Gait 2392 and 2354 Models

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The Gait 2392 Model and Gait 2354 Model are three-dimensional, 23 degree-of-freedom computer model of the human musculoskeletal system. The models were created by Darryl Thelen, Univ. of Wisconsin-Madison, and Ajay Seth, Frank C. Anderson, and Scott L. Delp, Stanford University. The models feature lower extremity joint definitions adopted from Delp et al. (1990), low back joint and anthropometry adopted from Anderson and Pandy et al. (1999), and a planar knee model adopted from Yamaguchi and Zajac etal. (1989).

The Gait 2392 model features 92 musculotendon actuators to represent 76 muscles in the lower extremities and torso. For the Gait 2354 model, the number of muscles was reduced by Anderson to improve simulation speed for demonstrations and educational purposes. Seth removed the patella to avoid kinematic constraints; insertions of the quadrucepts are handled with moving points in the tibia frame.

The default, unscaled version of the models a subject that is about 1.8 m tall and weighs 75.16 kilograms.

The models can be used and modified in OpenSim, an open source biomechanics simulation application. Some of the uses of the models include:

1. Computing the maximum isometric force and joint moment a muscle can develop at any body position
2. Studying how surgical changes in musculoskeletal geometry (e.g. origin-to-insertion path) and muscle-tendon parameters (e.g. optimal muscle-fiber length and tendon slack length) can affect the moment-generating capacity of the different muscles on the human body
3. Generating muscle drive forward simulations of walking and running to analyze how muscles contribute to motions (e.g. Induced Acceleration Analysis) or how joints are loaded (see Joint Reactions Analysis).

See the sections below for more information about the following components of these models:

- Gait 2392 and 2354 Models
    - Accessing the Models
    - Kinematics
        - Bone geometry
        - Joint geometry
        - Muscle geometry
    - Dynamics
        - Inertial properties
        - Actuators and Other Force-Generating Elements
    - Associated Publications

Accessing the Models

The musculoskeletal file (.osim), the setting files (.xml), and associated result files (.mot, .sto) for this model are provided free of charge with the OpenSim software for researchers interest in reproducing the result of the simulation. These files can be accessed via the **Models/Gait2392_Simbody or Models/Gait2354_Simbody** folder in the OpenSim 3.0 installation directory, and the **example/Gait2392_Simbody or Models/Gait2392_Simbody** folder in the OpenSim 2.4.0 installation directory.

Kinematics

Bone geometry

Bones surface data for the pelvis and the thigh are obtained by first marking the surfaces of bones with a mesh of polygons, and then determining the coordinates of the vertices with a three-dimensional digitizer. Data describing the shank and foot bones are adopted from Stredney et al (1982).

Joint geometry

The lower extremity has seven right-body segments: pelvis, femur, patella, tibia/fibula, talus, foot (which includes the calcaneus, navicular, cuboid, cuneiforms, metatarsals), and toes. Reference frames are fixed in each segment.

