordinate System for Measuring Kinematics in Normal and Abnormally Shaped Lumbar Vertebrae**. *Journal of applied biomechanics* 2015.

16. Pfirrmann CW, Metzdorf A, Zanetti M, Hodler J, Boos N: **Magnetic resonance classification of lumbar intervertebral disc degeneration**. *Spine* 2001, **26**(17):1873-1878.

17. Fujiwara A, Lim TH, An HS, Tanaka N, Jeon CH, Andersson GB, Haughton VM: **The effect of disc degeneration and facet joint osteoarthritis on the segmental flexibility of the lumbar spine**. *Spine* 2000, **25**(23):3036-3044.

18. Cobb JR: **Outline for the study of scoliosis** *American Academy of Orthopaedic Sugeons, Instructional Couse Lectures* 1948, **5**:261-275.

19. Geijer H, Beckman K, Jonsson B, Andersson T, Persliden J: **Digital radiography of scoliosis with a scanning method: initial evaluation**. *Radiology* 2001, **218**(2):402-410.

20. Schmitz A, Kandyba J, Koenig R, Jaeger UE, Gieseke J, Schmitt O: **A new method of MR total spine imaging for showing the brace effect in scoliosis**. *Journal of orthopaedic science : official journal of the Japanese Orthopaedic Association* 2001, **6**(4):316-319.

21. Al-Khabbaz YS, Shimada T, Hasegawa M: **The effect of backpack heaviness on trunk-lower extremity muscle activities and trunk posture**. *Gait & posture* 2008, **28**(2):297-302.

22. Bust PD, McCabe PT: **Proceeding of the International Conference on Contemporary Ergonomics (CE 2005)**. In: *Contemporary Ergonomics 2005: 5-7 April 2005 2005; Hatfield, UK*: Taylor & Francis; 2005.

23. Bloon D, Woodhull-McNeal P: **Postural adjustments while standing with two types of loaded backpack.** *Ergonomics* 1987(30):1425-1430.

24. Birrell SA, Haslam RA: **The effect of military load carriage on 3-D lower limb kinematics and spatiotemporal parameters**. *Ergonomics* 2009, **52**(10):1298-1304.

25. Pascoe DD, Pascoe DE, Wang YT, Shim DM, Kim CK: **Influence of carrying book bags on gait cycle and posture of youths**. *Ergonomics* 1997, **40**(6):631-641.

26. Smith B, Ashton KM, Bohl D, Clark RC, Metheny JB, Klassen S: **Influence of carrying a backpack on pelvic tilt, rotation, and obliquity in female college students**. *Gait & posture* 2006, **23**(3):263-267.

27. Neuschwander TB, Cutrone J, Macias BR, Cutrone S, Murthy G, Chambers H, Hargens AR: **The effect of backpacks on the lumbar spine in children: a standing magnetic resonance imaging study**. *Spine* 2010, **35**(1):83-88.

28. Herkowitz HN, International Society for Study of the Lumbar Spine.: **The lumbar spine**, 3rd edn. Philadelphia: Lippincott Williams & Wilkins; 2004.

29. Alexander LA, Hancock E, Agouris I, Smith FW, MacSween A: **The response of the nucleus pulposus of the lumbar intervertebral discs to functionally loaded positions**. *Spine* 2007, **32**(14):1508-1512.

30. Pal GP, Routal RV: **Transmission of weight through the lower thoracic and lumbar regions of the vertebral column in man**. *Journal of anatomy* 1987, **152**:93-105.

31. Lee ES, Ko CW, Suh SW, Kumar S, Kang IK, Yang JH: **The effect of age on sagittal plane profile of the lumbar spine according to standing, supine, and various sitting positions**. *Journal of orthopaedic surgery and research* 2014, **9**(1):11.

32. Barrey C, Jund J, Noseda O, Roussouly P: **Sagittal balance of the pelvis-spine complex and lumbar degenerative diseases. A comparative study about 85 cases**. *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 2007, **16**(9):1459-1467.

CHAPTER 4 : THE EFFECT OF LOAD MAGNITUDE AND DISTRIBUTION ON LUMBAR SPINE POSTURE IN ACTIVE-DUTY MARINES

4.1 Abstract

Low back pain has been associated with heavy load carriage among military personnel. Although there are data indicating that lumbar spine (LS) posture changes in response to load, there no data documenting how the magnitude or distribution of these loads impact postural alignment or disc compression in detail. Therefore, the purpose of this study was to quantify the effect of operationally relevant loads and distributions on LS postures in a group of active-duty Marines. We hypothesized that when loads are evenly distributed anterior and posteriorly, the deviation from the unloaded standing posture would be small. Additionally, we hypothesized that as load magnitude increases lumbar lordosis would decrease and lumbar flexion would increase, consistent with previous observations. Active-duty Marines (n=12) were scanned standing unloaded and while standing and carrying 22, 33 and 45kg of load distributed both 50%-50% and 20%-80% anteriorly and posteriorly using an upright magnetic resonance imaging (MRI) scanner. Images were used to measure LS and pelvic postures. Two-way repeated-measures ANOVA and Sidak *post-hoc* tests were used to compare posture as a function of load magnitude and distribution (α=0.05). No changes in posture were induced by any load magnitude carried in the 50%-50% configuration. When load was carried in the 20%-80% configuration lumbar flexion increased as a result of sacral anterior rotation and reductions in overall lumbar lordosis. This pattern was greater as load was increased between 22 and 33kg, but did not increase further between 33 and 45kg. When the

superior and inferior LS regions were compared, the inferior LS became uniformly less lordotic, regardless of load magnitude. However, the superior LS became progressively more lordotic as function of load in the 20%-80% configuration. Interestingly, changes in intervertebral distances at inferior lumbar levels varied in magnitude throughout regions of the disc. Postural adaptations were found only when load was carried in the 20%-80% configuration, which suggests that load-carriage limits based on "at-risk" postural changes are likely only relevant when loads are non-uniformly distributed. Although the tendency would be to interpret that loads should be carried symmetrically (50%-50%) to protect the spine, this should be done with caution, as the relationship between postural changes and injury are not yet clear. Furthermore, the operational efficiency of a Marine carrying load in this configuration needs to be tested.

4.2 Introduction

Back problems represent a major health and economic burden among military personnel, as they are the primary cause for medical encounters and lost work time.[1] For example, the incidence of moderate or severe low back pain (LBP) after a one year deployment to Afghanistan is around 22%, resulting in a reduction in life quality indicators.[2] Among soldiers, a self-reported cause of injury questionnaire revealed that the relative risk of developing LBP increased as a function of the magnitude of the carried load.[2] Military personnel carry loads of up to 68kg depending on duty position and nature of the mission.[3] The equipment and supplies that make up these loads are necessary to maintain soldiers' safety and to successfully fulfill their missions. Consequently, in order to maximize these operational outcomes, the effect of heavy load carriage on energy expenditure, situational awareness and combat readiness has been extensively studied.[4-6] Contrastingly, data on the relative position of the musculoskeletal components of the lumbar spine (LS) during load carriage is limited. In previous work from our group, we quantitatively described the deformation of the LS when carrying 50.8kg of load in a group of active-duty Marines with validated methods.[7, 8] These data indicated that heavy load carried with a posterior bias induced lumbar flexion and forward trunk lean. Other authors have described LS postural changes when carrying backpacks with increasing load magnitudes in the 10-30% of body weight (BW) range. However, these data were measured from an adolescent population using different loads and packs, which may not be representative of military personnel and operational conditions. Detailed postural adaptations of the LS in response to varying load magnitudes and anterior-posterior distributions have not been systematically studied in a military population.

This information may allow scientists to identify potential LS injury mechanisms associated with load carriage and allow the military to develop load carriage limits based on measurable changes in LS posture induced by operationally relevant loads. Both pieces of information may inform best practices to minimize LS injuries.

Therefore, the purpose of this study was to quantify the effect of operationally relevant loads and distribution on LS posture in a group of active-duty Marines. We hypothesize that when loads are evenly distributed anterior and posteriorly, the deviation from the standing posture is small compared to loads carried with a posterior bias. Additionally, we hypothesize that as load magnitude increases whole lumbar lordosis will decrease and lumbar flexion will increase.

4.3 Materials and Methods

Subjects

Twelve active-duty Marines volunteered to participate in this study. The University of California, San Diego and Naval Health Research Center institutional review boards approved this study, and all volunteers provided oral and written informed consent.

Imaging

Marines were scanned using an upright 0.6T magnetic resonance imaging (MRI) scanner (UPRIGHT® Multi-Position MRI, Fonar Corporation, Melville, NY, USA) and flexible planar coil. A soft sleeve was used to retain the coil behind the volunteer's back at the lumbar spine (L1–S1) level while standing without external load. The sleeve was tight enough to keep the coil in place yet loose enough not to alter the volunteer's natural standing position. During scans with load, the coil was placed between the volunteer's back and load carriage system. The bottom of the coil was carefully placed above the height of the sacroiliac joint. A three-plane localizer and sagittal T2-weighted images (repetition time 1974 msec; echo time 160 msec; field of view 32 cm; 224×224 acquisition matrix; 1.56×1.56 mm^2 pixel size; 4.5mm slice thickness; 0.5 mm gap; number of averages 1, scan duration 2 minutes 30 seconds) were acquired.

Load-Carrying Tasks

Marines were scanned standing unloaded and while carrying 22, 33 and 45kg of load distributed both 50%-50% and 20%-80% anteriorly and posteriorly (AP), respectively. The first scan was always standing unloaded and the other 6 scans were randomized for all participants. Marines were purposefully not given instructions on how to stand in the scanner, but they were instructed to remain still during the MRI acquisition.

Load magnitudes of 22 and 33kg were selected because they are the recommended load carriage limits for fighting and approach march loads, respectively.[6] Additionally, 45kg is on the lower end of sustainment loads carried by Marines during dismounted operation in Afghanistan in 2003.[9] The 50-50% and 20-80% AP load distributions were selected based on preliminary data (not shown) indicating that when loads are light (i.e. average 12.28kg) they are carried balanced both in the anterior-posterior and left-right directions. Moreover, it has been previously hypothesized that evenly distributed load induces minimal postural changes in the LS; therefore, we tested this concept. Conversely, heavier loads are typically carried using a backpack, during which a larger fraction of the load is located posteriorly. This load carriage paradigm has been reported to induce postural changes that deviate from the standing posture. Typically, load carried anteriorly is in the form of small pouches containing gear attached to belts and body armors. For that reason, Marines wore a body armor, which was considered part of the load carriage system.

Data Analysis

Each image data set was analyzed as previously described [7]. Briefly, a set of markers was manually placed at the corners of each vertebra (L1–S1) on all sagittal images, and on posterior elements on a single axial image per lumbar level. These data were used to describe vertebral endplate position and orientation. Relative rotations in the axial and coronal planes between contiguous vertebrae were removed, and the resulting vertebral end-plate representations were used to generate postural measurements in the sagittal plane.

Measurements

Postural measurements of the LS in the sagittal plane were generated from vertebral endplates as previously described in the *Data Analysis* section. Angle with respect to the horizontal was defined as the angle between the centroid of L1, S1 and the horizontal line; it quantifies LS flexion, but does not convey postural information among LS levels. A relationship between lumbar spine posture and *sacral slope (SS)* has been previously reported in the literature when standing.[10, 11] Therefore, in order to estimate the contribution of the pelvis to load carriage postural adaptations the SS was also measured. Sacral slope is defined as the angle between the superior endplate of S1 and the horizontal and it describes the orientation of the sacrum. Unfortunately, the images field of view was not large enough to include the hip joint, while maintaining the acquisition duration under 3 minutes.

Lumbar lordosis was measured using Cobb angle, defined it as the angle formed by the planes corresponding to the superior endplates of L1 and S1 in the sagittal plane. [12-14]. Due to the disparate behavior among the superior and inferior LS during load carriage, we defined the superior sagittal Cobb angle as the angle formed by the superior endplate of L1 and the inferior endplate of L3, and the inferior sagittal Cobb angle as the angle between the inferior endplates of L3 and S1 [7].

Intervertebral angles and regional disc heights were measured between the planes of the inferior and superior endplates of adjacent vertebrae. Intervertebral heights were measured as the shortest distance between inferior and superior endplates anteriorly, centrally, and posteriorly in the midsagittal plane.

