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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 reducing risk of injuries. Data describing the lumbar spine (LS) postural response to load carriage throughout training are limited. We hypothesized 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 (22.6kg) at the beginning, middle and end of School of Infantry (SOI) training. Disc degeneration was graded at L5−S1. ANOVA and *post-hoc* tests were used to compare posture across training and by tasks, and disc health (α=0.05). No changes in posture and disc degeneration were found throughout training. During load carriage the LS became less lordotic and the sacrum rotated anteriorly. 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 and IVD degeneration were found across training. Marines with degeneration at the L5-S1 level had larger sacral postural perturbations in response to load.

Keywords

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—5km, 10km, 15km, and 20km performed around days 12, 16, 28, and 40—under load. All marches are conducted with a standard fighting load, which is approximately 33.6kg. During the 15km and 20km training marches, Marines are also required to carry their designated weapon system during training.

Despite the association between LBP, load carriage and the structured SOI training paradigm (Glomsaker 1996), which progressively increases intensity of load carriage via increased hike duration, there are no data documenting the behavior of spinal structures as Marines progress through SOI. To date, one study by Aharony et al.

measured the impact of Israeli Navy Special Forces training on lumbar spine (LS) pathology through physical examination and radiological evaluation; however, no overuse changes or new injuries in the LS were noted (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 (Rodriguez-Soto et al. 2013). These changes appeared to be responses to center of mass realignment (subject and backpack). More locally in the lumbar spine, these observed changes originated from the disparate postural behavior 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 Rodriguez-Soto *et al.*) had measureable, pre-existing structural changes in muscles, vertebrae, and intervertebral discs (IVDs) that may have affected load-carrying posture (Rodriguez-Soto et al. 2013, Aharony et al. 2008).

The interaction between pelvic and LS posture has been previously investigated in the standing position (Legaye et al. 1998, Vaz et al. 2002, Jackson et al. 2000). 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 hypothesized 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 lose enough not to alter the volunteer's natural standing position. A three-plane localizer and sagittal T2-weighted images (repetition time 1974 msec; echo time 160 msec; field of view 32 cm; 224×224 acquisition matrix; 1.43×1.43 mm^2 pixel size; 4.5mm slice thickness; 0.5 mm gap; number of averages 1, scan duration 2 minutes 30 seconds) were acquired.

Load-Carrying Tasks

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

Data Analysis

Each image data set was analyzed as previously described (Rodriguez-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, 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 to 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 lordosis*: 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 lordosis* as the angle formed by the superior endplate of L1 and the inferior endplate of L3, and the *inferior lumbar lordosis* as the angle between the superior endplate of L4 and inferior endplate of S1 (Rodriguez-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 ± standard deviation (SD) values.

Results

Volunteer Characteristics

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

Measurement of IVD Degeneration

The distribution of the Pfirrmann grades by lumbar level is shown in 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 (Fig. 2, Supplemental Table 1). Simultaneously, the sacral slope significantly increased ($p<0.05$) when carrying load, compared to its orientation when standing unloaded (Fig. 3A). Marines with L5-S1 IVD degeneration had a larger ($p<0.05$) change in sacrum orientation (7.94°±4.17°) between unloaded and loaded tasks, compared to Marines without degeneration at the same lumbar level (4.13°±4.18°; Fig. 3B). Absolute SS values suggest that this difference is attributed to the orientation of the sacrum when loaded (degenerated 43.39°±4.01°, non-degenerated 38.38°±7.56°).

Additionally, we found that simultaneously to the LS flexion and sacrum orientation changes observed during load carriage, there was a reduction on whole LS lordosis ($p<0.05$, Fig. 4A). No significant differences were found within the unloaded to loaded conditions between L5–S1 IVD degeneration groups (Fig. 4B). However, we observed a trend ($p=0.07$, observed power 46%) towards reduced change in lordosis in the group with degeneration (Fig. 4B). This suggests that individuals with L5S1 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 (Fig. 5A).

The magnitude of change between tasks was not different between Marines regardless of the presence of degeneration at the L5-S1 IVD (Fig. 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 (Fig. S1). Specifically, the L1–L2 level became more lordotic (unloaded 5.05°±1.63°, loaded 6.01°±1.60°) in response to load—in contrast to inferior levels L3-L4 (unloaded 9.28°±1.80°, loaded 8.3°±2.45°), L4–L5 (unloaded 10.83°±2.23°, loaded 7.48°±3.56°) and L5–S1 (unloaded 10.83°±4.04°, loaded 6.82°±2.58°), which became less lordotic ($p<0.05$, Fig. S1 A–E). No postural changes were detected in response to load at the L2–L3, suggesting that it acts as a “transition” level.

