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A model for lever-arm length calculation of the flexor and extensor muscles at the ankle

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Abstract

A sagittal-plane mathematical model of joint mobility, including the mechanical effect of the extensor retinacula, was used to predict the lever arm lengths of the main flexor and extensor muscles of the human ankle over the range of movement. In plantarflexion, the centre of rotation lies posteriorly and distally, maximising the lever arm of the tibialis anterior. The action of the gastrocnemius and soleus is maximised in dorsiflexion. Traditional calculation of ankle joint moment based on a fixed centre of rotation is acceptable only in exercises such as level walking with a limited range of motion about the neutral position. The present model with a moving centre is particularly advised in exercises which take the joint nearer to the extremes of sagittal motion. © 2002 Elsevier Science B.V. All rights reserved.

Keywords: Ankle complex; Muscle force; Lever arm; Four-bar linkage; Centre of rotation

1. Introduction

The ankle–subtalar complex plays a fundamental role in the human locomotor system, being involved in virtually every activity. This unit provides the rocker of the shank with respect to the foot during the different phases of the walking cycle [1]: (1) from the terminal part of the swing phase, through the heel contact, until the foot is flat on the ground, it controls the lowering of the foot to the floor; (2) during the period in which the foot remains flat on the ground and the shank advances, it controls the continued forward movement of the body; (3) during the push-off phase, it allows the generation of power for progression of the limb. As in all other human joints, motion is guided by the osteoarticular and ligamentous structures and induced by the forces and moments of the extrinsic muscles. Muscles act by applying force through the muscle tendons with instantaneous lever arms relative to the joint centre, the tendons wrapping around bones and deflecting under retinacula when necessary. The lever arm lengths are measures of the ability of muscles to produce joint torque in order to generate rotation and/or to resist external forces. Any injury, lesion or neuromuscular disorder of this complex system affects this interaction between muscles, bones and ligaments and causes degradation, instability or disability of locomotion.

To enhance understanding of disorders and of relevant conservative and surgical treatments, a better knowledge of the physiological mechanics of the ankle complex still remains a crucial issue. All the previous models [2–6] assumed the ankle and subtalar joint to be simple hinge and fully congruent joints, although most of the recent experimental studies [7–13] strongly suggest that this assumption should be questioned. There is need for a mathematical model which can describe the changing orientations of the muscle tendon lines of action and positions of the instantaneous centre of rotation. Particularly for the definition of the lever arms of the main flexor and extensor muscles, there are no previous modelling studies aimed at analysing the mechanical effects of the retinacula constraining the tendons at the ankle joint.

Lever arm lengths have been estimated in experimental studies in vivo and in vitro by indirect measurements. The 'tendon excursion' method has been largely

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used in cadaver studies [14–17]. The instantaneous length of the muscle lever arms is calculated as the ratio between the instantaneous values of the tendon excursion and the joint rotation, the former being the distance moved by the tendon with respect to the underlying bone during both concentric and eccentric muscle contraction. This method of measurement accounts for the combined effect of (i) changing positions of the rotation axis; (ii) the spanning of the tendon over several joints; and (iii) action of retinacula, sheaths, or bony structures influencing muscle and tendon course. It has therefore been suggested that this length should be called the 'effective moment (lever) arm' [15]. The main concerns with this technique are the critical measurements of both tendon excursion and joint rotation, and the underlying assumption that the line of action of the muscle force and the axis of rotation are always perpendicular.

Measurements of lever arm lengths for the Achilles tendon and tibialis anterior were also performed in vivo [18–21] on a series of sagittal plane magnetic resonance images (MRI). Rugg et al. [18] found that moment arm lengths decrease by approximately 20% for the Achilles tendon and increase by approximately 30% for the tibialis anterior when the ankle is moved from maximal plantarflexion to maximal dorsiflexion. A 3.1 and 2.5% average increase of the moment arm was obtained for the two muscle groups, respectively when a fixed centre was used. It was claimed that the averaged moment arm lengths for these tendons were relatively unaffected by the use of a moving centre as opposed to a fixed. Maganaris et al. [19–21] measured the changes in the tibialis anterior and Achilles tendon moment arm lengths with increasing muscle force. Sagittal-plane MRI pictures were taken at rest and at maximum voluntary contraction in several different joint positions. The authors concluded that a substantial increase in the moment arms occurs because of the deformation of the tissues. They did not discuss the patterns of increasing of the tibialis anterior and Achilles tendon moment arms with respectively increasing dorsi- and plantarflexion of the ankle, nor the mechanical effects of the retinacular constraints.

