Passive Energy-Saving Mechanisms in Human Locomotion

by

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Abstract

Humans tend to value economy of locomotion, often choosing movement strategies that help minimize how hard their bodies must work to perform a task. In this thesis I explore passive mechanisms humans use to reduce the metabolic energy consumed by their muscles, specifically due to positive muscle work, which is metabolically expensive compared to negative work or force production. I use a combination of modeling and human movement analysis to investigate how work performed passively by the ankle and by distributed soft tissues can save energy.

During normal walking, ankle push-off work provides an economical way to transition between steps. Push-off prior to collision redirects the body's velocity upward, which reduces the energy dissipated by collision, and the positive muscle work that must be performed to compensate for these losses. Through computational modeling and an experimental study of amputees walking on a variable-stiffness prosthetic foot, I demonstrate that elastic energy storage and return at the ankle can passively perform this energy-saving push-off function.

Active muscle work can also be reduced by passive soft tissues, which can perform mechanical work without the metabolic cost. I found that during walking and jump landings people choose to reduce demands on muscles by performing negative soft tissue work during collisions, and through a damped-elastic rebound of soft tissues after collisions.

While passive ankle work may seem entirely distinct from wobbling soft tissues, I demonstrate that similar biomechanical principles underlie the benefits of each. To save energy during locomotion one should avoid negative work that is not freely

returned, otherwise it requires extra positive muscle work to compensate. Alternatively, passive mechanisms may provide a means to reduce both negative and positive muscle work.

However, since economy is not the only factor influencing movement, I also present a jump landing experiment to demonstrate how the amount of work people perform may reflect their subjective valuation of active muscle effort vs. other difficult-to-measure costs, such as pain. Ultimately, the long-term goal is to use these fundamental energy-saving principles elicited through simulation and human movement analysis to inform the design of assistive technology for individuals with locomotor impairments.

Chapter 1.

Introduction

Humans seem to value economy of locomotion, often moving in ways that reduce the energetic demands on the body (e.g., [1–6]). This may seem obvious in certain locomotor activities. For example, it is beneficial for a person running a marathon or cross-country skiing to move in a way that minimizes his metabolic energy expenditure, a measure of how hard the body is working, in order to maximize performance. However, even in less demanding tasks such as walking there is evidence that minimizing effort, specifically minimizing energy expenditure, is an important factor influencing how people move (e.g., [2], [4], [5]). One of the major energetic costs of movement is due to actively performing work with muscles, in particular positive work, which is more metabolically expensive than negative work or isometric force production [8–10]. Therefore, humans take advantage of a variety of different energy-saving mechanisms during walking to reduce muscle work, including passive pendular dynamics [11], [12], trailing limb push-off [5], elastic energy storage and return [1], arm swing [13] and step length selection [2]. In this thesis we use a combination of computational modeling and human movement analysis to investigate two passive energy-saving mechanisms in locomotion: ankle elasticity and soft tissue deformation. The long-term goal is to apply fundamental insights gained from the study of human locomotion to improve assistive technology for individuals with locomotor impairments.

Experimental and modeling studies presented in this dissertation investigate the role of the ankle elasticity in gait, amputees walking on prosthetic feet, contributions of passive soft tissues to locomotion and factors that influence preferred movement strategy. While studies presented here are necessarily focused and task specific, the deeper scientific endeavor is to elucidate fundamental principles that are applicable across various forms of human locomotion, and potentially relevant to other animals and other fields, such as assistive device design and walking robots. Based on the passive energy-saving mechanisms we studied we found two common, underlying principles:

1. To save energy, beware of negative work because it then requires extra positive work, which is metabolically costly when performed by muscles.

2. If negative work is necessary, perform it passively. The passive negative work is not only less costly than the equivalent work done actively, but it can also reduce the total mechanical work and the positive muscle work needed.

In Chapter 2 I present a computational model to investigate the effect of ankle elasticity on walking economy. It has been proposed that elastic energy storage and return at the ankle can improve walking economy by reducing the active work required of the calf muscles [14–17]. However, this explanation fails to explain why any ankle work actually needs to be performed. After all, less energy would be consumed by the calf muscles by simply not contracting them. I propose that the fundamental energy-saving benefit of ankle work is its ability to modulate collisions by redirecting the body center-of-mass (COM). Here, we use a simple walking model to show that ankle elasticity is an elegant solution that can perform this push-off function passively and reduce the total energetic demands of walking, including work required from more proximal knee and hip muscles. The model predicts that under certain conditions ankle elasticity could even completely remove the need for active muscle work by reducing collision losses to zero, highlighting the fundamental benefit of avoiding dissipative negative work.

In Chapter 3 I study the importance of ankle push-off in amputees during walking. Lower-limb amputees expend more energy to walk than non-amputees and have an elevated risk of secondary disabilities [18–24]. Insufficient push-off by the prosthetic foot as a person transitions from one step to the next is believed to be a major contributing factor because it leads to larger energetic losses (i.e., more negative work) in heelstrike collisions, which must then be offset by positive muscle work. We systematically studied the effect of prosthetic foot mechanics on gait by varying a single parameter in isolation, the energy-storing spring in a prototype prosthetic foot, which affects the magnitude of prosthetic push-off. The prosthesis, called the Controlled Energy Storage and Return (CESR) foot, uses a spring underneath the heel to store energy from heelstrike collisions, then locks this energy in place until near the end of stance phase of walking before allowing this energy to return in the form of plantarflexion push-off. This push-off is similar in function to the natural ankle during walking, and is of larger magnitude than conventional foot prostheses. We found that the softest springs tested led to the greatest elastic energy storage, energy return and prosthetic limb COM push-off work. But metabolic energy expenditure was lowest with a spring of intermediate stiffness, suggesting biomechanical disadvantages to the softest spring despite its greater push-off. These results indicated that the passive spring compliance could contribute to the prosthetic prototype's push-off, but with biomechanical trade-offs that limited the degree to which greater push-off improved walking economy.

In Chapter 3 I also present a study of non-amputees walking on the CESR foot by wearing prosthetic simulator boots. This experiment paralleled the amputee protocol, but we observed some substantial differences in joint kinetics between the groups. During prosthetic push-off, amputees exhibited reduced energy transfer from the prosthesis to the COM along with increased positive hip work, as compared to the non-amputees. This suggests that amputees dissipated energy from the prosthetic push-off at the knee, and partially compensated by performing additional positive work at the hip. The need to perform this extra positive work again highlights the dangers of unnecessary negative work on the economy of movement, and may explain why this CESR prototype yielded metabolic benefits in the non-amputees tested, but not in the amputee group.

In Chapter 4 I present a study on soft tissue contributions to human walking. Although the muscles and tendons of the lower extremity are generally considered the dominant producers of work during gait, passive soft tissues can also perform mechanical work. In our study of normal human walking we found indirect evidence for both negative and positive work performed by soft tissues. After walking heelstrikes some energy was dissipated in soft tissue deformations, while some was stored and returned as damped elastic bouncing. Over a range of walking speeds we found that soft tissues performed about 60% of the total negative collision work. Each collision was then followed by some positive soft tissue work, indicating that soft tissue deformation may save muscles both the effort of actively dissipating energy, and the effort of performing positive work.

In Chapter 5 I investigated mechanical work as an indirect measure of subjective costs that might influence preferred movement strategies. Although people value economy of movement, it is not the only factor influencing preference. The amount of work a person chooses to perform may reflect a subjective valuation of the trade-offs between active muscle effort and other difficult-to-measure costs, such as pain. The degree to which work is not minimized may, therefore, indirectly quantify the relative valuation of these costs that are otherwise difficult to measure. We used a simple jump landing experiment to investigate this premise. We quantified the work humans prefer to perform to dissipate the energy of landing. We found that people preferred a strategy that involved performing 37% more negative work than minimally necessary across a range of landing heights. This then required additional positive work to return to standing rest posture, highlighting the cost of this preference. We also found that subjects were able to modulate the distribution of work between active and passive tissues. By choosing to perform more work through passive soft tissue deformations subjects were able to reduce the total mechanical work needed during landings. So people could have chosen to minimize muscle work during jump landings by dissipating more energy through soft tissue deformations, but instead we observed that they preferred to perform a consistent amount of extra muscle work, presumably to avoid other subjective costs.

Chapter 2.

Modeling the Benefits of Ankle Elasticity in Human Walking

Introduction

It has been proposed that elastic energy storage and return at the ankle can improve walking economy by reducing the mechanical and metabolic energy demands on muscles [1], [16], [17], [25]. Work performed elastically by the Achilles tendon reduces active work required of the calf muscles [14], [15], [26], [27], but this does not explain why this ankle work actually needs to be performed. After all, less energy would be consumed by the calf muscles by simply not contracting them, since even isometric contractions cost energy. We propose that the fundamental benefit of performing ankle work is the ability to reduce collision losses by redirecting the body center-of-mass (COM). Here, we use a simple walking model to show that ankle elasticity provides an elegant means of performing this function passively, one that can reduce the total energetic demands of walking, including work required from more proximal knee and hip muscles.

Humans perform a burst of positive ankle work at the end of stance phase in walking (Figure 2.9). Much of this ankle push-off work is performed passively by the recoil of elastic tissues in the lower-limb, notably the Achilles tendon [14], [16], [26], [28] and the plantar fascia [29]. The long, compliant Achilles tendon in series with the short muscle fibers of the ankle plantarflexors is particularly well-suited for elastic energy storage and return [17]. The tendon work itself is free since mechanical function of the tissue is decoupled from metabolic processes [30].

Meanwhile, this series ankle elasticity allows the soleus and gastrocnemius muscles to operate efficiently, performing primarily isometric and eccentric contractions [14], [15], [26], [27]. By minimizing the positive muscle work, which is metabolically more expensive than negative work or isometric muscle contractions [8–10], the calf muscle-tendon unit can operate more efficiently than muscle alone.

However, muscles acting about the hip and knee still perform metabolically expensive positive work during gait. At the hip in particular, far more positive work is performed than negative work (e.g., [31]), indicating active muscle contributions since passive tendons cannot perform net positive work. This muscle work is due in part to differences in architecture between proximal and distal muscles [32]. In contrast to the ankle plantarflexors, the muscle-tendon units acting about the more proximal joints have longer muscle fibers, shallower pennation angles and shorter tendons [33], making them better suited for performing muscle fascicle work [32], [34]. This, however, comes at the expense of higher metabolic energy consumption and lower efficiency for mechanical work generation [28]. Ideally, to minimize the overall metabolic cost of walking, muscle work would be avoided in favor of less expensive isometric contractions and elastic energy storage and return.

Theoretically, no muscle work is needed to power level ground, steady-state walking since the body undergoes no net acceleration (thus no net work) from one stride to the next. But to achieve purely passive, zero-muscle-work walking one must avoid energetic losses due to heelstrike collisions. Energy dissipated in collisions must be replaced by positive muscle work [4], [5], [11] regardless of whether the negative collision work was performed actively by muscles or passively by other soft tissues in the body [35]. Fortunately, there are ways to avoid collision losses by preemptively redirecting the body's COM velocity before heelstrike. Simple walking models have demonstrated mechanisms that can reduce collision magnitudes, such as preemptive push-off [4], [36], rolling feet [37], trunk elasticity [38] and ankle elasticity [39], [40]. These energy-saving mechanisms reduce the total amount of active work that must be performed to locomote. A growing body of experimental evidence suggests that humans use a number of these energy-saving mechanisms to walk more economically [5], [14], [29], [37], [41–45]. In theory, mechanical work requirements of walking could even be reduced to zero if the COM velocity were redirected to be perfectly perpendicular to the direction of force at heelstrike, essentially avoiding collisions altogether [36], [38]. Unlike running, which due to its aerial phase requires alternating periods of negative and positive work, there is no fundamental reason why negative collision work must be performed in walking.