Anterior and posterior IV 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 (Fig. 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 and heavy load carriage), presumably improving Marines' endurance and strength while progressively exposing them to load carriage. Based on this paradigm, we hypothesized that it would become progressively easier to carry a fixed-load magnitude over the training period because of improvements in endurance, strength, and motor learning. Additionally, we hypothesized that the presence of IVD degeneration would alter LS postural adaptations to 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, 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 center of mass of the system (body + loaded pack) over the feet (Bloon 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±5.04° (or 17.26° anterior to vertical) when carrying a load of approximately 25% BW. In a previous study (Rodriguez-Soto et al. 2013), LS flexion was roughly 52° (or 38° anterior to vertical) when carrying a load of 50kg (~68% BW). These findings suggest that there is a proportional increase in trunk flexion with increasing load, which is again consistent with previous literature (Knapik, Harman, and Reynolds 1996, Knapik et al. 1990). For example, when using different methods (motion capture), Atwells *et al*. reported trunk flexion between 77° and 80° (with respect to the horizontal) when walking with loads of 15.95kg (22% BW) on a waist belt and 20kg (27% BW) in a backpack, respectively (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°±8.3°, 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 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 ~60° of lordosis when standing without load and ~55° of lordosis when carrying 30% BW (Neuschwander et al. 2010). These values were obtained using a similar definition of the lumbar lordosis angle used in this study, but they were measured two dimensionally. In this study, we have found 50° of lordosis when standing without load and 40° after 45 minutes of standing with ~25% BW. Such findings are also in agreement with our previous study, for which LS lordosis was 52° when standing without load and 40° after standing for 45 minutes with 50kg of load (Rodriguez-Soto et al. 2013). Interestingly, whole LS lordosis values reported by Neuschwander *et al.* (Neuschwander et al. 2010), Rodríguez-Soto *et al.* (Rodriguez-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

(Rodriguez-Soto et al. 2013) and those of the present study, we identified that the superior LS had a larger increase in lordosis when carrying 50kg of load than when carrying 26kg. Similarly, the reduction in inferior LS lordosis was larger when carrying the heavier load; resulting in a similar value in whole LS lordosis, but with different contributions from each lumbar level.

Local LS posture measurements indicate that the overall reduction in LS lordosis is primarily driven by the changes that occurred at the L4–L5 and L5–S1 levels. These data also suggest that the LS experiences two opposing motions under load-carrying conditions. 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 50kg of load. However, in that evaluation, Marines wore body armor as part of their total load, while in the present study they did not. This is a limitation to comparing relative changes of the LS since the body armor may (or may not) have affected how the LS changes with load. Future work is being conducted to elucidate the effects of body armor on LS posture both with and without load. Additionally, the location of transition levels during load carriage might be associated with the location of the lumbar lordosis apex of each person when standing unloaded. The variation of the apex location ranges from the base of L3 to the middle region of L5 depending on the pelvic and lumbar sagittal alignment of each person (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. All IVDs of Marines with complete and useful data sets were graded using the Pfirrmann scoring system for IVD degeneration. The incidence of IVD degeneration (at least one degenerated IVD) among these Marines was 47.5%, while the incidence of degeneration at the L5–S1 level was 40.7%. Analysis of a larger data set is needed to examine how the combination of multiple degeneration scores through lumbar levels in a single individual can predict LS postural load carriage behavior. Additionally, the fact that most significant postural changes and higher incidence of IVD degeneration occurred at the inferior LS may be related to the greater forces acting on these levels through the LS (Alexander et al. 2007, Pal and Routal 1987). It has been previously suggested that in the presence of IVD degeneration at inferior levels a compensatory mechanism of increased lordosis occurs at superior lumbar levels (Lee et al. 2014, Rodriguez-Soto et al. 2013). However, in the present study, we did not find evidence of this phenomenon.

We found that Marines with degeneration at the L5-S1 level demonstrated larger sacral postural perturbations in response to load as well as a trend of reduced change in lumbar lordosis. Absolute values of SS and LS lordosis of the L5-S1 degeneration group suggest that during load carriage, two postural differences exist compared to the non-degenerated group: 1) sacral 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). However, in the present study we did not find any indication of these differences while standing without external load.

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 fetal load increases (Whitcome et al 2007). Similar to the postural adaptations shown in the present study, it has been hypothesized that these adjustments allow realigning the position of the center of mass in order to maintain balance. Interestingly, in both cases (soldiers posteriorly carrying load and pregnant women anteriorly carrying fetal load) the LS curvature is at extreme standing flexion/extension, during which increased incidence of back pain has been reported (Ostgaard et al. 1993, Dumas et al 1995 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 6-8deg 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 ($1.43\times1.43\times4.5$ mm^3) used in this study (Rodriguez-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 22.6kg load in a standard military load carriage system without wearing body armor, there is an observable compensatory forward lean and an overall reduction in the LS lordosis. Locally, L1–L2 becomes more lordotic, L2–L3 does not change, and L3–L4, L4–L5, and L5–S1 become more kyphotic. Moreover, the anterior and central IVD regions of inferior lumbar levels experience compression, while the posterior disc region becomes distracted, leading to postural changes after standing for 45 min with load. The contribution of each intervertebral level is reflected in lumbar spine flexion and reduced lordosis during load-carrying tasks. Additionally, training did not induce further progression of IVD degeneration in any participant of this study. However, Marines with degenerated IVDs at L5–S1 exhibited a larger sacral postural perturbations and smaller lumbar lordosis changes in response to load. These data suggest that LS postural adaptations to load may not be regulated by physical conditioning as much as they are inherent strategies to manage the overall load over the base of support. However, this concept needs to be tested explicitly.

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Tables

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	

Supplemental Table 1— Values for all angular posture measurements in the unloaded and loaded conditions by visit.

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

Figure Captions

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

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

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

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

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

Supplement Fig. 1—Lumbar-level dependent lordosis measurements. Intervertebral disc (IVD) angles (A–E), change in IVD angle in response to load (F–J), and change in IVD angle in response to load per visit for subjects with and without degeneration at the L5–S1 level (K–O) per lumbar level. Horizontal bars represent statistical difference ($p<0.05$).

Supplement Fig. 2—Lumbar-level dependent regional intervertebral distances. Anterior (A–E), central (F–J), and (K–O) intervertebral disc (IVD) distances per task and visit. Horizontal bars represent statistical difference ($p<0.05$).