Postural adaptations to load carriage have been hypothesized to realign the center of mass (CoM) (soldier+backpack) over the base of support. Determining the CoM of the system requires knowledge of the position and mass of the body segments.[15] This information was not available in this experiment as only the lumbar spine was imaged and the use of x-rays was not approved in this population. However, it has been shown that during static activities the location of the CoP and CoM with respect to the base of support are distinct, but highly correlated.[15] Therefore, the center of pressure (CoP) was measured using a pressure mat (Tekscan Inc., South Boston, MA). Ideally, these measurements would be made during MRI acquisition; however, due to the ferromagnetic components of the mat, this was not possible. Alternatively, a mock scanner with the same dimensions as those of the upright MRI scanner was built and the pressure mat was placed between the structure walls. After each MRI acquisition Marines were asked to step on the mat and stand still for one minute during data collection, while still carrying the load. A minimum bounding box (MBB, Fig. 4−1A) is defined as the smallest rectangle that can fit all points of a determined dataset in space, in this case, the footprints on the pressure mat.[16] This analysis allowed to account for the differences in feet position (between and within subjects) and footprint shape (between subjects). The location of the CoP was expressed as the fraction of length and width of the MBB around the footprints. We have defined the left posterior corner of the MBB as the origin.

Statistical Analysis

All variables were compared using two-way repeated-measures analyses of variance (ANOVA) with Sidak *post-hoc* tests to identify significant differences as a function of load magnitude and configuration. The comparison between each load carriage configuration (50%-50% or 20%-80%) and the unloaded condition were identified using one-way repeated-measures ANOVA with Sidak *post-hoc* tests. The threshold for significance (α) was set at 0.05 for all analyses. Statistical analyses were performed using SPSS Statistics software (version 20.0, IBM, Armonk, NY), and all data are reported as mean ± standard deviation (SD) values.

4.4 Results

Volunteer Characteristics

A total of 12 active-duty Marines (mean±SD age, 23.41±4.71 years; age range 19–35 years; height, 177.8±5.41 cm; weight, 76.77±11.32 kg; body mass index, 24.15±2.19 kg/m^2) were scanned. The 22, 33 and 45kg loads were 32.68±4.82%, 34.85±5.14% and 37.27±5.48% of BW, respectively. The average time of service of this group was 48±39.96 months (range 10-120 months) and their occupations were 1 infantry officer, 8 riflemen, 2 machine gunners, 2 infantry assault men, and 1 infantry unit leader.

Measurement of the CoP Location

The average location of the CoP along the width (left to right) of the BMM was 46.78±4.91% and 46.38±6.06% along the height (posterior to anterior). There was no significant difference in the location of the CoP between load magnitudes and configurations. The variation of the location of the CoP during the different loaded conditions was less than 5% of that when standing without load (Fig. 4−1B).

Measurement of Lumbar Spine Load-Carriage Postural Changes

The effect of load carriage on LS posture is both magnitude and distribution dependent ($p<0.05$). Loads carried in the 50%-50% configuration did not have an effect on LS flexion. Contrastingly, the overall position of the LS was significantly more horizontal in the 20%-80% configuration compared to standing unloaded (82.28±4.14°; Fig. 4−2, solid bars). More specifically, these postural changes were significant different only when carrying 33 and 45kg, but not when carrying 22kg (75.23±7.79°, Fig. 4−2, asterisks). Interestingly, lumbar flexion values when carrying 33 and 45kg were not significantly different from each other (64.77±7.91° and 62.62±9.36°).

Sacral slope measurements had a similar response to load magnitude and configuration as lumbar flexion ($p<0.05$). In general, when loads were carried in the 20%-80% configuration the orientation of the sacrum became more horizontal (Fig. 4−3, solid bars). However, only 33 and 45kg load magnitudes had a significant effect on sacrum orientation compared to standing without external load (34.29±6.59°) and were not different from each other (46.40±6.40° and 50.76±8.35°; Fig. 4−3, asterisks).

Whole LS lordosis (L1-S1) was also influenced by both load magnitude and configuration ($p<0.05$); however, *post-hoc* tests revealed no differences between load magnitudes (Fig. 4−4A). Overall, the LS became less lordotic ($p<0.05$) when carrying load in the 20%-80% configuration. Surprisingly, only when carrying 22kg lumbar lordosis deviated from that of standing without external load (Fig. 4−4A, asterisks). In previous work, we showed that superior and inferior LS have different kinematic behavior during load carriage; therefore, lordosis in these two regions was measured. [7] Both load magnitude and configuration had a significant effect on superior LS lordosis

($p<0.05$). *Post-hoc* tests revealed that loads carried in the 50%-50% configuration did not have a significant effect on superior or inferior LS lordosis (Fig. 4−4B, clear bars). On the other hand, superior LS became more lordotic ($p<0.05$) when carrying 33 and 45kg in the 20%-80% configuration (17.80±6.28° and 16.88±5.49°) compared to the standing unloaded (10.49±5.18°, Fig. 4−4B, asterisks). The lordosis of the inferior LS was found to be affected solely by load configuration ($p<0.05$). Interestingly, the inferior LS became straighter (~10°) regardless of load magnitude when load was carried in the 20%-80% configuration (Fig. 4−4C, solid bars).

In agreement with the observed changes in regional lordosis, load magnitude had a significant effect on superior levels (L1−L2 and L2−L3), but load configuration influenced inferior levels (L3−L4 to L5−S1, Fig. 4−5). More specifically, load magnitude had a significant effect ($p<0.05$) on L1-L2 IVD. It became more lordotic as load increased, again, no differences were found between 33 and 45kg. At the L2-L3 level, both load magnitude and configuration had an effect on local lordosis ($p<0.05$), however, *post hoc* tests revealed no differences between load magnitudes. A significant interaction between load magnitude and distribution ($p<0.05$) was found at the L3-L4 level. These data revealed that differences between load distributions were observed at the 33 and 45kg loads at this lumbar level. Similarly to the behavior described by the inferior LS lordosis, only the effect of configuration was significant at L4-L5 and L5-S1 levels ($p<0.05$). Both levels became less lordotic in the same amount regardless of load magnitude, but in a configuration dependent manner.

Similar results were observed for changes in regional IVD distances, which reflect postural changes in IVD angle throughout lumbar levels (Fig. 4−7). For example, when a functional spinal unit became less lordotic in response to load carriage, anterior IVD distances decreased and posterior IVD distances increased. This was the case at L4-L5 and L5-S1, it is worth mentioning that at L4-L5 the magnitude of the change in anterior distance appears to be larger than the posterior change, while at the L5-S1 level the magnitude of change was larger at the posteriorly than anteriorly.

4.5 Discussion

The objective of this study was to measure the postural changes of the LS in response to loads carried in operationally relevant magnitudes and configurations. We hypothesized that the postural deviation of the LS when carrying load in a posterior bias would be larger than that when the load was evenly distributed anteriorly and posteriorly. Our results showed that in fact, regardless of load magnitude, the LS posture was not different from that when standing without external load when carried in the 50%-50% configuration. We also hypothesized that as load magnitude increased, so would lumbar flexion. Overall, when load was carried with a posterior bias, lumbar flexion progressively increased. Interestingly, when a load of 22kg was carried in this configuration, the LS posture was not significantly different from that of standing unloaded. Suggesting that when load is carried mostly posteriorly, the load magnitude that significantly alters LS posture is between 22 and 33kg. This is relevant because these two load magnitudes are the recommended load carriage limits for march and approach loads.[6] Furthermore, the lack of differences between 33 and 45kg suggests a postural adaptation plateau, indicating the contribution of active components of the musculoskeletal system to maintain the load carriage posture.

The observed increased trunk flexion appears to result from the contribution of two postural mechanisms: anterior rotation of the sacrum and reduced whole lumbar lordosis. Both SS and lumbar lordosis have been previously measured during load carriage using motion capture and springs.[17-21] However, in the present study we have measured these variables directly. Anterior rotation of the pelvis has been previously reported during walking[17] and static[22, 23] load carriage. These data have been reported as linear displacements and range of motion making it impossible to compare to our results. Additionally, Birrell et al showed that maximum pelvic rotation linearly increased as a function of load magnitude during gait up to 32 kg, which was the heaviest load measured in their study.[22]

In a previous work from our group Marines carried 50.8kg of load mostly posteriorly and found that lumbar lordosis was reduced. In the present study, our results show that lumbar lordosis was significantly decreased only when carrying the lightest load and was not different when carrying the heavier loads. We attribute this partially to the difference in load magnitude (~6kg) when carrying 33 and 45kg, but mostly to the disparity in load distribution. Here, for a total load of 45kg in the 20%-80% configuration 36kg were carried posteriorly, which is much smaller compared to 50.8kg carried mostly posteriorly. This emphasizes the importance of load distribution, suggesting that careful attention should be given to this parameter in load carriage recommendations.

In order to understand the several components of the overall LS posture, we measured regional and local lordosis. In agreement with our previous work, we found that when load is carried with a posterior bias superior and inferior LS have opposite postural adaptations to load.[7] Surprisingly, inferior LS lordosis was reduced by the same amount (~10°) independently of load magnitude. Simultaneously, the overall orientation of the LS was also more horizontal. This may suggest that it is the orientation of the inferior lumbar spine that determines the response of the superior LS to load, which increased as a function of load. We propose that this postural adaptation aims to maintain the rest of the trunk and head in a vertical position.

Intervertebral disc angles revealed that when standing without external load, lumbar lordosis increases caudally, peaking at L4-L5, however during load carriage, this behavior is reversed and lumbar lordosis increased cranially and peaks at L3-L4. In agreement with our previous work, the L3-L4 lumbar level behaved as a transition level between the superior and inferior LS. These data are comparable because in both studies Marines carried loads while wearing body armor. Furthermore, we compared local lordosis when carrying ~22kg in the 20%-80% distribution from the present study to another work from our group where Marines carried the same load in a similar configuration without body armor. We found that at this load magnitude, the use of body armor did not have an effect on LS local lordosis. However, the effect of body armor at heavier loads remains to be investigated as it may alter both trunk and LS kinematics during load carriage.

4.6 Conclusions

In conclusion, when Marines carry load in the 50%-50% configuration, no postural changes were detected. However, the interpretation of these data should be limited because whether this means that this load distribution is more protective (or has a less negative impact on) of the musculoskeletal system compared to a posteriorly biased distribution is unknown. During load carriage with a posterior bias, the overall position of the LS with respect to the ground was more horizontal, this posture resulted from anterior pelvic rotation and overall reduced lumbar lordosis; however, the LS remained in flexion at all lumbar levels. Further research is needed to investigate the adaptation of other musculoskeletal tissues to different degrees of lordosis and how the effect might throughout lumbar levels, as they are morphologically different. These data would allow narrowing down potential mechanisms of injury due to load carriage and to adjust physical training to further prevent low back-related injuries.

4.7 Acknowledgements

Chapter 4 is original to this dissertation. The work is being prepared for publication and will be submitted to the journal Spine. This project was founded by the U.S. Army Medical Research Acquisition Activity, Award No. W81XWH-13-2-0043, under Work Unit No. 1310. The views expressed in this article are those of the authors and do not necessarily reflect the official policy or position of the Department of the Navy, Department of Defense, or the U.S. Government. Approved for public release; distribution is unlimited. U.S. Government Work (17 USC 105). Not copyrighted in the U.S. This research has been conducted in compliance with all applicable federal regulations governing the protection of human subjects in research (Protocol NHRC.2014.0021).

