Muscle Preactivation Control: Simulation of Ankle Joint Adjustments at Touchdown During Running on Uneven Ground

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In locomotion, humans have to deal with irregularities in the ground. When they encounter uneven terrain with changes in vertical height, they adjust the geometry of their legs. Recent investigations have shown that the preactivation of the gastrocnemius muscle (GM) correlates with the ankle angle at touchdown, but it is as of yet unclear why these adjustments were achieved by the GM and not by the preactivation of the tibialis anterior (TA). To examine the differences between TA regulation and GM regulation regarding (1) ankle angle adjustment and (2) joint stiffness, we used a three-segment musculoskeletal model with two antagonistic muscles (GM, TA). During the GM regulation, the ankle angle was adjusted from 121° to 109° (dorsiflexion) by a 41% decrease in the GM activation. During the TA regulation, the activation of TA must be increased by about 52%. In addition, we found that the ankle stiffness was most sensitive to changes in activation of the GM and decreased by about 20% while adjusting the angle. In contrast, the ankle stiffness remains similar when using TA regulation. Thus, the GM regulation is more adequate for adjustment in the ankle joint, enabling sufficient regulation of angle and stiffness.

Keywords: biomechanics, human locomotion, musculoskeletal model, gastrocnemius, joint stiffness

During locomotion, humans have to deal with irregularities in the ground, such as pathways covered with stones, grass, or roots. When they encounter uneven terrain with changes in vertical height, they do not want to stumble and adjust the geometry of their legs consciously (Müller & Blickhan, 2010; Grimmer et al., 2008; Patla & Rietdyk, 1993). In a situation in which the runner is well aware of a step, leg length at touchdown is shortened in proportion to the height of the obstacle (Blickhan et al., 2007; Grimmer et al., 2008). Shorter leg length implies increased bending in the joints (Blickhan et al., 2007). Grimmer et al. (2008) found decreased ankle angles with increased step height at touchdown. A recent study has shown that during running on uneven ground the preactivation of the GM correlates with the ankle angle at touchdown in contrast to an approximately constant TA preactivation during this phase (Müller et al., 2010). Yet, it is unclear why the ankle joint angle was adjusted by the GM and not by the TA. In the same study, the authors identified decreased ankle joint stiffness with increasing step heights (Müller et al., 2010). Decreased joint stiffness is crucial to minimize the vertical lift of the runner’s center of mass (COM) during running over a step as well as to come back to a stable COM trajectory after the step or perturbation (Ferris et al., 1999; Grimmer et al., 2008).

In principle, the ankle joint angle can be adjusted by the activation of a feed forward controlled muscle crossing the joint (Duncan & McDonagh, 2000; Funase et al., 2001). The change in activation results in an altered ankle angle, but the amount of the change in the angle depends on the muscular force-length relation. To achieve greater dorsiflexion angles at touchdown, humans can (1) increase the preactivation of tibialis anterior (TA, ankle joint flexor) or (2) reduce the preactivation of gastrocnemius (GM, ankle joint extensor). We did not investigate the ankle joint adjustment by the soleus, as this is considered to be a “reflex” muscle (Duncan & McDonagh, 2000; Dyhre-Poulsen et al., 1991). Its preactivation at touchdown is, first, smaller than the GM preactivation (Swanson & Caldwell, 2000; Wank et al., 1998) and, second, its activation occurs later during swing phase (Kuitunen et al., 2002; Swanson & Caldwell, 2000; Wank et al., 1998). Considering an electromechanical delay of 30–100 ms between EMG and force (Cavanagh & Komi, 1979; Vos et al., 1991) and its lower preactivation, the contribution of the soleus to the ankle joint moment is considered to be negligible.

