

## Comparison of multifidus muscle intramuscular fat by ultrasound echo intensity and fat-water based MR images in individuals with chronic low back pain

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## **Abstract**

**Purpose:** The aim of this observational cross-sectional study was to examine correlations of intramuscular fat content in lumbar multifidus (LM) by comparing muscle echo intensity (EI) and percent fat signal fraction (%FSF) generated from ultrasound (US) and magnetic resonance (MR) images, respectively.

**Methods:** MRI and US images from 25 participants (16 females, 9 males) selected from a cohort of patients with chronic low back pain were (CLBP) used. Images were acquired bilaterally, at the L4 and L5 levels (e.g., 4 sites). EI measurements were acquired by manually tracing the cross-sectional border of LM. Mean EI of three US images per site were analyzed (e.g., raw EI). A correction factor for subcutaneous fat thickness (SFT) was also calculated and applied (e.g., corrected EI). Corresponding fat and water MR images were used to acquire %FSF measurements. Intra-rater reliability was assessed by intraclass coefficients (ICC). Pearson correlations and simple linear regression were used to assess the relationship between %FSF, raw EI and corrected EI measurements.

**Results:** The intra-rater ICCs for all measurements were moderate to excellent. Correlations between %FSF vs. raw EI and corrected EI were moderate to strong ( $0.40 < r < 0.52$ ) and ( $0.40 < r < 0.51$ ), respectively. Moderate correlations between SFT and EI were also identified.

**Conclusion:** US is a low-cost, non-invasive, accessible, and reliable method to examine muscle composition, and presents a promising solution for assessing and monitoring the effect of different treatment options for CLBP in clinical settings

**Keywords:** imaging, subcutaneous fat, correction factor, low back pain.

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## **Introduction**

Chronic low back pain (CLBP) is a significant health problem worldwide affecting an estimated 577 million people in 2017 (Wu et al., 2020). According to the global burden of disease 2010 study, low back pain (LBP) ranked first in terms of disability (Cheung et al., 2020; Hoy et al., 2014), it is the leading cause of years lived with disability (Vos et al., 2012), and has been associated with worse health-related quality of life (Nolet et al., 2015). Despite extensive research, the exact source of pain remains unknown in most people suffering from CLBP. Some studies have suggested excess body fat contributes to CLBP (Walsh et al., 2018). Furthermore, it has been shown that higher body fat percentage is associated with higher proportions of fat within skeletal muscle tissue i.e., intramuscular fat (IMF) (Rahemi et al., 2015), contributing to muscle degeneration and decreased function (Goubert et al., 2017). For example, dysfunction in the lumbar multifidus (LM), a paraspinal muscle identified as having an important role in spinal stability, has been linked to CLBP (Chan et al., 2012; Goubert et al., 2017; Nandlall et al., 2020). More specifically, morphological changes due to increased fat content within the muscle has been associated with decreased function, which can lead to instability and may contribute to LBP conditions (Goubert et al., 2017; Seyedhoseinpoor et al., 2021). Medical imaging modalities are relevant clinical tools that can help deepen our understanding of the relationship between variations in muscle morphology, function, and their connection to CLBP.

Ultrasound (US) imaging is gaining popularity in evaluating muscle quality. This technique operates under the premise that skeletal muscle tissue is composed of both high density, contractile tissue and low density, non-contractile tissue (adipocytes, fibrous tissue) (Stock & Thompson, 2021). Relative proportions within the muscle are quantitatively measured by analyzing echogenicity using a grey scale distribution of pixel intensity within a selected region of interest (ROI) (Almazán-Polo et al., 2020). Lean muscle tissue is hypoechoic (i.e., appears darker) whereas adipose tissue is hyperechoic (i.e., appears whiter). Muscle quality is reflected in the total echo intensity (EI) within a ROI where high values suggest higher adipose and connective tissue deposition within the muscle and low values indicate low adipose deposition suggesting poorer and superior muscle quality, respectively (Pillen & van Alfen, 2011). Previous studies have shown a negative relationship between muscle EI and physical fitness parameters including muscle force output (Belzunce et al., 2021; Cadore et al., 2012; Fukumoto et al., 2012), cardiovascular capacity (Cadore et al., 2012), and mobility (Goodpaster et al., 2008).

While MRI or computed tomography (CT) are the preferred imaging modalities for measuring IMF, these methods are costly, time consuming and not always feasible (Ghadimi & Sapra, 2022; Stecco et al., 2007). US provides relatively low-cost, low-risk, accessible and portable option for evaluating muscle morphology and quality. Previous evidence showed moderate to strong correlations between MRI measured IMF and EI in muscles of the thigh (rectus femoris and biceps femoris) and lower leg (medial gastrocnemius and tibialis anterior) (Young et al., 2015). However, to our knowledge, no study examined this comparison specifically for the LM muscle. While degenerative LM morphological changes are suspected contributors to LBP conditions for many years, challenges related to measuring these changes quantitatively in clinical settings remains a significant barrier in understanding their impact. The development and implementation

of accessible imaging tools such as US in clinical settings are critical to deepen our understanding of the role of the paraspinal musculature in the development and management of CLBP conditions. Given the increasing interest and research into the clinical assessment of muscle quality via US, it is imperative its accuracy is examined and maximized. Therefore, the purpose of this study was to compare LM muscle EI measurements with percent fat signal fraction (%FSF) measurements derived from IDEAL fat-water MR images. We hypothesized that there would be a strong correlation between EI and %FSF LM measurements.

## **Materials and Methods**

### *Study sample*

A sample of 25 subjects (16 female, 9 male) was selected from a cohort of patients participating in a randomized controlled trial (trial registration XXXXXXXXXX) evaluating the effect of two exercise interventions on paraspinal muscle morphology and function. The subjects' baseline MRI and ultrasound imaging assessments were used for the purpose of the current study. Inclusion criteria for the original trial (and current study) included having non-specific CLBP (3 months) with or without leg pain, having a “moderate” or “severe” score on the modified Oswestry Low Back Pain Disability Index (ODI) questionnaire and not engaging in sport or training specifically for the lower back musculature 3 months prior the beginning of the trial. Patients were excluded they were aged below 18 or above 65 years old, had evidence or nerve root compression (or motor sign deficits), had a previous history of spinal surgery or vertebral fractures, had major spine structure abnormalities (e.g., spondylolisthesis, scoliosis  $>10^\circ$ ), were pregnant or had comorbidities preventing them to safely participate in an exercise program. This study was

approved by the [REDACTED]

[REDACTED]. Written consent form was obtained from all subjects prior to any data collection.