In understanding the mechanics of a human joint, mobility studies can first elucidate the relationship of kinematics to the geometry of the passive structures, i.e. articular surfaces and ligaments. A previous extensive investigation [22–25] has shown, by virtue of a computer-based model of ankle mobility in the sagittal plane, that the articular surfaces and the ligaments prescribe a unique envelope for the instantaneous positions of the axis of rotation [25]. The final goal of this investigation was to analyse the changing directions of the lines of action of contact, ligament and muscle forces as the necessary preliminary information for the estimation of the magnitude of the corresponding internal forces during activity. The aim of the present study was to show these changing directions throughout the flexion range and to analyse the corresponding effects on muscle lever arms and forces during muscle strengthening and gait exercises.

2. Materials and methods

2.1. The four-bar linkage (4BL) model of the ankle joint complex

Experiments carried out by several authors on below-knee amputated specimens [22,24] have been later elucidated by a geometrical computer-based model [25]. In summary, during passive flexion, the motion of the ankle complex (calcaneus with respect to tibia/fibula segment) occurs mostly at the ankle (tibiotalar) level. This single degree-of-freedom (DOF) motion is guided by the isometric rotation of the most anterior fibres of the calcaneofibular (CaFi) and the tibiocalcaneal (TiCa) ligaments [22]. The instantaneous centre of rotation (IC), the point at which the two ligament fibres cross, moves anteriorly and proximally during dorsiflexion. The articular contact moves from the posterior part of the tibial mortise in maximal plantarflexion to the anterior part in maximal dorsiflexion. Rolling as well as sliding occurs at the ankle joint during flexion: the talus rolls forward while sliding backwards on the tibial mortise during dorsiflexion, and vice versa during plantarflexion.

The CaFi and TiCa ligaments are therefore essential to control normal kinematics. The orientation of these ligaments, the position of the instantaneous centre about which dorsi/plantarflexion occurs and the position of the contact point all change relative to both bony segments during flexion according to the ligament linkage. Because of the changing positions of the IC and of the points of tendon attachment and wrapping, muscle lever arm lengths must also change with flexion.

2.2. The muscle action model

The sagittal model of ankle mobility was developed to include the course of the main extensor and flexor muscle groups and the extensor retinacular bands. Fig. 1 provides a schematic geometrical representation of this model.

The geometry of this model in the neutral position was defined by sets of anatomical parameters. The course of the three extensor retinaculum bands and the central areas of origin and insertion of the soleus and tibialis anterior muscles were detected in three below-knee amputated specimens, using a stereophotogrammetric system and an anatomical landmark calibration procedure [24]. Three-dimensional (3-D) positions of

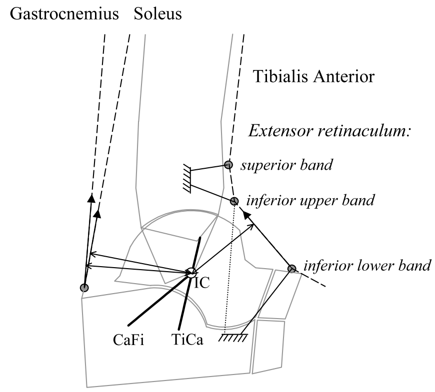

Fig. 1. The extended muscle action 4BL model in neutral position. The courses (dashed) and lines of action (single-arrow segments) of the muscles, the frictionless pulleys (grey circles), the extensor retinaculum bands (solid segments from the pulleys to the attachment areas) are depicted. The band not included in the mechanical effect of the extensor retinaculum is in dotted line. Muscle lever arm lengths (double-arrow segments) are the orthogonal distances between muscle force lines of action and instantaneous centre of rotation IC (empty circle, at the cross point of the model ligaments CaFi and TiCa).

these landmarks in the neutral position were projected in the sagittal plane. With respect to the gastrocnemius, the two-dimensional (2-D) insertion point was taken at the Achilles tendon attachment on the upper calcaneus, whereas the origin was assumed as it was taken at knee full extension in a geometrical sagittal model of this joint [26,27]. This knee model showed how small is the effect of knee flexion on the orientation of the gastrocnemius muscle in a tibial reference frame.

