scientific reports

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Influence of CT parameters upon the quantification of sarcopenia and myosteatosis: a study with human bodies

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Computed tomography (CT) is increasingly being used in the quantitative assessment of body composition (CT-assessed sarcopenia and myosteatosis), but the effects of varying technical CT parameters on muscle area and density measurements remain unclear. This study analyzed muscle area and density in six human bodies scanned with varying CT parameters (tube current, tube voltage, slice thickness, and iterative reconstruction levels). Significance of differences relative to our institution's standard CT abdomen protocol and among all individual parameters was assessed, with effect sizes calculated to evaluate the magnitude of observed changes. Small but significant differences were found within all groups of investigated parameters, with most statistical significant differences corresponding to large effect sizes. Compared to our institution's clinical protocol, tube current showed significant variation in measured muscle area ($p < 0.001$), but no significant changes in regard to muscle density. Tube voltage and reconstruction level type both showed a significant direct relationship with muscle area ($p < 0.001$) and inverse relationship with muscle density ($p < 0.001$). Also, a slice thickness of 0.6 mm, compared to 1 mm, showed significant differences in both muscle area and density ($p < 0.001$), where no significant variation was observed between 1 mm and thicker slices. In conclusion, CT technical parameters significantly affect muscle area and density measurements. Although individual differences are small, their cumulative effect may lead to clinically significant discrepancies.

Keywords Sarcopenia, Myosteatosis, Computed tomography, Tube current, Tube voltage, Slice thickness

Changes in body composition, such as the progressive loss of muscle mass and function known as sarcopenia, alongside the deposition of adipose tissue within and between muscle fibers, termed myosteatosis, have emerged as an important marker of frailty with considerate prognostic value across various pathological conditions[1–4]. Computed tomography (CT) is increasingly being used in the quantitative assessment of body composition, particularly in the evaluation of low muscle area (CT-assessed sarcopenia) and low muscle density (CT-assessed myosteatosis).

A comprehensive understanding of the influence exerted by various technical CT acquisition and reconstruction parameters, including contrast phase, tube voltage, tube current, slice thickness and reconstruction methodology, is essential for precise quantitative evaluations of body composition and ensuring interchangeability between studies. A meta-analysis conducted by Amini et al. revealed considerable heterogeneity in CT acquisition parameters within sarcopenia and myosteatosis research, suggesting that this variability may contribute to inconsistent results and highlighting the potential effects of such parameters on quantitative measurements of body composition[5]. Importantly, inconsistencies in reported muscle area and density values across the literature primarily reflect methodological heterogeneity in CT acquisition and reconstruction, rather than limitations of body composition assessment itself.

Several studies have examined the influence of various CT parameters on quantitative muscle measurements. Lortie et al. conducted a review demonstrating that tube voltage significantly affects measured radiodensity. However, the studies included in the review combined various kilovoltage levels, making it challenging to reach

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a consensus on the impact of each specific level on individual Hounsfield Units (HU). Interestingly, the greatest discrepancies in measured HU were found in studies that compared the narrowest kilovoltage range (100 to 120)[6–8]. A recent phantom study showed that quantitative measurements of muscle area were reliable with different technical CT parameters, but muscle density was not[9]. An in vivo study by Fuchs et al. however, showed significant differences in both area and density for changes in slice thickness and tube current[10]. This uncertainty underscores the need for further investigation to clarify the factors influencing these measurements.

Therefore, the aim of our study was to investigate the potential impact of variations in technical CT parameters on the quantitative assessment of muscle area and density.

Methods

Muscle area and density were measured in six human bodies scanned under various conditions to assess the impact of different technical CT parameters, including tube current, tube voltage, slice thickness, and iterative reconstruction levels. Individual changes in CT parameters were evaluated by altering one parameter at a time while keeping the others constant.

The study was performed in the University Medical Centre Groningen (UMCG). The UMCG serves as a tertiarily referral center for the northern regions of The Netherlands and hosts a large anatomy department, dedicated to both training medical specialists and conducting research. All participants donated their bodies for both training and scientific purposes, with full anonymity. Exclusion criteria upon the initial acceptance of human anatomical donations were unnatural causes of death, prior autopsy or organ transplantation, and a body mass index exceeding 30. The embalming process was required to commence within 24 hours of decease. For this study we additionally excluded patients who had metallic orthopedic implants within the scanning plane, as these cause extensive beam and streak artifacts which make quantitative measurements highly unreliable.

