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TITLE: Lumbar Spine Musculoskeletal Physiology and Biomechanics During Simulated Military Operations

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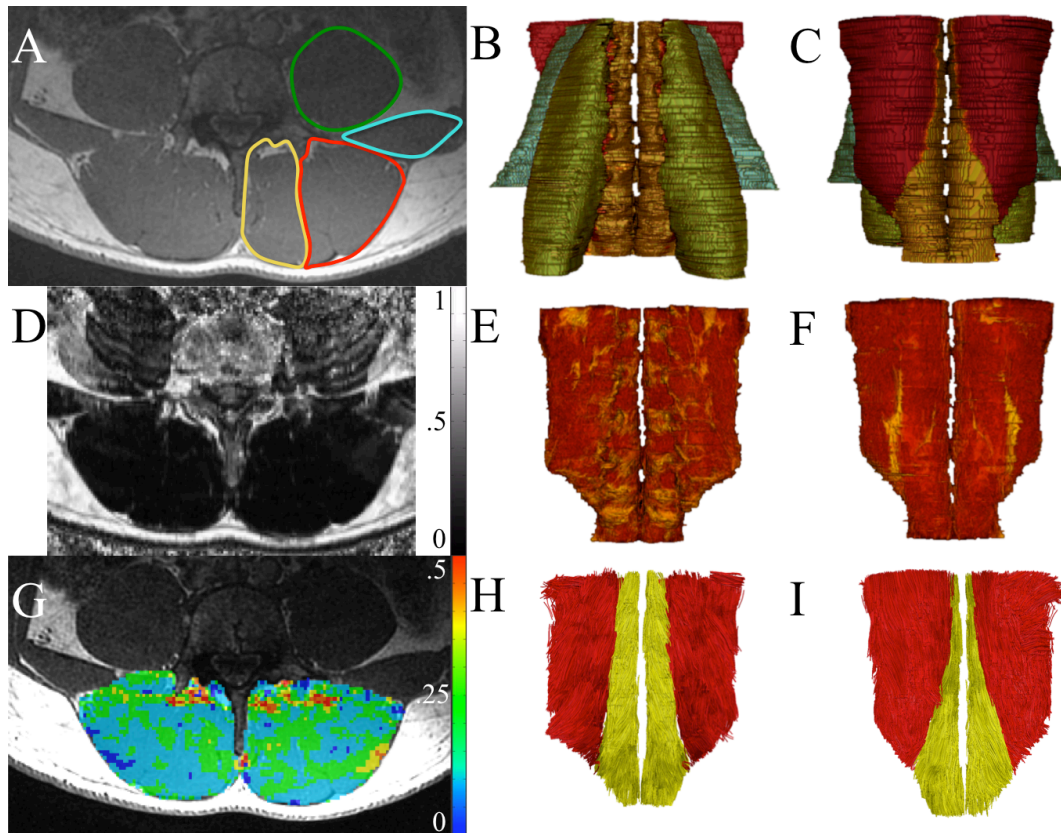
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14. ABSTRACT This study evaluated the relationship between 3D geometry of the lumbar spine, under different positions, and the pathophysiology of the intervertebral disc and lumbar trunk muscles. 43 Marines were scanned using upright MRI in operationally relevant loading conditions with minimum operational load, 39 of these Marines were also scanned in a high-resolution MRI to assess muscle and disc physiology. Local changes in posture were found in sitting and prone on elbows positions. A caudal increase in fat fraction of the multifidus was observed. No significant correlations were found between disc injury, low back pain, and muscle pathophysiology. Erector spinae fractional anisotropy was found to be a significant predictor of lumbar posture; increased fractional anisotropy predicts straighter, forward leaning posture.					
15. SUBJECT TERMS Spine, Lumbar, Posture, Muscle Architecture, Low Back Pain, Intervertebral Disc					
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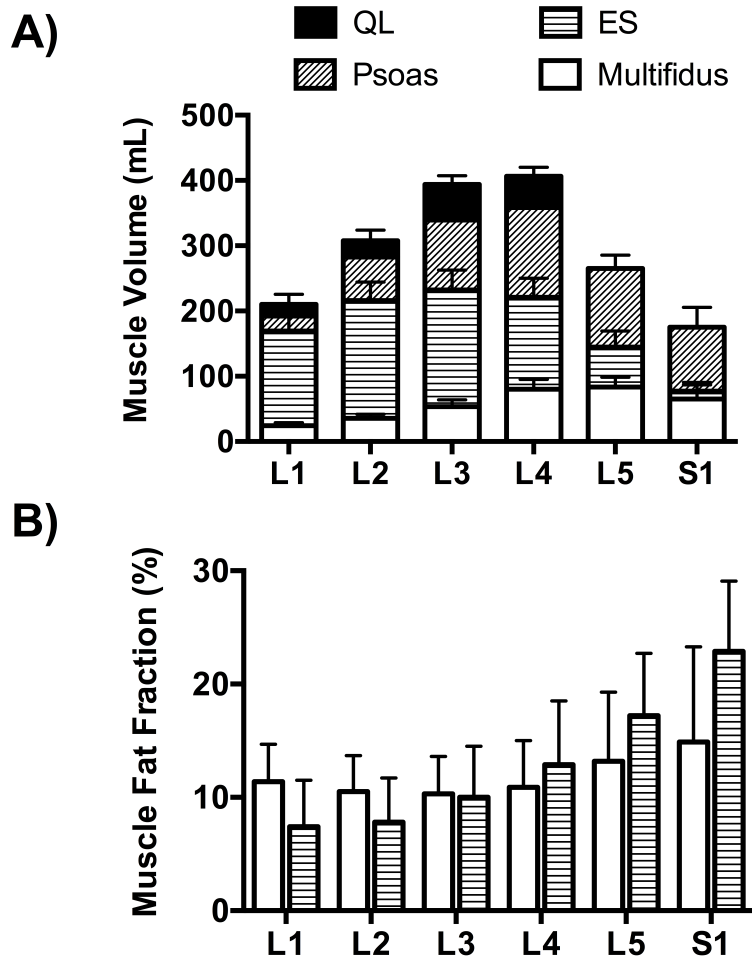
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1. **INTRODUCTION:** The weights of loads carried into battle pose an injury and performance problem for the US Marines. Marine Corps assault loads range from 44kg for the rifleman to 61kg for the squad leader. These are well in excess of the recommended assault load of 22.7kg; 30% of body weight. Two things are clear: (1) Marines routinely carry more weight than the recommended limit, and (2) the Veterans Administration shows an increasing trend in disabilities related to lower back injury as a result of carrying excessive loads in operational environments. We have implemented new MRI technologies enabling quantification of lumbar spine kinematics under simulated march conditions. From these investigations, we have observed a large number (37.5%) of Marines with some evidence of disc pathology. However, it is unclear if these disc changes, or the expected concomitant muscle changes, are related to kinematic changes in the lumbar spine. Further, it is not known whether pre-existing disc pathology predicts future injury and service life in U.S. Marines. Our central hypothesis is that lumbar disc and muscle degeneration alters the kinematic response of the lumbar spine to functional positions and loads, predisposing individuals to injury.
2. **KEYWORDS:** Spine, Lumbar, Posture, Muscle Architecture, Low Back Pain, Position
3. **ACCOMPLISHMENTS:**
 - **What were the major goals of the project?**
 - Major Goal 1: To compare lumbar spine kinematics in simulated operational conditions in Marines with measurable lumbar disc pathology
 - Task 1-Approved IRB Protocols
 - Local IRB Approvals
 - Complete: 01 August 2013
 - HRPO Approvals
 - Complete: 01 August 2013
 - Task 2-Subject battalions identified and coordinated
 - Complete: 01 July 2015
 - 55 Subjects recruited
 - Task 3-Data Collection/Analysis
 - Vertical data acquisition
 - Complete: 14 July 2015
 - 55 Subjects scanned
 - Vertical data analysis
 - Complete: 16 July 2015
 - Major Goal 2: To quantify changes in lumbar spine muscle architecture in Marines with measurable lumbar disc pathology.
 - Task 1-Approved IRB Protocols
 - Local IRB Approvals

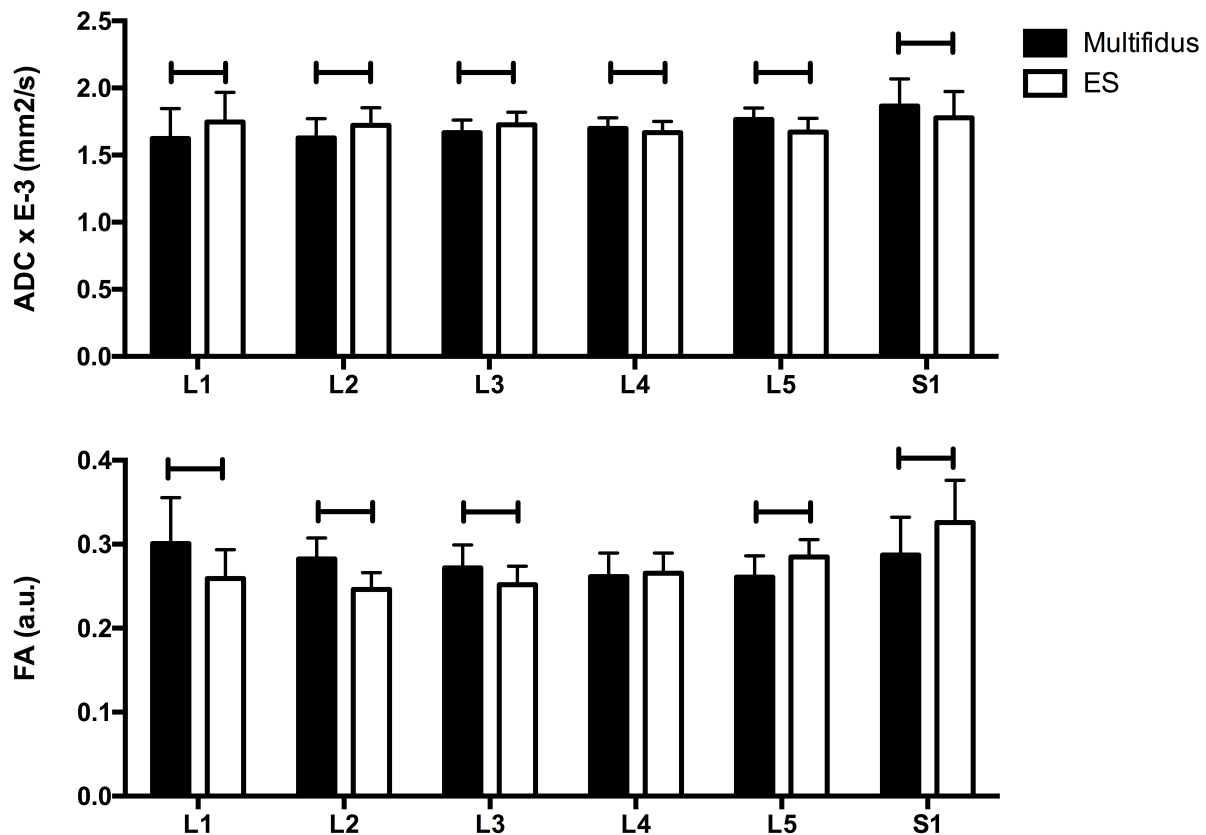
- Complete: 01 August 2013
- HRPO Approvals
 - Complete: 01 August 2013
- Task 2-Subject battalions identified and coordinated
 - Complete: 01 July 2015
 - 43 Subjects recruited
- Task 3-Data Collection/Analysis
 - Vertical data acquisition
 - Complete: 14 July 2015
 - 39 Subjects scanned
 - Supine data analysis
 - Complete: 15 June 2016
- **What was accomplished under these goals?**
- **Study 1: NONINVASIVE ASSESSMENT OF PARASPINAL MUSCLE COMPOSITION IN ACTIVE-DUTY MARINES**
 - 43 Marines (mean age = 26.8 ± 6.4 years, height = 1.78 ± 0.07 meters, weight = 82.04 ± 9.85 kg., BMI = 25.90 ± 2.86 kg/m²) volunteered to participate for this study. Magnetic resonance images of the lumbar spine (L1-S1) of this group of Marines were acquired using an upright 3T MRI scanner (GE MR350 Discovery, GE Healthcare) and spine array coil. The imaging protocol consisted of 1) an anatomical, 2) fat and water separation scan, 3) diffusion tensor imaging of the LS and 4) T2 mapping of each lumbar intervertebral disc. Anatomic images were imported into Osirix for semi-automatic segmentation. The contours of the multifidus, erector spinae group, psoas and quadratus lumborum muscles were manually traced from L1 to S1. The resulting segmentations were used to generate masks to quantify muscles volume, fat content and diffusion properties. Two subjects dropped out of this study due to claustrophobia in the MRI scanner. Respiratory artifacts were found in 6 masks of the psoas and these data were excluded from analysis. Additionally, breathing artifact was found in the DTI scans of 10 subjects and the data was excluded from analysis.



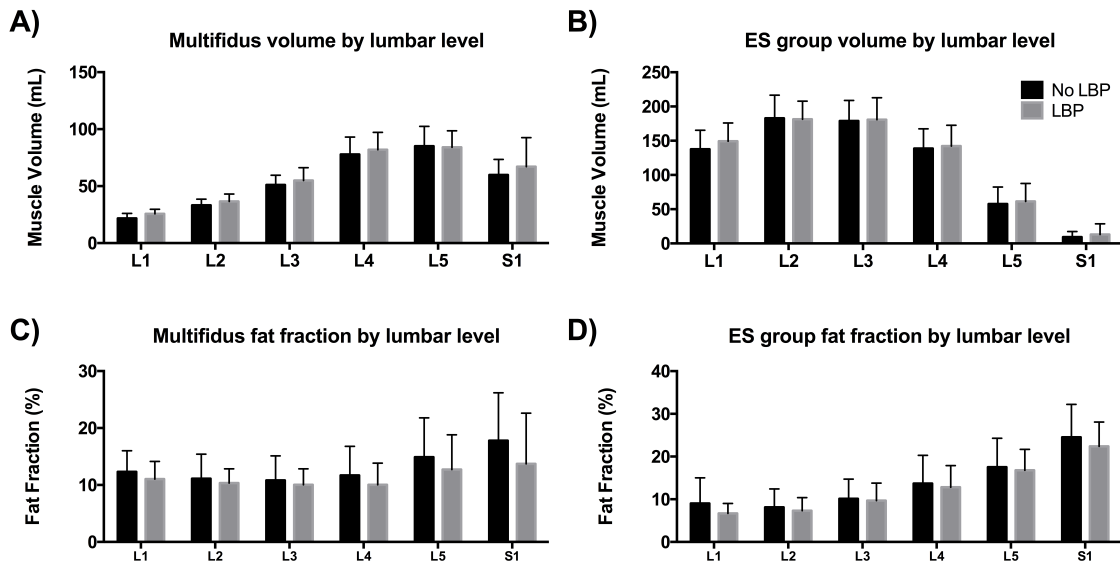
Study 1 Figure 1. **A)** Example of anatomical MR images used for segmentation of multifidus (yellow), erector spinae group (ES, red), psoas (green) and quadratus lumborum (QL, cyan). These segmentations were traced throughout the lumbar spine (L1-S1). **B)** Anterior and **C)** posterior (right) three-dimensional representation of these muscles. **D)** Fat fraction map of the paraspinal muscles. **E)** Anterior and **F)** posterior reconstruction showing fat distribution in multifidus and ES group throughout the lumbar spine. **G)** Fractional anisotropy (FA) map of the multifidus and ES groups overlay on anatomical image. **H)** Anterior and **I)** posterior views of tractography of the erector spinae (red) and multifidus (yellow).



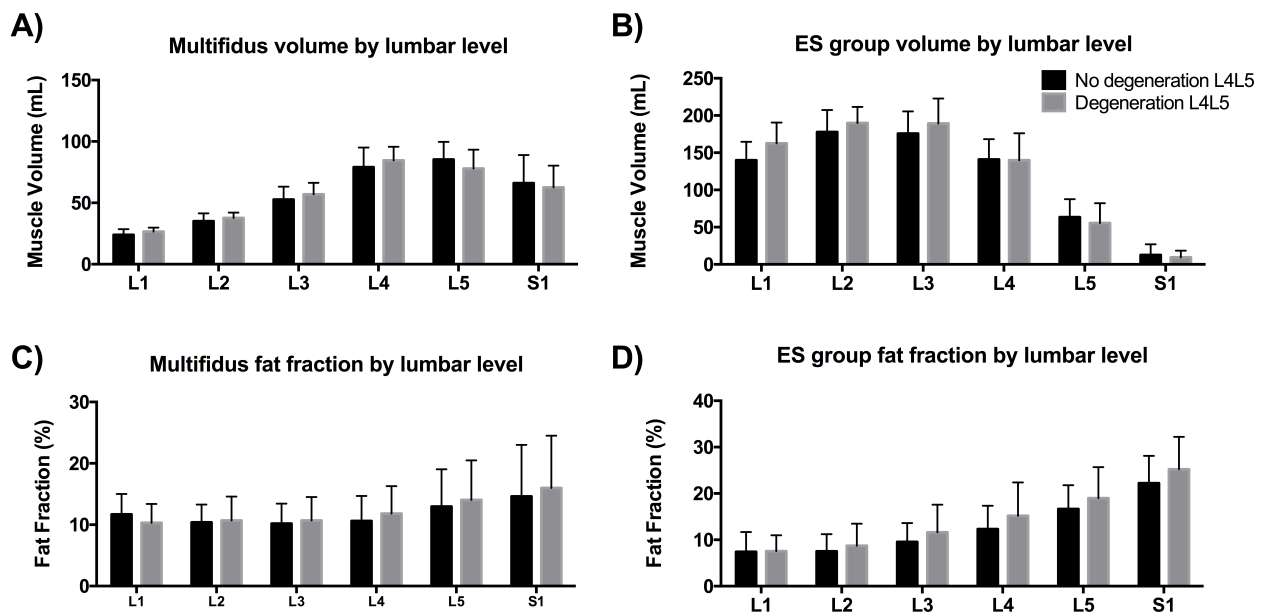
Study 1 Figure 2. A) Muscle volume of the multifidus (white), erector spinae group (ES, horizontal lines), psoas (inclined lines), and quadratus lumborum (QL, black), and **B)** multifidus and ES group fat content by lumbar level (L1-S1). The volume of the ES group was found to be significantly largest ($p < 0.05$) at L2 and L3 (no differences between these two levels), psoas at L4, QL at L3 and L4, and multifidus at L4 and L5 (Fig. 2A). In the superior LS (L1-L4) the largest volume was occupied by the ES group followed by psoas ($P < 0.05$), and finally by multifidus and QL. Contrastingly to superior lumbar levels, at L5 and S1 the ES group was found to be the smallest muscle; while the largest was psoas, which was also significantly ($p < 0.05$) larger than multifidus. Fat fraction of the multifidus was significantly increased at L5 and S1 lumbar levels with respect to the rest of the LS. In the ES, however, the fat fraction significantly increased caudally. The amount of fat fraction present at inferior lumbar levels (L4-S1) was found to be larger ($p < 0.05$) in the ES group, than in the multifidus.



Study 1 Figure 3. Diffusion tensor imaging (DTI) estimates: apparent diffusion coefficient (MD, top) and fractional anisotropy (FA, bottom) for multifidus (black bars) and erector spinae (ES) group (white bars) by lumbar levels. Horizontal lines reflect significant differences between ES muscle group and multifidus muscle. A significant ($p < 0.05$) interaction between muscles (multifidus and ES) and lumbar level were found for MD. The MD of the ES group at L4 and L5 was significantly lower than that of the rest of the spine, while the opposite was found in the multifidus muscle at L5 and S1 levels. Additionally, also in the multifidus, MD was significantly higher at S1 than at L5. Overall, at superior lumbar levels (L1-L3) the MD of the multifidus was smaller ($p < 0.05$) than that of the ES group. Contrastingly, at inferior lumbar level (L4-S1) the MD was significantly smaller in the ES group. A significant interaction between muscle group and lumbar level ($p < 0.05$) were found for FA. In the ES group, FA remained unchanged ($p > 0.05$) from L1 to L4 levels, however at L5 and S1 levels it was significantly higher than the rest of the lumbar levels. Fractional anisotropy was also higher at S1 than at L5. Contrary to these results, FA in the multifidus was found to be significantly lower at L4 and L5 compared to the rest of the lumbar levels. Additionally, FA was different between ES group and multifidus muscle at all levels except L4. In general, FA of the multifidus was higher than that of the ES group from L1 to L4, but lower at L5 and S1.



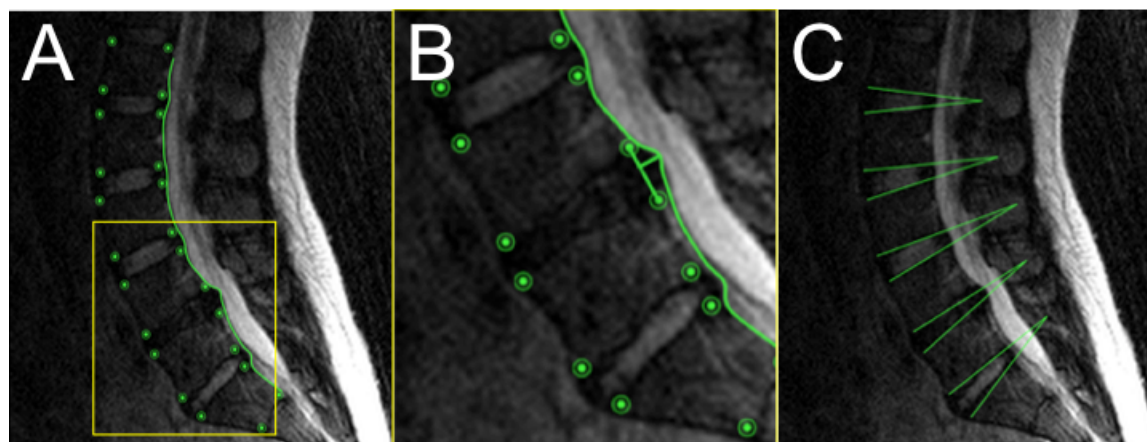
Study 1 Figure 4. Muscle volume and fat fraction of the multifidus (A and C) and ES group (B and D) at lumbar level (L1-S1) by LBP group. No difference in volume or fat fraction of the multifidus or ES when Marines were grouped by low back pain.



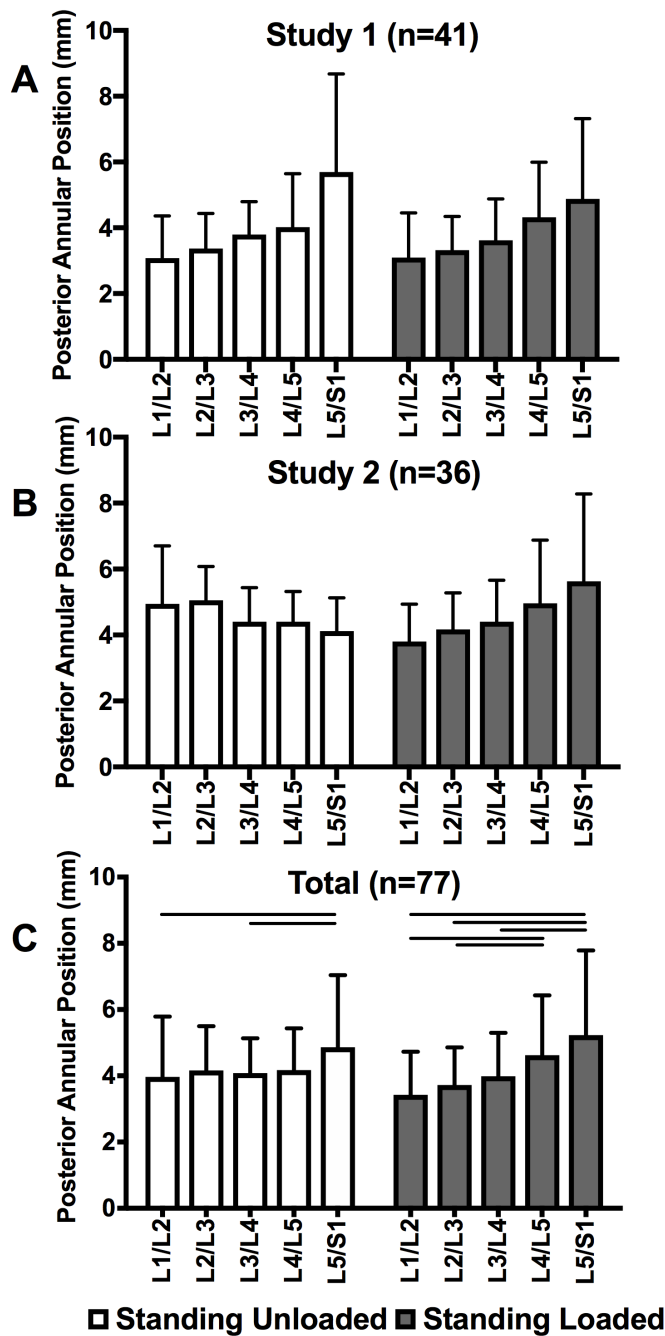
Study 1 Figure 5. Muscle volume and fat fraction of the multifidus (A and C) and ES group (B and D) at lumbar level (L1-S1) by L4-L5 degeneration group. No difference in volume or fat fraction of the multifidus or ES when Marines were grouped by L4-L5 intervertebral disc degeneration.

