Rapid calculation of bespoke body segment parameters using 3D infra-red scanning

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A B S T R A C T
Body segment parameters such as segment mass, centre of mass and moment of inertia, serve as important inputs for musculoskeletal modelling. These parameters are normally derived from regression tables; however, can be poorly representative of the study population with variations of up to 40% recorded between different tables. More recent methods, such as 3D scanning, present a rapid and accurate way to produce subject-specific body segment parameters for use in musculoskeletal models. An infra-red 3D scanner was used to produce full-body scans of 95 males and females. Each was put through an algorithm to calculate bespoke segment mass, centre of mass and inertial properties for each segment of the body, with results comparable to cadaveric data. These methods could be used to increase the specificity of musculoskeletal modelling outputs for individual subjects, improving the accuracy of modelling outputs in biomechanics-related research.

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1. Introduction

Musculoskeletal (MSK)1) modelling of human movement requires body segment parameters (BSP) such as segment mass, centre of mass (CoM) and inertial properties, to calculate intersegmental muscle and joint forces. BSPs are typically calculated through the use of regression-based anthropometric tables derived from cadaver [1–5] or imaging data [6–9], which feature both male and female subjects and a large range of ages. The segment mass and inertial properties predicted using these tables have been shown to vary by up to 40% [10]. Previous literature suggests BSP proportions change with age [8,11,12], sex [7,8,12–15] and morphology [12,16], yet studies only appear to consider differences in sex when forming regression tables, along with the ability to input height and mass, which will not consider the aforementioned proportion changes. Therefore, using generalised regression tables is ill-advised when focusing on a specific population, as the initial populating group of the regression is unlikely to reflect the subjects of any one study.

MSK modelling outputs have been shown to be significantly affected by BSP perturbations [17–22], with joint moments varying by up to 20% during gait [17,22]. An increased sensitivity at the hip to BSP error [10,17,20–23], generally more prominent during the swing phase, could be attributed to joint moment contributions. During swing, where no additional external reaction forces are present, BSPs will dominate contributions to joint torque requirements; whereas ground reaction forces perform a similar role in stance. The majority of studies use gait trials to determine the sensitivity of joint moments, however, some predict that for high acceleration activities the sensitivity may be higher [10,24]. Many applications of MSK modelling are for low acceleration activities, where the inertial tensor falls out of the equations as angular accelerations drop to near-zero. Therefore, it is expected that low acceleration activities would be less sensitive to perturbations in BSPs (particularly inertia) than those with high accelerations. One study showed high sensitivity in joint moments, again mostly in the hip, during cycling [25], however, these uncertainties were small at 10% of the peak hip joint power. There are few studies looking at high-demand activities and the effects of BSP variation, and further research is needed to confirm these hypotheses. Considering the potential effects on both quantified joint power and muscle force for certain activities, there is a need to develop protocols by which users can produce subject-specific BSPs rapidly. Bespoke methods of calculating BSPs should allow users to produce appropriately scaled models of individual subjects, as opposed to estimating these from regression tables produced from non-homogenous populations.

Approaches used to address the issue of bespoke MSK models are to either calculate subject-specific BSPs or create a regres-
sion table that is specific to the study population [12,26]. In both cases, the rapid acquisition of BSPs is required. Custom values can be achieved through geometric estimation or 3D imaging techniques. The former, such as that proposed by Hanavan [27], can produce errors of up to 37% [8] and is time consuming in a clinical setting. Three-dimensional medical imaging techniques such as magnetic resonance imaging [MRI, 28-30] and dual-energy x-ray absorptiometry [DEXA - 8,20,31,32] give good estimates of BSPs. However, full body MRI scans can take up to 90 min to complete, are limited by high costs for equipment, staffing and maintenance, extensive post-processing, assumptions of tissue density, and, in the case DEXA, exposure to ionising radiation.