As the study involved human bodies donated for science, it does not fall under the scope of the Dutch Medical Research Involving Human Subjects Act (WMO). Informed consent was obtained from all study participants, with all donors provided consent for their bodies to be used for both training and research purposes prior to their death. The study was conducted in accordance with institutional and national guidelines.

Embalming method

For body preservation the Thiel-embalming method was used. This method entails a multi-step procedure where chemicals are sequentially combined to form different kinds of embalming fluids. These fluids are then administered intravascularly and viscerally in multiple steps or utilized as a solution for full-body immersion. Through this process, tissue integrity is fixed and preserved, while natural color, flexibility, and texture are maintained[11–13].

CT scanning

All human bodies were scanned on a Siemens SOMATOM Definition Flash (Siemens Healthineers, Erlangen, Germany) CT scanner. The scan range extended from the 12th thoracic vertebra to the 5th lumbar vertebra. Our clinical CT abdomen protocol was used as 'default'. This protocol consisted of spiral scanning with a fixed tube voltage of 100 kV, a tube current of 100 quality reference (Qref) mAs on a 128 × 0.6 mm collimation with a rotation time of 0.5 s with a pitch of 0.8. Reconstruction parameters consisted of a standard slice thickness of 1 mm, Bv40 kernel, 512 × 512 pixel matrix and Advanced Modeled Iterative Reconstruction (ADMIRE) level 3 as reconstruction algorithm.

To investigate the effects of altering individual CT parameters, each parameter was systematically adjusted one at a time, while keeping the others constant, and the results were compared to those obtained using our institution's normal clinical CT abdomen protocol. The clinical protocol served as the reference for assessing the impact of these variations. Tube voltage settings included 80 kV, 100 kV, 120 kV, and 140 kV, tube current was set at 13, 25, 50, 100, and 200 Qref mAs, slice thicknesses was 0.6, 1, 2, 3, 5, and 10 mm, and ADMIRE levels 1, 3, and 5 were used.

Morphometric measurements

Muscle area and density were measured on a single slice in axial plane in the 12th thoracic vertebra and 1st to 4th lumbar vertebra, on the level were both transverse processes were shown. Density assessment relied on the standard CT calibration (with water equaling 0 HU and air − 1000 HU), with scans exported in DICOM format preserving all raw voxel attenuation values. Muscle measurements were performed using Sarcomeas v1.0, software developed in-house (UMCG, The Netherlands). This software facilitates the manual delineation of skeletal muscle areas based on tissue attenuation values obtained from the calibrated CT data preserved in the DICOM files[14]. Within these outlines, skeletal muscle was defined as all voxels with a radiodensity within the range of -29 HU to + 150 HU, while for adipose tissue was defined within the range of -190 HU to -30 HU[15]. The mean muscle density in HU was calculated by averaging the density of all voxels within the delineated muscle area that fell within the specified range for muscle tissue. All muscle contours were delineated on the default scan by an experienced board certified musculoskeletal radiologist with 15 years of experience (A.V.) and included the erector spinae, quadratus lumborum and psoas major muscles (Fig. 1). We excluded ventral abdominal wall muscles from delineation, as they were not fully intact due to prior surgical training procedures. To avoid bias from variation in manual delineation, all contours in the default scan were copied to the scans with the other parameters. For each vertebral level and body, the calculated values were summed and averaged, resulting in 30 measurements of muscle area and 30 measurements of muscle density per individual technical CT parameter.