The muscle–tendon units (dashed lines) were modelled as single lines connecting relevant origins and insertions approximated by single points. Each of the main three extensor retinaculum bands was modelled as a frictionless pulley (grey circles), around which the tibialis anterior tendon wraps. The superior and inferior lower bands were rigidly attached to the tibia/fibula and calcaneus segments, respectively. In nature, the inferior upper band of the extensor retinaculum runs from the medial aspect of the distal tibia to the lateral aspect of the calcaneus. In the present model, the relevant pulley was taken to be rigidly attached to the tibia/fibula segment, and the mechanical action of the part of the band joining the pulley and the calcaneus (dotted line) was ignored. The movement of the inferior lower pulley was calculated from 4BL kinematics.

In such a multiple attachment/wrapping mechanism, when the tibia/fibula is taken as the fixed body and the talus/calcaneus as the moving body, the effective line of action of a pulling force joins the relevant extreme connection points. The line of action of the corresponding muscle force which spans the ankle joint therefore coincides with the tendon course between the upper and lower inferior retinaculum (single-arrow segment on the tendon course). The lever arm length of the tibialis anterior muscle (double-arrow segment) was therefore calculated as the orthogonal instantaneous distance between this muscle force line of action and the location of the IC at any joint position. Lever arm lengths (double-arrow segments) of the gastrocnemius and soleus muscles were simply taken as the instantaneous distances between relevant muscle lines of action (single-arrow segments) and the location of the IC.

A diagrammatic sketch from the computer animation based on this model is given in Fig. 2. The changing positions of the muscles, retinaculum bands, ligaments, and relevant centre of rotation and common normal in the sagittal plane are depicted at three characteristic joint positions.

Because both the lines of action of the muscle tendons and the centre of rotation change position during flexion as guided by the ligament linkage, the lever arm lengths of the three muscles change accordingly. The lever arms of the tibialis anterior and gastrocnemius are depicted in the figure. Because the centre of rotation

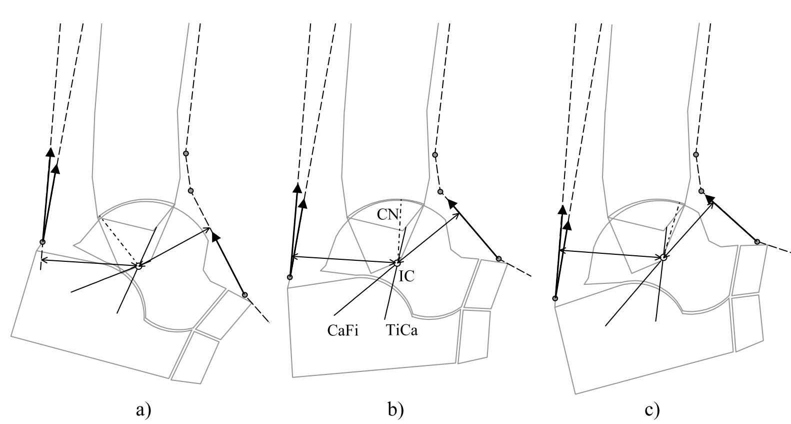

Fig. 2. A diagrammatic sketch of the mechanism of muscle leverage at the ankle complex in the sagittal plane as predicted by the muscle action development of the 4BL model. Lines of action (single-arrow segments) of the three muscles and lever arm lengths (double-arrow segment) of the tibialis anterior and gastrocnemius are shown at 20° plantarflexion (a), at neutral (b), and at 10° dorsiflexion (c) positions. Positions of the instantaneous centre (IC, empty circle) and of the contact point identified by the common normal (CN, dash–dotted line) are also reported.

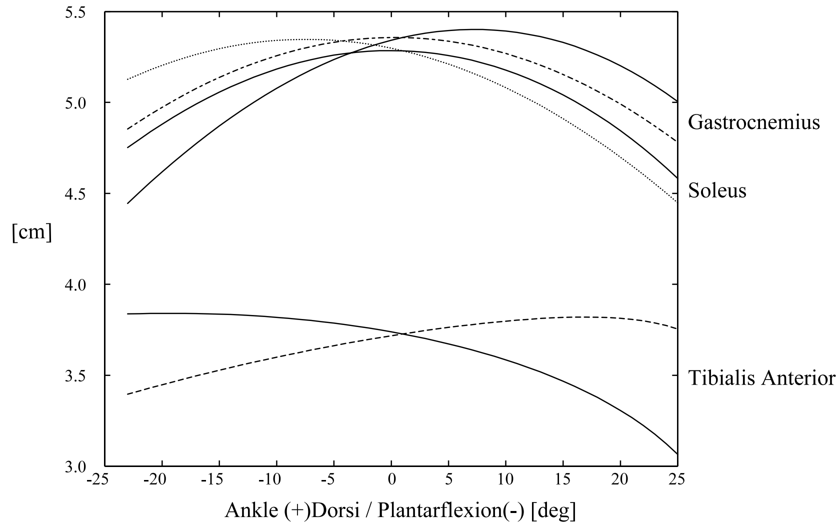

Fig. 3. Calculated lever arm lengths (solid lines) of the main flexor and extensor muscles against ankle dorsi/plantarflexion angle. Corresponding patterns for gastrocnemius (dash–dotted), soleus (dotted) and tibialis anterior (dashed) muscles calculated for a fixed-hinge ankle joint are superimposed.