3D light scanning has more recently been introduced as a method to obtain fast measurements of the body without expensive equipment or time-consuming scans. This can either be through the use of laser [22,33,34] or infra-red light [35], neither of which pose any harm to the body. The use of infra-red scanners, such as the Microsoft Kinect (Microsoft Corporation, Redmond, USA), has proven popular not just in the gaming industry but in robotics, body scanning and the healthcare environment [36]. Its low cost [37], readily available software for integration with motion capture environments, and promising results in estimating BSPs [38] and object scanning [39] make the Kinect a good alternative for full body scanning.

Despite the numerous advantages, many full body scanners are large and cumbersome, featuring multiple sensors and frames requiring assembly. A product by Styku (Los Angeles, USA), has been designed to minimise these disadvantages and would give a motion capture environment quick access to 3D anthropometric data. A full body scan takes less than a minute and produces a surface point cloud of between 40,000 and 60,000 vertices. The scan is produced by a Microsoft Kinect which remains stationary whilst the subject is rotated about a single point on a small platform. The platform rotates slowly and the area required is small; only $2.5 \times 1.7 \text{m}$ is sufficient for a full scan.

The aim of this study was to assess the use of a 3D infra-red scanner in producing subject-specific BSPs for use as bespoke inputs into MSK models. The methods for producing the BSPs will be presented, along with the proposed set up and measurement techniques. This study did not aim to assess the suitability for a specific population, but rather to evaluate the use for both males and females, and a range of ages, heights and masses. It was hypothesised that this new technology would be a more rapid method of producing subject-specific BSPs, with an accuracy comparable with empirical published data.

2. Methods

Ninety-five participants, 62 males and 33 females, volunteered for a full body scan using a Styku 3D body scanner, consisting of a Microsoft Kinect, tripod and rotating platform (Fig. 1). The age and sex of each subject was recorded, who were then asked to change into skin tight clothing (a form-fitting running kit was provided), removed their shoes and socks and either tied their hair or wore a swimming cap. The height and mass (for comparisons to estimated body mass - the sum of all segment masses) of each subject was recorded, after which each was asked to step onto the remotely-operated rotating platform with their feet shoulder-width apart and assumed the anatomical stance. The scan was initiated, taking approximately 60 s whilst the subject was rotated through 360°. Once complete, the platform was halted and the scan stopped. Approval for this study was obtained through the Joint Research Compliance Office and Imperial College Research Ethics Committee. All subjects gave written informed consent.

Each scan produced between 40,000 and 60,000 vertices points, which were used to manually identify anatomical landmarks when splitting into body segments (Table 1 and Fig. 2). Landmarks were chosen to best represent each part of the body as a functional segment, in line with published anatomical locations used to construct segment coordinate frames [40,41]. Due to a lack of visual external landmarks on a scan to define the right and left ASIS points, the distance from knee joint to hip joint centre was estimated based on the expected thigh length defined from the regression table of De Leva [13] and the Shank ratio of measured to expected length. The feet, shanks, forearms and hands were all segmented based on planes through anatomical joints; namely the ankle, knee, elbow and wrist. For the purposes of this study, the data representing the neck were discarded due to a lack of available data in the literature for comparison, leaving the head as a separate segment.

Once the body was partitioned into 14 individual segments, each was passed through an algorithm calculating segment mass, CoM and moments of inertia (MoI), as described in Fig. 3 and in more detail in the following sections.

2.1. Segment mass

The volume of each scan was used to calculate segment mass. As shown in Fig. 2, the scans were built from layered data. Each layer consisted of a 2D surface profile which was sectioned using Delaunay Triangulation. The area and centroids were calculated for each triangle:

$$\text{Area} = \sqrt{s * (s-a) * (s-b) * (s-c)}$$

(1)

where: $s = \frac{1}{2} * (a+b+c)$, and $a$, $b$, and $c$ are the lengths of the three sides of each triangle.

$$\text{Centroid}_{x,y} = \left( \frac{1}{3}(x_1 + x_2 + x_3), \frac{1}{3}(y_1 + y_2 + y_3) \right)$$

(2)

where $x_{1,2,3}$ and $y_{1,2,3}$ are the $x$ and $y$ coordinates for the three vertices of each triangle relative to the global axis.

