The Lower Limb Extremity Model 2010 (Arnold 2009) is a three-dimensional computer model of the lower limb extremity that can be examined and analyzed in OpenSim, a freely available biomechanics simulation application. The model includes geometric representation of the bones, kinematic descriptions of the joints, and Hill-type models for 44 muscle-tendon compartments in the lower limb extremity.

The Lower Limb Extremity Model 2010 builds on the lower limb model developed by Scott Delp et al. (1990) and refined by Allison Arnold et al. (2000). Compared to preceding models of the lower extremity, the Lower Limb Extremity Model 2010 more accurately reflects muscle fiber operating lengths and force generation properties of the lower limb muscles. It does so by incorporating muscle architecture data collected by Ward et al. (2009) for 21 adult cadavers. Using the Ward dataset means the model is based on a cohesive set of experimentally measured data that is not pieced together from separate sources. Furthermore, the number of subjects in the dataset is large enough to make the Lower Limb Extremity Model 2010 a generic lower extremity model.

The Lower Limb Extremity Model 2010 can be utilized to calculate muscle-tendon lengths and moment arms over a wide variety of body positions, and to examine of the force and moment generation capacities of the muscles about the ankle, knee, and hip.

Ellipsoidal wrapping surfaces are extensively used to define muscle-tendon paths that are constrained by bones, deeper muscles, and retinacula in the Lower Limb Extremity Model 2010. Employing these wrapping surfaces allows the model to reflect more accurately operating lengths and force generation properties of lower limb muscles. Unfortunately, it also makes dynamic simulation with the Lower Limb Extremity Model 2010 significantly slower than that with the Gait 2392 and 2354 Models and Full Body Running Model model. Due to this higher computational cost, the Lower Limb Extremity Model 2010 is currently unsuitable for Computed Muscle Control (CMC).

3 versions of the Lower Limb Extremity Model 2010 are currently available for download. Two of these include a model with 2 legs and a torso with the mass properties of a composite head-arms-torso (HAT). In the fifty-percent-stronger version of the two, the maximum isometric force for all lower limb muscles is increased by 50% to make the model strong enough for a CMC simulation of walking for an average-sized, healthy, adult male. The third version of the Lower Limb Extremity Model 2010 replaces the composite head/arms/torso body with a head/torso and torque actuated multi-body arms (Arnold 2010).

See the sections below for more information about the components of the model:

- Kinematics
  - Bone geometry
  - Joint geometry
  - Muscle Geometry
- Dynamics
  - Inertial Properties
  - Actuators and Other Force-Generating Elements
- Model testing
- References

## Kinematics

### Bone geometry

The default version has 14 body segments representing the limb’s skeletal geometry including the torso, the pelvis, and left and right femur, patella, tibia, calcaneus, talus, and toes. The dimensions of the segments are consistent with those of a 170-cm-tall adult male.

The rigid models for the bony segments featured in this model are adapted from A.S. Arnold et al. (2000), who contributed many modifications to the original Delp model (1990) to improve its accuracy. Noteworthy modifications include:

- Re-orientation of the right and left hemi-pelves
  In this model, spatial relationship between the hemipelves and sacrum are more consistent with the pelves constructed from MR images.

- Re-digitalization of the femur
  The femur from Delp’s model (Specimen A55, Stanford University) is re-digitized at a higher resolution.

- Replacement of the tibia
  The tibia in Delp’s model doesn’t match the Specimen A55 femur. The tibia from Specimen A55 is digitized and used to replace the tibia in the Delp model.

- Re-orientation and scaling of the fibula
  The fibula is scaled by 97% from that in the Delp model.