In this study, we use computational modeling to systematically investigate the biomechanical benefits of series ankle elasticity on walking economy, while considering practical trade-offs. In contrast to previous approaches that have modeled a single muscle-tendon unit in isolation and assumed a certain magnitude of ankle work must be performed during walking, our approach embeds ankle elasticity into the whole-body dynamics. Therefore, the magnitude of elastic energy storage affects and is affected by other body segments and joints. We use this dynamic walking approach to investigate the benefits of passive ankle elasticity and show how elastic energy storage and return at the ankle can improve the overall economy of gait.

Methods

We developed a springy-ankle walking model and performed simulations to systematically investigate the effects of various parameters on the energetics and mechanics of locomotion.

Walking Model

We created a walking model with series ankle elasticity that could be powered by any combination of hip and ankle work. We extended the simplest walking model [12], [46] to include flat-feet and torsional springs at each ankle and between the hips at the pelvis (Figure 2.1). The two-dimensional, six degree-of-freedom model has a concentrated point mass, M, at the pelvis, and feet of infinitesimally small point mass m. Straight, rigid legs of length l are connected by a frictionless hinge at the pelvis, and rigid feet of length l_f are connected to the legs by frictionless hinges at the ankles. Ankle and pelvis (hip) springs are modeled as linear rotational springs, with stiffness k_{ank} and k_p, respectively. For simplicity these springs are depicted in figures as extension springs, similar to elastic tissues in the human body such as the Achilles tendon. The ankle spring has an additional free parameter, q_0, used to set the engagement point of the spring. We also defined a nominal walker model with dimensionless parameters: l=1, l_f=0.15, M=1 and q_0 = 0 (referring to spring engagement when the stance leg is perpendicular to the foot). The model's foot length of 0.15 was selected as an approximation of the horizontal distance from the ankle to the first metatarsal joint on the human foot. This length was used instead of human foot length because the model has no heel and the center of pressure during walking does not move all the way to the distal end of the toes [47].

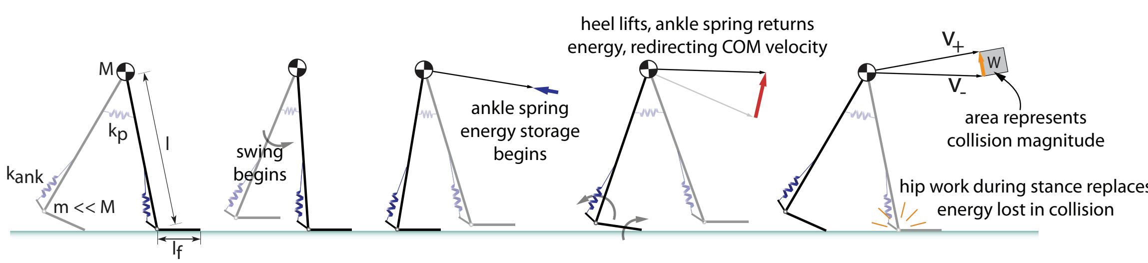


Figure 2.1: Springy ankle walking model, stop animation. The walker model has concentrated pelvic mass M, and feet of infinitesimal mass m, connected by leg length l, and with foot length l_f. Torsional springs are located at each ankle (k_{ank}), and between the legs at the pelvis (k_p), although these are depicted as extension springs for simplicity. The actuator in series with the ankle spring is not depicted here, but instead acts isometrically as a rigid strut. Although the model can perform ankle work (Figure 2.2), in this stop animation hip work during stance replaces energy lost in collision. The magnitude of energy stored in the ankle spring and the timing of this energy return affect the magnitude of collision, as defined by the difference in kinetic energy between the center-of-mass (COM) velocity immediately before (v_-) and after (v_+) heelstrike.

Walking Simulation

We performed steady-state walking simulations of this springy-ankle model. Collision with the ground was modeled as instantaneous and perfectly inelastic, with the subsequent state following impact found using conservation of angular momentum and impulse-momentum equations [11], [12], [46]. Scuffing of the foot during mid-swing phase was ignored. We defined nominal walking speed and step length as 0.4 (1.25 m/s) and 0.7 (0.7 m), respectively, based on typical human gait patterns [48].

Collisional energy losses were offset by some combination of hip and ankle work to maintain steady gait speed. Hip powering was approximated by positive work performed by gravity as the walker descended a gentle slope of angle γ, which has been shown to be similar to powering gait *via* constant hip torque [11], [49]. Hip work was approximately equal to γ for small slopes. Ankle work was performed by a virtual motor in series with the ankle spring. Ankle work loops were created using a step increase in torque at peak ankle flexion (Figure 2.2). During purely hip-powered gaits the ankle actuator acted as a rigid strut, and during purely ankle-powered gaits the ground slope was zero. We defined the sum of hip and ankle work per unit walking distance as the *mechanical cost of transport* (mCOT), and used this metric as an indicator of the minimum positive muscle work required to locomote. We defined the *optimal ankle stiffness* as the stiffness that minimized the mCOT for a given set of model parameters. A mCOT equal to zero indicated passive walking on level ground with no heelstrike collision losses and thus no need for active ankle or hip work.

First, we performed parameter studies on hip-powered walkers to investigate the effects of model parameters: spring stiffness, foot length and spring engagement point. Walking speed and step length were held constant by adapting the hip spring stiffness and ground slope. Other outcome variables of interest included ankle torque and power, energy stored in the ankle spring and energy returned before heelstrike (preemptive push-off). Next, we performed parameter studies on hip-powered walkers to explore the effects of gait parameters: speed, step length, and both speed and step length, assuming humans' preferred relationship (step length $\propto$

speed$^{0.42}$; [12], [48]). Hip stiffness and ground slope were again adapted to find gaits, while other physical parameters of the model were held constant, and speed and step length were systematically varied. Insight from these parameter studies was then used to investigate the possibility of zero-energy-cost gait.

Finally, we studied the effects of ankle vs. hip work on walking economy to determine which was more energetically optimal. We also compared the nominal method of ankle powering to an alternate method, which held peak ankle torque constant for a fixed duration after energy return began. We refer to these two methods as increasing-torque (nominal) and constant-torque ankle powering.

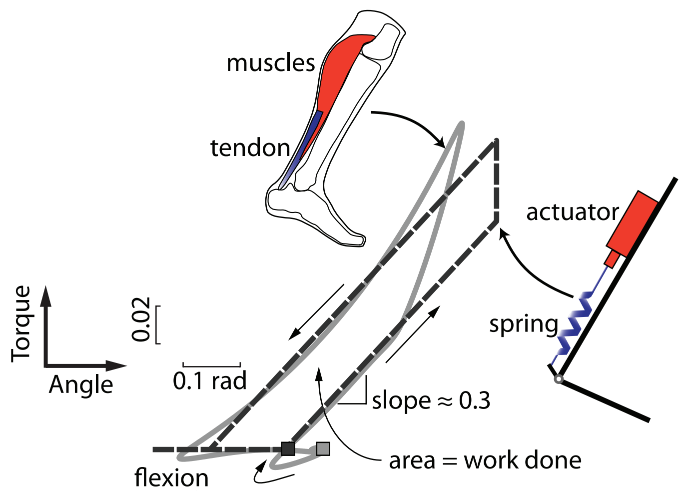


Figure 2.2: Ankle torque vs. angle during walking. Overlaid are ankle work loop profiles for walking at dimensionless speed of 0.4 (1.25 m/s). In humans (solid gray line), net positive work is performed about the ankle by the calf muscle-tendon unit. The model (dashed black line) uses a spring in series with a virtual motor to perform similar active powering. The model work loop is achieved by performing a step increase in torque at maximum ankle flexion, with the area between the curves representing the mechanical work done. Human tendon stiffness was estimated as about 0.3 (dimensionless), based on the slope of the torque vs. angle curve during initial loading. Heelstrikes are depicted for the human (gray box) and the model (black box). Normal human walking data (N=10) is from previous literature [35].

Results

We found that ankle work can reduce the energetic demands of walking by modulating heelstrike collisions, and that ankle elasticity offers an efficient means of performing this work. Maximizing the economy of walking depends on both ankle spring energy storage and the timing of energy return, which in turn rely on a careful balance of the model parameters (spring stiffness, foot length and spring engagement point) with gait characteristics (speed and step length). We discovered that there are indeed certain model configurations capable of walking with zero cost of transport by using elastic energy storage and return to redirect COM velocity and avoid collisions. We also observed that the optimal ankle stiffness remains roughly constant when speed and step length increase together along humans' preferred relationship. Finally, we found that ankle-powered gait is generally more economical than hip-powered gait.

Spring Stiffness

The mCOT was found to be strongly dependent on spring stiffness. We observed high mCOT for very low spring stiffness, followed by a fluctuating series of low and high mCOT as stiffness increases (Figure 2.3). Without an ankle spring, acceleration of the COM due to the falling inverted pendulum leads to a large collision (i.e., high mCOT). The addition of a soft spring allows some energy to be stored elastically and thus removed from the kinetic energy of the COM, essentially hiding that energy away and leading to a more moderate collision. The optimal spring bounces, storing and returning energy to redirect the COM velocity before heelstrike, thus minimizing collision. For instance, for the nominal walking model the minimum mCOT was 74% lower with optimal ankle spring than without a spring (mCOT 0.027 vs. 0.102). However, if the spring is too stiff, the preemptive energy return occurs too soon, and the COM velocity begins falling again before collision occurs, leading to a moderate to large collision magnitude (e.g., causing mCOT of nominal walker to be as high as 0.07). If the spring is even stiffer, the ankle will perform multiple bounces during a single stance phase. We found that the mCOT for these multi-bounce gaits was always higher than the minimum mCOT for gaits with a single cycle of energy storage and return (e.g., minimum mCOT of 0.027 for single-bounce vs. 0.030 for double-bounce, for nominal walker model).

The spring stiffness that maximizes peak ankle power was not found to maximize economy. This ankle stiffness tends to store substantial energy, but returns little preemptively. As the transition to the new stance leg unweights the trailing leg, elastic energy can be returned rapidly against the reduced load, leading to high ankle power. We found that the optimal ankle spring is stiffer than this power-maximizing stiffness and performs substantially more preemptive energy return (e.g., 56% vs. 6% for nominal walker model). The optimal spring is also slightly softer than the spring that returns all of its energy preemptively, consistent with prior literature [40].

