DAVID B. BERRY, PhD[1] • **BAHAR SHAHIDI,** PT, PhD[2,3] • **ANA E. RODRÍGUEZ-SOTO,** PhD[1]
JAN M. HUGHES-AUSTIN, PT, PhD[3] • **KAREN R. KELLY,** PhD[4] • **SAMUEL R. WARD,** PT, PhD[1-3]

Lumbar Muscle Structure Predicts Operational Postures in Active-Duty Marines

The muscles of the lumbar spine are crucial for stabilizing and supporting the upper trunk, especially during dynamic loading conditions. A muscle's 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.[27] However, it is difficult to precisely measure PCSA in vivo, as it includes 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.[7,9] However, muscle is a heterogeneous tissue, also consisting of fat and collagenous tissues, which can confound measures of muscle volume.

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

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

METHODS: Forty-three Marines were scanned in this cross-sectional study, using an upright magnetic resonance imaging scanner while standing without load and standing, sitting, and prone on elbows with body armor. Cobb, horizontal, and sacral angles were measured. Marines were then scanned while unloaded in supine using a supine magnetic resonance imaging 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 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 fractional anisotropy of the erector spinae predicted decreased lordosis, lumbosacral extension, and anterior pelvic tilt.

CONCLUSION: Fractional anisotropy is inversely related with muscle fiber size, which is associated with the isometric force-generating capacity of a muscle fiber. This suggests that stronger erector spinae muscles predict decreased lordosis, lumbosacral extension, and anterior pelvic tilt in a highly trained population. *J Orthop Sports Phys Ther 2018;48(8):613-621. Epub 17 May 2018. doi:10.2519/jospt.2018.7865*

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

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.[17,21,22] 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.[5,8,12,35]

With injury and age, atrophy of muscle fibers and replacement of muscle tis-

[1]Department of Bioengineering, University of California, San Diego, La Jolla, CA. [2]Department of Radiology, University of California, San Diego, La Jolla, CA. [3]Department of Orthopaedic Surgery, University of California, San Diego, La Jolla, CA. [4]Department of Warfighter Performance, US Naval Health Research Center, San Diego, CA. The University of California, San Diego (UCSD) and US Naval Health Research Center (NHRC) Institutional Review Boards approved this study (UCSD protocol 110483; NHRC.2013.0023). The opinions or assertions contained herein are the private views of the authors and are not to be construed as official or reflecting the views of the US Navy, US Army, US Air Force, the US Departments of Defense and Veterans Affairs, or the US Government. Any references to commercial organizations and trade names in this article do not constitute an official US Army endorsement or approval of the products or services of these organizations. Dr Kelly is a military service member. This work was prepared as part of her official duties. Title 17, USC, §105 provides that "Copyright protection under this title is not available for any work of the United States Government." Title 17, USC, §101 defines a US Government work as work prepared by a military service member or employee of the US Government as part of that person's official duties. The US Army Medical Research Acquisition Activity supported this manuscript through award number W81XWH-13-2-0043 (work unit number 1305). This work was also supported by US Department of Defense grant PR120576 (to Drs Ward and Kelly). Dr Ward is also supported by grants from the National Institutes of Health (R01AR070830, R01HD073180, R01HD088437). The authors certify that they have no affiliations with or financial involvement in any organization or entity with a direct financial interest in the subject matter or materials discussed in the article. Address correspondence to Dr Samuel R. Ward, University of California, San Diego, 9500 Gilman Drive (0863), La Jolla, CA 92093. E-mail: s1ward@ucsd.edu

sue with adipose and fibrotic tissue are typically observed compared to healthy muscle, further decreasing the overall volume of functional contractile tissue in the muscle.[7,36] As pathogenic (diseased) muscle becomes atrophied and 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.[20,26,39,48] The multifidus muscle is considered to be 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 pathogenic changes associated with injury, low back pain (LBP), or age.[46]

Changes to the orientation and position of bony structures of the spinal column are often observed simultaneously with these changes in muscle composition.[19,38,40] With age, gross changes in spinal posture, such as decreased lumbar lordosis, increased lumbar flexion, and increased pelvic tilt, are typically observed.[13,14,18,42] Decreased segmental range of motion has also been measured at vertebral levels with intervertebral disc (IVD) degeneration,[13,16] which is defined as decreased hydration of the nucleus pulposus with accompanying disc height loss.[25] 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, external stimuli, such as carrying load or whole-body position, may affect posture.[3,29-31] Military members are highly active and often required to carry heavy loads in unusual positions. Studies investigating how Marines adapt to load carriage suggest that they routinely operate under conditions that put them at risk for developing lumbar musculoskeletal injury and that they exhibit higher rates of LBP than civilians.[33,34] 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.[15,28] A noninvasive tool that can 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 individual's risk of lumbar spine injury.