### *Ultrasound imaging protocol*

LM assessments were performed using an Aixplorer ultrasound machine (Supersonic Imagine, Aix-en-Provence, France). The B-mode with musculoskeletal abdominal pre-set and a 1-6MHz curvilinear transducer was used. Gain and transducer frequency were adjusted to 60-dB and 5MHz, respectively. The depth was set at 8 cm and was only increased when subjects had greater subcutaneous fat and allow for the entire visualization of the muscle cross-sectional area (CSA). Focus areas were kept consistent for all subjects and no other ultrasound settings were changed during the examination. Subjects were placed in a prone position on a therapy table with a pillow under the abdomen to relax the paraspinal musculature and minimize lumbar lordosis. Prior to imaging, the spinous process of L5 was palpated. The ultrasound transducer was then placed longitudinally to confirm the location of L5. Once the location was confirmed, the transducer was rotated transversally over the L5 spinous process. Transverse images at L4 and L5 levels were obtained bilaterally, to assess the LM muscle size (cross-sectional area). All ultrasound images were acquired by a physical therapist with 5 years of experience (NN) in LM ultrasound imaging. Images were stored and analysed offline using the Horos DICOM viewer software (version 3.3.6). LM muscle EI measurements were obtained using grayscale analysis and tracing the ROI representing the entire CSA of the LM muscle while avoiding surrounding bone and myofascial tissue (Fortin et al., 2021). EI was defined as the average amount of gray within the ROI using the “grayscale histogram function” (e.g. Pixels expressed as a value between 0 = Black and 255 = white (Arts et al., 2010; Fortin et al., 2021)), where higher values signify higher amounts of IMF

and connective tissue. A total of 3 different images per level were analyzed and the average was used in the analysis. Image segmentation was performed by a single novice rater (JC) trained by an experienced rater (MF) with over 10 years of experience.

#### *Correction for subcutaneous fat thickness*

In accordance with a previous study, we examined the effect of subcutaneous fat thickness (SFT) on muscle EI (Young et al., 2015). Young et al. (2015) formulated a correction factor (CF) by comparing EI and SFT after applying various pressures to the transducer at the same site which resulted in small changes in SFT. In the present study, CF was obtained by selecting 6 participants (3 female, 3 male) with 1 participant of each gender within the age range from 20-30, 30-40 and 40-50 years old. Three SFT measurements were taken at each muscle site (e.g., right L4, left L4, right L5 and left L5) by tracing a straight line from the innermost border of epidermis to the myofascial line on the posterior aspect of the muscle. The association between EI and SFT for each participant was examined (**Figure 3**). The average of the three measures was compared to the associated mean EI computed for that site. The average slope and y-intercept of the 6 individuals were calculated. The following equation was used to determine the CF, where  $cf$  = correction factor and  $x$  = subcutaneous fat thickness. As such, the CF represents the addition of EI for each “1 cm” unit increase of SFT:

$$cf = -15.400 (1.0cm) + 67.059 = 51.661 \quad \text{(Equation 1)}$$



As proposed by Young et al. (2015), to reduce the potential influence of SFT on EI, raw EI values were adjusted using the CF and corrected values of EI were obtained by applying the following equation where,  $y_1$  = raw EI,  $x$  = SFT,  $cf$  = correction factor, and  $y_2$  = corrected EI.

$$y_2 = y_1 + (x \times cf) \quad \text{(Equation 2)}$$

### *MRI imaging protocol*

IDEAL (Lava-flex, 2 echo sequence) fat and water images of the entire lumbar spine (L1-L5) were obtained using a 3.0 Tesla GE scanner (Milwaukee, WI, USA). A standard phased-array body coil was used, with 4-mm slice thickness, 180-mm<sup>2</sup> field of view and 512x512 matrix. MR images were analyzed using Horos DICOM viewer software (version 4.0.0). Axial water and fat images at L4-L5 and L5-S1 level (mid-disc) were used to calculate LM % fat signal fraction (%FSF), bilaterally. The ROI representing the LM CSA was traced manually on the fat image and the ROI was then copied on the corresponding water image. Resulting %FSF for the right and left LM muscle at each level was calculated using the following formula: %FSF=(Signal<sub>fat</sub>/[Signal<sub>water</sub>+Signal<sub>Fat</sub>] $\times$ 100). MR image segmentation was performed by a single, novice rater (SM) trained by an experienced rater (MF) with over 10 years of experience.

### *Statistical Analysis*

No *a priori* sample size calculation was performed, as subjects included in the current study were selected from an ongoing clinical trial. The sample size of 25 is in accordance with previous related investigations (Cadore et al., 2012; Young et al., 2015). Data were analyzed using SPSS

version 26.0.0. Data are shown as mean  $\pm$  SD (range). Intraclass correlation coefficients (ICC) using a two-way mixed model and absolute agreement with average measures for the EI measurements (ICC<sub>3,3</sub>) and single measures for the %FSF measurements (ICC<sub>3,1</sub>) were used to assess intra-rater reliability. ICC interpretation was based on Portney and Watkins guidelines where ICC values  $< 0.50$  indicates poor reliability,  $0.50 < \text{ICC} < 0.75$  indicates moderate reliability, from  $0.75 < \text{ICC} < 0.9$  indicates good reliability and  $\text{ICC} > 0.9$  shows excellent reliability. The intra-rater reliability for the derived ultrasound EI measured and MRI %FSF measures were assessed on a random sample of 15 and 10 images, respectively. The correlation between ultrasound-derived EI (raw and corrected) and corresponding MRI %FSF values were assessed using Pearson correlation. Similarly, we also used Pearson correlation to examine the association between SFT and raw EI, as well as the association between muscle EI (raw and corrected) with both age and ODI scores. Strength of correlation was defined using Cohen's conventions where a correlation coefficient ( $r$ ) between 0.1 and 0.2 indicates a small/ weak relationship between variables,  $0.3 < r < 0.5$  and  $0.5 > r$  indicates medium/moderate and large/ strong correlations respectively. Simple linear regression was used to analyze the relationship between MRI %FSF and raw EI, and MRI %FSF and corrected EI.

## **Results**

### *Participants*

Physical characteristics: Age (yrs.), height (cm), weight (kg), BMI (kg/m<sup>2</sup>), and ODI are outlined in **Table 1**.

## US and MRI outcome measures

Outcome measures of US and MR image analysis are presented in Table 2.

### *Ultrasound and MRI intra-rater reliability*

Based on the ICCs and 95% CIs, the intra-rater reliability for the EI measurements was moderate to excellent with ICCs ranging between 0.92-0.98 (**Table 3**). The intra-rater reliability for the MRI %FSF measurements was excellent (ICCs=0.99).

### *Correlation between raw EI and MRI %FSF*

Correlations and related scatterplots illustrating the relationship between %FSF and raw EI measurements are presented in **Table 4** and **Figure 4a) and c) and Figure 5a) and c)**, respectively. A strong positive correlation was found at level L4 left ( $r = 0.52$ ). Moderate correlation was found at levels L4 right ( $r = 0.44$ ) and L5 left ( $r = 0.40$ ). However, the L5 right correlation was not significant ( $r = 0.20$ ,  $p = 0.33$ ). When comparing raw EI and MRI measured intramuscular fat of all sites combined, a moderate correlation was found ( $r = 0.41$ ) (**Figure 6a**).