The distance between layers (between 0.6 and 0.7 cm) was used to determine the volume of each small mass. This was multiplied by the relevant density, as discussed in the following section, to give individual triangle mass which was summed per layer, and then per segment, to give whole segment mass.

2.2. Density

A uniform density was used within each segment when calculating mass [22,34,35,42,43], however, due to the unique combination of bone, muscle and fat in individual segments of the body.
Table 1

Anatomical landmarks and planes used to partition the body into functional segments. Any bony landmarks stated here were visually identifiable on the 3D scans. Planes are shown in Fig. 2.

| Segment | First plane                                                                 | Second plane                                                                 |
|---------|-----------------------------------------------------------------------------|-----------------------------------------------------------------------------|
| Foot    | Plane connecting lateral/medial malleolus                                   | Horizontal plane passing through the most anterior point of the patella      |
| Shank   | Plane connecting lateral/medial malleolus                                   | Plane originating at the most medial part of the groin and passing through  |
|         |                                                                             | the estimated hip joint centre                                              |
| Thigh   | Horizontal plane passing through the most anterior point of the patella     | Plane connecting the jugular notch and the most posterior-inferior part of  |
|         |                                                                             | the neck (or the seventh cervical vertebrae)                                |
| Trunk   | Plane originating at the most medial part of the groin and passing through |
|         |                                                                             | the estimated hip joint centre                                              |
| Upper arm | Plane connecting the armpit and most superior-lateral point of the shoulder | Plane perpendicular to the arm long axis, passing through the olecranon      |
| Forearm | Plane perpendicular to the arm long axis, passing through the olecranon     | process                                                                        |
| Hand    | Plane connecting the wrist styloids                                         | Plane connecting the most posterior-inferior point of the occipital bone and |
| Neck    | Plane connecting the jugular notch and the most                            | a point where the mandible meets the neck line                              |
|         | posterior-inferior part of the neck (or the seventh cervical vertebrae)    |                                                                              |
| Head    | Plane connecting the most posterior-inferior point of the occipital bone    | –                                                                            |
|         | and a point where the mandible meets the neck line                          |                                                                              |

Fig. 2. Initial data (left) with segmentation planes indicated by dashed lines (top boxed image shows planes for the head and neck), zoomed in section of thighs to show detailed scan data (centre) and final segmented scan (right).

[44], the same assumption cannot be applied when assigning intersegment density values. Densities values were based on literature cadaver data (Table 2). Previous studies calculating whole-body mass found significant differences between a subject’s measured mass and the sum of the individual, estimated, segment masses [22,29,38]. Following the method of Clauser et al. [4], the ratio of estimated segment masses to estimated whole body mass was used to distribute the expected mass differences between measured and estimated whole body mass.

2.3. Centre of mass

Having determined the centroid of each 2D triangle in the previous section, it was then possible to find the centroid of the whole 2D layer. As a uniform density was used across each segment, centre of area was equivalent to the CoM. The centroid of the layer was calculated as:

$$x_{layer}, y_{layer} = \frac{\sum_{i=1}^{n} A_i \cdot (x_i, y_i)}{layer\,area}$$  (3)
Fig. 3. Flowchart describing the steps for calculating body segment parameters. Darkest boxes describe steps for inertial property calculation, the lightest boxes describe the steps for calculating centre of mass and the remaining boxes describe mass calculation. Numbers correspond to accompanying images/equations.
where \( A_i = \text{individual triangle area} \), \( n = \text{number of elements in triangulation} \) and \((x_i,y_i) = \text{distances from the global axis to the triangle centroid} \). An iteration of Eq. (3) was then used to calculate the coordinates for the CoM of the whole segment relative to the global axis:

\[
X_{\text{Segment}} \cdot Y_{\text{Segment}} \cdot Z_{\text{Segment}} = \sum_{\text{layer}} M_{\text{layer}} \cdot \left( X_{\text{layer}} \cdot Y_{\text{layer}} \cdot Z_{\text{layer}} \right) / \text{layer mass}
\]