The purpose of this study was to investigate the predictive capacity of muscle structure, IVD health, and anthropometric measures on lumbar spine posture in active-duty Marines. We hypothesized that multifidus muscle volume would predict lumbar posture in different positions, because the multifidus provides intersegmental lumbar support and muscle volume is related to muscle strength.

METHODS

The University of California, San Diego and US Naval Health 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 years of age, and healthy enough to perform their assigned duty. Marines were excluded from this study if they had undergone lumbar spine surgery or had the possibility of shrapnel in their bodies. Marines were not recruited based on LBP status or history. All Marines underwent standard magnetic resonance imaging (MRI) safety screening prior to scanning. All scans were performed early in the morning, between 4 am and 9 am.

Upright MRI

Marines were scanned using an upright 0.6-T 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 and S1 levels while standing. The band was secured to hold the coil in place without altering the volunteer's natural position. A 3-plane localizer (repetition time [TR], 1254 milliseconds; echo time [TE], 100 milliseconds; field of view (FoV), 34 cm; matrix, 256 × 256; in-plane resolution, 1.33 × 1.33 mm; thickness, 9 mm; number of excitations, 1; time, 0:17) and sagittal T2-weighted images (TR, 1974 milliseconds; TE, 160 milliseconds; FoV, 35 cm; matrix, 224 × 224; in-plane resolution, 1.56 × 1.56 mm; thickness, 3 mm; gap, 0 mm; number of excitations, 1; time, 2:12) were acquired.[4]

Upright MRI: Load Carriage and Position Tasks

Marines were scanned in the following positions: standing without load, standing with body armor, 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 selected positions were static positions that Marines are often required to maintain for extended periods, depending on military occupational specialty, and are often reported as provoking LBP.[3] The load magnitude of 11.3 kg was chosen based on the use of body armor, which is the minimum protective equipment Marines are required to wear 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.[31]

Upright MRI: Postural Measurements

Postural measurements were generated from upright MRI images in each position, using a previously validated algorithm.[3] 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).[32] The locations of the seed points were imported into MATLAB (The MathWorks, Inc, Natick, MA) and used to define an end plate–based joint coordinate system applied to the superior and inferior end plate 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 values for global measurements were measured previously and are 0.28°, 0.95°, and 0.95°, respectively.[4,31] Global measurements between the standing unloaded and the standing loaded (delta load) positions, and between the sitting loaded and prone on elbows loaded (delta position) positions, were also calculated to determine lumbar kinematics in response to load and dynamic movement, respectively.

Supine MRI

Magnetic resonance images of the lumbar spine (L1-S1) were acquired using a 3-T MRI scanner (Discovery MR750; GE Healthcare, Waukesha, WI) and spine array coil. The imaging protocol consisted of (1) an anatomical scan, (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, to mitigate motion and breathing artifacts. The anatomical scan was an axial, fast spoiled-gradient echo with the following scanning parameters: TR, 5 milliseconds; TE, 2.3 milliseconds; flip angle, 20°; FoV, 32 cm; acquisition matrix, 512 × 512; pixel size, 0.625 × 0.625 mm^2; slice thickness, 1 mm; no gap; number of averages, 3. Fat-water separation images were acquired utilizing a 3-point iterative decomposition of water and fat, with echo asymmetry and a least-squares estimation sequence in the sagittal plane (TR, 1974 milliseconds; TE, 160 milliseconds; flip angle, 20°; FoV, 25.6 cm; 176 slices; acquisition matrix, 256 × 256; voxel size, 1 × 1 × 1 mm^3; no gap; number of averages, 1). Scanning parameters of the axial DTI sequence were as follows: TR, 10 seconds; TE, 46 milliseconds; FoV, 19.2 cm; 82 slices; acquisition matrix, 128 × 128; pixel size, 1.5 × 1.5 mm^2; slice thickness, 3 mm; no gap; B value, 400 mm^2/s; 45 diffusion directions. Last, multispin-echo data (8 echoes; TE, 8.6 to 68.8 milliseconds; TR, 800 milliseconds; FoV, 16 cm; 5 slices; acquisition matrix, 256 × 256; voxel size, 0.625 × 0.625 × 5 mm^3; no gap; number of averages, 1) were 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 the 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 yielded 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 MRI signal. These data were then used to quantify the fat fraction (FF) of the multifidus and erector spinae group with the following relationship: $FF = S_F/(S_W + S_F)$.