moves forwards and upwards during dorsiflexion, the lever arm length of tibialis anterior becomes shorter as the ankle moves from maximal plantar- to maximal dorsiflexion, and vice versa for the two flexor muscles. For comparison, lever arm lengths were also calculated as though the ankle joint were a hinge with axis of rotation fixed at the location assumed in the neutral position.

2.3. The muscle strengthening exercises

To study the effect of the combined mechanical action of the muscle and of the external forces to be resisted during activity, muscle strengthening exercises were first simulated using the computer model configured with the same set of geometrical parameters. The tibia was thought to be held oblique at 45° from the horizontal facing up and down, respectively for the strengthening of the extensor and flexor muscles. A unit load was thought to be hung to the posterior aspect of the calcaneus and the anterior aspect of the talus, respectively. A 23° plantar /25° dorsiflexion range of motion was considered. Directions of the lines of action of the muscle forces and of the external load throughout the flexion range were calculated from the model.

2.4. The gait analysis test

A gait analysis test was carried out on an adult male volunteer (582 N weight, 1.74 m height), with leg dimensions roughly similar to that of one of the specimens analysed. Trajectories of anatomical landmarks on the shank and on several foot segments necessary to define the parameters of the model were tracked by a stereophotogrammetric system (ElitePlus, BTS, Milan) and a standard protocol [28,29]. Flexion angle and moment at the ankle were also calculated accordingly [29]. Ground reaction force (GRF) was measured by means of a force platform (Kistler, Switzerland). Electromyography (Telemg, BTS, Milan) was also used to assess muscle activation (on–off) timing, according to an established algorithm [30]. Kinematic, electromyographic and force data were collected simultaneously.

Sagittal projection of the instantaneous position of the GRF vector was expressed in the same tibial reference frame as the sagittal mechanical model and considered to be the external load resisted by the relevant muscle force. Patterns of change in IC position and muscle line of action orientation for every joint flexion angle were calculated. Muscle action was assumed to balance the joint moment induced by the GRF, therefore flexors act when GRF passes anterior to IC, extensors when GRF is posterior to IC.

3. Results

3.1. Prediction of muscle lever arm lengths

Fig. 3 shows the calculated lever arm lengths of the three muscles (solid lines) plotted against corresponding flexion angle of the ankle. A 23° plantar /25° dorsiflexion range of motion was imposed according to the relevant motion performed by one of the specimens analysed. With the model configured as in Fig. 1, both the anterior and proximal displacement of the IC was approximately 7 mm.

The figure shows that the lever arm of tibialis anterior increases from approximately 3.1 cm in dorsiflexion to 3.8 in plantarflexion, an increase of 23%. This trend was exactly opposite to that obtained from the fixed-axis model which predicted a decrease from a maximum of 3.8 to a minimum of 3.4 cm, 11%. The gastrocnemius has a maximum lever arm length of 5.4 cm at 7° dorsiflexion, with a total lever arm length excursion of 1.0 cm. The corresponding figures for the fixed-axis model are 5.3 cm at 0° flexion and 0.6 cm excursion, resulting in a 40% decrease of lever arm length excursion. The soleus has a maximum lever arm length of 5.3 cm at 0° flexion, with a total lever arm length excursion of 0.7 cm. The corresponding figures for the fixed-axis model are 5.3 at 7° plantarflexion, 0.9 cm excursion. Between $\pm 10°$ flexion from the neutral position, all these differences are small.

The calculation of lever arm lengths with sets of geometrical parameters taken from other specimens produced similar results.

3.2. Forces at the ankle in muscle strengthening exercises

Fig. 4 shows the lever arm length and the corresponding muscle force of the tibialis anterior muscle for the simulation of the strengthening exercise.