(4)

where \( M_{\text{layer}} = \text{mass of the whole layer} \) and \( Z_{\text{layer}} = \text{the height of the layer relative to the global axis} \). Previously, studies that calculated BSPs were restricted to single plane data [8] and therefore the CoM could only be given as an \( x \) and \( y \) coordinate, or relative to a proximal/distal joint. Using 3D scanning, the CoM can be calculated as a 3D coordinate which takes into account mass distribution in the anterior–posterior direction in addition to existing information for the medial-lateral and superior–inferior directions. One restriction of existing regression tables is that the proximal/distal segment ends are aligned with the centre of mass, despite evidence to the contrary [9,46]. The method presented here therefore provides a more anatomically-correct CoM position.

### 2.4. Inertial tensor

Firstly, the moment of inertia for each triangulation element about the layer CoM was found:

\[
I_{\text{element, CoM}} = \sum_{i=1}^{n} m_i \cdot \left( x_i^2 + z_i^2, z_i^2 + y_i^2, x_i^2 + y_i^2 \right)
\]

(5)

where \( i = \text{triangle (element)} \), \( m_i = \text{triangle mass} \) and \((x_i^2 + z_i^2, z_i^2 + y_i^2, x_i^2 + y_i^2)\) indicates the Euclidean distances between the triangle centroid and the layer \( y, x \) and \( z \) axes, whose origins all pass through the layer centroid. As each triangle lies within the same \( x\)-\( y \) plane as the centroid, \( z \) distances are equal to 0. The parallel axis theorem determined the moment of inertia of the layer about an arbitrary axis, in this case the global system as determined by the 3D scanner calibration:

\[
I_{\text{layer, axis}} = \left( \sum_{\text{element}=1}^{n} I_{\text{element, CoM}} \right) + m_{\text{layer}} \cdot \left( x_{\text{layer}}^2 + z_{\text{layer}}^2, z_{\text{layer}}^2 + y_{\text{layer}}^2, x_{\text{layer}}^2 + y_{\text{layer}}^2 \right)
\]

(6)

where \( m_{\text{layer}} = \text{total mass of the layer} \) and \((x_{\text{layer}}^2 + z_{\text{layer}}^2, z_{\text{layer}}^2 + y_{\text{layer}}^2, x_{\text{layer}}^2 + y_{\text{layer}}^2)\) indicates the Euclidean distances between the layer centroid and the \( y, x \) and \( z \) axes of the arbitrary system.

An iteration of the parallel axis theorem then determined the segment’s inertial tensor around its own CoM. As the sum of the moments of inertia of individual layers (which make up the segment), about an arbitrary axis, equals the whole segment’s moments of inertia about an arbitrary axis, then:

\[
I_{\text{segment, CoM}} = I_{\text{segment, axis}} - m_{\text{segment}} \cdot \left( X_{\text{segment}}^2 + Z_{\text{segment}}^2, Z_{\text{segment}}^2 + Y_{\text{segment}}^2, X_{\text{segment}}^2 + Y_{\text{segment}}^2 \right)
\]

(7)

where \( I_{\text{segment, axis}} = \sum_{\text{layer}=1}^{n} I_{\text{layer, axis}} \), \( m_{\text{segment}} = \text{total segment mass} \) and \((X_{\text{segment}}^2 + Z_{\text{segment}}^2, Z_{\text{segment}}^2 + Y_{\text{segment}}^2, X_{\text{segment}}^2 + Y_{\text{segment}}^2)\) indicates the Euclidean distances between the segment CoM and the \( y, x \) and \( z \) axes, respectively, of the arbitrary system.