The diffusion tensor was fitted using Analysis of Functional NeuroImages software (National Institutes of Health, Bethesda, MD) and function 3dDWItoDT.[6] Mean diffusivity, 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

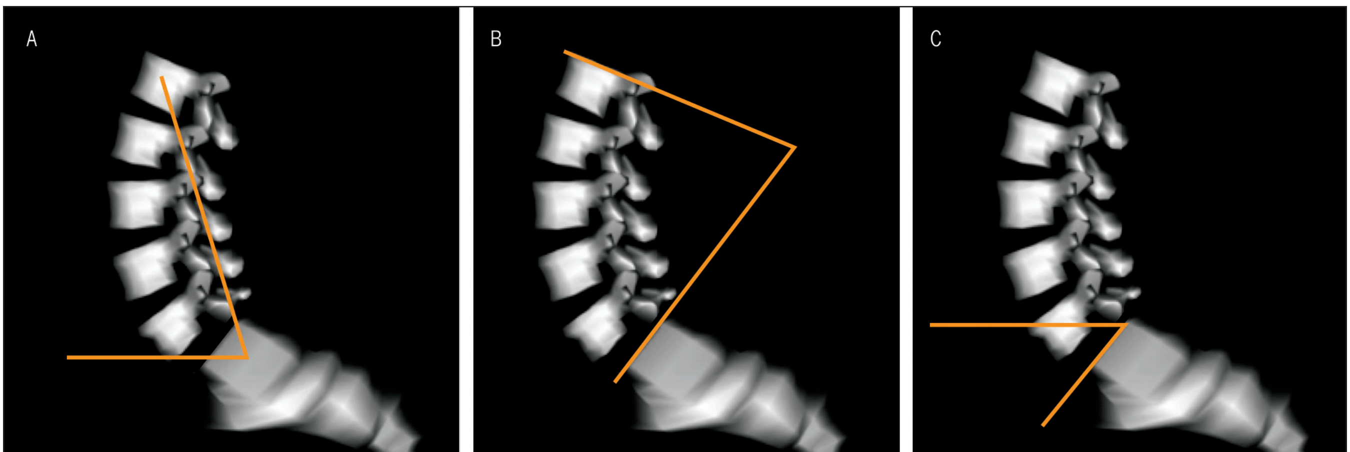

FIGURE 1. Schematic depicting lumbar spine postural measurements on a 3-dimensional model of the lumbar spine. Measurements include (A) angle with respect to the horizontal to assess lumbar flexion/extension, (B) sagittal Cobb angle to measure lumbar lordosis, and (C) sacral slope to assess rotation of the pelvis.

size.[5,8,12,35] Mean diffusivity describes the average restricted diffusion coefficient of $\lambda_{1\text{-}3}$ and is normally between 1×10^{-3} mm^2/s and 2×10^{-3} mm^2/s.[24] Fractional anisotropy 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). The eigenvalues ($\lambda_{1\text{-}3}$) 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 multiecho data to a monoexponential decay: $S_i = S_o e^{-t/T2}$.

Intervertebral disc health is often assessed by qualitatively assessing disc hydration from T2-weighted MRI scans. 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 IVD degeneration.[41]

Statistical Analysis

Dependent variables were global postural measurements (angle with respect to the horizontal, sagittal Cobb angle, and sacral angle) for all positions (standing unloaded and standing, sitting, and prone on elbows with load) and the change in load and flexion/extension positions (delta load, delta position). To assess variance, a 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, mean diffusivity, and $\lambda_{1\text{-}3}$), IVD health (T2 relaxation of each disc), and anthropometric (age, weight, height, and body mass index [BMI][43]) 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, the Pearson correlation coefficient was used to remove collinear variables (r>0.80). For collinear variables, the variable with the smallest eigenvector value was removed to avoid redundancy of variance across variables. Collinearity was also verified at this point by the variance inflation factor; any variable that had a variance inflation factor greater than 10 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 by 3 postural measurements). Statistical analyses were performed using SPSS Version 20.0 (IBM Corporation, Armonk, NY).