Fig. 4a shows that, in the present model of moving IC, the lever arm of the tibialis anterior is larger in maximal plantarflexion than in maximal dorsiflexion, and that the contrary occurs for the external load. These two contributions combine in Fig. 4b. It shows that in maximal dorsiflexion the value of the muscle force necessary to resist an external load with a moving centre joint is approximately 41% bigger than that obtained with a fixed centre, but approximately 22% less in maximal plantarflexion.

In maximal plantarflexion, the forces in the soleus and gastrocnemius muscles necessary to resist independently the external load in a moving centre joint was found, respectively to be approximately 20 and 21% bigger than that obtained with a fixed centre, but 4 and 5% less in maximal dorsiflexion. The calculation of lever arm lengths and simulation of the two strengthening exercises was repeated with sets of geometrical parameters taken from two other specimens with similar results.

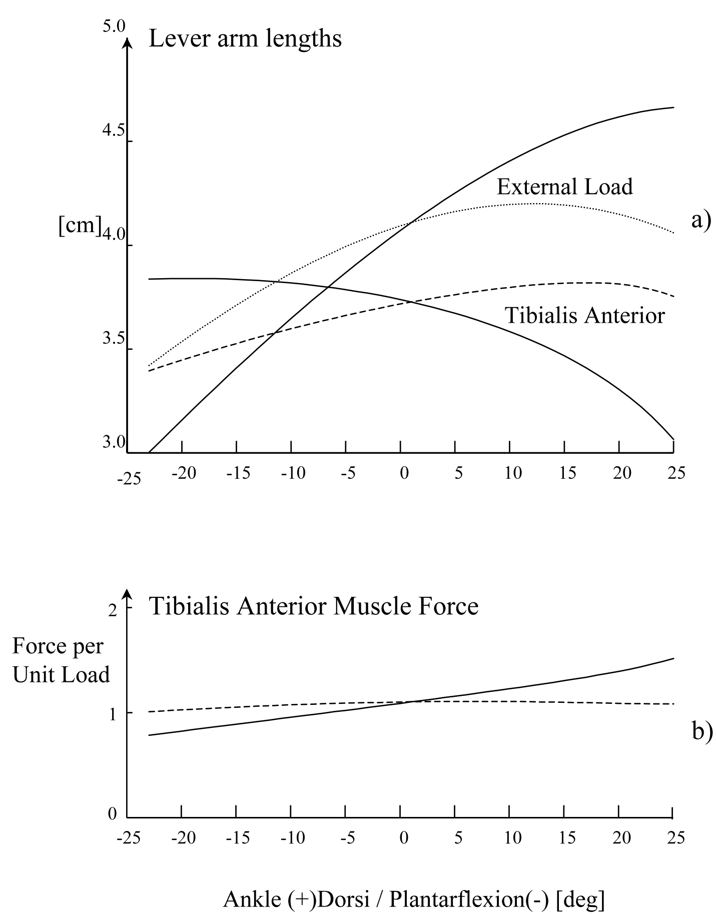

Fig. 4. Lever arm lengths of the tibialis anterior force and external load (a) and the magnitude of the tibialis anterior muscle force (b) necessary to counterbalance a unit external load in muscle strengthening exercise. The patterns for the present moving IC are reported with solid lines. Corresponding patterns for the tibialis anterior (dashed) and external load (dotted) for a fixed-hinge ankle joint are superimposed.

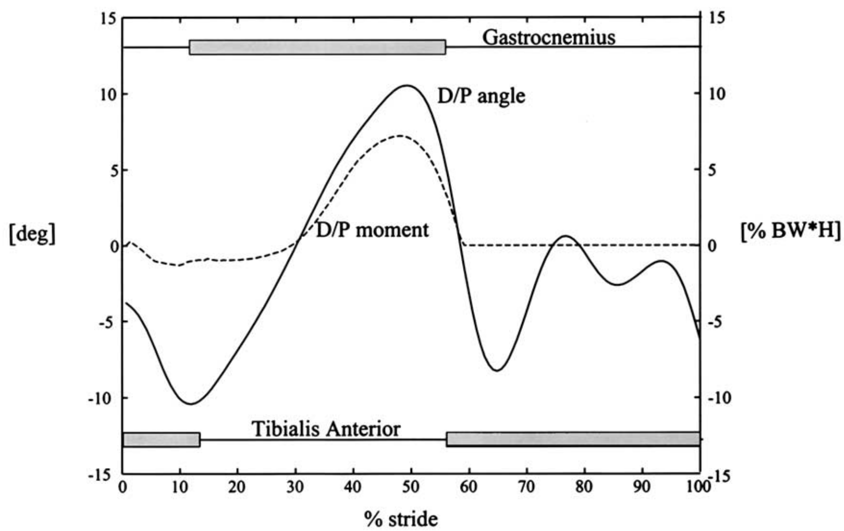

Fig. 5. Time history of muscle activation (grey bars) from EMG, of dorsi/plantarflexion angle (solid line, [°]) and of dorsi/plantarflexion moment associated to the GRF (dashed line, [% body weight × height]) at the ankle from the gait analysis test.