- **Study 2: INTERVERTEBRAL DISC KINEMATICS IN ACTIVE-DUTY MARINES WITH AND WITHOUT LUMBAR SPINE PATHOLOGY**

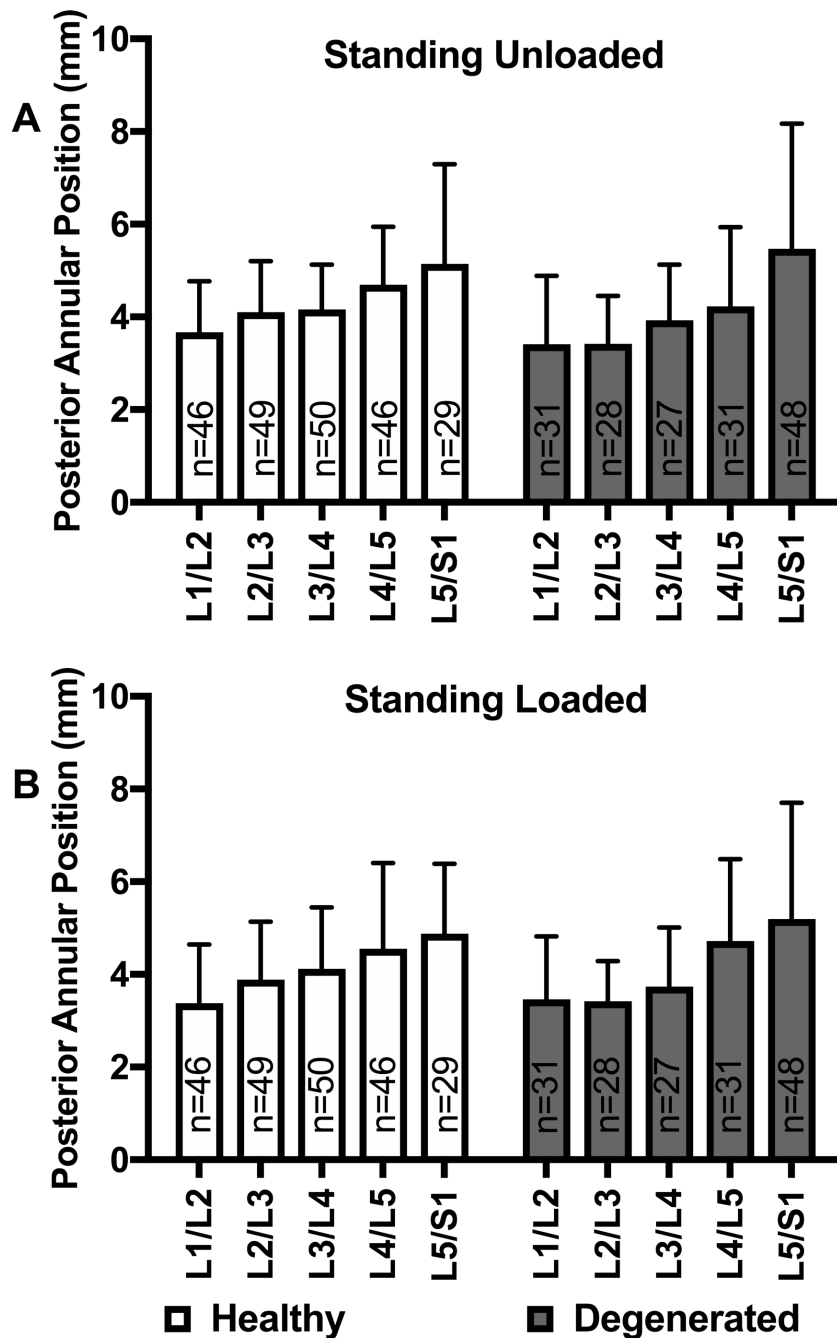
- This was a retrospective analysis of lumbar spine imaging data aimed at understanding IVD kinematics in active-duty Marines. T2-weighted images from an upright magnetic resonance imaging scanner were obtained in standing with and without load for two cohorts of Marines. One cohort of participants donned 11.3kg (n=41) and the other cohort donned 22.6kg (n=36) of operational equipment. IVD health and kinematic changes between loading conditions and across lumbar levels were evaluated using two-way repeated measures analysis of variance (ANOVA) tests in both cohorts. In one cohort of Marines, changes in IVD kinematics with loading were compared between individuals with and without LBP. Linear regression analyses were performed to determine associations between IVD position and kinematic changes with loading.



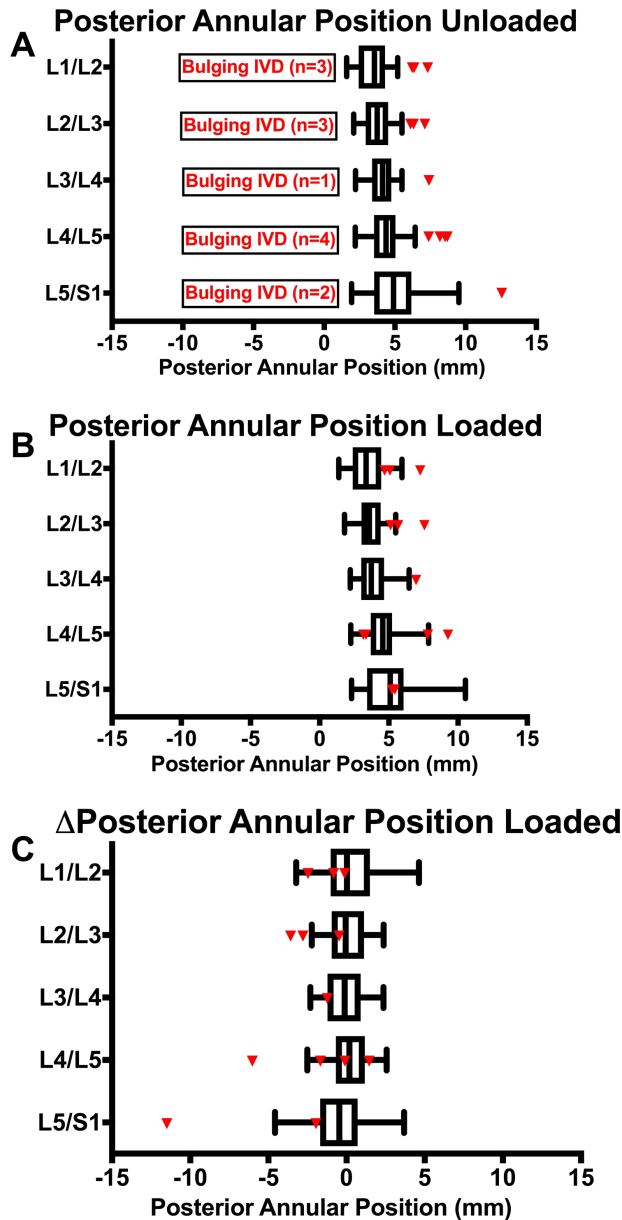
Study 2 Figure 1. Schematic depicting measurement of A.) Disc bulge, B.) Posterior annular position, and C.) Intervertebral angle. Postural measurements were generated from upright MRI images in each load configuration using a previously validated algorithm. Local measurements of lumbar spine posture including intervertebral angles and distances were obtained. *Sagittal intervertebral angle* was measured between the superior and inferior endplates of adjacent vertebrae to describe local changes in lordosis, and distribution of flexion/extension throughout the lumbar spine. Increases in IVD angle correspond to lumbar extension while decreases in IVD angle correspond to lumbar flexion. *IVD height* was measured as the anterior and posterior distances between contiguous vertebrae. Disc bulge was quantified as *posterior annular position*, measured as the maximum distance between a line connecting the posterior aspects of adjacent vertebrae and the posterior IVD border in Matlab.



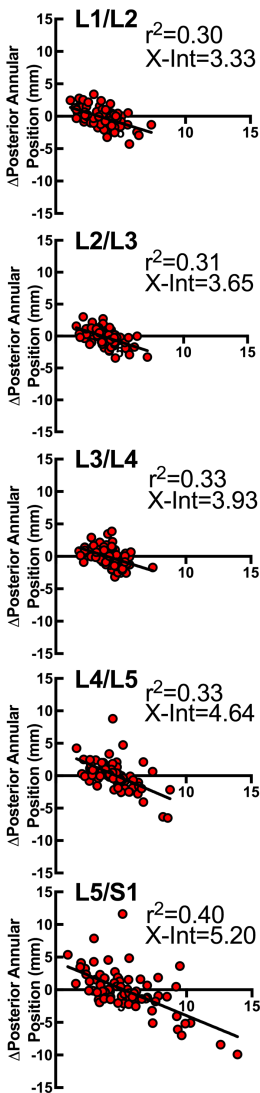
Study 2 Figure 2. IVD Kinematic differences between lumbar levels in 77 active-duty Marines. The Marines' posterior annular position was recorded in the standing unloaded (white) and standing loaded (gray) at all levels. A.) Displays the posterior annular positions with and without load from subjects in Study 1, B.) Displays posterior annular positions with and without load from subjects in Study 2, and C) Displays a significant effect of IVD level in both studies ($p < 0.0001$).



Study 2 Figure 3. IVD degeneration on posterior annular position in the standing unloaded and loaded conditions. A.) The effect of IVD degeneration on posterior annular position in degenerated IVDs is seen in the standing unloaded condition as compared to non-degenerated IVDs and B.) The effect of IVD degeneration on posterior annular position in degenerated IVDs is seen in the standing loaded conditions as compared to non-degenerated IVDs. There was no significant effect of IVD degeneration while standing without or with axial load ($p=0.34$).



Study 2 Figure 4. IVD behavior standing at baseline and with axial load. A.) Posterior annular position of all participants at all lumbar levels when standing unloaded. Individuals whose IVD posterior annular positions were greater than the 95th percentile while standing unloaded were characterized as bulging IVDs, shown in red. B.) Posterior annular position standing with load at all lumbar levels in normal (white) and bulging IVDs (red). C.) Difference between posterior annular position when standing unloaded and standing with load at all lumbar levels in normal (white) and bulging IVDs (red). In the standing unloaded condition, 13% of participants (10/77) showed evidence of baseline disc bulge, defined as posterior annular position greater than 2 standard deviations above the mean. Those individuals with bulging discs at baseline demonstrated a reduction in posterior annular position (decrease in disc bulge) with load.



Study 2 Figure 5. Relationship between posterior annular position when standing unloaded, and the difference in posterior annular position between standing unloaded and standing loaded positions. From top (L1/L2) to bottom (L5/S1), all lumbar levels are shown in all individuals under study with r^2 values. At all levels, there was a significant negative relationship between larger posterior annular position and a reduction in posterior annular position standing with axial load ($p < 0.0002$). There was a significant negative relationship between the baseline posterior annular position and the change in posterior annular position when loaded at all lumbar levels, although the L5/S1 level demonstrated the strongest relationship. Generally, the point of inflection (x-intercept) where IVD bulge began to reduce with axial load began at approximately 3mm posterior in the mid-sagittal plane. IVDs with posterior annular positions less than 3mm tended to exhibit increased magnitude of posterior annular position under axial load, whereas IVDs with posterior annular positions greater than 3mm tended to display increasing posterior annular position reduction under axial load.

- **Study 3:** Berry et al. "AN ENDPLATE-BASED JOINT COORDINATE SYSTEM FOR MEASURING KINEMATICS IN NORMAL AND ABNORMALLY-SHAPED LUMBAR VERTEBRAE." *JAB* 2015; (31): 499-503.
 - Appendix: Study3_JAB_2015.pdf
 - Abstract: Vertebral level-dependent, angular and linear translations of the spine have been measured in 2D and 3D using several imaging methods to quantify postural changes due to loading conditions and tasks. Here, we propose and validate a semi-automated method for measuring lumbar intervertebral angles and translations from upright MRI images using an endplate-based, joint coordinate system (JCS). This method was validated using 3D printed structures, representing intervertebral discs (IVD) at pre-determined angles and heights, which were positioned between adjacent cadaveric vertebrae as a gold standard. Excellent agreement between our measurements and the gold standard was found for intervertebral angles in all anatomical planes ($ICC > 0.997$) and intervertebral distance measurements ($ICC > 0.949$). The proposed endplate-based JCS was compared to the vertebral body based JCS proposed by the International Society of Biomechanics (ISB) using the 3D printed structures placed between three adjacent vertebrae from a cadaver with scoliosis. The endplate-based method was found to have better agreement with angles in the sagittal plane ($ICC = 0.985$) compared to the vertebral body based method ($ICC = 0.280$). Thus, this method is accurate for measuring 3D intervertebral angles in the healthy and diseased lumbar spine.
- **Study 4:** Berry et al. "LUMBAR SPINE POSTURES IN MARINES DURING SIMULATED OPERATIONAL POSITIONS." *JOR* 2017; 35(10).
 - Appendix: Study4_JOR_2017.pdf
 - Abstract: Low back pain has a 70% higher prevalence in members of the armed forces than in the general population, possibly due to the loads and positions soldiers experience during training and combat. Although the influence of heavy load carriage on standing lumbar spine posture in this population is known, postures in other operationally relevant positions are unknown. Therefore, the purpose of this study was to characterize the effect of simulated military operational positions under relevant loading conditions on global and local lumbar spine postures in active duty male US Marines. Secondary objectives were to evaluate if intervertebral disc degeneration and low back pain affect lumbar spine postures. Magnetic resonance images were acquired on an upright scanner in the following operational positions: natural standing with no external load, standing with body armor (11.3 kg), sitting with body armor, and prone on elbows with body armor. Custom software was used to measure global lumbar spine posture: lumbosacral flexion, sacral slope, lordosis, local measures of intervertebral angles, and intervertebral distances. Sitting resulted in decreased lumbar lordosis at all levels of the spine except L1–L2. When subjects were prone on elbows, a significant increase in local lordosis was observed only at L5–S1 compared with all other positions. Marines with disc degeneration (77%) or history of low back pain (72%) had decreased lumbar range of motion and less lumbar extension than healthy Marines. These results indicate that a male Marine's pathology

undergoes a stereotypic set of postural changes during functional tasks, which may impair performance.

- **Study 5:** Berry et al. "THE RELATIONSHIP BETWEEN SKELETAL MUSCLE KINEMATICS AND MUSCLE PHYSIOLOGY IN ACTIVE DUTY MARINES." *JOSPT* 2018.
- Appendix: Study5_JOSPT_2018.pdf
- Abstract: BACKGROUND: The relationship between lumbar spine posture and muscle structure is not well understood.

OBJECTIVES: Investigate the predictive capacity of muscle structure on lumbar spine posture in active duty Marines.

METHODS: Forty-three Marines were scanned using an upright magnetic resonance imaging (MRI) scanner while standing unloaded and standing, sitting, and prone on elbows with body armor. Cobb, horizontal, and sacral angles were measured. Marines were then scanned unloaded in supine using a supine MRI scanner. The imaging protocol consisted of T2 intervertebral disc mapping, high-resolution anatomical, fat-water separation, and diffusion tensor imaging to quantify disc hydration and muscle volume, fat fraction, and restricted diffusion profiles in the lumbar muscles. A stepwise multiple linear regression model was used to identify physiological measures predictive of lumbar spine posture.

RESULTS: The multiple regression model demonstrated that fractional anisotropy (FA) of the erector spinae was a significant predictor of lumbar posture for 7 of 18 dependent variables measured and explained 20% to 35% of the variance in each model. Decreased FA of the erector spinae predicted decreased lordosis, lumbosacral extension, and anterior pelvic tilt.

CONCLUSION: FA is inversely related to muscle fiber size, which is related to isometric force generating capacity of a muscle fiber. This suggests stronger erector spinae muscles predict decreased lordosis, lumbosacral extension, and anterior pelvic tilt in a highly-trained population.

- **Study 6:** Rodriguez-Soto et al. "THE EFFECT OF TRAINING ON LUMBAR SPINE POSTURE AND INTERVERTEBRAL DISC DEGENERATION IN ACTIVE-DUTY MARINES." *Ergonomics* 2017. 60(8): 1055-1063

- Appendix: Study6_Ergonomics_2017.pdf
- Abstract: BACKGROUND: 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 as military training progresses.

PURPOSE: 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.

Study Design: A short-term longitudinal study was conducted.

METHODS: 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 a 0.6T 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 ($\alpha=0.05$).

RESULTS: 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 a larger sacral postural perturbations in response to load.

CONCLUSIONS: Although the posture of the LS changes in response to loading, load-induced postural changes did not change during 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.

- **Study 7:** Rodriguez-Soto et al. "EFFECT OF LOAD MAGNITUDE AND DISTRIBUTION ON LUMBAR SPINE POSTURE IN ACTIVE-DUTY MARINES." *Spine* 2017. 42(5):345-351.
- Appendix: Study7_Spine_2017.pdf
- Abstract: BACKGROUND: Low back pain has been associated with heavy load carriage among military personnel. Although there are data indicating that lumbar spine (LS) postures change in response to load, there no data documenting how the magnitude or distribution of these loads impact postural alignment or disc kinematics.

PURPOSE: 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 hypothesize that when loads were 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.

METHODS: 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 lumbar spine and pelvic postures. Two-way repeated-measures ANOVA and Sidak post-hoc tests were used to compare LS posture as a function of load magnitude and distribution ($\alpha=0.05$).

RESULTS: No changes in LS 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 of the inferior lumbar levels varied in magnitude throughout regions of the disc.

CONCLUSIONS: 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 and the operational efficiency of a Marine carrying load in this configuration needs to be tested.

▪ **What opportunities for training and professional development has the project provided?**

- The grant supported the graduate student researcher stipend for PhD candidate David Berry.

▪ **How were the results disseminated to communities of interest?**

- STUDY 1 was presented at the Orthopaedic Research Society meeting in Las Vegas, NV in 2015, the Military Orthopaedic Surgeon Society meeting in Olympic Valley, CA in 2017, and the Orthopaedic Research Society meeting in San Diego, CA in 2017.
- STUDY 2 was presented at the UC San Diego Medical School Annual Research conference in 2017.
- STUDY 3 was presented at the Orthopaedic Research Society meeting in Las Vegas, NV in April 2015.
- STUDY 3 was accepted for publication in the Journal of Applied Biomechanics, 2015.
- STUDY 4 was presented at the World Congress on Biomechanics in Boston, MA in 2014, the Military Health System Research Symposium in Ft. Lauderdale, FL in 2014, and the Orthopaedic Research Society Conference in San Diego, CA in 2017.
- STUDY 4 was accepted for publication in the Journal of Orthopaedic Research, 2017.
- STUDY 5 was accepted for publication in the Journal of Orthopaedic and Sports Physical Therapy, 2018
- STUDY 5 was presented at the International Society of Biomechanics meeting in Brisbane, Australia in 2017.
- STUDY 6 was accepted for publication in Ergonomics, 2017.
- STUDY 7 was presented at the American Society of Biomechanics meeting in Cincinnati, OH in 2015.
- STUDY 7 was accepted for publication in SPINE, 2017.

▪ **What do you plan to do during the next reporting period to accomplish the goals?**

- We intend on submitting the manuscript for STUDY 1 to SPINE and STUDY 2 to The Spine Journal.

4. **IMPACT:**

▪ **What was the impact on the development of the principal discipline(s) of the project?**

- The results of this study may inform a set of load carriage guidelines to be put in place, through changes in training practices, gear design and/or implementation of exercises to strengthen the

musculature of the spine. Interestingly, the finding that light to moderate axial loading does not exacerbate disc bulge and actually reduces large posterior annular positions does not support that axial load modification is necessary when loads are within the load carriage limit. Overall, the results of these studies will allow researchers to better relate the complex 3D geometry of the lumbar spine in subjects with different levels of lumbar disc and degeneration and muscle health. The role of idiopathic lumbar back pain on lumbar spine kinematics will also allow us to investigate differences in the shape of the lumbar spine between a wide range of positions. The finding that fractional anisotropy (measure of muscle microstructure) is able to predict postural responses in several positions, along with the absence of association between muscle volume and lumbar spine posture, suggests muscle microstructure, but not quantity, is an important predictor of lumbar spine posture. The predicative capacity of a noninvasive measure of muscle quality on maladaptive posture in operational conditions may allow clinicians to tailor rehabilitation protocols to prevent injury, and could potentially be used to predict individuals at risk for lumbar spine injury.

- **What was the impact on other disciplines?**
 - Nothing to report
- **What was the impact on technology transfer?**
 - Nothing to report
- **What was the impact on society beyond science and technology?**
 - The broad impact is that the load recommendations determined from this study can be applied to anyone who works with their spine under awkward or loaded positions.

5. **CHANGES/PROBLEMS:**

- **Changes in approach and reasons for change**
 - Nothing to report
- **Actual or anticipated problems or delays and actions or plans to resolve them**
 - Nothing to report
- **Changes that had a significant impact on expenditures**
 - Nothing to report
- **Significant changes in use or care of human subjects, vertebrate animals, biohazards, and/or select agents**
 - Nothing to report
- **Significant changes in use or care of human subjects**
 - Nothing to report
- **Significant changes in use or care of vertebrate animals.**
 - Nothing to report
- **Significant changes in use of biohazards and/or select agents**
 - Nothing to report

6. **PRODUCTS:**

▪ **Publications, conference papers, and presentations**

Poster Presentation - Berry DB, Rodriguez-Soto AE, Gombatto SP, Jaworski R, Kelly KR, Ward SR. Lumbar spine postures in Marines during simulated operational conditions. Word Congress of Biomechanics, Boston, 2014.

Poster Presentation - Berry DB, Rodriguez-Soto AE, Gombatto SP, Jaworski R, Kelly KR, Ward SR. Lumbar spine postures in Marines during simulated operational conditions. Military Health System Research Symposium, Ft. Lauderdale, 2014. *Honorable Mention: Best Poster.*

Poster Presentation - 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. Orthopaedic Research Society, Las Vegas, 2015.

Poster Presentation - Rodriguez-Soto AE, Stambaugh JR, Su J, Berry DB, Gombatto SP, Kelly KR, Ward SR. Spinal muscle quality changes in physically active individuals with disc degeneration. Orthopaedic Research Society, Las Vegas, 2015.

Poster Presentation - Rodriguez-Soto AE, Berry DB, Palombo L, Valaik E, Kelly KR, Ward SR. The effect of load magnitude and distribution on lumbar spine posture in active-duty Marines. American Society of Biomechanics, Cincinnati, 2015.

Podium Presentation - Stambaugh JR, Rodriguez-Soto AE, Berry DB, Parra C, Gombatto SP, Palombo L, Kelly KR, Ward SR. Structural adaptations of paraspinal musculature in active-duty Marines with intervertebral disc degeneration. Military Orthopedic Surgeon Society, Olympic Valley, 2016.

Poster Presentation - Berry DB, Rodriguez-Soto AE, Parra C, Stambaugh J, Valaik E, Chung C, Ward SR. Noninvasive Assessment of Lumbar Muscle Architecture in Active-Duty Marines. Orthopaedic Research Society Meeting, San Diego, 2017.

Podium Presentation - Berry DB, Rodriguez-Soto AE, Gombatto S, Shahidi B, Chung C, Ward SR. Lumbar Spine Postures in Marines During Simulated Operational Positions. Orthopaedic Research Society Meeting, San Diego, 2017.

Podium Presentation - Berry DB, Shahidi B, Rodriguez-Soto AE, Hughes-Austin J, Kelly KR, Ward SR. Lumbar muscle structure predicts operational postures in active duty Marines. International Society of Biomechanics, Brisbane, 2017.

▪ **Journal publications.**

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; 31(6).

Rodriguez-Soto AE, Berry DB, Jaworski R, Jensen A, Chung CB, Niederburger B, Qadir A, Kelly KR, Ward SR. The effect of training on lumbar spine kinematics and intervertebral disc degeneration in active-duty Marines. *Ergonomics*, 2016; 60(8).

Rodriguez-Soto AE, Berry DB, Palombo L, Kelly KR, Ward SR. The effect of load magnitude and distribution on lumbar spine posture in active-duty Marines. *Spine*, 2017; 42(5).

Berry DB, Rodriguez-Soto AE, Su J, Gombatto SP, Shahidi B, Palombo L, Chung C, Kelly KR, Ward SR. Lumbar spine postures in Marines during simulated operational positions. *Journal of Orthopaedic Research*. 2017; 35(10).

Berry DB, Shahidi B, Rodriguez-Soto AE, Hughes-Austin J, Ward SR. The relationship between skeletal muscle kinematics and muscle physiology in active duty Marines. *JOSPT*. 2018.

Onodera K, Berry DB, Shahidi B, Kelly KR, Ward SR. Intervertebral disc kinematics in active-duty Marines with and without lumbar spine pathology. *In Review: The Spine Journal*.

Berry DB, Rodriguez-Soto AE, Kelly KR, Ward SR. Noninvasive assessment of paraspinal muscle composition in active-duty Marines. *In preparation: Journal of Orthopaedic Research*.

▪ **Books or other non-periodical, one-time publications.**

- Nothing to report

▪ **Other publications, conference papers, and presentations.**

- Nothing to report

▪ **Website(s) or other Internet site(s)**

- Nothing to report

▪ **Technologies or techniques**

- Nothing to report

▪ **Inventions, patent applications, and/or licenses**

- Nothing to report

▪ **Other Products**

- Nothing to report.

7. PARTICIPANTS & OTHER COLLABORATING ORGANIZATIONS

▪ **What individuals have worked on the project?**

Name:	David Berry, PhD
Project Role:	Postdoctoral Scholar
Researcher Identifier (e.g. ORCID ID):	0000-0001-8275-8322
Nearest person month worked:	12
Contribution to Project:	<i>Dr. Berry has participated in upright MRI data collection, data analysis and interpretation of results</i>
Funding Support:	
Name:	Ana Rodriguez-Soto, PhD

Project Role:	Postdoctoral Scholar
Researcher Identifier (e.g. ORCID ID):	0000-0002-9544-547X
Nearest person month worked:	12
Contribution to Project:	<i>Dr. Rodriguez-Soto has participated in supine data collection, data analysis and interpretation of results</i>
Funding Support:	UC MEXUS-CONACYT Doctoral Fellowship Cohort2010

Name:	Dr. Karen Kelly PT, PhD
Project Role:	Primary Investigator
Researcher Identifier (e.g. ORCID ID):	0000-0001-8003-6123
Nearest person month worked:	6
Contribution to Project:	<i>Dr. Kelly has assisted with subject recruitment, subject scheduling, subject consent, and interpretation of results</i>
Funding Support:	Award #new award 03/01/2015-09/30/2016 2.4 calendar Office of Navy Research Program Evaluation of SEAL Delivery Vehicles Unit Level Training The primary aim of this project is to describe the physiological, physical, cognitive and sleep changes that occur during unit level training in a specialized military operational specialty. Award #N1421 10/01/2014-09/30/2015 2.4 calendar Army Special Operation Command Validating Gender-Neutral Standards for Army Special Forces (SFAS) and Ranger (RASP) Selection The primary aims of this project are (1) To determine whether the selection criteria for entry into SFAS/RASP training are accurate predictors of success during SFAS/RASP training; and (2) To establish task-dependent selection and training criteria, which could then be viewed as “gender neutral,” thus meeting the directive of the Secretary of Defense. Award #N1336 09/01/2013-06/30/2015 9.0 calendar Naval Special Warfare-WARCOM 265,000.00 Validating Gender-Neutral Standards for SEAL and SWCC Selection The primary aims of this project are (1) To determine whether the selection criteria for entry into SEAL/SWCC training are accurate predictors of success during SEAL/SWCC training; and (2) To establish task-dependent selection and training criteria, which could then be viewed as “gender neutral,” thus meeting the directive of the Secretary of Defense.