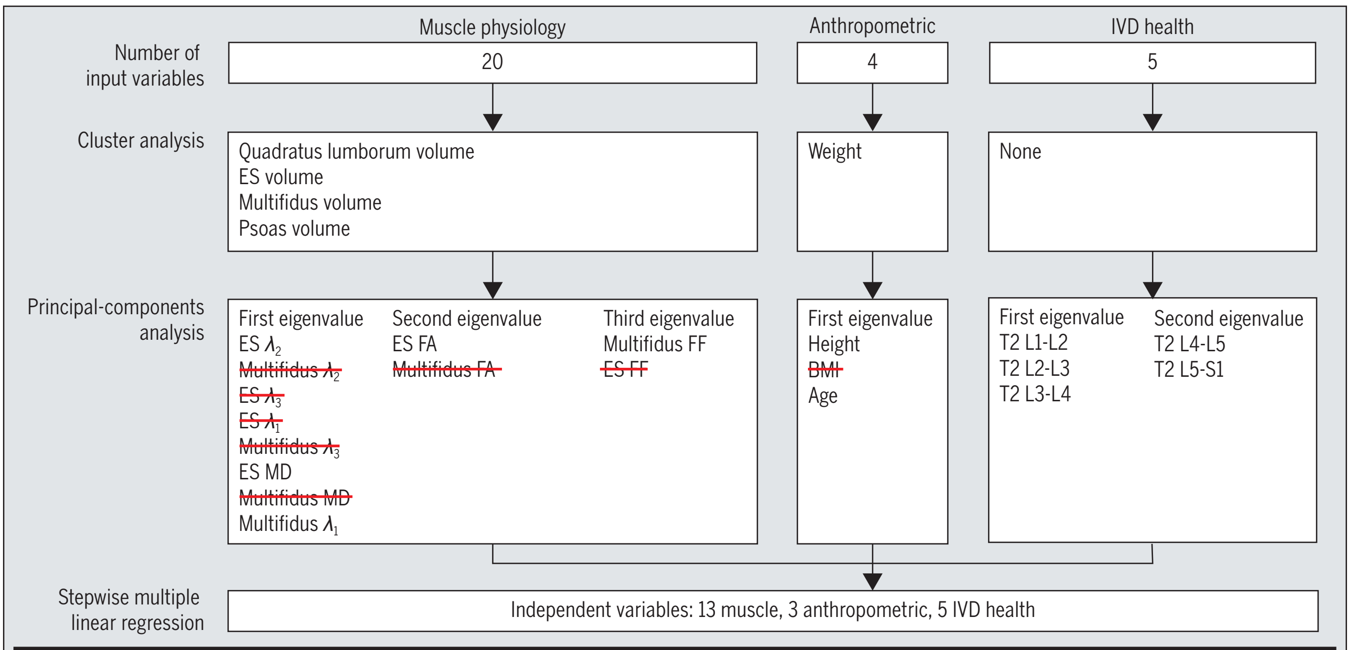

FIGURE 2. Schematic depicting the reduction of collinear independent variables for input into the stepwise multiple regression model. Initially, models were sorted into measures of muscle physiology, anthropometric measures, and IVD health. Cluster analysis was used to identify similar measures. For similar variables, principal-components analysis was used to separate like variables into groups (components). Within each component, Pearson correlations were used to identify collinear variables. If 2 variables were collinear (r>0.80 or variance inflation factor greater than 10), then the variable with the weaker contribution to the eigenvector was removed (crossed out). Abbreviations: BMI, body mass index; ES, erector spinae; FA, fractional anisotropy; FF, fat fraction; IVD, intervertebral disc; λ, eigenvalue; MD, mean diffusivity.

RESULTS

Volunteer Demographics

Forty-three male Marines (mean ± SD age, 26.8 ± 6.4 years; 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 data sets of 10 subjects were deemed unusable due to breathing or motion artifact. Therefore, 31 Marines were included in this analysis (mean ± SD age, 27.3 ± 6.9 years; 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.

Coefficients of variation were relatively low for dependent and independent variables (range, 0.04-10.61; median, 0.16) (**APPENDIX**, available at www.jospt.org). 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 the multifidus or erector spinae, erector spinae FF, and BMI. Surprisingly, 9 of 18 dependent variables were found from the stepwise multiple linear regressions to have a significant predictor. 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 the erector spinae was predictive of increased lumbar lordosis, lumbosacral extension, 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.

TABLE — Results From Stepwise Multiple Linear Regression

Dependent Variable	Significant Independent Variable	β*	R^2	P Value
Cobb angle				
Standing unloaded	None			
Standing loaded	ES FA	0.453	0.205	.02
Sitting loaded	None			
Prone on elbows loaded	Weight	0.468	0.219	.016
Delta load†	None			
Delta position‡	None			
Angle with respect to horizontal				
Standing unloaded	T2 L4-L5	–0.439	0.192	.025
Standing loaded	ES FA	0.514	0.264	.007
Sitting loaded	None			
Prone on elbows loaded	ES FA	–0.480	0.23	.013
Delta load†	None			
Delta position‡	ES FA	0.455	0.207	.02
Sacral angle				
Standing unloaded	ES FA	0.442	0.195	.024
Standing loaded	ES FA	0.587	0.345	.002
Sitting loaded	None			
Prone on elbows loaded	ES FA	0.562	0.316	.003
Delta load†	None			
Delta position‡	None			

Abbreviations: ES, erector spinae; FA, fractional anisotropy.
**Standardized coefficient.*
†Standing unloaded to standing loaded.
‡Sitting loaded to prone on elbows 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. Fractional anisotropy 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.[10,19] Together, the ability of FA to predict postural behavior in several positions and the absence of association between muscle volume and lumbar spine posture suggest that muscle microstructure, but not quantity—both measures associated with force-generating capacity of muscle—is an important predictor of lumbar spine posture.

