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BIOMECHANICS

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Study Design. Feasibility study on the acquisition of lumbar spine kinematic data from upright magnetic resonance images obtained under heavy load carrying conditions.

Objective. To characterize the effect of the load on spinal kinematics of active Marines under typical load carrying conditions from a macroscopic and lumbar-level approach in active-duty US Marines.

Summary of Background Data. Military personnel carry heavy loads of up to 68 kg depending on duty position and nature of the mission or training; these loads are in excess of the recommended assault loads. Performance and injury associated with load carriage have been studied; however, knowledge of lumbar spine kinematic changes is still not incorporated into training. These data would provide guidance for setting load and duration limits and a tool to investigate the potential contribution of heavy load carrying on lumbar spine pathologies.

Methods. Sagittal T2 magnetic resonance images of the lumbar spine were acquired on a 0.6-T upright magnetic resonance imaging scanner for 10 active-duty Marines. Each Marine was scanned without load (UN1), immediately after donning load (LO2), after 45 minutes of standing (LO3) and walking (LO4) with load, and after 45 minutes of side-lying recovery (UN5). Custom-made software was used to measure whole spine angles, intervertebral angles, and regional disc heights (L1–S1). Repeated measurements analysis of variance and *post hoc* Sidak tests were used to identify significant differences between tasks ($\alpha = 0.05$).

Results. The position of the spine was significantly ($P < 0.0001$) more horizontal relative to the external reference frame and lordosis was reduced during all tasks with load. Superior levels became more lordotic, whereas inferior levels became more kyphotic. Heavy load induced lumbar spine flexion and only anterior disc and posterior intervertebral disc height changes were observed. All kinematic variables returned to baseline levels after 45 minutes of side-lying recovery.

Conclusion. Superior and inferior lumbar levels showed different kinematic behaviors under heavy load carrying conditions. These findings suggest a postural, lumbar flexion strategy aimed at centralizing a heavy posterior load over the base of support.

Key words: low back pain, backpack, military, load carriage, MRI, upright MRI, lumbar spine kinematics.

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

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

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

MATERIALS AND METHODS

Subjects

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

Imaging

Marines were scanned using a 0.6-T MRI scanner (Upright MRI, Fonar Corporation, Melville, NY) and a planar RF coil. A brace was used to place the coil behind the volunteers' back at the lumbar spine (L1–S1) level while standing (Figure 1A). The brace was tight enough to keep the coil in place and avoid as much as possible any alteration of the volunteer's natural standing position. A 3-plane localizer and sagittal T2 weighted images (repetition time = 610 mS, echo time = 17 mS, field of view = 24 cm, 210 × 210 acquisition matrix, 4-mm slice thickness, no gap, scan duration 2 min) were acquired.

Load-Carrying Tasks

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

Data Analyses

A set of points was manually placed at the corners of the each vertebra on the images acquired in the upright MR scanner using the region of interest point tool available in Osirix (Pixmeo, Geneva, Switzerland) (Figure 2A).[30] The location of the seed points were used to fit planes to the inferior and superior endplates of each vertebra $\mathcal{L}$. To compare sagittal measurements between Marines, due to the differences in standing positions, Procrustes analysis was used to find the rotation matrix R between the inferior and superior vertebrae contiguous vertebrae with the inferior vertebra as reference. The rotation matrix R follows the x-y-z convention defined as follows: where ψ, θ, and ϕ are the rotations in radians around the x, y, and z axes, respectively. These are the Euler angles that