- Revision of the pelvis coordinate system
  In this model, the “neutral position” is such that the ASISs and pubic tubercle lie in the frontal plane. In Delp’s model, the “neutral position” is the anatomical position. This difference yields an approximate 5-degree offset in pelvis tilt and hip flexion angle. Because this may
Joint geometry

The model includes the metatarsophalangeal, subtalar, ankle, knee, hip, pelvis, and lumbar joints. Adapted from the original Delp model, the joints included are modeled as follows:

1. The metatarsophalangeal joint
   The metatarsophalangeal joints are modeled as revolute joints, with the axes defined by Delp based on Inman (1976). The metatarsophalangeal joint axis is rotated -8° around a vertical axis from the description by Inman, and its range is -30° (extension) to 30° (flexion).

2. The subtalar joint
   Similar to the metatarsophalangeal joints, subtalar joints are modeled as revolute joints, with the axes defined by Delp based on Inman. The subtalar range is -20° (eversion) to 20° (inversion).

3. The ankle joint
   The ankle is modeled as a revolute joint between the tibia and the talus, and is defined by one degree of freedom (dorsiflexion/plantarflexion), which has a range of -40° (plantarflexion) to 20° (dorsiflexion).

4. The knee joint
   The knee joint has one degree of freedom (flexion/extension). In the original Arnold model, it is modeled using the equations reported by Walker et al. (1988) for anterior-posterior and medial-lateral translation, and internal-external and varus-valgus rotation. The knee angle ranges from 0° (full extension) to 100° (flexion).

Arnold et al. (2000), Buford et al. (1997), and Grood et al. (1984) have tested this model by comparing the moment arms of the knee muscles to those measured in cadaver subjects.

In an update released on April 12th, 2011, the patellar joint and its corresponding coupler constraints are altered to rectify a problem with coordinate coupler constraints that occurs during scaling in OpenSim. Prior to the fix, translational coordinate coupler constraints are scaled incorrectly in the OpenSim 2.2 Scale tool. This affects the constraints that are used to define patellar motion as a function of knee_angle_l and knee_angle_r. The issue is rectified by replacing the three coordinates defining rotational and translational motion in the left and right patella bodies with a single rotational coordinate in each, knee_angle_beta_* , and coupling the new coordinates to equal knee_angle_* .

5. The hip joint
   The hip joint is modeled as a ball and socket joint with three degrees of freedom: flexion/extension, adduction/abduction, and internal/external rotation. The joint ranges are -20° (extension) to 90° (flexion), -40° (abduction) to 10° (adduction), and -40 (external rotation) to 40 (internal rotation).

6. The pelvis joint
   The pelvis joint has 6 degrees of freedom (3 translational: x, y, z and 3 rotational: tilt, list, and rotation).

7. The lumbar joint
   The lumbar joint has 3 degrees of freedom: extension, bending, and rotation. The joint ranges are -90° to 90° for all degrees of freedom.

Figure 1 illustrates the coordinate systems defined in the lower limb extremity.
The coordinate systems of the bone segments. The systems are oriented so that when all joint angles are 0° the x-axes points anteriorly, the y-axes points superiorly, and the z-axes points to the right (laterally for the right leg). The joints in the model are defined as translations and rotations between these coordinate systems. (Arnold 2010)

Figure 1: The coordinate systems of the bone segments. The systems are oriented so that when all joint angles are 0° the x-axes points anteriorly, the y-axes points superiorly, and the z-axes points to the right (laterally for the right leg). The joints in the model are defined as translations and rotations between these coordinate systems. (Arnold 2010)

In general, the coordinate systems of the bone segments are defined such that in anatomical position, the x-axis points anteriorly, the y-axis superiorly, and the z-axis to the right.

Adapted from the original Delp model, the locations of the coordinate systems are as follows:

- **Calcaneus:** the calcaneus coordinate system is located at the most inferior, lateral point on the posterior surface of the calcaneus.
- **Toe:** the toe coordinate system is located at the distal end of the second metatarsal.
- **Talus:** the talus coordinate system is located at the midpoint of the line between the apices of the medial and lateral maleoli.
- **Tibia:** the tibia coordinate system is fixed in the tibia and located at the midpoint of the femoral condyles with the knee in full extension.
- **Patella:** the patella coordinate system is located at the distal pole of the patella.
- **Femur:** the femur coordinate system is located at the center of the femoral head.
- **Pelvis:** the pelvis coordinate system is located at the midpoint of the left and right anterior superior iliac spines (ASIS) so that the two ASISs and pubic tubercles were in the frontal (y–z) plane.