	Award #N1324 03/01/2013-09/30/2016 0.6 calendar Office of Navy Research Independent Evaluation of the ETOWL software program The Office of Navy Research funded Iowa University to develop a virtual Marine that will predict the effect of load on physiological and biomechanical parameters. This project aims to independently test the developed software using “live” active duty military personnel to determine the accuracy of the software program. Award #N1310 03/01/2013-09/30/2016 2.4 calendar Program Manager-Infantry Combat Equipment Personal Protection Equipment Development The purpose of this project is to test a prototype of a novel plate carrier system against the existing plate carrier system in a variety of different load conditions as well as in various environments. Award #N1305 02/01/2012-09/30/2015 1.0 calendar Congressionally Directed Medical Research Programs Lumbar Spine Musculoskeletal Physiology and Biomechanics During Simulated Military Operations The purpose of this project is to determine if lumbar disc and muscle degeneration alters the kinematics response of the lumbar spine to functional positions and loads, predisposing individuals to injury. Award # N1301 10/01/2012-09/30/2015 0.6 calendar Naval Special Warfare-Center Evaluation of VASPER The purpose of this project is to evaluate whether the VASPER system can be used as an adjunctive mode of training in U.S. Navy SEALs. Award # 21A839 03/01/2012-12/30/2015 1.0 calendar Naval Special Warfare-Group 1 Androgen Deficiency in Navy SEALs The purpose of this project is to identify risk factors associated with hormonal imbalance in our elite forces. The study address caloric intake, expenditure, sleep hygiene, alcohol consumption as well as lifestyle choices that may
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	influence hormone production in Navy SEALs. Award # 21A843 09/30/2012-9/30/2015 0.4 calendar Naval Special Warfare-Group 1 Warrior Development The purpose of this project is to identify risk factors associated with decrements in performance and to evaluate means by which to improve the Special Warfighter. Award # N1263 09/30/2012-09/30/2016 0.6 calendar Space and Naval Warfare Systems Command Bureau of Medicine and Surgery Prescriptive Exercise Therapy Program to Reduce Hyper-arousal in Residential Treatment for Active Duty Service Members Diagnosed with Post-Traumatic Stress Disorder To improve the quality of exercise/physical training the service members are receiving while at OASIS, as well as to develop a new exercise program and assist in training the staff at OASIS to maintain the quality of physical training after the project is complete. It is hypothesized that with better exercise and physical training there will be improvements in symptoms of PTSD, better compliance with exercise, increased mood and thus functionality of the persons obtaining treatment at OASIS.
Name:	Dr. Sara Gombatto PT, PhD
Project Role:	Project Advisor
Researcher Identifier (e.g. ORCID ID):	0000-0002-8284-4789
Nearest person month worked:	3
Contribution to Project:	<i>Dr. Gombatto has assisted with upright MRI data collection and interpretation of results</i>
Funding Support:	University Grants Program, San Diego State

Name:	Dr. Samuel Ward PT, PhD
Project Role:	Primary Investigator
Researcher Identifier (e.g. ORCID ID):	0000-0002-4470-155X
Nearest person month worked:	6
Contribution to Project:	Dr. Ward has assisted with project oversight and interpretation of results
Funding Support:	<u>Ongoing Research Support</u> R01 HD073180 (Ward, Samuel) 05/01/13 – 04/30/18 NIH(NICHD) The Physiological Basis of Rotator Cuff Muscle Rehabilitation The goal of this project is to elucidate the structural, mechanical, and physiological consequences of tendinopathy-related muscle atrophy and degeneration after rotator cuff tears in humans. Architectural, passive mechanical, and gene expression profiles will be measured and compared amongst patients with different rotator cuff tear severities. W81XWH-13-2-0043 (Ward, Samuel) 06/01/13 – 05/31/18 (USAMRMC) DOD Lumbar Spine Musculoskeletal Physiology and Biomechanics During Simulated Military Operations The goals of this project are to; 1) understand lumbar spine and lumbar disc kinematics during simulated operational conditions, 2) understand the effect of load and body position on spine and disc kinematics when pathology is present, and 3) understand the influence of muscle structure and physiology on lumbar spine kinematics. R01 HD088437-01A1 (Ward, Samuel) 07/01/2017–03/31/2022 NIH (NICHD) Lumbar Spine Muscle Degeneration Inhibits Rehabilitation-Induced Muscle Recovery The goal of this project is to characterize the structural, physiological, and adaptive potentials of pathological lumbar spine muscles in response to exercise in an attempt to identify sources of resistance to recovery in patients with disc injury.

R01 AR070830-01 (MPI: Ward, Samuel primary) 08/11/2017–7/31/2022
NIH (NICHD)

Non invasive measurements of muscle microstructure assessed by diffusion tensor imaging

Skeletal muscle is important for function, metabolism, and cardiovascular health, and therefore, injury and disease adversely affect quality of life and healthcare costs. Currently, muscle biopsy is the gold standard for diagnosis and monitoring muscle disease and recovery, but is prone to sampling errors and it is invasive. Here we propose a novel strategy to harness diffusion MRI to accurately assess muscle health and pathology.

UCSD Award (PI: Ward and Kalunian) 10/01/2017 – 06/30/2018
Sanford Stem Cell Clinical Center, UCSD Health

Bone Marrow Aspirate Concentrate (BMAC) Treatment for Knee Osteoarthritis

The goal of this project is to understand the clinical and biological effects of a bone marrow aspirate concentrate injection into the knees of patients with moderate or severe osteoarthritis. At this stage, the primary targets are an understanding the variability of clinical and biological measurements as we power a larger clinical trial proposal.

R56 AG052593-01A1 (PI: Villarreal) 09/30/2017 – 08/31/2018
NIH/NIA

Epicatechin Treatment for Sarcopenia

The goal of this project is to understand the effect of epicatechin on skeletal muscle function and recovery in humans.

R01 HD092515A (PI: Alperin) 08/01/2017 – 07/31/2022
NIH (NICHD)

Mechanisms and impact of pregnancy-induced adaptations in pelvic floor muscles.

The goal of this project is to determine whether pregnancy-induced structural and mechanical adaptations in the pelvic floor muscles protect them from birth injury and to elucidate mechanisms that govern pelvic floor muscles' plasticity during pregnancy.

DelNova LLC (PI: Ward) 07/01/2017 – 06/30/2022
Del NOVA

Recovery of Muscle Function after BTX Reverse Agents

The purpose of this project is to develop a reversal agent for BTX-induced muscle weakness.

CTSA Bioinformatics Grant Subaward (PI: Firestein) 06/01/17 – 05/31/18
NIH (UL1TR001442, PI Firestein)

The Physiological Basis of Rotator Cuff Muscle Rehabilitation: The influence of strain and exercise on muscle transcriptional activity and epigenetic control of growth.

	R21HD094566-01 (PI: Alperin) 12/01/2017 – 11/30/2019 NIH/NICHD Injectable Tissue-Specific Extracellular Matrix Hydrogel for Pelvic Skeletal Muscle Regeneration Following Birth Injury The goals of the project are to understand; 1) the histological and biological effects of ECM hydrogel on muscle recovery, and 2) determine if there are positive functional effects of ECM hydrogel on a recovering skeletal muscle after tendon repair. R03 HD094598-01 (PI: Shahidi) 12/01/2017 – 11/30/2019 NIH/NICHD 3D Quantification of Exercise-Induced Recruitment in Pathological Paraspinal Muscles Low back pain (LBP) is a condition that affects a majority of the US population and is responsible for a significant proportion of health care costs and utilization due to the high incidence of recurrent and persistent symptoms. Lumbar spine muscle is compromised in LBP and is a key target for rehabilitation because it is a pivotal component of stability and function. Here, we will characterize the muscle activation patterns during common rehabilitation exercises in a spatially and compositionally sensitive manner in an attempt to identify sources of resistance to recovery in patients with chronic LBP. R21 AR072523-01 (MPI: Ward, Christman) 03/07/2018 – 12/31/2019 NIH/NIAMS Promotion of Skeletal Muscle Recovery Using an Extracellular Matrix Hydrogel in a Rabbit Model of Chronic Rotator Cuff Injury The goals of this project are to measure the biological, biochemical, and histological changes associated with a decellularized matrix injection into degenerated skeletal muscles of the rabbit shoulder. Secondly, the physiological changes associated with muscle recovery will also be explored between.

- **Has there been a change in the active other support of the PD/PI(s) or senior/key personnel since the last reporting period?**
 - Nothing to report
- **What other organizations were involved as partners?**
 - **Organization Name:** San Diego State University
 - **Location of Organization:** San Diego, CA
 - **Partner's contribution to the project**
 - **Collaboration:** Sara Gombatto PT, PhD. Provided assistance during data collection and data interpretation

8. SPECIAL REPORTING REQUIREMENTS

- **COLLABORATIVE AWARDS:** None

- **QUAD CHARTS:** See attached FY1718_QuadChart_W81XWH-13-2-0043

9. **APPENDICES:**

- Quad Chart: FY1617_QuadChart_W18XWH-13-2-0043.pptx
- Study3_JAB_2015.pdf
- Study4_JOR_2017.pdf
- Study5_JOSPT_2018.pdf
- Study6_Ergonomics_2017.pdf
- Study7_SPINE_2017.pdf
- Bibliography_ W81XWH-13-2-0043.pdf

PUBLICATIONS

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; 31(6).

Rodriguez-Soto AE, Berry DB, Jaworski R, Jensen A, Chung CB, Niederburger B, Qadir A, Kelly KR, Ward SR. **The effect of training on lumbar spine kinematics and intervertebral disc degeneration in active-duty Marines.** *Ergonomics*, 2016; 60(8).

Rodriguez-Soto AE, Berry DB, Palombo L, Kelly KR, Ward SR. **The effect of load magnitude and distribution on lumbar spine posture in active-duty Marines.** *Spine*, 2017; 42(5).

Berry DB, Rodriguez-Soto AE, Su J, Gombatto SP, Shahidi B, Palombo L, Chung C, Kelly KR, Ward SR. **Lumbar spine postures in Marines during simulated operational positions.** *Journal of Orthopaedic Research*. 2017; 35(10).

Berry DB, Shahidi B, Rodriguez-Soto AE, Hughes-Austin J, Ward SR. **The relationship between skeletal muscle kinematics and muscle physiology in active duty Marines.** *JOSPT*. 2018.

Onodera K, Berry DB, Shahidi B, Kelly KR, Ward SR. **Intervertebral disc kinematics in active-duty Marines with and without lumbar spine pathology.** *In Review: The Spine Journal*.

Berry DB, Rodriguez-Soto AE, Kelly KR, Ward SR. **Noninvasive assessment of paraspinal muscle composition in active-duty Marines.** *In preparation: Journal of Orthopaedic Research*.

CONFERENCE PRESENTATIONS

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

Berry DB, Rodriguez-Soto AE, Gombatto SP, Jaworski R, Kelly KR, Ward SR. **Lumbar spine postures in Marines during simulated operational conditions.** Military Health System Research Symposium, Ft. Lauderdale, 2014. *Honorable Mention: Best Poster*.

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.** Orthopaedic Research Society, Las Vegas, 2015.

Rodriguez-Soto AE, Stambaugh JR, Su J, Berry DB, Gombatto SP, Kelly KR, Ward SR. **Spinal muscle quality changes in physically active individuals with disc degeneration.** Orthopaedic Research Society, Las Vegas, 2015.

Rodriguez-Soto AE, Berry DB, Palombo L, Valaik E, Kelly KR, Ward SR. **The effect of load magnitude and distribution on lumbar spine posture in active-duty Marines.** American Society of Biomechanics, Cincinnati, 2015.

Stambaugh JR, Rodriguez-Soto AE, Berry DB, Parra C, Gombatto SP, Palombo L, Kelly KR, Ward SR. **Structural adaptations of paraspinal musculature in active-duty Marines with intervertebral disc degeneration.** Military Orthopedic Surgeon Society, Olympic Valley, 2016.

Berry DB, Rodriguez-Soto AE, Parra C, Stambaugh J, Valaik E, Chung C, Ward SR. **Noninvasive Assessment of Lumbar Muscle Architecture in Active-Duty Marines.** Orthopaedic Research Society Meeting, San Diego, 2017.

Berry DB, Rodriguez-Soto AE, Gombatto S, Shahidi B, Chung C, Ward SR. **Lumbar Spine Postures in Marines During Simulated Operational Positions.** Orthopaedic Research Society Meeting, San Diego, 2017.

Berry DB, Shahidi B, Rodriguez-Soto AE, Hughes-Austin J, Kelly KR, Ward SR. **Lumbar muscle structure predicts operational postures in active duty Marines.** International Society of Biomechanics, Brisbane, 2017.

An Endplate-Based Joint Coordinate System for Measuring Kinematics in Normal and Abnormally-Shaped Lumbar Vertebrae

David B. Berry,¹ Ana E. Rodríguez-Soto,¹ Jana R. Tokunaga,² Sara P. Gombatto,² and Samuel R. Ward¹

¹University of California San Diego; ²San Diego State University

Vertebral level-dependent, angular, and linear translations of the spine have been measured in 2D and 3D using several imaging methods to quantify postural changes due to loading conditions and tasks. Here, we propose and validate a semiautomated method for measuring lumbar intervertebral angles and translations from upright MRI images using an endplate-based, joint coordinate system (JCS). This method was validated using 3D printed structures, representing intervertebral discs (IVD) at predetermined angles and heights, which were positioned between adjacent cadaveric vertebrae as a gold standard. Excellent agreement between our measurements and the gold standard was found for intervertebral angles in all anatomical planes (ICC > .997) and intervertebral distance measurements (ICC > .949). The proposed endplate-based JCS was compared with the vertebral body-based JCS proposed by the International Society of Biomechanics (ISB) using the 3D printed structures placed between 3 adjacent vertebrae from a cadaver with scoliosis. The endplate-based method was found to have better agreement with angles in the sagittal plane (ICC = 0.985) compared with the vertebral body-based method (ICC = .280). Thus, this method is accurate for measuring 3D intervertebral angles in the healthy and diseased lumbar spine.

Keywords: lumbar spine, posture, intervertebral angle, upright MRI, scoliosis

Load carriage, tasks, and pathology have been shown to alter lumbar spine posture.^{1,2} High-resolution studies have quantified the kinematics of the lumbar spine in vitro, however the effect of active muscles and passive soft tissue tension are difficult to replicate,³ which makes their correlation to the in vivo condition questionable. In vivo vertebral posture changes have been measured using computerized tomography (CT),⁴ biplanar radiography,⁵ magnetic resonance imaging (MRI),^{6,7} and combinations of multiple imaging modalities.⁸ The development of upright MRI machines has allowed for the study of in vivo postural changes in the bony elements and soft tissues of the spine in relevant and functional positions. Therefore, images acquired using upright MRI include the effects of gravity, muscle activity, motor control, pain, and pathology on the body's posture. Several upright MRI studies of the lumbar spine have focused on only midsagittal angular and translational changes.^{7,9} The kinematic behavior of the intervertebral joint in the axial and coronal planes is often omitted, although altered biomechanical behavior of the spine has been associated with low back pain and other pathologies.³ Therefore, quantifying spine three-dimensional (3D) biomechanical behavior is crucial for understanding the progression of the aging and load-related pathology in the lumbar spine.

In 2002, the International Society of Biomechanics (ISB) proposed a standard joint coordinate system (JCS) for the

spine¹⁰—similar to JCSs used to study the lumbar spine^{3,11}—to facilitate comparisons among various studies of the spine, including abnormally-shaped vertebrae that present in scoliosis. The ISB JCS is centered within the body of each vertebra in the spine, and is based on distinct anatomical landmarks. However, a vertebral body-based approach may not accurately reflect intervertebral angles when vertebrae are nonnormal in shape, as is frequently observed in pathology. Two studies have used an endplate-based JCS to measure 3D kinematics of the lumbar spine using MRI,^{12,13} however the method has not been validated in healthy or pathological conditions. Therefore, the goal of this study was to describe and validate a robust imaging-based method to measure sagittal, coronal, and axial intervertebral angles and distances. To achieve this, we propose a modified endplate-based version of the ISB JCS for the intervertebral joints that allows for accurate measurement of intervertebral angles and distances in 3D. We validated and tested this modified JCS in typically developed and scoliotic cadaveric vertebrae.

Methods

Intervertebral disc models were designed with specified angles and heights at predetermined angles (Figure 1A) and were printed with ABS plastic in 0.01 in (0.025 cm) thick layers using a 3D printer. These models were used as the gold standard for validation of the kinematic measurements in 3 planes of motion: flexion–extension, lateral bending, and axial rotation. Flexion–extension angles ranged from –80° to 80°, in 5° increments from 0° to 40° (and 0° to –40°), and 10° increments from 40° to 80° (and –40° to –80°). Lateral bending angles ranged from –40° (left) to 40° (right) in increments of 5°. Similarly, axial rotation ranged from –15° (left) to 15° (right) in increments of 5°. Additional models were fabricated as gold standards for intervertebral distances; these ranged from 10 mm to 22.5 mm in increments of 2.5 mm, without angulation.

David B. Berry, Ana E. Rodríguez-Soto, and Samuel R. Ward are with the Department of Bioengineering, University of California San Diego, La Jolla, CA. Jana R. Tokunaga and Sara P. Gombatto are with the Doctor of Physical Therapy Program, San Diego State University, San Diego, CA. Samuel R. Ward is also with the Departments of Radiology and Orthopaedic Surgery, University of California San Diego, La Jolla, CA. Address author correspondence to Samuel R. Ward at srward@ucsd.edu.

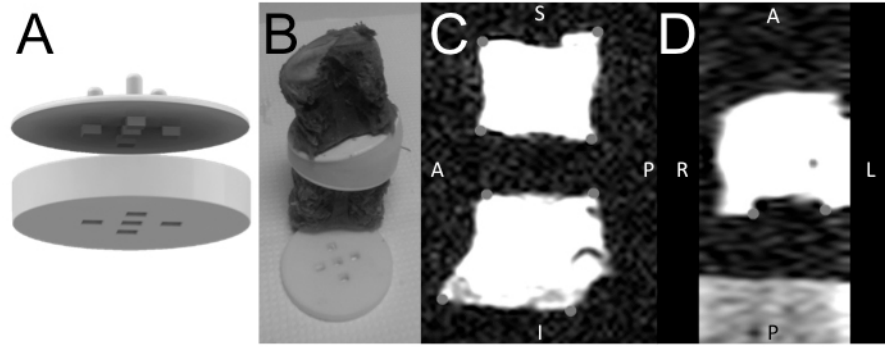


Figure 1 — (A) The 3D printed intervertebral disc model (bottom) and vertebral endplate (top). The endplate model has 5 square pins that press into the holes in the intervertebral disc model, and 3 posts that press into holes drilled into the endplates of the cadaver vertebrae. (B) The vertebra-model-vertebra construct viewed posteriorly. Markers placed on the corners of the vertebrae in sagittal (C) and axially-reformatted (D) T1-weighted magnetic resonance imaging scans.

Four functional spinal units (FSUs) (L2–L3 [$n = 2$], L3–L4 [$n = 2$]) were excised from 4 cadavers (80.0 ± 17.6 years, 2 male) to evaluate the accuracy of the proposed method. All soft tissues surrounding the FSUs, including the intervertebral disc (IVD), were removed. The posterior elements were removed midpedicle to allow for supra-physiological range of motion measurements. Three perforations were drilled on the adjacent endplates of 2 vertebrae (ie, inferior endplate of L2 and superior endplate of L3) in an equilateral triangular shape. These perforations allowed for securing printed IVD models to adjacent vertebrae (Figure 1B). All angular and distance IVD models were placed between the 2 vertebrae of each FSU, one at a time, and scanned using an upright MRI scanner. These data were used to validate the angular and distance measurements taken with the proposed method in normally-shaped vertebrae.

To assess the accuracy of both the proposed modification and current ISB JCS definitions on intervertebral joint angular measurements in the presence of pathology, we used 2 adjacent FSUs (L1–L3) excised from a cadaver with left-convexity scoliosis. The L2 vertebra was wedged, sloping down from the right-posterior to left-anterior corner. The superior (L1–L2) and inferior (L2–L3) FSUs were combined and scanned as a stacked construct in multiple known positions. The angular position of the superior FSU (L1–L2) was measured from 5° of flexion to 5° of extension and from 5° of left lateral bending to 5° of right lateral bending, both in 5° increments. The angular position of the inferior FSU (L2–L3) was measured from 10° of flexion to 10° of extension and 10° of left lateral bending to 10° of right lateral bending, in increments of 10° . Angular measurements of the intervertebral joint were taken using the proposed endplate-based JCS and ISB-recommended vertebral body-based JCSs, applied to abnormally-shaped vertebrae.

Each vertebra-model-vertebra construct was scanned with a 0.6-T upright MRI (Fonar Corporation, Melville, NY). Sagittal T1-weighted images (TR: 427 ms, TE: 17 ms, NEX: 1) were obtained with a matrix size of 256×256 , field of view of 35 cm, thickness of 3 mm, and scan time of 2:30.⁶

Markers were placed on the corners of each vertebra on each sagittal magnetic resonance image (Figure 1C) using OsiriX (Geneva, Switzerland).¹⁴ A second set of markers was placed on each pedicle at the same distance from the posterior vertebral body in axially-reformatted images (Figure 1D). The position of each marker was imported into Matlab (Mathworks, Natick, MA) and postprocessed using custom software to fit planes to both endplates

of each vertebra using principle component analysis (PCA). A right-handed coordinate system was defined for the superior C_s and inferior C_i endplates of each vertebra. The origin of the coordinate system was placed at the geometric centroid of the endplate. The Y-axis was defined as the vector normal to the plane fit to each endplate from the PCA, pointing cephalad. The orientation of the Z-axis was defined as a line parallel to a line connecting markers on the pedicles, projected onto the plane fit to the endplate, with the positive axis pointing to the right. The X-axis was defined as the vector orthogonal to the Y- and Z-axes, pointing anterior (Figure 2). For a given FSU, the rotation matrix (R) between the superior endplate of the inferior vertebra, and the inferior endplate of the superior vertebra, was calculated as $R = C_s \cdot inv(C_i)$. The rotation matrix follows the Y-Z-X convention defined as follows: where θ , ϕ , and ψ are the rotations in radians around the Y-, Z-, and X-axes, respectively.

$$R = R_y(\theta) R_z(\phi) R_x(\psi) = \begin{bmatrix} R_{11} & R_{12} & R_{13} \\ R_{21} & R_{22} & R_{23} \\ R_{31} & R_{32} & R_{33} \end{bmatrix} \quad (1)$$

$$= \begin{bmatrix} \cos \phi \cos \theta & \sin \psi \sin \theta - \cos \psi \cos \theta \sin \phi & \cos \psi \sin \theta + \cos \theta \sin \phi \sin \psi \\ \sin \phi & \cos \phi \cos \psi & -\cos \phi \sin \psi \\ -\cos \phi \sin \theta & \cos \theta \sin \psi + \cos \psi \sin \phi \sin \theta & \cos \psi \cos \theta - \sin \phi \sin \psi \sin \theta \end{bmatrix}$$

The Euler angles that describe the orientation of the superior vertebra with respect to the inferior vertebra are θ , ϕ , and ψ . We solved for the Euler angles θ , ϕ , and ψ from the rotation matrix R and obtained the following:

$$\phi = \sin^{-1}(R_{21}), \theta = \sin^{-1}\left(\frac{R_{31}}{-\cos^{-1}(\phi)}\right), \psi = \sin^{-1}\left(\frac{R_{23}}{-\cos^{-1}(\phi)}\right) \quad (2)$$

The rotations around each off-plane axis were removed to maintain the true in-plane rotation.

Anterior and posterior intervertebral distances were defined as the Euclidean distance between the centroid of the superior and inferior corners of adjacent vertebra, respectively. Central intervertebral distance was defined as the Euclidean distance between the origins of adjacent endplates.

A power analysis was conducted on pilot data to determine the number of samples needed to observe a correlation between

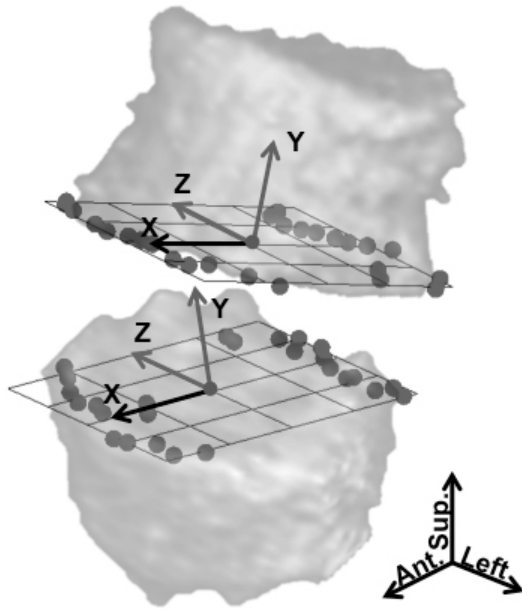


Figure 2 — Joint coordinate systems for adjacent vertebral endplates defined from markers placed on the corners of the vertebrae. The Y-axis is normal to the best-fit plane for the endplate of the vertebrae. The Z-axis is the projection of a line connecting similar landmarks on the pedicles of the vertebra onto the plane. The X-axis is normal to the Y- and Z-axis, pointing anterior.

expected and measured intervertebral angles, which yielded a necessary sample size of 4 ($p = 0.98$, $\alpha = .05$, $\beta = 0.20$). The degree of agreement between expected and measured intervertebral angles and IVD distances was assessed using intraclass correlation coefficients (2,1) (ICC[2,1]). For the abnormally-shaped vertebrae, ICC(2,1) were calculated between the gold standard of the intervertebral angles and measurements taken with both the proposed and the ISB JCSs. Average absolute error (AAE) was calculated for each intervertebral angle and intervertebral distance. The coefficient of determination (r^2) was calculated to determine the total variance accounted for by the proposed method. Root mean square error (RMSE) was calculated to determine the magnitude of the error in the measurements. All statistics were performed using SPSS (Version 21, IBM, Armonk, NY). All data are reported as mean $\pm$ 95% confidence interval (CI) unless otherwise stated.

Results

In normally-shaped vertebrae, excellent agreement and low absolute error were observed when compared with the gold standard (Table 1, Figure 3). Intervertebral angles in the sagittal, coronal, and axial planes had an ICC $> .990$, and anterior, center, and posterior intervertebral distance had an ICC $> .949$. The average absolute errors for the sagittal, coronal, and axial intervertebral angle measurements were 0.77° , 1.34° , and 0.78° , respectively.

In addition, angular positions were measured using the cadaveric vertebrae of 2 adjacent scoliotic FSUs and IVD models of known angles. We found an excellent agreement between the gold standard and measured intervertebral angles using the proposed modification to the ISB JCS in the sagittal and coronal planes for both functional spinal units (ICC $> .985$; Figure 4A). Axial angles were identical since the Z-axis definition is the same in both JCSs. However, the ISB JCS definition yielded poor agreement between expected and measured intervertebral angles for the superior FSU in the sagittal plane (ICC = .280) and good agreement in the coronal plane (ICC = .896). The intervertebral angle for the superior FSU in the sagittal plane of motion was underestimated by approximately 13° flexion in every position with the ISB JCS definition (Figure 4B). For the inferior FSU, there was excellent agreement between the expected and measured angles using the ISB JCS definitions in the sagittal and coronal planes (ICC $> .940$).

Discussion

The proposed endplate-based version of the ISB JCS for the intervertebral joint yields accurate measurements of angles and distances in all 3 planes of motion for normally- and abnormally-shaped vertebrae. When the standard ISB JCS was used to measure joint angles in samples with severe shape abnormalities, measurement errors were profound. These errors are due to the fact that the ISB JCS defines the inferior–superior vector by the line that passes through the center of the inferior to the superior endplate, forming a perpendicular vector to the endplate planes only when they are parallel, which is not the case in pathological vertebrae.

There are a few potential limitations to this study. First, markers defining the vertebral body boundaries could have been placed inaccurately on anatomical landmarks. The coefficient of variation (CV) has previously been calculated to investigate intra- and interuser precision using this method;⁶ intervertebral angles and distances had a CV of 0.58° and 0.38 mm, respectively. This result, coupled

Table 1 Results of validation of intervertebral angles and intervertebral distances

	Sagittal	Coronal	Axial	Anterior Intervertebral Distance	Center Intervertebral Distance	Posterior Intervertebral Distance
r^2	.999	.995	.990	.947	.946	.942
ICC (95% CI)	1 (1–1)	0.999 (0.998–0.999)	0.997 (0.994–0.999)	0.955 (0.243–0.989)	0.954 (0.229–0.989)	0.949 (0.156–0.988)
RMSE $\pm$ SD	$0.95^\circ \pm 2.16^\circ$	$1.66^\circ \pm 2.12^\circ$	$1.09^\circ \pm 1.63^\circ$	1.08 mm ± 2.08 mm	1.10 mm ± 2.08 mm	1.14 mm ± 2.13 mm
AAE $\pm$ SD	$0.77^\circ \pm 0.55^\circ$	$1.34^\circ \pm 1.03^\circ$	$0.78^\circ \pm 0.79^\circ$	1.60 mm ± 1.08 mm	1.62 mm ± 1.09 mm	1.74 mm ± 1.13 mm

Abbreviations: ICC = intraclass correlation coefficient; CI = confident interval; RMSE = root mean square error; SD = standard deviation; AAE = average absolute error.