R_{11}

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

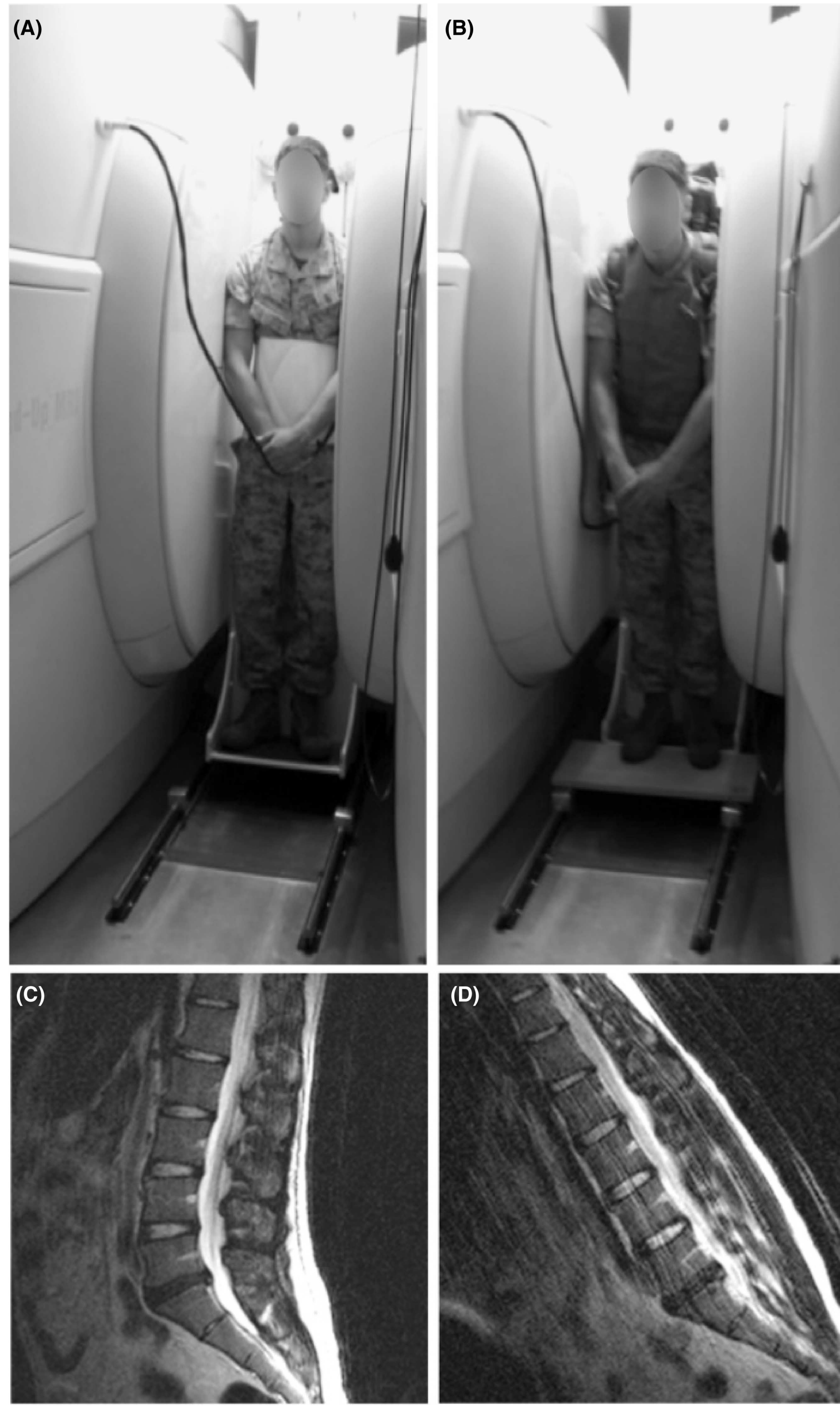

Figure 1. Photographs of a Marine standing in the MRI scanner without load (**A**) and with load (**B**). Representative midsagittal MR image of the lumbar spine without load (**C**) and with load (**D**). MRI indicates magnetic resonance imaging.

describe the orientation of a vertebra (L1–L5) with respect to the inferior contiguous vertebra (L2–S1). We solved for the Euler angles θ, ψ, and ϕ from the rotation matrix R and obtained the following:

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

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

Measurements

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

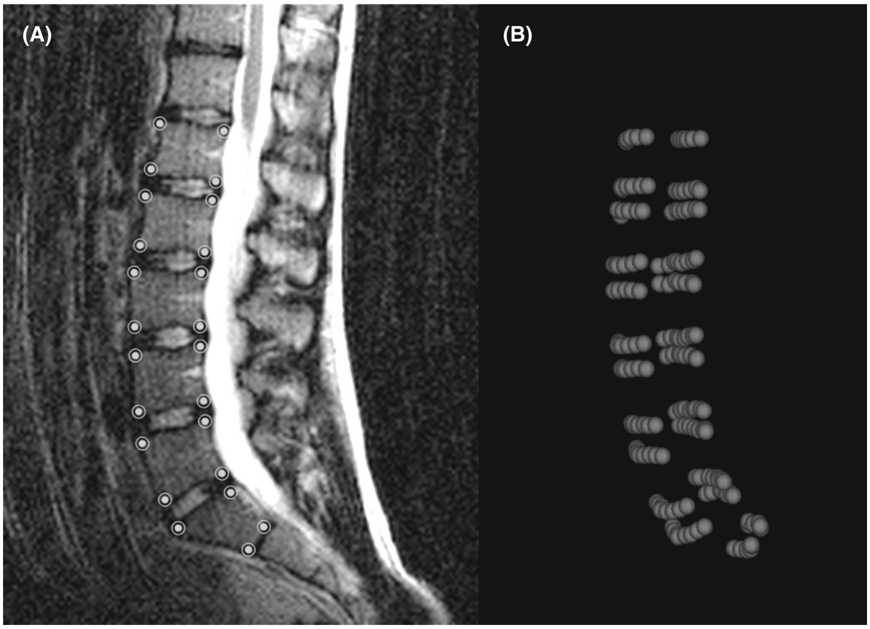

Figure 2. Spinal MR images in upright posture. (**A**) Example of upright MRI T2-weighted image of the lumbar spine with ROI points at the corners of each vertebra (L1–S1). (**B**) 3D representation of the lumbar spine vertebrae. MRI indicates magnetic resonance imaging; ROI, region of interest; 3D, 3-dimensional.