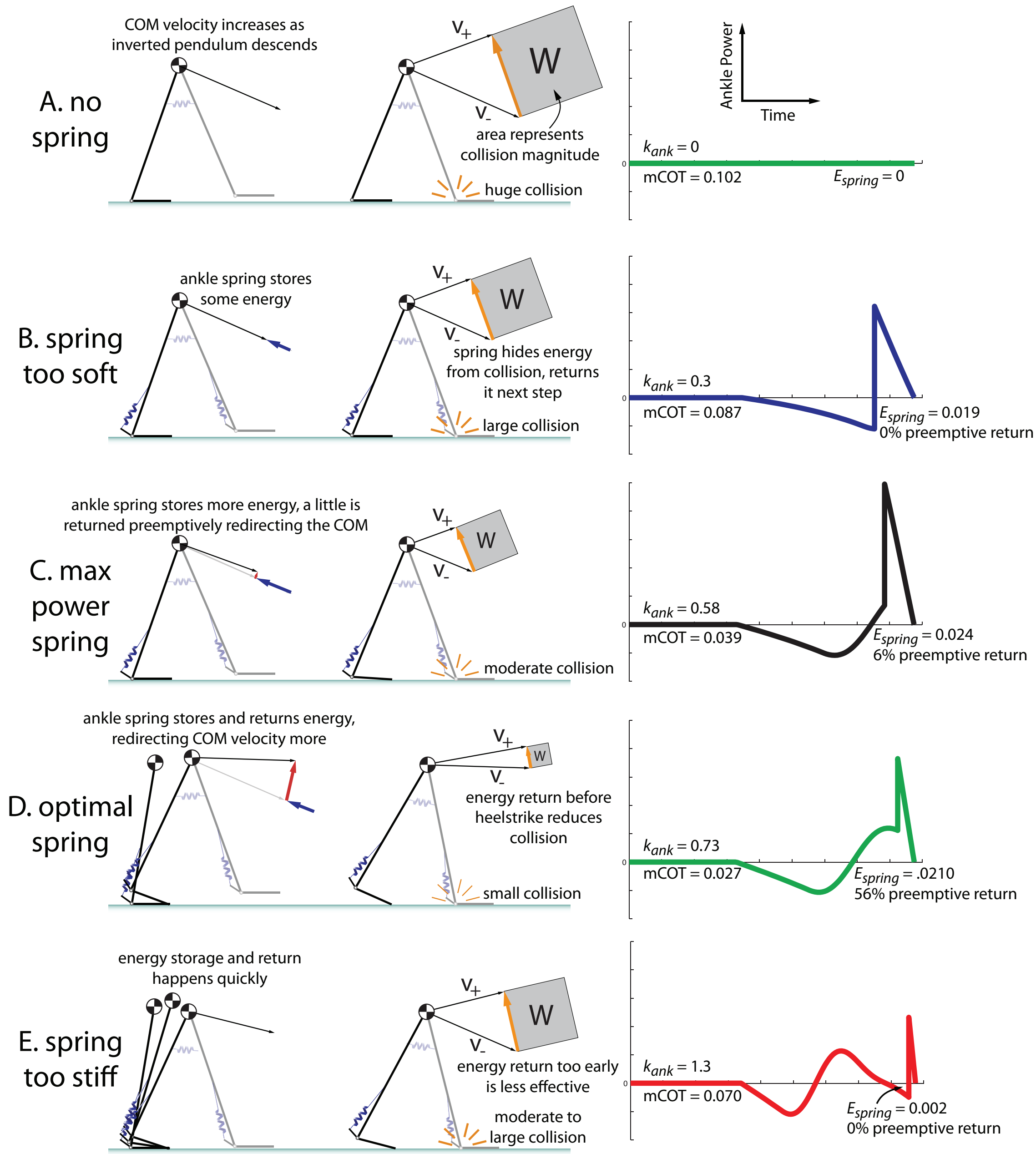


Figure 2.3: Effect of ankle spring stiffness. Optimal ankle spring stiffness is a balance between energy storage and timing of energy return. (A) Without a spring, acceleration of the center-of-mass (COM) due to inverted pendulum rotation leads to a very large collision. (B) The addition of a soft spring allows some energy to be stored in the spring and thus removed from the kinetic energy of the COM, but still leads to a relatively large collision. (C) The spring that maximizes the rate of energy return (power) stores substantial energy, hiding it from collision. It may also release a small amount of energy preemptively, but most of the return occurs after heelstrike when, due to unweighting of the trailing limb, it can be returned rapidly. (D) An optimal spring stores and returns energy with appropriate timing to redirect the COM velocity and minimize the magnitude of collision. (E) However, if the spring is too stiff, the energy return occurs too soon, and the COM velocity begins falling again before collision occurs, leading to a larger collision magnitude. The right column shows examples of ankle power for each case, based on the nominal model walking at 0.4 speed and 0.7 step length (both dimensionless). The mCOT, spring stiffness and energy storage are also reported to demonstrate an example of the trends. All values are reported in dimensionless units.

Foot Length

The mCOT is also strongly dependent on foot length, with minimum mCOT decreasing as foot length increases (Figure 2.4). We found that the optimal spring stiffness increases approximately linearly with foot length (Figure 2.10). In addition, longer feet lead to higher maximum energy storage, preemptive energy return, hip spring stiffness and peak ankle torques (Figure 2.4). We also observed that zero mCOT gaits are achievable for feet of sufficient length, typically greater than about half of step length (Figure 2.5). Long feet ensure sufficient energy can be stored in the spring to fully redirect the COM velocity to be perpendicular to the new stance leg at heelstrike. Once feet are long enough, there are many combinations of model parameters that can achieve passive, collision-free walking.

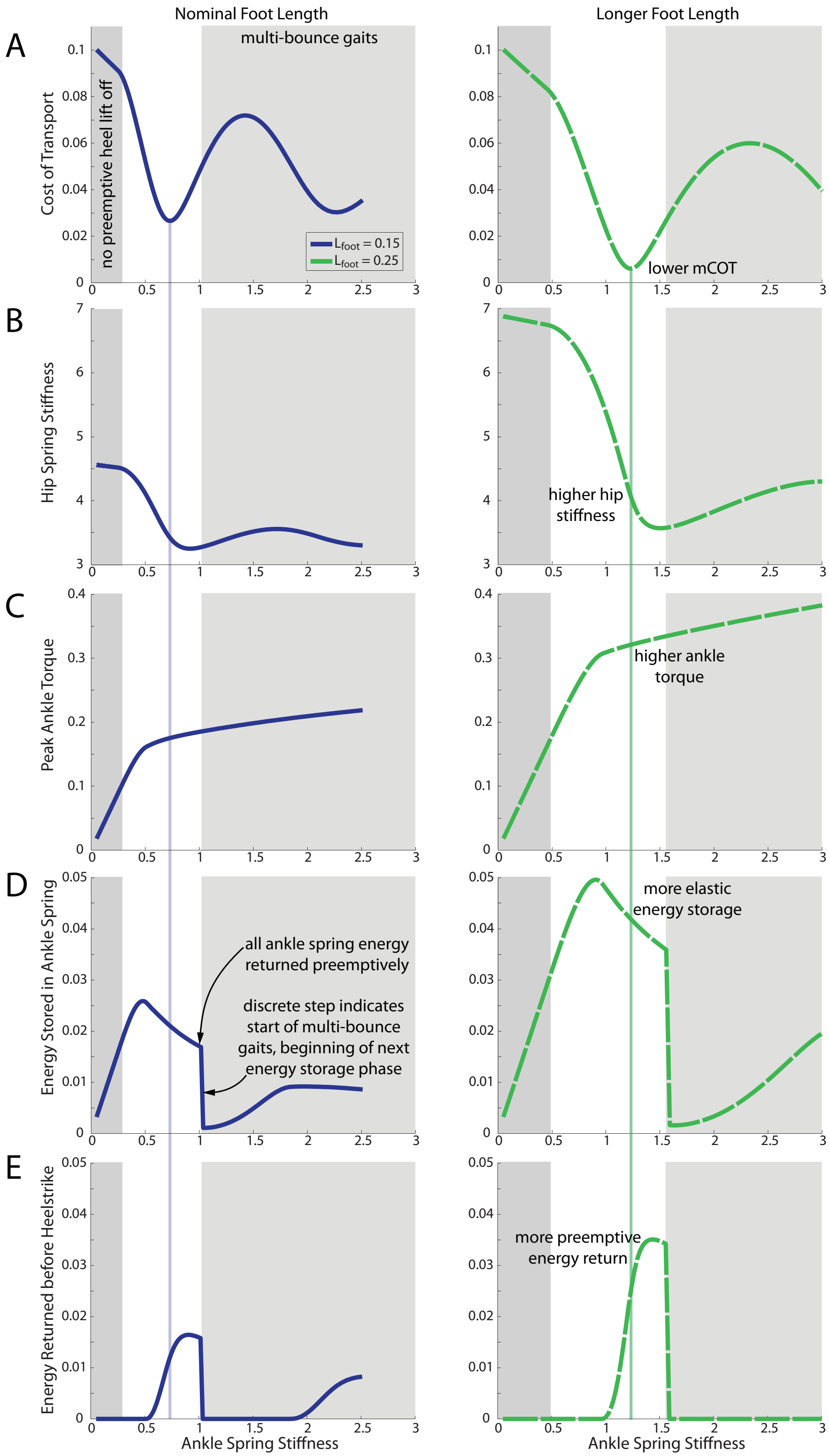

Nominal Foot Length
Longer Foot Length
A
B
C
D
E
Cost of Transport
Hip Spring Stiffness
Peak Ankle Torque
Energy Stored in Ankle Spring
Energy Returned before Heelstrike
Ankle Spring Stiffness
multi-bounce gaits
no preemptive heel lift off
$L_{foot} = 0.15$
$L_{foot} = 0.25$
lower mCOT
higher hip stiffness
higher ankle torque
all ankle spring energy returned preemptively
discrete step indicates start of multi-bounce gaits, beginning of next energy storage phase
more elastic energy storage
more preemptive energy return

Figure 2.4: Effects of spring stiffness and foot length. The left column shows results for the walker model with nominal foot length (0.15) and the right column for longer foot length (0.25). (A) Mechanical cost of transport (mCOT) is initially high, and then follows an oscillating pattern of decreasing and increasing. The minimum mCOT achievable decreases as foot length increases. Vertical lines show optimal ankle stiffness (i.e., stiffness that yields minimum mCOT) for each walker model. (B) Hip stiffness tends to follow fluctuations in mCOT. (C) Peak ankle torque increases with spring stiffness, changing slope due to heel lift and elastic energy return. For the case of optimal spring stiffness, (B) hip spring stiffness and (C) ankle torque increase with longer foot length. (D) Energy stored in the spring increases with spring stiffness to a point, but then decreases when energy begins to be returned preemptively. (E) There is no energy returned before heelstrike until a point after the heel lift condition is met. Then preemptive energy return increases with spring stiffness until all the energy is returned before heelstrike. The cycle then resets to zero, following a similar pattern for multi-bounce gaits. (D) Energy stored in the spring and (E) energy returned before heelstrike both increase with longer feet. All models walked at fixed speed (0.4) and step length (0.7) and used the nominal (0 rad) engagement angle, corresponding to spring engagement when the stance leg is perpendicular to the foot. Shaded regions show when ankle stiffness is too low to allow preemptive heel lift, and when it is stiff enough to induce multi-bounce gaits. All values are reported in dimensionless units.