3.3. Forces at the ankle during gait

In Fig. 5, muscle action timings of the gastrocnemius and tibialis anterior muscles (grey bars) are superimposed on flexion angle and flexion moment of the ankle joint over a gait cycle. Figs. 3 and 4 showed that the mechanical advantage increases in some joint positions and decreases in others. The joint positions in which the two muscle groups fire are exactly those in which they were predicted to be mechanically advantaged.

Fig. 6a and b show lever arm lengths and the necessary force magnitudes for both the moving (solid lines) and fixed (dashed lines) centre joint model. A maximum advantage of approximately 15 and 4% was found, respectively, for the tibialis anterior and gastrocnemius muscle forces, accounted for by the combined action of a shorter lever arm for the GRF and a longer lever arm for the muscle forces.

4. Discussion

4.1. Main results and relevance of the present work

In the present study, the lever arm length variations for the gastrocnemius, soleus, and tibialis anterior muscles are predicted in a systematic way by an extension of a previously validated 4BL model of ankle mobility [24,25]. The previous study showed how a 4BL model defines the changing geometry of the joint. It was demonstrated that the ligament-based linkage guides the axis of rotation to move anteriorly and proximally during dorsiflexion. The present study now provides directions of the muscle forces and the relevant lever arm lengths, through the guided relative motion of the two bones. It is demonstrated here that, because of the changing positions of both muscle lines of action and IC, the lever arm lengths change significantly with flexion. Lever arms of the flexor muscles are maximised in dorsiflexion, that of the extensor muscle are maximised in plantarflexion.

The path of IC displacement makes the action of the tibialis anterior advantaged in plantarflexion, the action of the gastrocnemius and soleus advantaged in dorsiflexion. The mechanical benefit is further improved by considering that the action of the external forces to be resisted during activities may be inversely advantaged. The combination of the two factors makes the action of the tibialis anterior even more advantageous in ankle positions close to maximal plantarflexion and vice versa for the two flexor muscles.

The ability of a muscle to produce joint rotation in the presence of external forces depends on the relevant lever arm length, defined as the perpendicular distance from the axis of joint rotation to the line of action of the muscle–tendon unit force. Knowledge of instantaneous muscle leverage is therefore fundamental in human movement analysis and joint modelling. No previous mathematical models of the human ankle have addressed the mechanical effect of the retinacula or have presented the changing positions of the instantaneous centre of rotation, the directions of the lines of action of the main flexor and extensor muscles and the resulting lever arm lengths.

4.2. Limitations of the present model

The present model demonstrates the mechanical effects of the extensor retinaculum action in constraining the tibialis anterior tendon. Several simplifications were still made here. The elastic stretching of the extensor retinacula as the tibialis anterior contracts was not considered. This effect would produce an increase of the muscle lever arm [21]. The present model also does not completely consider the mechanical action of the inferior upper band of the extensor retinaculum. However, it is reasonably expected that this area of wrapping of the tibialis anterior tendon does not move with respect to the tibia during ankle flexion, as it does the inferior lower band, rigid with the foot.

Further limitations of the present study are associated to the 2-D nature of the model and to the assumed rigidity of the joint structures. Particularly, this sagittal model does not include the significant medio-lateral component of the tibialis anterior tendon course and the associated further effect in inverting/everting the foot. Although foot and ankle have 3-D functions, the activities analysed occur mainly in the sagittal plane, and the present 2-D model has elucidated the key mechanical features of the flexors and extensor muscles. Extension of the model to 3-D is still under development and relevant results will enhance present knowledge. The effects of deformable articular surfaces and extensible ligament fibres have been extensively investigated for the knee joint [31,32]. It has been demonstrated that although tissue deformation is critical for the calculation of ligament forces, it does not affect much calculation of muscle forces.