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

Resolution

To understand errors associated with resolution we acquired data at multiple in-plane resolutions and slice thicknesses. A volunteer was scanned using a 3.0-T MRI supine scanner (GE Discovery MR750) and GE 8ch CTL Spine Array Coil (Waukesha, WI). A 3-plane localizer and sagittal T2 Fast-recovery fast spin-echo (repetition time = 5000 ms, echo time = 17.2 ms, number of averages = 2, echo train length = 16, field of view = 25.6 cm, 1-mm thickness, no gap) were acquired at 8 different in-plane resolutions (0.5×0.5, 0.6×0.6, 0.7×0.7, 0.8×0.8, 0.9×0.9, 1.0×1.0, 1.1×1.1, and 1.2×1.2 mm^2). Additionally, these data sets were averaged in the slice direction to generate 2-mm-, 3-mm-, and 4-mm slice-thickness data. The $0.5 \times 0.5 \times 1.0$ mm^3 image set was analyzed 5 times by a single user; all kinematic data were averaged and used as a reference data set. The kinematic variables were measured from all other data sets and the root mean square errors (RMSE) between these and the reference values were calculated to investigate the effect of the in-plane and slice- thickness resolution on the precision of the kinematic measurements. The coefficient of variation (CV) was computed to assess the precision of the technique within and between users. For this, 5 data sets from the upright MRI (standing without load) were analyzed 3 times by 2 different users. The RMSE was calculated between the reference data set and a total of 31 data sets of varying in-plane resolution (0.5×0.5, 0.6×0.6, 0.7×0.7, 0.8×0.8, 0.9×0.9, 1.0×1.0, 1.1×1.1, and 1.2×1.2 mm^2) and slice thickness (1 mm, 2 mm, 3 mm, and 4 mm). The resolution of the images acquired in the upright MRI scanner was $1.14 \times 1.14 \times 4.0$ mm^3; therefore, we report the results at 1.1×1.1 mm^2 and 1.2×1.2 mm^2

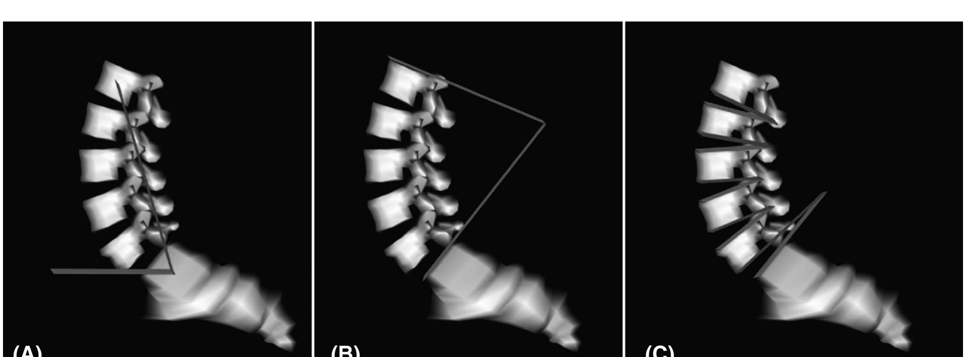

Figure 3. Lumbar spine kinematic measurements on a graphical representation of the lumbar spine. Angle with respect with the horizontal (**A**), sagittal Cobb angle (**B**), and intervertebral sagittal angles (**C**). These images were generated using OpenSim model of lumbar spine.[40,41]

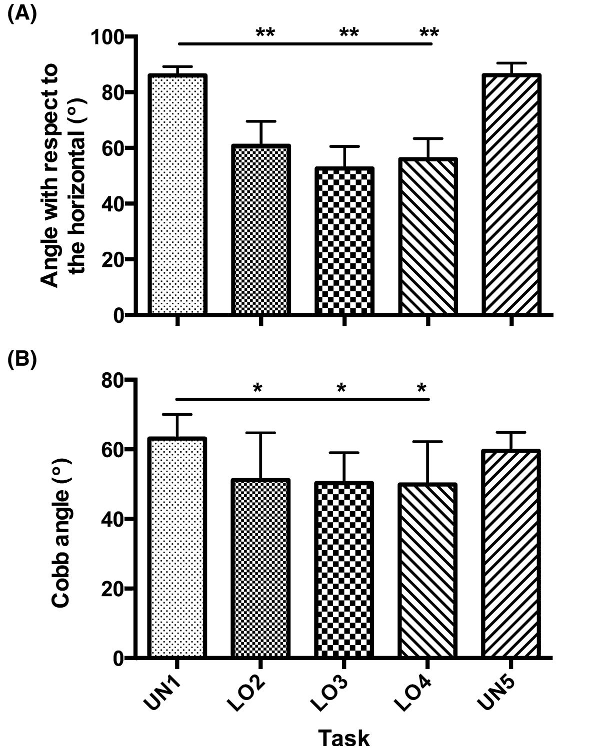

Figure 4. Whole lumbar spine kinematic measurements. (**A**) Angle with respect to the horizontal, (**B**) sagittal Cobb angle, per task. Significant differences were found between loaded and unloaded tasks ($P < 0.0001$) for both angle with respect to the horizontal and Cobb angle. UN1 indicates unloaded;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

in-plane resolutions and 4.0-mm slice thickness. The RMSE values for the angle with respect with the horizontal were 0.16° and 0.28°, whereas for the Cobb angle the values were 0.60° and 0.99°. The lumbar level dependent kinematic variables were averaged to report a single value per variable. The calculated RMSE values for the IV angles and heights were 0.39° and 0.54°, and 0.79 mm and 0.83 mm, respectively.

The within and between user CVs were calculated for 2 user and all kinematic variables, lumbar level dependent variables were averaged to obtain a single value. The within-user CVs for angle with respect to the horizontal, Cobb angle, and IVD angles, and heights were on average 0.2% (0.02°), 1.4% (0.44°), 3.9% (0.58°), and 3.3% (0.38 mm), respectively. The between-user CVs for angle with respect to the horizontal, Cobb angle, and IVD angles, and heights were 0.4% (0.04°), 2.7% (0.90°), 4.9% (0.72°), and 4.5% (0.55 mm).

Statistical Analysis

Angle with respect to the horizontal and Cobb angle were compared using 1-way repeated measurements analysis of variance and *post hoc* Sidak tests to identify significant differences between tasks ($\alpha = 0.05$). Additionally, IVD angle and distance measurements were compared by 2-way repeated measurements analysis of variance and *post hoc* Sidak tests to identify significant differences between lumbar levels through tasks. All data in plots are reported as means ± standard deviation unless otherwise stated.