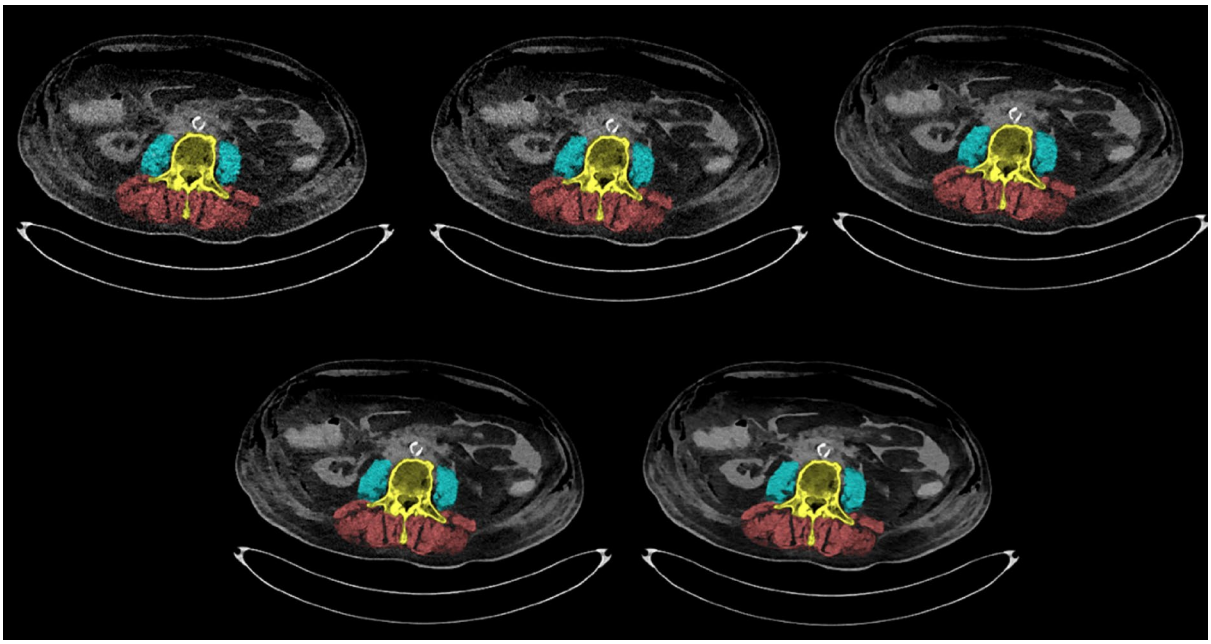

Fig. 1. Delineated muscle contours at different levels of tube current (Qref mAs 13 up left, 25 up middle, 50 up right, 100 lower left, 200 lower right). Note intra-abdominal and subcutaneous free air.

Statistics

Given the small sample size, data normality was assessed using the Shapiro–Wilk test. All measurements related to muscle area were normally distributed. Repeated measures ANOVA was employed to test for significant differences within these groups. Equal variances were evaluated with Mauchly's test of sphericity, which indicated violations of sphericity in all groups (epsilon < 0.750); therefore, the Greenhouse-Geisser correction was applied. All individual parameters were compared using the paired Student *t* test, with a Bonferroni correction to adjust for multiple comparisons. All muscle density measurements were not normally distributed. To assess significant differences within these groups, the Friedman test was employed. Comparisons of individual parameters were conducted using the Wilcoxon signed-rank test, also with a Bonferroni correction applied to account for multiple comparisons. For the comparison of individual CT parameters with the default protocol, mean differences (MD) were calculated in both absolute terms and as percentages. Additionally, 95% confidence intervals (CI) were determined for each comparison. For non-parametric comparisons, the Hodges-Lehmann estimator of the median difference was used to calculate the CI's. Effect sizes were additionally calculated to aid interpretation of the magnitude of observed differences. For normally distributed muscle area measurements, Cohen's *d* was used, with thresholds ranging from small (0.3–0.5), medium (0.5–0.8), and large (> 0.8) effect. For non-parametric muscle density outcomes, effect size *r* was derived from the Wilcoxon test using Rosenthal's formula $r = Z / \sqrt{N}$, with thresholds ranging from small (0.1–0.3), medium (0.3–0.5), and large (> 0.5) effect[16,17]. For all statistical tests, a p-value of < 0.05 was considered statistically significant.

Results

A total of six human bodies, consisting of two females and four males, were included in the study. Due to anonymized donation procedures, the ages of the donors were unknown, but all were estimated to be over 60 years old. Each body had a standard body composition without signs of severe cachexia. Table 1 presents the mean values for both the area and density groups across all parameters.