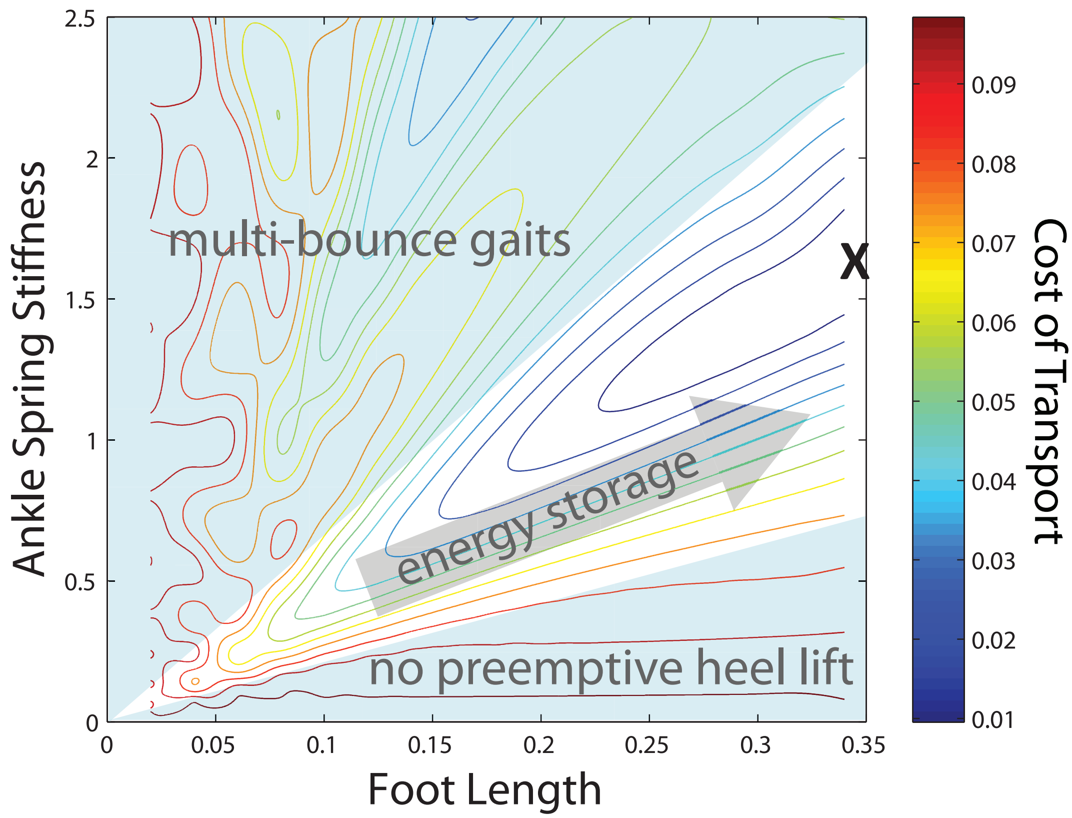


Figure 2.5: Contour plot of mechanical cost of transport (mCOT) as a function of foot length and ankle spring stiffness. To achieve zero mCOT a spring must store sufficient energy and return it with appropriate timing to avoid collisions. If the spring is too soft the heel will not lift off the ground and no preemptive energy release will occur. If the spring is too stiff multiple cycles of elastic energy storage and return will occur, also degrading economy. The maximum amount of energy that can be stored by the spring increases with foot length, and generally achieving zero collision gaits required

foot length greater than about half of step length. The black X on the contour plot shows the approximate region of zero energy cost gaits. The contour plot contains oscillatory features due to spring stiffness and a generally decreasing trend with longer feet. The concave-up "bowls," observed by varying spring stiffness, become a "valley" along the direction of foot length; however, this "valley" is oriented in a diagonal direction because the optimal ankle stiffness tends to increase with foot length. Other "valleys" exist at higher spring stiffness due to multi-bounce gaits. The "valley" at the lowest spring stiffness represents a single cycle of energy storage and return for the ankle spring. Gait speed is 0.4 and step length is 0.7. All values are reported in dimensionless units.

Spring Engagement

In general, it is more economical for the ankle spring to engage early in stance. The earlier the engagement, the softer the spring can be. And the softer the spring, the more energy can be stored elastically. However, the effect of spring engagement point on the minimum achievable mCOT is small compared to the effects of foot length and spring stiffness (Figure 2.11). For the nominal walker model, the mCOT only changed from 0.030 for late spring engagement (at approximately 80% of stance phase) to 0.024 for early spring engagement (at approximately 20% of stance phase). Nevertheless, in order to walk economically the spring engagement must still be selected appropriately to complement the other model parameters. As spring engagement becomes later, optimal stiffness increases (Figure 2.10). Early spring engagement also leads to slightly lower peak ankle torque, but does require higher hip spring stiffness to achieve equivalent speed and step length. As feet get longer, there is a limit to how early the spring can engage and still perform preemptive push-off. This is because near the end of energy return, when long feet are plantarflexed, the COM height can be greater than leg length, preventing leading leg from reaching the ground. Although this configuration could lead to some other mode of locomotion, we did not investigate it in this study of walking.

Speed & Step Length

Optimal ankle stiffness increases with speed and decreases with step length (Figure 2.6). The increase with speed was roughly linear for the ranges and models tested, whereas the decrease in optimal stiffness with step length appeared to fit a function of form (step length)$^{-N}$, where N was less than 1. Interestingly, these competing trade-offs largely offset when speed and step length increase along humans' preferred relationship (step length ∝ speed$^{0.42}$; [12], [48]). The optimal ankle stiffness was therefore found to be approximately constant across a wide range of speeds (approximately 0.6 to 1.9 m/s). For the nominal walking model we found optimal ankle stiffness only varied by 0.05 (Figure 2.6, 0.80 to 0.75 from slow to fast gait speed). We also found this result to be fairly independent of foot length. For much longer feet (0.25) the optimal ankle stiffness only changed by 0.06 (1.27 to 1.33, slow to fast), and for much shorter feet by 0.09 (0.25 to 0.16, slow to fast).

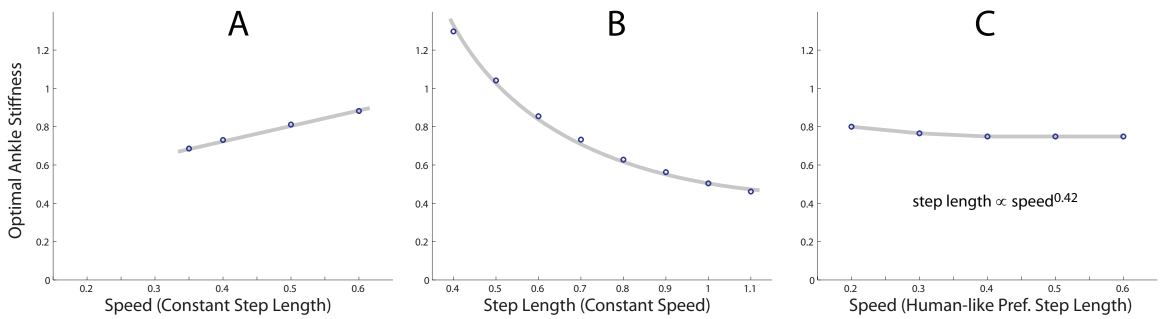


Figure 2.6: Optimal ankle stiffness as a function of speed and step length. Optimal ankle stiffness (A) increases with speed and (B) decreases with step length. (C) When speed and step length increase together along human's preferred relationship (step length ∝ speed$^{0.42}$), optimal ankle stiffness is approximately constant across a wide range of gaits. All values are reported in dimensionless units. Dimensionless speed 0.2 corresponds to approximately 0.6 m/s and 0.6 to 1.9 m/s.

Ankle Work

In general, powering gait *via* ankle work is more economical than hip powering because it can increase push-off work and cause redirection of the COM to occur more preemptively (Figure 2.7). The benefit is small for softer ankle springs when elastic energy is not returned preemptively (Figure 2.12). Near optimal ankle stiffness, when preemptive energy release does occur, the benefit of ankle work is greater (Figure 2.12). The new optimal stiffness, given the ankle work assistance, is slightly softer than the passive optimal stiffness since the ankle work causes earlier redirection of COM velocity. For the nominal walker model with optimal ankle spring, a purely ankle-powered gait (walking on level ground) saved about 40% of the mCOT compared to a purely hip-powered walker (0.016 vs. 0.027 mCOT; Figure

2.7). The energy-saving effect of ankle work becomes even larger as foot length decreases. As springs become stiffer than optimal the benefits of ankle work are reduced. In these cases, ankle work can cause push-off to be too early (i.e., poorly timed), leading to larger collisions. The type of ankle work loop performed had little effect on the results. The mCOT and optimal ankle stiffness for constant-torque ankle work loops were found to be nearly identical to results from increasing-torque work loops. We did not explore active powering for ankle stiffness that led to multi-bounce gaits.

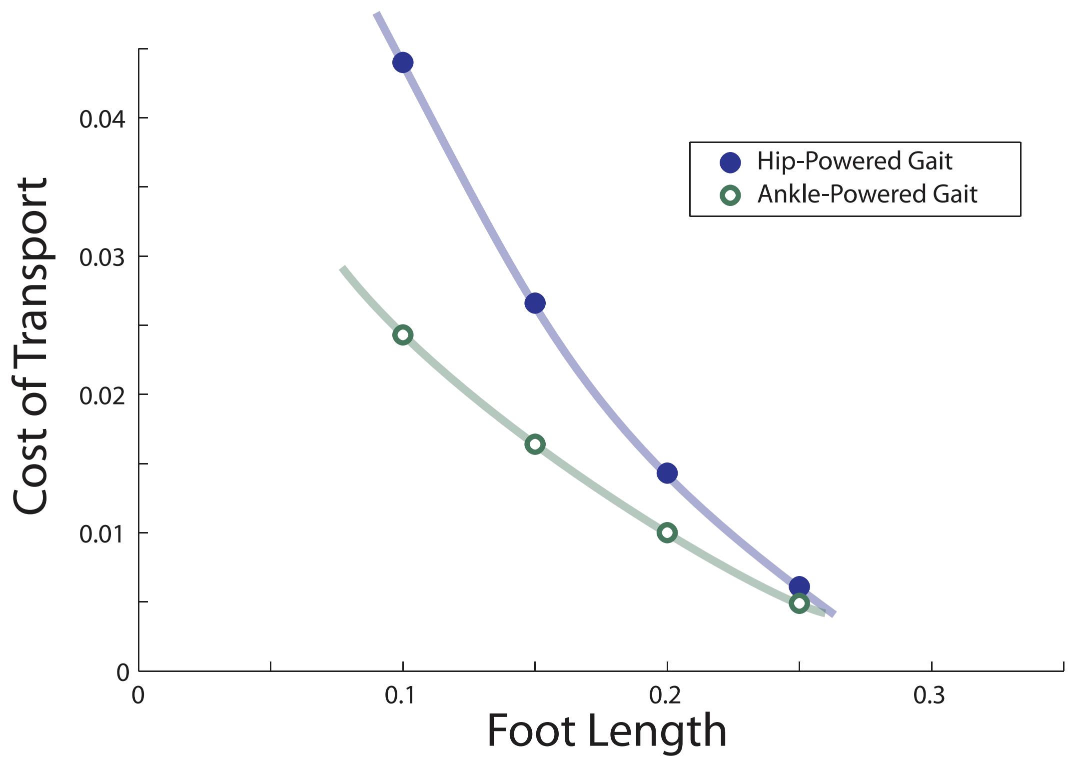


Figure 2.7: Ankle-powered vs. hip-powered gait. Ankle-powered walkers were generally found to have a lower mechanical cost of transport (mCOT) than hip-powered walkers. This difference was largest for short foot length. Both walker models approached zero mCOT for longer foot length, typically greater than about half step length. Step length was 0.7 and gait speed 0.4 for all walkers. Optimal passive ankle stiffness was used for each foot length. All values are reported in dimensionless units.