### 2.5. Rotation of segments

Participants were asked to stand in the anatomical stance. This meant that the arm segments (upper arm, forearm and hand) were not positioned in parallel with the global vertical axis. Finding the angle of each segment relative to the vertical axis (Fig. 4) allowed the calculated inertial tensors to be correctly aligned with the appropriate anatomical coordinate system, and transformation equations were used to re-calculate inertial properties in line with the global system vertical axis. This gave the inertial tensor aligned with the newly positioned segments.

Transformations were only required in the \( z \) (superior–inferior) and \( x \) (medial–lateral) axes:

\[
I_{\text{new}, z} = I_{\text{z}} \cdot \frac{I_{\text{z}}}{2} - \frac{I_{\text{z}}}{cos2\theta + I_{\text{y}} sin2\theta}
\]

(8)

\[
I_{\text{new}, z} = I_{\text{z}} \cdot \frac{I_{\text{z}}}{2} + \frac{I_{\text{z}}}{cos2\theta - I_{\text{y}} sin2\theta}
\]

(9)

where \( I_{\text{z}} = \text{segment original moment of inertia about CoM} \), \( I_{\text{new}, z} = \text{corrected moment of inertia about CoM and } \theta = \text{angle of seg-}
ment relative to vertical axis. \( I_{xz} \) = product of inertia, and was calculated as follows:

\[
I_{xz, \text{layer, axis}} = \left( \sum_{\text{element}=1}^{n} I_{\text{element, CoM}} \right) + m_{\text{layer}} \cdot (X_{\text{layer}} \cdot Z_{\text{layer}}) \tag{10}
\]

\[
I_{xz, \text{segment, CoM}} = \left( \sum_{\text{layer}=1}^{n} I_{xz, \text{layer, axis}} \right) - m_{\text{segment}} \cdot (X_{\text{segment}} \cdot Z_{\text{segment}}) \tag{11}
\]

### 2.6. Statistical analysis

Table 3 lists the literature studies used for comparison against the current study’s 3D scanning data.

Independent \( t \)-tests were used to test for any significant differences between the 3D scanning values and the literature for segment volume, mass and CoM. Only studies using similar segment boundary designations to that of the current study were considered. Paired-sample \( t \)-tests were used when comparing measured mass to estimated body mass. Reviewed studies either did not present Mol values [13], instead presenting radii of gyration relative to height, or failed to specify segment axes and therefore statistical comparisons were not conducted.

Hand CoM values were presented as a percentage of the whole hand, including fingers, and literature values, if stating otherwise, were adjusted to match this definition. All other segment CoM values are given relative to the proximal joint except the head which is given relative to the most superior point.

All inertial properties (MoI about the: transverse axis = \( I_{xx} \), sagittal axis = \( I_{yy} \) and coronal axis = \( I_{zz} \)) were presented as absolute values and are given relative to the segment CoM. Additional studies by McConville et al. [9] and Durkin & Dowling [8], who computed Mol via medical imaging methods (stereophotometry and DEXA, respectively) on living subjects, have been included where available. Local coordinate frames for each segment match those used by Chandler et al. [5], except the feet where the current study defined the positive \( Z \)-axis in the superior direction and the positive \( X \)-axis in the anterior direction.

### 3. Results

Subject data are presented in Table 4. Segment volume is presented relative to total subject volume, and segment mass is presented relative to total subject mass. CoM is presented as percentages of segment length.

This study had significantly higher segment volumes than the literature for most segments, where the feet and trunk had lower volumes (Table 5).

Segment mass data are shown in Table 5. The mean estimated body mass was 103.1 (±5.7) % of the measured mass (\( p < 0.0005 \)), prior to applying the correction factor described in the methods. Significant differences in segment mass were found for the feet, hands, upper arms and head (\( p < 0.0005, p < 0.022, p < 0.0005, p < 0.0005 \), respectively). The upper arm, hands and head were overestimations, whereas the feet values were underestimated compared to the literature. For CoM, the current study was significantly larger for the thigh (\( p < 0.0005 \)) and forearm (\( p < 0.0005 \)) and significantly smaller for the shank (\( p < 0.0005 \)), hands (\( p < 0.0005 \)) and upper arm (\( p < 0.0005 \)).

For visual comparisons of the moments of inertia, Fig. 5 shows the current study data about all axes for each segment’s local coordinate frame along with literature absolute values.

### 4. Discussion

The methods in this study represent a rapid and accurate way to produce subject-specific BSPs. In terms of segment designation, the current study made use of known anatomical points to create anatomical boundaries between segments. Some studies were found to draw a cutting plane parallel to the transverse plane through the groin [42,48,49]. This would not accurately represent the thigh as a functional segment as the most superior point of the groin does not align with the hip joint centre. The current study follows a more representative superior boundary line for an accurate definition of the thigh.

#### 4.1. Speed of scan and BSP calculation

Only one previous study was found which used the Microsoft Kinect to estimate BSPs; Kudzia scanned 21 subjects (10 male, 11 female) in a static pose whilst a handheld sensor was moved around the body [38]. Each scan took around 30 s, but additional time was required for anatomical land-marking (30 min) and extensive post-processing (45 min). This compares to around 60s for a scan using the method outlined in this study, followed by 10–15 min for manually finding landmarks on the point cloud, and less than a minute post-processing for the BSP calculation.

#### 4.2. Volume

The results from the 3D scanning method outlined in this study are in good agreement with cadaveric data. The differences shown are all below 3% and can be attributed to fluid loss from the cadavers prior to immersion testing [50] and the difference in demographics used for this study. Head and thigh volumes were larger than those observed in the literature data: hair was not accounted for separately, and as a result the scanner would have interpreted this data as extra head volume. Whilst defining the thigh using estimations of the hip joint centre is a more suitable method of defining the functional segment, it may have included some distal parts of the trunk and is methodologically different from other studies in the literature [42,48,49]; this also explains the difference in trunk volume. Feet volumes were lower than the literature. Visual inspection of the scans showed a loss of the inferior parts of the segment due to scanner field-of-view limitations, making it difficult to produce a comprehensive view of the foot. An alternative method, where subjects stand on a slanted transparent block, may overcome this in future design iterations [38].

The cadaver studies consisted mostly of elderly male subjects, whereas this study has both male and female subjects, and a large range of ages, heights and weights. Considering the loss of fluid
Table 4
Subject data.

| Sex          | Age (SD, min, max), years | Height (SD, min, max), m | Body mass (SD, min, max), kg |
|--------------|---------------------------|--------------------------|-----------------------------|
| Male (62)    | 34 (± 13, 18, 71)         | 1.78 (± 0.07, 1.62, 1.92) | 78.2 (± 10.5, 59.2, 106.7)  |
| Female (33)  | 32 (± 10, 19, 64)         | 1.66 (± 0.06, 1.53, 1.80) | 62.1 (± 9.6, 48.8, 89.0)    |

Table 5
Comparisons between this study and literature segment volume, mass and CoM. All literature data were combined into one group and averaged. * (p < 0.05) and ** (p < 0.0005) indicate significant differences from the current study.