### *Correlation between corrected EI and MRI %FSF*

The correlations and related scatterplots illustrating the relationship between corrected EI and MRI %FSF are presented in **Table 4** and **Figure 4b) and d) and Figure 5b) and d)**, respectively. After the CF was applied, a strong positive correlation was found at level L5 left ( $r = 0.51$ ). Moderate correlation was found at levels L4 right ( $r = 0.40$ ) and L4 left ( $r = 0.44$ ). However, the L5 right correlation was not significant ( $r = 0.30$   $p = 0.15$ ). When comparing

corrected EI and MRI measured intramuscular fat of all sites combined, a moderate correlation was found ( $r = 0.43$ ) (**Figure 6b**).

#### *Correlation between raw EI and SFT*

Correlation between raw EI and SFT are shown in **Table 5**. Significant moderate correlations were found at L4 right ( $r = .41$ ) and L5 right ( $r = .42$ ).

#### *Correlation between raw and corrected EI with Age and ODI*

Correlations between raw and corrected EI with age and ODI are presented in Table 6. No relationship was found between both raw and corrected EI and age (all  $p > 0.05$ ). A moderate negative correlation was found between raw EI and ODI at level L4 right ( $r = -0.44$ ). Correlation at all remaining sites for EI (raw and corrected) and ODI were not significant.

## **Discussion**

In this study, we examined the correlation between IMF proportions of the LM muscle measured via %FSF derived from fat-water MRI and US EI. Overall, our results showed a moderate correlation between MRI %FSF and US EI across the four sites investigated. This is consistent with previous studies. Young et al. (2015) examined the comparison between IMF of four leg muscle groups and found a moderate correlation between %IMF acquired via T1-weighted MR images and US EI. We are not aware of any previous studies that have assessed the correlation between skeletal muscle EI and derived fat-water MRI images.

High muscle EI has been associated with decreased quality and function of muscle (Belzunce et al. 2021; Cadore et al. 2012; Fukumoto et al. 2012). Increased IMF is associated with metabolic disorders (Shaw et al. 2010), decreased muscle strength and power output, balance deficits in ageing populations (Cadore et al. 2012; Rahemi et al. 2015), as well as decreased mobility and function (Rahemi et al. 2015). Historically, US has proved a valuable tool in providing clinical information in examining muscle quality. However, due the arbitrary units of output, it is very difficult to draw comparisons and conclusions to gold standard methods such as MRI and CT (Khil et al. 2020). This is the primary limiting factor in implementing US as an effective method of examining muscle composition. Given that the properties of muscle morphology may contribute to LBP conditions findings practical ways to measure such changes in clinical settings may led to improvements in specific approaches to treatment.

To minimize the effects of subcutaneous fat on EI a CF was formulated in accordance with previous study by Young et al. (2015). Indeed, increased subcutaneous fat may negatively affect the quality of US images. Storchle et al. (2016) state that using high-frequency US can produce quality imaging up to 0.1mm but greater SFT may require lower frequency to decrease attenuation which has strong direct relationship to frequency. Low frequency US generates lower quality images and decreases image resolution to approximately 0.3mm making border detection of deeper tissue less accurate. More recently, Neto Muller et al. (2021) examined the confounding overestimation of EI due to subcutaneous fat. They report a possible overestimation of EI of over 39 arbitrary units (AU) for every 1 cm increase in subcutaneous fat and claim that results obtained in the absence of this consideration are biased. The application of the CF in our study generated mixed results in terms of the relationship between US corrected EI and %FSF. Corrected EI

improved the correlation between %FSF and US EI at LM levels L5 left and L5 right from  $r = 0.40$  (-0.04, 0.73) to  $r = 0.51$  (0.23, 0.78) and  $r = 0.20$  (-0.21, 0.54) to  $r = 0.30$  (-0.06, 0.61), respectively. Conversely, a corrected EI resulted in a weaker correlation at levels L4 right from  $r = 0.44$  (.02, .75) to  $r = 0.40$  (0.06, 0.67) and L4 left from  $r = 0.52$  (0.07, .81) to  $r = .44$  (.02, .74). Our clinical population and associated high level of intramuscular fat present in our sample may have contributed to the mixed findings, (Pillen and Van Alfen 2011; Ziskin 1993) coupled with possible deficits in the CF model proposed by Young et al. (2015). Nevertheless, our results suggest that simple correction factors may not provide adequate adjustments for the effect of subcutaneous fat on EI LM measurements. Additional methodological studies are needed to further clarify the effect of subcutaneous fat on skeletal muscle EI measurements in different clinical populations.

Our findings corroborate existing literature and demonstrated moderate to excellent intra-rater reliability for image measurement using US and MRI (Fortin et al. 2021; Mansur et al. 2022). Although inter-rater reliability was not assessed for EI measurements, a recent study examining the effect on rater experience and reliability of measure of LM EI measurements and found moderate to excellent interrater reliability between novice and experienced raters (Fortin et al. 2021). Furthermore, studies by Valera-Calero et al. (2021) demonstrate good to excellent intra-rater reliability of EI features of cervical multifidus in healthy and in a clinical populations. (Valera-Calero, Arias-Buría, et al., 2021; Valera-Calero, Fernández-de-las-Peñas, et al., 2022) and acceptable inter-rater reliability in experienced examiners (Valera-Calero, Fernández-de-las-Peñas, et al., 2022). As such, rater reliability does not appear to be a barrier in the application of US imaging for experienced raters in clinical settings as the superior reproducibility of results speaks to its ease of use and remains a leading quality supporting its potential as a clinical

assessment tool. Some studies have demonstrated high reliability using automated segmentation methods calculated to be as high as 95% in accordance with radiologists (Gupta et al. 2014). This falls outside the scope of this paper but it provides an interesting avenue for future studies investigating potential improvement to clinical US use.

Observations from previous studies regarding potential interference at higher %FSF (>15%) prompted a small inquiry into our results as well (Young et al., 2015). US images are generated by recording the transmission, scattering and absorption of sound waves encountering tissue (Pillen & van Alfen, 2011). As the sound waves penetrate deeper tissue, the degree of the above-mentioned processes increases resulting in attenuation of the intensity of the beam. Furthermore, muscle tissue containing higher proportions of intramuscular fat causes a higher degree of reflection of sound waves potentially amplifying beam attenuation (Pillen & van Alfen, 2011; Ziskin, 1993). Young et al. (2015) observed higher variation between EI of participants with %FSF above 15%. The average %FSF at all sites for participants in this study surpasses this theoretical threshold meaning the level of attenuation may influence the quality of image and underestimation of EI. More investigation is required to assess effects of IMF on US image quality to develop standardize protocols that account for this factor.