Tube current

Compared to the default protocol (100 Qref mAs), muscle area measurements showed statistically significant differences at 13 mAs (MD: -2.88 cm^2; 5.2%; 95% CI: -3.8 to -1.96 cm^2; d = -1.17; $p < 0.001$), 25 mAs (MD: -1.09 cm^2; 1.9%; 95% CI: -1.48 to -0.91 cm^2; d = -1.69; $p < 0.001$) both with large effect sizes, and 50 mAs with medium effect size (MD: -0.37 cm^2; 0.6%; 95% CI: -0.62 to -0.13 cm^2; d = -0.56; $p < 0.001$) (Fig. 2). No significant differences in measured density were observed between the default protocol at 100 Qref mAs and the other parameters, as shown in Fig. 3.

All individual muscle area values differed significantly, except between Qref mAs values of 100 and 200 ($p = 1.00$), while significant differences in muscle density were only found between 13 and 25 mAs with a medium effect size (r = -0.4; $p = 0.002$). These results are detailed in Supplementary Table 1.

Reconstruction algorithm type

Significant differences and large effect sizes were observed for both muscle area and density depending on the reconstruction algorithm used. Compared to the default ADMIRE level 3, ADMIRE level 1 showed a reduction

	Mean area cm^2	Mean density HU
Qref mAs 13	55.8	60.1
Qref mAs 25	57.6	58.8
Qref mAs 50	58.3	59.8
Qref mAs 100	58.7	59.2
Qref mAs 200	58.8	59.3
IR 5	59.0	59.1
IR 3	58.3	60.2
IR 1	57.3	61.1
0.6 mm	58.3	59.7
1 mm	59.2	59.0
2 mm	59.3	58.8
3 mm	59.3	59.0
5 mm	59.3	58.9
10 mm	59.2	58.7
80 kV	57.1	59.7
100 kV	58.7	59.4
120 kV	59.4	58.5
140 kV	59.8	57.9

Table 1. Mean muscle area and density values by tube current (Qref mAs), reconstruction level ADMIRE (IR), slice thickness (mm), and tube voltage (kV).