Differences in structure and function between GM and TA may result in a better suitability of one muscle to adjust the ankle angle and the ankle joint stiffness (regarding reduced neuronal effort). Both muscles work within different working ranges of their force-length relation during running (Herzog et al., 1991; Ishikawa et al., 2007; Maganaris, 2001, 2003). During isometric contraction at ankle and knee joint angles found while
important design criteria for neuro-muscular systems during cyclical locomotion include reduced CNS control and energetic effort (Blickhan et al., 2007). To regulate ankle angle and joint stiffness, both strategies (GM and TA) may be used during external electrical stimulation of paraplegia to enable automotive locomotion (Agarwal et al., 2003; Kojovic et al., 2009; Marsolais & Kobetic, 1987; Popovic, Curt, et al., 2001; Popovic, Keller, et al., 2001; Sharma et al., 1998) or in active leg-prosthesis to enable sufficient ankle joint adjustment (Au et al., 2008; Eilenberg et al., 2010). The force-length curve of available musculature or its technical equivalent, a joint spring, relates force to deflection without neuronal feedback. This allows adjustment of joint angles by changing muscle activity (stimulation amplitude) in a paraplegic leg or by changing force in a quasi-elastic prosthesis. The same input signals, stimulation amplitude or force in the prosthesis, can be also used to adjust joint stiffness if after touchdown ground reaction force is fed back activation. Nevertheless, in an antagonistic system the question arises which of the muscles should be addressed for adjustment. Here, a closer look at the human ankle joint may illuminate the situation.

The purpose of our study was, first, to quantify the required GM or TA muscle activation to adjust ankle angles found while running on uneven ground; second, to calculate ankle joint stiffness using GM and TA regulation; and third, to compare the two regulation strategies with respect to experimentally observed changes in ankle joint stiffness. Therefore, we employed a three-segment musculoskeletal model with two antagonistic muscles (GM, TA).

Methods

Description of the Musculoskeletal Model

The musculoskeletal model used in this investigation consisted of a skeletal and a muscular submodel (Figure 1), which were modeled in Matlab/Simulink (R2006a, Mathworks, Inc., Natick, MA, USA). The skeletal submodel was comprised of a chain of three rigid segments, representing foot, shank, and thigh. The segments were interconnected by frictionless hinge joints. Parameter values for the skeletal submodel were adopted from a two-dimensional, forward-dynamic model, which has extensively been described by van Soest and Bobbert (van Soest & Bobbert, 1993; van Soest et al., 1993). In addition, it has been used successfully in simulation studies of human vertical jumping (van der Krogt et al., 2009; van Soest et al., 1993) and cycling (van Soest et al., 2005).

The muscular submodel (Figure 1b) consisted of two antagonistic muscles: the monoarticular ankle joint flexor TA (spanning the ankle joint) and the biarticular ankle joint extensor GM (spanning the ankle and knee joint). The soleus was not included in the model due to negligible muscle preactivation. The origins and insertions of the GM and the TA represented in Figure 1b were taken from van Soest et al. (2003).

The muscle force was described as

\[ F_m = A \cdot F_{im} \cdot F_1 \]  

where \( A \) is the muscle activation, \( F_{im} \) is the maximal isometric force, and \( F_1 \) is a factor representing the force-length relation of the muscle (Gordon et al., 1966). Within this model we assumed close to static conditions before touchdown. Including the force-velocity effects would not affect the general result (see discussion).

In the transition from \( 120^\circ \) to \( 70^\circ \), the working range of the GM is limited to the ascending limb of the force-length relation and the muscle force increased approximately linearly (Maganaris, 2003). The force-length relation of the GM was determined by linear regression to experimental GM force-length data (Maganaris, 2003):

\[ F_{L,\text{GM}} = k \cdot l_{\text{GM}} + F_0; \quad F_{L,\text{GM}} \in [0,1] \]  

Figure 1 — Musculoskeletal model. (a) Schematic representation of the skeletal submodel including definition of joint angles (ankle angle: \( \alpha \); knee angle: \( \beta \); \( \alpha \) represents the angle between the shank and a line (*) that is parallel to the sole of the foot (angles greater than 90° represent plantar flexion and angles less than 90° represent dorsiflexion), \( \beta = 180^\circ \) knee joint fully extended. (b) The musculoskeletal model with the origins (GM: +0.017 m and TA: –0.017 m relative to the knee joint) and insertions (GM: –0.07 m and TA: –0.07 m relative to the ankle joint) of the gastrocnemius and the tibialis anterior; \( r_{\text{GM}} \) and \( r_{\text{TA}} \) are the perpendicular distances between the force vector of the GM and the ankle joint axis and the force vector of the TA and the ankle joint axis, respectively.
where \( k \) and \( F_0 \) are specific parameters of the linear equation, and \( l_{GM} \) is the GM length. The specific \( F_l \) parameters of the GM were shown in Table 1.