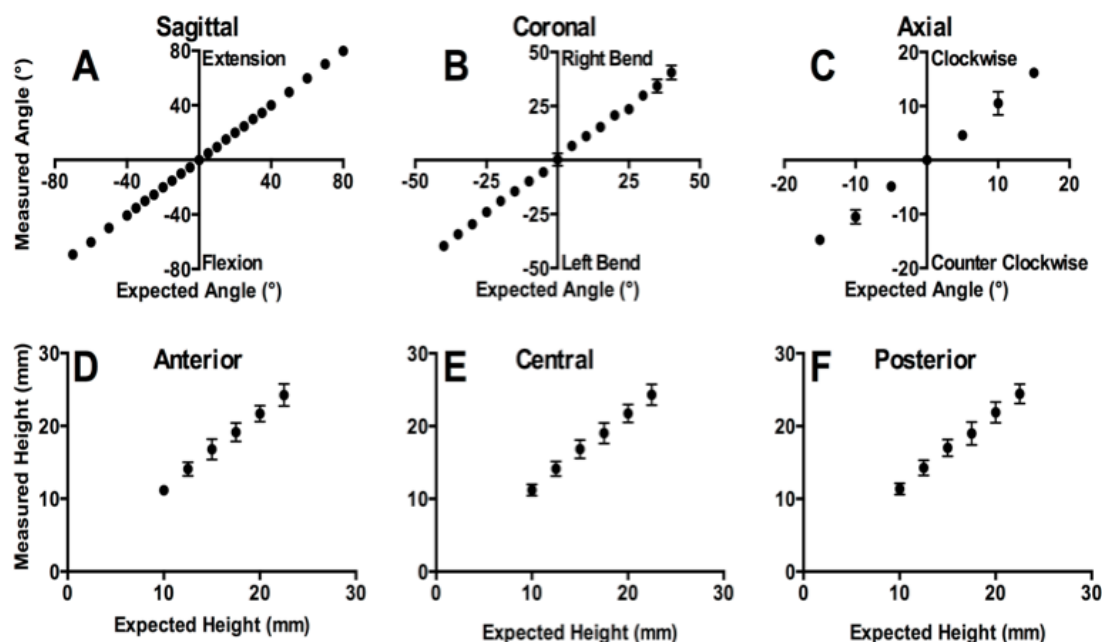


Figure 3 — Measured vs expected sagittal (A), coronal (B), and axial (C) intervertebral angles, and anterior (D), central (E), and posterior (F) disc distances.

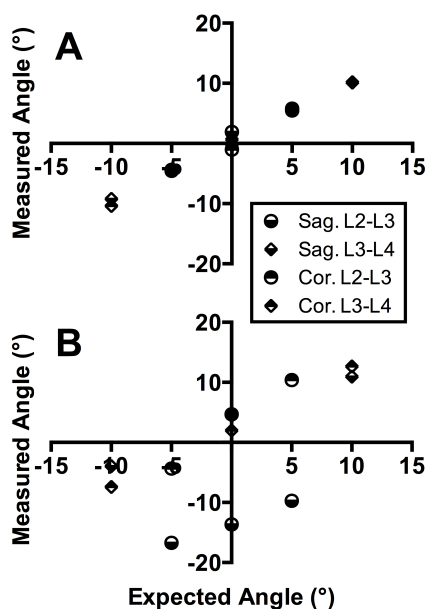


Figure 4 — Scoliotic stacks consisting of 3 vertebrae and 2 intervertebral disc models were scanned at known angles and evaluated using the proposed (A) and International Society of Biomechanics (B) joint coordinate system definitions. Superior (circle) and inferior (triangle) intervertebral angles were measured in the sagittal (bottom filled) and coronal (top filled) planes. Axial measurements were identical between the two joint coordinate systems and are not shown. Data are shown as single measurement. Sag. = sagittal; Cor. = coronal.

with the highly-accurate ICC shown in the study, indicates that anatomical landmark identification does not limit this algorithm's ability to measure intervertebral angles. Second, we chose to validate this algorithm with a clinically-relevant pulse sequence in an upright MRI machine. This pulse sequence has a short scan time to minimize motion artifact and patient pain, resulting in decreased

image resolution. While we did not evaluate the influence of voxel dimensions on the accuracy of the algorithm, we suspect accuracy will decline with increasing voxel dimensions. The small lower limit of the 95% CI observed for the height measurements (Table 1) is likely due to the large in-plane voxel dimensions used in this study; the accuracy of distances is limited by in-plane resolution. In addition, the small sample size used decreases the generalizability of the results, particularly for the scoliotic vertebrae. For a case of extreme pathology, the difference between the 2 methods is apparent. However, these techniques will measure the same angle for normally-shaped vertebrae. The relationship between vertebral body deformation and error measuring intervertebral angles using the ISB JCS is unknown and needs further investigation.

The proposed endplate-based modification to the ISB JCS can be used to quantify postural or kinematic changes that occur in the lumbar spine, both in people with healthy and pathologic spines. This study is the first to validate a method of measuring intervertebral angles and distances from MRI images, using an endplate-based JCS and a well-established gold standard. In the future, this approach can be used to study postural changes that result from load, specific tasks, and back pain in patients with normal and abnormal spine morphology.

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Research Article

Lumbar Spine Postures in Marines During Simulated Operational Positions[†]

Running title: Lumbar Spine Postures in Marines

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AUTHORS' CONTRIBUTIONS

DB, AR, JS, SG, BS, LP, and AJ collected data. DB, AR, JS, SG, BS, LP, and AJ designed the research question. DB, AS, JS, BS, and CC processed the data. DB processed statistics and drafted the paper. All authors have read and approved the final submitted manuscript.

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ABSTRACT: Low back pain has a 70% higher prevalence in members of the armed forces than in the general population, possibly due to the loads and positions soldiers experience during training and combat. Although the influence of heavy load carriage on standing lumbar spine posture in this population is known, postures in other operationally relevant positions are unknown. Therefore, the purpose of this study was to characterize the effect of simulated military operational positions under relevant loading conditions on global and local lumbar spine postures in active duty male US Marines. Secondary objectives were to evaluate if intervertebral disc degeneration and low back pain affect lumbar spine postures. Magnetic resonance images were acquired on an upright scanner in the following operational positions: natural standing with no external load, standing with body armor (11.3 kg), sitting with body armor, and prone on elbows with body armor. Custom software was used to measure global lumbar spine posture: lumbosacral flexion, sacral slope, lordosis, local measures of intervertebral angles, and intervertebral distances. Sitting resulted in decreased lumbar lordosis at all levels of the spine except L1–L2. When subjects were prone on elbows, a significant increase in local lordosis was observed only at L5–S1 compared with all other positions. Marines with disc degeneration (77%) or history of low back pain (72%) had decreased lumbar range of motion and less lumbar extension than healthy Marines. These results indicate that a male Marine's pathology undergoes a stereotypic set of postural changes during functional tasks, which may impair performance. This article is protected by copyright. All rights reserved

Keywords: upright MRI, lumbar spine posture, intervertebral disc, low back pain

INTRODUCTION

Low back pain (LBP), one of the most common musculoskeletal complaints, affects between 70% and 85% of the population at any given time,¹ and is a particularly large problem in the military. In 2013, military medical encounters due to back pain were greater than any other major medical condition (975,000).² Additionally, the Veterans Health Administration has seen an increasing trend in the incidence of LBP in recent years.³ These trends yield a large financial burden on the Departments of Defense and Veterans Affairs to provide health care to these individuals, costing over \$2 billion annually.^{4,5}

Members of the US Armed Forces are a unique population due to the physical demands required by their job. Increased rates of intervertebral disc (IVD) degeneration have been associated with increased military service, time, and age,⁶ and individuals with IVD degeneration have been shown to have a higher incidence of LBP.⁷ The increased rates of reported LBP in Marines may be attributed to pathophysiologic changes occurring in the lumbar spine, resulting from heavy loads and unusual positions experienced in training and combat.⁸ Until recently, these ideas had not been measured directly. Rodriguez-Soto et al.⁹ observed that heavy loading creates postural changes in standing that are not further exacerbated by the duration of loading or activity (e.g., marching with load). Although standing with heavy loads is not necessarily an injury- or pain-provoking activity in these subjects, the effect of other operationally relevant tasks and positions on posture have not been studied. For example, military members are often required to maintain static positions such as standing, sitting, and prone for extended periods of time (depending on military occupational specialty), which may contribute to LBP. In particular, members of the military have reported LBP induced by prolonged sitting in vehicles, suggesting sitting posture may also be related to back pain in this population.¹⁰

The influence of IVD degeneration and LBP on posture has also been studied in a male civilian population. Subjects with idiopathic LBP have been found to have less overall lordosis, but increased local lordosis in the upper lumbar spine compared with healthy controls.¹¹ In subjects with disc

degeneration and LBP, decreased segmental range of motion between extension and flexion positions was found at levels with degenerated IVDs compared to levels with nondegenerated IVDs.^{12,13}

Understanding the relationship between the health and posture of the lumbar spine is crucial to understanding why subjects may be at risk of developing LBP. However, the results of the previous studies may not translate to the military due to the difference in age, loads worn in training, and physical activity level. The influence of IVD degeneration and LBP on posture in operationally relevant positions has not been studied in a military population, and may provide insight into whether task modification would be recommended for military personnel with these pathologies.

Therefore, the purpose of this study was to investigate changes in lumbar spine posture, in a variety of operationally relevant positions, with the minimum operationally relevant load, in active duty male Marines. Secondary analyses were performed to compare lumbar spine postures between Marines with and without IVD degeneration, and with and without LBP. We hypothesized that Marines with IVD degeneration or LBP would have less lower lumbar lordosis in operationally relevant positions than healthy Marines.

METHODS

Participants

Forty-three active duty male Marines from the Marine Corps Base Camp Pendleton 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 Marines gave oral and written consent to participate.

Low Back Pain Evaluation

The Baecke questionnaire on habitual physical activity was administered to classify a Marine's daily work, leisure, and sport-related activity levels.¹⁴ The Roland-Morris Disability Questionnaire data were collected to assess how a participant's LBP limits the ability to perform everyday activities.¹⁵ We

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used another questionnaire to assess self-reported LBP on an 11-point numeric rating scale (0–10; 0 = no pain, 10 = worst imaginable pain) at the time of data collection, duration of the current episode of LBP (if applicable), duration of LBP since the first episode (if applicable), and number of LBP episodes within the last year. Additionally, volunteers were asked to report the activities that aggravated or eased their LBP symptoms.

Imaging

Marines were scanned using an upright 0.6T magnetic resonance imaging (MRI) scanner (Upright Multi-Position MRI, Fonar Corporation, Melville, NY) and a planar coil. An elastic band was used to hold the coil against the volunteer's lumbar spine between the L1–S1 levels while standing. The band was secured to hold the coil in place while retaining the volunteer's natural position. A three-plane localizer (TR = 1254 ms, TE = 100 ms, FOV = 34 cm, matrix = 256×256 , in-plane resolution = $1.33\text{mm} \times 1.33\text{mm}$, THK = 9 mm, NEX = 1, time = 0:17) and sagittal T2-weighted images (TR = 1974 ms, TE = 160 ms, FOV = 35 cm, matrix = 224×224 , in-plane resolution = $1.56\text{mm} \times 1.56\text{mm}$, THK = 3 mm, gap = 0 mm, NEX = 1, time = 2:12) were acquired.

Load Carriage and Position Tasks

Marines were scanned in the following positions: standing without load, standing with body armor (11.3 kg), sitting with body armor, and prone on elbows with body armor (Fig. 1). Positions with external load were randomized to control for the cumulative effects of loading over time. Body armor was selected for the load configuration in the current study, because it is the minimum protective equipment Marines are required to wear at all times during military operations/training. Marines were not instructed on how to assume each position, but were asked to hold each position steady for the duration of the MRI acquisition. At the end of each scan, volunteers were asked to rate their level of LBP in each position on an 11-point numeric rating scale. Volunteers were classified as having increased pain in a position if LBP on the numeric rating scale was greater than baseline levels when standing without load.

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Disc Grading

Lumbar IVD degeneration was assessed by an experienced radiologist (C.C.) using the Pfirrmann grading scale.¹⁶ The Pfirrmann classification system categorizes IVD degeneration from Grade I (normal) to Grade V (severely degenerated), based on nucleus pulposus signal strength and homogeneity, and intervertebral distance. Marines were separated into degenerated or nondegenerated groups based on the Pfirrmann grade of the L5–S1 IVD. A threshold of Pfirrmann Grade III was used to define spine degeneration because data support that IVD biomechanics begin to change at this grade.^{17–19}

Kinematic Measurements

Postural measurements were generated from upright MRI images in each position, using a previously validated algorithm.²⁰ Briefly, digital seed points were manually placed on the corners of the vertebral body and on the posterior elements of each vertebra using OsiriX.²¹ The locations of the seed points were imported into MATLAB (MathWorks Inc., Natick, MA) and used to define an endplate-based joint coordinate system applied to the superior and inferior endplate of each vertebra (L1–S1).

Global and local measurements of lumbar spine posture, intervertebral angles, and distances were calculated for all positions (Fig. 2). Global measurements included angle with respect to the horizontal, sacral slope, and sagittal Cobb angle. The angle with respect to the horizontal is defined as the angle between a line connecting the geometric centroids of L1 and S1, and a line perpendicular to gravity, pointing anteriorly. This measurement indicates the degree of lumbosacral flexion or extension of the entire lumbar spine in the sagittal plane. For the prone loaded position, images were rotated 90 degrees to allow for comparison of the measurements from the prone position to other positions (Fig. 1). The sacral slope is defined as the angle between the superior endplate of S1 and the horizontal, and describes the orientation of the sacrum to provide an estimate of pelvic rotation. The sagittal Cobb angle is used extensively to measure the curvature of the spine and is defined from the superior endplate of L1 to the superior endplate of S1.²² Sagittal intervertebral angles were measured between

the superior and inferior endplates of adjacent vertebrae to describe local changes in lordosis and distribution of lordosis throughout the lumbar spine. Intervertebral distances were measured as the anterior, central, and posterior Euclidean distance between planes fit to the endplates of each vertebra.

Statistical Analysis

Demographic comparisons of age, height, weight, and body mass index (BMI) between volunteers with and without IVD degeneration or LBP were evaluated using independent t-tests. Global kinematic measurements (angle with respect to the horizontal and Cobb angle) were compared using one-way repeated measures analysis of variance (ANOVA) tests with post hoc Sidak tests to identify differences between positions. Local measurements (intervertebral angles and heights) were analyzed using two-way repeated measures ANOVA tests with post hoc Sidak tests to identify differences between positions and intervertebral levels. Secondary analyses were conducted to determine kinematic differences in Marines with and without LBP or IVD degeneration. Volunteers were grouped based on presence of LBP on the day of data collection (yes, no), when the current episode of LBP began (chronic >6 months, acute <6 months, none), history of LBP (yes, no) and positional back pain (extension, neutral, sitting, none), and degeneration of the lumbar spine assessed by Pfirrmann grading (L5–S1 IVD > Grade II). Separate two-way ANOVAs (position × LBP or IVD variable) with post hoc Sidak tests were performed to identify differences for each variable describing LBP or disc degeneration. To investigate the relationship between disc degeneration, LBP, and segmental range of motion, we calculated the difference in postural measurements between the sitting and prone positions. Differences in segmental range of motion for each variable describing LBP or disc degeneration were compared using independent t-tests. The threshold for significance (α) was set to 0.05 for all analyses. η^2 and Cohen's d effect sizes are reported for ANOVA and post hoc tests respectively. Analyses were conducted using IBM SPSS Statistics 20.0 (IBM, Armonk, NY). All data are reported as mean $\pm$ standard deviation.

RESULTS

Marine Demographics

Complete image data sets were obtained from 43 male Marines (mean age = 26.8 ± 6.4 years, height = 1.78 ± 0.07 meters, weight = 82.04 ± 9.85 kg, BMI = 25.90 ± 2.86 kg/m²; Table 1). IVD degeneration assessed using the Pfirrmann classification system (>2) was observed in 125 out of 215 (58%) of all graded lumbar IVDs (Fig. 3). Based on the Pfirrmann grades (>2) of the L5–S1 IVD, 33 of 43 (77%) Marines were in the degenerated group, and 10 were in the nondegenerated group.

The Baecke and Roland-Morris questionnaires were collected from 32 Marines. The average Baecke score was 9.96 ± 1.61 (range: 7.25–13.25). The Roland-Morris average score was 2.69 ± 3.25 (range: 0–13, 22/32 had score >0). Back pain history information was collected from 39 Marines. At the time of data collection, 16 of 39 (41%) Marines reported current back pain (7 chronic LBP, 9 acute LBP), and 28 of 39 (72%) Marines reported previously experiencing at least one episode of LBP. Marines self-reported that sitting, standing for extended periods of time, and hiking with a pack resulted in increased LBP. Additionally, lying supine and positions of lumbar spine extension were reported to decrease LBP in this population. Positional back pain information was collected from 32 Marines. Seven participants reported LBP when standing unloaded (22%), 9 when standing loaded (28%), 6 when sitting loaded (19%), and 11 when prone loaded (34%).

Effect of Position on Global Posture

Significant differences occurred in lumbosacral flexion ($p < 0.001$, $\eta^2 = 0.22$), sacral slope ($p < 0.001$, $\eta^2 = 0.51$), and lumbar lordosis ($p < 0.001$, $\eta^2 = 0.60$) between tasks (Fig. 4). Compared to standing without load, standing wearing body armor resulted in a small decrease in sacral slope (3° , $p = 0.006$, $d = 0.19$), indicating slight posterior tilt of the pelvis, while lumbosacral flexion and lumbar lordosis remained constant. Decreased sacral slope ($>21^\circ$, $p < 0.001$, $d = 0.33$ -1.03) and lumbar lordosis ($>30^\circ$, $p < 0.001$, $d = 0.92$ -1.18) were observed when sitting loaded compared with other positions. No change in lumbosacral flexion was observed during sitting relative to standing positions.

Compared with standing unloaded, when Marines were prone on elbows, lumbosacral flexion increased (5.5° , $p = 0.016$, $d = 0.40$), sacral slope decreased (14° , $p < 0.001$, $d = 0.98$), and lumbar lordosis increased (6° , $p = 0.001$, $d = 0.33$). Interestingly, both sitting and prone on elbows positions resulted in significantly decreased sacral slope compared to standing without load, indicating posterior pelvic tilt.

Effect of Position on Local Posture

Local intervertebral angles were measured to identify regional contributions to overall lumbar posture. A significant effect of level ($p < 0.001$, $\eta^2 = 0.49$), position ($p < 0.001$, $\eta^2 = 0.54$), and an interaction of level x position ($p < 0.001$, $\eta^2 = 0.30$) for intervertebral angles was found. When volunteers were sitting, a significant decrease in lordosis was evident at all levels of the lumbar spine except at L1–L2 (Fig. 5). Larger decreases in lordosis were found in the inferior lumbar spine than in the superior lumbar spine when seated. When Marines were prone on elbows, the only level where a significant increase in lordosis was found was L5–S1, indicating that overall lordosis changes in this position originate from L5–S1.

Intervertebral distance was measured to identify local IVD compression and excursion in different postures (Fig. 6). A significant effect of level ($p < 0.001$), position ($p < 0.001$), and an interaction of level x position ($p < 0.001$) for anterior ($\eta^2_{\text{level}} = 0.31$, $\eta^2_{\text{position}} = 0.37$, $\eta^2_{\text{interaction}} = 0.13$), central ($\eta^2_{\text{level}} = 0.06$, $\eta^2_{\text{position}} = 0.21$, $\eta^2_{\text{interaction}} = 0.01$), and posterior ($\eta^2_{\text{level}} = 0.07$, $\eta^2_{\text{position}} = 0.12$, $\eta^2_{\text{interaction}} = 0.05$) intervertebral distance was found. Intervertebral distance changes paralleled changes in intervertebral lordosis at all levels. The sitting position resulted in anterior compression from L2–L3 to L5–S1 ($p < 0.009$, $d = 0.62 - 0.90$), and posterior distraction from L3–L4 to L5–S1 ($p < 0.017$, $d = 0.20 - 0.38$), with magnitude of effect increasing inferiorly through the lumbar spine. Additionally, sitting resulted in a slight loss of central intervertebral distance from L2–L3 to L5–S1 ($p < 0.011$, $d = 0.15 - 0.31$) compared with standing unloaded, with magnitude of effect increasing inferior through the lumbar spine. Similarly, the prone on elbows position induced an anterior intervertebral distance distraction at L5–S1 ($p < 0.001$, $d = 0.28 - 0.68$).

Effect of Intervertebral Disc Degeneration on Lumbar Spine Posture

No lumbar spine postural differences were found between Marines with and without disc degeneration at L5–S1. However, Marines with IVD degeneration at L5–S1 were found to have less whole lumbar range of motion, measured as the Cobb angle between sitting loaded and prone on elbows loaded positions, than Marines with no IVD degeneration (9° , $p = 0.002$, $d = 0.13$).

Interestingly, this difference between groups appears to be driven by decreased intervertebral segmental range of motion at L2–L3 (2.1° , $p = 0.031$, $d = 0.29$), L3–L4 (1.0° , $p = 0.015$, $d = 0.14$), and L4–L5 (1.9° , $p = 0.039$, $d = 0.19$).

Effect of Low Back Pain on Lumbar Spine Posture

There were no significant differences in lumbar spine posture, intervertebral angle, or intervertebral distance between Marines with and without back pain at the time of data collection, or with and without reported back pain in each position. However, there was a difference in angle with respect to the horizontal ($p < 0.001$, $\eta^2 = 0.27$) when the current episode of LBP was subclassified into chronic, acute, or no LBP. Marines with chronic LBP displayed more extension of the lumbosacral spine than those without LBP ($\sim 4^\circ$, $p = 0.024$, $d = 0.61$) and Marines with an acute episode of LBP ($\sim 3^\circ$, $p = 0.048$, $d = 0.08$) when standing without external load. Additionally, when sitting with body armor, Marines with chronic LBP displayed more extension of the lumbosacral spine than those with an acute episode of LBP (5° , $p = 0.021$, $d = 0.24$), but their posture was the same as Marines without LBP ($p = 0.105$, $d = 0.21$). When subjects were classified by history of LBP, a significant effect was found for angle with respect to the horizontal ($p < 0.001$, $\eta^2 = 0.12$). Participants who reported a history of LBP displayed less intervertebral extension at L5–S1 (6° , $p = 0.024$, $d = 0.78$) when prone on elbows, with corresponding less anterior distraction (4 mm, $p = 0.028$, $d = 0.50$) than those who did not have an episode of LBP.

DISCUSSION

In this study, we evaluated the lumbar spine posture in 43 active duty male Marines in simulated, relevant operational positions and loading conditions. Position-dependent changes in global lumbar spine postures were found in lumbosacral extension, sacral slope, and lumbar lordosis. Sitting with body armor resulted in a reduced lumbar lordosis of over 30°, but no changes in the forward inclination of the lumbar spine with respect to standing positions was observed. The addition of load did not change standing posture as was previously observed.⁹ This is likely related to the small load magnitude used in this study (11.3 kg) compared with the previous study (50.8 kg). This load magnitude was used because tasks involving sitting and being prone on elbows typically do not involve the heavy packs used in previous studies. Additionally, the load from the body armor was evenly distributed anterior-posterior, while the load distribution in previous studies was distributed with a posterior bias, which may have an effect on the response of the lumbar spine to loading.

To determine the location of global position-dependent differences in posture, we measured sagittal intervertebral angles and intervertebral heights. The largest decrease in lordosis relative to standing was observed between L2 and S1 when volunteers were sitting, with magnitudes of change increasing caudally, similar to previous studies.²³ Interestingly, when volunteers were prone on elbows, the only intervertebral segment with a measured increase in lordosis was found in L5–S1. Additionally, this level had the largest segmental range of motion between sitting and prone positions, and the largest number of discs that were classified as degenerated (33/43). Decreases in anterior intervertebral distance, driven by intervertebral flexion, were observed between L2 and S1, with anterior intervertebral distance progressively decreasing caudally. This follows previous observations that changes in intervertebral distance appear to track changes in sagittal intervertebral angle, not task or load.⁹

During this study, we did not specifically recruit Marines with LBP. However, we observed a relationship between history of LBP and posture in Marines. In standing and sitting positions, Marines

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with chronic LBP (current episode lasting >6 months) had more lumbosacral extension than Marines with no back pain or Marines with an acute (<6 months) case of LBP. O'Sullivan identified that LBP subjects with lumbar segmental instability can be classified into four groups, based on the manifestation of the patients' symptoms and motor dysfunction: flexion, extension, lateral shift, and multidirectional patterns.^{24,25} Of those subclassifications, the extension pattern has been reported to be due to repetitive low impact trauma associated with sporting activities, such as hiking with a heavy pack.²⁴ These data support the idea that postural changes in soldiers with chronic LBP may be due to heavy loads carried in training and combat.

In the present study, approximately half of the Marines reported LBP as a result of sitting for extended periods of time. It was previously reported that volunteers who sit for extended periods of time are at risk for developing LBP.²⁶ A 2007 review paper by Lis et al. determined that people with occupations requiring them to sit for over half of the workday, and who are exposed to factors such as whole body vibration and/or awkward postures, are at an increased risk of developing LBP.²⁶ The Marines in this study are often exposed to whole body vibration during transport in tactical vehicles and helicopters, and remain in awkward positions such as prone on elbows, for extended periods of time. Therefore, the high rate of LBP in this young population of Marines is likely attributable to their high occupational demands during both training and combat situations.

In this study, 58% of all IVDs were classified as degenerated, which is greater than among older patients imaged for LBP (45.5 years old, 44% degenerated).¹³ Marines with IVD degeneration experienced less whole lumbar range of motion between sitting and prone positions. It has been established that disc degeneration decreases local segmental range of motion using upright MRI.¹³ As a disc degenerates, functional spinal units become hypermobile (Pfarrmann Grades III and IV) and then hypomobile (Pfarrmann Grade V).¹⁷⁻¹⁹ Although we did not identify local changes in segmental range of motion, possibly because they were below our threshold of detection, we hypothesize that small changes at each level all contributed to the overall observed change in lumbar range of motion. Several

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studies have found trends toward differences in the posture of the lumbar spine with disc degeneration; however, it is unclear if disc degeneration leads to increased or decreased lordosis.^{12,13,27,28} Disc degeneration occurs in the normal progression of aging, but it can be accelerated by lifestyle factors such as excessive loading or smoking.¹¹ People with physically intensive jobs are at risk for accelerated IVD degeneration. Marines in our study had higher levels of habitual physical activity, measured by the Baecke questionnaire, than civilians of similar age^{14,29} because they routinely engage in physical activity and training per their infantry program of instruction.