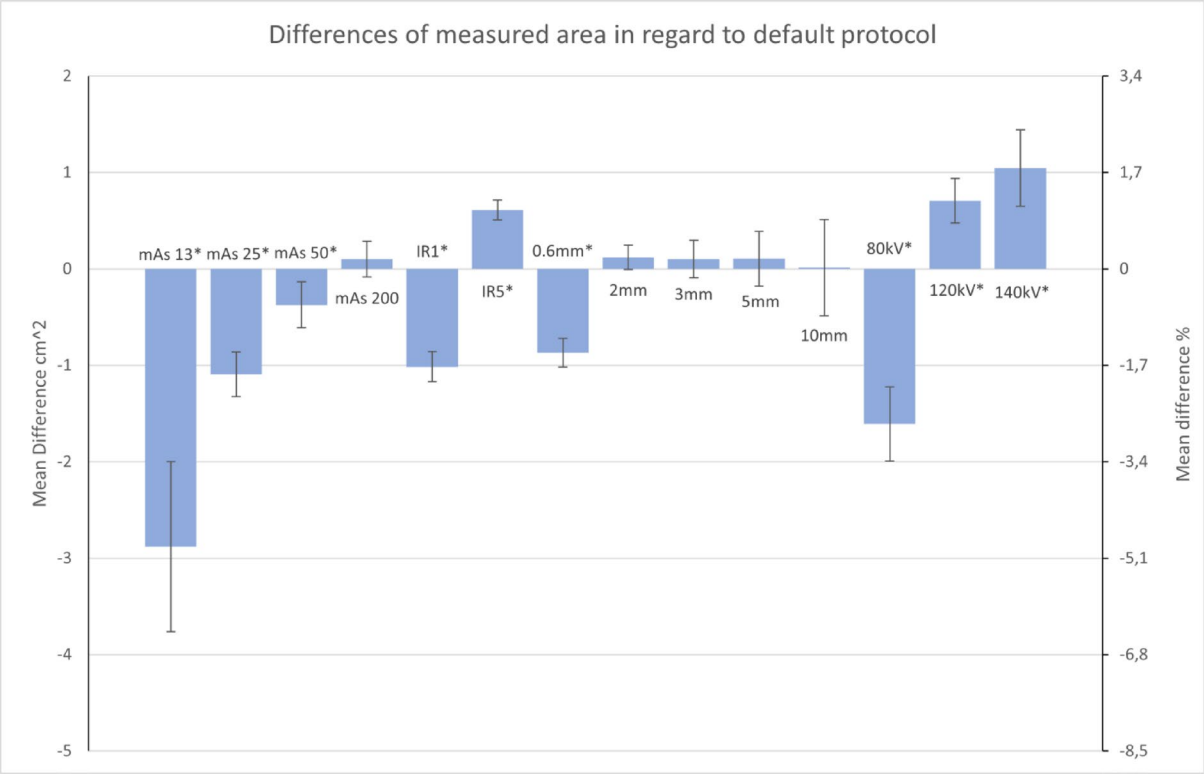

Fig. 2. Differences in measured muscle area relative to the default protocol (which consisted of a tube current of 100 Qref mAs, reconstruction algorithm ADMIRE level 3, tube voltage 100 kV and slice thickness of 1mm). The left y-axis shows absolute differences in cm^2, and the right y-axis shows the corresponding percentage differences. Negative values indicate a relative decrease compared to the default protocol, while positive values indicate a relative increase. Error bars represent the 95% confidence intervals.* = statistical significant ($p < 0.05$).