Within an ankle angle from 135° to 60°, the TA muscle operates on the ascending limb and plateau region of the force-length relationship (Maganaris, 2001). Therefore, the force-length relation was described using parabolic function (Siebert et al., 2010; Woittiez et al., 1984) as follows:

\[
F_l = 1 - \left( \frac{1 - l_0}{l_w} \right)^2 ; \quad F_l \in [0,1]
\]

where \( l_w \) is the depth of the parabola, \( l \) is the muscle length and \( l_0 \) is the optimum length, where the muscle produces \( F_{im} \). The specific \( F_l \) parameter of TA (Table 2) was determined by nonlinear regression to experimental TA force-length data (Maganaris, 2001).

The passive ankle joint elasticity was included by fitting a quadratic function to experimental moment-angle data (Weiss et al., 1986):

\[
M_{pass} = 0.4533 \cdot \sigma^2 - 118.47 \cdot \sigma + 7691 \quad [Nm] \quad (90° \leq \alpha \leq 140°)
\]

Within the observed ankle angle range, there were only marginal passive forces at maximal plantar flexion, but passive forces rose with dorsiflexion.

**Experimental Data**

In the study of Grimmer et al. (2008), 11 subjects ran along uneven ground with steps of adjustable height. They have found that the ankle and knee angles at touchdown ranged from 109° to 121° and from 144° to 161°, respectively. Kinematic data of the ankle joint and normalized electromyographic data that could range between 0 and 1 for maximal activation (\( A_{max} \)) of the TA and the GM of the same experiments (Müller et al., 2010) were used to set initial condition (Figure 2). In this study, the initial activation of the GM from experimental data (running without step) was set fixed (\( A_{initial,GM} = 0.25 \)).

### Table 1
**Fl parameter and Fim of the GM**

<table>
<thead>
<tr>
<th>( k )</th>
<th>( F_0 )</th>
<th>( R^2 )</th>
<th>( F_{im} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>GM (70° ≤ ( \alpha ) ≤ 120°)</td>
<td>16.09</td>
<td>-7.63</td>
<td>0.969</td>
</tr>
</tbody>
</table>

### Table 2
**Fl parameter and Fim of TA**

<table>
<thead>
<tr>
<th>( l_0 )</th>
<th>( l_w )</th>
<th>( R^2 )</th>
<th>( F_{im} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>TA (60° ≤ ( \alpha ) ≤ 135°)</td>
<td>0.50</td>
<td>0.0524</td>
<td>0.952</td>
</tr>
</tbody>
</table>

**Figure 2** — Normalized EMG of (a) GM and (b) TA during running upward a step of different height; solid line: without step, i.e., while running across level ground, dashed line: step of 5 cm, dotted line: step of 10 cm, and chained line: step of 15 cm. The gray shaded area is one standard deviation of the reference run on the undisturbed track. The beginning of the ground contact (touchdown) is marked by a vertical line and the curves are continued till the end of the longest contact phase. The selected initial activation of the GM is marked by a circle.
Model Calculations

Due to the assumed quasi-static conditions at touchdown, inertial moments can be neglected and the equilibrium of moments at the ankle joint can be written as

$$M_{GM} + M_{purs} = M_{TA}$$  \hspace{1cm} (5)

with

$$M_{GM} = F_{M,GM} \cdot r_{GM}$$  \hspace{1cm} (6)

for GM, and

$$M_{TA} = F_{M,TA} \cdot r_{TA}$$  \hspace{1cm} (7)

for the TA, where $r_{GM}$ and $r_{TA}$ are the perpendicular distances (taken from Maganaris, 2001, 2003) between the force vector of the GM and the ankle joint axis and the force vector of the TA and the ankle joint axis, respectively (Figure 1). To calculate the initial state of the TA ($A_{initial,TA}$), an equilibrium situation was defined using the experimental activation of the GM ($A_{initial,GM} = 0.25$) at touchdown and the corresponding experimental ankle ($\alpha_{ws} = 121^\circ$) and knee ($\beta_{ws} = 161^\circ$) angles at touchdown observed in running across level ground. During GM regulation, we calculated the necessary GM activation to adjust the ankle angle, whereas $A_{initial,TA}$ was held constant. In the other case, during TA regulation the TA activation was varied whereas $A_{initial,GM}$ was held constant. In the calculations, we examined ankle angle changes from $121^\circ$ to $101^\circ$. Starting from a point of equilibrium ($121^\circ$), this range includes the ankle angle range ($109^\circ \pm 8^\circ$) from our observations while running over steps elevated by 15 cm. Furthermore, the stiffness of the ankle joint in the model was calculated as a change in joint moment divided by a change in joint angle. To examine the influence of the combined movement (experimental ankle and knee angles) on ankle angle adjustment and ankle stiffness, we examined (1) the adjustment of the ankle angle while the knee angle ($\beta_{ws} = 161^\circ$) was fixed. Then (2) we included experimentally observed changes in the ankle and knee angle in the calculations of the ankle angle equilibrium state.