Our findings revealed a moderate correlation between SFT and raw EI. There is limited research on the relationship between SFT and EI, however the relationship between IMF and body composition has been widely researched. Recently, one group of researchers pioneered an investigation into the association between SFT and paraspinal muscle fatty infiltration and the occurrence of LBP (Berikol et al., 2022). They found that increased SFT reliably predicted the

occurrence of vertebral degeneration as well as increased fatty infiltration of lower paraspinal muscle (Berikol et al., 2022). It has also been shown that total body fat percentage is associated with LM EI (Fortin et al., 2019). Such results confirm that body composition (total body fat mass, total lean body mass and SFT) should be taken into consideration when measuring muscle quality (Fortin et al. 2019). There is clear evidence suggesting SFT thickness may influence EI. To elaborate on the study by Neto Muller et al. (2021), differences between individual's SFT presents significant confounding effects to EI. They examined the effects of exogenous SFT of various thickness, applied to the skin over the tibialis anterior muscle. The purpose of their study was to examine the effects the of tissue alone in absence of confounds inadvertently introduced by other groups of researchers such as positional inconsistencies of focus adjustment (Fukumoto et al. 2012), and tissue density variation by transducer pressure manipulation (Young et al. 2015). While this group of researchers encourages the use of correction factors to buffer these effects, they emphasize the fact that there are several properties of SFT that contribute to the underestimation of EI and that additional studies are required to provide more comprehensive correction methods to account for the influence of SFT on EI.

#### *Clinical applicability, ODI and age*

Lastly, in addition to the aforementioned effects of body composition and several socio-demographic factors including sex, BMI, age and disability may contribute to the increase in IMF, resulting in hyperechoic (i.e., appears whiter) US images (Valera-Calero, Fernández-de-las-Peñas, et al., 2022; Valera-Calero, Al-Buqain-Ortega, et al., 2021; Rummens et al. 2020). When the muscle contains high fat content, the muscle tissue presents with EI similar to the surrounding connective tissue, blurring the borders of the ROI and introducing error potential during



segmentation process. Investigation into trends within specific populations may improve predictability of appropriate clinical US candidates as well as development of tailored protocols for specific demographic samples. We examined the relationship between disability (ODI) and age with EI. Our results showed moderate correlation between ODI and EI at only one site (L4 right), suggest that disability is likely not a significant factor in predicting LM EI. Supporting our results, one study found no association between US assessed cervical multifidus muscle morphology or quality and level of disability in patients with fibromyalgia syndrome (Valera-Calero, Úbeda-D'Ocasar, et al., 2022; Valera-Calero, Navarro-Santana, et al., 2022). Additionally, we found no relationship between age and EI. While Arts et al. (2010) showed muscle-specific non-linear correlation between EI and age, this was not translated in our data. Recent research suggests that age may be the most important factor related to error variance of EI measures (Valera-Calero, Navarro-Santana, et al., 2022). There is no consensus as to which factors most contribute to EI measurement error and mixed results hinder our current understanding of the influence of each factor (Valera-Calero, Al-Buqain-Ortega, et al., 2021; Valera-Calero, Navarro-Santana, et al., 2022). Future investigation of more focused effects of specific sociodemographic factors on EI are required to improve the potential clinic application of US in examining muscle morphology and function.

### *Limitations*

There are several limitations in this study, including the relatively small sample size. As highlighted above, the images quality of some subjects with greater percentage body fat (e.g., higher BMI) or greater IMF resulted in hyperechoic US images, presenting challenges in precisely

identifying ROI. While the US and MRI images were acquired at the same site/location, the images acquired with both modalities were not registered. Lastly, the effects of IMF and connective tissue on US attenuation are poorly understood and require further investigation.

## **Conclusion**

Our findings showed moderate to strong correlations between LM EI and %FSF measurements acquired via US and MRI, respectively. A moderate correlation between SFT and EI was also identified. The application of a CF to account for the influence of SFT led to a small improvement in the correlation between both measurements when all 4 sites were combined and analysed together. Future methodological studies are necessary to investigate the effect of SFT on EI measurements. US is a low-cost, non-invasive, accessible, and reliable method to examine muscle composition, and presents a promising solution for assessing and monitoring the effect of different treatment options for CLBP in clinical settings.

## **Conflict of interest statement**

The authors have no conflict of interest to declare.

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## **Figure captions list**

**Figure 1:** Bilateral transverse US images at L5 vertebral level tracing of right lumbar multifidus muscle cross sectional area (left) and corresponding subcutaneous fat thickness (right).

**Figure 2:** MR images at L4-L5 vertebral level tracing of lumbar multifidus muscle cross sectional area water image (left) and fat image (right).

**Figure 3:** Correlation Between Subcutaneous Fat Thickness and LM Muscle EI (Raw)

**Figure 4:** Scatterplot illustration the correlation between EI (raw and corrected) and MRI %FSF at L4.

**Figure 5:** Scatterplot illustration the correlation between EI (raw and corrected) and MRI %FSF at L5.

**Figure 6:** Correlation between MRI %FSF and a) US Raw EI and b) US corrected EI of all sites

## Tables

**Table 1: Participant demographics** (mean  $\pm$  standard deviation, (range)).

	Men (n = 9)	Women (n = 16)
Age, yrs	40.6 $\pm$ 12.8 (23-63)	39.6 $\pm$ 10.7 (21-59)
Height, cm	177.4 $\pm$ 8.7 (170.0-193.0)	166.6 $\pm$ 5.4 (160.0-178.0)
Weight, kg	92.7 $\pm$ 22.9 (62.0-140.0)	68.1 $\pm$ 11.3 (46.0-83.0)
BMI, kg/m <sup>2</sup>	29.2 $\pm$ 5.5 (21.5-40.9)	24.7 $\pm$ 4.7 (15.4-32.0)
ODI	29.1 $\pm$ 9.8 (16-40)	27.2 $\pm$ 9.5 (12-44)

Values expressed as mean  $\pm$  SD (range).

BMI = body mass index.