RESULTS

Volunteers' Characteristics

Complete and usable images were obtained from 8 Marines (average age = 20.50 ± 1.17 yr, age range 19–22, average height = 179.3 ± 10.5 cm, average weight = 73.5 ± 8.0 kg, body mass index = 22.8 ± 1.6). The images from one Marine contained motion artifacts severe enough to make measurements impossible and were therefore not included in the reported results. Another Marine had a large body habitus and did not fit in the scanner; therefore, data were not collected.

Measurement of Lumbar Spine Kinematics

Regional kinematic measurements reflect that while carrying load (LO2, LO3, and LO4) the overall position of the spine was significantly more horizontal (25°–34°, $P < 0.001$) than at baseline and recovery; indicating forward flexion of the trunk (Figure 4A). Simultaneously, lumbar lordosis was also reduced (10°–13°, $P < 0.05$) when compared with unloaded tasks (Figure 4B).

Local IV angles and regional heights were measured to investigate their individual contribution to the observed regional changes. Lordosis was significantly ($P < 0.05$) increased from baseline (UN1) to after 45 minutes of standing with load (LO3) at L1–L2 and L2–L3 levels, 5° ± 2° and 3° ± 2°, respectively (Figures 5A–B). Posteriorly, L1–L2 was significantly distracted 2.5 ± 1.1 mm after 45 minutes of walking on the treadmill (LO4) when compared with the IV height after 45 minutes of standing with load (LO3, Figure 6K). This is in agreement with the increase in kyphosis observed from between these 2 tasks. Additionally, L2–L3 was significantly compressed centrally (Figure 6G) and posteriorly (Figure 6L) (0.8 ± 0.7 mm and 1.3 ± 1.4 mm, respectively) after standing for 45 minutes (LO3) with load in comparison with values at baseline (UN1) and immediately after donning load (LO2). No significant changes in anterior heights were found at either L1–L2 or L2–L3 (Figures 6A, B). In contrast, at L3–L4 level local lordosis was significantly decreased 4° ± 3° after 45 minutes of standing and walking with load (LO3 and LO4, Figure 5C). The magnitude of the kinematic changes was the largest at both L4–L5 and L5–S1 where local lordosis was significantly decreased (Figures 5D, E) 7° ± 2° immediately after donning load (LO2), 11° ± 3° after 45 minutes of standing with load (LO3), and 9° ± 4° after 45 minutes walking on the treadmill (LO4, Figure 5). In summary, we measured significant anterior and central compression, and posterior distraction during loaded tasks at these levels (Figure 6).

It was observed that superior lumbar levels (L1–L2, L2–L3, and L3–L4) show significant changes only after 45 minutes of standing with load, whereas L4 and L5 decrease immediately after donning load and during all loaded tasks (Figure 5). Local lordosis at all lumbar levels recovered to baseline values after the recovery period. Overall, through different tasks,