Results

Using the experimental activation of the GM at touchdown ($A_{initial,GM} = 0.25$) and the experimental ankle and knee angles at touchdown ($\alpha_{ws} = 121^\circ$, $\beta_{ws} = 161^\circ$), the initial state of the TA was 0.252 and corresponded well with the experimental data (Figure 2b). Based on this equilibrium situation, the adjustment of the ankle joint angle can be regulated by preactivating the GM as well as by preactivating the TA.

In a first attempt the knee joint angle was set fixed ($\beta_{ws} = 161^\circ$). During GM regulation the ankle angle $\alpha$ was adjusted from $121^\circ$ to $109^\circ$ (dorsiflexion) by a 46% decrease in the GM activation from 0.25 to 0.134 whereas the TA activation was held constant (Figure 3). During TA regulation the activation of the TA must be increased by about 65% from 0.252 to 0.416 to adjust the ankle angle (Figure 3). Across the ankle angle range between $121^\circ$ and $109^\circ$ the ankle angle activation relation was linear for the GM ($\alpha = 133.7A_{GM} + 90.1$, $R^2 = .990$) and the TA ($\alpha = -75.3A_{TA} + 140.2$, $R^2 = .997$).

In addition, we found that the ankle stiffness was most sensitive to changes in the activation of the GM (Figure 4). For example, if the ankle angle was flexed from $121^\circ$ to $109^\circ$ using GM regulation, the relative ankle joint stiffness decreased by about 26.5%. In contrast, changing the activation of TA had a much smaller effect and ankle joint stiffness decreased by about 7.7%.
In a second attempt we included the experimentally observed change in the knee angle from 161° to 144° (Figure 5; Grimmer et al. 2008). To adjust an ankle angle of 109° when the knee angle is 144°, the TA activation has to be increased by about 51.6% (ATA = 0.382) during TA regulation. During GM regulation, on the other hand, the GM activation has to be decreased by about 40.8% (AGM = 0.148), and the ankle stiffness was reduced to 80.4%. In contrast, the ankle stiffness of the model remains similar (differences <0.3%) when using TA regulation.

The total activation $A_{ges}$ (sum of GM and TA) to adjust the experimental ankle angle increased in TA regulation ($A_{ges} = 0.632$) compared with initial condition ($A_{ges} = 0.502$). However, during GM regulation the total activation decreased ($A_{ges} = 0.4$). Static and dynamic optimization (minimization of muscle activations) has been used extensively to estimate in vivo muscle forces during gait (Anderson & Pandy, 2001; Brand et al., 1986; Glitsch & Baumann, 1997; Patriarco et al., 1981; Yokozawa et al., 2007). Under these circumstances, the regulation by the GM portends a lower applied load in the observed ankle joint, expressed by a lower $A_{ges}$. Thus, it can be considered to be more efficient for ankle flexing than the TA regulation.

Adjustment of Ankle Stiffness by Activation

To adjust an ankle angle of 109° when the knee angle is 144° (experimental ankle and knee angle at highest step), the ankle stiffness was reduced to 80.4% and corresponded well with experimental data (79%, Müller et al., 2010) during GM regulation. In contrast, the ankle stiffness of the model remained similar (differences <0.3%) using TA regulation.

Joint stiffness strongly depends on the level of activation of the muscles acting on the joint (Weiss et al., 1988). Thus, the ankle could be made stiffer by increasing the activation of the GM or by increasing the coactivation of the TA (Hortobagyi & DeVita, 2000; Kuitunen et al., 2002; Nielsen et al., 1994; Weiss et al., 1988). However, a decreased activation of the GM or a decreased coactivation of the TA resulted in a decreased joint stiffness. During GM regulation this is in accordance with our model. In case of an increased activation of the TA to flex the ankle joint (TA regulation), coactivation increases but the ankle stiffness of the model remains similar. Thus, the GM regulation is more adequate for adjustment in the ankle joint, enabling sufficient regulation of angle and stiffness.