The bone geometry for the torso is adapted from the Gait_2392 model (see Gait 2392 and 2354 Models). The arms included in the version with simple arms are adapted from the Full Body Running Model developed by Samuel Hamner et al. (2010).

**Muscle Geometry**

The model includes 35 muscles of the lower limb. In the case of muscles with complex geometry, such as broad attachments, multiple muscle paths are used (e.g., gluteus maximus), resulting in 44 muscle–tendon compartments.

Table 1 lists the muscle-tendon compartments in the lower limb contained in the model and their abbreviations.
Table 1: List of muscles and their abbreviations (Arnold 2010)

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Abbreviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adductor brevis</td>
<td>addbrev</td>
</tr>
<tr>
<td>Adductor longus</td>
<td>addlong</td>
</tr>
<tr>
<td>Adductor magnus c,d</td>
<td></td>
</tr>
<tr>
<td>Adductor magnus distal</td>
<td>addmagDist</td>
</tr>
<tr>
<td>Adductor magnus ischial</td>
<td>addmagIsch</td>
</tr>
<tr>
<td>Adductor magnus middle</td>
<td>addmagMid</td>
</tr>
<tr>
<td>Adductor magnus proximal</td>
<td>addmagProx</td>
</tr>
<tr>
<td>Biceps femoris long head</td>
<td>bfh</td>
</tr>
<tr>
<td>Biceps femoris short head</td>
<td>bfsh</td>
</tr>
<tr>
<td>Extensor digitorum longus</td>
<td>edl</td>
</tr>
<tr>
<td>Extensor hallucis longus</td>
<td>ehl</td>
</tr>
<tr>
<td>Flexor digitorum longus</td>
<td>fdl</td>
</tr>
<tr>
<td>Flexor hallucis longus</td>
<td>fhl</td>
</tr>
<tr>
<td>Gastrocnemius lateral head</td>
<td>gaslat</td>
</tr>
<tr>
<td>Gastrocnemius medial head</td>
<td>gasmed</td>
</tr>
<tr>
<td>Gemelli d</td>
<td>gem</td>
</tr>
<tr>
<td>Gluteus maximus c,g</td>
<td></td>
</tr>
<tr>
<td>Gluteus maximus superior</td>
<td>glmax1</td>
</tr>
<tr>
<td>Gluteus maximus middle</td>
<td>glmax2</td>
</tr>
<tr>
<td>Gluteus maximus inferior</td>
<td>glmax3</td>
</tr>
<tr>
<td>Gluteus medius g,f</td>
<td></td>
</tr>
<tr>
<td>Gluteus medius anterior</td>
<td>glmed1</td>
</tr>
<tr>
<td>Gluteus medius middle</td>
<td>glmed2</td>
</tr>
<tr>
<td>Gluteus medius posterior</td>
<td>glmed3</td>
</tr>
<tr>
<td>Gluteus minimus e</td>
<td></td>
</tr>
<tr>
<td>Gluteus minimus anterior</td>
<td>glmin1</td>
</tr>
<tr>
<td>Gluteus minimus middle</td>
<td>glmin2</td>
</tr>
<tr>
<td>Gluteus minimus posterior</td>
<td>glmin3</td>
</tr>
<tr>
<td>Gracilis</td>
<td>grac</td>
</tr>
<tr>
<td>Iliacus</td>
<td>iliacus</td>
</tr>
<tr>
<td>Pectineus h</td>
<td>pect</td>
</tr>
<tr>
<td>Peroneus brevis</td>
<td>perbrev</td>
</tr>
<tr>
<td>Peroneus longus</td>
<td>perlong</td>
</tr>
<tr>
<td>Peroneus tertius e</td>
<td>pertert</td>
</tr>
<tr>
<td>Piriformis e</td>
<td>piri</td>
</tr>
<tr>
<td>Psoas</td>
<td>psoas</td>
</tr>
<tr>
<td>Quadratus femoris e</td>
<td>quadfem</td>
</tr>
<tr>
<td>Rectus femoris</td>
<td>recfem</td>
</tr>
<tr>
<td>Sartorius</td>
<td>sart</td>
</tr>
<tr>
<td>Semimembranosus</td>
<td>semimem</td>
</tr>
<tr>
<td>Semitendinosus</td>
<td>semiten</td>
</tr>
<tr>
<td>Soleus</td>
<td>soleus</td>
</tr>
<tr>
<td>Tensor fascia latae e</td>
<td>tfl</td>
</tr>
<tr>
<td>Tibialis anterior</td>
<td>tibant</td>
</tr>
<tr>
<td>Tibialis posterior</td>
<td>tibpost</td>
</tr>
<tr>
<td>Vastus intermedius</td>
<td>vasint</td>
</tr>
<tr>
<td>Vastus lateralis</td>
<td>vaslat</td>
</tr>
<tr>
<td>Vastus medialis</td>
<td>vasmed</td>
</tr>
</tbody>
</table>