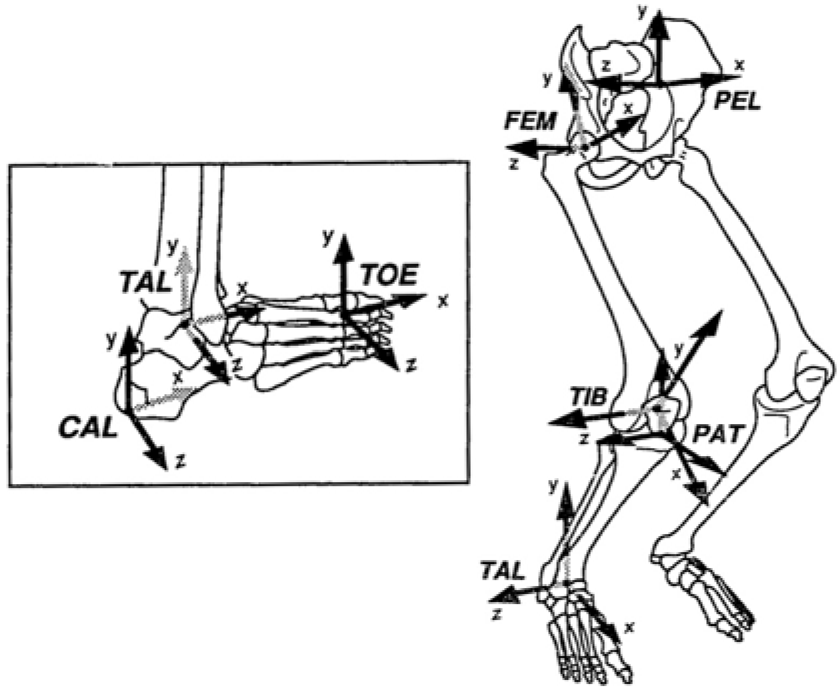

Figure 1 Location of the body-segmental reference frames (Delp et al., 1990).

- **Pelvis**: The pelvic reference frame is fixed at the midpoint of the line connecting the two anterior superior iliac spines
- **Femur**: The femoral frame is fixed at the center of the femoral head
- **Tibia**: The tibial frame is located at the midpoint of the line between the medial and lateral femoral epicondyles
- **Patella**: The patellar frame is located at the most distal point of the patella
- **Talus**: The talar frame is located at the midpoint of the line between the apices of the medical and lateral malleoli
- **Calcanus**: The calcaneal frame is located at the most interior, lateral point on the posterior surface of the calcanus
- **Toe**: The toe frame is located at the base of the second metatarsal

Models of the hip, knee, ankle, subtalar, and metatarsophalangeal joints define the relative motions of these segments.

Hip Joint

The hip is characterized as a ball-and-socket joint. The transformation between the pelvic and femoral reference frame is thus determined by successive rotations of the femoral frame about three orthogonal axes fixed in the femoral head.

Knee Joint

Because of its three-bone, multi-ligamented structure, the knee presents a challenge for the determination of the moment arm of the quadriceps muscles. In order to calculate the extensor moment arm of the knee in a computationally inexpensive way, Yamaguchi et al. (1989) developed a simplified model of the knee. The single-degree-of-freedom model provided by Yamaguchi et al. accounts for the kinematics of both the tibiofemoral joint and the patellafemoral joint in the sagittal plane as well as the patellar levering mechanism. Delp et al. adopted this planar knee model and specified the transformations between the femoral, tibial, and patellar reference frames as functions of the knee angle. Figure 2 illustrates how the planar knee model is adopted in the Delp model of lower limb extremity (1990). In the Delp model, the femoral condyles are represented as ellipses, and the tibial plateau is represented as a line segment. The transformation from the femoral reference frame to the tibial reference frame is specified such that the femoral condyles remain in contact with the tibial plateau throughout the range of knee motion. The tibiofemoral contact point depends on the knee angle and is specified according to data reported by Nisell *et al.* (1986).

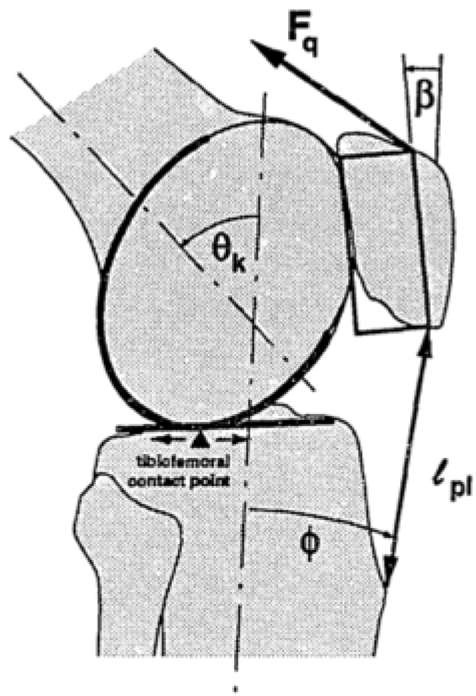

Figure 2: Geometry for determining knee moments and kinematics in the sagittal plane in the Delp model (Delp et al., 1990)

Ajay Seth adapted the Delp model, removing the patella to avoid kinematic constraints. In the Gait 2392 and Gait 2354 models, the insertions of the quadriceps on the tibia are modeled as moving points in the tibial frame.

Ankle, subtalar, and metatarsophalangeal joints

The ankle, subtalar, and metatarsophalangeal joints are modeled as frictionless revolute joints (as seen in **Figure 3**).