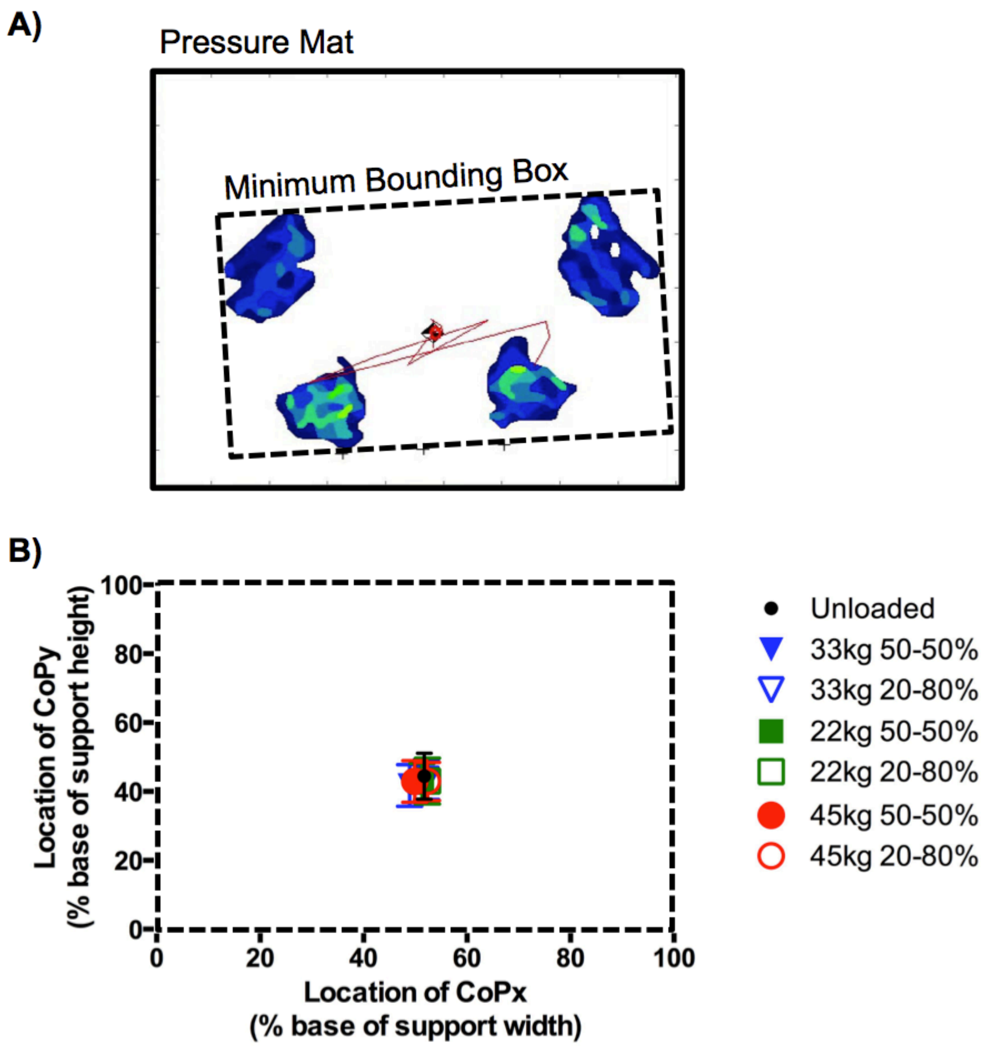

Figure 4–1 A) Representative image showing footprints acquired using pressure mat (solid rectangle) indicating minimum bounding box (MBB, dashed rectangle). The center of pressure (CoP) trajectory is shown in red. Red circle indicates the average location of the CoP. B) Plot of the location of the CoP as a percentage of the width and height of the MBB. No differences were found across load magnitudes and distributions.

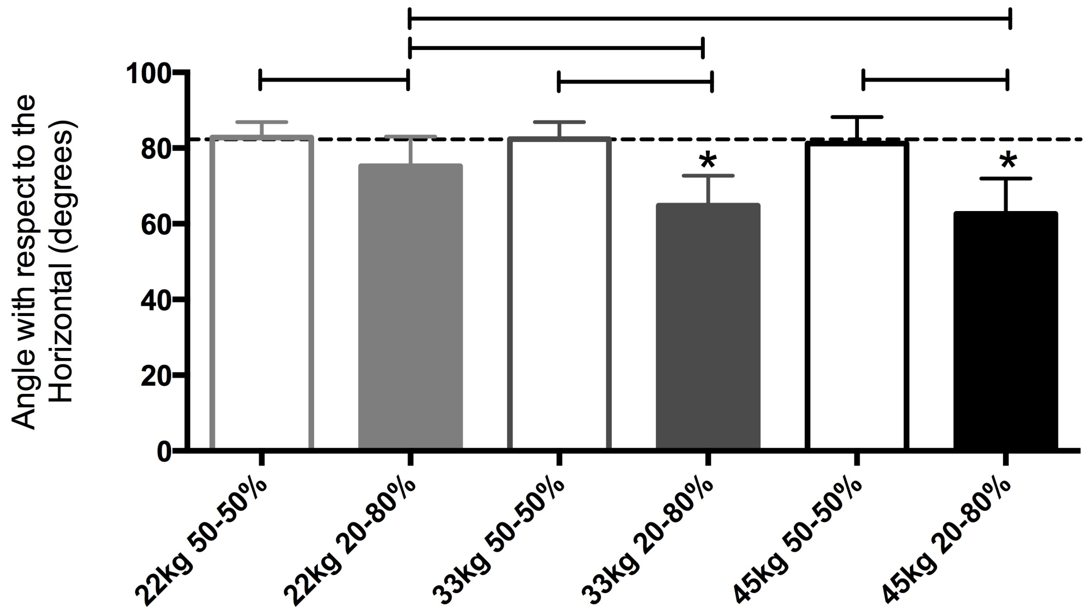

Figure 4–2 Lumbar spine flexion results for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22kg (clear grey), 33kg (dark grey) and 45kg (black). The dashed line represents trunk flexion when standing without external load. Solid horizontal bars represent significant differences ($p<0.05$) between load magnitudes and configurations. Asterisks represent significant differences when compared to the standing unloaded position.

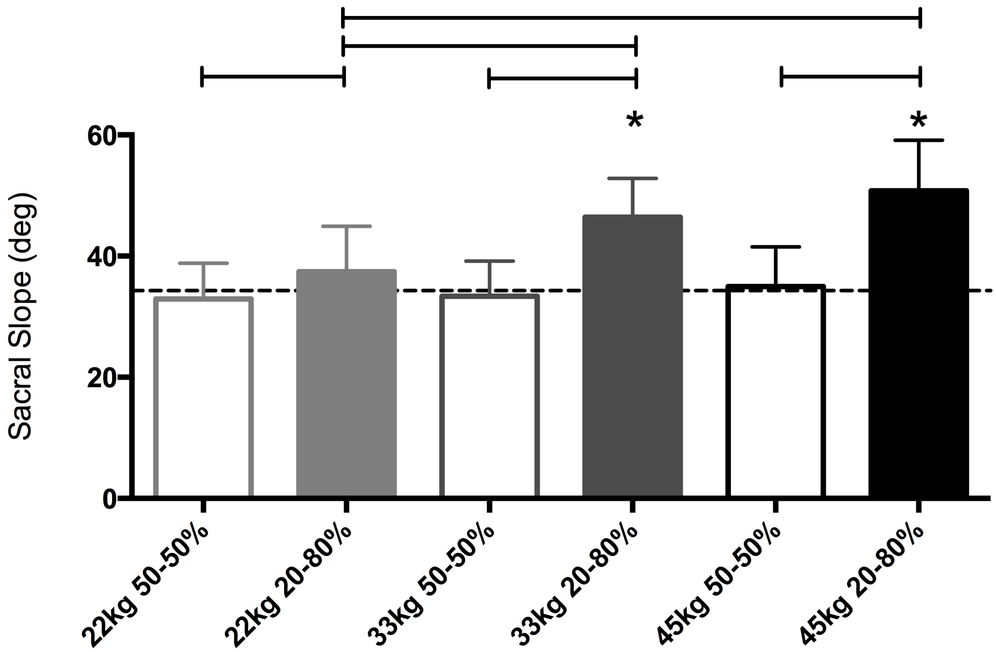

Figure 4–3 Sacral orientation results for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22kg (clear grey), 33kg (dark grey) and 45kg (black). The dashed line represents sacral orientation when standing without external load. Solid horizontal bars represent significant differences ($p<0.05$) between load magnitudes and configurations. Asterisks represent significant differences when compared to the standing unloaded position.

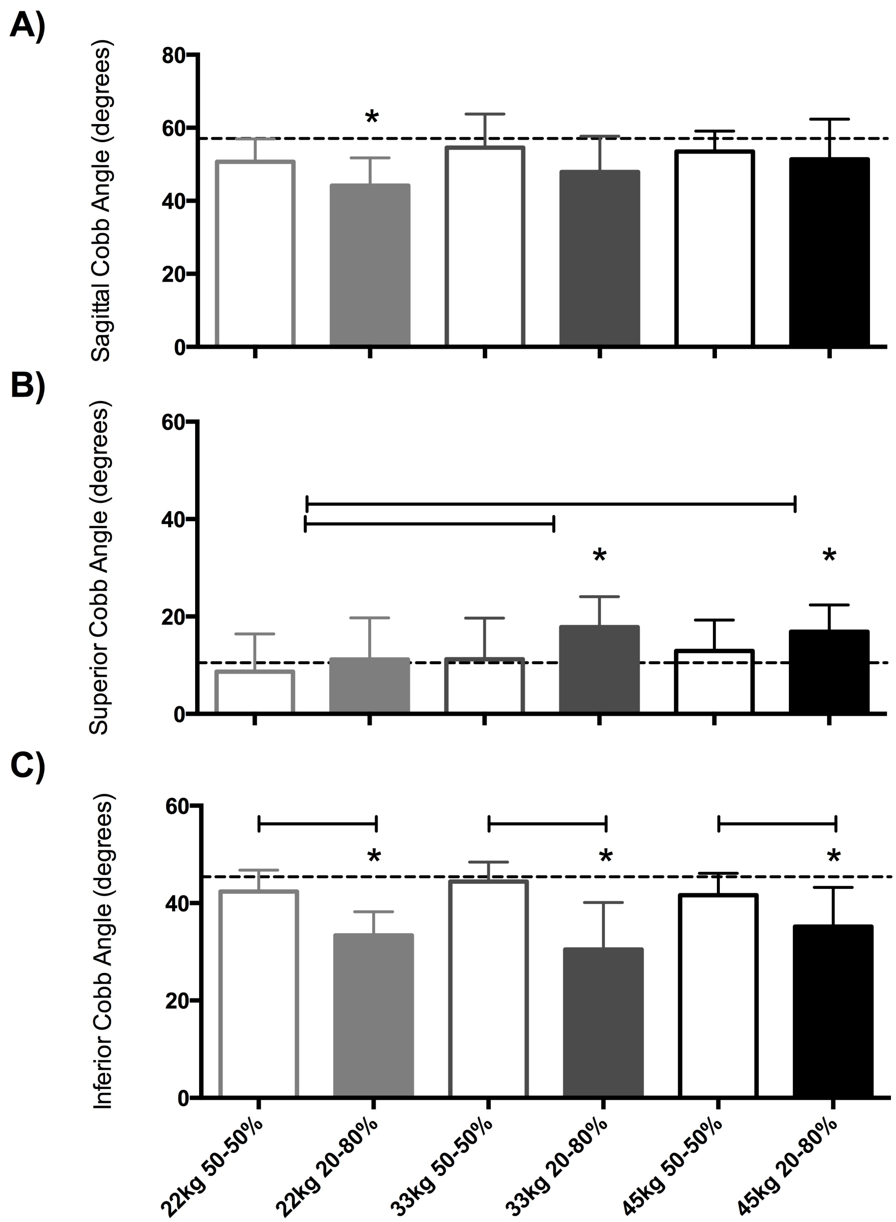

Figure 4–4 A) Whole, B) superior, and C) inferior lumbar lordosis results for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22kg (clear grey), 33kg (dark grey) and 45kg (black). The dashed line represents lordosis when standing without external load. Solid horizontal bars represent significant differences ($p<0.05$) between load magnitudes and configurations. Asterisks represent significant differences when compared to the standing unloaded position.

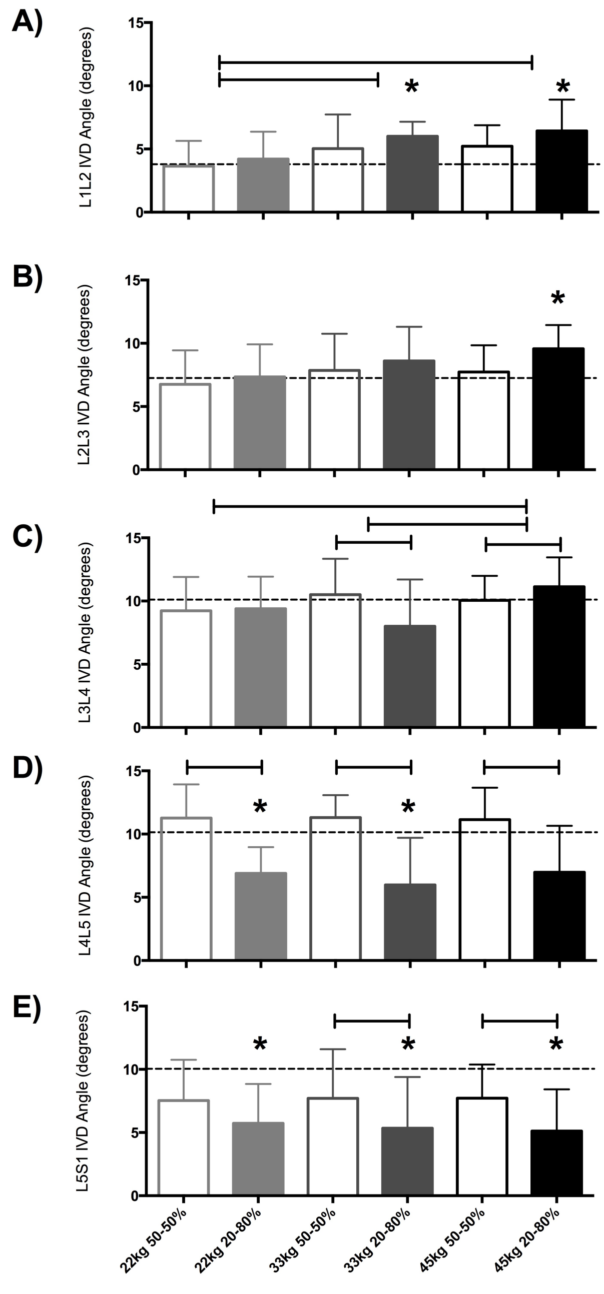

Figure 4–5 Lumbar lordosis results at A) L1-L2, B) L2-L3, C) L3-L4, D) L4-L5 and E) L5-S1 for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22kg (clear grey), 33kg (dark grey) and 45kg (black). The dashed line represents lordosis when standing without external load. Solid horizontal bars represent significant differences ($p<0.05$) between load magnitudes and configurations. Asterisks represent significant differences when compared to the standing unloaded position.