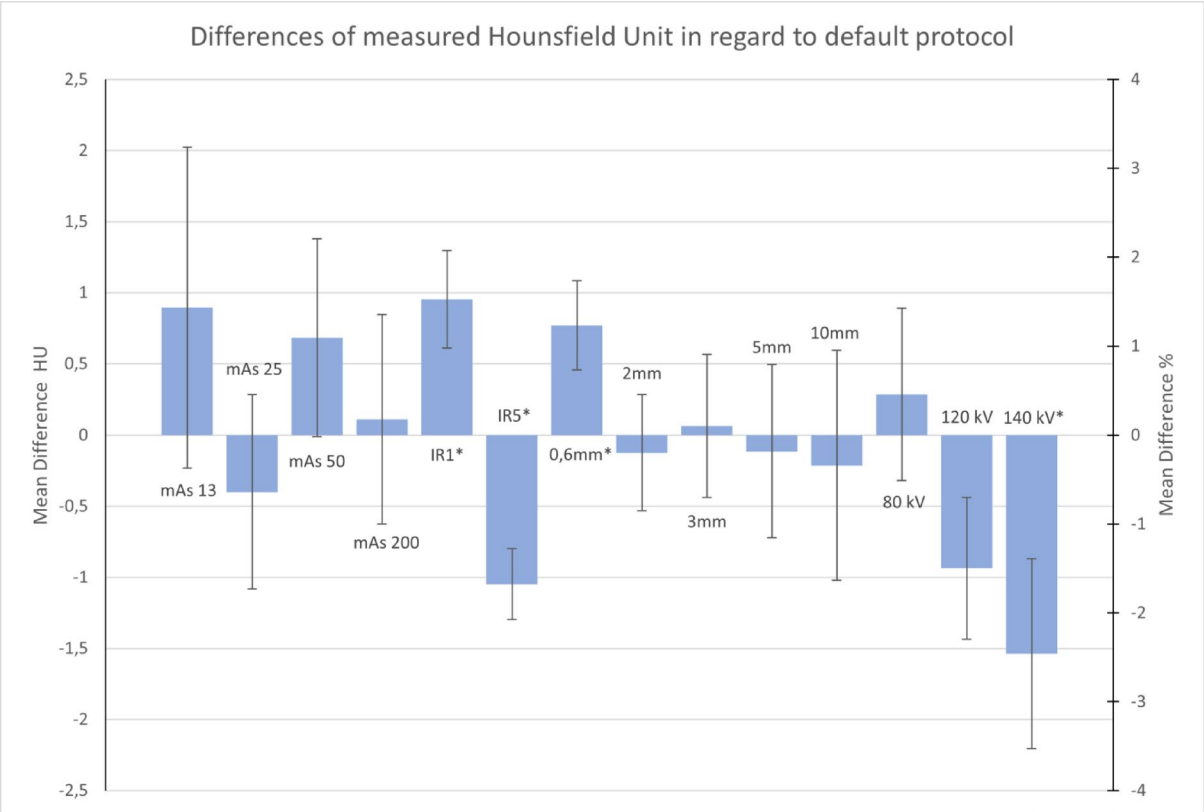

Fig. 3. Differences in measured muscle density relative to the default protocol (which consisted of a tube current of 100 Qref mAs, reconstruction algorithm ADMIRE level 3, tube voltage 100 kV and slice thickness of 1mm). The left y-axis shows absolute differences in cm^2, and the right y-axis shows the corresponding percentage differences. Negative values indicate a relative decrease compared to the default protocol, while positive values indicate a relative increase. Error bars represent the 95% confidence intervals.* = statistical significant ($p < 0.05$).

in muscle area (MD: -1.01 cm^2; 1.7%; 95% CI: -1.17 to -0.85 cm^2; d = -2.35; $p < 0.001$), while ADMIRE level 5 resulted in an increased area (MD: 0.61 cm^2; 1.0%; 95% CI: 0.50 to 0.71 cm^2; d = 2.1; $p < 0.001$) (Fig. 2). For muscle density, ADMIRE level 3 showed significant differences compared to both ADMIRE level 1 (MD: 0.95 HU; 1.5%; 95% CI 0.59 to 1.31 HU; $r = 0.61$; $p < 0.001$) and ADMIRE level 5 (MD: -1.05 HU; 1.7%; 95% CI -0.79 to -1.31 HU; r = -0.62; $p < 0.001$) (Fig. 3). All differences between parameters are listed in Supplementary Table 1.

Slice thickness

Only the 0.6 mm slice thickness showed both statistically significant differences as large effect sizes compared to the default 1 mm protocol, resulting in a reduction in muscle area (MD: -0.87 cm^2; 1.5%; 95% CI: -1.02 to -0.71 cm^2; d = -2.11; $p < 0.001$) and an increase in muscle density (MD: 0.77 HU; 1.3%; 95% CI: 0.44 to 1.10 HU; r = -0.54; $p < 0.001$) (Figs. 2 and 3).

No significant differences were observed between the default 1 mm protocol and thicker slices. Details of the comparisons between individual slice thicknesses are shown in Supplementary Table 1.

Tube voltage

When compared to the 100 kV of the default protocol, significant differences in muscle area were noted at 80 kV (MD -1.61 cm^2 (2.7%); 95% CI -2.00 to -1.21 cm^2; d = -1,5; $p < 0.001$), 120 kV (MD 0.71 cm^2 (1.2%); 95% CI 0.46 to 0.95 cm^2; $d = 1.08$; $p < 0.001$) and 140 kV (MD 1.04 cm^2 (1.8%); 95% CI 0.63 to 1.46 cm^2; $d = 0.95$; $p < 0.001$), as is shown in Fig. 2. All of these parameters corresponded to a large effect size. For muscle density, significant differences were found only at 140 kV (MD -1.54 HU (2.6%); 95% CI -0.84 to -2.23; r = -0.47; $p < 0.001$), see Fig. 3. Although there was a noticeable difference in muscle density between the default protocol and 120 kV (MD: -0.94 HU; 1.6%; 95% CI -0.41 to -1.46; r = -0.44), this difference was not statistically significant after adjustment for multiple comparisons ($p = 0.075$). Both these parameters had a medium effect size. A detailed breakdown of all comparisons is available in Supplementary Table 1.

Discussion

This study investigated the influence of various CT technical parameters on the quantitative assessment of muscle area and density using human bodies. Our findings show that even small changes in these parameters

can lead to significant differences in measurements. Importantly, most of the observed statistical significant differences corresponded to large effect sizes, with some showing medium effects, suggesting that the observed changes are not only statistically significant but also potentially practically relevant.