The muscle architecture for the torso is adapted from the Gait_2392 model (see Gait 2392 and 2354 Models), and the muscle architecture for the arms is adapted from the Full Body Running Model developed by Samuel Hamner et al. (2010) (see Full Body Running Model).

In modeling the muscle architecture, line segments are used to approximate the muscle–tendon path from origin to insertion. Wrapping surfaces and via points define muscle–tendon paths that are constrained by bones, deeper muscles, or retinacula.

In the Lower Limb Extremity Model 2010, many new wrapping surfaces and via points are added to the Delp model to increase its physiological...
accuracy in representing muscle paths and moment arms. Currently, the model employs 17 wrapping surfaces per limb: 3 at the hip, 1 at the femur, 5 at the tibia, and 8 at the shaft axis. These additions are a combination of surfaces adapted from A.S. Arnold et al. (2000) and new surfaces.

With respect to the original Delp model, the Lower Limb Extremity Model 2010 also contains the following notable modifications in muscle geometry:

- More accurate fiber length-joint angle relationship
  Ward et al. explicitly link measured fiber lengths to joint angles. This allows examination of the fiber length operating range in relation to the joint angle.

- Experimentally based tendon lengths
  In Delp’s model, tendon lengths are set so that resultant passive moments match experimental results. In this model they are set using an experimentally measured relationship between fiber length and joint angle.

The model offers two options for semimembranosus tendon length:

a. 0.348 is representative of the experimental joint angle-fiber length relationship, but results in unrealistic passive forces at high hip flexion. This value is the default as it is most true to the architecture data.

b. 0.378 gives more realistic passive behaviors in high hip flexion angles, but fibers become very short in less extreme configurations (like walking).

- Longer plantarflexor fibers
  In Delp’s model, the fiber lengths of the soleus and the gastrocnemii are lengthened beyond the data reported by Wickiewicz et al (1983). This was necessary to achieve physiologically reasonable results. The plantarflexor fiber lengths measured by Ward et al. were longer and thus lengthening was not necessary.

**Dynamics**

**Inertial Properties**

The inertial parameters for the body segments in the model are adopted from a 10-segment, 23 degree-of-freedom model developed by Frank C. Anderson and Marcus G. Pandy (1999). In the Anderson and Pandy model, mass and inertial properties for all segments, except the hindfeet and toes, are based on average anthropometric data obtained from five subjects (age 26 3 years, height 177 3 cm, and weight 70.1 7.8 kg). All data are recorded according to the method described by McConville et al. (1980). The lengths of the body segments are taken from the Delp model (1990).