Influence of Knee Angle Change on Ankle Stiffness

By changing the knee angle from 161° to 144°, the effect of the muscle activation on ankle adjustments was reduced. Thus, part of the adjustments can be regulated by knee flexion. However, a decreased activation of the GM or an increased activation of the TA was required to adjust an ankle angle of 109°.

As an alternative to ankle joint adjustment by activation, it is conceivable that ankle stiffness is adjusted by changing leg geometry—because the stiffness of the muscle-tendon complex (MTC) crossing the ankle varies depending on muscle-tendon length (Gottlieb & Agarwal, 1978; Weiss et al., 1986). Farley et al. (1998) found that in hopping on different surface stiffness the ankle stiffness is affected by the touchdown knee angle without changes in activation. A shorter MTC of the
biarticular GM arises from a more flexed knee and an unaltered ankle angle at touchdown. Thus, the GM muscle force would decrease without changes in activation of the GM resulting in decreased ankle stiffness. During running on uneven ground, the leg length at touchdown was shortened proportionally to the height of the obstacle (Grimmer et al., 2008). Although the knee and ankle angle decreased simultaneously, the MTC length of the GM increased by about 1.1 cm. With a fixed knee angle (β≤ 161°) the GM length change is about 1.4 cm. Therefore knee flexion reduced the GM lengthening only by 20% and the ankle stiffness by about 7%. Thus, the effect of muscle activation on ankle stiffness seems to be larger than the effects of muscle length. This is in accordance to the results of van der Krogt et al. (2009).

**Limitations of the Model**

Within this three-segment musculoskeletal model we assumed quasi-static conditions. Thus, the force-velocity relation was neglected. Before touchdown the angle of the ankle joint does not change (for –0.001 s < t < 0 s: Δα ≤ 50°; Müller et al., 2010). The length of the GM is also affected by the altering knee angle (for –0.040 s < t < 0 s: Δβ ≤ 150°/s at β = 170°; Müller & Blickhan, 2010). In our model this results in an additional decrease in GM force and length (see above)—in addition to the pantograph coupling taken into account, too. Consequently, the dynamics of knee movement supports ankle dorsiflexion.

During the foot ground contact phase, the ankle joint angle changes (Grimmer et al., 2008; Müller & Blickhan, 2010) thereby shortening and lengthening the antagonistic muscles spanning the ankle joint, respectively. Due to the force-velocity relation (Edman, 1988; Hill, 1938), the lengthening muscle generates higher force while the shortening muscle generates lower force (lower than its isometric force at a given activation and muscle length). This would result in an enhanced ankle joint stiffness during ground contact.

Furthermore, the introduction of elasticity in series (SEC) to the contractile component (Hill, 1938) introduces an additional degree of freedom to the muscle model (Mörl et al., 2012; Siebert et al., 2008). Therefore, length and velocity of the CC (contractile component) can be decoupled from the entire muscle length and velocity. The influence of the SEC increases for increasing ratio of SEC and CC lengths, which is reported for larger compared with smaller animals and for more distal compared with more proximal muscles (Biewener, 1998; Zajac, 1989). To examine the ankle joint adjustments under dynamic conditions (e.g., during ground contact) it is essential to include the force-velocity relation and the influence of elasticity in series to the contractile component.

Several studies show that a three-dimensional model can produce different results compared with that of a two-dimensional model (Glitsch & Baumann, 1997). In a first approach we used a two-dimensional musculoskeletal model, which considers the movement of the upper ankle joint only (e.g., Geyer & Herr (2010). The contribution of GM and TA to lower ankle joint movement (inversion/eversion) was neglected, which may be a limitation of the model.

Furthermore, the model used in this study is a generic model (van der Krogt et al., 2009; van Soest & Bobbert, 1993; van Soest et al., 2003) and does not include any subject specifications (e.g., segment length or muscle properties).

Nevertheless, this model approach provides an insight into the mechanism of ankle joint adjustments at touchdown and stresses the role of GM during running on uneven ground.

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