Diffusion tensor imaging is an MRI technique that measures the restricted diffusion of water in tissues with anisotropic microstructure.[1] 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.[44,45] While it has been shown that FA and fiber area are inversely related,[2,5,8,12,35] it is important to note that the exact relationship has not been validated. However, it is well established that muscle fiber area and isometric force are directly related.[17,21,22] Therefore, it appears that there is likely an inverse relationship between FA and isometric force-generating capacity of muscle. As such, it is inferred that when FA increases, the force-generating capacity of a muscle decreases (ie, the muscle is weaker). For example, if the multifidus muscles in 2 Marines were imaged using DTI and 1 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) the erector spinae, not the multifidus, and (2) muscle microstructure, not volume, were found to be significant predictors of lumbar posture. First, FA of the multifidus and FA of the erector spinae were found to be collinear, with FA of the erector spinae being a stronger descriptor of the eigenvector from the PCA. Therefore, the multifidus was not included in the final statistical model. To verify that FA of the multifidus was not removed from the model because it had less variability than FA of the erector spinae, a coefficient of variation was calculated for both variables. Fractional anisotropy of the erector spinae had less variability relative to the mean than did FA of the multifidus (0.07 versus 0.08), further supporting the latter as a stronger descriptor of the eigenvector. While there is a small difference in variability of these measures, the variability values are both 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,[46,47] the erector spinae may play a role in determining gross lumbar posture.

Second, while muscle volume is proportional to muscle strength,[17,27] 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 that 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 noncontractile 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 in 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, the authors[4] reported no significant difference in lumbosacral extension in Marines when categorized by degeneration at L5-S1 (Pfirrmann grade greater than 2). 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 that health of the L4-L5 IVD 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 highlights 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 that increased lumbar lordosis might be found in individuals with increased BMI[11,23]; however, other studies have shown no difference.[49] 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 in BMI in this relatively homogeneous population. If a more representative cross-section of the population were used, then these findings may have been different.

In this study, the researchers made several attempts to decrease the complexity of the model to decrease the amount of type I error that can be associated with making multiple comparisons. First, this study does not include individual vertebral-level measures of muscle structure or lumbar posture. Second, the authors removed collinear variables with clustering and PCA to minimize the number of independent variables representing similar constructs that were entered into the model. Third, this study 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 that it was the least conservative regression approach. Therefore, the authors 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 one third of the Marines who were included in this study reported LBP. It is important to note that no Marines had an episode of LBP 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

LBP at the time of data collection.[3] 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 had relatively normal muscle, with no underlying pathology observable. In patients with pathology or age-related atrophic changes in muscle, the volume or FF of muscle may be more important in predicting lumbar posture. Therefore, the results of this study may only extend to a highly active population. Second, the positions measured in this study place relatively small challenges on the muscles of the lumbar spine. A future direction of this research is to investigate whether muscle microstructure can predict posture, given the heavy loading conditions under which Marines routinely operate.

Finally, the model used in the present study incorporated 21 variables, with only 31 full data sets to include. This was a retrospective analysis of 2 studies investigating (1) the effect of operationally relevant positions on lumbar posture[3] and (2) normative paraspinal muscle composition in active-duty Marines. It was determined that 43 participants were needed to provide adequate power to these studies. However, to mitigate type I error associated with multiple comparisons, the authors used the most conservative statistical approach. While more participants may provide an increase in the amount of variance explained by the model, this study still reached significance with 31 complete data sets.

CONCLUSION

THE AUTHORS BELIEVE THAT 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, because many clinicians believe that short-term postural positions are more related to motor control than to strength or end organ–dependent behavior.

This study 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 when loading the lumbar spine.[37] 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 of this study support the idea that muscle strengthening/exercise may influence posture, although this cause-and-effect relationship needs to be substantiated in prospective clinical research. As this relationship was found in a healthy population with relatively little variance in muscle quality, it is likely that these relationships may be stronger in patients with LBP or injury. Understanding the influence of microstructural features of muscle on posture may allow clinicians to prognostically categorize patients into groups that may respond better to exercise-based treatments. Future studies should take a more controlled approach to determine whether targeted exercise of the erector spinae muscles increases muscle quality (measured with DTI) and can elicit a postural response. ◉

KEY POINTS

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

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

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

ACKNOWLEDGMENTS: *The authors thank the Marines from the 1st and 5th Regiments who supported this effort.*