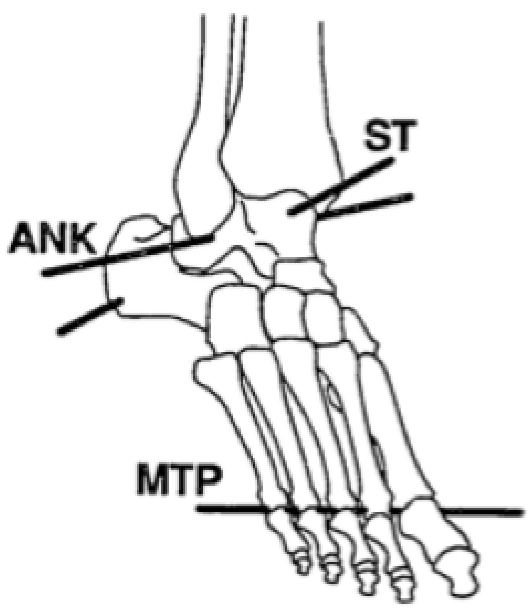

Figure 3. The ankle, subtalar, and metatarphalangeal joints are modeled as revolute joints with axes oriented as shown. (Delp et al., 1990)

The location and orientation of the axes for each of the joints are modeled after the descriptions provided by Inman (1976), with one modification. When displayed, the axes produce realistic motion of the ankle and subtalar joints (i.e. the bone surface models do not collide or disarticulate), but exhibit unrealistic motion of the metatarsophalangeal joint (i.e. the phalanges separate from the metatarsals). To fix this problem, the metatarsophalangeal axis is rotated by – 8 degree on a right-handed vertical axis to minimize disarticulation of the joint.

Muscle geometry

The paths (i.e. the lines of action) of the muscle-tendon actuators in the lower extremity portion of the model are defined based on the anatomical landmarks on the bone surface models. Each muscle-tendon path is represented by a series of line segments. In some cases, for example the soleus, origin and insertion landmarks are sufficient for describing the muscle path. In other cases, where muscle wraps over bone or is constrained by retinacula, intermediate via points are introduced to represent the muscle path more accurately. The number of via points activated for the muscle can depend on body position. For example, because the quadriceps tendon wraps over the distal femur when the knee is flexed beyond 80 degrees, additional via points, also known as “wrapping points,” are defined for the knee flexion angles greater than 80 degrees so that the quadriceps tendon can wrap over the bone, instead of passing through it, in that range of knee motion.

Despite the effort to define accurate muscle paths in the lower extremity, there are some muscles that pass through the bones or deeper muscles with extreme hip flexion and extension, and thus yield unrealistic moment arms. Specifically, GMAX3 (the most interior of the gluteus maximus) passes through the ischial tuberosity beyond 60 degree of hip flexion. GMAX1 and GMAX2 (the superior and the middle components of the gluteus maximus) pass through the deeper muscles beyond 80 degree of hip flexion.

For details about what muscles are included in each of the model, refer to the following PDF: Gait 2392 vs. Gait 2354.pdf

Dynamics

Inertial properties

The inertial parameters for the body segments in the model are adapted 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 shoe. 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.

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 Gait 2392 Model.

Table 2: Inertial parameters for the body segments included in the model

Body segment	Mass (kg)	Moments of inertia		
		xx	yy	zz
Torso	34.2366	1.4745	0.7555	1.4314
Pelvis	11.777	0.1028	0.0871	0.0579
Right femur	9.3014	0.1339	0.0351	0.1412
Right tibia	3.7075	0.0504	0.0051	0.0511
Right patella	0.0862	0.00000287	0.00001311	0.00001311
Right talus	0.1000	0.0010	0.0010	0.0010
Right calcaneus	1.250	0.0014	0.0039	0.0041
Right toe	0.2166	0.0001	0.0002	0.0010
Left femur	9.3014	0.1339	0.0351	0.1412
Left tibia	3.7075	0.0504	0.0051	0.0511
Left patella	0.0862	0.00000287	0.00001311	0.00001311
Left talus	0.1000	0.0010	0.0010	0.0010
Left calcaneus	1.250	0.0014	0.0039	0.0041
Left toe	0.2166	0.0001	0.0002	0.0010