There are several limitations to this study. First, imaging resolutions were 1.56 mm in plane and 3.0 mm thick. We know from prior validation work²⁰ that this yields low angular error rates (~1 deg), but it does mean that our confidence in linear changes in disc height is relatively coarse. Second, our secondary analyses included a number of univariate comparisons aimed at identifying kinematic differences between Marines with and without LBP or with and without IVD degeneration. Because of the structure of these variables and our statistical approach, type I error rates could have been elevated. However, the subtle effects of LBP and IVD degeneration on whole spine kinematics are consistent with prior observations in civilian populations.

This study was performed on active, young Marines, and their postural responses to loading may not apply to a civilian population. Due to the intense training and demands of their jobs, the Marines in this study were extremely fit, and trained on how to adapt their posture in different positions while wearing body armor to minimize their risk of injury. One of the long-term goals of this line of research is to identify Marines who were at risk for developing LBP based on their posture profile. We observed that Marines with chronic LBP exhibit an extension pattern of LBP, which may be related to the loads they carry during training and combat. Marines carry required equipment to ensure they can successfully complete their job, a load that sometimes exceeds the standard 32 kg. Since we have previously investigated the effect of heavy load on lumbar spine posture, and have identified postural differences in Marines with pathology in this study, further studies on LBP and disc degeneration in

Marines should focus on managing the anterior-posterior distribution of load around the torso center of mass. We have previously noted that balanced loads minimize postural changes in standing³⁰, but their influence on non-standing postures (e.g. sitting) have not been investigated. A second long-term goal of this research is to understand position-related lumbar spine disease and to understand the influence of loading and position on existing lumbar spine pathology. For example, future work should test the influence of relevant positions and loads on stable and unstable disc protrusions/fragmentations in the civilian population. These studies may shed light onto more appropriate, or patient specific, exercise regimes for individuals with anatomically identifiable sources of low back pain.

CONCLUSION

This study investigated the effect of operationally relevant positions on lumbar spine posture in active duty male Marines. Sitting and prone on elbows positions resulted in large global postural differences in the lumbar spine, with the greatest local change found at L5–S1. This level also had the highest incidence of IVD degeneration in the lumbar spine. Overall, we found that Marines have a high incidence of lumbar disc degeneration and high rates of LBP, particularly in the sitting position. Marines with disc degeneration at L5–S1 had decreased lumbar range of motion and Marines with chronic LBP had more lumbar extension. Subtle postural changes were seen in subjects with disc degeneration or LBP, resulting in increased overall lumbar lordosis and decreased lumbar range of motion. These results provide the framework for postural interventions aimed at reducing LBP, and perhaps, risk factors for LBP in Marines during operational tasks.

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Human subjects participated in this study after giving their free and informed consent. This research has been conducted in compliance with all applicable federal regulations governing the protection of human subjects in research (University of California, San Diego protocol 110483, and NHRC.2013.0023).

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Table 1. Participant characteristics by intervertebral disc degeneration, and current and past low back pain. No significant differences were found between groups

	Aggregate	Degeneration		Low Back Pain			
		Yes = 33	No = 10	“Are you currently experiencing LBP?”		“Have you ever experienced LBP?”	
				Yes = 16	No = 23	Yes = 28	No = 11
Age (years)	26.8 ± 6.4	27.6 ± 6.8	24.0 ± 3.7	26.9 ± 5.7	27.1 ± 7.4	27.7 ± 6.6	25.4 ± 6.6
Height (m)	1.78 ± 0.07	1.78 ± 0.07	1.76 ± 0.07	1.78 ± 0.07	1.78 ± 0.08	1.79 ± 0.07	1.77 ± 0.09
Weight (kg)	81.9 ± 9.8	83.8 ± 9.5	75.6 ± 8.6	81.8 ± 10.6	82.3 ± 10.0	82.5 ± 9.9	81.0 ± 11.2
BMI (kg/m ²)	25.9 ± 2.9	26.4 ± 2.8	24.3 ± 2.4	25.7 ± 3.1	26.1 ± 2.9	25.9 ± 2.8	26.0 ± 3.5

BMI, body mass index.

Figure Legends

Figure 1. T2-weighted midsagittal magnetic resonance images (MRIs) of the lumbar spine (top) and photographs of Marines in the MRI scanner (bottom). Volunteers were scanned standing unloaded (A, E), standing with body armor (B, F), sitting with body armor (C, G), and prone on elbows with body armor (D, H).

Figure 2. Distribution of Pfirrmann-graded intervertebral discs by level.

Figure 3. Schematic depicting lumbar spine postural measurements on a 3D model of the lumbar spine. Measurements included A.) Angle with respect to the horizontal to assess lumbar flexion/extension. B.) Sagittal Cobb angle to measure lumbar lordosis. C.) Sacral slope to assess rotation of the pelvis. D.) Intervertebral angles to provide a measurement of local lumbar lordosis.

Figure 4. Global measures of lumbar spine posture. Angle with respect to horizontal (top), sacral slope (middle), Cobb angle (bottom). Statistically significant difference between measurements ($p > 0.05$) indicated by line. Data reported as mean (standard deviation).

Figure 5. Local measures of lumbar spine posture. Intervertebral angles from L1–L2 (top) to L5–S1 (bottom) are shown. Statistically significant difference between measurements ($p > 0.05$) indicated by line. Data reported as mean (standard deviation).

Figure 6. Anterior (left column), central (middle column), and posterior (right column) intervertebral distance measurements for each level of the lumbar spine. Intervertebral heights from L1–L2 (top) to L5–S1 (bottom) are shown. Statistically significant difference between measurements ($p > 0.05$) indicated by line. Data reported as mean (standard deviation).

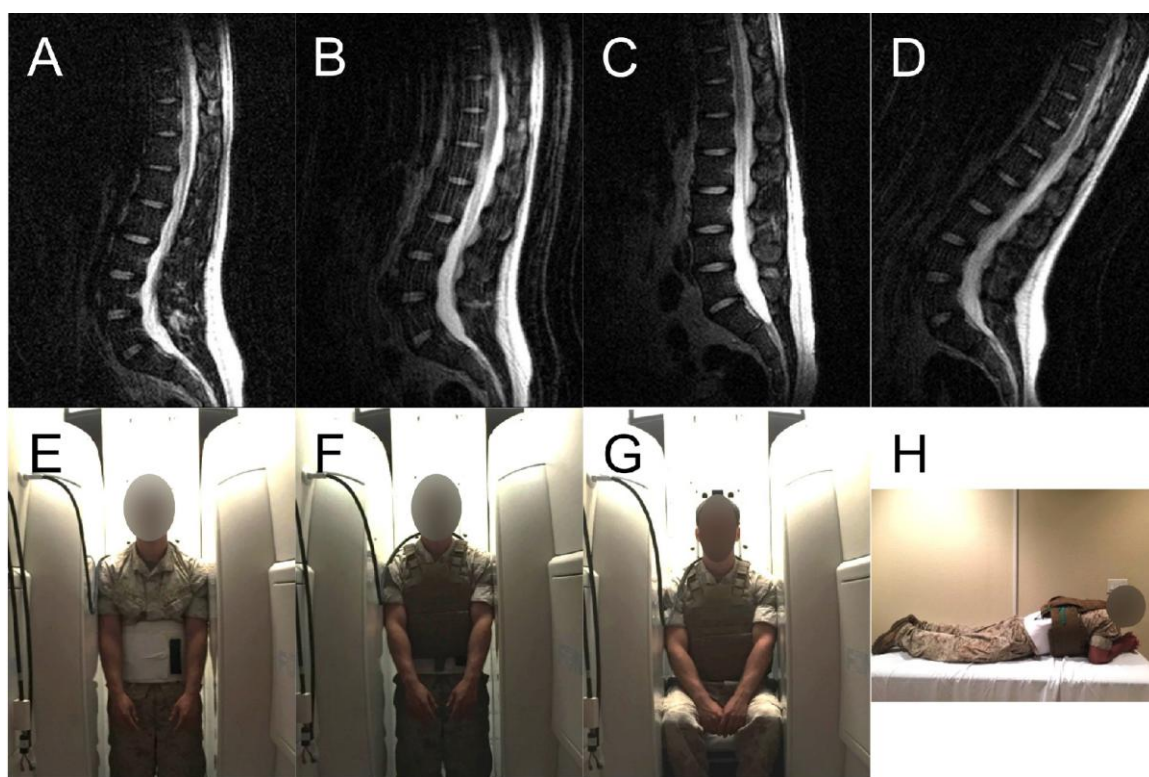


Figure 1

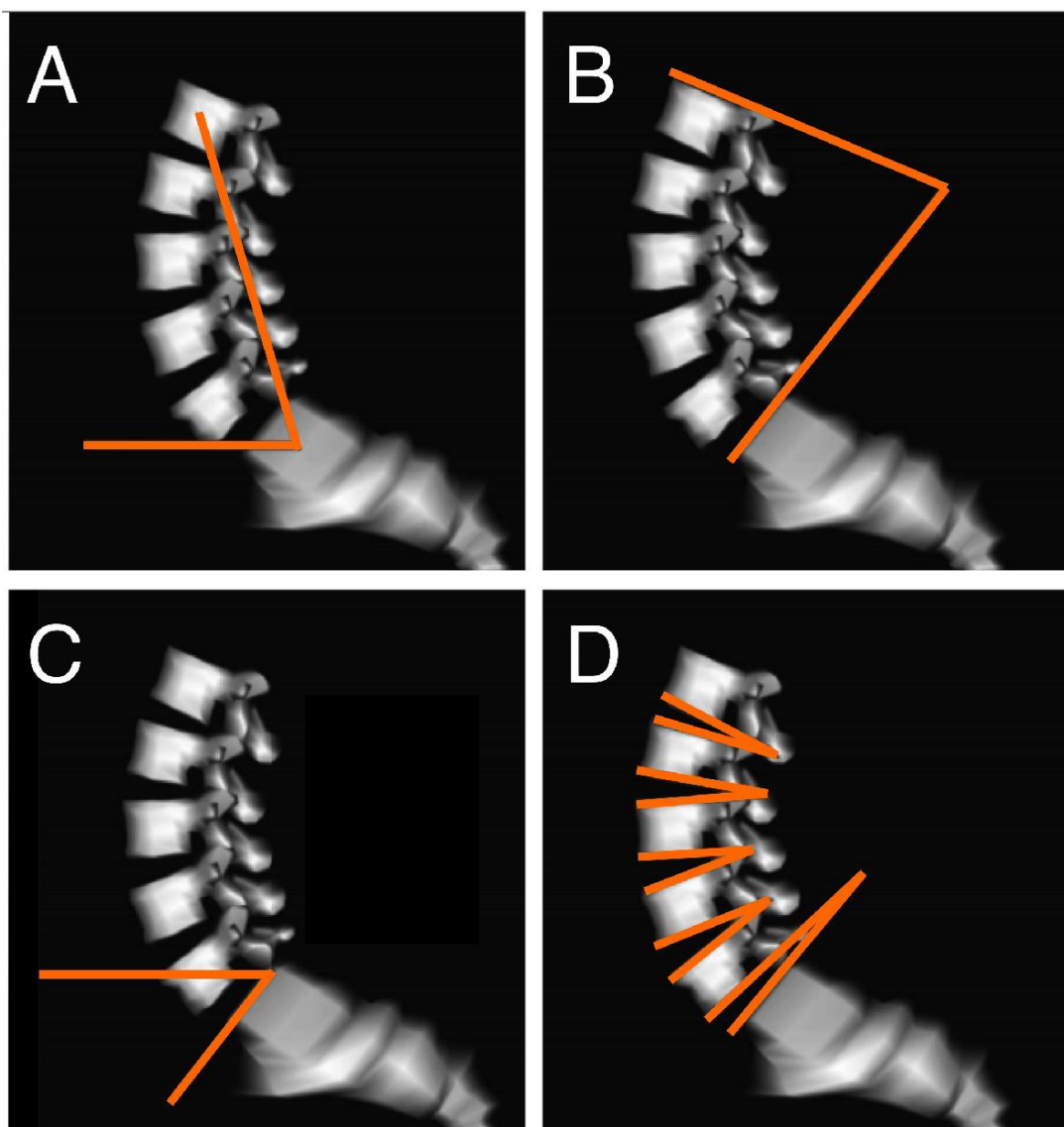


Figure 2

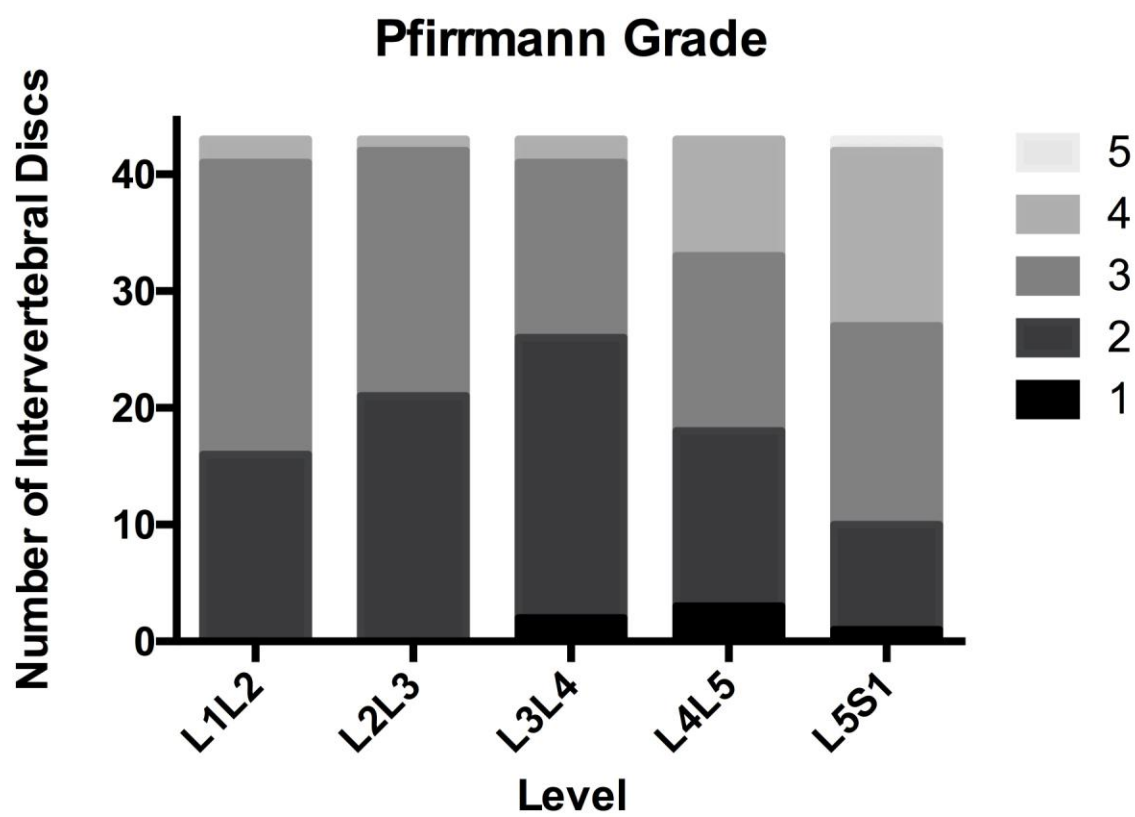


Figure 3

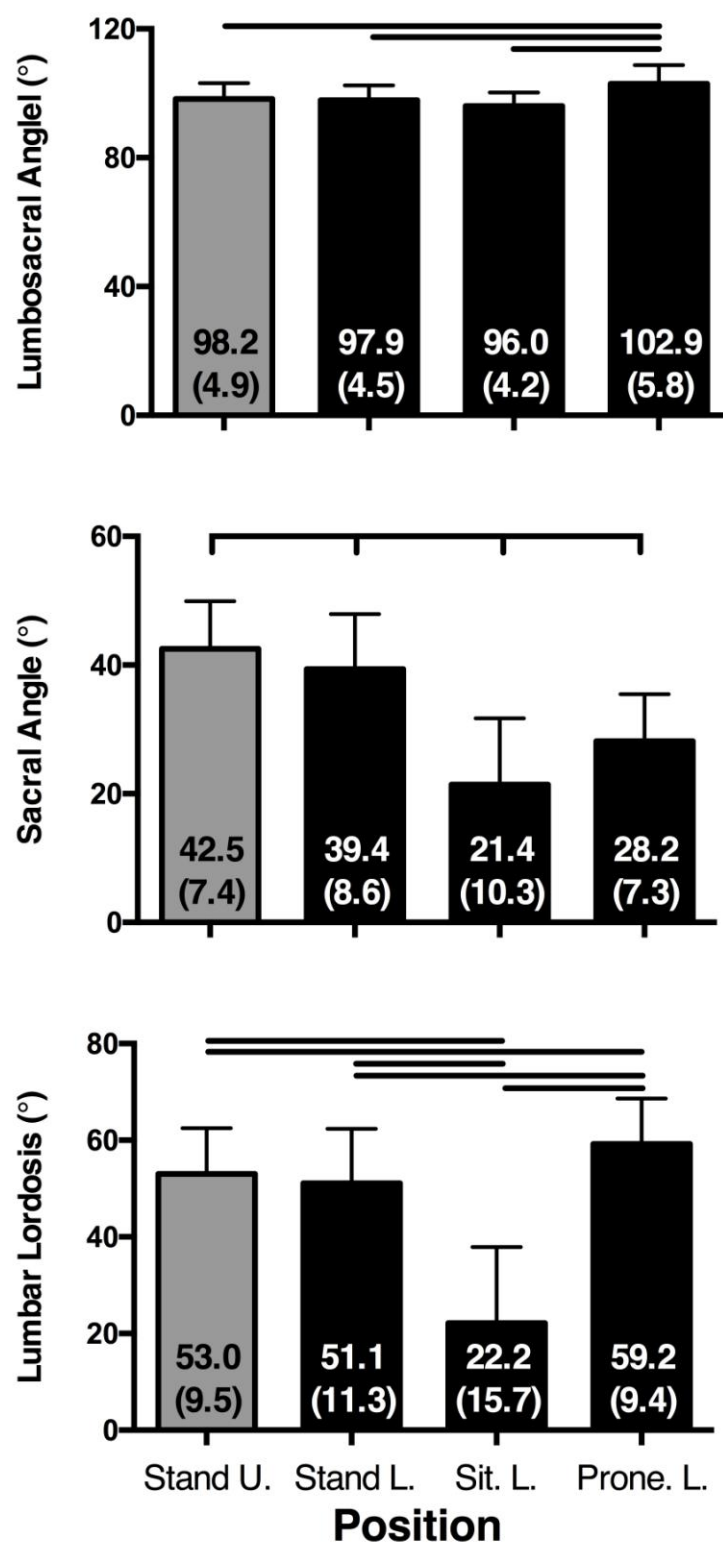


Figure 4

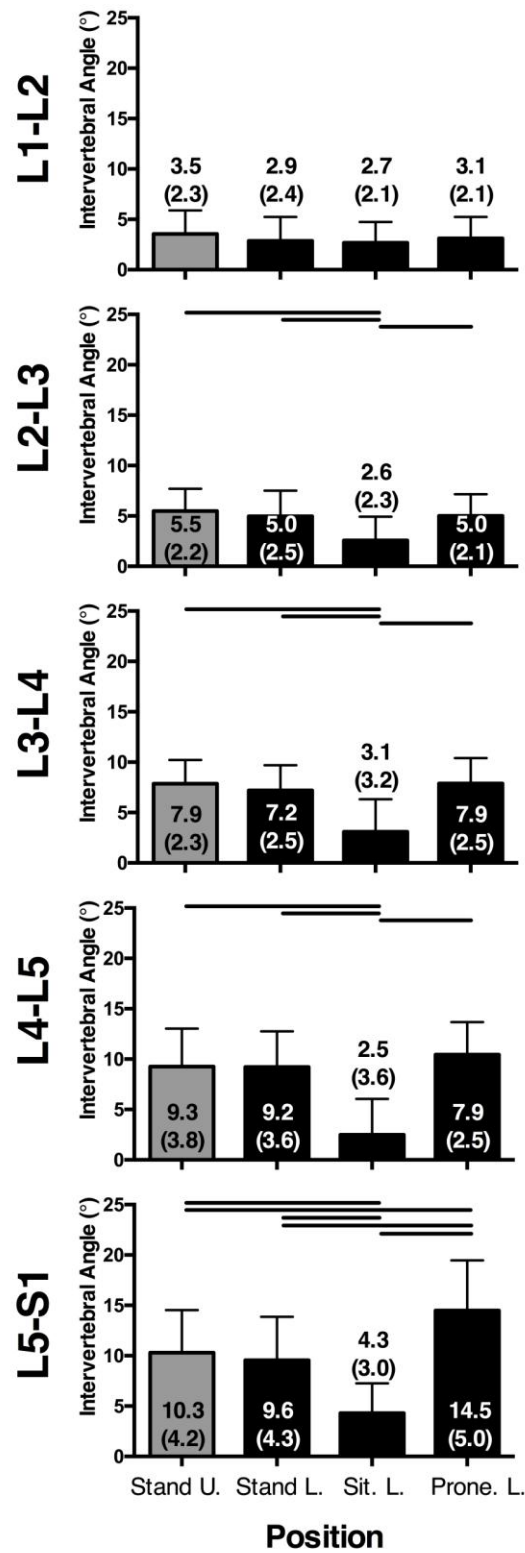


Figure 5

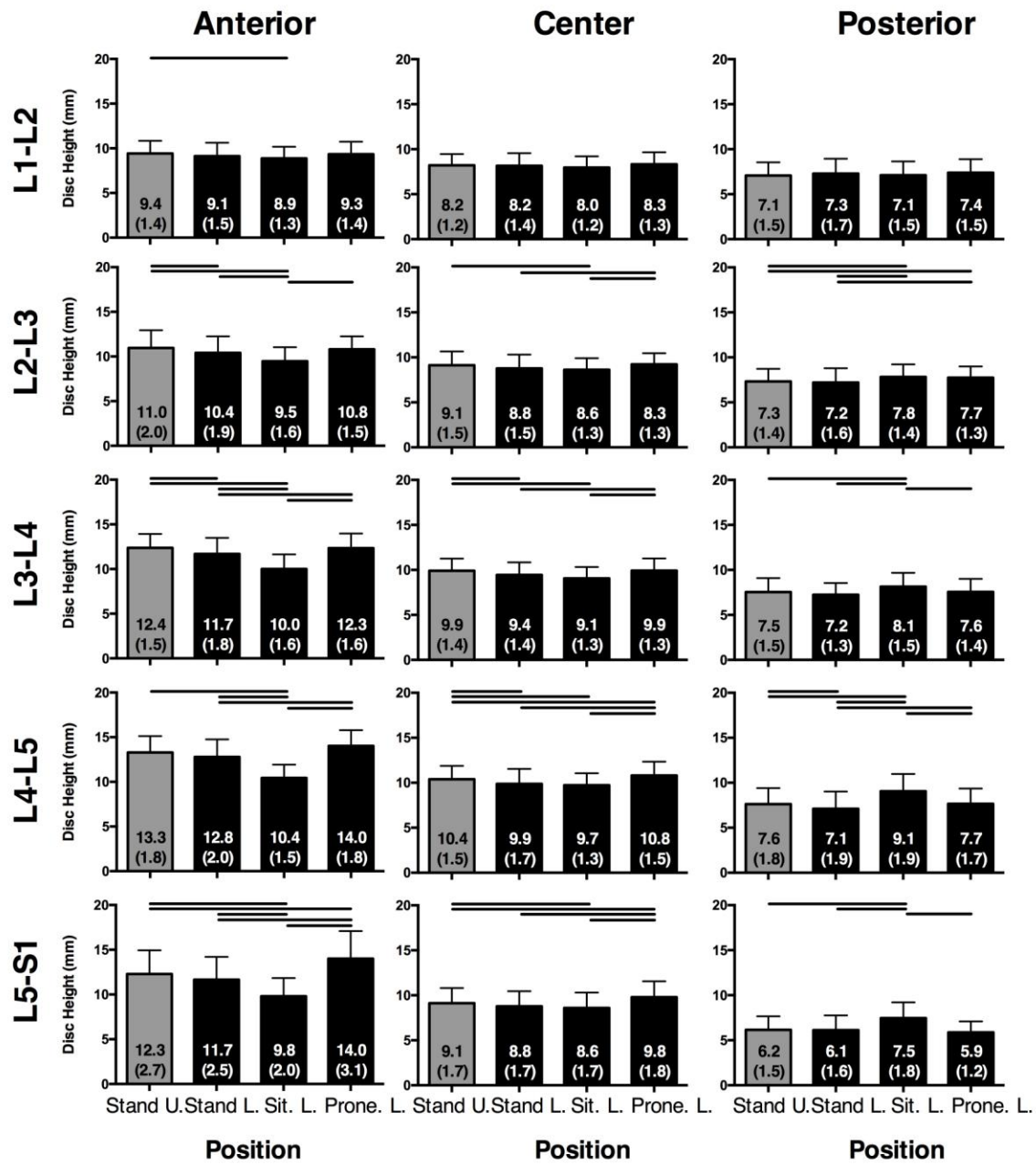


Figure 6

Lumbar Muscle Structure Predicts Operational Postures in Active Duty Marines

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Abstract

STUDY DESIGN: Cross-sectional.

BACKGROUND: The relationship between lumbar spine posture and muscle structure is not well understood.

OBJECTIVES: Investigate the predictive capacity of muscle structure on lumbar spine posture in active duty Marines.

METHODS: Forty-three Marines were scanned using an upright magnetic resonance imaging (MRI) scanner while standing unloaded and standing, sitting, and prone on elbows with body armor. Cobb, horizontal, and sacral angles were measured. Marines were then scanned unloaded in supine using a supine MRI scanner. The imaging protocol consisted of T2 intervertebral disc mapping, high-resolution anatomical, fat-water separation, and diffusion tensor imaging to quantify disc hydration and muscle volume, fat fraction, and restricted diffusion profiles in the lumbar muscles. A stepwise multiple linear regression model was used to identify physiological measures predictive of lumbar spine posture.

RESULTS: The multiple regression model demonstrated that fractional anisotropy (FA) of the erector spinae was a significant predictor of lumbar posture for 7 of 18 dependent variables measured and explained 20% to 35% of the variance in each model. Decreased FA of the erector spinae predicted decreased lordosis, lumbosacral extension, and anterior pelvic tilt.

CONCLUSION: FA is inversely related to muscle fiber size, which is related to isometric force generating capacity of a muscle fiber. This suggests stronger erector spinae muscles predict decreased lordosis, lumbosacral extension, and anterior pelvic tilt in a highly-trained population.