REFERENCES

1. Basser PJ, Mattiello J, LeBihan D. MR diffusion *tensor* spectroscopy and imaging. *Biophys J*. 1994;66:259-267. https://doi.org/10.1016/S0006-3495(94)80775-1
2. Berry DB, Regner B, Galinsky V, Ward SR, Frank LR. Relationships between tissue microstructure and the diffusion tensor in simulated skeletal muscle. *Magn Reson Med*. 2018;80:317-329. https://doi.org/10.1002/mrm.26993
3. Berry DB, Rodríguez-Soto AE, Su J, et al. Lumbar spine postures in Marines during simulated operational positions. *J Orthop Res*. 2017;35:2145-2153. https://doi.org/10.1002/jor.23510
4. Berry DB, Rodríguez-Soto AE, Tokunaga JR, Gombatto SP, Ward SR. An endplate-based joint coordinate system for measuring kinematics in normal and abnormally-shaped lumbar vertebrae. *J Appl Biomech*. 2015;31:499-503. https://doi.org/10.1123/jab.2015-0008
5. Berry DB, You S, Warner J, Frank LR, Chen S, Ward SR. A 3D tissue-printing approach for validation of diffusion tensor imaging in skeletal muscle. *Tissue Eng Part A*. 2017;23:980-988. https://doi.org/10.1089/ten.tea.2016.0438
6. Cox RW. AFNI: software for analysis and visualization of functional magnetic resonance neuroimages. *Comput Biomed Res*. 1996;29:162-173. https://doi.org/10.1006/cbmr.1996.0014
7. Demoulin C, Crielaard JM, Vanderthommen M. Spinal muscle evaluation in healthy individuals and low-back-pain patients: a literature review.