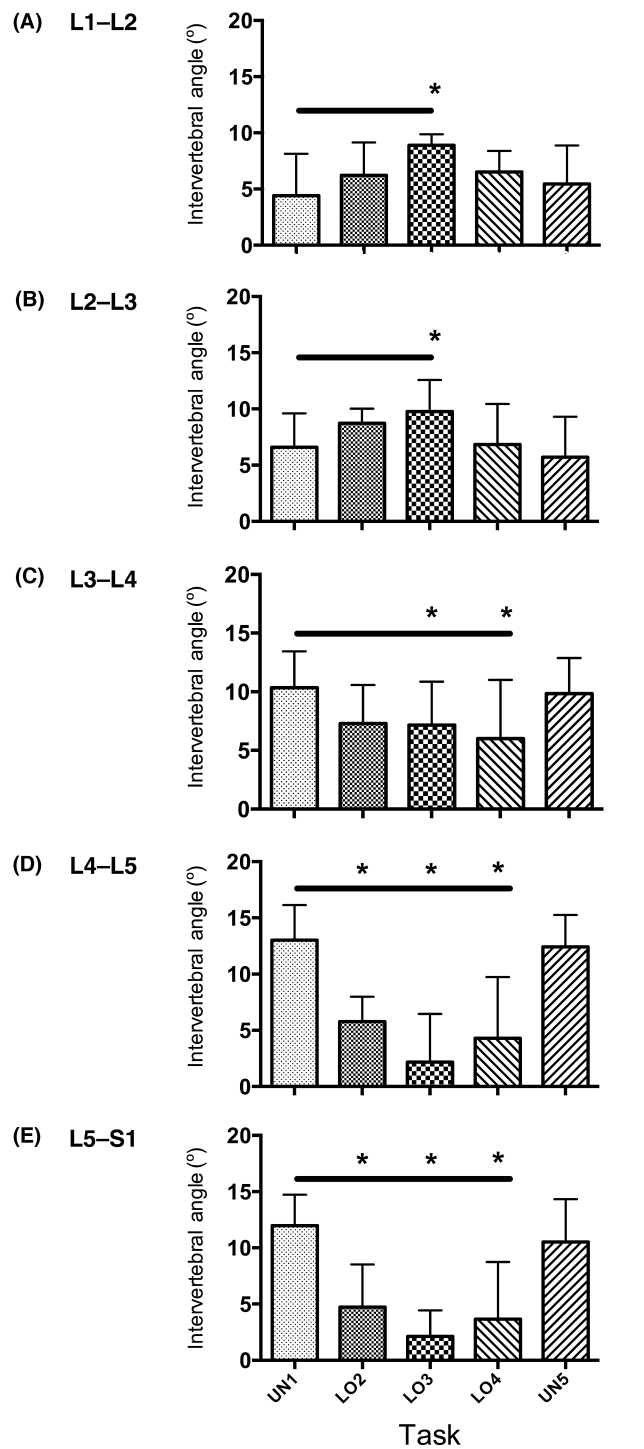

Figure 5. Intervertebral sagittal angle per lumbar level (L1–L5) and task. Most significant differences were found after LO3 ($P < 0.05$) through all lumbar levels: (**A**) L1-L2, (**B**) L2-L3, (**C**) L3-L4, (**D**) L4-L5 and (**E**) L5-S1. Additionally, L4–L5 and L5–S1 became significantly more kyphotic during all tasks with load ($P < 0.05$). UN1 indicates unloaded; LO2, immediately after donning load; LO3, after 45 minutes of standing with load on; LO4, after walking for 45 minutes with load on; UN5, after side-lying for 45 minutes.

the kinematic changes of superior lumbar levels (L1–L2 and L2–L3) are different from inferior levels (L4–L5 and L5–S1). Interestingly, L3–L4 showed significantly different ($P < 0.05$) kinematic behavior from L1–L2 and L2–L3 at baseline and after the recovery period, and from L4–L5 and L5–S1 after standing with load for 45 minutes. No significant differences were found between L3–L4 and other lumbar levels immediately after donning load or after walking for 45 minutes with load. In summary, the kinematic behavior of L3–L4 is similar to inferior levels during tasks without load and similar to superior levels under load-carrying conditions.

DISCUSSION

The objective of this study was to investigate the kinematic behavior of the lumbar spine of active-duty US Marines while carrying a heavy load. To our knowledge, this is the first study to measure level dependent lumbar spine kinematics in active-duty US Marines under load-carrying conditions using an upright MRI scanner. It was possible to scan Marines with backpack LCS loaded with 50.8 kg, an amount of weight that is typical of that carried in training and combat situations. However, a constraint of the technique is that the shoulder width of the subjects is limited to 31 inches. Although it is possible to scan a person with wider shoulders, the acquired images would not represent the true state of the spine when carrying load because they are supported. For this reason, it was verified that Marines were as comfortable as possible and standing on their own through the duration of the scans. Because Marines had to stand still in the scanner with the donned load, motion artifacts were found in some images; however, we considered these images to still be measurable (Figures 1C, D). In one case, severe motion artifacts were present and the data set was removed from the analysis. We measured the following lumbar spine kinematic variables: overall angle with respect to the horizontal, sagittal Cobb angle, IVD sagittal angles, and regional IVD heights. From these, the angle with respect to the horizontal has been previously reported to be progressively reduced in proportion to the amount of load been carried by soldiers.[16,35] Our results indicate a change in magnitude of angle with respect to the horizontal between unloaded and loaded conditions of 25° to 34° depending on the loaded task, which is larger than that reported by Attwells *et al*[16] of approximately 18°. The maximum load in both studies is about 50 kg. The variation in reported magnitude of this angle may be because of the different LCSs, measuring techniques and the location of body markers used by Attwells *et al.*[16] It has been suggested that this motion is aimed to reorient the center of mass of the system over the feet to keep balance.[9,36] Similarly, our results indicating a reduced lordosis when carrying load are in agreement with those in the literature, however, the magnitude of the results cannot be compared because of the differences in techniques and load weights.[37] The overall spine reduction in lordosis seems to be driven by the kinematic changes occurred at the L4–L5 and L5–S1 levels. The magnitude of the changes in these levels is significantly larger than that of L1–L2, L2–L3, and L3–L4 during load carrying tasks (data not shown).

Lumbar level-dependent lordosis data indicate that the superior and inferior lumbar spine has different behavior under load-carrying conditions. Superior lumbar spine