Actuators and Other Force-Generating Elements

Peak isometric force

In the original lower limb model developed by Delp et al. (1990), values for the muscle-tendon parameters are determined with a procedure similar to that used by Hoy et al. (1990). Values for muscle physiological cross-sectional area (PCSA), which determine the peak isometric force, are taken from Friederich et al. (1990) and Wickiewicz (1983). Because the measurements reported by Friederich et al. (1990) [25 N-m^2] are obtained from experiments on young cadavers, and those reported by Wickiewicz et al. (1983) [61 N-m^2] are obtained from experiments on elderly cadavers, a factor that is larger than the “specific tension” reported by Spector at al. (1980) [23 N-m^2] is used to scale the PCSA values from the elderly cadavers.

While constructing the Gait 2392 Model from the original Delp model, Anderson noticed that the muscle strengths in the Delp model were still weak compared to the experimental results from Anderson and Pandy (1999) and Carhart (2000) on healthy, living subjects. To better match the strength of the Delp model to the joint torque-angle relationships measured in living subjects, additional strength scaling was employed. Despite efforts to keep the scaling factor consistent across all muscles, a different scaling factor is needed for bi-articular muscles because they span two joints. In many cases, the muscle strength parameters from Anderson and Pandy are used instead, as they are more physiologically accurate. For details, refer to the following PDF of the maximum isometric muscle forces from Gait2392/Gait2354, Delp1990, and Carhart2000, along with the scale factors: MuscleMaxIsometricForces.pdf

Optimal fiber length and pennation angle

For most muscles, values for the optimal fiber length and pennation angle are taken from Wickiewicz et al. (1983). The fiber lengths reported are scaled by a factor 2.8/2.2, which is the ratio of the sarcomere length at which muscle fibers develop peak force based on the sliding filament theory of muscle contraction (2.8 micrometers) to the sarcomere length measured by Wickiewicz et al. (2.2 micrometers).

For muscles not reported by Wickiewicz et al., the muscle-fiber length and pennation angles measured by Friederich et al. (1990) in the anatomical position are used instead.

Associated Publications

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

Delp, S.L., Loan, J.P., Hoy, M.G., Zajac, F.E., Topp E.L., Rosen, J.M.: An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures, IEEE Transactions on Biomedical Engineering, vol. 37, pp. 757-767, 1990. (Download PDF)

Yamaguchi G.T., Zajac F.E.: A planar model of the knee joint to characterize the knee extensor mechanism." J . Biomecl7. vol. 21. pp. 1-10. 1989. (Download PDF)

Anderson F.C., Pandy M.G.: A dynamic optimization solution for vertical jumping in three dimensions. Computer Methods in Biomechanics and Biomedical Engineering 2:201-231, 1999.(Download PDF)

Anderson F.C., Pandy M.G.: Dynamic optimization of human walking. Journal of Biomechanical Engineering 123:381-390, 2001. (Download PDF)

Carhart, M. R. "Biomechanical Analysis of Compensatory Stepping: Implications for Paraplegics Standing Via FNS," Ph.D Dissertation, Arizona State University, 2000.

Publications supplying anatomical data for the model:

Stredney, D. L. “The representation of anatomical structures through computer animation for scientific, educational and artistic applications,” Master Thesis, The Ohio State University, 1982.

Inman, V.T. The Joints of the Ankle. Baltimore: Williams & Wilkins, 1976.

Carhart, M. R. "Biomechanical Analysis of Compensatory Stepping: Implications for Paraplegics Standing Via FNS," Ph.D Dissertation, Arizona State University, 2000.

Friederich, J.A. and Brand, R.A. "Muscle fiber architecture in the human lower limb," J. Biomech., vol. 23, pp. 91-95, 1990.

Wickiewicz, T. L., Roy, R. R., Powell, P. L., and Edgerton, V. R., "Muscle architectuie of the human lower limb," Clin. Orthop. Rei. Res., vol. 179, pp. 275-283, 1983.
Hoy, M. G., Zajac, F. E., and Gordon, M. E., "A musculoskeletal model of the human lower extremity: the effect of muscle, tendon, and moment ann on the moment-angle relationship of musculotendon actuators at the hip, knee, and ankle," J. Biomech., vol. 23, pp. 157-169, 1990.