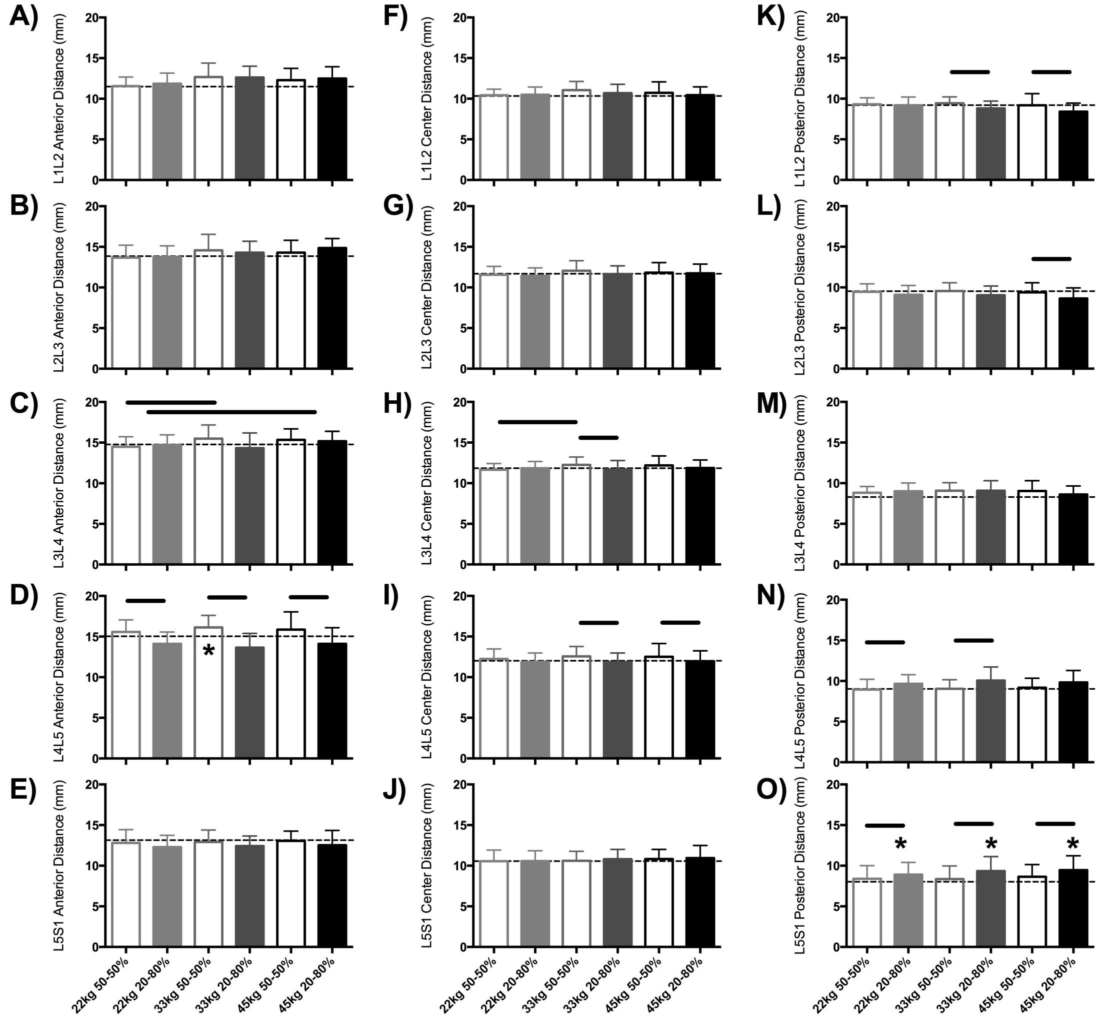

Figure 4–6 Anterior (A-E), central (F-J), and posterior (K-O) intervertebral distances at L1-L2, L2-L3, L3-L4, L4-5 and L5-S1 (from top to bottom) for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22kg (clear grey), 33kg (dark grey) and 45kg (black). The dashed line represents lordosis when standing without external load. Solid horizontal bars represent significant differences ($p<0.05$) between load magnitudes and configurations. Asterisks represent significant differences when compared to the standing unloaded position.

4.8 References

1. Army Medical Surveillance Activity. **Low Back Pain, Active Component, U.S. Armed Forces, 2000-2009**. MSMR. 2010;17(6):2-7.

2. Roy TC, Lopez HP, Piva SR: **Loads worn by soldiers predict episodes of low back pain during deployment to Afghanistan**. *Spine* 2013, **38**(15):1310-1317.

3. Knapik JJ, Reynolds KL, Harman E: **Soldier load carriage: historical, physiological, biomechanical, and medical aspects**. *Military medicine* 2004, **169**(1):45-56.

4. Knapik JJ, Harman EA, Steelman RA, Graham BS: **A systematic review of the effects of physical training on load carriage performance**. *Journal of strength and conditioning research / National Strength & Conditioning Association* 2012, **26**(2):585-597.

5. US Army Development and Employment Agency (ADEA): **Report on the ADEA Soldier's Load Initiative**. Ft Lewis, WA, USA; 1987.

6. Field Manual 21-18, **Foot Marches**. Washington, DC: Department of the Army; 1990.

7. Rodriguez-Soto AE, Jaworski R, Jensen A, Niederberger B, Hargens AR, Frank LR, Kelly KR, Ward SR: **Effect of load carriage on lumbar spine kinematics**. *Spine* 2013, **38**(13):E783-791.

8. Berry DB, Rodriguez-Soto AE, Tokunaga JR, Gombatto SP, Ward SR: **An Endplate-Based Joint Coordinate System for Measuring Kinematics in Normal and Abnormally Shaped Lumbar Vertebrae**. *Journal of applied biomechanics* 2015.

9. Dean CD, FJ; : **The Modern Warrior's Combat Load**. In. Fort Leavenworth, KS, USA: Center for Army Lessons Learned Report; 2008.

10. Legaye J, Duval-Beaupere G, Hecquet J, Marty C: **Pelvic incidence: a fundamental pelvic parameter for three-dimensional regulation of spinal sagittal curves**. *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 1998, **7**(2):99-103.

11. Vaz G, Roussouly P, Berthonnaud E, Dimnet J: **Sagittal morphology and equilibrium of pelvis and spine**. *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 2002, **11**(1):80-87.

12. Cobb JR: **Outline for the study of scoliosis** *American Academy of Orthopaedic Sugeons, Instructional Couse Lectures* 1948, **5**:261-275.

13. Geijer H, Beckman K, Jonsson B, Andersson T, Persliden J: **Digital radiography of scoliosis with a scanning method: initial evaluation**. *Radiology* 2001, **218**(2):402-410.

14. Schmitz A, Kandyba J, Koenig R, Jaeger UE, Gieseke J, Schmitt O: **A new method of MR total spine imaging for showing the brace effect in scoliosis**. *Journal of orthopaedic science : official journal of the Japanese Orthopaedic Association* 2001, **6**(4):316-319.

15. Benda BR, PO; Krebs, DE;: **Biomechanical relationship between center of gravity and center of pressure during standing**. *Rehabilitation Engineering, IEEE Transactions on* 1994, **2**(3):10.

16. O'Rourke J: **Finding minimal enclosing boxes**. *International Journal of Computer & Information Sciences* 1985, **14**(3):16.

17. Filaire M, Vacheron JJ, Vanneuville G, Poumarat G, Garcier JM, Harouna Y, Guillot M, Terver S, Toumi H, Thierry C: **Influence of the mode of load carriage on the static posture of the pelvic girdle and the thoracic and lumbar spine in vivo**. *Surgical and radiologic anatomy : SRA* 2001, **23**(1):27-31.

18. Harman EH, Han KH, Frykman P, Pandorf C: **The Effects of Backpack Weight on the Biomechanics of Load Carriage**. Natick, MA, USA: Military Performance Division of the US Army Research Institute of Environmental Medicine 2000.

19. Martin PE, Nelson RC: **The effect of carried loads on the walking patterns of men and women**. *Ergonomics* 1986, **29**(10):1191-1202.

20. Attwells RL, Birrell SA, Hooper RH, Mansfield NJ: **Influence of carrying heavy loads on soldiers' posture, movements and gait**. *Ergonomics* 2006, **49**(14):1527-1537.

21. Orloff HA, Rapp CM: **The effects of load carriage on spinal curvature and posture**. *Spine* 2004, **29**(12):1325-1329.

22. Birrell SA, Haslam RA: **The effect of military load carriage on 3-D lower limb kinematics and spatiotemporal parameters**. *Ergonomics* 2009, **52**(10):1298-1304.

23. Smith B, Ashton KM, Bohl D, Clark RC, Metheny JB, Klassen S: **Influence of carrying a backpack on pelvic tilt, rotation, and obliquity in female college students**. *Gait & posture* 2006, **23**(3):263-267.

CHAPTER 5: SUMMARY AND SIGNIFICANCE

5.1 Biomechanical Analysis

A simple biomechanical analysis of the moments exerted by external loads on the LS would allow us to understand, to some extent, the muscle force moments required to maintain the loaded upright position. First, we can estimate the moments around L5-S1, due to the following forces: total trunk, arms and head weight (W_t), anterior (W_A) and posterior (W_P) external loads, and spinal muscles (F_x). Second, by taking into account the postural adaptations in response to load, we can estimate the compressive and shear force components induced by different load configurations throughout the levels of the LS. Finally, these force components will allow us to calculate the net joint reaction force at each functional spinal unit in response to different load magnitudes and configurations.

Assumptions

In order to estimate the location of the CoM of the trunk (CoM_t), anterior (CoM_A) and posterior (CoM_P) loads with respect to the center of L5-S1, the following assumptions were made:

1) The average body mass, load magnitudes and distributions used in this analysis are those from the experiment described in Chapter 4.
2) The axis origin (*O)* for this analysis is located at the center of S1 (Fig. 5–1).
3) The length of the human trunk is approximately 30% of the adult body height (Table 5–1).[1]
4) The percentage of BW carried at L5-S1 is ~55% (Table 5–1).[2]

5) The CoM of the trunk, arms and head (*CoM_t*) is located at ~55% of the trunk length (cranially) and at ~50% anterior-posteriorly (Fig. 5–1).[1]

6) The CoM of the trunk is aligned with a line that joins the centroids of L1 and S1 (Fig. 5–1). This is the same line fragment used to measure lumbosacral flexion (ξ) throughout this dissertation. This was defined so that it was possible to find the position of the CoM_t, while taking into account the overall orientation of the LS.