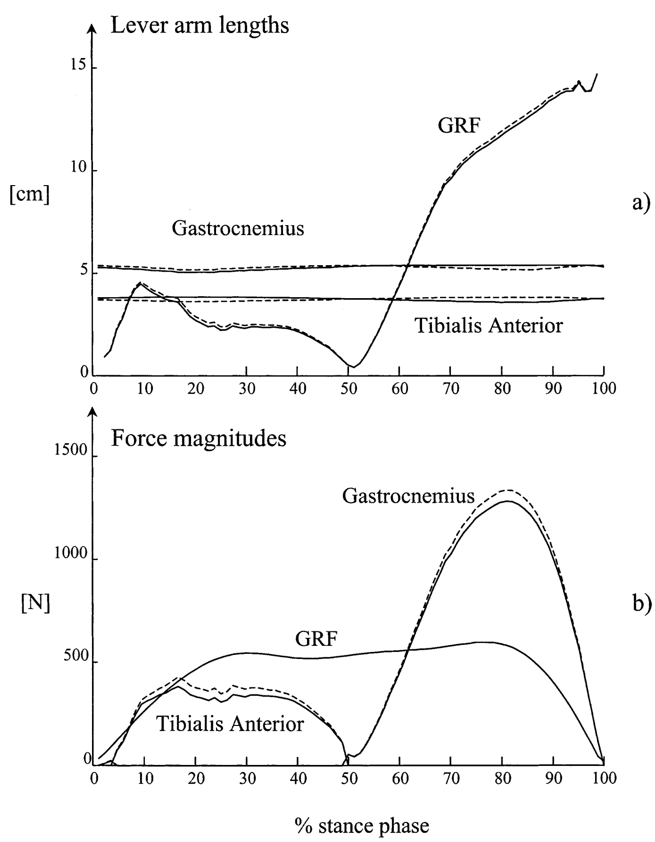

Fig. 6. Lever arm lengths (a) and magnitudes of the relevant forces (b) necessary to counterbalance the external GRF, expressed for both the 4BL (solid lines) and fixed centre of rotation (dashed lines) joint models throughout the stance phase of a gait cycle.

In the present determination of the forces transmitted by the structures at the ankle joint, a unique solution with a single muscle action has been only considered for any loading situation. Possible isometric simultaneous contractions of extensor and flexor muscles are in addition to those required to balance the external loads, and form systems of muscle, ligament and compressive contact forces with no resultant. Arbitrary values of these systems of forces can be added to the single muscle solution to give further solutions valid over the entire range of flexion. The present single muscle solutions therefore represent the minimal response of the joint to external load, at least in terms of muscle and contact force.

The predicted mechanical advantage during muscle activity must be considered here as only associated to the production of movement and as only accounted for the displacement, though small, of IC. The present considerations have no reference to any features associated to the production of force at the muscle–tendon units, such as the muscle–tendon interaction, the length–tensions and force–velocity relationships, and to the muscle fibre recruitment. Whatever would be the tension force developed at the muscle–tendon unit in a gait cycle, the associated generated torque is maximised by the pattern of motion of the centre of rotation. There could be situations, for example, in which the muscle force generated and the lever arm are not optimised at the same joint angle.

Finally, the present study is limited by the small number of experiments. However, the few observations on specimens confirmed a consistent pattern of antero-proximal translation of IC during dorsiflexion, the main cause for muscle lever arm changes. The only one volunteer analysed showed patterns of muscle activation and joint motion well within standard gait analysis reports.

4.3. Choice of muscle fulcrum point

Any system of forces, and in particular the system of muscle, ligament and articular contact forces transmitted across a joint can always be reduced to a single resultant force on any chosen line of action, plus a couple. The magnitude of the resultant force is given by the vector summation of the muscle, ligament and articular contact forces and the moment of the couple is given by the summation of the moments of the forces.

In moment equilibrium equations, any point (in 2-D) or axis (in 3-D) can be taken as the fulcrum point about which torque can be derived for every force acting on the joint (muscular, external, ligament and contact). It is recognised that the centre of rotation moves with respect to both bones during rotation in several human joints and it is therefore not the most simplest point for the calculation of joint moments in in vivo movement analysis tests. Although this centre is also not easily determined, it is preferred to the even more difficult evaluations of other characteristic points such as the articular contact or the centre of curvature. Effective moment arm for the muscle forces can be estimated as the partial derivative of the tendon excursion with respect to the joint rotation as in previous studies. The external force, typically the ground reaction force, has such a long moment arm that the corresponding moment is not too much affected by movement of the centre of rotation. The ligament and contact forces are ignored in moment equilibrium equations because they pass through the centre of rotation in frictionless joints [25,26]. A more accurate representation of the instantaneous positions of the centre of rotation is therefore certainly fundamental in both routine human movement analysis and in modelling studies of joint mechanics. The present study is the first attempt to define changing positions of the lines of action of the internal forces at the ankle joint in a systematic way.