KEY WORDS: *lumbar spine, skeletal muscle, diffusion tensor imaging, magnetic resonance imaging, military, posture*

INTRODUCTION

The muscles of the lumbar spine are crucial for stabilizing and supporting the upper trunk, especially during dynamic loading conditions. A muscles force generating capacity is directly related to its architectural and microstructural features, which are therefore variables of interest when trying to assess muscle health. Physiological cross-sectional area (PCSA) is a measure of muscle architecture that can be measured to estimate muscle force.²⁹ However, it is difficult to precisely measure PCSA *in vivo* as it takes into account measures of muscle architecture, such as pennation angle and normalized fiber length. Volume is a dominant input variable to measure muscle PCSA, and is commonly used as a proxy for muscle force producing capacity.^{8, 10} Although, muscle is a heterogeneous tissue, also consisting of fat and collagenous tissues, which can confound measures of muscle volume. Skeletal muscle exhibits a classic structure function relationship, where its microstructural properties are closely related to whole muscle function. For example, muscle fiber isometric force generating capacity is directly related to fiber cross sectional area.^{19, 23, 24} It is also difficult to measure muscle microstructure *in vivo*, although there is some evidence that diffusion based imaging techniques are sensitive to different features of muscle microstructure, in particular fiber area.^{5, 9, 13, 37} With injury and age, atrophy of muscle fibers and replacement of muscle tissue with adipose and fibrotic tissue is typically observed compared to healthy muscle, further decreasing the overall volume of functional contractile tissue in the muscle.^{8, 38} As pathogenic (diseased) muscle becomes atrophied, fibrotic, and contains more adipose tissue, the active and passive force generating potential of the whole muscle changes, which can have a direct and negative effect on joint stability, range of motion and posture.^{22, 28, 41, 50} The multifidus muscle is considered one of the primary muscular stabilizers of the lumbar spine, due to its ability to produce high forces over a narrow range of

lengths and often undergoes the aforementioned pathogenic changes associated with injury, low back pain, or age.⁴⁸

Changes to the orientation and position of bony structures of the spinal column are often observed simultaneously with the aforementioned changes in muscle composition.^{21, 40, 42} With age, gross changes in spinal posture, such as decreased lumbar lordosis, increased lumbar flexion, and increased pelvic tilt, are typically observed.^{15, 16, 20, 44} Decreased segmental range of motion has also been measured at vertebral levels with intervertebral disc (IVD) degeneration,^{15, 18} which is defined as decreased hydration of the nucleus pulposus with accompanying disc height loss.²⁷ However, changes in muscle structure, lumbar posture, and IVD health are not independent of one another, and their effects are confounded by age, sex, activity level, and the timing of disease progression.

In addition to associated changes with age and disease, posture is affected by external stimuli such as carrying load or whole body position.^{4, 31-33} Military members are a highly active population and are often required to carry heavy loads in unusual positions. Studies investigating how Marines adapt to load carriage suggest they routinely operate under conditions that put them at risk for developing lumbar musculoskeletal injury and exhibit higher rates of low back pain than civilians.^{35, 36} This may be attributed to pathophysiologic changes of the lumbar spine structures, as a result of the heavy loads and unusual postures experienced in training and combat.^{17, 30} A noninvasive tool that is able to correlate musculoskeletal health to posture under relevant loading conditions would allow clinicians to tailor rehabilitation protocols to target specific musculoskeletal components involved in regulating posture to mitigate an individuals risk of lumbar spine injury. Therefore, the purpose of this study is to investigate the predictive capacity of muscle structure, IVD health, and anthropometric measures on lumbar spine posture

in active duty Marines. We hypothesize that multifidus muscle volume can predict lumbar posture in different positions, since multifidus provides intersegmental lumbar support, and muscle volume is related to muscle strength.

METHODS

The UC San Diego and Naval Hhealth Research Center Institutional Review Boards approved this study, and all volunteers gave verbal and written consent to participate. Marines were included in this study if they were male, over 18, and healthy enough to perform their assigned duty. Marines were excluded from this study if they have undergone lumbar spine surgery, or had the possibility of shrapnel in their body. Marines were not recruited based on low back pain status or history. All Marines underwent standard MRI safety screening prior to scanning. All scans were performed early in the morning, between 4am and 9am.

Upright MRI-Imaging

Marines were scanned using an upright 0.6T MRI scanner (Upright Multi-Position MRI, Fonar Corporation, Melville, NY) and a planar coil. An elastic band was used to hold the coil against the volunteer's lumbar spine between the L1–S1 levels while standing. The band was secured to hold the coil in place, while not altering the volunteer's natural position. A three-plane localizer (TR = 1254 ms, TE = 100 ms, FOV = 34 cm, matrix = 256×256 , in-plane resolution = 1.33mm x 1.33mm, THK = 9 mm, NEX = 1, time = 0:17) and sagittal T2-weighted images (TR = 1974 ms, TE = 160 ms, FOV = 35 cm, matrix = 224×224 , in-plane resolution = 1.56mm x 1.56mm, THK = 3 mm, gap = 0 mm, NEX = 1, time = 2:12) were acquired.³

Upright MRI-Load Carriage and Position Tasks

Marines were scanned in the following positions: standing without load, standing with body armor (11.3 kg), sitting with body armor, and prone on elbows with body armor. Positions with

external load were randomized to control for the cumulative effects of loading or time. The positions chosen are common static positions Marines are often required to maintain for extended periods of time, depending on military occupational specialty, and are often self-reported as provoking LBP.⁴ Load magnitude (11.3 kg) was chosen based on use of body armor, which is minimum protective equipment Marines are required to wear at all times during military operations and training. Marines were not provided instruction on how to assume each position but were asked to hold each position steady for the duration of the MRI acquisition. A previous study has shown no statistically significant difference in test-retest variation in posture within a subject, even after performing heavy load and activity tasks.³³

Upright MRI-Postural Measurements

Postural measurements were generated from upright MRI images in each position, using a previously validated algorithm.⁴ Briefly, digital seed points were manually placed on the corners of the vertebral body and on the posterior elements of each vertebra using OsiriX version 3.9.3 imaging software (Pixmeo SARL, Bernex, Switzerland).³⁴ The location of the seed points were imported into MATLAB (The MathWorks, Inc, Natick, MA) and used to define an endplate-based joint coordinate system applied to the superior and inferior endplate of each vertebra (L1–S1).

Global measurements of lumbar spine posture were calculated for each position to characterize the posture of the lumbar spine. Global measures included: angle with respect to the horizontal to assess lumbosacral flexion/extension, sacral slope to assess sacral tilt, and sagittal Cobb angle to assess lumbar lordosis (FIGURE 1). Root mean square error for global measurements have been measured previously and are 0.28°, 0.95° and 0.95° respectively.^{3,33} In addition to static positions, global measurements were calculated between the standing unloaded

to standing loaded (Δ load) positions, and sitting loaded to prone loaded (Δ position) positions to determine lumbar kinematics in response to load and dynamic movement, respectively.

Supine MRI-Imaging

MRIs of the lumbar spine (L1–S1) were acquired using a 3T MRI scanner (GE MR750 Discovery, GE Healthcare, Madison, WI) and spine array coil. The imaging protocol consisted of: 1) an anatomical, 2) fat-water separation scan, 3) diffusion tensor imaging (DTI) of the lumbar spine, and 4) T2 mapping of each lumbar IVD. Marines were scanned supine with the lumbar muscles relaxed in order to mitigate motion and breathing artifacts. The anatomical scan was an axial fast spoiled gradient echo (FSPGR) with the following scanning parameters: repetition time (TR) 5msec, echo time (TE) 2.3msec, flip angle 20°, field of view (FOV) 32cm, 512×512 acquisition matrix, 0.625×0.625mm² pixel size, 1mm slice thickness, no gap, and three averages. Fat and water separation images were acquired utilizing a three-point iterative decomposition of water and fat with echo asymmetry and least-squared estimation (IDEAL) sequence in the sagittal plane (TR 1974msec, TE 160msec, flip angle 20°, FOV 25.6cm, 176 slices, 256×256 acquisition matrix, 1×1×1mm³ voxel size, no gap, and number of averages 1). Scanning parameters of the axial DTI sequence were: TR 10sec, TE 46msec, FOV 19.2cm, 82 slices, 128×128 acquisition matrix, 1.5×1.5mm² pixel size, 3mm slice thickness, no gap, b-value 400mm²/s, and 45 diffusion directions. Lastly, multi-spin echo data (8 echoes, TEs 8.6 to 68.8msec, TR 800msec, FOV 16 cm, 5 slices, 256×256 acquisition matrix, 0.625×0.625×5mm³ voxel size, no gap, and number of averages 1) was acquired and used to estimate the T2 of each lumbar IVD. The scanning plane was axial oblique, parallel to each lumbar IVD.

Supine MRI-Lumbar Physiology Measurements

Anatomical images were imported into the OsiriX imaging software for segmentation. Contours of the multifidus, erector spinae group, psoas, and quadratus lumborum muscles were manually traced from L1 to S1 lumbar levels. The resulting segmentations were used to generate masks to quantify muscle volumes, fat fraction, and diffusion properties of Marines in the supine position.

Images acquired using the fat-water separation sequence yield 2 sets of images: 1 where both fat and water MRI signals are in phase and 1 where they are out of phase. This allows for isolating the independent contributions of water (S_W) and fat (S_F) to the total MR signal. These data were then used to quantify fat fraction (FF) of the multifidus and erector spinae group with the following relationship:

$$FF = \frac{S_F}{S_W + S_F}$$

The diffusion tensor was fitted using Analysis of Functional NeuroImages software (National Institutes of Health, Bethesda, MD) function 3dDWItoDT.⁷ Mean diffusivity (MD), fractional anisotropy (FA), and the 3 eigenvalues (λ_{1-3}) of the diffusion tensor are reported. The quantitative relationship of diffusion variables to specific features of muscle microstructure is the focus of current work, although there is some evidence that they are related to muscle fiber size.^{5, 9, 13, 37} MD describes the average restricted diffusion coefficient of λ_{1-3} and is normally between $1 \cdot 10^{-3} \text{ mm}^2/\text{s}$ - $2 \cdot 10^{-3} \text{ mm}^2/\text{s}$.²⁶ FA is a unitless measurement from 0 to 1 that indicates the shape of the diffusion tensor. An FA value of 0 corresponds to isotropic diffusion (unrestricted), and an FA value of 1 corresponds to diffusion along a line (highly restricted). λ_{1-3} define the magnitude of diffusion along (λ_1) and radial to ($\lambda_{2,3}$) the main direction of the muscle fiber.

The T2 values for each IVD were estimated by fitting the magnitude of the multi-echo data to a monoexponential decay:

$$S_i = S_o e^{-t/T_2}$$

IVD health is often assessed by qualitatively assessing disc hydration from T2 weighted MRIs. Quantitative T2 mapping provides a quantitative measurement of IVD hydration; T2 is inversely proportional to Pfirrmann grade, which is a common ordinal scale to assess intervertebral disc degeneration.⁴³

Statistical Analysis

Dependent variables were global postural measurements (angle with respect to the horizontal, sagittal Cobb, and sacral angles) for all positions (standing unloaded and standing, sitting, and prone on elbows with load) and the change in load and flexion-extension positions (Δ load, Δ position). In order to assess variance, coefficient of variation was calculated for each dependent and independent variable.

An *a priori* approach was used to minimize the number of independent variables input into each model (FIGURE 2). First, independent variables were empirically grouped into 3 separate domains: muscle structure (volume, FF, FA, MD, and λ_{1-3}); IVD health (T2-relaxation of each disc); and anthropometric (age, weight, height, and body mass index⁴⁵) measures. Hierarchical cluster analysis was used to verify domain groupings. Within each domain grouping, an additional hierarchical analysis was performed. Variables that did not cluster were entered into a stepwise multiple linear regression model for each dependent variable to identify physiologic measures predictive of lumbar spine posture. Variables that did cluster were then sorted into like variables (eigenvectors) using principal components analysis (PCA). Within each eigenvector, Pearson's correlation was used to remove collinear variables ($r > 0.80$). For collinear variables, the variable with the smallest eigenvector value was removed in order to avoid redundancy of variance across variables. Collinearity was also verified at this point by variance inflation factor. If a variable had a variance inflation factor > 10 it was removed from the model.

Remaining variables were then entered into the stepwise multiple linear regression model for each dependent variable. A stepwise multiple linear regression was run for each individual dependent variable (18 models: 6 positions x 3 postural measurements). Statistical analyses were performed using SPSS Version 20.0 (IBM Corporation, Armonk, NY).

RESULTS

Volunteer Demographics

Forty-three male Marines (mean age = 26.8 ± 6.4 y; height = 1.8 ± 0.1 m; weight = 82.0 ± 9.9 kg) volunteered for this study. Two subjects dropped out during supine imaging due to claustrophobia in the MRI scanner. Additionally, DTI datasets of 10 subjects were deemed unusable due to breathing or motion artifact. Therefore, 31 Marines were included in this analysis (mean age = 27.3 ± 6.9 y; height = 1.8 ± 0.1 m; weight = 80.6 ± 8.7 kg). Marines excluded from the study had no differences in anthropometric measures compared with those included. Of these volunteers, 10 Marines self-reported experiencing LBP at the time of the scan. Coefficient of variation was relatively low for dependent and independent variables (SUPPLEMENTARY TABLES 1 and 2; range: 0.04-10.61; median: 0.16). On average, the greatest variation was found for the IVD health measures.

Regression Model

After initial grouping of independent variables, collinearity resulted in the removal of 8 of the 29 independent variables from the model (FIGURE 2). Collinear variables that were removed included diffusion measurements from either multifidus or erector spinae, erector spinae FF, and BMI. Surprisingly, 9 of 18 dependent variables were found to have a significant predictor from the stepwise multiple linear regressions. In fact, FA of the erector spinae was a significant predictor of lumbar posture for 7 of the 18 dependent variables measured and explained 20% to

35% of the variance for each outcome (TABLE). In general, increased FA in erector spinae was predictive of increased lumbar lordosis, lumbosacral flexion, and pelvic tilt in each position. Additionally, decreased T2 relaxation of the L4–L5 IVD was a significant predictor of increased lumbosacral extension when standing unloaded ($p = .025$, $R^2 = 0.192$). When prone on elbows, increasing subject weight was a significant predictor of increased lumbar lordosis ($p = .016$, $R^2 = 0.219$). No muscle volume, muscle microstructure, IVD health, or anthropometric measures were significant predictors of posture when subjects were sitting loaded.

DISCUSSION

In this study, we evaluated the relationship between lumbar spine posture and muscle structure, IVD health, and anthropometric measures in 31 active duty male Marines in simulated, relevant, operational positions and loading conditions. FA of the erector spinae was a significant predictor in 7 of the 18 measures of lumbar spine posture across several different positions. For the standing loaded condition, FA of the erector spinae was a significant predictor of all 3 measures of lumbar posture; Marines with increased FA of the erector spinae had a more lordotic, extended lumbar posture with greater sacral tilt. Muscle volume was not a significant predictor of any postural measurements, despite being a commonly used proxy for muscle strength.^{11,21} Together, the ability of FA to predict postural behavior in several positions, along with the absence of association between muscle volume and lumbar spine posture, suggests muscle microstructure, but not quantity, - which are both measures associated with force generating capacity of muscle - is an important predictor of lumbar spine posture.

DTI is an MRI technique that measures the restricted diffusion of water in tissues with anisotropic microstructure.² As the sarcolemma is considered to be the primary barrier to diffusion, DTI is believed to be most sensitive to changes in fiber size because radial diffusion of

water across a muscle fiber is more restricted (by the sarcolemma) than longitudinal diffusion within a muscle fiber.^{46, 47} While it has been shown that FA and fiber area are inversely related,^{5, 6, 9, 13, 37} it is important to note the exact relationship has not been validated. However, it is well established that muscle fiber area and isometric force are directly related.^{19, 23, 24}. Therefore, it appears there is likely an inverse relationship between FA and isometric force generating capacity of muscle. As such, it is inferred when FA increases, the force generating capacity of a muscle decreases (i.e., the muscle is weaker). For example, if the multifidus muscle in two Marines were imaged using DTI and one had a larger FA (smaller fiber size), that muscle would be expected to generate less overall force.

Two unique relationships between posture and muscle structure were found in this study: 1) erector spinae and not multifidus, and 2) muscle microstructure, not volume, were found to be significant predictors of lumbar posture. First, FA of the multifidus and the erector spinae were found to be collinear, with FA of the erector spinae as a stronger descriptor of the eigenvector from the PCA. Therefore, multifidus was not entered into the final statistical model. To verify FA of the multifidus was not removed from the model because it had less variability than FA of the erector spinae, coefficient of variation was calculated for both variables. FA of the erector spinae had less variability relative to the mean than FA of the multifidus (0.07 vs 0.08), further supporting the latter as a stronger descriptor of the eigenvector. While there is a small difference in variability of these measures, they are greater than the associated measurement error (0.03 and 0.04 respectively). This finding suggests that while the multifidus stabilizes the individual segments of the spinal column,^{48, 49} the erector spinae may play a role in determining gross lumbar posture. Second, while muscle volume is proportional to muscle strength,^{19, 29} muscle microstructure has been shown to be a more accurate predictor of muscle force generating

capacity. Clinically, the findings from this study are important because they suggest microstructural quality of the lumbar muscles is more important to whole lumbar posture in functionally loaded positions than the quantity or volume of muscle. This is not surprising given that measures of whole muscle size and volume are confounded by non-contractile tissue, such as fat and fibrosis. Importantly, FA may be a noninvasive composite measure of the functional contractile tissue present in a whole muscle, which seems to explain much of the variance much postural responses to body position.

In this study, T2 of the L4–L5 IVD was found to be inversely proportional to lumbosacral extension when Marines were standing without load. This suggests that Marines with decreased IVD T2 values (increased IVD degeneration) at L4–L5 have increased lumbosacral extension. Previously, using the Pfirrmann grading scale, we reported no significant difference in lumbosacral extension in Marines when categorized by degeneration at L5–S1 (Pfirrmann grade >2).³ As L5–S1 is the base of support of the lumbar spine, it was assumed that degeneration at this level would have whole lumbar postural consequences. However, our findings demonstrate L4–L5 IVD health is related to whole lumbar posture and, therefore, should be considered an important structural level for whole lumbar stability. The finding that single-level disc health has the potential to influence lumbosacral flexion reiterates the importance of the lower lumbar spine as a transition zone of load between the trunk and body. Changes to the health of this region have the potential to affect support of the torso.

Several studies have previously attempted to determine the relationship between lumbar lordosis and BMI. It appears increased lumbar lordosis might be found in subjects with increased BMI,^{12, 25} however, other studies have shown no difference.⁵¹ In this study, weight and BMI were found to be collinear, with weight being the stronger predictor of the eigenvector from

PCA, therefore BMI was dropped from the final statistical model. However, this is likely due to a larger variance in subject weight rather than BMI in this relatively homogeneous population. If a more representative cross section of the population was used, these findings may be different.

In this study, we made several attempts to decrease the complexity of this model in order to decrease the amount of type I error, which can be associated from making multiple comparisons. First, we chose not to include individual vertebral level measures of muscle structure or lumbar posture. Second, we removed collinear variables with clustering and PCA to minimize the number of independent variables representing similar constructs that were entered into the model. Third, we evaluated forward, backward, and stepwise multiple linear regression models to determine which model was the most conservative approach. Results were the same with forward and stepwise elimination techniques, and backward elimination allowed for several more independent variables to be retained in the model, suggesting it was the least conservative regression approach. Therefore, we chose to use a stepwise multiple linear regression technique as it appeared to be the most conservative model.

The Marines in this study were not recruited based on history or presence of LBP at the time of the study, and approximately 1/3 of the Marines who were included in this study reported LBP. It is important to note that no Marines had an episode of low back pain so severe that they were relieved of duty. In a previous study, no difference in lumbar spine posture was found between Marines with and without low back pain at the time of data collection.⁴ No differences have been observed between Marines with and without LBP at the time of data collection for muscle physiology, IVD health, or anthropometric measures (data not published). As LBP did not result in differences in the dependent or independent variables measured, it is unlikely that the inclusion of Marines with and without LBP affected the findings of this study.

There are several limitations to this study. First, the Marines in this study had relatively, normal muscle, with no underlying pathology observable. We would expect that in patients with pathology or age related atrophic changes in muscle, the volume or fat fraction of muscle may become more important in predicting lumbar posture. Therefore, when considering the results of this study, it is important to frame that these findings may only extend to a highly-active population. Second, the positions that were measured in this study place relatively small challenges on the muscles of the lumbar spine. A future direction of this study is to investigate whether muscle microstructure can predict posture under heavy loading conditions that Marines routinely operate under. Last, we had 21 variables included in our model, with only 31 full datasets to include. This was a retrospective analysis of two studies investigating 1.) the effect of operationally relevant positions on lumbar posture⁴ and 2.) normative paraspinal muscle composition in active-duty Marines; 43 subjects was determined to provide adequate power to these studies. In order to mitigate type I error associated with multiple comparisons, we used the most conservative statistic approach. While more subjects may provide an increase in the amount of variance explained by the model, we still reached significance with 31 complete data sets.

CONCLUSION

To our knowledge, this study is the first to measure the predictive capacity of lumbar muscle structure, IVD health, and anthropometric measures on lumbar spine posture in different positions. It is surprising that any structural variable in muscle predicted any of the variance in posture, since many clinicians believe that short-term postural positions are more related to motor control then strength or end-organ dependent behavior. It was found that FA of the erector spinae was a significant predictor of several lumbar postural measures. In general, decreased FA of the erector spinae resulted in decreased lordosis, lumbosacral extension, and anterior pelvic

tilt. This posture results in decreased shear stress at lower lumbar levels during hyperlordosis, and may be considered a more protective posture for preventing injury and LBP while load bearing in the lumbar spine.³⁹ Decreased FA of the erector spinae can be physiologically interpreted as larger muscle fibers with more capacity to generate force. Due to the intense training and demands of their jobs, the Marines in this study were extremely active and trained on how to adapt their posture in different positions, while wearing body armor to minimize their risk of injury. Therefore, these findings may not translate to a civilian population.

The findings from this study lend credence to the idea that muscle strengthening/exercise may influence posture, although, this cause-effect relationship needs to be fleshed out in prospective clinical research. As this relationship was found in a healthy population with relatively little variance in muscle quality, it is likely these relationships may be stronger in patients with low back pain or injury. Understanding the influence of microstructural features of muscle on posture may allow clinicians to prognostically categorize patients into groups, which may respond better to exercise based treatments. Future studies should take a more controlled approach, to determine if targeted exercise of the erector spinae muscles increases muscle quality (measured with DTI) and can elicit a postural response.

336 **KEY POINTS**

337 **FINDINGS:** Fractional anisotropy of the erector spinae was a significant predictor of lumbar
338 lordosis, lumbar flexion, and sacral tilt in several different operationally relevant positions in
339 active duty Marines.

340 **IMPLICATIONS:** The finding that fractional anisotropy is able to predict postural responses in
341 several positions, along with the absence of association between muscle volume and lumbar
342 spine posture, suggests muscle microstructure, but not quantity, is an important predictor of
343 lumbar spine posture.

344 **CAUTION:** These findings were found in a group of highly-active Marines and may not
345 translate to a civilian population.

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501

502 **Table Legend**

503 **TABLE 1.** Results from stepwise multiple linear regression.

504 **Supplemental Table Legend**

505 **SUPPLEMENTAL TABLE 1.** Coefficient of variation calculated for each dependent variable.

506 **SUPPLEMENTAL TABLE 2.** Coefficient of variation calculated for each independent
507 variable.

508

509 **TABLE 1.**

Dependent Variables	Significant Independent Variable(s)	β^*	R^2	P Value
Cobb angle: standing unloaded	none			
Cobb angle: standing loaded	ES FA	0.453	0.205	0.02
Cobb angle: sitting loaded	none			
Cobb angle: prone on elbows loaded	weight	0.468	0.219	0.016
Cobb angle: Δ load	none			
Cobb angle: Δ position	none			
Angle w.r.t. horizontal: standing unloaded	T2-L4/L5	-0.439	0.192	0.025
Angle w.r.t. horizontal: standing loaded	ES FA	0.514	0.264	0.007
Angle w.r.t. horizontal: sitting loaded	none			
Angle w.r.t. horizontal: prone on elbows loaded	ES FA	-0.480	0.23	0.013
Angle w.r.t. horizontal: Δ load	none			
Angle w.r.t. horizontal: Δ position	ES FA	0.455	0.207	0.02
Sacral angle: standing unloaded	ES FA	0.442	0.195	0.024
Sacral angle: standing loaded	ES FA	0.587	0.345	0.002
Sacral angle: sitting loaded	none			
Sacral angle: prone on elbows loaded	ES FA	0.562	0.316	0.003
Sacral angle: Δ load	none			
Sacral angle: Δ position	none			

Abbreviations: ES, erector spinae; FA, fractional anisotropy; Δ load, standing unloaded–standing loaded; Δ position, sitting loaded–prone on elbows loaded; w.r.t., with respect to.