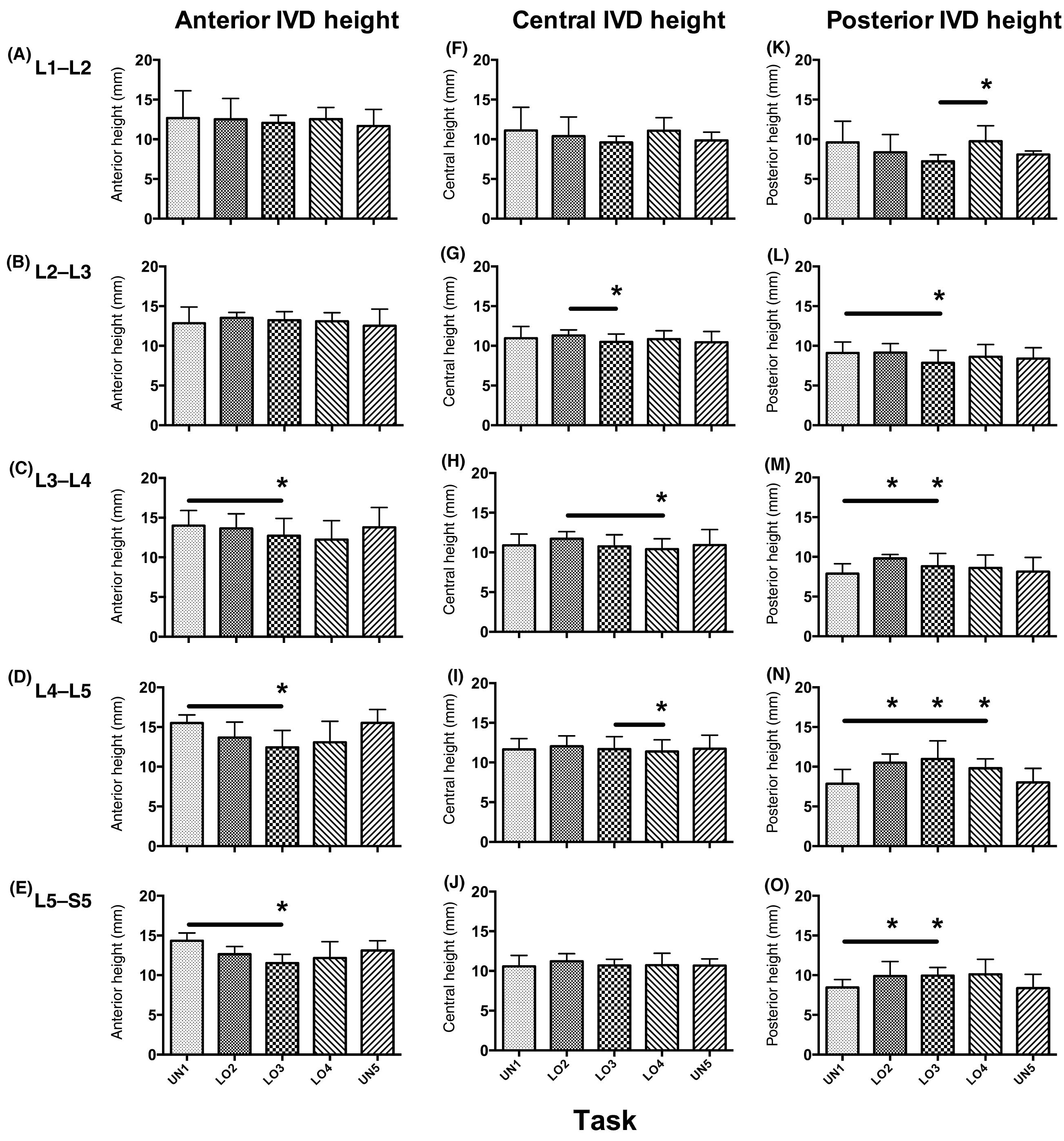

Figure 6. Anterior (**A–E**), central (**F–J**), and posterior (**K–O**) IVD heights at L1–L2, L2–L3, L3–L4, L4–L5, and L5–S1, per task. The IVDs of lumbar levels L3–L4, L4–L5 and L5–S1 were anteriorly compressed after LO3 and posteriorly distracted in most tasks with load ($P < 0.05$). Central IVD heights were significantly reduced between load-carrying tasks with ($P < 0.05$). UN1 indicates unloaded; LO2, immediately after donning load; LO3, after 45 minutes of standing with load on; LO4, after walking for 45 minutes with load on; UN5, after side-lying for 45 minutes; IVD, intervertebral disc.

levels present increased lordosis, whereas inferior levels become straighter when carrying load. However, the kinematics of L3–L4 seem to indicate a transition level between superior and inferior lumbar spine, whose behavior depends on the presence of load. Correspondingly, IVD height is anteriorly decreased and posteriorly increased in inferior lumbar levels. The fact that most significant changes occur in the lower lumbar spine might be related to the greater forces acting on inferior levels through the lumbar spine[38] and that IVDs of inferior levels undergo greatest posterior migration.[39]

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

LIMITATIONS

Upright MRI has the advantage of acquiring images while subjects are in functional and relevant positions. However, this system also has characteristics that impose limitations on this study. Although this technique does not allow measuring the kinematic changes during gait, it permits evaluation of the kinematic changes over time of exposure to load and the response to tasks with load in a natural standing posture. Importantly, the data acquired during this study reflect changes in kinematics after a task. We recognize that these standing postures may involve changes in muscle activity and orientation compared with the kinematics of the spine during the task.

The low-strength magnetic field of this system directly limits the in-plane resolution, slice thickness and scan time. In this study, we have shown that there are no significant differences between the kinematic variables measured from high-resolution ($0.5 \times 0.5 \times 1.0$ mm^3) images acquired using a 3-T supine MRI scanner and those from images collected with the 0.6-T upright MRI scanner ($1.14 \times 1.14 \times 4.0$ mm^3). Scan time was the main constraint of in-plane resolution and slice thickness to reduce the period of time that Marines had to stand still with donned load of 50.8 kg. The effect of acquiring thinner slices would be an increase in the acquisition time, which should remain as short as possible to reduce motion artifacts and assure the safety of Marines when carrying load. A balance between voxel size and scan time was then established to acquired images with tolerable motion artifact in standing position. Unfortunately, the selected slice thickness does not allow the IVD movement (*i.e.*, bulging, herniation) to be quantitatively assessed. Another disadvantage of this system is that the biochemical state of the IVDs cannot be described using techniques readily available on a high-resolution supine scanner (*i.e.*, T2 mapping, T1ρ, spectroscopy).

The field of view was limited by the size of the available planar RF coil, which did not permit the acquisition of other bony anatomical references and the lumbar spine in a single image set. Therefore, the rotational corrections applied in the axial and coronal planes were made with reference to the position of the superior plane of S1. This might result in intervertebral angles measured in the sagittal oblique plane; however, these angles still reflect the relative position between vertebrae of the lumbar spine. A supplemental imaging method that would yield useful information about sagittal and coronal balance is the use of long-dimension x-rays.