Discussion

Although there is no fundamental need to perform ankle work during walking, we use a computational model to show that elastic ankle work can reduce the total mechanical work of walking, in some cases even completely removing the need for active muscle work. The primary mechanism that enables these zero-energy-cost gaits is the ability of the ankle to perform work that preemptively redirects COM velocity, minimizing collision energy losses. In our model, the two key requirements for using ankle elasticity to achieve passive gait are: (1) heel lift and elastic energy release must occur before contralateral heelstrike and (2) the spring must store sufficient energy to redirect the COM velocity. When these conditions are met there are many combinations of model parameters that yield walking with zero active work. When model parameters are insufficient to achieve purely passive gait, ankle powering can help improve walking economy.

Our model suggests that series ankle elasticity can benefit walking economy in two main ways: by redirecting COM velocity before heelstrike or to a lesser extent by effectively hiding energy from collisions. The model elucidates the fundamental benefits of ankle elasticity in reducing collisions and thus the minimum muscle work required during gait. As elastic energy is stored it is siphoned out of the kinetic energy of walking, slowing the person down. The return of this elastic energy speeds the person back up, but since energy is roughly conserved (i.e., elastic tissues in the body perform nearly zero, but slightly negative net work), this energy return only replaces the kinetic losses that the spring itself caused. Therefore, elastic energy storage and return does not directly replace work done by muscles, but rather by modulating heelstrike collisions it reduces the total amount of active positive work that must be performed. While there is no fundamental need for ankle work during walking, our model offers an explanation for the energy-saving benefits of performing elastic energy storage and return. This explanation differs from previous literature [14], [15], [26], [27], which has primarily focused on the advantages of tendon elasticity in reducing calf muscle work. Our findings suggest that energy savings from elastic ankle work outweigh the metabolic costs associated with isometric contractions of the muscles in series. Ankle elasticity seems to be an efficient way to reduce the total work required of muscles and improve the overall economy of walking.

We found that it is possible for the springy-ankle model to walk with zero active work, but there may be other factors that explain why humans do not. For passive walking the ankle torque must become large enough to lift the heel off the ground and release elastic energy before contralateral heelstrike. The ankle spring must also store sufficient energy to fully redirect the COM velocity. The simulation indicates that these requirements can only be achieved when the foot is sufficiently long, greater than about half of step length. Therefore, for most physiologically relevant speeds and step lengths the model required long feet (relative to human anatomy) to walk passively. The longer feet lead to larger ankle flexion torque in the model, which in humans would generally correspond to higher knee extension torque. People would, therefore, need to counteract this torque actively using their knee flexor muscles to avoid knee hyperextension during the stance phase of gait, which could be painful. Alternatively, humans could choose to walk with a more flexed knee during stance to reduce the moment arm due to ground reaction forces near the toe, but we expect both of these options would incur a metabolic penalty. This suggests the importance of biarticular muscles such as the gastrocnemius, which help counteract knee extension as ankle torque increases. The higher hip spring stiffness for longer feet suggests there may also be additional metabolic costs related to leg swing, specifically due to the high rate of muscle force needed during early swing [12], [50], [51]. Finally, long feet introduce practical issues such as toe clearance during swing phase that could heighten the risk of tripping. An alternative to long feet might be to take shorter steps, but this approach is also limited by the leg swing costs needed to maintain speed.

Maximizing power output using series ankle elasticity does not appear to be a critical aspect of maximizing walking economy, rather simple bouncing seems to be most economical. Simple bouncing due to the natural dynamics of walking happens when ankle torque due to compression of the passive spring exceeds the torque due to body mass, thus allowing the spring to recoil (before the contralateral limb heelstrike). This causes the preemptive redirection of the COM that minimizes collisions. It has been suggested that slow elastic energy storage followed by a rapid rate of energy return may be a vital feature for minimizing energy consumption during walking [14]. But our model suggests that maximizing the rate of energy return may be sub-optimal during gait because the timing of energy return is not sufficiently preemptive. Using series elasticity to return energy at rates higher than muscle capacity, a process known as power amplification, is a mechanism used by some insects, frogs and birds to improve jump performance [52–56]. However, as noted by Roberts and Azizi [30] the use of series elasticity in human walking is distinct from the power amplification used in these creatures, since human muscle contractions are not the direct source of elastic strain energy. Our model suggests that maximizing the rate of energy return may even be detrimental to walking because it actually tends to throw the walker model into the air, requiring negative vertical ground reaction forces to "hold" the stance leg on the ground (Figure 2.14). This effect may partially be due to our modeling of collisions as instantaneous and inelastic. Finite-time collisions would cause a slower unloading of the trailing limb, which would reduce peak powers compared to the instantaneous transition in the model, but would still facilitate the return of elastic energy at a rate higher than it was stored. It appears that in the absence of mechanical latches, humans and other biological systems can use passive dynamics to modulate the storage and release of elastic energy.

In humans, the return of elastic ankle energy does not appear to be from simple spring bouncing, suggesting that the Achilles tendon stiffness may be softer than optimal. Elastic energy return instead seems to be initiated by another means. Ankle torque-angle curves (Figure 2.8) suggest that the rapid stiffening of the ankle joint may be the primary cause, enabling heel lift off and elastic energy return at higher walking speeds (>1.25 m/s). This stiffening is likely due to muscle contractions, but could also be influenced by passive stiffening of the joint. At lower speeds the timing of energy return closely corresponds to the timing of limb unloading, which would facilitate heel lift and energy release at a lower ankle torque (Figure 2.15). This limb unloading contributes to the increase in rate of elastic energy return as compared to the rate of energy storage, highlighting how natural walking dynamics can modulate the power of elastic energy return. Interestingly, simulation studies using humans' preferred speed-step length relationship suggest that constant ankle stiffness, such as provided by the Achilles tendon, could be equally well suited for walking at both slow and fast speeds. Our findings that suggest the Achilles tendon may be sub-optimal for walking differ from the conclusions drawn by Lichtwark and Wilson [27], who used a Hill-type muscle model to predict that Achilles stiffness is approximately optimal to maximize efficiency of the gastrocnemius. However, a limitation of their approach is that minimizing energy consumption of a single muscle does not guarantee the overall economy of locomotion is optimal. Furthermore, their assumption that varying Achilles stiffness does not change the kinetics of the ankle or more proximal joints may discount concomitant changes in gait dynamics. We demonstrate that walking at a fixed speed and cadence with zero calf muscle work can be achieved using a wide range of ankle stiffness, but that the stiffness chosen affects the amount of work that must be performed by muscles elsewhere in the body. We do expect that tasks other than level-ground walking could potentially benefit from different or adjustable ankle stiffness. In particular, tasks such as landing that involve energy absorption may benefit from a more compliant tendon [57], [58].

When foot length and spring stiffness are not adequate for zero-energy-cost gait, active ankle work can be used to reduce the mCOT. The ability of ankle elasticity to perform preemptive push-off work that redirects the COM upward seems to be the crucial beneficial factor. Thus, the major benefits of ankle work were only found for spring stiffness that yielded preemptive timing of energy return, and the type of work loop (increasing-torque vs. constant-torque) did not have a large effect on the model's walking economy. The ankle may be unique amongst the lower-limb joints given its ability to redirect the COM, since the knee is already near full extension during stance and hip powering is largely orthogonal to the desired redirection. However, hip powering may have more substantial benefits for a more complex model with distal trunk mass. A previous model by Gomes et al (2011) has demonstrated that substantial oscillations of an elastic trunk can be used to modulate collision losses. Without a trunk the hip work has little ability to preemptively redirect the COM vertically. There may of course be other powering alternatives, such as using hip work to increase kinetic energy of the body early in stance, which is then transferred into the ankle spring; however, it is unclear if there would be advantages to approaches such as this.

The advantage of simple walking models is that they promote the understanding of fundamental principles, which can then be applied to other areas. In particular, modeling the benefits of series ankle elasticity may have implications for prosthetic foot design. Conventional prosthetic feet have significantly reduced push-off capabilities compared to the natural ankle/foot [59], [60], which affects heelstrike collisions and may contribute to higher metabolic cost of walking for amputees [43], [44], [61]. Our simulation findings suggest that this lack of push-off, the inability to help redirect the COM velocity, increases heelstrike collision losses and thus the mechanical work required for walking. The importance of COM velocity redirection in the design of prosthetic feet has been supported by empirical evidence demonstrating reductions in metabolic cost with increased push-off [43], [45]. However, it is still unclear why Dynamic Elastic Response (DER) prostheses, which store and return energy in spring composite forefoot keels, do not provide metabolic savings over less elastic conventional feet (e.g., Solid-Ankle Cushioned Heel prosthesis) [18], [62], [63]. DER feet span a wide range of stiffness, including the optimal stiffness estimates from our model, but little quantifiable evidence exists to suggest one stiffness is superior to another. One possibility is that potential benefits are mitigated by energy dissipated in DER prostheses due to hysteresis (Figure 2.13), which has been estimated as 10-40% in these feet [64], [65]. Another possibility is that the lack of biarticular linkage (e.g., gastrocnemius) across the ankle and knee joints may make prosthetic feet more energetically expensive to walk on given the need for amputees to recruit other muscles to counteract knee extension moments. Certainly other complications and limitations may be due to the physical interface between amputees and their prosthetic feet.

There are also limitations to our simple walking model. The idealized 2-D model uses many simplifications, assuming point mass, instantaneous collisions, and no heels, knees or upper body. The model provides trends, but no absolute values that can be directly compared to experimental findings. We focused on the mechanical work of the body's COM during walking as a surrogate for metabolic cost due to positive muscle work. The simplified walker model had an infinitesimally small foot mass, thus we did not investigate costs associated with deceleration of the foot or leg at foot strike. We did not model muscles directly, nor did we account for costs due to force or rate of force production (aside from some limited discussion of potential costs associated with leg swing). However, these simplifications seem reasonable given the much lower efficiency of positive muscle work compared to negative work or isometric force production [8–10], [28]. We also did not examine the effects of ankle work and elasticity on other outcomes, such as stability or energy absorption abilities. There are practical limitations in applying model predictions as well. For example, with regards to assistive devices there is a limit to how stiff springs can be to interface with the deformable human body and still achieve desired performance. Additionally, the human body has other segments, such as the trunk, and other degrees of freedom, such as the knee, that could enable behaviors outside the capabilities of the model. Nevertheless, we believe this dynamic walking approach is useful for elucidating the fundamental mechanisms underlying the locomotion of more complex systems. While our focus was on ankle elasticity, we also recognize that there are other mechanisms that could enable zero-energy-cost walking without a preemptive redirection of the COM. For instance, there are models that can theoretically walk with zero energy loss by recycling energy from collision, such as with an axial spring in series with the leg [66], [67]. Overall, the modeling here is meant to give fundamental insight into walking with series ankle elasticity, but ultimately these predictions must be evaluated empirically.

In summary, we suggest a fundamental mechanism to explain the energy-saving benefits of ankle elasticity during walking. Although work does not need to be performed about the ankle, choosing to do so can greatly reduce collisions and thus the total muscle work required to walk. Ankle elasticity provides a simple and efficient means of performing this ankle work.