7) The distance from the anterior and posterior loads (CoM_A and CoM_P) to CoM_t is 24.7cm. This number was defined based on average trunk (24cm) and gear depth (25.4cm).[2]

Moment equilibrium analysis – Standing unloaded position

The orientation of the LS used in this first analysis was that of standing without external load. This we can estimate the moments exerted by external loads on the LS before any postural adaptation. During static loading, the sum of moments *M* around L5-S1, for each load configuration *k* equals zero. Therefore, we have:

$$\sum M_{k-L5S1} = 0 \qquad (5\text{-}1)$$

where *k* is the unloaded standing position and the 6 combinations of 50%-50% and 20%-80% load distributions, at 22kg, 33kg and 45kg load magnitudes. The moments acting on the LS are:

$$M_{tk} + M_{Ak} + M_{Pk} + M_{xk} = 0 \qquad (5\text{-}2)$$

These moments are due to BW, anterior, posterior and spinal muscles, and they are the product of each force and the corresponding lever arm:

$$W_{tk}\, d_{tk} + W_{Ak}\, d_{Ak} - W_{Pk}\, d_{Pk} + F_{xk} d_x = 0 \qquad (5\text{-}3)$$

The variables d_{tk}, d_{Ak}, d_{Pk}, and d_x are the lever arms of each of the aforementioned loads (W_t, W_A, W_P, and F_x) for configuration k, in this case, with respect to the center of L5-S1 (Fig. 5–2). The negative sign assigned to the third and fourth terms of equation 5-3 follow the conventional definition of clockwise moment (counter-clockwise moments are defined as positive). The magnitudes of W_t, W_A, and W_P are known, and their lever arms can be calculated from the listed assumptions. Moreover, the lever arm d_x has been previously reported at the L5-S1 level (Table 5–2).[3] Therefore, the moment around L5-S1 induced by the trunk and external loads, and resisted by the spinal extensor muscles is:

$$M_{xk} = F_{xk}d_x = -W_{tk}\,d_{tk} - W_{Ak}\,d_{Ak} + W_{Pk}\,d_{Pk} \qquad (5\text{-}4)$$

Additionally, we can now solve for F_{xk} from equation 5-4:

$$F_{xk} = \frac{W_{tk}d_{tk}+W_{Ak}d_{Ak}-W_{Pk}d_{Pk}}{d_{xk}} \qquad (5\text{-}5)$$

Note that at each lumbar level, F_x is perpendicular to the line, of slope $tan(\beta)$, that bisects each IVD (grey line, Fig. 5–2). The magnitude of the estimated moments exerted by the spinal muscles around the LS for all load configuration were calculated using equation 5-4. First, the dashed line represent the moment exerted by the spinal muscles, necessary to maintain the upright position when standing without external load, and is constant, as we are not considering changes in posture (Fig. 5–3). Second, when load was carried in an even distribution (50%-50%), and assuming no postural changes, the spinal muscles were shown to exert an extension inducing moment (black bars). In this case, the moments of anterior and posterior loads mostly cancel each other out. The small difference across magnitudes is due to the fact that the position of the loads is not perfectly horizontal, but is inclined by ~8°. Consequently, the moment induced by the anterior load in an even distribution is slightly larger than that induced by the posterior load. The resulting moment is, therefore, mostly attributed to trunk, arms and head weight alone. This would suggest that in order to maintain the upright posture while standing unloaded or with evenly distributed load, spinal extensor muscles must be activated to counteract the natural moment of the upper body. In contrast, when load was carried with a posterior bias, the moments exerted by the spinal muscles induced flexion of the LS (grey bars). Meaning that load carriage in this distribution causes extension motion of the trunk. Additionally, the magnitude of these moments increased as a function of load magnitude.

Moment equilibrium analysis – Loaded postures

The results from this dissertation show that when carrying load the LS changes in both curvature and orientation to maintain balance, instead of the upright posture. Therefore, in order to estimate more realistic total moments around the loaded LS, the changes in overall LS flexion (ξ) and orientation of L5-S1 with respect to the horizontal (for each configuration) were incorporated into this analysis. These were re-calculated using equation 5-4 and the results were somewhat similar to when load was carried in the 50%-50% distribution (Fig. 5–4, black border bars). The magnitude of the moment due to spinal muscles increased slightly as a function of load magnitude. However, the magnitude of these moments was larger when the actual orientation of the LS and L5-S1 were taken into account. These differences were due to incremental changes in both lever arms and W_t as a result of variations of the L5-S1 orientation with respect to the ground. During posteriorly biased load carriage (20%-80%), the smallest torque around L5-S1 was found when carrying 22kg of load magnitude (Fig. 5–4, grey border bars).

Furthermore, the estimated moments at 33kg and 45kg during posteriorly biased load carriage (20%-80%) were much larger than those at 22kg. Interestingly, when load was carried with a posterior bias, the estimated moment exerted by spinal muscles activity was negative (Fig. 5–4, grey border bars). This means that the resulting moment exerted by exterior loads is positive when postural adaptations are taken into account. Additionally, the magnitude of these results were smaller than those estimated to be the induced solely by the carried loads (Fig. 5–3). This suggests that postural changes may be aimed not only at maintaining balance during load carriage, but to do so in a more efficient manner.

We can take this a step further and estimate the moment M_{xk} around each lumbar level. In order to do this, it is necessary to find the orientation of each IVD (i = L1-L2, L2-L3, L3-L4, L4-L5) at each load carriage configuration. The endplate orientation data, from Chapter 4, was used to calculate the angle of each IVD with respect with the horizontal (β_i). These results are shown in Figure 5–5 for each load carriage configuration. The estimated moments induced by the spinal muscles at each lumbar intervertebral joint are shown in Figure 5–6A for all load carriage configurations. When load was evenly distributed, spinal muscles exert a flexion-inducing moment throughout the LS to maintain the standing loaded posture. The magnitude of the moment at inferior lumbar levels (L4-L5 and L5-S1, 50%-50%) increased as a function of load magnitude, while that at superior levels the magnitude of the moment appears to remain unchanged. Contrastingly, when loads were carried with a posterior bias (20%-80%), extension-inducing moments appear to be experienced at superior lumbar levels while the opposite was found at inferior levels. Additionally, the magnitude of the moments seems to progressively decrease from L1-L2 to L4-L5, however, it was highest at L5-S1. The rate at which this change in moment magnitude occurs was largest and similar for 33kg and 45kg, while less pronounced for 22kg of load. Interestingly, the magnitude of the moment across all loads is very similar at L1-L2, however differences in moment magnitude become apparent between different load conditions at other levels, and in fact increases caudally throughout the LS. The postural adaptation of the LS appears to increase the magnitude of the moments experienced at inferior lumbar levels (Fig. 5–6B), while decreasing the magnitude of the moments due to posterior load at inferior levels (Fig. 5–6D). These results suggest that muscles at superior levels may be exerting an anterior moment around S1, while inferior levels may exert a posterior moment to maintain balance.

Force equilibrium analysis – Loaded postures

Using equation 5-5, we calculated the force exerted by the spinal extensor muscles by lumbar level and load configuration (Table 5–2). The estimated forces exerted by spinal muscles F_x (Fig. 5–**7**), matched the shape of the moments experienced at each IVD (Fig. 5–6A). This might be attributed to the small variations in paraspinal muscles' lever arms throughout the LS.[3] The magnitude of F_x is under 1.1kN throughout all lumbar levels and similar to moment results, the largest forces seem to be exerted at L1-L2 and L5-S1.

Intervertebral disc position and orientation information also allowed us to find the lever arms of each load with respect to the center of each lumbar IVD. These data were then used to estimate the compressive (C_t, C_A, C_P and C_x) and shear (S_t, S_A, S_P and S_x) force components, acting at each IVD (black, blue, green and red in Fig. 5–2):

$$C = W\cos(\beta_i) \qquad (5\text{-}6)$$

$$S = W\sin(\beta_i) \qquad (5\text{-}7)$$

Since the angles β_i are known (Fig. 5–5), we can calculate compressive and shear components of W_t, W_A, W_P, *and* F_x. During standing equilibrium the sum of forces perpendicular and parallel to the plane of each IVD is zero. Therefore, the sum of perpendicular or compressive components is:

$$\sum C_{ik} = C_{jx} + C_{xk} + C_{Ak} + C_{Pk} + C_{tk} = 0 \qquad (5\text{-}8)$$

while the sum of parallel or shear force components to the same plane is:

$$\sum S_{ik} = S_{jx} + S_{xk} + S_{Ak} + S_{Pk} + S_{tk} = 0 \qquad (5\text{-}9)$$

The subscripts t, A, P and x indicate the force components of trunk weight, anterior and posterior loads, and spinal muscle forces, respectively. Similarly, x and j subscripts indicate spinal muscle force and joint reaction force. Furthermore, from equations 5-8 and 5-9, we can solve for the components of the joint reaction force:

$$C_{jk} = -C_{Ak} - C_{Pk} - C_{tk} - C_{xk} \qquad (5\text{-}10)$$

$$S_{jk} = -S_{Ak} - S_{Pk} - S_{tk} - S_{xk} \qquad (5\text{-}11)$$

Additionally, the magnitude of the joint reaction forces is:

$$F_j = \sqrt{S_{jk}^{\ 2} + C_{jk}^{\ 2}} \qquad (5\text{-}12)$$

The compressive components of the joint reaction force experienced at each IVD were calculated using equation 5-10. These values were found to increase with load magnitude, independently of load distribution (Fig. 5–8A). The estimated compressive component of the joint reaction force was found to be larger during evenly distributed load carriage compared to posteriorly biased load carriage (Fig. 5–8A). During load carriage in the 20%-80% distribution, the reorientation of the LS in response to load appears to minimize the magnitude of the compressive forces exerted by W_{tk} (Fig. 5–8B) and W_{Pk} (Fig. 5–8AD). Contrastingly, those exerted by the anterior load remain mostly unchanged (Fig. 5–8C).

Previous *ex vivo* studies have shown that lumbar vertebrae compressive strength is in the range of 5-6kN, while cyclic loading has shown to reduce this value by up to 50%.[4] Therefore, lumbar compressive strength during prolonged loaded gait may be in the range of 2.5-3kN. Furthermore, the loads experienced by the lumbar spine during walking are reported to be up to 2.5 times BW.[5] This means that the estimated compressive component of the joint reaction forces of 0.6-1.2kN, while carrying 45kg in the 20%-80% distribution may be in the range of 1.5-3kN during gait. These compressive forces may be sufficient to induce vertebral damage during loaded gait. Similarly, if loads were to be carried with an even distribution, the compressive component of the joint reaction of 0.3-0.7kN may be in the range of 0.75-1.75kN during gait. Suggesting that this load carriage distribution may be more protective of the LS. However, this needs to be tested explicitly.

The shear components of the joint reaction force experienced at each IVD were calculated using equation 5-11. During standing without external load these shear forces act posteriorly at L1-L2 and L2-L3 and anteriorly at inferior lumbar levels (Fig. 5–9A). This is in agreement with what has been previously reported in the literature.[5, 6] Overall, the shear components of the joint reaction forces across load configurations were larger at inferior lumbar levels and increased as a function of load magnitude (Fig. 5–9A). In contrast to its compressive components, the shear components of the joint reaction forces per lumbar level were overall smaller in magnitude when load was carried in an even distribution than when it was carried with a posterior bias (Fig. 5–9A). During posteriorly biased load carriage, total shear forces increased mostly due to those induced by posterior loads (Fig. 5–9D). Interestingly, changes in shear forces exerted due to trunk load and anterior load appeared to be small, despite the changes in overall LS orientation (Fig. 5–9B and C).

The range of maximum shear force in the 20%-80% load distribution at L4-L5 and L5-S1 is 0.3-0.7kN. Meanwhile, IVDs and posterior bone elements have been reported to fail in shear at approximately 2-5.4kN and 2kN, respectively.[5] Assuming that failure strength is decreased by 50% and that shear forces are increased by 2.5 times due to gait, the estimated shear component of the joint reaction forces experienced by the LS is 1-1.4kN. During a long march, the LS may only be able to resist 1-2.7kN before fatigue damage is induced. Although this analysis estimates the compressive and shear forces experienced by the LS to be in the same range as its potential cyclic failure strength, these data may not be reflective of the situation *in vivo.* Potentially, due to the lack of information on the role of important factors such as trunk muscles, ligaments and abdominal pressure on LS stabilization.

The magnitude of the joint reaction force at each IVD can be now calculated using equation 5-12. In general, these forces were found to be smaller during evenly distributed load carriage (Fig. 5–10). Note that the direction of this force is perpendicular to the plane across each IVD and therefore it has no shear components. These results are smaller than those previously reported by more complete and complex models that have studied other load lifting tasks.[7] Differences might be attributed to the fact that we are not taking into account the process of donning load or loaded gait, and only studying but static standing (in equilibrium). Additional discrepancies might arise from differences in analysis assumptions and model components.[8]

5.2 Significance of Findings

Effect of evenly distributed load carriage on LS posture

During AP balanced load carriage, no postural changes were induced in the LS. Other authors have proposed that this load distribution protects the LS, as spinal loading is minimized when the vertebral column is balanced vertically on the pelvis.[4] However, previous studies have shown that compressive loading is more likely to affect the vertebral bodies than the IVDs.[9,10] Therefore, load carried in an even distribution may increase the risk of damaging intervertebral endplates and trabeculae of the vertebral bodies. In fact, several patterns of microfractures and healing trabeculae are commonly found in cadaveric vertebral bodies and endplates.[11] However, we did not find evidence of this in our subjects, but we did find evidence of IVD degeneration.

Effect of load carried with a posterior bias on LS posture

In all experiments presented in this dissertation during which load was carried with a posterior bias increased LS flexion and overall reduction of LS lordosis were observed. Interestingly, disparate postural behaviors of inferior and superior LS were measured as a function of load magnitude and distribution, respectively. Additionally, changes in lordosis distribution throughout the LS were observed between unloaded and loaded tasks.