**Standardized coefficient.*

510

511

512 **SUPPLEMENTAL TABLE 1.**

Position	Angle with Respect to Horizontal	Sacral Angle	Cobb Angle
Standing Unloaded	0.05	0.17	0.18
Standing Loaded	0.05	0.22	0.22
Sitting Loaded	0.04	0.48	0.16
Prone on Elbows Loaded	0.06	0.26	0.16
Delta Load	10.61	1.86	4.11
Delta Position	1.18	1.30	0.37

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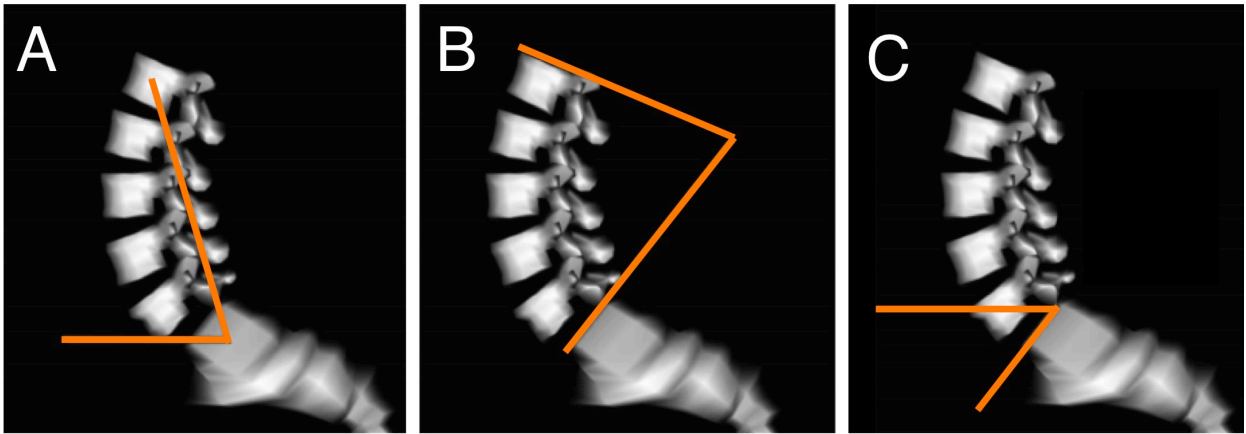
515 **SUPPLEMENTAL TABLE 2.**

		Independent Variable	Coefficient of Variation
Muscle Measures		Multifidus Volume	0.14
		Multifidus Fat Fraction	0.41
		Multifidus Mean Diffusivity	0.05
		Multifidus Fractional Anisotropy	0.08
		Multifidus $\lambda 1$	0.04
		Multifidus $\lambda 2$	0.04
		Multifidus $\lambda 3$	0.06
		Erector Spinae Volume	0.22
		Erector Spinae Fat Fraction	0.41
		Erector Spinae Mean Diffusivity	0.05
		Erector Spinae Fractional Anisotropy	0.07
		Erector Spinae $\lambda 1$	0.04
		Erector Spinae $\lambda 2$	0.04
		Erector Spinae $\lambda 3$	0.05
		Psoas Volume	0.13
		Quadratus Lumborum Volume	0.19
IVD Measures		T2-L1L2	0.24
		T2-L2L3	0.27
		T2 -L3L4	0.29
		T2-L4L5	0.35
		T2-L5S1	0.41
Anthro. Measures		Age	0.24
		Height	0.04
		Weight	0.12
		Body Mass Index	0.11

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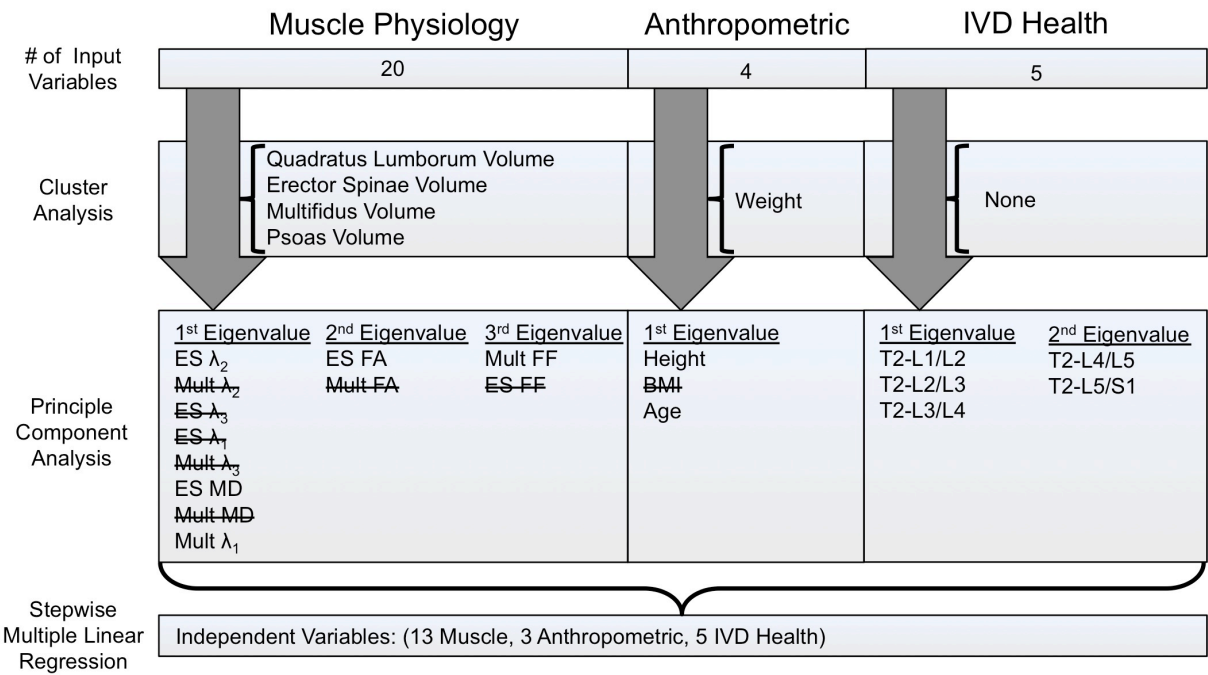
518 **FIGURE 1.**



519

520

521 **FIGURE 2.**



522



The effect of training on lumbar spine posture and intervertebral disc degeneration in active-duty Marines

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The effect of training on lumbar spine posture and intervertebral disc degeneration in active-duty Marines

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ABSTRACT

Military training aims to improve load carriage performance and reduce risk of injuries. Data describing the lumbar spine (LS) postural response to load carriage throughout training are limited. We hypothesised that training would reduce the LS postural response to load. The LS posture of 27 Marines was measured from upright MR images: with and without load (22.6 kg) at the beginning, middle, and end of School of Infantry (SOI) training. Disc degeneration was graded at L5–S1. No changes in posture and disc degeneration were found throughout training. During load carriage the LS became less lordotic and the sacrum more horizontal. Marines with disc degeneration had larger sacral postural perturbations in response to load. Our findings suggest that the postural response to load is defined more by the task needs than by the physical condition of the Marine.

Practitioner Summary: The effect of military training on lumbar spine posture is unknown. The lumbar posture of 27 Marines was measured from upright MR images, with and without load throughout infantry training. No changes in posture or IVD degeneration were found across training. Marines with degeneration at the L5–S1 level had larger sacral postural perturbations in response to load.

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Military; training; load carriage; posture; lumbar spine; sacral slope

Introduction

Low back pain (LBP) in the military population has been associated with carrying heavy loads during training and operational tasks (Attwells et al. 2006; Heir and Glomsaker 1996; Knapik, Harman, and Reynolds 1996; Taanila et al. 2009). 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 (Harman et al. 2008; Knapik et al. 1990; Swain et al. 2010). It is widely accepted that to improve high intensity load carriage performance, military training should consist of a combination of aerobic and resistance exercises (Knapik et al. 1990).

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 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, 10, 15 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 15 and 20 km 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 (Heir and Glomsaker 1996), which progressively increases intensity of load carriage via increased hike duration, there are no data documenting the behaviour 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

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physical examination and radiological evaluation; however, no overuse changes or new injuries in the LS were noted (Aharony et al. 2008). In a previous study, whole LS and lumbar level-dependent postural changes were measured in active-duty Marines while posteriorly carrying a load of 50.8 kg (Rodríguez-Soto et al. 2013). These changes appeared to be responses to centre of mass realignment (subject and backpack). More locally in the lumbar spine, these observed changes originated from the disparate postural behaviour of the superior and inferior LS. However, the Marines who were evaluated in this 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 Rodríguez-Soto et al.) had measurable, pre-existing structural changes in muscles, vertebrae and intervertebral discs (IVDs) that may have affected load-carrying posture (Aharony et al. 2008; Rodríguez-Soto et al. 2013).

The interaction between pelvic and LS posture has been previously investigated in the standing position (Jackson et al. 2000; Legaye et al. 1998; Vaz et al. 2002). The strongest association found exists between sacral slope and LS lordosis, which reveals that these two variables are proportional to each other (Jackson et al. 2000; Vaz et al. 2002). Meaning that in people with a more anteriorly rotated sacrum, the LS is more lordotic, and vice versa. Furthermore, reduced sacral slope and LS lordosis have been reported in the presence of IVD degeneration and LBP (Berthonnaud et al. 2005; Schwab et al. 2009). In the context of posterior 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 posterior 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 hypothesised the following: (1) load carriage-induced LS posture will change with training, and (2) Marines with IVD degeneration will manage loads and adapt to training differently than Marines without IVD degeneration.

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 ms; echo time 160 ms; field of view 32 cm; 224 × 224 acquisition matrix; 1.43 × 1.43 mm² pixel size; 4.5 mm slice thickness; 0.5 mm gap; number of averages 1, scan duration 2 min 30 s) 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 min (Figure 1). This load mass (22.6 kg) 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 but were instructed not to lean on surfaces or against the wall. After the 45 min 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 data-set was analysed as previously described (Rodríguez-Soto et al. 2013). 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,

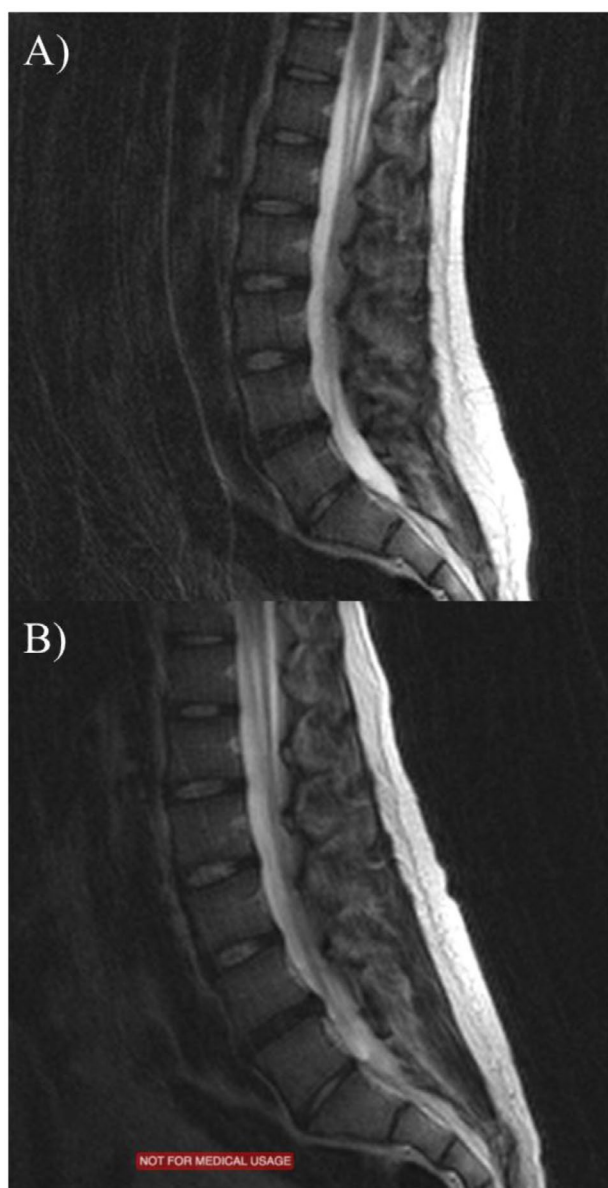


Figure 1. Representative sagittal magnetic resonance images of the lumbar spine without load (A) and with load (B).

and the resulting vertebral end-plate representations were used to generate postural measurements in the sagittal plane (Berry et al. 2015).

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 (Pfirrmann et al. 2001). 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 (Fujiwara et al. 2000).

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–S1) with respect to the ground (i.e. flexion, extension); however, it does not convey relative postural information between LS levels. When the LS is flexed this angular variable is reduced; extension has the opposite effect on this variable.
- *Sacral slope (SS)*: defined as the angle between the superior endplate of S1 and the horizontal. We consider SS a surrogate measurement of pelvic tilt, assuming that the motion between sacrum and pelvis is negligible. This variable describes sacral inclination: a small SS value indicates that the overall orientation of the sacrum is close to the vertical (S1 endplate is more horizontal), while a larger value describes a more horizontal sacrum (S1 endplate is more vertical).
- *Lumbar angle*: defined as the angle formed by the planes corresponding to the superior end-plates of L1 and S1 in the sagittal plane. As such, an increase in LS lordosis will be reflected by an increase in this angular variable, and viceversa. It has been previously reported that superior and inferior sections of the LS have different postural adaptations to load; therefore, we defined the *superior lumbar angle* as the angle formed by the superior endplate of L1 and the inferior endplate of L3, and the *inferior lumbar angle* as the angle between the superior endplate of L4 and inferior endplate of S1 (Rodríguez-Soto et al. 2013).
- *Segmental intervertebral (IV) 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 distribution was tested for normality using Shapiro-Wilk test. The absolute values of all variables were compared over 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 $\pm$ standard deviation (SD) values.

Results

Volunteer characteristics

Complete image data-sets for each time point were obtained from 27 Marines (mean $\pm$ 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/m²). 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 Table 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

None of the measured variables changed between loading tasks throughout training. The overall position of the spine was significantly ($p < 0.05$) more flexed when carrying load compared with those without load, at all time points (Figure 2, Supplemental Table 1). Simultaneously, the sacral slope significantly increased ($p < 0.05$) when carrying load, compared to its orientation when standing unloaded (Figure 3A). Marines with L5–S1 IVD degeneration had a larger ($p < 0.05$) change in sacrum orientation ($7.94^\circ \pm 4.17^\circ$) between unloaded and loaded tasks, compared to Marines without degeneration at the same lumbar level ($4.13^\circ \pm 4.18^\circ$; Figure 3B). Absolute SS values suggest that this difference is attributed to the orientation of the sacrum when loaded (degenerated $43.39^\circ \pm 4.01^\circ$, non-degenerated $38.38^\circ \pm 7.56^\circ$).

Additionally, we found that simultaneously to the LS flexion and sacrum orientation changes observed during load carriage, there was a reduction on whole LS lordosis ($p < 0.05$, Figure 4A). No significant differences were

Table 1. Distribution of 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	

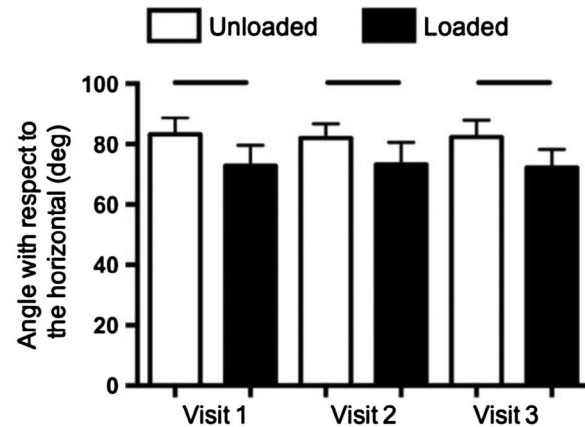


Figure 2. Trunk flexion measurements per task and visit.

Notes: Significant differences ($p < 0.001$), were found between unloaded (white) and loaded (loaded) tasks but not throughout training. Horizontal bars represent statistical difference ($p < 0.05$).

found within the unloaded to loaded conditions between L5–S1 IVD degeneration groups (Figure 4B). However, we observed a trend ($p = 0.07$, observed power 46%) towards reduced change in lordosis in the group with degeneration (Figure 4B). This suggests that individuals with L5–S1 degeneration may change posture in response to loading less than individuals without degeneration.

In order to investigate which LS regions contributed to the 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 (Figure 5A). The magnitude of change between tasks was not different between Marines regardless of the presence of degeneration at the L5–S1 IVD (Figure 5B).

The IV angles across lumbar levels are shown in Supplement Figure 1. We observed that overall, the magnitude of the response to load is larger at inferior lumbar levels than at superior levels (Figure S1). Specifically, the L1–L2 level became more lordotic (unloaded $5.05^\circ \pm 1.63^\circ$, loaded $6.01^\circ \pm 1.60^\circ$) in response to load – in contrast to inferior levels L3–L4 (unloaded $9.28^\circ \pm 1.80^\circ$, loaded $8.3^\circ \pm 2.45^\circ$), L4–L5 (unloaded $10.83^\circ \pm 2.23^\circ$, loaded $7.48^\circ \pm 3.56^\circ$) and L5–S1 (unloaded $10.83^\circ \pm 4.04^\circ$, loaded $6.82^\circ \pm 2.58^\circ$), which became less lordotic ($p < 0.05$, Figure S1 A–E). No

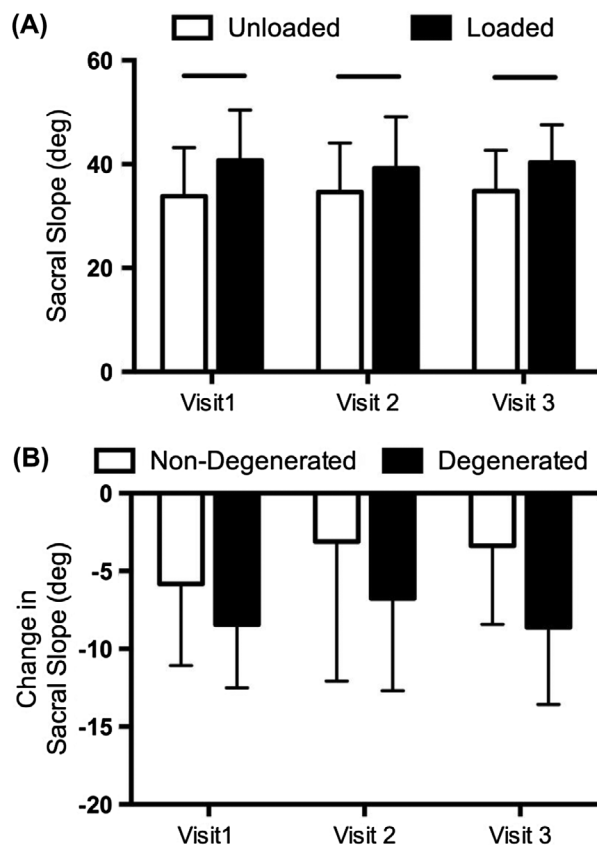


Figure 3. (A) Sacral slope (SS) per task and visit. Significant differences ($p < 0.001$), were found between unloaded (white) and loaded (loaded) tasks but not throughout training. (B) Change in sacral slope between tasks by L5–S1 IVD degeneration, throughout training.

Notes: 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 ($p < 0.05$).

postural changes were detected in response to load at the L2–L3, suggesting that it acts as a 'transition' level.

Anterior and posterior IVD distances at the L1–L2, L4–L5 and L5–S1 levels changed significantly ($p < 0.05$) in response to load. Overall, changes in regional IVD distances reflect postural kinematics throughout lumbar levels (Figure S2). The L1–L2 IVD was anteriorly distracted ($p < 0.05$) and posteriorly compressed ($p < 0.05$); while, L4–L5 and L5–S1 were anteriorly compressed and posteriorly distracted when carrying load ($p < 0.05$). Centrally, only L4–L5 became significantly more compressed when carrying load ($p < 0.05$).

Discussion

The main objective of this study was to measure the postural changes of the LS with and without posterior load throughout USMC SOI training. In terms of physical condition, the School of Infantry training includes both aerobic and resistance exercise (e.g. long training marches

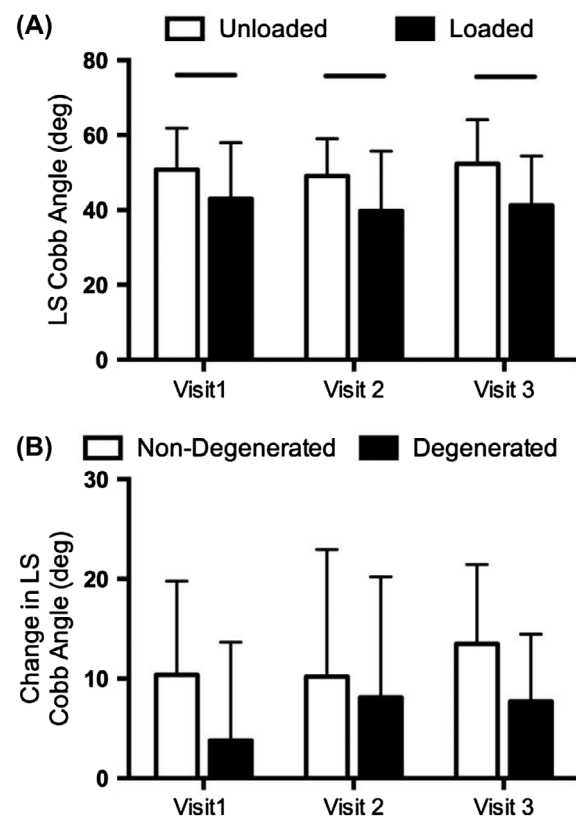


Figure 4. (A) Results for whole lumbar spine (LS) lordosis per task and visit. Overall LS became more straight during load carriage. (B) Change in LS lordosis between tasks by L5–S1 IVD degeneration groups.

Note: Horizontal bars represent statistical difference ($p < 0.05$).

and heavy load carriage), presumably improving Marines' endurance and strength while progressively exposing them to load carriage. Based on this paradigm, we hypothesised 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 hypothesised that the presence of IVD degeneration would alter LS postural adaptations to posterior load carriage. Other authors have evaluated the outcomes of military training in terms of physical condition testing and radiological evaluation of the IVDs [5–8]. 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,

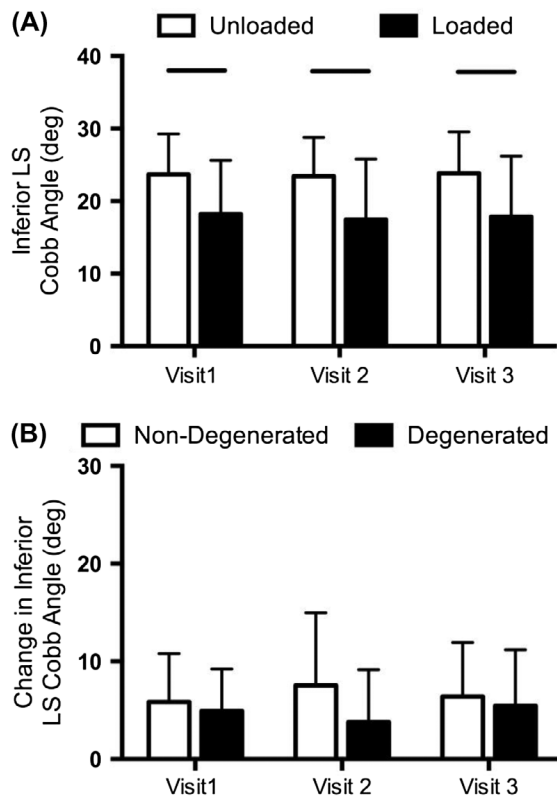


Figure 5. (A) Results for inferior lumbar spine (LS) lordosis per task and visit. Inferior LS became more straight during load carriage. (B) Change in inferior LS lordosis between tasks by L5–S1 IVD degeneration groups.

Notes: These data show that postural response to load is driven by changes in the inferior LS. Horizontal bars represent statistical difference ($p < 0.05$).

subjects with IVD degeneration demonstrated larger sacral perturbations and trended towards a smaller change in LS lordosis in response to load.

To quantify global LS posture, we measured LS flexion, whole LS and regional lordosis, and sacral slope. 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 (Al-Khabbaz, Shimada, and Hasegawa 2008; Attwells et al. 2006; Bust and McCabe 2005). This increase in lumbar flexion may be a compensatory response used to reorient the centre of mass of the system (body + loaded pack) over the feet (Bloom and Woodhull-Mcneal 1987; Knapik, Harman, and Reynolds 1996); however, this idea needs to be tested explicitly. In this study, LS flexion was on average $72.74 \pm 5.04^\circ$ (or 17.26° anterior to vertical) when carrying a load of approximately 25% BW. In a previous study (Rodríguez-Soto et al. 2013), LS flexion was roughly 52° (or 38° anterior to vertical) when carrying a load of 50 kg ($\sim 68\%$ BW). These findings suggest that there is a proportional increase in trunk flexion with increasing load, which is again consistent with previous literature (Knapik,

Harman, and Reynolds 1996; Knapik et al. 1990). For example, when using different methods (motion capture), Attwells et al. reported trunk flexion between 77° and 80° (with respect to the horizontal) when walking with loads of 15.95 kg (22% BW) on a waist belt and 20 kg (27% BW) in a backpack, respectively (Attwells et al. 2006). 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 $34.43^\circ \pm 8.3^\circ$, whereas most of the values previously reported in the literature range between 39° and 42° (Jackson et al. 2000; Vaz et al. 2002). The discrepancy between these data might be caused by the difference in measuring tools; 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 be 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 were 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 (Birrell and Haslam 2009; PASCOE et al. 1997; Smith et al. 2006). 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 slope, 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 $\sim 60^\circ$ of lordosis when standing without load and $\sim 55^\circ$ of lordosis when carrying 30% BW (Neuschwander et al. 2010). These values were obtained using a similar definition of the lumbar 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 min of standing with $\sim 25\%$ 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 min with 50 kg of load (Rodríguez-Soto et al. 2013). Interestingly, whole LS lordosis values reported by Neuschwander et al. (2010), Rodríguez-Soto et al. (2013), 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 (Rodríguez-Soto et al. 2013) and those of the present study, we identified that the superior LS had a larger increase in lordosis when carrying 50 kg of load than when carrying 26 kg. 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. The L1–L2 increment in lordosis may potentially cause the inferior endplate of L1 and the superior endplate of L2 to become parallel, as superior lumbar vertebral bodies are commonly kyphotic in nature (taller posterior than anteriorly). Contrastingly, inferior levels (L3–L4, L4–L5 and L5–S1) become more straight. 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 50 kg 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 in 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 (Herkowitz and International Society for Study of the Lumbar Spine 2004).

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. 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%. This has been previously attributed to the greater forces acting on these levels through the LS (Alexander et al. 2007; Pal and Routal 1987).

Marines were grouped into 'degeneration' and 'no-degeneration' based on the degenerative state of L5–S1. Although L5–S1 may not reflect the degenerative state of all lumbar IVDs this level was chosen because during load carriage the largest motion in the LS occurs at L4–L5 and L5–S1. Therefore, the idea that the presence of degeneration at these levels may influence the

posture of other regions of the LS was explored. In fact, 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 angle. 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 slope 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 (more horizontal S1 endplate) and reduced LS lordosis were reported (Barrey et al. 2007). Additionally, it has been suggested that in the presence of IVD degeneration at inferior lumbar levels a compensatory mechanism of increased lordosis occurs at superior lumbar levels (Lee et al. 2014; Rodríguez-Soto et al. 2013). However, no evidence of this phenomenon was found in this population.

Although only young men were included in the present study, it is relevant to discuss the LS postural response in other contexts; for example, the postural adaptations that the LS of women undergo during pregnancy. In this case, the LS of women progressively extends as the magnitude of the foetal load increases (Whitcome, Shapiro, and Lieberman 2007). Similar to the postural adaptations shown in the present study, it has been hypothesised that these adjustments allow realigning the position of the centre of mass in order to maintain balance. Interestingly, in both cases (soldiers posteriorly carrying load and pregnant women anteriorly carrying foetal load) the LS curvature is at extreme standing flexion/extension, during which increased incidence of back pain has been reported (Dumas et al. 1995; Östgaard et al. 1993 and Taanila et al. 2009). Consequently, future work should focus on identifying the biomechanical effect of LS posture. These data would help elucidate injury mechanisms and their impact on health outcomes.

In order to understand the biomechanical association between postural responses, it is necessary to consider both general orientation and lordosis of the LS. For instance, in the results presented here, although there were changes in IVD flexion that may alter the fraction of total compressive load resisted by IVD and facet joints, overall all levels remained in approximately 6–8° of flexion. This suggests that compressive load distribution between these two structures may be similar throughout the LS. However, in this case, the role of the orientation of each IVD with respect to the ground may be more relevant in determining the distribution of compressive and shear forces between

these structures. Nonetheless, the interplay between local lordosis and overall orientation of a vertebral joint with respect to the horizontal remains unknown.

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 ($1.43 \times 1.43 \times 4.5 \text{ mm}^3$) used in this study (Rodríguez-Soto et al. 2013). 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.

Conclusions

In conclusion, when Marines carry a 22.6 kg 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 angle 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.

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Disclosure statement

No potential conflict of interest was reported by the authors.

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Effect of Load Magnitude and Distribution on Lumbar Spine Posture in Active-duty Marines

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Study Design. Repeated measures.

Objective. The purpose of this study was to quantify the effect of operationally relevant loads and distributions on lumbar spine (LS) in a group of active-duty Marines.

Summary of Background Data. Low-back pain has been associated with heavy load carriage among military personnel. Although there are data describing the LS posture in response to load, the effect of varying load characteristics on LS posture remains unknown.

Methods. Magnetic resonance images of Marines (n = 12) were acquired when standing unloaded and when carrying 22, 33, and 45 kg of load distributed both 50% to 50% and 20% to 80% anteriorly and posteriorly. Images were used to measure LS and pelvic postures. Two-way repeated-measures ANOVA and posthoc tests were used to compare LS posture across load magnitudes and distributions ($\alpha = 0.05$). This project was funded by the US Army Medical Research Acquisition Activity, Award No. W81XWH-13-2-0043, under Work Unit No. 1310.