Supplementary Material

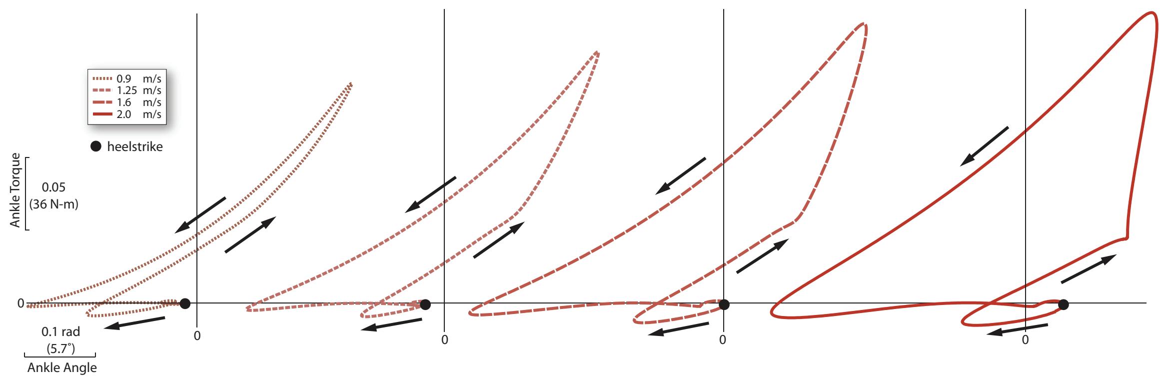


Figure 2.8: Human ankle torque vs. angle work loops across walking speed. The counter-clockwise work loops show positive work performed about the ankle during gait. The magnitude of positive ankle work (area between the curves) increases with speed. Stiffening of the joint during load up also becomes more drastic as speed increases. Data from normal human walking [35] is reported here for reference (N=10).

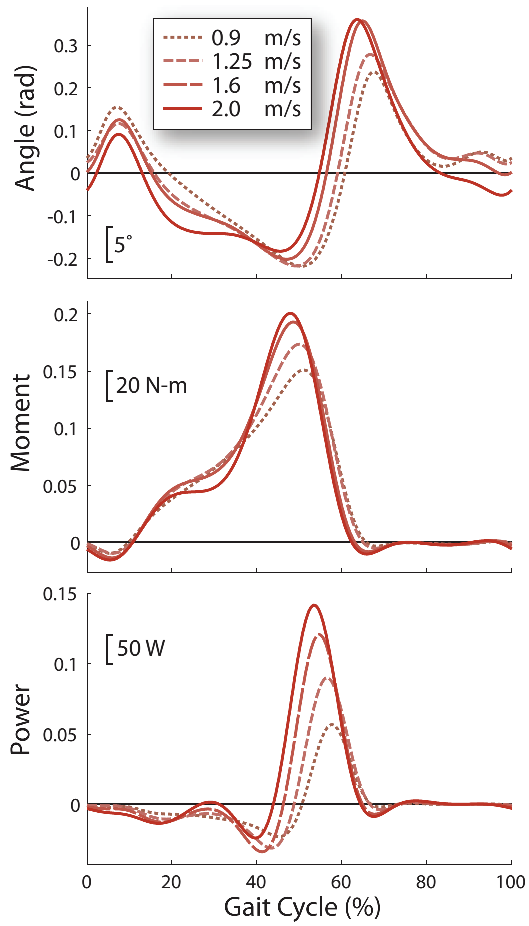


Figure 2.9: Human ankle angles, moments and powers across walking speeds. Data from normal human walking [35] is reported here for reference (N=10).

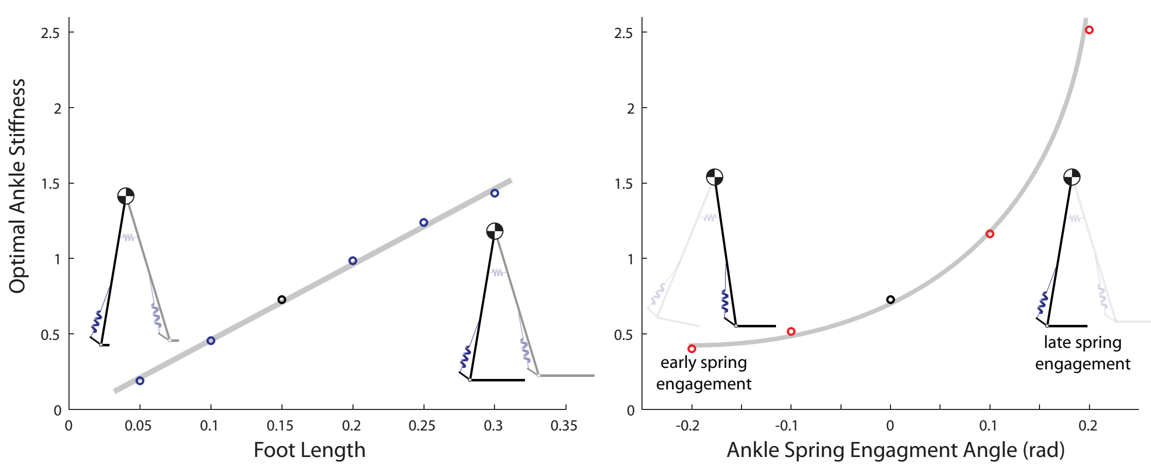
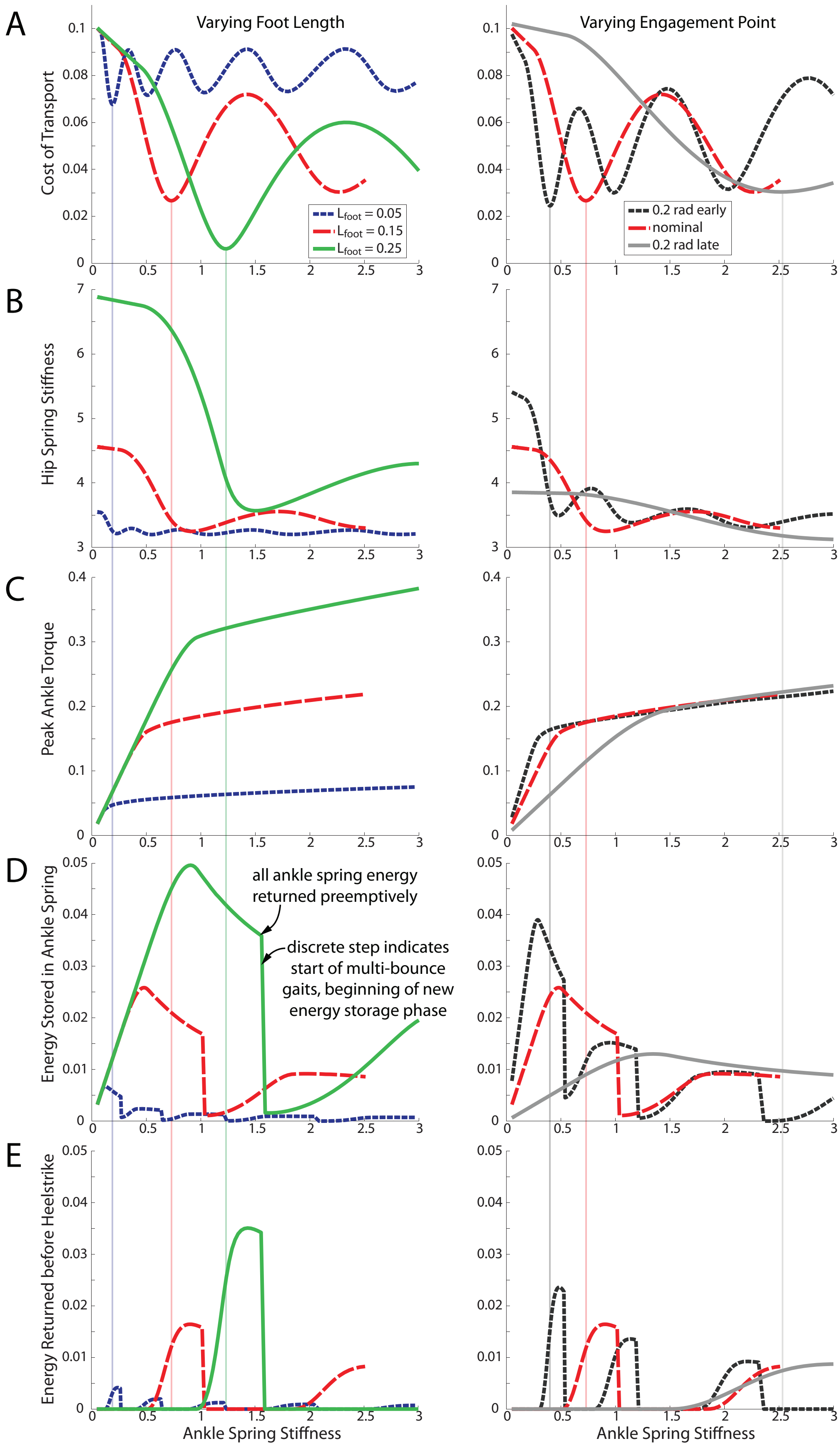


Figure 2.10: Effects of foot length and spring engagement angle on optimal ankle stiffness. Optimal ankle stiffness increases with (A) foot length and (B) later spring engagement. The foot length parameter study was performed using a nominal spring engagement point (0 rad), which refers to

the part of the gait cycle when the leg is perpendicular to the foot. The spring engagement parameter sweep was performed with a nominal foot length of 0.15. All models walked at fixed speed (0.4) and step length (0.7). Values are reported in dimensionless units.



A
B
C
D
E
Varying Foot Length
Varying Engagement Point
Cost of Transport
Hip Spring Stiffness
Peak Ankle Torque
Energy Stored in Ankle Spring
Energy Returned before Heelstrike
L_{foot} = 0.05
L_{foot} = 0.15
L_{foot} = 0.25
0.2 rad early
nominal
0.2 rad late
all ankle spring energy returned preemptively
discrete step indicates start of multi-bounce gaits, beginning of new energy storage phase
Ankle Spring Stiffness
Ankle Spring Stiffness

Figure 2.11: Effects of foot length and spring engagement angle. (A) The minimum mechanical cost of transport (mCOT) decreases as foot length increases and as spring engagement becomes earlier. Vertical lines show optimal ankle stiffness for each walker model. (B) For the optimal spring case, hip spring stiffness increases with longer foot length and earlier spring engagement, and (C) peak ankle torques increase with longer foot length and to a lesser extent with later spring engagement. (D) Energy stored in the spring and (E) energy returned before heelstrike are higher for longer feet and earlier spring engagement. All models walked at fixed speed (0.4) and step length (0.7). The foot length parameter sweep was performed using the nominal (0 rad) engagement angle, corresponding to engagement when the stance leg is perpendicular to the foot. And the spring engagement sweep was performed with the nominal foot length (0.15). Therefore, the red dashed lines in each parameter sweep are identical. Values are reported in dimensionless units.