ODI = Oswestry disability index

**Table 2: US and MRI measurements for each site** (mean  $\pm$  standard deviation)

	LM L4 Right	LM L4 Left	LM L5 Right	LM L5 Left
	(n=25)			
<b>Ultrasound</b>				
Cross sectional area, cm <sup>2</sup>	8.8 $\pm$ 1.7	8.6 $\pm$ 1.4	9.4 $\pm$ 1.9	9.2 $\pm$ 1.5
Echo intensity (Raw), AU	67.8 $\pm$ 15.7	69.2 $\pm$ 15.9	76 $\pm$ 15.6	78.1 $\pm$ 14.6
Echo intensity (corrected), AU	94.9 $\pm$ 28.5	94.4 $\pm$ 26.8	106 $\pm$ 30.3	105.8 $\pm$ 28.2
<b>MRI</b>				
Measured Fat SI	63.0 $\pm$ 27.3	60.4 $\pm$ 26.1	67.0 $\pm$ 27.6	67.71 $\pm$ 26.8
Measured Water SI	218.3 $\pm$ 23.0	205.1 $\pm$ 25.7	210.1 $\pm$ 24.2	200.1 $\pm$ 26.2
% Fat signal fraction	22.0 $\pm$ 8.5	22.5 $\pm$ 9.1	23.8 $\pm$ 8.5	25.1 $\pm$ 9.2

Values are expressed as mean  $\pm$  SD. AU = arbitrary units. MRI = magnetic resonance imaging. SI = signal intensity.

**Table 3: Intra-rater Reliability**

	ICC	95% CI
<b>L4 Right</b>		
US EI	0.959 <sup>a</sup>	[0.875, 0.987]
MRI %FSF	0.993 <sup>b</sup>	[0.973, 0.998]
<b>L4 Left</b>		
US EI	0.963 <sup>a</sup>	[0.864, 0.989]
MRI %FSF	0.995 <sup>b</sup>	[0.982, 0.999]
<b>L5 Right</b>		
US EI	0.990 <sup>a</sup>	[0.919, 0.997]
MRI %FSF	0.991 <sup>b</sup>	[0.962, 0.998]
<b>L5 Left</b>		
US EI	0.978 <sup>a</sup>	[0.933, 0.993]
MRI %FSF	0.995 <sup>b</sup>	[0.980, 0.999]

<sup>a</sup> ICC average measure

<sup>b</sup> ICC single measure

**Table 4: Correlation between raw and corrected EI with MRI %FSF**

	Pearson correlation (r) (95% CI)	
	Raw EI	Corrected EI
<b>MRI %FSF vs. Muscle EI</b>		
L4 Right	.44* (.02, .75)	.40* (.06, .67)
L4 Left	.52** (0.07, .81)	.44* (.02, .74)
L5 Right	.20 (-.21, .54)	.30 (-.06, .61)
L5 Left	.40* (-.04, .73)	.51** (.23, .78)

\*Correlation is significant at the <0.05 level.

\*\*Correlation is significant at the <0.01 level.

**Table 5: Correlation between SFT and Raw EI**

	Pearson correlation (r) (95% CI)
<b>SFT vs raw EI</b>	
L4 Right	.41* (.01, .68)
L4 Left	.35 (-.11, .71)
L5 Right	.42* (.01, .69)
L5 Left	.39 (-.003, .70)

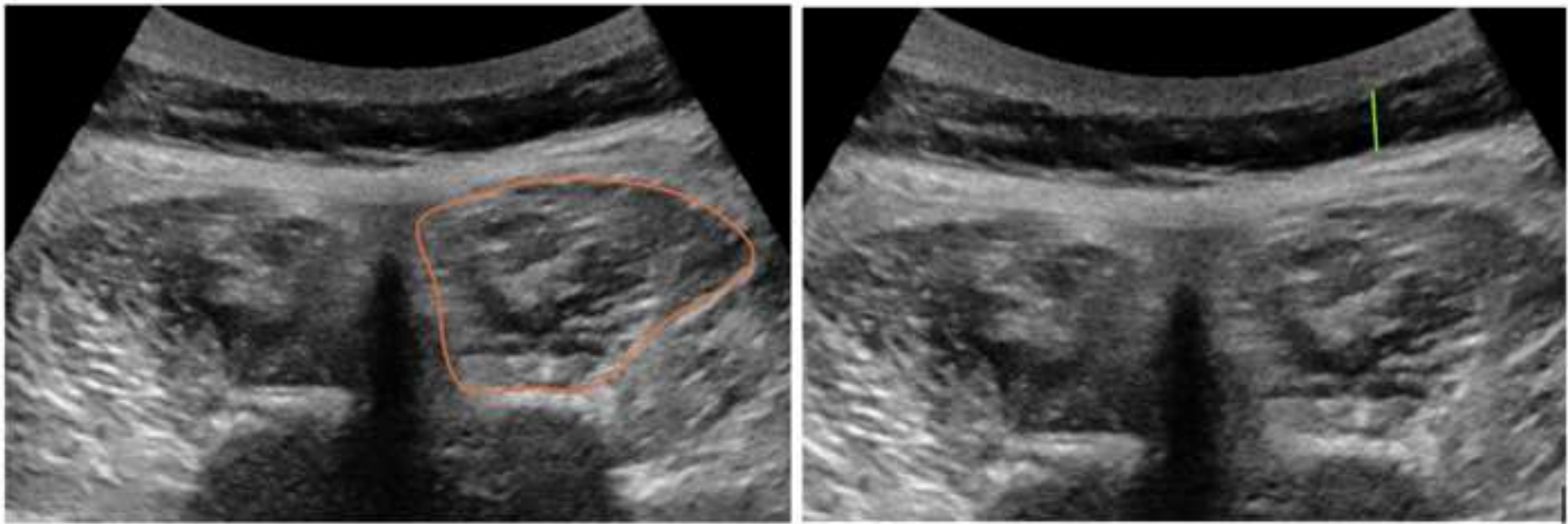
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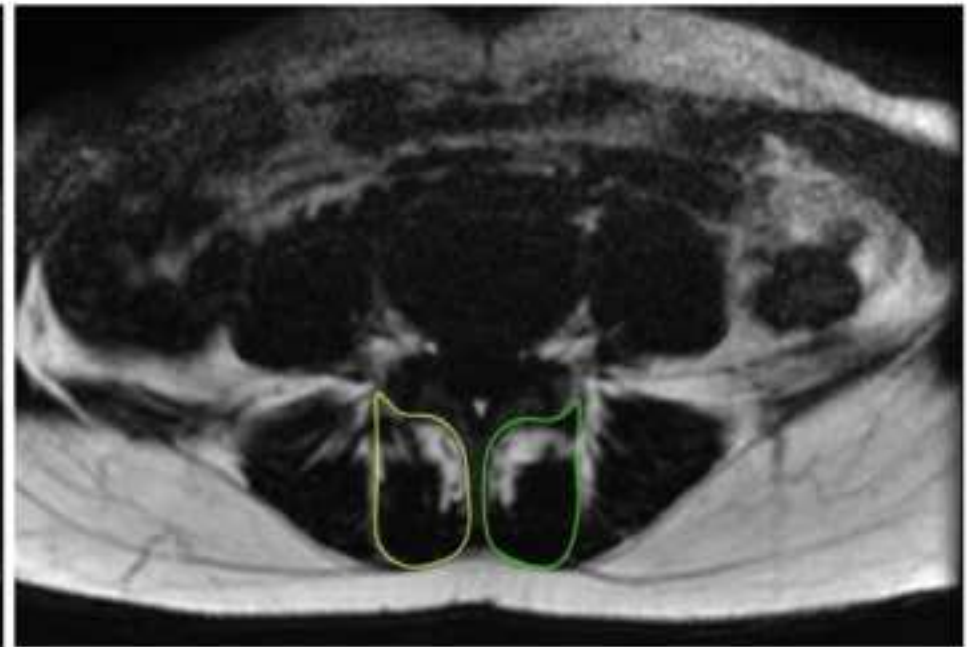
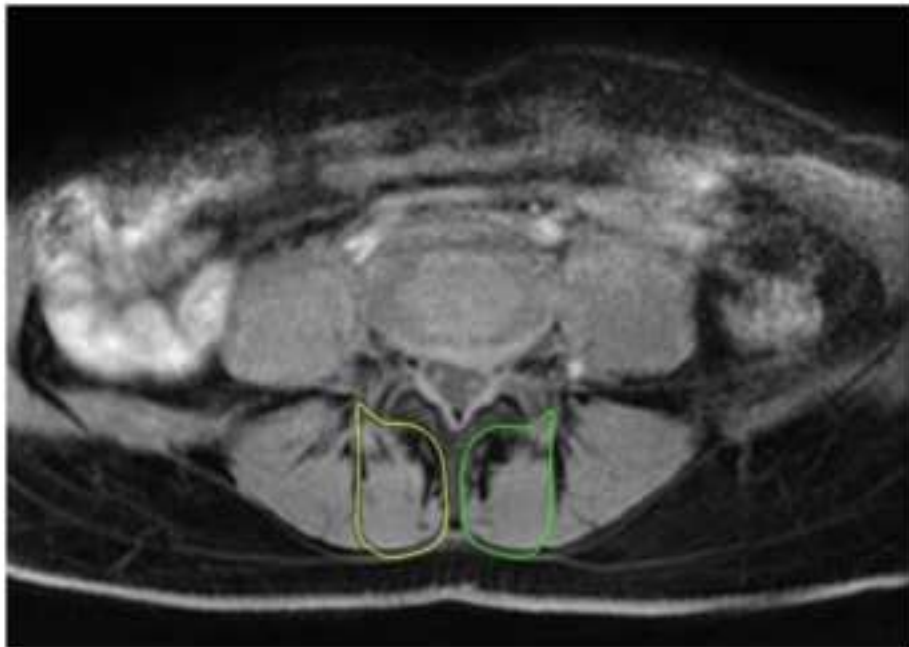
**Table 6:** Correlation between raw and corrected EI with Age and ODI