This study was performed in nonobese male Marines, which makes it difficult to compare our results with populations of different age, sex, and body habitus. Additionally, Marines in this study wore a body armor, which may reduce the range of motion of the lumbar spine. Assuming that the effect of the body armor on lumbar spine kinematics is negligible, it is possible that in subjects with increased range of motion such as children[17] and females[40] the magnitude of the changes observed are greater than those observed here. Inversely, in an older population with known decreased range of motion,[41] the magnitude of level-dependent kinematics is expected to decrease, therefore compromising the capacity of the lumbar spine to accommodate kinematic changes. In terms of body habitus, if our speculation that kinematic changes are driven by the need to maintain body center of mass, then obese individuals may be in a more lordotic posture without load and might lean forward less when load is applied.

CONCLUSION

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

➢ Key Points

- ❑ In a standing position, without carrying load over hips or shoulders, the inferior lumbar spine (L3–L5) is more lordotic than the superior lumbar spine (L1–L2).
- ❑ Immediately after load is donned the superior lumbar spine increases its lordosis and the inferior lumbar spine becomes more kyphotic; this effect increases with time of exposure to load in standing position.
- ❑ These changes drive a decrease of the overall lumbar spine (L1–L5) lordosis together with trunk flexion.

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References

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

2. Dean C. *The Modern Warrior's Combat Load. Dismounted Operations in Afghanistan*. Ft Leavenworth, KS: US Army Center for Army Lessons Learned; 2004.
3. US Department of the Army. *Foot Marches* 21-18. *Field Manual*. Washington, DC: Department of the Army; 1990.
4. Datta SR, Ramanathan NL. Ergonomic comparison of seven modes of carrying loads on the horizontal plane. *Ergonomics* 1971;14:269–78.
5. Bhambhani Y, Buckley S, Maikala R. Physiological and biomechanical responses during treadmill walking with graded loads. *Eur J Appl Physiol Occup Physiol* 1997;76:544–51.
6. May B, Tomporowski PD, Ferrara M. Effects of backpack load on balance and decisional processes. *Mil Med* 2009;174:1308–12.
7. Polcyn AF, Besseel CK, Harman EA, et al. *The Effects of Load Weight: A Summary Analysis of Maximal Performance, Physiological, and Biomechanical Results From Four Studies of Load-Carriage Systems*. Natick, MA: US Army Soldier Systems Center; 2007.
8. Johnson RF, Knapik JJ, Merullo DJ. Symptoms during load carrying: effects of mass and load distribution during a 20-km road march. *Percept Mot Skills* 1995;81:331–8.
9. Knapik J, Harman E, Reynolds K. Load carriage using packs: a review of physiological, biomechanical and medical aspects. *Appl Ergon* 1996;27:207-16.
10. Cohen SP, Gallagher RM, Davis SA, et al. Spine-area pain in military personnel: a review of epidemiology, etiology, diagnosis, and treatment. *Spine J* 2012;12:833–42.
11. Ulaska J, Visuri T, Pulkkinen P, et al. Impact of chronic low back pain on military service. *Mil Med* 2001;166:607–11.
12. Toblin RL, Riviere LA, Thomas JL, et al. Grief and physical health outcomes in US soldiers returning from combat. *J Affect Disord* 2012;136:469–75.
13. Meakin JR, Smith FW, Gilbert FJ, et al. The effect of axial load on the sagittal plane curvature of the upright human spine in vivo. *J Biomech* 2008;41:2850–4.
14. Yen SC, Gutierrez GM, Ling W, et al. Coordination variability during load carriage walking: can it contribute to low back pain? *Hum Mov Sci* 2012;31:1286–301.
15. Bettany-Saltikov J, Cole L. The effect of frontpacks, shoulder bags and handheld bags on 3D back shape and posture in young university students: an ISIS2 study. *Stud Health Technol Inform* 2012;176:117–21.
16. Attwells RL, Birrell SA, Hooper RH, et al. Influence of carrying heavy loads on soldiers' posture, movements and gait. *Ergonomics* 2006;49:1527–37.
17. Neuschwander TB, Cutrone J, Macias BR, et al. The effect of backpacks on the lumbar spine in children: a standing magnetic resonance imaging study. *Spine (Phila Pa 1976)* 2010;35:83–8.
18. Schiffman JB, Bensel CK, Hasselquist L, et al. *The Effects Of Soldiers' Loads on Postural Sway*. Natick, MA: US Army Soldier Systems Center; 2004.
19. Filaire M, Vacheron JJ, Vanneuville G, et al. Influence of the mode of load carriage on the static posture of the pelvic girdle and the thoracic and lumbar spine in vivo. *Surg Radiol Anat* 2001;23: 27–31.
20. Fujii R, Sakaura H, Mukai Y, et al. Kinematics of the lumbar spine in trunk rotation: in vivo three-dimensional analysis using magnetic resonance imaging. *Eur Spine J* 2007;16:1867–74.
21. Kimura S, Steinbach GC, Watenpaugh DE, et al. Lumbar spine disc height and curvature responses to an axial load generated by a compression device compatible with magnetic resonance imaging. *Spine (Phila Pa 1976)* 2001;26:2596–600.
22. Hioki A, Miyamoto K, Sakai H, et al. Lumbar axial loading device alters lumbar sagittal alignment differently from upright standing position: a computed tomography study. *Spine (Phila Pa 1976)* 2010;35:995–1001.
23. Willen J, Danielson B, Gaulitz A, et al. Dynamic effects on the lumbar spinal canal: axially loaded CT-myelography and MRI in patients with sciatica and/or neurogenic claudication. *Spine (Phila Pa 1976)* 1997;22:2968–76.
24. Alyas F, Connell D, Saifuddin A. Upright positional MRI of the lumbar spine. *Clin Radiol* 2008;63:1035–48.
25. Jinkins JR, Dworkin J. Upright, weight-bearing, dynamic-kinetic MRI of the spine: pMRI/kMRI. In: *Proceedings of the State-of-the-Art Symposium on Diagnostic and Interventional Radiology of the Spine*; September 7, 2003 (Part two); Antwerp, Belgium. *JBR-BTR* 2003;86:286–93.
26. Wang S, Passias P, Li G, Wood K. Measurement of vertebral kinematics using noninvasive image matching method-validation and application. *Spine (Phila Pa 1976)* 2008;33:E355–61.
27. Janssen M, Nabih A, Moussa W, et al. Evaluation of diagnosis techniques used for spinal injury related back pain. *Pain Res Treat* 2011;2011:478798.
28. Knapik JJ, Harman EA, Steelman RA, et al. A systematic review of the effects of physical training on load carriage performance. *J Strength Cond Res* 2012;26:585–97.
29. Williams AG, Rayson MP, Jones DA. Training diagnosis for a load carriage task. *J Strength Cond Res* 2004;18:30–4.
30. Rosset A, Spadola L, Ratib O. OsiriX: an open-source software for navigating in multidimensional DICOM images. *J Digit Imaging* 2004;17:205–16.
31. Cobb JR. Outline for the study of scoliosis. *AAOS, Instr Course Lect* 1948;5:261–75.
32. Schmitz A, Kandyba J, Koenig R, et al. A new method of MR total spine imaging for showing the brace effect in scoliosis. *J Orthop Sci* 2001;6:316–9.
33. Geijer H, Beckman K, Jonsson B, et al. Digital radiography of scoliosis with a scanning method: initial evaluation. *Radiology* 2001;218:402–10.
34. Asghar J, Samdani AF, Pahys JM, et al. Computed tomography evaluation of rotation correction in adolescent idiopathic scoliosis: a comparison of an all pedicle screw construct versus a hook-rod system. *Spine (Phila Pa 1976)* 2009;34:804–7.
35. Bust PD, McCabe PT. Contemporary Ergonomics 2005. In: *Proceedings of the International Conference on Contemporary Ergonomics (CE2005)*, April 5–7, 2005; Hatfield, UK. Taylor & Francis; 2005.
36. Bloon DW-MP. Postural adjustments while standing with two types of loaded backpacks. *Ergonomics* 1987;30.
37. Chow DH, Leung KT, Holmes AD. Changes in spinal curvature and proprioception of schoolboys carrying different weights of backpack. *Ergonomics* 2007;50:2148–56.
38. Pal GP, Routal RV. Transmission of weight through the lower thoracic and lumbar regions of the vertebral column in man. *J Anat* 1987;152:93–105.
39. Alexander LA, Hancock E, Agouris I, et al. The response of the nucleus pulposus of the lumbar intervertebral discs to functionally loaded positions. *Spine (Phila Pa 1976)* 2007;32:1508–12.
40. Van Herp G, Rowe P, Salter P, et al. Three-dimensional lumbar spinal kinematics: a study of range of movement in 100 healthy subjects aged 20 to 60+ years. *Rheumatology (Oxford)* 2000;39:1337–40.
41. Keorochana G, Taghavi CE, Lee KB, et al. Effect of sagittal alignment on kinematic changes and degree of disc degeneration in the lumbar spine: an analysis using positional MRI. *Spine (Phila Pa 1976)* 2011;36:893–8.