Significant differences in muscle area were observed across different tube current settings, with a variation of up to 5.2% between 13 and 100 Qref mAs. This aligns with previous studies, such as those by Fuchs et al. and Kim et al., which reported similar differences in muscle area measurements[9,10]. In contrast, a study by Zwart et al. compared muscle area measurements between low-dose PET-CT and diagnostic quality CT of the neck at the level of the third cervical vertebra and reported almost no change in measured area[18]. This discrepancy is possibly due to the reduced tissue volumes required for scanning in the head and neck region compared to the abdominal area.

Interestingly, when examining the differences in reported mean measured muscle density in relation to tube current, we observed a maximum variation of only 1.3 HU (2.2%). This is in sharp contrast with the average difference of 8.1 HU (46.5%) reported by Fuchs et al. on non-contrast enhanced CT images[10]. The reason for this substantial discrepancy is unclear. Although Fuchs et al. did not report their mean muscle density in HU on standard quality CT, the large percentage difference implies that their mean muscle density would be significantly lower than 30 HU, which is generally considered the lower boundary for normally attenuating muscle[19]. This indicates that Fuchs' study population could have been highly myosteatotic, potentially introducing a population bias that explains the observed differences. In contrast, our findings are more in line with the phantom study by Kim et al., which reported differences of up to 1.8 HU[9]. However, they did not provide absolute values, making it difficult to calculate precise percentage differences for comparison.

In contrast to muscle area, there does not appear to be a correlation between tube current and muscle density. The only significant difference between parameters was observed between Qref mAs values of 13 and 25, which is unexpected. Examination of the individual bodies with varying Qref mAs revealed that, unlike the mean area values, the mean muscle density values were quite heterogeneous, suggesting that the results are not due to a single outlier (see Supplementary Figs. 1 and 2). This heterogeneity is consistent with existing literature, where Fuchs et al. reported substantial effects, Kim et al. found small changes, and Morsbach et al. observed no effect of tube current on mean measured density[9,10,20].

Both muscle area and density were significantly affected by changes in the reconstruction algorithm. The results indicated that higher iterative reconstruction (ADMIRE) levels increased muscle area while reducing muscle density. To our knowledge no previous studies have looked at differences within distinct levels of Iterative Reconstructions (IR) algorithms. There have been studies examining variances between Filtered Back Projection reconstructions and IR, reporting minor differences with ranges comparable as those observed in our study, suggesting that changes in these algorithms can influence body composition measurements[9,21]. Although the specific algorithms and implementation details vary among vendors, the underlying principle remains the same. This highlights the importance of considering reconstruction algorithms in standardizing CT protocols for sarcopenia and myosteatosis research.

For slice thickness, only the 0.6 mm slice thickness, when compared to the default 1 mm, showed significant differences in both muscle area and density. No significant variations were observed between thicker slices. This is possibly due to the decrease in noise and increase in partial volume effect offsetting each other. This finding is consistent with previous research, such as Lee et al., which found minimal differences between 1.25 mm and 5 mm slices[22]. The significant discrepancies at 0.6 mm emphasizes the need to avoid using slice thicknesses below 1 mm to ensure accurate measurements.

Variations in tube voltage had a notable effect on both muscle area and density. Elevating the tube voltage corresponded to an increase in mean muscle area, with discrepancies of 2.7% between the default protocol of 100 kV and 80 kV, and even up to 4.7% observed difference between 80 and 140 kV. This finding parallels the outcomes reported by Morsbach et al. who likewise observed a 5.1% difference in muscle area using dual energy CT within the same voltage spectrum[23].

There is an inverse relationship between voltage and measured muscle density. As voltage increases, density decreases, with a difference of 1.5 HU (2.6%) between the default protocol of 100 kV and 140 kV, and a maximum difference of 1.7 HU (3.1%) between 80 kV and 140 kV. This is expected as with increasing voltage the x-ray beam becomes less attenuated by tissues which causes a shift in HU. Available literature reported mean differences are much bigger, ranging between 5.2 HU (11.5%;Lennartz et al.) and 22.3 HU (40%;Ippolito et al.)[6,7,24]. However, these studies all used contrast-enhanced CT's, suggesting that contrast agents may further amplify the effects of voltage changes. Adjusting the HU range used to identify skeletal muscle based on the applied tube voltage could help improve the accuracy of muscle measurements in future studies.