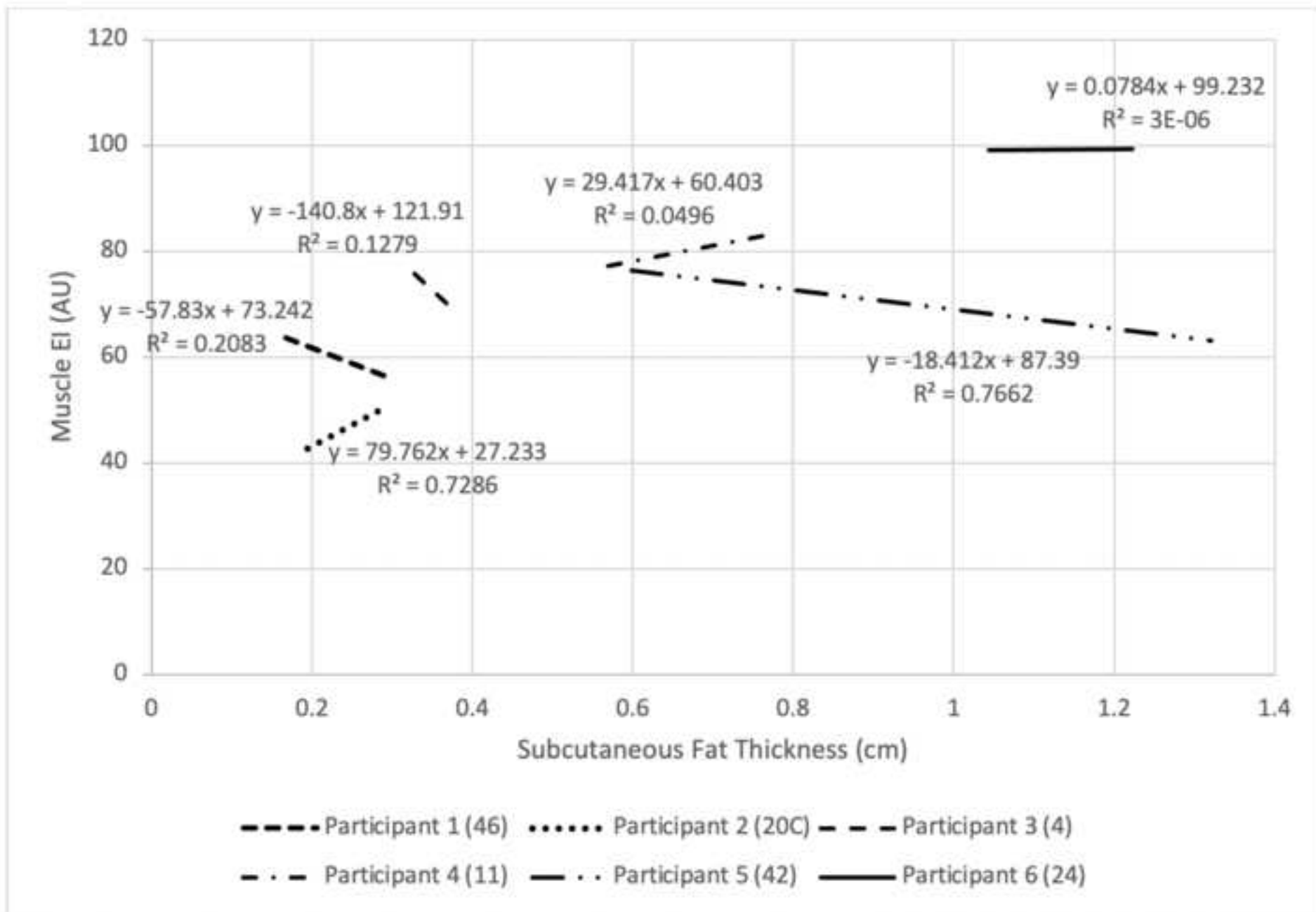
	Pearson correlation (r) (95% CI)	
	Raw EI	Corrected EI
Age vs. Muscle EI		
L4 Right	.24 (-.02, .51)	.16 (-.19, .44)
L4 Left	.24 (-.57, .53)	.16 (-.19, .50)
L5 Right	.01 (-.38, .41)	.02 (-.36, .38)
L5 Left	.07 (-.32, .41)	.07 (-.30, .41)
ODI vs. Muscle EI		
L4 Right	-.44* (-.73, -.05)	-.27 (-.59, .13)
L4 Left	-.34 (-.69, .15)	-.19 (-.55, .27)
L5 Right	-.31 (-.73, .21)	-.15 (-.55, .31)
L5 Left	-.13 (-.62, .41)	-.05 (-.50, .44)

\*Correlation is significant at the <0.05 level.