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14. ABSTRACT
PURPOSE: To measure kinematic changes in the lumbar spines of Marines carrying heavy loads. We hypothesized that heavy loads would increase lumbar lordosis and disc compression.
METHODS: Sagittal T2 MR images of the lumbar spine were acquired on a 0.6T vertical MRI scanner in 10 active duty Marines. Each was scanned without load (SW1), immediately after donning body armor and a backpack (SL2), after 45 min of standing with the loa after walking for 45 min with load (SL4), and after 45 min of side-lying recovery (SW5). In order to characterize the kinematic behavi functional spinal units (L1-S1) and overall lumbar spine during each task a 3D geometric representation of the spine was generated processing of the images acquired using an upright MRI scanner. Repeated measurements ANOVAs and post hoc Sidak tests were identify significant differences between tasks ($\alpha = 0.05$).
RESULTS: The position of the spine was significantly ($p<0.0001$) more horizontal relative to the external reference frame and lord reduced during all tasks with load compared with those without load. Superior and inferior lumbar levels showed different kinematic be When carrying load, superior levels became more lordotic while inferior levels became more kyphotic. All kinematic variables ret baseline levels after 45 min of side-lying recovery.
CONCLUSIONS: Heavy load carrying induced lumbar spine flexion and only anterior disc and posterior intervertebral disc height were observed, contrary to our initial hypothesis. These findings suggest a postural, lumbar flexion strategy aimed at centralizing posterior load over the base of support. (BUMED-WII42, WU61016, US ARMY RDECOM N6311610MP00182)

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