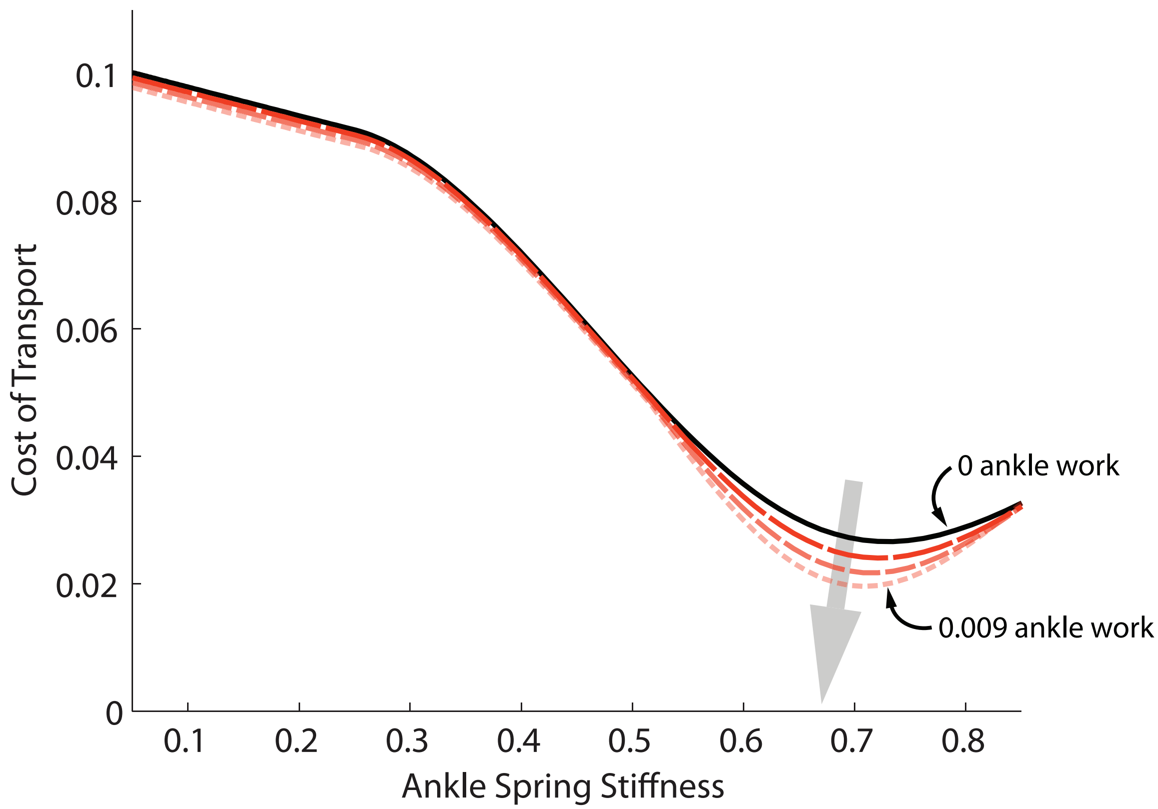


Figure 2.12: Effect of positive ankle work. Increasing ankle work reduces the minimum cost of transport (mCOT) substantially for spring stiffness near optimal, and reduces it slightly for softer springs. However, for ankle springs that are too stiff, benefits are reduced because timing of energy return becomes too preemptive. Parameter sweep was performed using the nominal walker model walking at 0.4 speed and 0.7 step length. Values are reported in dimensionless units. Note that the mCOT curves indicate the sum of active work performed by both the hip and the ankle.

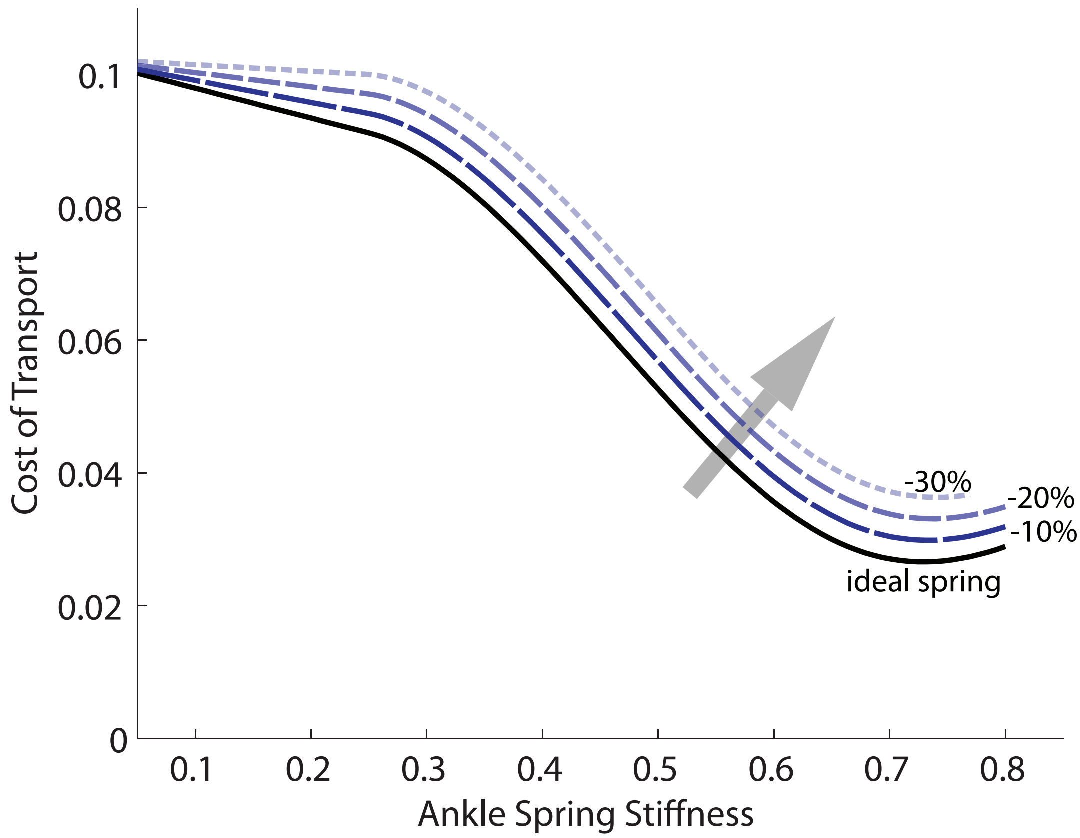


Figure 2.13: Effect of ankle spring hysteresis. Energy losses in the spring reduce the benefits of ankle elasticity. The effect appears to be relatively consistent across most spring stiffness, with the exception of extremely soft springs. The effect of hysteresis may be relevant to the efficacy of prosthetic feet that use springy composite forefoot keels to store and return energy during gait, which are known to exhibit hysteresis of about 10-40% [64], [65].

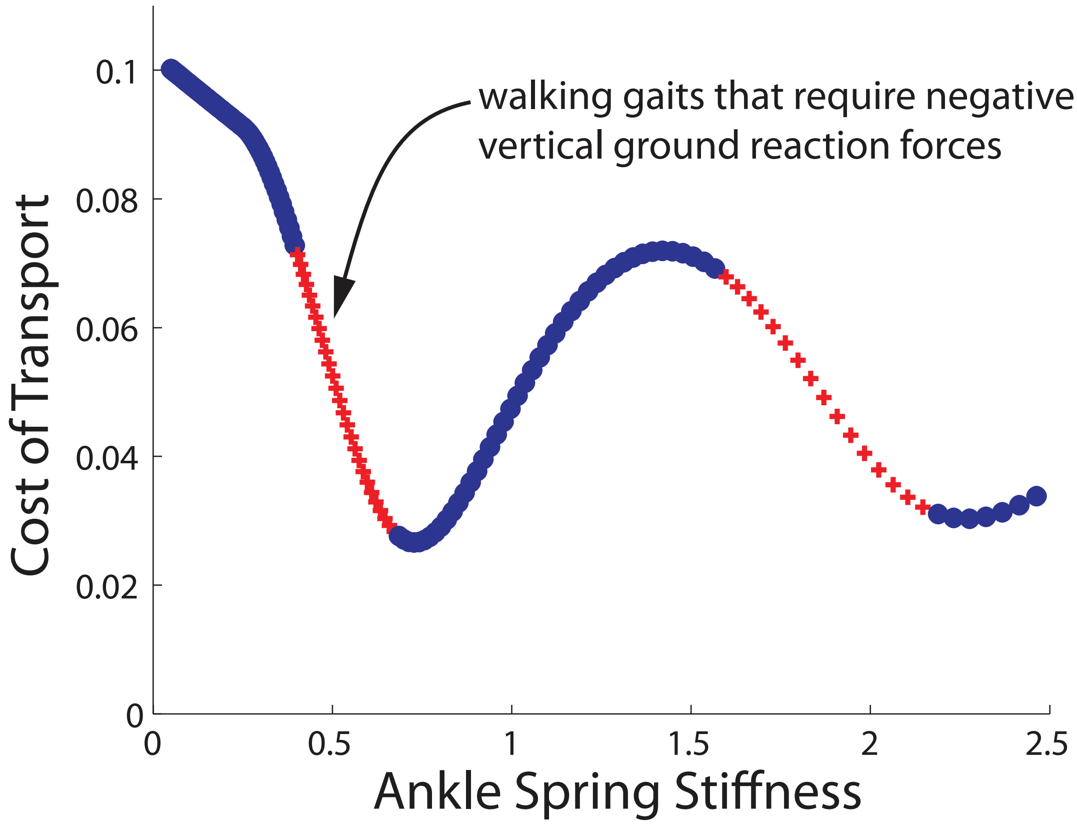


Figure 2.14: Walking model gaits that require negative vertical ground reaction forces. In some cases, the return of energy from the trailing limb's ankle spring would tend to throw the walker model into the air. We allowed the model to perform negative ground reaction forces, essentially using the leading stance limb to hold onto the ground. This is not possible in normal human gait. Here, we report these sets of solutions that perform this "illegal" gait. This does not affect gaits at minimum mCOT, but may suggest why slightly softer springs, for instance ones that maximize the rate of energy return, may not be most conducive for walking economically. Representative data shown here is for the nominal walker model, speed (0.4) and step length (0.7). All values are reported in dimensionless units.

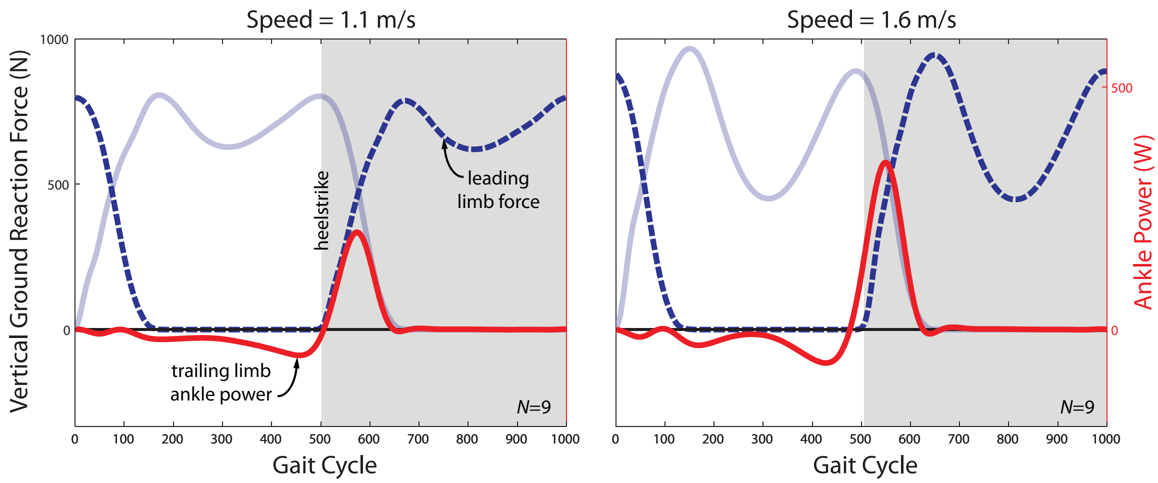


Figure 2.15: Timing of positive ankle power relative to heelstrike in human walking. Vertical ground reaction forces for both limbs, and trailing limb ankle power are shown. The white to gray background transition delineates leading limb heelstrike. At slower speeds (e.g., 1.1 m/s) positive ankle power coincides with contralateral heelstrike, suggesting the transition of weight from trailing to leading limb may facilitate the release of elastic ankle energy. At faster speeds (e.g., 1.6 m/s) the positive ankle power precedes contralateral heelstrike, suggesting that a different mechanism allows the ankle to return elastic energy before the limb begins unloading. Human ankle torque vs. angle data (Figure 2.8) suggests that stiffening of the joint, likely from muscle contraction, allows this preemptive push-off. Force and ankle power data is from normal human walking (*N*=10) [35].