Results. No changes in LS posture were induced when load was evenly distributed. When load was carried in the 20% to 80% distribution lumbosacral flexion increased as a result of sacral anterior rotation and overall reduced lumbar lordosis. This

pattern was greater as load was increased between 22 and 33 kg, but did not increase further between 33 and 45 kg. We observed that the inferior LS became uniformly less lordotic, independently of load magnitude. However, the superior LS became progressively more lordotic with increasing load magnitude.

Conclusion. Postural adaptations were found only when load was carried with a posterior bias, suggesting that load-carriage limits based on postural changes are relevant when loads are nonuniformly distributed. Although the tendency would be to interpret that loads should be carried symmetrically to protect the spine, the relationship between postural changes and injury are not clear. Finally, the operational efficiency of carrying load in this distribution needs to be tested.

Key words: load carriage, load distribution, load magnitude, lumbar spine, military, posture, sacral slope.

Level of Evidence: 3

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Spine

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 The incidence of moderate or severe low-back pain (LBP) after a 1 year deployment to Afghanistan is around 22%.² Among soldiers, a questionnaire study 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 68 kg depending on duty position and nature of the mission.³ These loads are necessary to maintain soldiers' safety and to successfully fulfill their missions. Consequently, the effect of heavy load carriage on energy expenditure, situational awareness, and combat readiness has been extensively studied.^{4–6} Contrastingly, postural data of the musculoskeletal components of the lumbar spine (LS) during load carriage is limited. In previous work from our group, we quantified the deformation of the LS when carrying 50.8 kg in a group of active-duty Marines.⁷ These data indicated that heavy load carriage induced lumbosacral flexion and forward trunk lean. Other authors have described LS postural changes when carrying backpacks with increasing load magnitudes in the

10% to 30% of body weight (BW) range. However, these data have been measured in a pediatric population using different loads, which may not be representative of military personnel and operational conditions.^{9,10} Postural adaptations of the LS in response to load magnitudes and different anterior-posterior distributions have not been systematically studied in a military population.

This information may allow identifying potential LS injury mechanisms associated with load carriage and contribute to developing load carriage systems and limit recommendations based on measurable changes in LS posture. 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 active-duty Marines. We hypothesize that when loads are evenly distributed anterior and posterior, the deviation from the standing posture would be small compared with loads carried with a posterior bias. In addition, we hypothesize that as load magnitude increases lumbar lordosis would decrease and lumbosacral flexion would increase.

MATERIALS AND METHODS

Subjects

Twelve active-duty Marines (mean $\pm$ SD age, 23.41 ± 4.71 yrs; height, 177.8 ± 5.41 cm; weight, 76.77 ± 11.32 kg; body mass index [BMI], 24.15 ± 2.19 kg/m²) from the Marine Corps Base Camp Pendleton volunteered in this study. The average time of service of this group was 48 ± 39.96 months and their occupations were all of the

infantry rating (one officer, eight riflemen, two machine gunners, two assault men, and one unit leader). The University of California, San Diego, and Naval Health Research Center institutional review boards approved this study. 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) and flexible planar coil. A soft sleeve was used to retain the coil posteriorly at the LS level. During loaded scans, the coil was placed between the volunteer's back and load carriage system. A localizer and sagittal T2 weighted images (repetition time 1974 ms; echo time 160 ms; field of view 32 cm; 1.56×1.56 mm² pixel size; 4.5 mm slice thickness; 0.5 mm gap; scan duration 2:30 mins) were acquired.

Load-carrying Tasks

Marines were scanned standing unloaded (Figure 1A) and when carrying 22, 33, and 45 kg of load distributed both 50% to 50% (Figure 1B–D) and 20% to 80% (Figure 1E–G) anteriorly and posteriorly (AP), respectively. The first scan was always standing unloaded and the other six scans were randomized for all participants. Marines were not given instructions on how to stand, but were instructed to remain during image acquisition.

Load magnitudes of 22 and 33 kg were selected because they are the recommended load carriage limits for fighting and approach march loads, respectively.⁶ In addition, 45 kg is on the lower end of sustainment loads carried by Marines

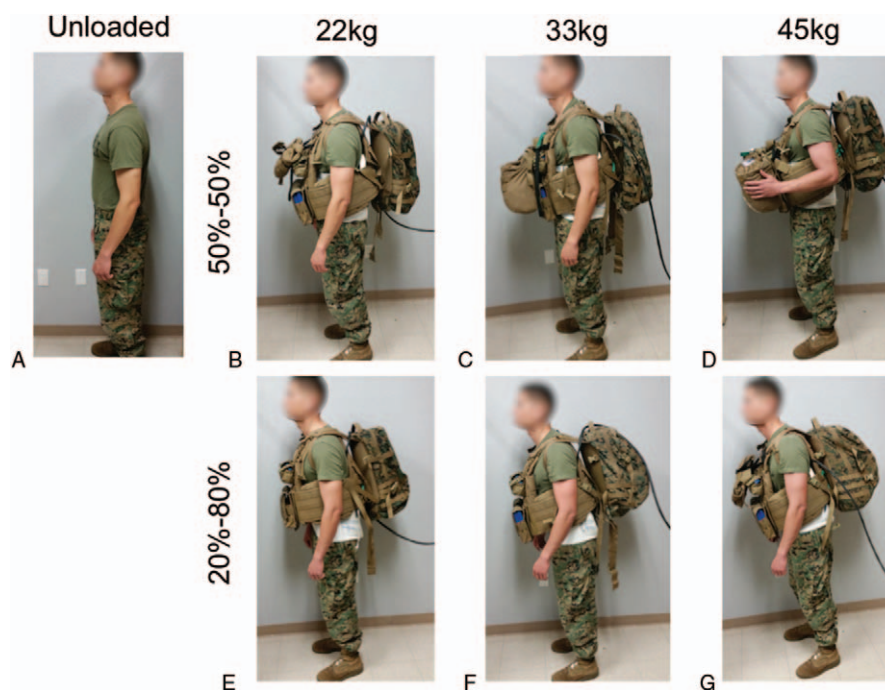


Figure 1. Representative pictures of a Marine standing (A) without external load, and carrying 22 kg (B and E), 33 kg (C and F), and 45 kg (D and G) in the 50% to 50% (top row) and 20% to 80% (bottom row) distributions.

during dismounted operation in Afghanistan.¹¹ The 50% to 50% and 20% to 80% AP load distributions were selected based on preliminary data (not shown) indicating that when loads are light (ie, average 12.3 kg) they are carried evenly. Moreover, it has been previously hypothesized that this load distribution induces minimal postural changes in the LS; therefore, we tested this concept.¹² Further, heavier loads are typically carried using a backpack. This load carriage paradigm has been reported to induce postural changes that progressively deviate from the standing posture as a function of load magnitude.^{12,13} Typically, load carried anterior is in the form of small pouches attached to the body armor; therefore, Marines wore body armor.

Data Analysis

Each data set was analyzed as previously described.^{7,8} 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 orientation.

Measurements

Postural measurements of the LS in the sagittal plane were generated based on vertebral endplates orientation. Angle with respect to the horizontal was defined as the angle between the centroid of L1, S1 and the horizontal line; it quantifies lumbosacral flexion. A relationship between LS posture and sacral slope (SS) has been previously reported.^{14,15} Therefore, 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, which describes the orientation of the sacrum.

Lumbar lordosis was measured using Cobb angle, defined as the angle between the superior endplates of L1 and S1 in the sagittal plane.^{16–18} Because of the disparate behavior among the superior and inferior LS during load carriage,⁷ 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 disc (IVD) angles and regional heights were measured between the planes of the inferior and superior endplates of adjacent vertebrae. Intervertebral heights were measured as the shortest distance between these endplates anteriorly, centrally, and posteriorly.

Postural adaptations to load carriage have been hypothesized to realign the center of gravity (CoG) over the base of support. Determining the system's CoG requires knowledge of the position and mass of the body segments.¹⁹ This information was not available and the use of x-rays was not approved. However, it has been shown that during static activities the location of the center of pressure (CoP) and CoG with respect to the base of support are highly correlated.¹⁹ Therefore, the CoP was measured using a pressure mat (Tekscan Inc., South Boston, MA). Ideally, these

measurements would be made during MRI acquisition; however, because of the ferromagnetic components of the mat, this was not possible. Alternatively, a mock scanner was built and the pressure mat was placed between the structure's wall. After the image acquisition for each configuration, Marines were asked to step on the mat and stand still for one minute. A minimum bounding box (MBB) (Figure 2A) is defined as the smallest rectangle that can fit all points of a determined dataset, in this case, the footprints.²⁰ This analysis allowed to account for the differences in feet position and size. The location of the CoP was expressed as the percentage of length and width of the MBB.

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 distribution. The comparison between each load carriage distribution (50%–50% or 20%–80%) and the unloaded condition were identified using one-way repeated-measures ANOVA with Sidak posthoc tests ($\alpha=0.05$). Statistical analyses were performed using SPSS Statistics software (version 20.0, IBM, Armonk, NY). All data are reported as mean $\pm$ SD values.

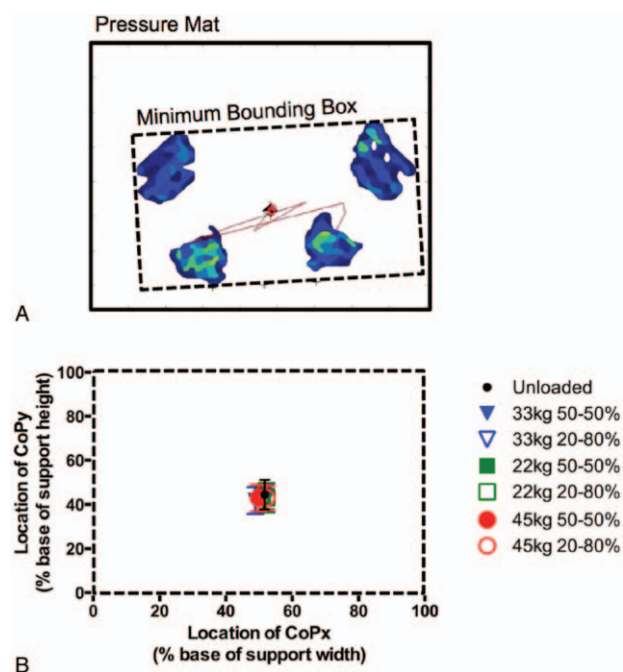


Figure 2. A, Representative image showing footprints acquired using pressure mat (solid rectangle) indicating minimum bounding box (MBB, dashed rectangle). The center of pressure (CoP) trajectory is shown in red. Red circle indicates the average location of the CoP. B, Plot of the location of the CoP as a percentage of the width and height of the MBB. No differences were found across load magnitudes and distributions.

RESULTS

Measurement of the CoP Location

The average location of the CoP along the width (left to right) of the BMM was $46.78 \pm 4.91\%$ and $46.38 \pm 6.06\%$ along the height (posterior to anterior). There was no significant difference in the location of the CoP between configurations (Figure 2B).

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% to 50% distribution did not have an effect on LS posture. Contrastingly, the overall position of the LS was significantly more horizontal in the 20% to 80% distribution compared with standing unloaded ($82.28 \pm 4.14^\circ$; Figure 3, solid bars). More specifically, these changes were significant different only when carrying 33 and 45 kg, but not when carrying 22 kg ($75.23 \pm 7.79^\circ$, Figure 3, asterisks). Interestingly, lumbosacral flexion values when carrying 33 and 45 kg were not different from each other ($64.77 \pm 7.91^\circ$ and $62.62 \pm 9.36^\circ$).

Sacral slope measurements had a similar response to load magnitude and distribution as lumbosacral flexion. In general, when loads were carried in the 20% to 80% distribution the orientation of the sacrum became more horizontal (Figure 4, solid bars). However, only 33 and 45 kg load magnitudes had a significant effect on sacrum orientation compared with standing without external load ($34.29 \pm 6.59^\circ$) and were not different from each other ($46.40 \pm 6.40^\circ$ and $50.76 \pm 8.35^\circ$; Figure 4, asterisks).

Whole LS lordosis (L1-S1) was also influenced by both load magnitude and distribution ($P < 0.05$); however, post-hoc tests revealed no differences between load magnitudes (Figure 5A). Overall, the LS became less lordotic ($P < 0.05$) when carrying load in the 20% to 80% distribution.

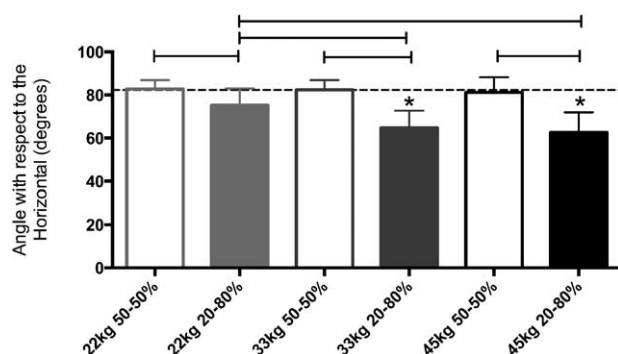


Figure 3. Lumbar spine flexion results for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22 kg (clear grey), 33 kg (dark grey), and 45 kg (black). The dashed line represents trunk flexion when standing without external load. Solid horizontal bars represent significant differences ($P < 0.05$) between load magnitudes and distributions. Asterisks represent significant differences when compared with the standing unloaded position.

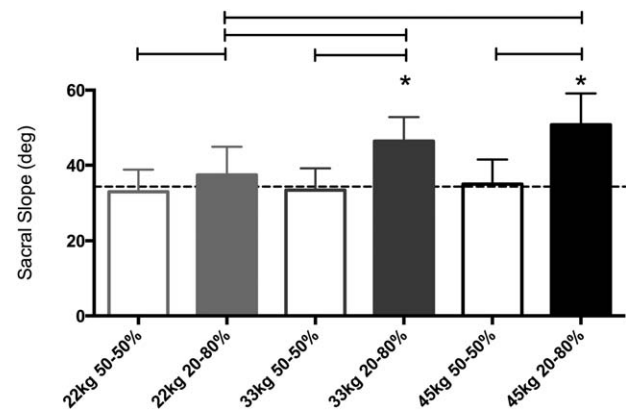


Figure 4. Sacral orientation results for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22 kg (clear grey), 33 kg (dark grey), and 45 kg (black). The dashed line represents sacral orientation when standing without external load. Solid horizontal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

Contrary to our initial hypothesis, only when carrying 22 kg, lumbar lordosis deviated from that of standing without external load (Figure 5A). No changes were found in whole LS lordosis when carrying 33 and 45 kg. However, regional lordosis measurements revealed that the upper and lower lumbar spine behave differently.

Both load magnitude and distribution had a significant effect on superior LS lordosis ($P < 0.05$). Superior LS became more lordotic when carrying 33 and 45 kg in the 20% to 80% distribution ($17.80 \pm 6.28^\circ$ and $16.88 \pm 5.49^\circ$) compared with the standing unloaded ($10.49 \pm 5.18^\circ$, Figure 5B, asterisks). The lordosis of the inferior LS was found to be affected solely by load distribution ($P < 0.05$). Interestingly, the inferior LS became straighter ($\sim 10^\circ$) independently of load magnitude when load was carried with a posterior bias (Figure 5C, solid bars).

In agreement with the changes in regional lordosis, load magnitude had a significant effect on superior levels, whereas load distribution influenced inferior levels (Sup. Figure 1, <http://links.lww.com/BRS/B179>). More specifically, L1-L2 IVD became more lordotic as load increased, again, no differences were found between 33 kg and 45 kg. At the L2-L3 level, both load magnitude and distribution had an effect on local lordosis ($P < 0.05$), however, posthoc tests revealed no differences between load magnitudes. A significant interaction between load magnitude and distribution was found at the L3-L4 level. At L4-L5 and L5-S1 levels, a significant effect of load distribution was observed. Both levels became less lordotic in the same amount independently of load magnitude.

Similar results were observed for changes in regional IVD distances, which reflect postural changes in IVD angle throughout lumbar levels (Figures S2, <http://links.lww.com/BRS/B179>). For example, when a functional spinal unit became less lordotic in response to load carriage,

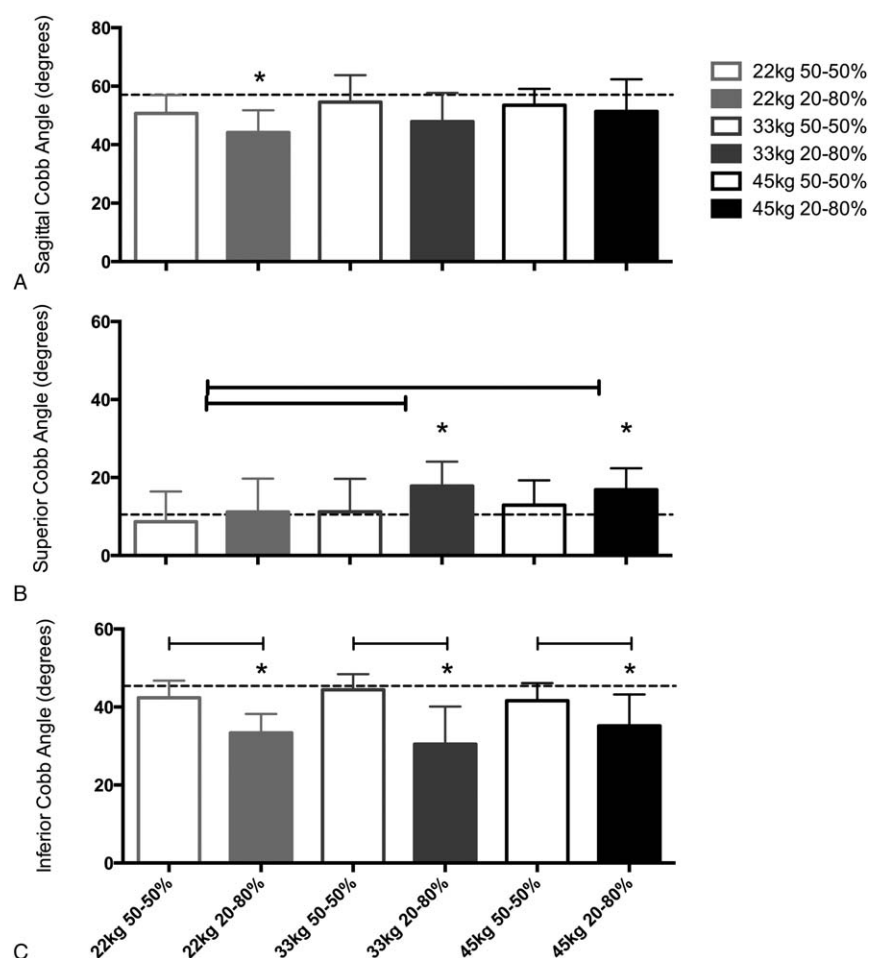


Figure 5. A, Whole, (B) superior, and (C) inferior lumbar lordosis results for loads carried with equal anterior-posterior distribution (clear bars) and with posterior bias (solid bars) for 22 kg (clear grey), 33 kg (dark grey), and 45 kg (black). The dashed line represents lordosis when standing without external load. Solid horizontal bars represent significant differences ($P < 0.05$) between load magnitudes and distributions. Asterisks represent significant differences when compared with the standing unloaded position.

anterior IVD distances decreased and posterior IVD distances increased.

DISCUSSION

The objective of this study was to measure the postural changes of the LS in response to operationally relevant load carried magnitudes and distributions. We hypothesized that the postural deviation of the LS when carrying load in a posterior bias would be larger than when the load was evenly distributed. Our results showed that independently of load magnitude, the LS posture was not different from that when standing without external load when carried in an even distribution. In addition, when load was carried with a posterior bias, lumbosacral flexion progressively increased. Interestingly, when a load of 22 kg was carried in this distribution the only postural difference in the LS is at inferior lumbar levels. Furthermore, the lack of differences between 33 and 45 kg suggests a postural adaptation plateau, indicating the contribution of active components of the musculoskeletal system to maintain the load carriage posture.

The observed increased lumbosacral flexion appears to result from the contribution of two postural mechanisms: anterior sacral rotation and reduced lordosis. Both SS and

lumbar lordosis have been previously measured during load carriage using motion capture and springs.^{12,21–24} However, in the current study we have measured these variables directly. Previously reported values of pelvic rotations were described as linear displacements and range of motion making it impossible to compare with our results. In addition, Birrell *et al*²⁵ showed that pelvic rotation linearly increased as a function of load magnitude.

In previous work from our group, Marines carried 50.8 kg in a backpack and found reduced lordosis. In the current 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 (~6 kg), but mostly to the disparity in load distribution. In the current study, a total load of 45 kg in the 20% to 80% distribution means that 36 kg were carried posteriorly, which is much smaller compared with 50.8 kg carried mostly posteriorly. This emphasizes the importance of load distribution, suggesting that careful attention should be given to this parameter in load carriage recommendations.

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 ($\sim 10^\circ$) 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 LS that determines the response of the superior LS to load, which is load dependent. 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, and during load carriage it increases cranially. In agreement with our previous work, the L3-L4 lumbar level behaved as a transition level between the superior and inferior LS. Furthermore, we compared local lordosis when carrying ~ 22 kg in the 20% to 80% distribution from the current study to another work from our group (unpublished data) where Marines carried the same amount of load in a similar distribution without body armor. We found that when carrying this load magnitude, the use of body armor did not affect lordosis distribution. 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.

An additional factor contributing to the LS posture during load carriage is the location of the load CoG in the superior-inferior direction. Although this was not measured in this experiment, its effect on overall trunk flexion has been previously studied.^{26,27} The response in trunk flexion has been shown to be larger when loads are carried more inferiorly, potentially leading to increased energy expenditure and paraspinal muscles contraction.²⁶ Reducing trunk flexion during load carriage would also result in reduced moments around all lumbar levels (biomechanical analysis not shown). These data may suggest that carrying load more superiorly has several advantages in terms of LS biomechanics and energy expenditure. However, further field studies are needed to incorporate LS biomechanics in response to load into military recommendations. For example, subjects with load carriage experience report that during marches carrying load superiorly is more unstable (more sway) and that this effect is increased in uneven terrains.²⁷

In conclusion, when Marines carry load in an evenly distribution 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 of the LS compared with a posteriorly biased distribution is unknown. During load carriage with a posterior bias, the position of the LS with respect to the ground was more horizontal. This resulted from anterior pelvic rotation and overall reduced lordosis; however, on average, all lumbar levels remained in flexion. Further research is needed to investigate the adaptation of other musculoskeletal tissues to different degrees of lordosis and how this might vary throughout lumbar levels, as they are morphologically different. These data would allow

narrowing down potential mechanisms of injury and to adjust physical training to further prevent low back-related injuries.

➤ Key Points

- ❑ Evenly distributed AP load carriage does not induce significant postural changes in the lumbar spine.
- ❑ Load carriage with a posterior bias induced lumbosacral flexion and overall reduced lumbar lordosis.
- ❑ Postural response of superior lumbar levels to posteriorly biased load carriage was load magnitude dependent, whereas the response of inferior lumbar levels was load distribution dependent.
- ❑ No significant postural differences were found between 33 and 45 kg, suggesting a postural adaptation plateau.

Supplemental digital content is available for this article. Direct URL citations appearing in the printed text are provided in the HTML and PDF version of this article on the journal's Web site (www.spinejournal.com).

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Lumbar Spine Musculoskeletal Physiology and Biomechanics During Simulated Military Operations

PR120576:

W81XWH-13-2-0043



PI: Samuel Ward, Karen Kelly

Org: UC San Diego, NHRC

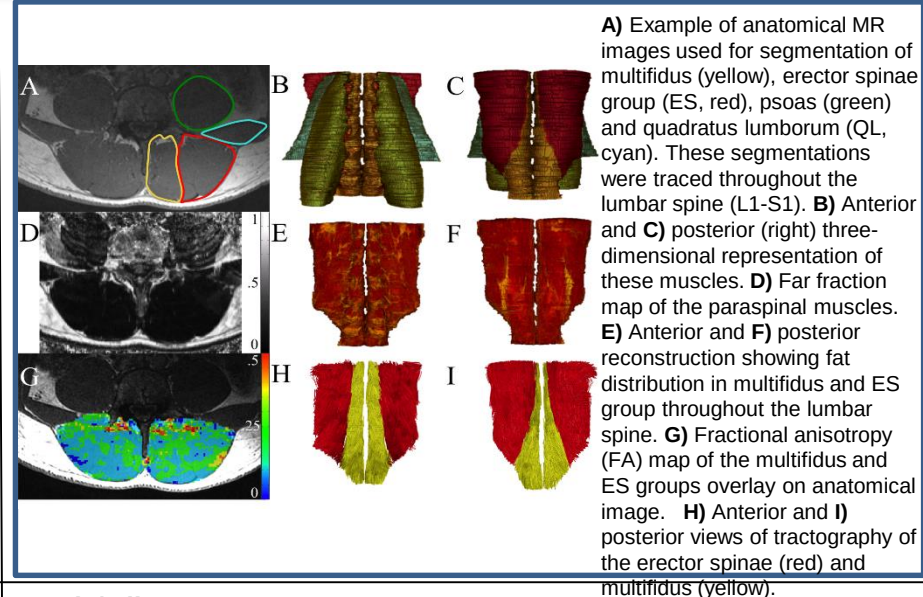
Award Amount: \$1,250,000.00

Study/Product Aim(s)

- To compare lumbar spine kinematics in simulated operational conditions in Marines with measureable disc pathology
- To quantify changes in lumbar spine muscle architecture in Marines with measureable lumbar disc pathology

Approach

Each Marine undergoes high-resolution structural and physiological imaging of the lumbar vertebrae and discs in a high strength (3T) supine MRI. Marines then undergo a 3D imaging protocol in an upright, low strength (0.6T) MRI in standing unloaded, standing loaded (25lb body armor), sitting loaded and prone on elbows positions. To investigate the effect of load magnitude and distribution, some Marines were scanned in standing unloaded and standing loaded conditions (6 conditions: 22kg, 33kg, 45kg x 50/50, 20/80 anterior/posterior). Back pain history is recorded for each subject at time of data collection.



Timeline and Cost

Activities	CY	13	14	15	16	17	18
Subject Battalion ID and Coordination							
MRI Data collection							
Data Analysis							
Dissemination of Findings							
Estimated Budget (\$K)		\$96	\$221	\$186	\$111	\$67	\$36

Goals/Milestones

CY13 Goal – Initiate subject recruitment/scanning

☒ Begin acquiring data on minimum 8 subjects

CY14 Goals – Develop measurement tools

☒ Validate tool for measuring kinematics, ☒ Scan >40 subjects

CY15 Goals – Data Analysis

☒ Scan remaining subjects, ☒ Analyze data from 80% of the subjects

CY16 Goals – Data Analysis/Disseminate findings

☒ Correlate kinematic physiologic data

☒ Disseminate Findings

CY17 Goal – Disseminate findings

☒ Disseminate Findings

CY18 Goal – Disseminate findings

☐ Disseminate Findings

Comments/Challenges/Issues/Concerns

Nothing to Report

Budget Expenditure to Date

Projected Expenditure: \$719,948 total cost (fully expend award through NCE)

Actual Expenditure: \$717,371 total cost

Updated: (20170601)