Increased lumbar flexion causes the orientation of all LS components to become more horizontal with respect to the ground. The overall effect of this change in orientation, independently of differences in lordosis, increases the risk of shear injury of the LS. Studies *in vitro* have shown that during flexion soft tissues primarily resist shear stresses. For example, IVD experiencing shear stresses exhibit creep resulting in increased anterior translation of a vertebral body relative to the inferior vertebra.[12] Consequently, shear stresses in the facet joints also increase, as they resist this anterior translation motion. Different structural and postural factors that contribute to the amount of shear stresses resisted by the facet joints will be discussed later.

The magnitude of LS lordosis change between unloaded and loaded tasks was progressively reduced as load magnitude increased. This was due to increased extension of the superior LS, while inferior LS lordosis was reduced by the same amount independently of load magnitude. Overall, the contribution of each IVD to the LS response to load carriage ranges from 0° to 5°. Although the magnitude of these changes might appear to be small, the biomechanical implications of local lordosis deviating from those of the 'more natural' standing position are important. For example, the separation between spinous processes of contiguous vertebrae when standing is in the range of 4 to 6 mm.[13] Therefore, in cases of extreme extension, as observed in the superior LS during load carriage, may cause neighboring spinous processes or facet joints to become in contact.[14] This results in transmission of compressive forces down the spine, which may cause bone bruising, cartilage damage, or fracture at high loads. Extreme extension may also result in a larger fraction of the total compression stress to be absorbed by the facet joints. The exact proportion of the total compressive loads resisted by the facet joints and IVDs *in vivo* as a function of the degree of FSU flexion/extension is unknown. However, Adams *et al* showed *ex vivo* that 2° of flexion increases the proportion of compression load carried by the facet joints from 1% to 16%.[15] Similarly, in extreme extension, the superior articular surface of the facet joints come in contact with the inferior margin of the joints near the lamina of the vertebra below. These areas lack articular cartilage and may potentially produce articular impingement during a loaded task. Overall, extension of the superior LS may increase the risk of injury of the neural arch during heavy load carriage.

In contrast to the potential effects of extreme extension of the superior LS, the observed flexion of the inferior LS may cause partial unloading of the facet joints. This postural adaptation to load is complex as it has multiple biomechanical consequences. First, the facet joints experience increased shear forces (perpendicular to facet joint plane with respect to the coronal plane) as they prevent anterior translation of L4 over L5 and L5 over S1. Second, in the flexed position, the surfaces of the facet joints slide past each other, reducing the articular contact area. The total shear stress at the facet joint potentially increases considerably as it is directly proportional to the magnitude of the force and inversely proportional to the total contact area between the articular surfaces. Finally, the orientation of the facet joints varies along lumbar levels. In the inferior LS they are closer to the coronal plane increasing their capacity to resist shear stresses. However, the relative contribution of these three factors to the overall shear stresses resisted by the facet joints is unknown. Furthermore, unloading of the facet joints also increases the amount of compression and shear stresses resisted by the IVDs.

Overall, the results presented in this dissertation suggest that intervertebral distances follow the level-dependent angular responses to load carriage. However, at inferior levels, where local lordosis changes were the largest, the corresponding intervertebral distances varied between anterior and posterior regions. At the L4-L5 level there was more anterior compression than posterior distraction, while at the L5-S1 IVD the opposite was observed. These results suggest that the vertebral joint center of rotation is not in the middle of the IVDs. The location of the centers of rotation of the L4-L5 and L5-S1 levels during extension/flexion movements while standing is slightly posterior (3-5mm) to the AP midline of the IVD.[16] The disparate behavior between L4-L5 and L5-S1 suggests that the center of rotation of L5-S1 may shift forward during load carriage. Therefore, the center of rotation of each FSU may vary as a function of relative orientation with respect to the ground. This is relevant in the context of load carriage because it potentially affects the position of the nucleus pulposus within the IVD. Proposed injury models developed based on *in vitro* studies, suggest that repetitive loading in the flexion position may increase the risk of shear injury of the IVD.[17] The proposed mechanism is that as anterior lamellae of the annulus fibrosus becomes more slack and the posterior region are stretched, therefore reducing its strength. Consequently, when the pressure rises in the NP due to compression it may rupture the annulus, causing a herniation or chronically stretch the posterior lamellae resulting in IVD protrusion.

Sacral orientation was measured in the experiments presented on chapters 3 and 4 to further understand the relation between the postural adaptations of the LS and the pelvic region. Here, we showed that during load carriage there is an increase in LS flexion, anterior rotation of the sacrum and changes in regional lumbar lordosis. Previous studies have reported a moderate association between lumbar lordosis and sacral slope in the standing position. Together, these data suggest that the lumbar flexion during load carriage results from the postural contribution of sacral orientation and regional LS lordosis.

Furthermore, the sacrum and superior LS have a postural response that is load magnitude dependent. Interestingly, the posture of the inferior LS appears to be configuration dependent, as it remained constant across load magnitudes while they were carried in the 20%-80% configuration. Likely, this posture is the most efficient, in terms of energy expenditure, to maintain the balance of the system. Previous authors have proposed that at significant lumbar flexion the contribution of passive structures (i.e. lumbodorsal fascia) to total muscle tension becomes substantial. These data, together with the fact that the largest postural adaptation is in sacral orientation, suggest the recruitment of muscles in the gluteal region may act in cooperation with spinal fascial planes to assist maintain load carriage posture.

Postural differences between loaded tasks

Although the overall lordosis of the LS did not differ across loaded tasks presented in chapter 2 (50.8kg immediately after donning load, after standing and walking for 45mins), local lordosis differences were observed. Inferior lumbar levels (L4-5, L5-S1) became significantly less lordotic immediately after donning load. Contrastingly, lordosis at superior lumbar levels (L1-L2 and L2-L3) was different from unloaded tasks only after 45 minutes of exposure to load. Interestingly, after 45 minutes of walking with 50kg of load the pattern of lordosis distribution throughout the length of the LS was more similar to that measured immediately after donning load than after 45 minutes of standing with load. These data suggest that compared to loaded gait, maintaining a static posture may require continuous engagement of the LS musculature or may induce a significant amount of creep on its passive elements.[18] Therefore, sustained flexion may reduce the stabilization capacity of muscles and ligaments,[19, 20] which may explain the differences in lordosis distribution across loaded tasks.

Effect of intervertebral disc degeneration on lumbar spine load carriage

The prevalence of IVD degeneration in at least one lumbar IVD among the group of active-duty Marines studied in chapter 2 was 47.5%, while the prevalence at the L5–S1 level was 40.7%. Posturally, small differences in sacral orientation during load carriage between Marines with and without IVD degeneration at the L5-S1 level were observed in the study presented in chapter 3. Intervertebral discs with Pfirrmann grades of III and IV were considered ‘degenerated’, which have been shown to be unstable compared to grades I, II and V. In the presence of degeneration, IVDs have low osmotic pressure of the nucleus pulposus and reduced height. Consequently, the IVD resists less of the applied compressive forces, thus more load falls on the annulus fibrosus and facet joints.[21] This causes instability of the inner annulus fibrosus lamellae, which may lead to structural damage and increase the risk of injury of the IVD. It has been proposed by other authors that if injuries of this kind occur at the IVD at a faster pace than the body can heal, this may cause the beginning the degeneration process.

5.3 Limitations

Unfortunately, the resolution of the images was not high enough to allow for quantification of IVD deformation in the presence of degeneration or protrusions as a function of load magnitude, distribution or duration of exposure to load. These data would allow estimating IVD creep *in vivo* and furthering our understanding of the interaction between spinal muscles and deformation of other LS soft tissues. Another relevant variable that was not possible to measure was facet joint orientation, which has been shown to be associated with excessive anterior translation of a vertebra over adjacent vertebra and may lead to vertebral arch fractures.

An important limitation of this dissertation work, specifically in chapter 3, is the attrition rate of 34%. Unfortunately, it is not possible for us to discern between volunteers who decided to stop participating in the study and those who were unable to complete SOI training due to injury. Most importantly, we cannot identify those who were discharged due to back injuries.

5.4 Conclusions

Several other authors have explored the effect of training, different load magnitudes and distribution on performance, gait mechanics and joint angles. However, data describing LS postural response to varying these variables was not available. In the work presented here LS posture was measured as loaded tasks, Marine's fitness level, and load magnitude and distribution varied. Although increasing fitness level of Marines has been shown to reduce overall musculoskeletal injuries, it appear to have no effect on LS posture during load carriage. Moreover, we found that the main driver of LS posture is the magnitude of carried load and perhaps more importantly, the way this load is distributed.

Previous *ex vivo* studies suggest that marching long distance with a heavy backpack could be a common scenario of shear-induced failure of the neural arch.[16] In conclusion, the data presented in this dissertation furthers our understanding of the range of potential injury mechanisms that may affect of the lumbar spine during load carriage.

5.5 Future Directions

The work presented in this dissertation has allowed us to propose mechanisms of injury of the LS due to heavy load carriage. However, future work should focus on understanding the interaction between FSU flexion/extension and differences in lumbar level morphology of the facet joints with respect to shear forces. In addition, while shape variations of the facet joints have been studied in relation to lower back pain, disc degeneration and spondylolisthesis, understanding its postural and biomechanical consequences during load carriage remains unknown.

In this work we have shown that carrying load with an even anterior-posterior distribution does not cause postural changes. Future work in this line of research should include prototyping of load carriage systems that are specifically designed to allow this kind of load distribution. In addition, studying the performance and discomfort caused by such a load carriage system should be evaluated. Studying the medium and long-term consequences of carrying load more evenly distributed in dedicated load carriage systems would further our understanding of the effect of load distribution on LS health. These data together would allow the development of more effective injury preventive standards.

Table 5−1 Variables used to define location of center of mass (CoM) of trunk, anterior and posterior loads.

Definition	**Value**
Average Height	177.80 cm
% of HT that corresponds to trunk	30%
Calculated trunk length	53.34 cm
Location of CoM_t (% of trunk length)	55.14%
Distance from CoM_t from origin O	29.41 cm
Trunk depth (cm)	24.00 cm
Distance from trunk CoM_t to CoM_A	24.70 cm
Distance from trunk CoM_t to CoM_P	24.70 cm

Table 5−2 Percentage of body weight (BW) carried at each lumbar level and the equivalent in Newtons based on the average BW from Chapter 4 is 76.77kg.[3] The right column contains the level arm of the erector spinae and multifidus muscle group at each lumbar level.[2]

Lumbar Level	**BW (%)**	**W_t (N)**	**d_t, Spinal Muscles Lever Arm (cm)**
L1-L2	40%	301.25	3.07
L2-L3	43%	323.84	3.45
L3-L4	48%	361.49	3.68
L4-L5	52%	391.62	3.99
L5-S1	55%	414.21	3.92

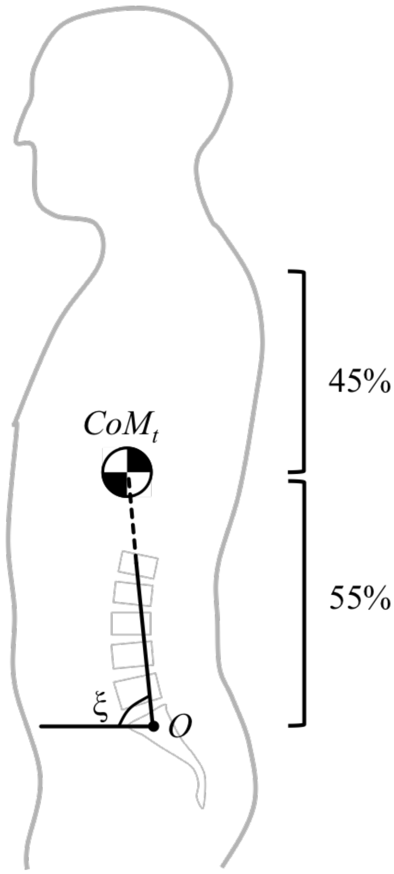

Figure 5–1 Diagram showing the assumption that the center of mass of the trunk (CoM_t) is located approximately at 55% the trunk length. This diagram also shows the assumption that the COM_t is aligned with the line that joins the centroids of L1 and S1. This is the same line used to calculate angle with respect to the horizontal (ξ), a measurement of lumbosacral flexion.