For the hindfoot and toes, the mass, position of the center of mass, and moments of inertia are found by representing the volume of each segment by a set of interconnected vertices, the coordinates of which are derived from measuring the surface of a size-10 tennis shoes. Assuming a uniform density of 1.1 g/cm$^3$ for the feet, the density is numerically integrated over the volume of each segment to find the mass.

It should be noted that the mass of the torso varies between the version with arms and the version without. The mass of the torso in the version without arms correlates to the mass of the head-arm–torso complex in the original Anderson and Pandy model, and is, thus, heavier. In the version with arms, the mass of the torso is reduced to compensate for the added mass for the arms. The arms contain mass properties adapted from de Leva (1996) and are driven by torque actuators.

All inertial parameters for the model are scaled by a factor of 1.05626 from those reported by Anderson and Pandy (1999). Table 2 summarizes the mass and moments of inertia for each body segment in the Lower Extremity Model 2010.

<table>
<thead>
<tr>
<th>Body segment</th>
<th>Mass (kg)</th>
<th>Moments of inertia</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>xx</td>
<td>yy</td>
</tr>
<tr>
<td>Torso</td>
<td>34.2366</td>
<td>1.4745</td>
</tr>
<tr>
<td></td>
<td>26.8266</td>
<td>1.4745</td>
</tr>
<tr>
<td>Pelvis</td>
<td>11.777</td>
<td>0.1028</td>
</tr>
<tr>
<td>Right femur</td>
<td>9.3014</td>
<td>0.1339</td>
</tr>
<tr>
<td>Right tibia</td>
<td>3.7075</td>
<td>0.0504</td>
</tr>
<tr>
<td>Right patella</td>
<td>0.0862</td>
<td>0.00000287</td>
</tr>
<tr>
<td>Right talus</td>
<td>0.1000</td>
<td>0.0010</td>
</tr>
<tr>
<td>Right calcaneus</td>
<td>1.250</td>
<td>0.0014</td>
</tr>
<tr>
<td>Right toe</td>
<td>0.2166</td>
<td>0.0001</td>
</tr>
</tbody>
</table>
### Table 2: Inertial parameters for the body segments included in the model

* These segments are included only in the version with simple arms.

** The top number corresponds to the value used in the version without arms. The bottom number corresponds to the value used in the version with arms.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Left femur</th>
<th>Left tibia</th>
<th>Left patella</th>
<th>Left talus</th>
<th>Left calcaneus</th>
<th>Left toe</th>
<th>Right humerus</th>
<th>Right ulna</th>
<th>Right radius</th>
<th>Right hand</th>
<th>Left humerus</th>
<th>Left ulna</th>
<th>Left radius</th>
<th>Left hand</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>9.3014</td>
<td>3.7075</td>
<td>0.0862</td>
<td>0.1000</td>
<td>1.250</td>
<td>0.2166</td>
<td>2.0325</td>
<td>0.6075</td>
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<td></td>
<td>0.1339</td>
<td>0.0504</td>
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<td>0.0010</td>
<td>0.0014</td>
<td>0.0001</td>
<td>0.011946</td>
<td>0.002962</td>
<td>0.002962</td>
<td>0.000892</td>
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<tr>
<td></td>
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<td>0.0051</td>
<td>0.000001311</td>
<td>0.0010</td>
<td>0.0039</td>
<td>0.0002</td>
<td>0.004121</td>
<td>0.000618</td>
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<tr>
<td></td>
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<td>0.001311</td>
<td>0.0010</td>
<td>0.013409</td>
<td>0.003213</td>
<td>0.003213</td>
<td>0.00134</td>
<td>0.013409</td>
<td>0.003213</td>
<td>0.003213</td>
<td>0.00134</td>
</tr>
</tbody>
</table>