In a simplified configuration with only the external, the muscular and the contact forces, this latter can be easily determined from the force equilibrium. The magnitude of this force was found to be very similar to the sum of the magnitudes of the external and muscular forces because their orientations are almost parallel throughout the flexion range, and therefore not reported. The present anatomically based geometrical model is also able to provide the changing directions of the forces transmitted by muscles, ligaments and articular surfaces. These are fundamental data for further mechanical analyses aimed at calculating the magnitude of these forces.

4.4. Comparison with previous works

The present predictions of lever arm length changes are in general agreement with the results reported by Spoor et al. [15]. The present results are in disagreement with the other previous studies based on MRI measurements, where the position of the instantaneous centre of rotation obtained by Reuleaux's graphical method [18–21] is liable to significant errors [33]. Using this latter technique, the patterns of lever arm changes reported for both a fixed and a moving centre of rotation were found to be very similar. These studies in fact failed to show a systematic path of motion for the centre of rotation. These patterns of lever arm changes are instead similar to those reported here when simulating a fixed centre of rotation (dashed lines in Fig. 3), demonstrating the general weakness of this technique.

4.5. Mechanics of the ankle in the gait cycle

Though estimated in the sagittal plane only, the prediction of muscle action during gait was found to be attractive. When looking at the mechanics of the gait cycle, the GRF is the external load to be balanced. IC moves in the same direction of the displacement of GRF vector during the stance phase of walking, causing a generally smaller lever arm and a corresponding smaller generated torque to be resisted than what it would be for a fixed centre of rotation. These changes affect disadvantageously GRF action during the walking cycle. The firing muscles generally lie opposite to the GRF and therefore are inversely advantaged by the movement of IC. Action of the flexor muscles is maximised in ankle dorsiflexion, action of the extensor in plantarflexion, exactly those joint positions in which the two muscle groups fire in the gait cycle. The mechanical advantage of the gastrocnemius (Fig. 6b) is a little smaller than corresponding tibialis anterior advantage because of the much larger lever arm of the GRF at push-off phase (Fig. 6a). A slightly smaller muscle force is therefore necessary to resist GRF and produce joint movement, and a smaller contact force is also expected. This advantage can elucidate the role of the moving centre in reducing muscle forces and therefore joint loading during level walking. Although further gait analysis tests would be necessary to confirm these observations and in vitro tests would better support experimentally these results, the present observation of a mechanical, though small, advantage at the ankle joint in the gait cycle supports the assumptions underlying the formulation of the model.

4.6. Final recommendations

Comparison of muscle force estimation in strengthening (Fig. 4) and gait (Fig. 6) exercises points out a difference in the mechanical advantage of the muscle action associated to the moving centre of rotation. In the stance phase of level walking, the advantage may be marginal being limited to a maximum of 4% for the gastrocnemius, although 15% for the tibialis anterior. In the muscle strengthening exercise, the tibialis anterior can reach an advantage of 22%. Fig. 3 reveals that this difference is accounted for the much larger range of motion imposed in muscle strengthening (23° plantar/25° dorsiflexion) than that observed in the gait cycle (10° plantar/10° dorsiflexion).

In the clinical context, it is here advised that careful restoration of the original geometry of the ligamentous and muscle–tendon apparatus would allow not only a more physiologic pattern of joint kinematics, but also a more physiologic pattern of joint loading. The present model would advance the estimation of the resulting external moment and power at the ankle joint in movement analysis and the estimation of the internal forces at the single anatomical structures. A more accurate estimation of the effects of muscle–tendon unit lengthening can certainly be obtained by the present geometrical model for a better assessment of the relevant surgical interventions.

5. Conclusions

This study has shown how a previous model of ankle mobility enhanced with a model of the mechanical effect of the extensor retinacula is effective in a systematic prediction of flexor and extensor lever arm lengths at the human ankle joint. Displacement of the centre of rotation significantly affects estimation of the muscle lever arm lengths. Traditional calculation of ankle joint moment based on a fixed centre of rotation model are acceptable only in exercises with a limited range of motion near the neutral position (gait). A fixed centre of rotation model for the ankle should not be used in the analysis of human exercises which take the joint nearer extremes of motion in the sagittal plane (muscle strengthening but also stair climbing/descending, raising from a chair, deep squat, etc.). In these latter, the present 4BL mechanical model is particularly advised, pending development of a 3-D anatomical model of the joint system.

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