# TLEM 2.0 - A comprehensive musculoskeletal geometry dataset for subject-specific modeling of lower extremity 

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#### Abstract

When analyzing complex biomechanical problems such as predicting the effects of orthopedic surgery, subject-specific musculoskeletal models are essential to achieve reliable predictions. The aim of this paper is to present the Twente Lower Extremity Model 2.0, a new comprehensive dataset of the musculoskeletal geometry of the lower extremity, which is based on medical imaging data and dissection performed on the right lower extremity of a fresh male cadaver. Bone, muscle and subcutaneous fat (including skin) volumes were segmented from computed tomography and magnetic resonance images scans. Inertial parameters were estimated from the image-based segmented volumes. A complete cadaver dissection was performed, in which bony landmarks, attachments sites and lines-ofaction of 55 muscle actuators and 12 ligaments, bony wrapping surfaces, and joint geometry were measured. The obtained musculoskeletal geometry dataset was finally implemented in the AnyBody Modeling System ${ }^{\mathrm{TM}}$ (AnyBody Technology A/S, Aalborg, Denmark), resulting in a model consisting of 12 segments, 11 joints and 21 degrees of freedom, and including 166 muscle-tendon elements for each leg. The new TLEM 2.0 dataset was purposely built to be easily combined with novel image-based scaling techniques, such as bone surface morphing, muscle volume registration and muscle-tendon path identification, in order to obtain subject-specific musculoskeletal models in a quick and accurate way. The complete dataset, including CT and MRI scans and segmented volume and surfaces, is made available at http://www.utwente.nl/ctw/bw/research/projects/TLEMsafe for the biomechanical community, in order to accelerate the development and adoption of subject-specific models on large scale. TLEM 2.0 is freely shared for non-commercial use only, under acceptance of the TLEMsafe Research License Agreement.


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## 1. Introduction

Musculoskeletal models of the lower extremity represent a valuable tool to explore various biomechanical problems, where accurate knowledge muscle and joint reaction forces is necessary. At the turn of this century, Rik Huiskes was one of the initiators to link musculoskeletal models with finite element models in a European project entitled 'Pre-clinical testing of cemented hip replacement implants: Prenormative research for a European

[^0]standard'. In that project a consortium of academic and industrial partners tried to establish simplified and validated loading protocols to be used as input for finite element models and experimental testing set-ups. The project was rather successful although the protocols were not accepted as tests by the ISO-standardizing committee. It was concluded that there was still a lot of work to be done to improve the robustness of the finite element simulations and the applicability of the experimental protocols. Nevertheless, Rik was very satisfied with the results of the project as it gave a lot of information to unravel the failure scenarios that were involved. Typically Rik, with many others, was not interested in the individual patient, but focused more on the general phenomena which dominated failure of these implants. However, times are changing and over the last 10 years the demand to explain differences amongst patients has grown tremendously. Hence,
the modeling community is challenged to incorporate the huge variability amongst patients in terms of anatomy, activity levels, loading conditions, etc. To do that, patient-specific musculoskeletal modeling tools need to be developed and this paper contributes to that goal. We can only guess how Rik would feel about this development of patient-specific simulations. One thing is for sure: without his work on hip biomechanics, we would not be at this stage where we are able to utilize these new modeling tools to assess biomechanical issues at the hip joint for an individual patient.

In the past, musculoskeletal models of the lower extremity have been used in several disparate disciplines, such as in orthopedic surgery to simulate the effects of joint replacements (Delp et al., 1994; Piazza and Delp, 2001) and tendon transfers (Piazza et al., 2003; Reinbolt et al., 2009); in neurology to model the effects of a stroke (Higginson et al., 2006), disorders of the central nervous system (Steele et al., 2012; Van der Krogt et al., 2013), and spinal cord injuries (Paul et al., 2005; To et al., 2005); in sport to optimize athletes performances (Pandy et al., 1990; Rasmussen et al., 2012), and analyses and prevent injuries (McLean et al., 2003; Manal and Buchanan, 2005); or in ergonomics for prevention of work-related musculoskeletal disorders (Wu et al., 2009).

Unfortunately, the reliability of force predictions is affected by the accurateness of many model inputs. In particular, one of the most sensitive parameters of the musculoskeletal geometry is represented by muscle moment arms (Hoy et al., 1990; Out et al., 1996), whose estimation depends on the identification of the muscle-tendon lines-of-action (Rohrle et al., 1984; Pal et al., 2007); moreover, errors in the estimated position of muscle attachment sites have been shown to affect muscle force predictions (Carbone et al., 2012).

To represent different subjects, simple linear scaling laws are usually applied to generic models, which are based on one or more cadaver specimens (Delp et al., 1990b; Klein Horsman et al., 2007; Arnold et al., 2010). However, these scaling procedures do not take into account the inter-individual variability present in musculoskeletal geometry (White et al., 1989a; Duda et al., 1996). For these reasons, subject-specific models have been shown to be necessary when exploring complex biomechanical problems, such as representing pathologies in the musculoskeletal anatomy and predicting the outcome of orthopedic surgery (van der Krogt et al., 2008; Lenaerts et al., 2009; Scheys et al., 2009; Taddei et al., 2012).

Constructing subject-specific models without intensive manual intervention represents a significant challenge. Indeed, several recent studies have focused on developing such subject-specific models based on imaging or functional measurements (Blemker et al., 2007; Scheys et al., 2011; Hainisch et al., 2012; Hausselle et al., 2014) but their clinical application on a large scale has not been demonstrated.

An interesting approach to obtain subject-specific models is to register or morph the medical images of the subject to a previously built template or atlas, containing muscle-tendon attachment sites and lines-of-action (Pellikaan et al., 2014), or muscle volumes (Carbone et al., 2013). However, no musculoskeletal model in literature is linked to such a template or atlas. The Twente Lower Extremity Model (Klein Horsman et al., 2007) represents so far the most complete and consistent dataset of the lower extremity, including both musculoskeletal geometry and muscle-tendon architecture based on one single cadaver specimen. Unfortunately, lack of detailed medical images of that cadaver specimen makes it impossible to apply any image-based subject-specific scaling technique.

The aim of this paper is to present a new comprehensive musculoskeletal geometry dataset of the lower extremity, based on medical images and dissection measurements of a single cadaver specimen. This dataset, named Twente Lower Extremity Model 2.0,
consists of a coherent set of medical imaging data (CT and MRI), segmented bone, muscle and subcutaneous fat (including skin) volumes, coordinates of muscle attachment sites and lines-of-action, ligament attachment sites and lines-of-action, bony wrapping surfaces, and joint centers and axes of rotation. TLEM 2.0 was purposely built to be easily combined with image-based scaling techniques, in an attempt to accelerate the application of subject-specific models. The complete dataset is made freely available at $h t t p: / / w w w . u t w e n t e$. $\mathrm{nl} / \mathrm{ctw} / \mathrm{bw} /$ research/projects/TLEMsafe to the scientific community to be used for non-commercial use only, under acceptance of the TLEMsafe Research License Agreement.

## 2. Methods

### 2.1. Cadaver specimen

Measurements were performed on the right lower extremity of a fresh cadaver (male, white, age 85 years, estimated mass 45 kg ), with no clear pathologies affecting the musculoskeletal system. The leg length, measured from the anterior superior iliac spine to the medial malleolus, was 813 mm .

In the specimen we distinguished 6 segments: pelvis, femur, patella, tibia (including fibula), talus and foot (consisting of hindfoot, midfoot and phalanges). During the whole measurement session, the foot bones were fixed to each other and the foot was fixed to a wooden plate, in order to avoid internal movements.

### 2.2. Medical imaging

Prior to the dissection of the specimen, computed tomography (CT) and magnetic resonance images (MRI) of both lower extremities, from the most proximal extremity of the iliac crest to the most distal tip of the foot, were acquired at the Department of Radiology of the Radboud University Medical Center (Fig. 1A). For the CT, a Siemens SOMATOM ${ }^{\circledR}$ Sensation 16 Scanner (Siemens AG, Munich, Germany) was used, with voxel size of $0.977 \mathrm{~mm} \times 0.977 \mathrm{~mm} \times 0.75 \mathrm{~mm}$. For the MRI, T1 weighted axial spin echo (SE) scan was taken using a Siemens 3T MAGNETOM ${ }^{\mathbb{R}}$ Skyra (Siemens AG, Munich, Germany), with different slice thickness between series covering the joint regions ( 3 mm ) and series covering the shaft of femur and tibia ( 8 mm ), and an in-plane resolution of $1 \mathrm{~mm} \times 1 \mathrm{~mm}$. To improve the quality of the images and avoid crystallization damage of soft tissues, the scans were performed before freezing of the cadaver specimen.

### 2.3. Cadaver measurements

After thawing of the cadaver, a complete dissection of the lower extremity specimen was performed at the Department of Anatomy of the Radboud University Medical Center. The cadaver was divided at the level of L5, then the two lower extremities were separated. The right lower extremity specimen was not fixed in a specific position, so that segments and joints could be moved freely (except for the foot being fixed to a wooden plate) in order to facilitate the measurements. First, skin and subcutaneous fat were removed (Fig. 1B). Then, reference frames with retro-reflective markers were attached to the pelvis, femur, patella, tibia and foot segments. The Brainlab Kolibri ${ }^{\text {TM }}$ image-guided surgery platform (Brainlab AG, Munich, Germany) was used to measure the position of points in threedimensional space with respect to the corresponding reference frame fixed to the bones. This 3-D navigation system had a spatial accuracy of $0.231 \pm 0.137 \mathrm{~mm}$ (RMS $\pm$ SD) and an average orientation error of $0.383^{\circ}$ (Wiles et al., 2004).

### 2.3.1. Muscle attachment sites, lines-of-action, mass and volume

For each muscle, fat at the intermuscular connection was removed, resulting in muscles that were only connected to the bones at origin and insertion. After the identification, each muscle was excised and contours of its origin and insertion were measured with the Brainlab Kolibri ${ }^{\mathrm{TM}}$ system (Fig. 1C). The number of points measured to define each muscle attachment site depended on its shape and size. In total, 55 muscle actuators were analyzed, and 98 muscle-tendon attachment sites were measured. In case of a curvature of the muscle line-of-action, via point and underlying bone contours were measured. Then, tendon, remaining fat and excessive connective tissue were removed from the dissected muscle. Muscle weight was measured using a scale with an accuracy of 1.0 g . Muscle volumes were measured using the water dislocation method, using a scaled cylinder with an accuracy of 1.0 ml .

### 2.3.2. Joint geometry

After removal of all muscles, but with ligaments still intact, geometrical behavior of hip, knee, patellofemoral, talocrural and subtalar joints were measured. Each joint was manipulated by hand, the movement being limited by bone contact


Fig. 1. Measurements performed on the cadaver specimen. A. CT scan (left) and MRI scan (right) of the lower extremities of the cadaver specimen, from the most proximal extremity of the iliac crest to the most distal tip of the foot. B. Right lower extremity specimen after removal of skin and subcutaneous fat. The specimen was not fixed in a specific position, so that segments and joints could be moved freely (except for the foot being fixed to a wooden plate) in order to facilitate the measurements. C. Dissection session using the Brainlab KolibriTM image-guided surgery platform (Brainlab AG, Munich, Germany). In this example, a reference frame with retro-reflective markers was attached to the femur, and coordinates of muscle attachment sites were measured in three-dimensional space with respect to the reference frame fixed to the bone. Frame attachment pins remained fixed throughout the measurement.
or ligaments. Throughout the complete joint range of motion, the position of three points on the bone surface of the distal segment was measured in the reference frame of the proximal segment.

### 2.3.3. Ligaments

Similarly to muscles, attachment sites and lines-of-action of 5 ligaments of the hip joint (ischiofemoral, iliofemoral medial, iliofemoral lateral, pubofemoral and ligament of the head of the femur) and 5 ligaments of the knee joint (tibial collateral, fibular collateral, anterior cruciate, posterior cruciate and patellar ligament) were measured.

### 2.3.4. Bony landmarks and bone surfaces

After all the ligaments and remaining soft tissues had been removed, the bones were separated and 22 bony landmarks were measured on the bone surface, based on the definition by the Standardization and Terminology committee of the International Society of Biomechanics (Wu et al., 2002). Finally, at least 100 additional registration points were collected on the complete surface of each bone in order to facilitate accurate registration to segmented bone surface later.

### 2.4. Post processing

### 2.4.1. Image post-processing

Bone surfaces were automatically segmented from CT into STL files and remeshed to obtain a higher resolution in regions with a high curvature. Muscle volumes were segmented from MRI using a semi-automatic registration technique. Subcutaneous fat and skin volumes were manually segmented from MRI. All the image segmentation processing was performed using Mimics ${ }^{\circledR} 17.0$ (Materialise N . V., Leuven, Belgium).

### 2.4.2. Registration

To register the cadaver measurements to the CT-based bone surface STL files, the iterative closest point method (Besl and Mckay, 1992) was used to minimize the sum of the squared errors (SSE) between the registration points and the closest points on the faces of the STL:
$\operatorname{SSE}=\sum_{i=1}^{\mathrm{n}}\left(\mathrm{y}_{\mathrm{i}}-\hat{\mathrm{y}}_{\mathrm{i}}\right)^{2}$
where $y_{i}$ represents the registration points, $\hat{\mathrm{y}}_{\mathrm{i}}$ represent the closest point on the face of the STL to $y_{i}$, and $n$ represents the number of registration points. To improve results, $5 \%$ of the worst registration points were rejected after the first 50 iterations. After registration, the measured bony landmarks and muscle attachment points were projected to the closest point on the face of the STL.

### 2.4.3. Local reference frames

For each segment, the following local reference frames were defined (see Fig. 2):

- Pelvis

O: the origin coincident with the right (or left) hip joint. Z: the line parallel to the line connecting the right and left anterior superior iliac spine, and pointing
to the right. X : the line parallel to a line lying in the plane defined by the two anterior superior iliac spines and the midpoint of the two posterior superior iliac spines, perpendicular to the $Z$-axis and pointing anteriorly. $Y$ : the line perpendicular to both $X$ - and $Z$-axis, pointing cranially.

- Femur

O : the origin coincident with the midpoint between the medial and lateral epicondyles of the femur. Y: the line connecting the origin and the hip joint, pointing cranially. Z : the line lying in the plane defined by the medial and lateral epicondyles of the femur and the hip joint, perpendicular to the $Y$-axis, pointing to the right. $X$ : the line perpendicular to both $Y$ - and $Z$-axis, pointing anteriorly.

- Patella

O : the origin coincident with the center of mass of the patella. $\mathrm{X}, \mathrm{Y}, \mathrm{Z}$ : coordinate system parallel to the coordinate system of the femur when the knee joint angle is equal to $0^{\circ}$, with position and orientation of the patella being estimated during cadaver dissection and using MRI.

- Tibia

O : the origin coincident with the midpoint between the tips of the medial and lateral malleoli. Y: the line connecting the midpoint between the tips of the medial and lateral malleoli, and the midpoint between the most medial point of the medial condyle of the tibia and the most lateral point of the lateral condyle of the tibia. Z: the line lying in the plane defined by the most medial point of the medial condyle of the tibia, the most lateral point of the lateral condyle of the tibia and the midpoint between the tips of the medial and lateral malleoli, perpendicular to the $Y$-axis, pointing to the right. X: the line perpendicular to both $Y$ - and $Z$-axis, pointing anteriorly.

- Talus

O : the origin coincident with the center of mass of the talus. $\mathrm{X}, \mathrm{Y}, \mathrm{Z}$ : coordinate system parallel to the coordinate system of the tibia when talocrural joint angle is equal to $0^{\circ}$, with position and orientation of the talus being estimated during cadaver dissection and using MRI.

- Foot

O : the origin coincident with the center of the subtalar joint. Y : the line perpendicular to the plane defined by the contact points of heel, first metatarsal and fifth metatarsal, pointing cranially. X : the line perpendicular to the $Y$-axis, pointing toward the contact point of the second metatarsal. Z: the line perpendicular to both $Y$ - and $X$-axis, pointing to the right.

### 2.4.4. Inertial parameters

Segment mass, center of mass, principal axes of inertia and principal moment of inertia were calculated for each segment, based on the segmented bone, muscle and fat volumes, using SolidWorks ${ }^{\mathbb{B}} 2013$ (Dassault Systèmes S.A., Vélizy-Villacoublay, France). The following average density parameters were used: bone $1500 \mathrm{~kg} / \mathrm{m}^{3}$ for bone, $1060 \mathrm{~kg} / \mathrm{m}^{3}$ for muscle and $900 \mathrm{~kg} / \mathrm{m}^{3}$ for fat. Inertial parameters were calculated with respect to the local reference frames defined above.

### 2.4.5. Modeling of muscle and ligament attachment sites and lines-of-action

To accurately describe their mechanical effect, muscle actuators were divided into a sufficient number of muscle-tendon elements, in accordance with the original TLEM dataset (Klein Horsman et al., 2007). The contours of the measured


Fig. 2. Local coordinate frames of the bone segments: A. Pelvis (ASIS: anterior superior iliac spine, PSIS: posterior superior iliac spine). B. Femur and patella (ME: medial epicondyle of the femur, LE: lateral epicondyle of the femur). C. Tibia and talus (MC: most medial point of the medial condyle of the tibia, LC: most lateral point of the lateral condyle of the tibia, MM: medial malleolus, LM: lateral malleolus). D. Foot (HC: heel contact point, 1C: first metatarsal contact point, 2C: second metatarsal contact point, 5C: fifth metatarsal contact point).
muscle attachment sites were modeled either as points, straight or curved lines, or areas, as described by Pellikaan et al. (2014); afterwards, all the modeled muscle attachment sites were projected to the closest node point of the bone surfaces STL.

In case of curved muscle line-of-action, when the muscle was not free to shift over the underlying structures, via points were defined based on the measured coordinates of the line-of-action, dividing the muscle in a series of straight line segments.

When a free shift of the muscle over the underlying structure (usually bone) was possible, cylindrical surfaces were defined to represent the bony contours, based on the measured muscle line-of-action and CT-based bone surfaces. Such wrapping surfaces were defined for gluteus maximus, iliopsoas, quadriceps femoris and gastrocnemius.

Similarly to the muscle-tendon elements, ligaments were modeled as straight lines and their attachment sites and via points modeled from the cadaver measurements.