### Actuators and Other Force-Generating Elements

The isometric force-generating properties for each muscle-tendon actuator modeled are derived from scaling a generic Hill-type model. To scale the generic model, for each muscle-tendon compartment, four parameters are supplied: peak isometric muscle force, optimal muscle-fiber length, pennation angle, and tendon slack length. The specific values for each of these parameters are obtained from the measurements made in 21 cadaver subjects by Ward et al. (2009). The average age of the subjects (12 female and 9 male) in Ward’s study is 82.5 +/- 9.42. Parameters for the six small muscles not studied by Ward et al. (gemelli, gluteus minimus, peroneus tertius, piriformis, quadratus femoris, and tensor fascia latae) are adapted from the model described by Delp et al. (1990).

1. **Optimal fiber length and pennation angle**
   - Optimal fiber lengths and pennation angles for the muscles modeled are obtained from direct measurements on the cadavers. For gluteus medius, one of the two big muscle groups represented by multiple muscle-tendon compartments in the model, the physical locations of the fiber measurements performed by Ward et al. (2009) do not match the lines of action specified in the model. To minimize simulation error, average optimal fiber length and average pennation angle are used instead to represent each of the three muscle-tendon compartments.

2. **Maximum isometric force**
   - Maximum isometric force is calculated from measured PCSA and a specific tension of 61 N/cm² for all muscles. This specific tension is higher than the range of values (11 – 47 N/cm²) reported previously by Fukunaga et al. (1996) from magnetic resonance studies on 8 male subjects, but is identical to the value used by Delp et al. in an earlier model (1990). The adjustment is to compensate for age-related muscle atrophy and further atrophy in the cadavers.

3. **Tendon slack length**
   - Tendon slack length is based on the measured relationship between fiber length and joint position. Ward et al. (2009) measure fiber lengths and sarcomere lengths from subjects at an average position of 7° hip extension, 2° hip abduction, 0° knee flexion, and 40° plantarflexion.

   - From defined muscle-tendon paths, tendon slack length is calculated for each muscle-tendon compartment. This method works well for all muscles except for the following two exceptions:
     - **The Ankle**
       - In the ankle group, the resultant passive forces are physiologically unreasonable (i.e., passive forces were excessive). This is likely a result of a mismatch between the high degree of plantarflexion in the cadaver ankles and less extreme lengths at which the muscles are fixed. To adjust for this, tendon slack lengths for all muscles crossing the ankle are based on a joint angle of 20° plantarflexion.
     - **The semimembranosus**
       - The semimembranosus fibers are the shortest among all hamstring muscles. The tendon length calculation method used predicts unrealistically long fibers in the semimembranosus when the hip flexes and the knee extends. This discrepancy is corrected by increasing tendon slack length in the model.
Model testing

Arnold et al. (2009) has tested the accuracy of the muscle paths in the lower limb extremity by qualitatively comparing model predicted moment arms to experimentally measured moment arms. The maximum isometric joint moments predicted by the model do not exactly match experimental measurements of joint moments. Although it is possible to obtain a closer fit by varying parameters such as tendon slack lengths and PCSA to tune the model, doing so would sacrifice one of the strengths of the model: that it is based on a cohesive set of experimentally measured data.

As of June 2012, The Lower Limb Extremity Model 2010 has been cited in 48 publications. Researchers have been using the model for a variety of applications, from developing a musculoskeletal model for the lumbar joint (Christophy et al. 2012) to evaluating different knee models (Sandholm et al. 2011). An up-to-date and complete list of the publications citing the model can be obtained from Google Scholar.

References

An overview of OpenSim and its input file structure:


Publications specifying how the kinematic and dynamic properties of the model are defined:


Delp, Surgery Simulation: A computer graphics system to analyze and design musculoskeletal reconstructions of the lower extremity, Ph.D.Dissertation, Stanford University, 1990.


Publications featuring the experimental data from which the model is developed:


Publications that test different features of the model:
