Energetics and Passive Dynamics of the Ankle in Downhill Walking

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This study investigated the energetics of the human ankle during the stance phase of downhill walking with the goal of modeling ankle behavior with a passive spring and damper mechanism. Kinematic and kinetic data were collected on eight male participants while walking down a ramp with inclination varying from 0° to 8°. The ankle joint moment in the sagittal plane was calculated using inverse dynamics. Mechanical energy injected or dissipated at the ankle joint was computed by integrating the power across the duration of the stance phase. The net mechanical energy of the ankle was approximately zero for level walking and monotonically decreased (i.e., became increasingly negative) during downhill walking as the slope decreased. The indication is that the behavior of the ankle is energetically passive during downhill walking, playing a key role in dissipating energy from one step to the next. A passive mechanical model consisting of a pin joint coupled with a revolute spring and damper was fit to the ankle torque and its parameters were estimated for each downhill slope using linear regression. The passive model demonstrated good agreement with actual ankle dynamics as indicated by low root-mean-square error values. These results indicate the stance phase behavior of the human ankle during downhill walking may be effectively duplicated by a passive mechanism with appropriately selected spring and damping characteristics.

Keywords: ankle, modeling, biomechanics, gait analysis, robotics

Recent work has shown the net mechanical energy observed at the ankle during the stance phase of level walking is approximately zero at self-selected speeds and is negative at slower speeds (Hansen et al., 2004), indicating the behavior of the ankle is energetically passive during these modes of locomotion. This result is of particular interest to engineers designing efficient ankles for bipedal robots as well as prosthetic and orthotic devices, as it suggests it may be possible to reproduce the stance behavior of the ankle during level walking with a passive mechanical mechanism, that is, a pin joint coupled with either a revolute spring alone or with a revolute spring and damper. However, while much is known about the energetics of the ankle during level walking, comparatively few studies have considered the ankle during downhill walking. Therefore, the purpose of this study was to investigate the energetics of walking on a range of downhill slopes with the goal of reproducing the downhill walking behavior of the ankle during stance phase with energetically passive mechanical mechanisms.

Studies of level walking at self-selected speed have shown the ankle absorbs energy (i.e., the mechanical work performed by the ankle is negative) during the early portions of stance from heel strike to foot flat and from foot flat to heel off, and it supplies mechanical energy to the step (i.e., the ankle performs positive mechanical work) during the latter portion of stance from heel off to toe off (Winter, 1983; Kuster et al., 1995; Palmer, 2002). The net energy of the ankle during the stance phase, the sum of these two components, has been recently reported to be approximately zero at self-selected speeds and negative at slower speeds (Hansen et al., 2004). This finding suggests ankle behavior throughout stance at normal or slower than normal speeds is similar to a passive mechanism—dissipating or storing energy early in stance and releasing stored energy late in stance.

We emphasize that this zero or negative net energy result does not suggest the ankle itself is, in general, a passive mechanism. While passive tissues such as tendons and ligaments play a role in the motion of the ankle (Alexander, 2003), electromyography (EMG) analyses have made it clear that muscle activation in ankle flexors and extensors is present during locomotion (Radcliffe, 1962; Stern, 1997; Gottschall & Kram, 2003). Rather, the zero net energy result indicates the behavior of the ankle under these circumstances (the stance phase of level walking at self-selected speeds or slower) is energetically passive.
This finding is significant insofar as these modes of locomotion are common and as modeling and duplicating this behavior with a passive mechanical unit would be highly useful for the development of orthotics, prosthetics, and bipedal robots.

We have reason to suspect the behavior of the ankle is energetically passive during the stance phase of downhill walking as well. While less is known about walking on slopes, existing studies show downhill walking at self-selected speeds corresponds to reductions in both the overall energetic cost of walking (volume of oxygen consumed per unit time) and the positive work performed by the ankle late in stance compared with the ankle behavior during level walking. Margaria's (1976) investigation of O₂ consumption, recently duplicated by Minetti et al. (2002), demonstrated that the metabolic cost of walking diminishes with slope angle and is minimized on a downhill slope of about 5.7°. Mitsui et al. (2001) reported walking downhill resulted in decreased EMG activity in the ankle extensor gastrocnemius, exhibiting minimum activity on 5.7° slope, while activity in the ankle extensor soleus remained relatively constant. As ankle extensor activity is responsible for the positive work performed late in the stance phase (Radcliffe, 1962; Stern, 1997; Winter, 1983; Kuster et al., 1995), we suspect that a reduction in extensor activity in downhill walking will correspond to a decrease in the positive work performed by the ankle during stance. This may lead to a net negative ankle energy during locomotion at self-selected speeds, indicating the behavior of the ankle remains passive during downhill walking albeit more dissipative than in level walking.

Weiss et al. (1986) and Palmer (2002) have each investigated passive models of the ankle. Weiss et al. studied the dynamics of the ankles of reclined participants and reported spring and damper coefficients that were relatively constant across midrange motions. However, the model of Weiss et al.—which captured the behavior of the ankle while no weight was being supported—is not an effective model for the behavior of the ankle during the weight-bearing task of locomotion. Indeed, the model coefficients reported by Palmer (2002), who fitted coefficients of passive mechanical models to the behavior of the ankle during level walking, differ significantly from those reported by Weiss et al. (1986). Palmer (2002) modeled the behavior of the ankle during the first two portions of the stance phase of level walking—from heel-strike to foot-fall (the impact absorption portion) and from foot-fall to heel-off (the dorsiflexion portion)—as separate passive mechanical models, and reported spring and damper coefficients an order of magnitude larger than those found by Weiss et al. We note, however, that Palmer did not include the final portion of the stance phase from heel-off to toe-off (the propulsion portion) in the passive model parameters he reported and did not consider nonlevel walking.

Presently, it is unknown whether the stance behavior of the ankle is energetically passive during downhill walking and whether this behavior may be effectively duplicated by a passive mechanism with appropriately tuned springs and dampers. To this end, this study seeks to (1) investigate the energetics of the ankle during downhill walking at self-selected speed and determine whether its stance behavior is energetically passive and (2) to determine whether the behavior of the ankle in this mode of walking may be effectively reproduced by a passive mechanical device, if in fact the ankle dynamics are energetically passive. These results are significant to those designing ankle prosthetic and orthotic devices for humans as well as ankle joints for bipedal robots. If the stance behavior of the ankle during level and downhill walking may be replicated by a passive mechanism, ankle assistive devices and robotic ankle joints may be rendered simpler and lighter as there is reduced or eliminated need for a power supply.

Methods

Participants

Eight healthy males ranging in age from 22 to 27 with an average mass of 74 ± 4.7 kg and an average height of 178 ± 4.7 cm with no known gait impairments participated in this study. Participants were selected to minimize variation due to gender, age, mass, and stature. Each participant was informed of the experimental protocol and provided consent in compliance with the University Institutional Review Board. Each participant wore bicycling shoes (U.S. size 10.5, within one-half size of the participant’s normal shoe size) with rigid soles that restricted the motion of the foot to approximate a single rigid segment.

Experimental Setup

A variable slope apparatus was constructed consisting of three walking surfaces joined by hinges (Figure 1a). All walking surfaces were painted with nonslip paint. A force platform (model 02172; Advanced Medical Technology Inc., Watertown, MA, USA) was embedded in the center section level with the walking surface. The starting section of the walkway was mounted on four hydraulic jacks (model F-2365; Prolift, Kansas City, MO, USA). As the jacks were raised, hinges between the sections allowed the starting and ending platforms to remain horizontal while the center section assumed various angles. A number of structural reinforcements were used to minimize movement of the walking surfaces during data collection.

An optoelectric 6-camera system (model 460; Vicon Motion Systems Ltd., Oxford, UK) was used to collect kinematic data. Reflective markers were placed over the acromion, sacrum, and bilaterally over the anterior-superior iliac spine, lateral epicondyle of the femur (LEF), and lateral malleolus (LM). Four additional markers were placed on each shoe over the calcaneus (CAL) and first metatarsal of each foot (TO1) as shown in Figure 1.

Experimental Protocol

Participants practiced walking on level ground until they felt comfortable in the rigid-soled shoes (approximately
5 min. For each slope angle, the participants were instructed to walk the length of the walkway at a self-selected pace. No mention was made of the force platform and participants were instructed to look straight ahead to prevent them from biasing their gait based on its location. A successful trial consisted of heel-strike and toe-off of either foot completely within the boundaries of the force plate without contact from the other foot. The participant’s starting position was adjusted until a successful trial was achieved. Participants completed two successful trials at each ramp angle and were provided breaks while the angle was adjusted between trials.

The set of ramp angles used in this study was selected with the most resolution between 2° and 6°, a range that contained slopes corresponding to minimum metabolic cost of human walking (Margaria, 1976; Minetti et al., 2002) and minimum energetic cost of robot walking (Yamakita & Asano, 2001; Goswami et al., 1996; McGeer, 1990). Participants walked down the ramp in the order of 0°, 2°, 3°, 4°, 5°, 6°, and 8° not exceeding the greatest angle based on the legal maximum angle of a wheelchair ramp (United States Access Board, 2004). The order was selected to allow the participants to acclimate to rigid-foot walking with increases in slope angle.

**Data Analysis**

Kinematic and kinetic data were sampled at 100 Hz and were conditioned using a second-order zero-phase Butterworth low-pass filter with 10 Hz cutoff using MATLAB (version 6.5; The MathWorks, Natick, MA, USA). Analysis was computed in the sagittal plane and was deemed sufficient since 93% of the work done at the ankle during walking occurs in this plane (Eng & Winter, 2002). The ankle angle was defined as shown in Figure 1b, in agreement with (Redfern & DiPasquale, 1997). The angular velocity was computed using a numeric approximation of the derivative, that is, the change in angle divided by the time between frames.

Measures from the two successful trials of each ramp angle for each participant were averaged. Exceptions were made in two instances in which one of the two successful trials for a given angle produced data not representative of biological motion likely due to incorrect electronic synchronization of kinematic and kinetic data streams. In these instances, only data from the trials without technical errors were considered. Out of the 112 successful trials, the two exception trials occurred in different participants and on different angles, suggesting the errors were not related.

The ankle torque $\tau(t)$ was computed using the measured ground reaction force, center of pressure, and the motion data in a bottom-up inverse dynamics procedure (Winter, 2005). Mechanical power at the ankle was calculated by multiplying the torque and angular velocity of the ankle. To compute the mechanical energy of the ankle, the mechanical power was integrated

$$E = \int_{t_0}^{t_1} \tau(t) \dot{\theta}(t) \, dt$$

where time limits of integration $t_0$, $t_1$ were chosen by the specific gait phases; for example, to compute the energy during the dorsiflexion portion of stance, we integrated from $t_0 = \text{foot-fall}$ to $t_1 = \text{heel-off}$.

The following linear model for the ankle torque was used

$$\tau_{\text{an}}(t) = k_p \theta(t) + k_d \dot{\theta}(t) + c$$

$$= \begin{bmatrix} k_p & k_d \end{bmatrix} \begin{bmatrix} \theta(t) & \dot{\theta}(t) \end{bmatrix} + c$$

where $k_p$ and $k_d$ represent the unknown spring and damper coefficients, respectively. In the equation above, $c$ is a constant offset term that accounts for the angle of spring relaxation; for instance, $c = 0$ would indicate the revolute spring is relaxed when $\theta = 0°$, that is, when the sole of
the foot is perpendicular to the shank. Parameters \( k_p, k_d, \) and \( c \) were estimated by fitting the model-generated ankle torque \( \tau_{sd}(t) \) to the experimental data using linear regression, that is, by computing

\[
[k_p \ k_d \ c]^T = [\theta(t) \ \dot{\theta}(t) \ 1]^{-1} \tau_{sd}(t)
\]

where \([ \cdot ]^{-1}\) denotes the Moore–Penrose pseudo-inverse. To judge the quality of fit we computed the root-mean-square (RMS) error between the model-generated torque \(\tau_{sd}(t)\) and the experimentally measured torque \(\tau(t)\)

\[
error_{RMS} = \sqrt{\frac{1}{t_1 - t_0} \int_{t_0}^{t_1} (\tau_{sd}(t) - \tau(t))^2 \, dt}
\]

where again the time limits of integration \(t_0, t_1\) were chosen by the specific gait phases.

**Statistical Analysis**

A repeated measures analysis of variance was performed using SPSS (version 15.0; SPSS Inc., Chicago, IL, USA) to evaluate the significance of slope variation on the measures of mechanical power and energy and the passive model parameters. The degrees of freedom were Huynh–Feldt corrected when appropriate.

**Results**

Both step speed and step length increased nearly monotonically as the steepness of the downhill slope increased (Figure 2). The amount of variation in step length between slope angles was similar to the variations found in step speed. Both step speed and step length were significantly affected by slope angle (\(p < .0001\) and \(p = .007\) respectively).

Ankle power and torque were computed from the experimental data using the methods described above and representative plots have been provided (Figure 3). Four peak power values were identified during stance phase: a power minimum during impact absorption, a minimum and maximum power during dorsiflexion, and a power maximum during propulsion. These minima and maxima were averaged across participants for each angle (Figure 4). The greatest variation occurred in the power minimum during absorption, which ranged from –56.8 W on level ground to –148.9 W on 8°. The power maximum during propulsion remained relatively constant and was not significantly affected by slope. Slope had a significant effect on the power minima during absorption (\(p < .0001\)) and during dorsiflexion (\(p < .0001\)). Slope also had a significant effect on the power maximum during absorption (\(p = .003\)).

![Figure 2](image)

**Figure 2** — Forward speed and step length with variation ramp angle.
Parts of the stance phase when ankle power was positive (respectively, negative) were integrated to compute the positive (respectively, negative) mechanical energy produced (respectively, dissipated) during stance (Figure 5). The net mechanical energy of the ankle during the stance phase was equivalent to the sum of the positive and negative energy values. The positive energy ($p = .006$), negative energy ($p < .001$), and net energy ($p < .0001$) were all significantly affected by slope. The magnitude of the negative energy contribution dominated that of the positive energy for all angles tested, resulting in net negative ankle energy that decreased monotonically as the downhill angle increased.

The ankle power was also integrated across the three portions of stance to compute the net mechanical energy during impact absorption, dorsiflexion, and propulsion (Figure 6). Slope had a significant effect on dorsiflexion energy ($p < .0001$) as well as impact absorption energy ($p = .003$), while energy during propulsion was not significantly influenced by slope. Ankle energy during absorption and dorsiflexion decreased with slope and exhibited minimum values at 8° and 6°, respectively. The greatest variation occurred in dorsiflexion energy, which ranged from –9.03 J on level ground to –19.8 J at 6°.

The passive model was fit to the experimental data using the linear regression method described above (Figure 3). The quality of the fit was judged by computing the RMS error between the modeled torque and the experimentally measured torque during the impact absorption, dorsiflexion, and propulsion portions of stance (Figure 7 a). The net RMS error across all of the stance phase was also computed, and the percentage error between the model and actual torque was calculated for each slope (Figure 7 b). Slope had a significant effect on the RMS error during dorsiflexion ($p < .001$) and propulsion ($p = .006$) as well as the net RMS error across the stance phase ($p = .001$). The RMS error during impact absorption was not significantly affected by slope.

Evolution of the parameters of the passive model ($k_p, k_d, c$) revealed nearly monotonic trends as slope varied (Figure 8 a, b, c). The spring relaxation angle corresponding to the value of the linear offset $c$ was also computed.
Figure 4 — Trends in peak ankle power with variation in ramp angle: (a) Power minima during impact absorption and dorsiflexion portions of stance, (b) Power maxima during dorsiflexion and propulsion portions of stance.
Figure 5 — Positive (propulsive) work, negative (dissipative) work, and net work performed by the ankle during stance with variation in ramp angle.

Figure 6 — Trends in ankle energy during impact absorption (heel-strike to foot-fall), dorsiflexion (foot-fall to heel-off), and propulsion (heel-off to toe-off) with variation in ramp angle.
Figure 7 — RMS error between the torque of the passive ankle model and the experimentally measured ankle torque versus ramp angle: (a) RMS error during various portions of the stance phase, (b) overall RMS error for all of stance phase as a percentage of the RMS ankle torque.
All model parameters were significantly influenced by slope ($k_p: p < .0001$, $k_d: p < .0001$, offset $c: p < .0001$, relaxation angle: $p = .025$).

**Discussion**

This study provided ankle data on slopes not previously examined. However, on those angles where this work and prior studies overlapped, we noted general agreement in ankle torque and power profiles (Eng & Winter, 2002; Kuster et al., 1995; Redfern & DiPasquale, 1997; Winter, 1983).

The net mechanical energy of the ankle during stance was approximately zero for level walking, confirming the observation of Hansen et al. (2004). Net ankle energy was negative for all slopes tested, indicating the ankle is responsible for dissipating energy during downhill walking. Furthermore, the net ankle energy decreased monotonically (i.e., became increasingly negative) as the slope became steeper, indicating an increasing role in mechanical damping as slope increased; this observation corresponded to an increase in the damping coefficient of the passive model as slope increased (Figure 8 b). Measures of peak power (Figure 4) and energy (Figure 6) during the impact absorption and dorsiflexion portions of stance trended increasingly negative with steeper slopes, indicating that the increased energy dissipation during downhill walking occurs in these early portions of stance.

More energy was dissipated during dorsiflexion than any other portion of stance (Figure 6). We note that dorsiflexion energy exhibited a local minimum at 6°, roughly the same slope as that of the reported minimum EMG activity in the ankle extensors (Mitsui et al. 2001) and that of the reported minimum energetic cost of walking (Margaria 1976; Minetti et al. 2002), both of which occurred on 5.7°.

The RMS error of the model across the stance phase remained low for all angles, bounded below 14 N-m and never surpassing 25% of the RMS torque value (Figure 7). The relatively low error values suggest that the spring and damper mechanism effectively models the ankle dynamics for level walking, confirming the prediction of Hansen et al. (2004). Furthermore, all measures of error were minimized on slopes of 2°–4°, indicating the passive
model was most accurate on these slopes—even more accurate than on level ground. We note that this range of angles correspond to the angles on which passive robot walking has been reported (Yamakita & Asano, 2001; Goswami et al., 1996; McGeer, 1990).

Error during the dorsiflexion portion of stance was less than the error during impact absorption for all angles tested and less than the error during propulsion for all but two of the angles tested, indicating the passive model is most effective during dorsiflexion.

While the value of the damping parameter $k_d$ increased as the steepness of the slope increased, all values of the damping parameter were more than one order of magnitude smaller than values of the spring parameter (Figure 8). This is consistent with the relative magnitudes of ankle spring and damper coefficients reported in related studies (Palmer, 2002; Weiss et al., 1986). We conclude that ankle damping is negligible for walking at self-selected speed on the range of slopes tested. However, if the trends we observed continue, the damping role will become increasingly significant (and non-negligible) on steeper slopes. Furthermore, as damping magnitude is a function of speed, we anticipate that the significance of damping will be higher for faster walking speeds.

The angle of spring relaxation was nonzero for all slopes, indicating the spring of a massive ankle mechanism is not relaxed when the sole of the foot is perpendicular to the shank. Moreover, the variation in the spring relaxation angle with slope implies that the configuration of the ankle spring would vary depending on the target walking slope.

These results bear significance for the design of prosthetic and human assistive devices as well as biomimetic ankle joints for bipedal robots. One of the major concerns in using active prosthetic and orthotic devices is the consumption and management of power, which may make the device heavier and more cumbersome. It is tentatively suggested that during the stance phase of level and downhill walking the behavior of the ankle can be reproduced by a passive mechanical system. It may be possible to design an assistive or prosthetic device that utilizes this notion to minimize the power required for such a device. For additional implications for level walking, the interested reader is encouraged to consider the discussion of Hansen et al. (2004). Furthermore, as these results indicate the net mechanical energy during downhill walking is negative, it is also proposed that particular phases of gait during downhill walking could actually provide power to the device. Similar to regenerative braking used in hybrid cars on the market today, the energy absorbed during downhill slope walking could be regenerated to provide power to the device.

In addition, these results may be useful in the development of efficient ankle joints for bipedal robots, which have been shown to bear some energetic similarities to human locomotion on slopes. Simple two-limbed robots have demonstrated remarkably anthropomorphic gait requiring little energy input during downhill walking, with required energy input minimized on slopes of $2^\circ$–$4^\circ$ (Collins et al., 2005; Goswami et al., 1996; McGeer, 1990). The ankles of these robots, however, have been designed ad-hoc and in many cases do not flex at all. This fact provided part of the inspiration for this investigation. Mimicking human leg behavior above the ankle has been shown to correspond to minimal energy consumption in the bipedal robot model of Srinivasan & Ruina (2006); in a similar fashion, we anticipate that implementing a mechanism that reproduces or mimics the stance behavior of the human ankle during level or downhill walking will be of use to developers working to minimize the energy consumption of bipedal robots.

While the low RMSE indicates a fully passive mechanism could effectively reproduce the stance behavior of the ankle during level or downhill walking at normal speed, we note that such a mechanism would only minimize RMS error on one particular slope. The significant effect of slope on $k_s$, $k_d$, and the angle of spring relaxation indicates a fully passive mechanism with fixed spring and damping parameters will not optimally approximate the stance behavior of the ankle on all slopes. Optimization of the behavior for locomotion on varying slopes calls for an ankle mechanism with tunable spring and damper parameters. Such a device would not be fully passive but would require some form of active control to adjust the parameters of the spring and damper mechanism when the slope on which the human or bipedal robot is walking changes.

Although an attempt was made to model the foot as a single rigid segment, the use of the rigid soled bicycle shoes provided only an approximation. This type of shoe is not typically used for walking and may have made the relevance to a normal human foot more challenging. Because the foot was modeled as a single rigid segment, the unorthodox placement of the toe marker provided the best resolution for the kinematic data collection without compromising the integrity of the rigid foot model.

We note the linear passive model we have proposed and analyzed here is only one of many possible passive models that could be tested. Indeed, there are a number of nonlinear passive models that could also be studied.

Although an attempt at limiting morphological variations between participants was imposed, minor variations in physical parameters between participants played a role in influencing all measures of power, energy, and passive model parameters. This suggests that the energetics and dynamics of the ankle are particular to individual morphological characteristics and such parameters must be considered in the design of bipedal robots and the development of lower limb prosthetics.

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