### 2.4.6. Estimation of joint geometry

Hip rotation center was calculated based on a spherical fit through the trajectory of the femur with respect to the pelvis. Knee rotation center and axis were calculated based on a cylindrical fit through the trajectory of the tibia-fibula with respect to the femur. Patellofemoral rotation center and axis were calculated based on a cylindrical fit through the trajectory of the patella with respect to the femur. Talocrural rotation center and axis were calculated based on a cylindrical fit through the trajectory of the talus with respect to the tibia-fibula. Subtalar rotation center and axis were calculated based on a cylindrical fit through the trajectory of the foot with respect to the talus. The accuracy of the fitting was assessed with the average root mean squared error (RMSE) of the acquired data points to the fitted sphere or cylinder.

### 2.5. Musculoskeletal model

The obtained musculoskeletal geometry dataset was implemented in the AnyBody Modeling System ${ }^{\text {TM }}$ ver. 6.0.3 (AnyBody Technology A/S, Aalborg, Denmark). The muscle-tendon architecture dataset was adapted from the original TLEM dataset (Klein Horsman et al., 2007): nominal fiber lengths were individually scaled, comparing the total length of the muscle-tendon elements in the original TLEM and in the new TLEM 2.0 dataset; tendon slack lengths of each muscletendon element were then calculated to reproduce the relative sarcomere length as measured in the original TLEM dataset; physiological cross-sectional areas (PCSA) were calculated taking into account the scaled nominal fiber lengths, the nominal pennation angles, and the measured muscle volumes. Finally, the obtained musculoskeletal model of the lower extremity was integrated with the full-body model of the AnyBody Managed Model Repository ${ }^{\mathrm{TM}}$ ver. 1.6.4 (AnyBody Technology A/S, Aalborg, Denmark). This integration involved connection to the upper extremity spine model's geometry and muscles, using a set of morphing methods so that the pelvic geometry of the upper extremity models, arising from a different dataset, could fit with the pelvic geometry of TLEM 2.0

## 3. Results

The complete list of the measured muscle actuators is presented in Table 1. For each muscle actuator, the table contains the number of muscle-tendon elements representing that muscle actuator, the type of path line (straight line, passing through via points or curving around a wrapping surface), how the origin and
insertion sites were modeled (point, line, or surface), and the measured mass (g) and volume ( ml ). The dataset contains in total 55 actuators described by 166 muscle-tendon elements. In a similar way, Table 2 contains the list of the measured ligaments.

Segmentation of 6 bone segments (pelvis, femur, patella, tibia and fibula, talus, and foot), 55 muscle volumes, and subcutaneous fat (including skin) volumes were obtained from CT and MRI scans (Fig. 3A).

Inertial parameters (segment mass, center of mass, principal axes of inertia and principal moment of inertia) of each bone segment and coordinates of 22 bony landmarks, with respect to the relative local reference frame, are contained in Table A1 and Table A2, respectively.

Table A3 and A4 contain the coordinates of origin, insertion and via points of each muscle-tendon element and ligament, with respect to the relative local reference frame.

The geometrical description of the cylindrical wrapping surfaces used to represent the curved line-of-action of gluteus maximus, iliopsoas, quadriceps femoris and gastrocnemius muscles is contained in Table A5.

Table A6 contains the estimated joint rotation centers and axes expressed in the relative local reference frames. The average RMSE fitting errors were $0.86,2.52,1.83,2.30$ and 2.60 mm for the hip, knee, patellofemoral, talocrural and subtalar joint respectively.

Fig. 3B shows the final musculoskeletal model based on TLEM 2.0, implemented in the AnyBody Modeling System ${ }^{\mathrm{TM}}$ ver. 6.0.3. The model consists of 12 body segments: head-arms-trunk, pelvis, and right and left femur, patella, tibia, talus and foot. The model comprises 11 joints: L5S1 and left and right hip, knee, patellofemoral, talocrural and subtalar. The L5S1 and hip joints are modeled as a ball-and-socket, defined by a rotation center and three orthogonal axes. The knee, patellofemoral, talocrural and subtalar joints are defined as a hinge, with a fixed rotation center and axis. The patellar tendon is defined as a non-deformable element that connects the patella to the tibia, therefore the orientation and position of the patella depends solely on the knee flexion angle, without introducing an extra degree of freedom (DOF). The orientation and position of the center of mass of the pelvis with respect to a 3D global frame, together with the joint rotations of the L5S1, hip, knee, talocrural and subtalar joints, results in a model with 21 DOFs. The model contains 55 muscle actuators, described by 166 Hill-type elements. Nominal fiber length, tendon slack length, nominal pennation angle, and PCSA of each muscle-tendon element is presented in Table A7.