| Segment   | N (segment volume) | N (%) | Mean segment mass (% total body mass ± SD) | N (%) | Mean segment mass (% body weight ± SD) | N (%) | Mean CoM (% segment length from proximal end ± SD) |
|-----------|--------------------|-------|------------------------------------------|-------|----------------------------------------|-------|--------------------------------------------------|
| Feet      | This study 190     |       | 0.6 (0.1)                                | 190   | 0.64 (0.14)                            | 190   | 44.2 (1.8)                                      |
| Literature 26 | 1.2 (0.2) **   | 53    | 1.48 (0.29) **                          | 23    | 43.8 (1.3)                             |       |                                                  |
| Shank     | This study 190     |       | 3.9 (0.6)                                | 190   | 4.19 (0.62)                            | 190   | 37.2 (1.6)                                      |
| Literature 26 | 3.9 (0.6)       | 54    | 4.23 (0.57)                              | 24    | 41.8 (1.8) **                         |       |                                                  |
| Thigh     | This study 190     |       | 10.4 (1.3)                               | 190   | 10.76 (1.34)                          | 190   | 43.1 (2.1)                                      |
| Literature 26 | 9.4 (1) **      | 54    | 11.13 (2.62)                             | 24    | 41.0 (3.3) **                        |       |                                                  |
| Trunk     | This study 95      |       | 52 (4.1)                                 | 95    | 49.59 (4.16)                          | 95    | 51.3 (1.6)                                      |
| Hands     | This study 190     |       | 55.3 (2.9) *                            | 95    | 50.62 (5.10)                          | 10    | 50.7 (5.9)                                      |
| Literature 13 | 0.6 (0.1)       | 53    | 0.71 (0.16)                              | 190   | 41.3 (1.7)                            | 17    | 48.3 (7.1) **                                    |
| Forearm   | This study 190     |       | 1.6 (0.3)                                | 190   | 1.68 (0.33)                           | 190   | 42.8 (1.7)                                      |
| Literature 26 | 1.4 (0.2) *     | 54    | 1.61 (0.27)                              | 21    | 41.6 (2.1) *                         |       |                                                  |
| Upper arm | This study 190     |       | 3 (0.4)                                  | 190   | 3.08 (0.42)                          | 190   | 39.8 (2.3)                                      |
| Literature 26 | 2.4 (0.3)        | 53    | 2.84 (0.34) **                          | 22    | 49.2 (4) **                          |       |                                                  |
| Head      | This study 95      |       | 7.7 (1.3)                                | 95    | 8.26 (1.36)                          | 95    | 60 (5.4)                                        |
| Literature 13 | 6.6 (1.2) *    | 27    | 7.08 (0.84) **                          | 10    | 58.4 (12)                             |       |                                                  |

in cadavers and the known decrease in muscle mass with age, it appears logical that the volume data presented from the 3D scanning method of living subjects would be larger than that of the literature, and highlights the improvement of using 3D scanning to determine bespoke volumes.

4.3. Segment mass

The differences between measured whole-body mass and summation of estimated segment mass are comparable to other studies [22,29,38]. The correction factor described in the methods was applied to account for these differences. Significant differences in segment mass between 3D scanning and the literature mass values appear to follow the same pattern as segment volume, mostly falling below 10%.

4.4. Centre of mass

Despite the differences in foot mass, the segment CoM aligns well confirming that the loss of inferior segment data is the most likely cause for disparity in segment mass. Discrepancies of 10–25% were found between the 3D scanning and literature data for the shank, hands and upper arm. This study produced significantly different values for upper arm CoM, with the CoM positioned more towards the proximal end of the segment which can be attributed to the extra shoulder data included in the scanned data. Many cadaver studies appear to have discarded the mass around the glenohumeral head and lateral parts of the scapula, creating a purely humeral segment. However, the shoulder complex as a functional segment consists of much of the tissue surrounding the lateral parts of the scapula and as such, has been included in this study's designation of the upper arm.

4.5. Moments of inertia

Few studies report experimentally-obtained Mols, so direct comparisons were limited. Mol cannot be normalised to height and mass directly, however, De Leva [13], through adjustments to Zatsiorsky et al.'s [6] work, instead reports radii of gyration. This makes the final values sensitive to any discrepancies in segment mass and length and could result in large differences between estimated and regression-predicted values.

There is good visual agreement between the results of this study and previous literature for most segments, except the foot and hand. Differences between the current study and Dempster [3] for the trunk may relate to inconsistencies in the CoM values, as Dempster's are higher than those found by other studies. The hands and feet do show differences, which are most likely a result of errors from mass estimation.

4.6. MSK modelling outputs

This work has relevance for MSK modelling, for which BSPs act as bespoke inputs. This study found differences between the 3D scanning and cadaver data, however, these were not large enough to have a significant effect on MSK modelling outputs for low acceleration activities. Ganley & Powers [20] developed a DEXA-related method which over-predicted thigh mass by 12% but did not have a significant effect on hip joint moments during gait. Rao et al. [17] found that hip joint moments showed significant differences of up to 20% when perturbations in thigh mass reached up to 37% when comparing 6 different regression tables. This study produced a smaller error of 3%, demonstrating that it not only is an advance on the literature for modelling gait, but may also be a significant improvement for high acceleration activities where model outputs are more sensitive to errors in BSPs [10,24,25].

4.7. Limitations

This study had a major limitation in the measurement of head and feet BSPs. Further work on how to address field-of-view limitations must improve the representation of these segments. During the scan, each participant was expected to remain still for around 60 s. For certain age groups, this may prove problematic leading
Fig. 5. Moments of Inertia about the centre of mass, where $I_{xx}$ = horizontal stripes, $I_{yy}$ = white fill and $I_{zz}$ = diamond pattern. Study indicators on X-axis: 1 = Current study, 2 = Dempster (1955), 3 = Drillis and Contini (1966), 4 = Durkin and Dowling (2003), 5 = Chandler et al. (1975) and 6 = McConville et al. (1980).

to some sway of the body. There were no discernible movements in the participants during this study, however, future iterations of the design may want to consider this when scanning older participants.

The identification of externally-visible landmarks for segment designation will be subject to the individual performing the segmentation, however, any error in finding landmarks is expected to be minimal owing to the ease in which the majority could be located. The precise location of the knee joint and proximal end of the thigh, the latter being ideally based on a plane through the groin and ASIS, were two exceptions to this. The former was taken as the most anterior (and visible) point of the patella which forms a clear apex around the knee joint. Subjects’ ASIS points were not visible landmarks on the scan data, which would only be rectified by manual palpation. The alternative, estimating the hip joint centre, was not entirely based on regression but on the ratio of measured to expected shank length as per De Leva’s [13] regression tables. Nevertheless, the partial use of regression means the specificity of the thigh is diminished. Future protocols may look to include a measurement of the ASIS
positions for each subject. This should also be considered for the knee.

The assumption of using a uniform density along the length of a segment remains a limitation when calculating body mass from volume data. Previous work has shown that using a uniform density, over density profiles, has little effect on the calculation of inertial properties, and that inaccurate segment volumes were instead more detrimental [51]. The current study showed good comparisons with cadaveric volumes, however, future work using density profiles and/or values for BSPs may consider developing more robust methods for estimating density, whilst also determining the necessity of such work. The data representing the neck was not considered in this work, however, MSK modelling studies looking at the spine or neck may warrant further investigation. This would be important in defining the correct inertial properties of the accelerating head and neck segments, which would be of interest when considering head/neck interaction.

5. Conclusions

This study has presented a new method for calculating BSPs with an accuracy in line with empirical data from cadaver studies. Such methods can easily be integrated into a clinical or research setting, providing users with rapid access to subject-specific BSPs or the potential to create population-specific regression tables. Comparisons of absolute data with the literature serve as confirmation that the BSPs calculated from the 3D scanning method are accurate to the point at which they will not significantly affect the outputs from MSK modelling for low acceleration activities. The BSPs were found to be within the ranges found in the literature, however, some differences existed due to limitations of the scanner and natural variations between the group scanned in this study compared to those in the literature.

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Conflict of interest statement

The authors declare no conflict of interest with the content of this study.

Ethical approval

Approval for this study was obtained through the Joint Research Compliance Office and Imperial College Research Ethics Committee. Reference 16IC3224.

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