Joint Bone Spine. 2007;74:9-13. https://doi.org/10.1016/j.jbspin.2006.02.013

8. Deux JF, Malzy P, Paragios N, et al. Assessment of calf muscle contraction by diffusion tensor imaging. *Eur Radiol*. 2008;18:2303-2310. https://doi.org/10.1007/s00330-008-1012-z
9. Fortin M, Macedo LG. Multifidus and paraspinal muscle group cross-sectional areas of patients with low back pain and control patients: a systematic review with a focus on blinding. *Phys Ther*. 2013;93:873-888. https://doi.org/10.2522/ptj.20120457
10. Fukunaga T, Miyatani M, Tachi M, Kouzaki M, Kawakami Y, Kanehisa H. Muscle volume is a major determinant of joint torque in humans. *Acta Physiol Scand*. 2001;172:249-255. https://doi.org/10.1046/j.1365-201x.2001.00867.x
11. Guo JM, Zhang GQ, Alimujiang. [Effect of BMI and WHR on lumbar lordosis and sacrum slant angle in middle and elderly women]. *Zhongguo Gu Shang*. 2008;21:30-31.
12. Hatakenaka M, Yabuuchi H, Sunami S, et al. Joint position affects muscle proton diffusion: evaluation with a 3-T MR system. *AJR Am J Roentgenol*. 2010;194:W208-W211. https://doi.org/10.2214/AJR.09.3167
13. 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;36:893-898. https://doi.org/10.1097/BRS.0b013e3181f4d212
14. Kim YB, Kim YJ, Ahn YJ, et al. A comparative analysis of sagittal spinopelvic alignment between young and old men without localized disc degeneration. *Eur Spine J*. 2014;23:1400-1406. https://doi.org/10.1007/s00586-014-3236-8
15. Knapik JJ, Reynolds KL, Harman E. Soldier load carriage: historical, physiological, biomechanical, and medical aspects. *Mil Med*. 2004;169:45-56. https://doi.org/10.7205/MILMED.169.1.45
16. Lao L, Daubs MD, Scott TP, et al. Effect of disc degeneration on lumbar segmental mobility analyzed by kinetic magnetic resonance imaging. *Spine (Phila Pa 1976)*. 2015;40:316-322. https://doi.org/10.1097/BRS.0000000000000738
17. Lieber RL. *Skeletal Muscle Structure, Function, and Plasticity: The Physiological Basis of Rehabilitation*. 2nd ed. Philadelphia, PA: Wolters Kluwer/Lippincott Williams & Wilkins; 2002.
18. Mac-Thiong JM, Roussouly P, Berthonnaud É, Guigui P. Age- and sex-related variations in sagittal sacropelvic morphology and balance in asymptomatic adults. *Eur Spine J*. 2011;20 suppl 5:572-577. https://doi.org/10.1007/s00586-011-1923-2
19. Meakin JR, Fulford J, Seymour R, Welsman JR, Knapp KM. The relationship between sagittal curvature and extensor muscle volume in the lumbar spine. *J Anat*. 2013;222:608-614. https://doi.org/10.1111/joa.12047
20. Meyer GA, Lieber RL. Elucidation of extracellular matrix mechanics from muscle fibers and fiber bundles. *J Biomech*. 2011;44:771-773. https://doi.org/10.1016/j.jbiomech.2010.10.044
21. Miller AE, MacDougall JD, Tarnopolsky MA, Sale DG. Gender differences in strength and muscle fiber characteristics. *Eur J Appl Physiol Occup Physiol*. 1993;66:254-262. https://doi.org/10.1007/BF00235103
22. Minamoto VB, Hulst JB, Lim M, et al. Increased efficacy and decreased systemic-effects of botulinum toxin A injection after active or passive muscle manipulation. *Dev Med Child Neurol*. 2007;49:907-914. https://doi.org/10.1111/j.1469-8749.2007.00907.x
23. Murrie VL, Dixon AK, Hollingworth W, Wilson H, Doyle TA. Lumbar lordosis: study of patients with and without low back pain. *Clin Anat*. 2003;16:144-147. https://doi.org/10.1002/ca.10114
24. Oudeman J, Nederveen AJ, Strijkers GJ, Maas M, Luijten PR, Froeling M. Techniques and applications of skeletal muscle diffusion tensor imaging: a review. *J Magn Reson Imaging*. 2016;43:773-788. https://doi.org/10.1002/jmri.25016
25. Pfirrmann CW, Metzdorf A, Zanetti M, Hodler J, Boos N. Magnetic resonance classification of lumbar intervertebral disc degeneration. *Spine (Phila Pa 1976)*. 2001;26:1873-1878. https://doi.org/10.1097/00007632-200109010-00011
26. Pontén E, Fridén J, Thornell LE, Lieber RL. Spastic wrist flexors are more severely affected than wrist extensors in children with cerebral palsy. *Dev Med Child Neurol*. 2005;47:384-389. https://doi.org/10.1111/j.1469-8749.2005.tb01158.x
27. Powell PL, Roy RR, Kanim P, Bello MA, Edgerton VR. Predictability of skeletal muscle tension from architectural determinations in guinea pig hindlimbs. *J Appl Physiol Respir Environ Exerc Physiol*. 1984;57:1715-1721. https://doi.org/10.1152/jappl.1984.57.6.1715
28. Reynolds KL, Kaszuba J, Mello RP, Patton JF. Prolonged Treadmill Load Carriage: Acute Injuries and Changes in Foot Anthropometry. Natick, MA: US Army Research Institute of Environmental Medicine; 1990.
29. Rodriguez-Soto AE, Berry DB, Jaworski R, et al. The effect of training on lumbar spine posture and intervertebral disc degeneration in active-duty Marines. *Ergonomics*. 2017;60:1055-1063. https://doi.org/10.1080/00140139.2016.1252858
30. Rodríguez-Soto AE, Berry DB, Palombo L, Valaik E, Kelly KR, Ward SR. Effect of load magnitude and distribution on lumbar spine posture in active-duty Marines. *Spine (Phila Pa 1976)*. 2017;42:345-351. https://doi.org/10.1097/BRS.0000000000001742
31. Rodríguez-Soto AE, Jaworski R, Jensen A, et al. Effect of load carriage on lumbar spine kinematics. *Spine (Phila Pa 1976)*. 2013;38:E783-E791. https://doi.org/10.1097/BRS.0b013e3182913e9f
32. Rosset A, Spadola L, Ratib O. OsiriX: an open-source software for navigating in multidimensional DICOM images. *J Digit Imaging*. 2004;17:205-216. https://doi.org/10.1007/s10278-004-1014-6
33. Roy TC, Knapik JJ, Ritland BM, Murphy N, Sharp MA. Risk factors for musculoskeletal injuries for soldiers deployed to Afghanistan. *Aviat Space Environ Med*. 2012;83:1060-1066. https://doi.org/10.3357/ASEM.3341.2012
34. Roy TC, Lopez HP, Piva SR. Loads worn by soldiers predict episodes of low back pain during deployment to Afghanistan. *Spine (Phila Pa 1976)*. 2013;38:1310-1317. https://doi.org/10.1097/BRS.0b013e31829265c4
35. Schwenzer NF, Steidle G, Martirosian P, et al. Diffusion tensor imaging of the human calf muscle: distinct changes in fractional anisotropy and mean diffusion due to passive muscle shortening and stretching. *NMR Biomed*. 2009;22:1047-1053. https://doi.org/10.1002/nbm.1409
36. Shahidi B, Parra CL, Berry DB, et al. Contribution of lumbar spine pathology and age to paraspinal muscle size and fatty infiltration. *Spine (Phila Pa 1976)*. 2017;42:616-623. https://doi.org/10.1097/BRS.0000000000001848
37. Shirazi-Adl A, El-Rich M, Pop DG, Parnianpour M. Spinal muscle forces, internal loads and stability in standing under various postures and loads—application of kinematics-based algorithm. *Eur Spine J*. 2005;14:381-392. https://doi.org/10.1007/s00586-004-0779-0
38. Sinaki M, Itoi E, Rogers JW, Bergstralh EJ, Wahner HW. Correlation of back extensor strength with thoracic kyphosis and lumbar lordosis in estrogen-deficient women. *Am J Phys Med Rehabil*. 1996;75:370-374. https://doi.org/10.1097/00002060-199609000-00013
39. Smith LR, Lee KS, Ward SR, Chambers HG, Lieber RL. Hamstring contractures in children with spastic cerebral palsy result from a stiffer extracellular matrix and increased *in vivo* sarcomere length. *J Physiol*. 2011;589:2625-2639. https://doi.org/10.1113/jphysiol.2010.203364
40. Sparrey CJ, Bailey JF, Safaee M, et al. Etiology of lumbar lordosis and its pathophysiology: a review of the evolution of lumbar lordosis, and the mechanics and biology of lumbar degeneration. *Neurosurg Focus*. 2014;36:E1. https://doi.org/10.3171/2014.1.FOCUS13551
41. Stelzeneder D, Welsch GH, Kovács BK, et al. Quantitative T2 evaluation at 3.0 T compared to morphological grading of the lumbar intervertebral disc: a standardized evaluation approach in patients with low back pain. *Eur J Radiol*. 2012;81:324-330. https://doi.org/10.1016/j.ejrad.2010.12.093
42. Sullivan MS, Dickinson CE, Troup JD. The influence of age and gender on lumbar spine sagittal plane range of motion. A study of 1126 healthy subjects. *Spine (Phila Pa 1976)*. 1994;19:682-686. https://doi.org/10.1097/00007632-199403001-00007
43. Takatalo J, Karppinen J, Taimela S, et al. Body mass index is associated with lumbar disc