The complete TLEM 2.0 dataset is freely shared with the scientific community to be used for non-commercial use only. The complete Electronic Appendix (Tables A1-A7) and the

Table 1
List of muscle actuators analyzed: number of muscle-tendon elements representing the muscle actuator, type of the path line (straight line (S), passing through via points (VP) or curving around a wrapping surface (WS)), type of the origin and insertion sites (Point, Line (order), LineArea (order) or Area), mass (g) and volume (ml).

| Muscle | \# Elements | Type line | Origin | Insertion | Mass (g) | Volume (ml) |
| :---: | :---: | :---: | :---: | :---: | :---: | :---: |
| Adductor Brevis Distal | 2 | S | Line (2) | Line (1) | 20 | 20 |
| Adductor Brevis Mid | 2 | S | Line (2) | Line (1) | 20 | 20 |
| Adductor Brevis Proximal | 2 | S | Line (2) | Line (1) | 20 | 20 |
| Adductor Longus | 6 | S | LineArea (2) | Line (2) | 67 | 66 |
| Adductor Magnus Distal | 3 | S | Line (2) | Point | 183 | 181 |
| Adductor Magnus Mid | 6 | S | Line (2) | Line (2) | 106 | 102 |
| Adductor Magnus Proximal | 4 | S | Line (2) | Line (1) | 30 | 30 |
| Biceps Femoris Caput Breve | 3 | S | Line (2) | Point | 61 | 60 |
| Biceps Femoris Caput Longur | 1 | S | Point | Point | 116 | 111 |
| Extensor Digitorum Longus | 4 | VP | Line (2) | Point | 36 | 35 |
| Extensor Hallucis Longus | 3 | VP | Line (3) | Point | 17 | 16 |
| Flexor Digitorum Longus | 4 | VP | LineArea (3) | Point | 26 | 25 |
| Flexor Hallucis Longus | 3 | VP | Line (2) | Point | 30 | 30 |
| Gastrocnemius Lateralis | 1 | WS | Point | Point | 54 | 54 |
| Gastrocnemius Medialis | 1 | WS | Point | Point | 111 | 107 |
| Gemellus Inferior | 1 | S | Point | Point | 2 | - |
| Gemellus Superior | 1 | S | Point | Point | 2 | - |
| Gluteus Maximus Inferior | 6 | WS | Area | Line (1) | 336 | 316 |
| Gluteus Maximus Superior | 6 | WS | Area | Line (1) | 134 | 130 |
| Gluteus Medius Anterior | 6 | S | Area | Area | 77 | 75 |
| Gluteus Medius Posterior | 6 | S | Area | Area | 154 | 150 |
| Gluteus Minimus Anterior | 2 | S | Area | LineArea (2) | 26 | 26 |
| Gluteus Minimus Mid | 2 | S | Area | LineArea (2) | 26 | 26 |
| Gluteus Minimus Posterior | 2 | S | Area | LineArea (2) | 26 | 26 |
| Gracilis | 2 | S | Line (1) | Point | 58 | 58 |
| Iliacus Lateralis | 2 | WS | Area | LineArea (2) | 30 | 29 |
| Iliacus Medialis | 2 | WS | Area | LineArea (2) | 30 | 29 |
| Iliacus Mid | 2 | WS | Area | LineArea (2) | 30 | 29 |
| Obturator Externus Inferior | 2 | VP | Line (3) | Point | 15 | - |
| Obturator Externus Superior | 3 | VP | LineArea (1) | Point | 26 | - |
| Obturator Internus | 6 | VP | Area | Point | 32 | 31 |
| Pectineus | 4 | S | Line (2) | Line (3) | 38 | 37 |
| Peroneus Brevis | 3 | VP | Line (3) | Point | 20 | 19 |
| Peroneus Longus | 3 | VP | Line (3) | Point | 43 | 42 |
| Piriformis | 1 | S | Point | Point | 26 | 25 |
| Plantaris | 1 | WS | Point | Point | 6 | 5 |
| Popliteus | 3 | S | Line (2) | LineArea (3) | 19 | 18 |
| Psoas Major | 5 | WS | - | Point | - | - |
| Quadratus Femoris | 4 | S | Line (1) | Line (2) | 34 | 33 |
| Rectus Femoris | 2 | WS | Point | Line (2) | 118 | 114 |
| Sartorius | 1 | VP | Point | Point | 101 | 98 |
| Semimembranosus | 3 | S | Line (2) | Line (2) | 120 | 116 |
| Semitendinosus | 1 | S | Point | Point | 111 | 106 |
| Soleus Lateralis | 3 | S | Line (2) | Point | 150 | 146 |
| Soleus Medialis | 3 | S | Line (3) | Point | 82 | 80 |
| Tensor Fasciae Latae | 2 | S | Line (2) | Point | 33 | 34 |
| Tibialis Anterior | 3 | VP | LineArea (2) | Point | 77 | 75 |
| Tibialis Posterior Lateralis | 3 | VP | Line (2) | Point | 45 | 43 |
| Tibialis Posterior Medialis | 3 | VP | Line (2) | Point | 45 | 43 |
| Vastus Intermedius | 6 | WS | Area | Line (2) | 104 | 101 |
| Vastus Lateralis Inferior | 6 | WS | Line (2) | Line (3) | 84 | 84 |
| Vastus Lateralis Superior | 2 | WS | Line (2) | Point | 338 | 330 |
| Vastus Medialis Inferior | 2 | WS | Line (3) | Line (3) | 47 | 46 |
| Vastus Medialis Mid | 2 | WS | Line (3) | Line (3) | 87 | 88 |
| Vastus Medialis Superior | 4 | WS | Line (2) | Point | 111 | 112 |

segmented bone surfaces are available at http://www.utwente.nl/ ctw/bw/research/projects/TLEMsafe under acceptance of the TLEMsafe Research License Agreement. CT and MRI scans, and segmented muscle and subcutaneous fat (including skin) volumes are available upon request to be sent to TLEMsafe Project coordinator, Prof. Dr. Ir. Nico Verdonschot (n.verdonschot@utwente.nl), after approval of the TLEMsafe consortium.

## 4. Discussion

In this paper, we presented the Twente Lower Extremity Model 2.0, a new comprehensive musculoskeletal geometry dataset of the lower extremity. Most existing models have been based on one
or more cadaver studies to represent the musculoskeletal geometry of an average adult subject (Delp et al., 1990b; Klein Horsman et al., 2007; Arnold et al., 2010), but no prior dataset in the literature has been accompanied by detailed medical images and post-processing data. To the best of our knowledge, TLEM 2.0 represents the first consistent and most complete 'atlas' model, which includes a set of CT and MRI scans, segmented bone, muscle and subcutaneous fat (including skin) volumes, inertial parameters, coordinates of bony landmarks, muscle and ligaments attachment sites and lines-of-action, bony wrapping surfaces, and joint centers and axes of rotation.

TLEM 2.0 is freely shared so that the scientific community can utilize the presented dataset for their own research purposes, for instance to develop new personalization techniques, in order to help
to accelerate the development and adoption of subject-specific models on large scale. For this reason, data not used yet in the presented musculoskeletal model (such as fat and skin volumes, or ligament attachment sites) was also included in the shared dataset. Further data that was beyond the scope of this study, such as identification of muscle fiber direction, segmentation of articular cartilage and articular capsule, or a more sophisticated and realistic description of the knee joint and the foot model, could be also estimated in the future, in order to extend and improve the TLEM 2.0 dataset. Nevertheless, several limitations affect the presented dataset.

Table 2
List of ligaments analyzed: number of elements representing the ligament, type of the path line (straight line (S) or passing through via points (VP)), type of the origin and insertion sites (Point or Line (order)).

| Ligament | $\#$ <br> Elements | Type <br> line | Origin | Insertion |
| :--- | :--- | :--- | :--- | :--- |
| HIP |  |  |  |  |
| Ischiofemoral | 2 | S | Point | Line (2) |
| Iliofemoral medial | 3 | S | Point | Line (2) |
| Iliofemoral lateral | 3 | S | Point | Line (2) |
| Pubofemoral | 3 | S | Line (2) | Line (2) |
| Ligament of the head of the femur | 1 | S | Point | Point |
|  |  |  |  |  |
| KNEE | 1 |  |  |  |
| Tibial collateral | 1 | VP | Point | Point |
| Fibular collateral | 2 | S | Point | Point |
| Anterior cruciate | 2 | S | Point | Point |
| Posterior cruciate | 1 | S | Point | Point |
| Patellar ligament |  |  |  |  |

Firstly, medical images were taken with the cadaver in a supine position, resulting in a compression of muscles and other soft tissues in the gluteal region. Although unavoidable, we think that this phenomenon had low effect on the calculation of the total volume of muscle and fat tissue, and subsequently on the calculation of the inertial parameters of the pelvis segment. Furthermore, muscle lines-of-action and bony wrapping surfaces were measured during the cadaver dissection, and were not affected by tissue compression. However, we presume that future studies attempting to estimate muscle fiber direction and moment arms in the gluteal region, in particular for gluteus maximus, from the TLEM 2.0 datasets could yield inaccurate results.

Secondly, similarly to the original TLEM (Klein Horsman et al., 2007) and others lower extremity musculoskeletal geometry dataset in literature (Delp et al., 1990a), TLEM 2.0 is based on a single Caucasian white male cadaver. The wide inter-individual anatomical variability in size and shape of muscle attachment sites (White et al., 1989b; Duda et al., 1996) and the gender and ethnical variation (Kepple et al., 1998) reported in literature suggest that linear scaling of a model based on a single specimen may not be representative for individual applications. In these cases, imagebased subject-specific models that take into account non-linear differences are more advisable and could be used to create additional 'atlas' models in order to represent different age, gender or ethnical variations.

Another limitation of this dataset is that parameters of the muscle-tendon architecture were not measured on the cadaver specimen. This would have required a much longer timespan to perform the measurements, and was beyond the scope of this study. We were aware of the fact that inaccuracies in muscle-tendon


Fig. 3. A. Image-based segmentation using Mimics ${ }^{\circledR} 17.0$ (Materialise N.V., Leuven, Belgium). From left to right: bone surfaces and single muscle volumes, muscle volumes per segment, and subcutaneous fat and skin volumes per segment. B. TLEM 2.0 implemented in the AnyBody Modeling System ${ }^{\text {TM }}$ ver. 6.0 .3 (AnyBody Technology A/S, Aalborg, Denmark). The obtained model consisted of 12 body segments (head-arms-trunk, pelvis, and right and left femur, patella, tibia, talus and foot), 11 joints (L5S1 and left and right hip, knee, patellofemoral, talocrural and subtalar) and 21 DOFs.
parameters (in particular tendon slack length) can largely affect musculoskeletal model prediction (Scovil and Ronsky, 2006; Redl et al., 2007). For this reason, the new cadaver study was planned so that the resulting dataset was compatible with the original TLEM model, which represents the most complete and consistent muscle-tendon architecture dataset of the lower extremity to date. Moreover, muscle-tendon parameters were not simply scaled linearly with bone length, but they were individually adapted from the original TLEM to the new TLEM 2.0, taking into account the non-linear differences in bone size and muscle-tendon lengths, and preserving the originally measured relative sarcomere lengths, in order to guarantee consistency in muscle function between the two models.

It is important to note that TLEM 2.0 was not created with the main scope to be used as a generic musculoskeletal model, but was purposely built as a template to obtain subject-specific model. The advantage of TLEM 2.0 is that it can be easily combined with medical imaging scaling methods, allowing to create personalized musculoskeletal geometry, including better estimation of muscletendon total length, line-of-action and moment arm, that in turn can allow to obtain better estimation of muscle-tendon parameters. For instance, several scaling techniques were developed parallel to TLEM 2.0 within the TLEMsafe project. Pellikaan et al. (2014) used a morphing based method to estimate the muscle attachment sites of the lower extremity, based on TLEM 2.0 and a second cadaver dissection dataset, showing that for $69 \%$ of the muscle attachment sites the estimation error was smaller than 15 mm , and that the largest errors affected only the least sensitive attachment sites. Then, Carbone et al. (2013) combined TLEM 2.0 with morphing of bone surfaces, non-rigid registration of muscle volumes and functional optimization of muscle-tendon architecture in a streamlined modeling workflow, showing that subject-specific models resulted in more reliable outcome, while conventional anthropometric scaling laws were inadequate and caused unrealistic muscle activity predictions. Furthermore, the combination of patient-specific joint and muscle forces models with geometrically consistent bone geometry into finite element analyses is expected to be essential in the near future for predicting the individual functional outcome of patient treatments, allowing for example to obtain a better prediction of bone density remodeling and healing (Vahdati et al., 2014), or individualized predictions of fracture risk or peri-prosthetic micromotions (van der Ploeg et al., 2012). Moreover, the techniques applied to obtain personalized musculoskeletal models could also be applied to develop a population of subject-specific models to be used in statistical shape modeling of bone geometry (Baldwin et al., 2010).

However, obtaining personalized models that accurately reproduce the musculoskeletal system and the force-generating characteristics of a subject represents only one of several aspects to consider when aiming at reliable model predictions. For example, inverse dynamics-based simulations are sensitive to inaccuracies in the measured kinematic and kinetic and data (Pàmies-Vilà et al., 2012), and the resulting dynamic inconsistency can lead to unrealistic model predictions (Kuo, 1998). Deriving the force plates data from three-dimensional full-body motion (Robert et al., 2013; Fluit et al., 2014a) represents a promising technique to both improve the dynamic consistency as well as remove the model's dependency on measured external forces. Moreover, for individual applications such as prediction of functional outcome after a complex orthopedic surgery, kinematic data of the patient are missing and using pre-recorded measurements from different subjects would lead to obvious inconsistency. Many forwarddynamics methods to have been developed in recent years to predict gait movements (Fluit et al., 2012; Wang et al., 2012), but their complexity and large computational cost prevented their application in a clinical setting. To deal with this restriction, recently Principal Component Analysis (PCA) has been proposed
to interpret and evaluate gait data (Daffertshofer et al., 2004) and predict new gait movements (Safonova et al., 2004; Fluit et al., 2014b), by eliminating dependency on measured kinematic input data. We expect such techniques to evolve in the near future, increasing our confidence in the individual predictions of musculoskeletal models, and we believe that a consistent and comprehensive dataset like TLEM 2.0 represents the ideal foundation for such complex applications.

## Conflict of interest statement

The authors do not have any financial or personal relationships with other people or organization that could inappropriately influence their work.

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