Supplementary Material

Balanced Foot Dorsiflexion Requires a Coordinated Activity of the Tibialis Anterior and the Extensor Digitorum Longus: a Musculoskeletal Modelling Study

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1 Supplementary Data S1

The model was implemented on the SimWise-4D platform (Design Simulation Technologies, DST, Canton, MI, USA).

The following bones were modelled: tibia, fibula, talus, calcaneus, navicular, cuboid, the three cuneiform bones, the five metatarsi, the five proximal phalanges, five middle phalanges, the four distal phalanges.

The tibiotalar joint was reproduced by imposing a 'collision' condition between the distal tibia and the upper surface of the talus. By this constrain the two contact surfaces could slide one relative to the other according to their morphology, without a predefined rotation axis. Other joints were reproduced by rotational hinges. In particular, the axis of rotation of the talocalcaneal joint, the Henke axis [1], was identified as an axis passing through the superior part of the talus and the lateral tuberosity of the calcaneus. The corresponding revolute joint was fixed with the talus, and, when the foot was on the ground, was oriented 42° over the horizontal plane and 16° medially about the longitudinal foot axis. The talonavicular joint was reproduced by a spherical joint, allowing three rotational degrees of freedom (DoF), since the talus head has a spherical shape indeed. The cuboid-calcaneal joint is a typical 'saddle' joint, allowing rotation around two orthogonal axes. It was reproduced by two rotational hinges oriented following the anatomy of this joint. Navicular, cuneiforms and the first three metatarsi, rigidly connected each other, constituted the medial forefoot. Cuboid and fourth and fifth metatarsi constituted the lateral forefoot. The proximal phalanges were hinged with their respective metatarsi by revolute joints, allowing the relative dorsal/plantarflexion of the fingers. No movement was allowed at the interphalangeal joints. The ligaments were reproduced by springs having no linear characteristics. According to [2] their force/deformation curve was composed of a quadratic tract and a subsequent linear tract. The following formulation was adopted:

$$F = \begin{cases} 0.25K\,\varepsilon^2/\varepsilon_l\,, & 0 \le \varepsilon \le 2\varepsilon_l \\ K(\varepsilon - \varepsilon_l), & \varepsilon > 2\varepsilon_l \\ 0, & \varepsilon < 0 \end{cases}$$

Where K is the stiffness (force per unitary deformation N/ ε) in the linear section of the curve; ε is the strain: $\varepsilon = (L - L_0)/L_0$; ε_l is the strain limit, that corresponds to the intercept of the linear tract with the abscissa. According to [2] the strain limit was assumed $\varepsilon_l = 0.03$. The stiffness K and the rest length L_0 were initially computed as follows:

$K = AE$ (where A is the cross sectional area of ligaments provided in [3] and E is the Young modulus of the ligaments, assumed as $E = 260\ MPa$ [4]).

$$L_0 = L_{ref}/(\varepsilon_{ref} + 1)$$

(where L_{ref} was the ligament length in a reference position in which a ligament deformation ε_{ref} could be reasonably assumed). For those ligaments that lengthened during dorsiflexion, the reference position was defined as the one in which the foot was lying on the ground subjected to a light compression force due to the weight of the bones; for those ligaments that lengthened during plantarflexion, the reference position was defined when the foot was suspended off the ground as a pendulum hung at the tibia. In that reference positions the deformation was assumed to be $\varepsilon_{ref} = 0.05$ (as in [2]).
During the first simulations we realized that for certain ligaments the tension achieved was exceedingly high and the resulting movement was sometimes erratic or unnatural. So we decided to modify the values of K and L_0 iteratively, taking into account the work of [5], in order to achieve a physiological range of movement with relatively low ligaments tension. Table I reports the values adopted, together with the stiffness $k = K/L_0$ and the ligament forces in their two reference positions: foot on the ground and foot off the ground.

2 Supplementary Table

Table S1. *Mechanical parameters of the foot ligaments adopted in the model.*

	L0 (mm)	K (N)	k (N/mm)	F (foot on the ground) (N)	F (foot off the ground) (N)
Talocalcaneal external	27.8	5021.0	180.6	0.0	3.0
Talocalcaneal posterior	26.7	5081.6	190.3	5.4	0.0
Calcaneo-cuboid dorsal	16.8	657.7	39.1	0.0	4.0
Calcaneo-navicular inferior	32.4	14221.9	438.9	38.0	0.0
Cuboid-navicular dorsal	16	626.4	39.1	41.0	0.0
Calcaneo-cuboid surface	49.1	5164.2	105.2	8.5	0.0
Talofibular anterior	17	635.3	35.3	0.6	42.4
Talofibular posterior	24	1979.7	82.5	4.5	0.0
Calcaneofibular	29.9	3772.4	126.2	0.0	0.0
Talocalcaneal interosseus	10.5	2447.5	233.1	0.0	3.7
Talonavicular dorsal	12	3200.0	266.7	0.0	29.0
Tibiocalcaneal	27.8	1744.6	62.7	9.5	0.0
Tibionavicular dorsal	32.9	642.1	19.5	0.0	0.0
Tibiospring	18.8	2299.4	122.3	0.0	6.0
Tibiotalar posterior	17.1	1995.0	116.7	3.1	0.0

The plantar aponeurosis was considered composed of three bundles: they all started from the lower surface of the calcaneus and were attached to the first, third and fifth metatarsal head respectively. Their mechanical properties were defined according to [6].

The following muscles were modelled: Soleus (SO), Gastrocnemius (GA), Peroneus Longus (PL), Peroneus Brevis (PB), Tibialis Anterioris (TA), Tibialis Posterioris (TP), Extensor Digitorum Longus (EDL), Extensor Allucis Longus (EAL); Flexor Digitorum Longus (FDL), Flexor Allucis Longus (FAL). They were represented by force actuators having the origin in a proximal point of the tibia and the insertion distally on the respective bones. With the exception of the SO and GA, for which the line of action was straight, the other muscles had the line of action bent over ‘via-points’. This deviation was obtained by splitting the model in two parts: the proximal one represented by the force actuator, and a distal one, at constant length, representing the tendon. The connection point of the two was constrained to lay on a circular path representing the convex surface of a bone contact (for example the lateral malleolus for the peronei) or a concave surface corresponding to the retinaculum constrain (for TA and EDL).

3 References

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