\*\*Correlation is significant at the <0.01 level.

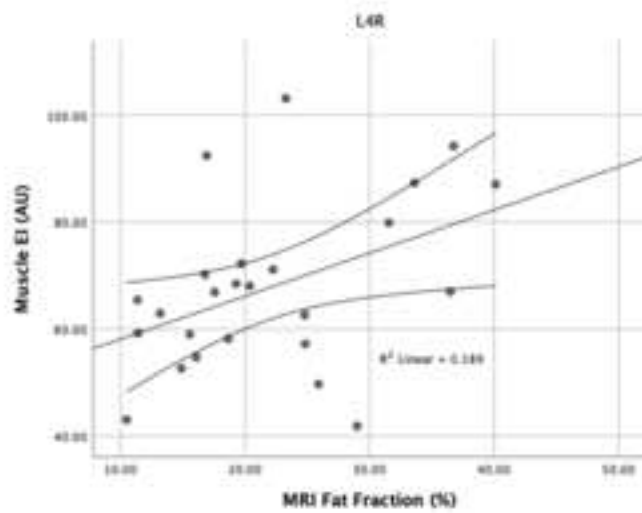




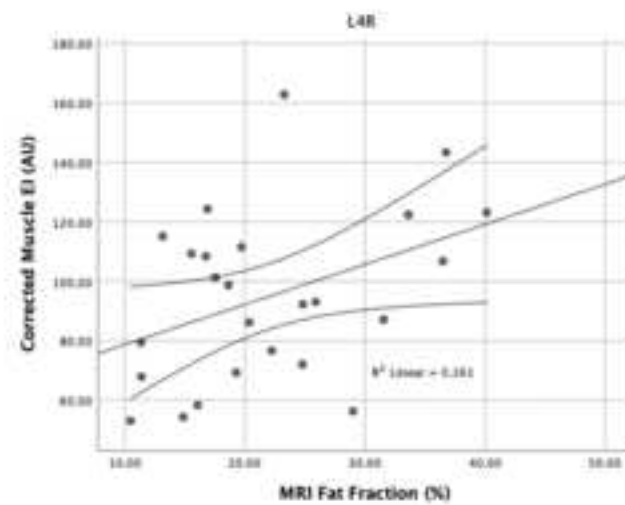




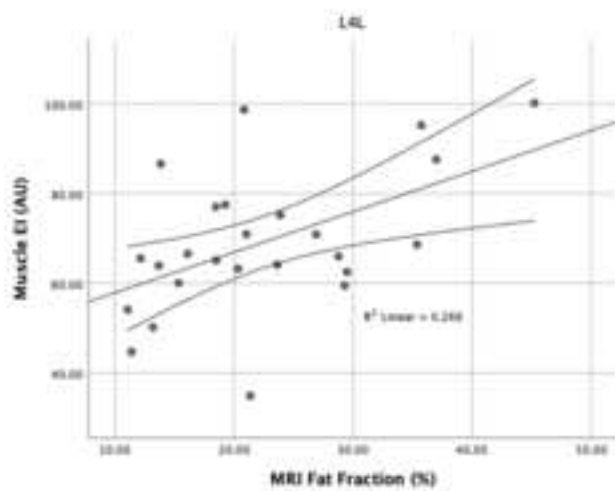
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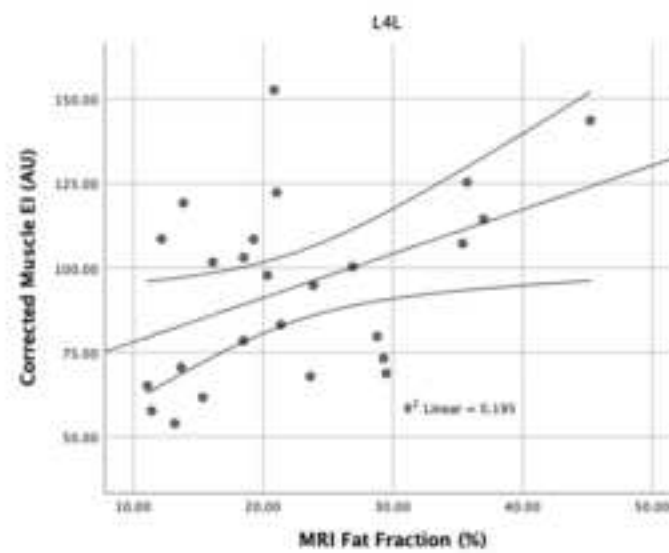
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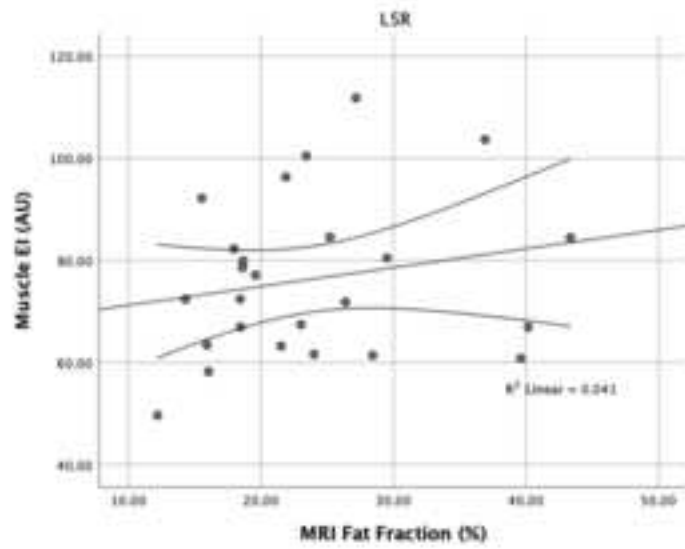
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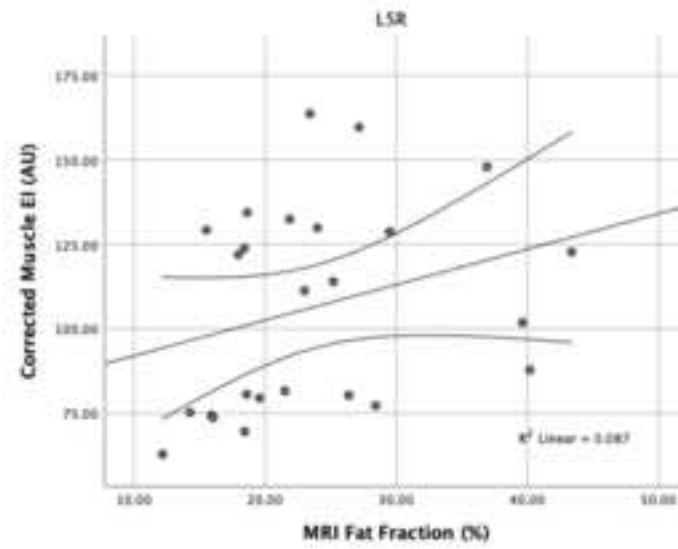