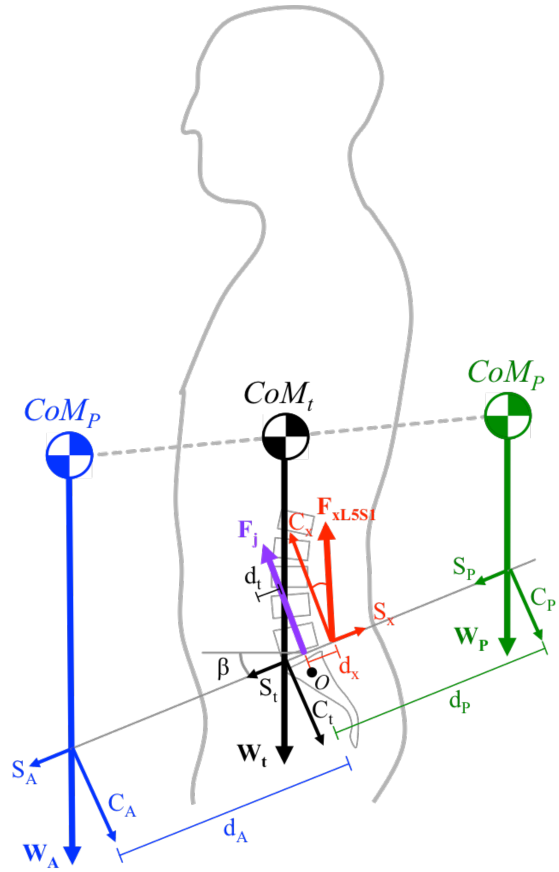

Figure 5–2 Free body diagram showing the forces acting on an intervertebral disc. Here, forces around L5-S1 are shown, however, similar calculations were applied to all lumbar levels. The origin O is located at the centroid of S1. The grey line represents the reference axis across L5-S1, which is at an angle β from the horizontal. The location of the center of mass of the trunk (CoM_t) is shown in black, and center of mass of anterior and posterior loads (CoM_A and CoM_P) are shown in blue and green, respectively. The forces exerted by these loads are W_t, W_A and W_P, and follow the same color scheme. Additionally, the force exerted by the spinal muscles F_x is shown in red. The compressive (C_t, C_A, C_P and C_x) and shear (S_t, S_A, S_P and S_x) components of these forces are also shown. Each of these forces act at a distance d_t, d_A, d_P and d_x from the center of each IVD (L5-S1 shown). Finally, the net joint reaction force acting on the L5-S1 functional spinal unit is shown in purple.

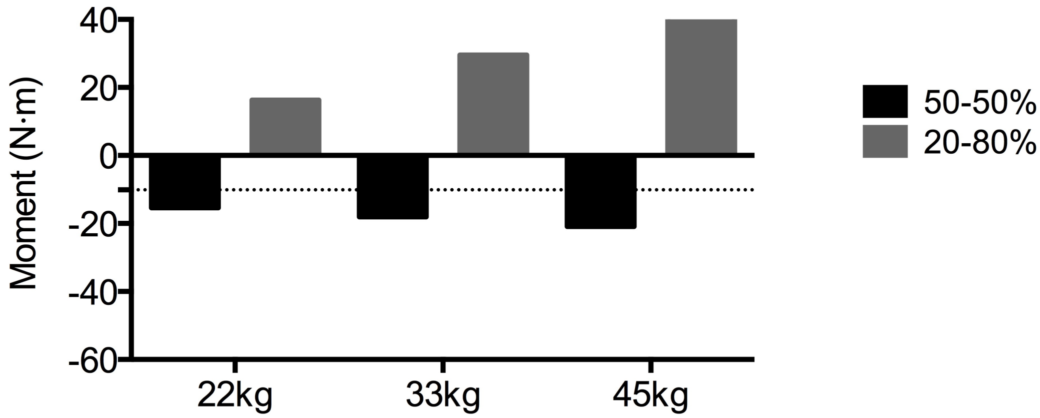

Figure 5–3 Estimated moment exerted by trunk, arms and head weight, and anterior and posterior load around L5-S1. These values were estimated using equation 5-4. The dashed line represents the moment exerted by the spinal muscles, necessary to maintain the upright position when standing without external load. Solid black bars represent these results for loads carried in an evenly distributed configuration. Solid grey bars show total torque results for loads carried with a posterior bias. Positive torques are anterior or flexion inducing, while negative torque values are posterior or extension inducing.

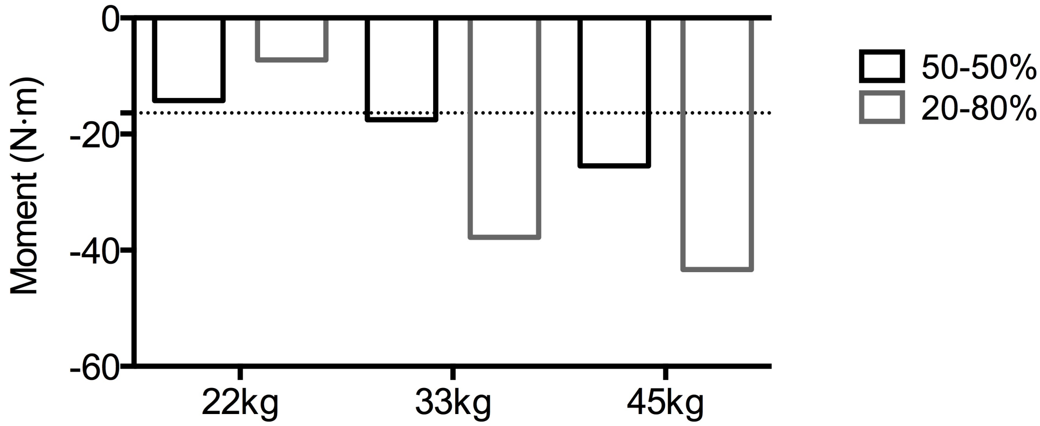

Figure 5–4 Estimated moment exerted by trunk, arms and head weight, and anterior and posterior load around L5-S1. These values were estimated using equation 5-4. This analysis takes into consideration changes in overall lumbar spine orientation with respect to the ground (ξ). The dashed line represents the moment exerted by the spinal muscles, necessary to maintain the upright position when standing without external load. Black bines represent these results for loads carried in an evenly distributed configuration. Grey bars show total torque results for loads carried with a posterior bias. Positive torques are anterior or flexion inducing, while negative torque values are posterior or extension inducing.

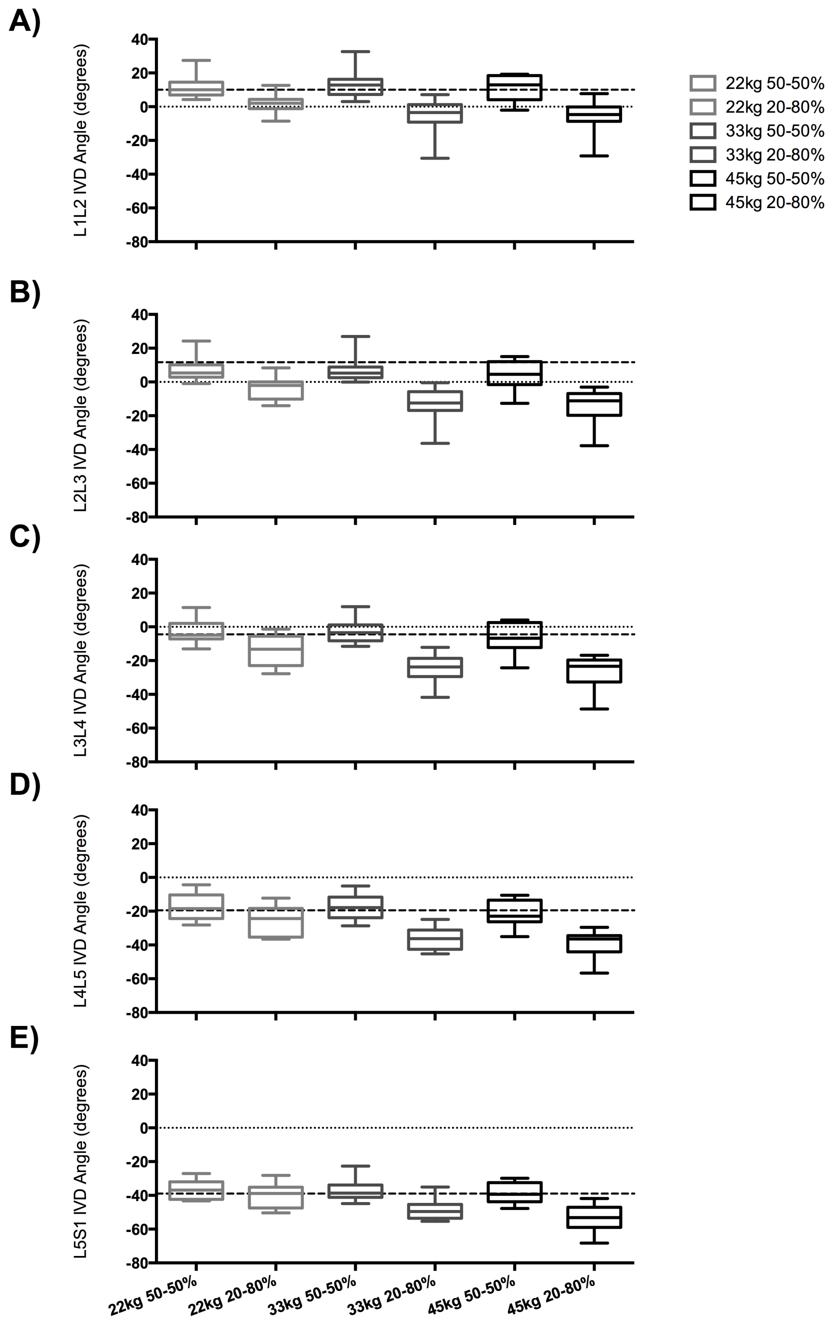

Figure 5–5 Orientation of each intervertebral disc by load magnitude and distribution, per lumbar level A) L1-L2, B) L2-L3, C) L3-L4, D) L4-L5 and E) L5-S1. The dashed line represents the orientation when standing without external load. Small dashed line represents zero degrees.

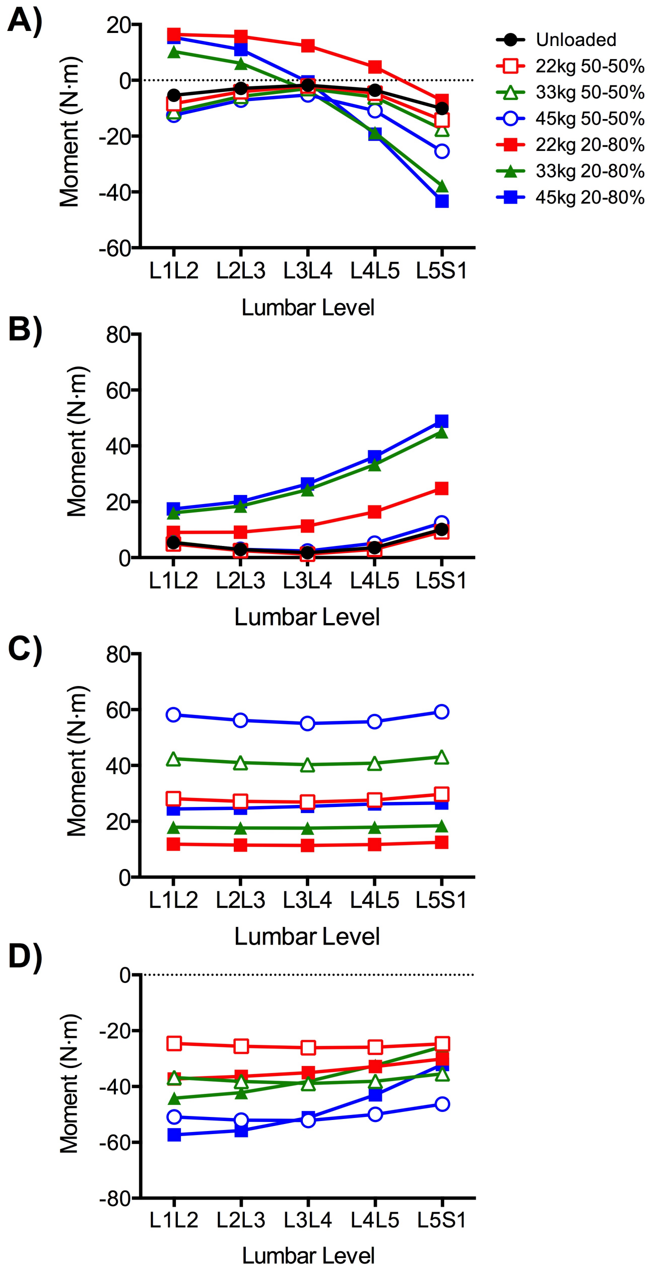

Figure 5–6 Biomechanical analysis results for calculated moment arms (N•m). Results for moment arms exerted by A) spinal muscles, B) trunk weight, C) anterior, and D) posterior loads around each lumbar IVD. These values were estimated using equation 5-4. Open shapes represent evenly distributed loads (50%-50%) and solid shapes represent loads carried with a posterior bias (20%-80%). Red = 22kg, green = 33kg, and blue = 45kg.