While the individual differences observed in this study were small, their cumulative effect may lead to clinically significant discrepancies, particularly when using threshold-based cut-offs for diagnosing conditions like sarcopenia and myosteatosis. Even minor variations in muscle area or density could cause a patient to fall above or below diagnostic thresholds, potentially resulting in misclassification. For example, a recent study comparing changes in morphometric measurements of body composition across different contrast phases showed small differences and near-perfect correlation coefficients; however, when applying binary cutoff values for sarcopenia or visceral obesity, a substantial number of false negatives were still observed[25]. This highlights that small measurement differences can have outsized clinical impact when relying on absolute cutoff values. Therefore, it is critical to standardize CT protocols to minimize variability and improve the reliability of body composition measurements in clinical and research settings. Our study indicates that a standardized CT protocol for body composition analysis should avoid slice thicknesses below 1 mm, refrain from using low-dose (reduced mAs) techniques, employ a fixed kV setting, and ensure consistency in the reconstruction algorithm used.

We used human bodies which have been embalmed by the method as first described by Walter Thiel[11,12]. To our knowledge, no studies have specifically examined differences in X-ray attenuation between living patients

and Thiel-embalmed bodies. Given that our measured HU values are similar to those of living patients, we expect that tissue attenuation in Thiel-embalmed bodies is consistent with that of living tissue, thereby minimizing any substantial bias. In our institution these bodies serve primarily for surgical training purposes. In the abdominal area they are mainly being used for intraperitoneal procedures, both laparoscopy as laparotomy (e.g. body in Fig. 1 has intra-abdominal air due to recent laparoscopy training). Consequently, the muscles of the ventral abdominal wall were not fully intact. Thus to prevent bias we decided to only delineate the psoas, erector spinae and quadratus lumborum muscles.

This study has several limitations. First, the sample size was small, limiting the generalizability of the findings and preventing the development of regression equations for different vertebral levels. In addition, all included bodies were estimated to be over 60 years old, of European descent, and there was a limited representation of body compositional variability, which may further restrict the applicability of our findings to younger populations, other ethnic groups, or those with markedly different body habitus. Furthermore, no information was available regarding comorbidities. Although the number of donor bodies was limited, the repeated-measures design with extensive parameter variation allowed robust assessment of within-subject effects of CT acquisition and reconstruction settings. Second, to mitigate bias from manual delineation, we replicated the contours drawn on the 'default protocol' scan onto the other scans within the same parameter. However, lower quality images pose challenges for accurate muscle delineation, impacting both manual human delineation and AI scripts, a factor we did not account for[26]. Third, a potential source of bias is that, because our study excluded the ventral abdominal wall muscles, our quantitative values are not directly comparable to studies that included all muscles. However, as our primary focus was on differences (both absolute and percentual) between CT parameters, this bias is expected to be minimal, though slight effects may persist due to partial volume differences. Fourth, the study did not account for variations across different CT scanner vendors, models, or vendor's reconstruction algorithms, nor did it consider differences in patient positioning, which may limit the generalizability of the results to other CT systems or clinical settings[27–29]. Future research should aim to include larger populations, especially those with higher BMI, and evaluate the impact of different CT systems and contrast-enhanced scans. Additionally, developing standardized protocols for CT acquisition in sarcopenia and myosteatosis research would be beneficial.

Conclusion

The aim of this study was to investigate the impact of variations in technical CT parameters on the quantitative assessment of muscle area and density. Our findings show that changes in tube current, tube voltage, slice thickness, and reconstruction algorithms produced statistically significant differences, with effect sizes suggesting many are also of practical importance. While individual differences were relatively small, their cumulative effects may result in clinically relevant discrepancies, especially when applying threshold-based cut-offs for sarcopenia or myosteatosis. These findings provide practical guidance for standardizing CT acquisition and reconstruction protocols for body composition analysis, which may improve comparability of measurements across studies and clinical settings.

Data availability

The datasets generated during and/or analysed during the current study are available from the corresponding author on reasonable request.

Received: 9 October 2024; Accepted: 23 December 2025
Published online: 08 January 2026

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Declarations

Competing interests

The author(s) declare no competing interests.

Additional information

Supplementary Information The online version contains supplementary material available at https://doi.org/10.1038/s41598-025-34140-x.

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