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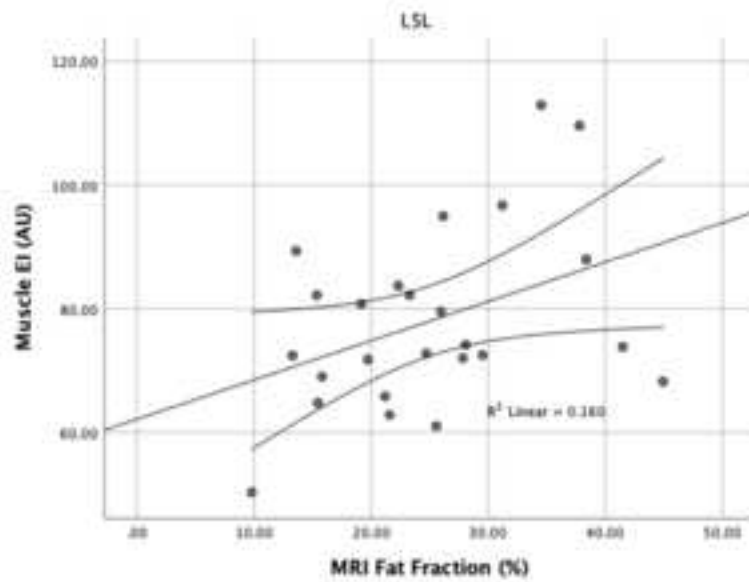
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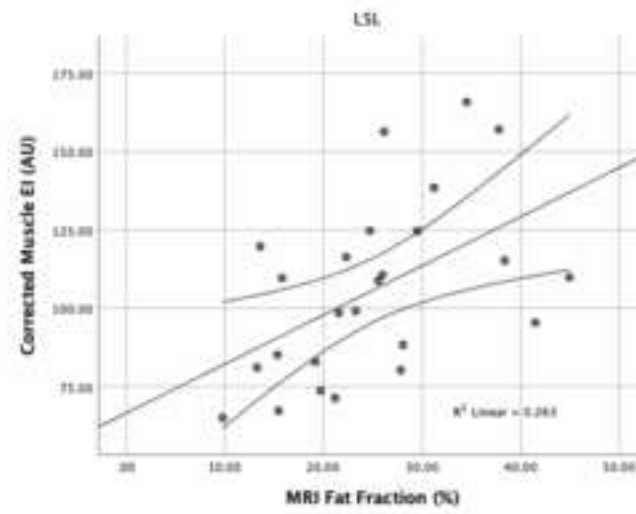
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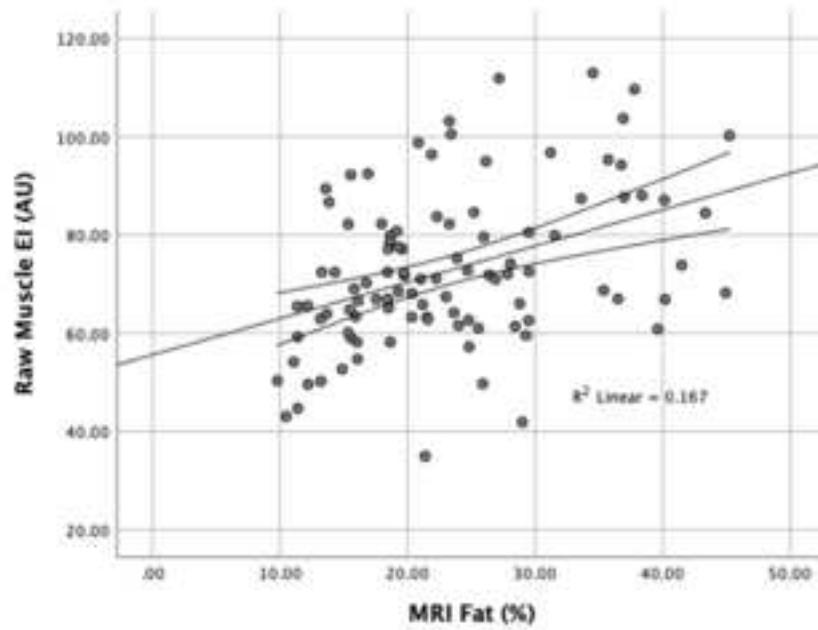
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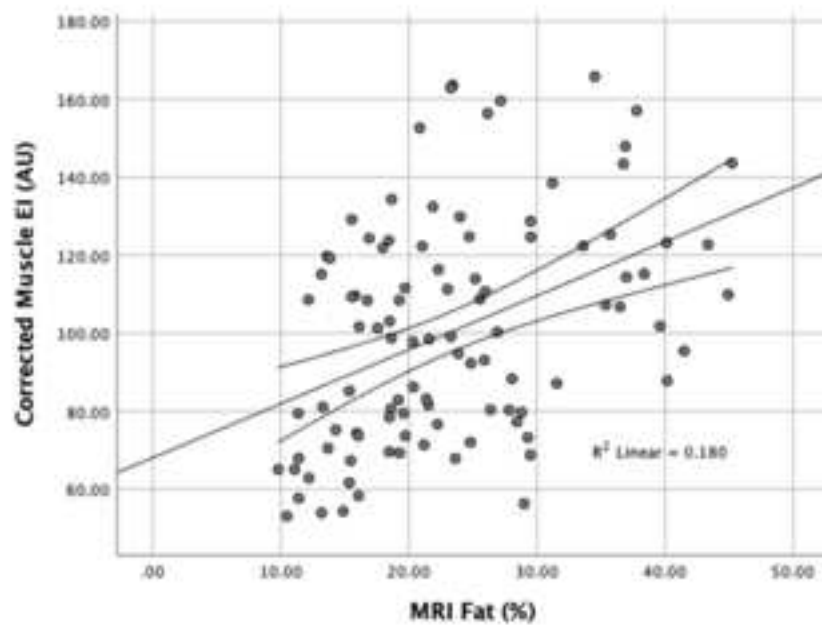


a.



$$r = .409^{**}, p < 0.001$$

b.



$$r = 0.425^{**}, p < 0.001$$