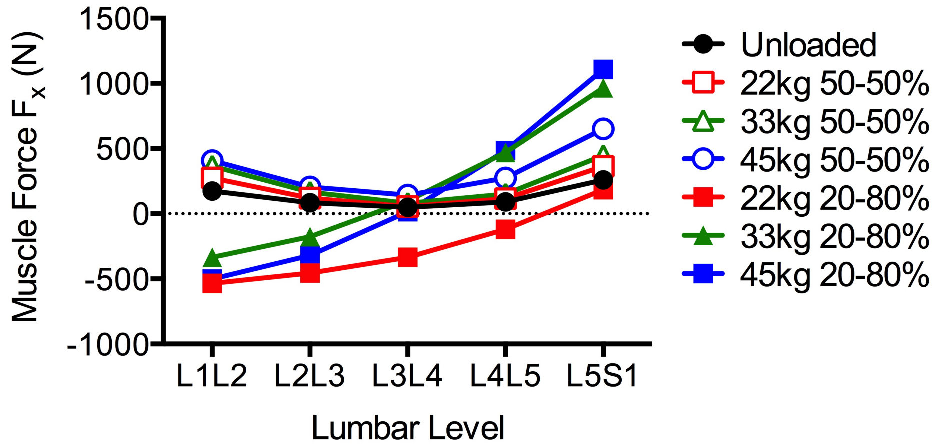

Figure 5–7 Biomechanical analysis results for estimated force exerted by spinal muscles. These values were estimated using equation 5-5. Open shapes represent evenly distributed loads (50%-50%) and solid shapes represent loads carried with a posterior bias (20%-80%). Red = 22kg, green = 33kg, and blue = 45kg.

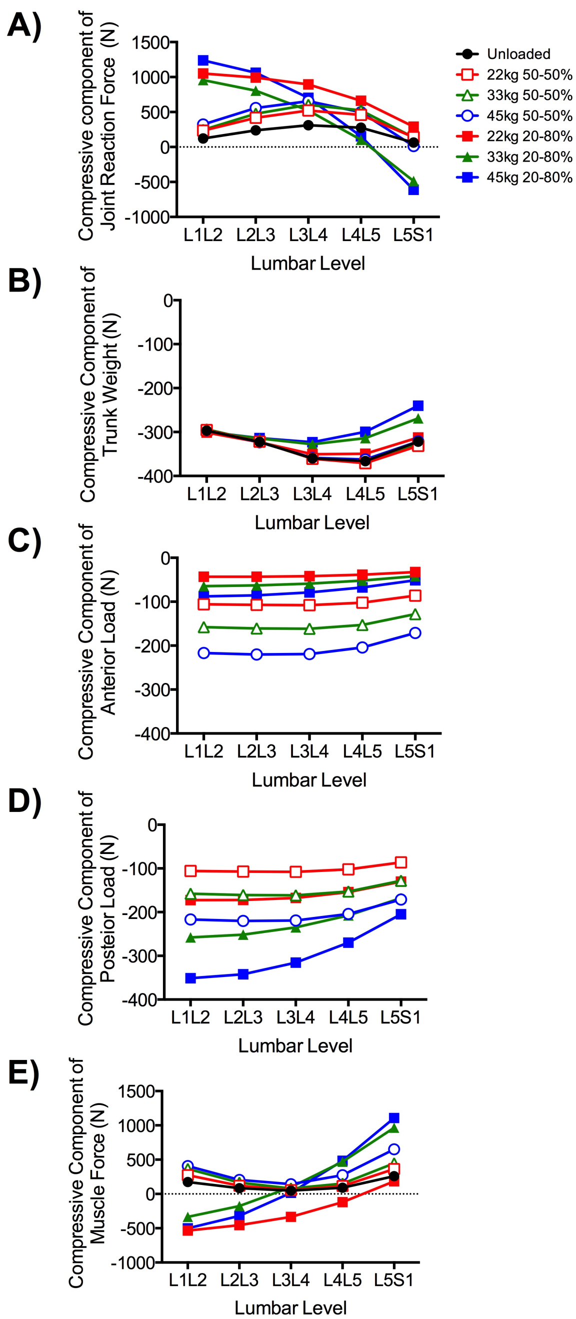

Figure 5–8 Biomechanical analysis results for estimated compressive component forces: A) joint reaction force, B) trunk weight, C) anterior, D) posterior, and E) muscle force at each lumbar IVD. These values were estimated using equations 5-5, 5-6 and 5-10. Open shapes represent evenly distributed loads (50%-50%) and solid shapes represent loads carried with a posterior bias (20%-80%). Red = 22kg, green = 33kg, and blue = 45kg.

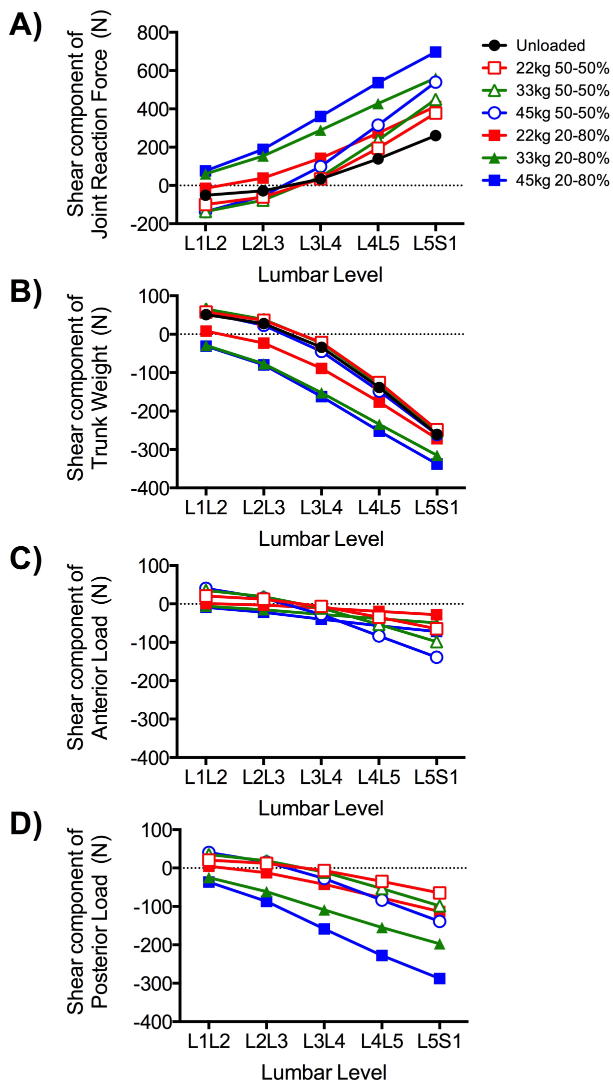

Figure 5–9 Biomechanical analysis results for estimated compressive component forces: A) joint reaction force, B) trunk weight, C) anterior, D) posterior. These values were estimated using equations 5-7 and 5-11. Open shapes represent evenly distributed loads (50%-50%) and solid shapes represent loads carried with a posterior bias (20%-80%). Red = 22kg, green = 33kg, and blue = 45kg.

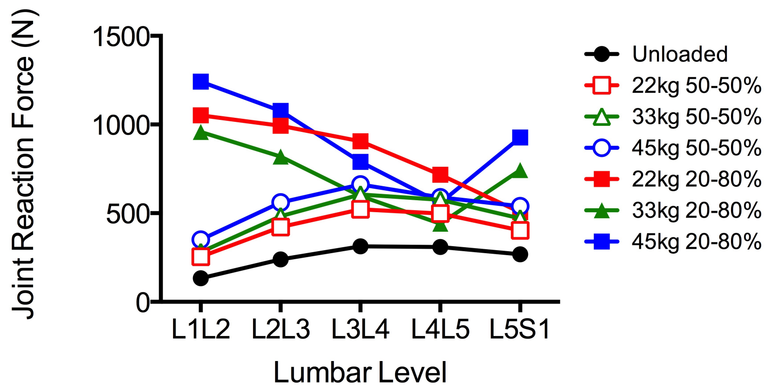

Figure 5–10 Biomechanical analysis results for estimated joint reaction force magnitude. These values were estimated using equation 5-12. Open shapes represent evenly distributed loads (50%-50%) and solid shapes represent loads carried with a posterior bias (20%-80%). Red = 22kg, green = 33kg, and blue = 45kg.

5.6 References

1. de Leva P: **Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters**. *Journal of biomechanics* 1996, **29**(9):1223-1230.

2. Liu YW, JK: **Estimation of the Inertial Property Distribution of the Human Torso from Segmented Cadveric Data**. *Reprinted from Perspectives in Biomedical Engineering* 1973.

3. Watanabe K, Miyamoto K, Masuda T, Shimizu K: **Use of ultrasonography to evaluate thickness of the erector spinae muscle in maximum flexion and extension of the lumbar spine**. *Spine (Phila Pa 1976)* 2004, **29**(13):1472-1477.

4. Brinckmann P, Johannleweling N, Hilweg D, Biggemann M: **Fatigue fracture of human lumbar vertebrae**. *Clinical biomechanics* 1987, **2**(2):94-96.

5. Adams MA: **The biomechanics of back pain**, 3rd edn. Edinburgh ; New York: Churchill Livingstone/Elsevier; 2013.

5. Hutton WCS, JRR; Cyron, BM;: **is spondylolysis a fatigue fracture?** *Spine* 1977, **2**(202):9.

6. Marras WS, Davis KG, Kirking BC, Granata KP: **Spine loading and trunk kinematics during team lifting**. *Ergonomics* 1999, **42**(10):1258-1273.

7. Rose JD, Mendel E, Marras WS: **Carrying and spine loading**. *Ergonomics* 2013, **56**(11):1722-1732.

8. Brinckmann P, Biggemann M, Hilweg D: **Prediction of the compressive strength of human lumbar vertebrae**. *Spine (Phila Pa 1976)* 1989, **14**(6):606-610.

9. Adams MA, Freema BJ, Morrison HP, Nelson IW, Dolan P: **Mechanical initiation of intervertebral disc degeneration.** *Spine* 2000, **23**(13):1625-1636.

10. Vernon-Roberts B, Pirie CJ: **Healing trabecular microfractures in the bodies of lumbar vertebrae**. *Annals of the rheumatic diseases* 1973, **32**(5):406-412.

11. Cyron BM, Hutton WC: **The behaviour of the lumbar intervertebral disc under repetitive forces**. *International orthopaedics* 1981, **5**(3):203-207.

12. Xia Q, Wang S, Passias PG, Kozanek M, Li G, Grottkau BE, Wood KB, Li G: **In vivo range of motion of the lumbar spinous processes**. *European spine journal : official publication of the European Spine Society, the European Spinal Deformity Society, and the European Section of the Cervical Spine Research Society* 2009, **18**(9):1355-1362.

13. Adams MA, Dolan P, Hutton WC: **The lumbar spine in backward bending**. *Spine (Phila Pa 1976)* 1988, **13**(9):1019-1026.

14. Adams MA, Hutton WC: **The effect of posture on the role of the apophysial joints in resisting intervertebral compressive forces**. *The Journal of bone and joint surgery British volume* 1980, **62**(3):358-362.

15. Alexander LA, Hancock E, Agouris I, Smith FW, MacSween A: **The response of the nucleus pulposus of the lumbar intervertebral discs to functionally loaded positions**. *Spine (Phila Pa 1976)* 2007, **32**(14):1508-1512.

16. Adams MA, Hutton WC: **Prolapsed intervertebral disc. A hyperflexion injury 1981 Volvo Award in Basic Science**. *Spine (Phila Pa 1976)* 1982, **7**(3):184-191.

17. McGill SM, Brown S: **Creep response of the lumbar spine to prolonged full flexion**. *Clinical biomechanics* 1992, **7**(1):43-46.

18. Adams MA, Dolan P: **Time-dependent changes in the lumbar spine's resistance to bending**. *Clinical biomechanics* 1996, **11**(4):194-200.

19. Yahia LH, Audet J, Drouin G: **Rheological properties of the human lumbar spine ligaments**. *Journal of biomedical engineering* 1991, **13**(5):399-406.

20. Adams MA, Freeman BJ, Morrison HP, Nelson IW, Dolan P: **Mechanical initiation of intervertebral disc degeneration**. *Spine (Phila Pa 1976)* 2000, **25**(13):1625-1636.