degeneration in young Finnish males: subsample of Northern Finland birth cohort study 1986. *BMC Musculoskelet Disord*. 2013;14:87. https://doi.org/10.1186/1471-2474-14-87

44. Van Donkelaar CC, Kretzers LJ, Bovendeerd PH, et al. Diffusion tensor imaging in biomechanical studies of skeletal muscle function. *J Anat*. 1999;194 pt 1:79-88. https://doi.org/10.1046/j.1469-7580.1999.19410079.x

45. van Doorn A, Bovendeerd PH, Nicolay K, Drost MR, Janssen JD. Determination of muscle fibre orientation using Diffusion-Weighted MRI. *Eur J Morphol*. 1996;34:5-10.

46. Ward SR, Kim CW, Eng CM, et al. Architectural analysis and intraoperative measurements demonstrate the unique design of the multifidus muscle for lumbar spine stability. *J Bone Joint Surg Am*. 2009;91:176-185. https://doi.org/10.2106/JBJS.G.01311

47. Ward SR, Tomiya A, Regev GJ, et al. Passive mechanical properties of the lumbar multifidus muscle support its role as a stabilizer. *J Biomech*. 2009;42:1384-1389. https://doi.org/10.1016/j.jbiomech.2008.09.042

48. Wokke BH, van den Bergen JC, Versluis MJ, et al. Quantitative MRI and strength measurements in the assessment of muscle quality in Duchenne muscular dystrophy. *Neuromuscul Disord*. 2014;24:409-416. https://doi.org/10.1016/j.nmd.2014.01.015

49. Youdas JW, Hollman JH, Krause DA. The effects of gender, age, and body mass index on standing lumbar curvature in persons without current low back pain. *Physiother Theory Pract*. 2006;22:229-237. https://doi.org/10.1080/09593980600927864

APPENDIX

Coefficient of Variation Calculated for Each Dependent Variable

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

Coefficient of Variation Calculated for Each Independent Variable

Independent Variable	Coefficient of Variation
Muscle measures	
Multifidus	
Volume	0.14
Fat fraction	0.41
Mean diffusivity	0.05
Fractional anisotropy	0.08
Lambda 1	0.04
Lambda 2	0.04
Lambda 3	0.06
Erector spinae	
Volume	0.22
Fat fraction	0.41
Mean diffusivity	0.05
Fractional anisotropy	0.07
Lambda 1	0.04
Lambda 2	0.04
Lambda 3	0.05
Psoas volume	0.13
Quadratus lumborum volume	0.19
IVD measures	
T2	
L1-L2	0.24
L2-L3	0.27
L3-L4	0.29
L4-L5	0.35
L5-S1	0.41
Anthropometric measures	
Age	0.24
Height	0.04
Weight	0.12
Body mass index	0.11

Abbreviation: IVD, intervertebral disc.