Chapter 3.

Systematic Variation of Prosthetic Foot Spring Affects Center-Of-Mass Mechanics and Metabolic Cost during Walking

Published in IEEE TNSRE (2011).

Abstract

Lower-limb amputees expend more energy to walk than non-amputees and have an elevated risk of secondary disabilities. Insufficient push-off by the prosthetic foot may be a contributing factor. We aimed to systematically study the effect of prosthetic foot mechanics on gait, to gain insight into fundamental prosthetic design principles. We varied a single parameter in isolation, the energy-storing spring in a prototype prosthetic foot, the Controlled Energy Storage and Return (CESR) foot, and observed the effect on gait. Subjects walked on the CESR foot with three different springs. We performed parallel studies on amputees and on non-amputees wearing prosthetic simulators. In both groups, spring characteristics similarly affected ankle and body center-of-mass (COM) mechanics and metabolic cost. Softer springs led to greater energy storage, energy return and prosthetic limb COM push-off work. But metabolic energy expenditure was lowest with a spring of intermediate stiffness, suggesting biomechanical disadvantages to the softest spring despite its greater push-off. Disadvantages of the softest spring may include excessive heel displacements and COM collision losses. We also observed some differences in joint kinetics between amputees and non-amputees walking on the prototype foot. During prosthetic push-off, amputees exhibited reduced energy transfer from the prosthesis to the COM along with increased hip work, perhaps due to greater energy dissipation at the knee. Nevertheless, the results indicate that

spring compliance can contribute to push-off, but with biomechanical trade-offs that limit the degree to which greater push-off might improve walking economy.

Introduction

Prosthetic foot technology has undergone substantial transformation over recent decades, most notably with the introduction of elastic materials and mechanisms. Despite a wide variety of implementations, there remain disadvantages to amputee gait that lead to greater fatigue, secondary disabilities and reduced mobility [18], [19], [20], [21], [22], [23], [24]. Further improvement of prostheses is hindered by a lack of quantitative principles for design and prescription. By systematically studying specific mechanical parameters and quantifying the effects, it might be possible to extract design principles that could help to improve comfort and reduce fatigue in amputee gait.

Ankle push-off mechanics may be an important factor affecting amputee mobility and walking economy. Healthy non-amputees typically perform a large burst of positive ankle work during the end of the stance phase of gait, whereas amputees wearing conventional prosthetic feet exhibit much less ankle/foot push-off [68], and less center-of-mass (COM) push-off work [69]. Reduced push-off may lead to greater collisional energy losses of the leading limb after heelstrike [49], [36], [37], and may therefore require additional muscle work from other joints to compensate. Observed differences in joint kinetics between amputees and non-amputees [59] may reflect these and other compensatory actions and perhaps explain some of the increase in metabolic cost [60].

Push-off can be affected by elastic features of prosthetic feet. Dynamic elastic response feet [70], [71] incorporate passive spring-like components that return elastic energy during unloading of the foot and may thus contribute to push-off. Other devices attempt to augment push-off more directly using active control. For example, a prosthetic foot prototype, the controlled energy storage and return (CESR) foot, captures energy from the heelstrike collision and releases it elastically at push-off [43]. Another prototype, the MIT powered ankle/foot prosthesis, drives push-off both actively and elastically [72]. Such devices raise the question of what compliance should be specified in the design process.

Although various prosthetic foot designs have been evaluated against each other in comparison studies (e.g., [18], [19], [62], [73]), it is difficult to extract specific design principles regarding mechanical compliance (e.g., [74]). This is in part because even superficially similar feet (e.g. Seattle Lightfoot2 and FlexFoot, both categorized as dynamic elastic response feet) vary not only in compliance, but also in geometry, mass, alignment, and material type. These differences confound the ability to directly attribute differences in measured outcomes to specific design characteristics.

An alternative to comparing different brands or designs of prosthetic feet is to test one foot while systematically varying a single isolated parameter. Systematic manipulation of prosthetic parameters have only been performed in a few previous studies (e.g., [75], [76]), but could lead to greater insight regarding design principles for improving amputee gait, particularly those related to ankle push-off mechanics. This is especially the case for more complex prostheses that employ active mechanisms which supplement or control passive compliance. Of particular interest in these devices is the selection of the passive compliant components that contribute to push-off and govern much of the mechanical behavior of the ankle/foot. Systematic studies would allow for direct attribution of altered gait mechanics to a specific prosthetic foot parameter.

The potential efficacy of parametric studies is illustrated by previous experiments performed on non-amputee subjects wearing prosthetic simulator boots. Simulator boots immobilize the ankle and allow for attachment of a prosthesis beneath the foot [77]. These prosthetic simulators have been used to systematically study how convex foot bottom shape can affect energy economy [37] and how prosthetic alignment relates to roll-over shape [78]. Of course, even with ankle fixation, there remain numerous differences with amputees, including interface, musculature, and sensory information. Non-amputees walking on prosthesis simulators are not intended to be a model of amputee gait, but may nonetheless represent one way individuals walk in the absence of an articulating ankle joint and the associated musculature. Comparing amputees and non-amputees walking on the same prosthetic foot may therefore provide additional insight into how and why amputation affects gait.

The main purpose of this study was to systematically vary a single component within a prosthetic foot and experimentally measure the resulting effect on gait mechanics and metabolic energy expenditure. We studied amputee subjects to test how spring characteristics in the CESR foot affected their gait. We also studied non-amputees with prosthetic simulator boots to test for general effects that apply to walking without an active ankle, even without amputation. We measured push-off work and other variables as a function of systematically varied stiffness of an energy-recycling spring component within a prototype CESR prosthesis. The CESR spring strongly affects ankle push-off work and is therefore amenable to systematic variation. We then observed how this altered prosthetic ankle work affected gait. We predicted that the energy returned for push-off would be proportional to the amount stored in the spring during the foot's collision with ground, with that amount in turn dependent on spring stiffness. An improved understanding of specific prosthetic foot parameters and design principles could offer important insights that may enable improvement in amputee gait.

Methods

We studied level-ground walking mechanics and metabolic energy expenditure of unilateral, transtibial Amputees (N = 5) walking on the CESR foot (Figure 3.1), while varying its energy-recycling spring between three stiffness levels. We also performed a closely paralleled study on Non-Amputees wearing simulator boots (N = 11). We tested the effect of stiffness on work performed on the COM, joint work measures, and oxygen consumption. We also measured subjects walking on a representative conventional prosthesis, the Seattle LightFoot2 [79] (Seattle Systems, Poulsbo, Washington), as a qualitative control. To reduce possible confounds from prosthesis weight we added mass to the conventional foot to match the CESR foot (1.37 kg). Prior to the study, all subjects provided their informed consent according to Institutional Review Board procedures. Methods are outlined below and further details are reported in the supplementary material.

The CESR prototype is designed to recycle energy that is largely dissipated in walking collisions [43]. It stores energy elastically at the heel during loading of the foot in early stance (termed Collision), locks the spring throughout mid-stance with a one-way clutch, and releases the spring energy near terminal stance in the form of plantarflexion push-off work (Figure 3.1). Unlike conventional passive prosthetic feet that store and return elastic energy in the heel only during early stance, heel energy stored in the CESR can be controlled and released to help increase prosthetic limb push-off during terminal stance. The amount of CESR energy return was expected to be dependent on the energy-recycling spring properties varied in this study. The CESR foot is not intended to represent conventional prosthetic feet, but is a useful tool for studying the response of amputee gait to push-off related parametric variation in a prosthesis.

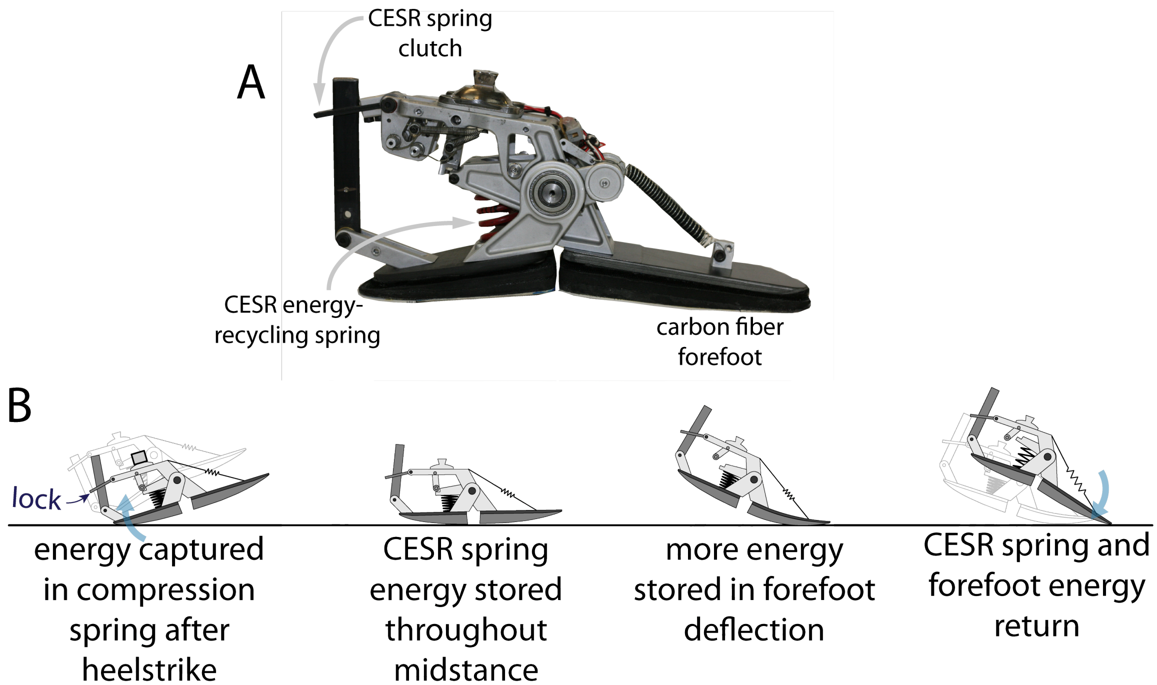


Figure 3.1. Controlled Energy Storage and Return (CESR) prosthesis prototype. The CESR foot stores energy in a compression spring after heelstrike, locks the spring energy into place with a clutch and then returns the energy during terminal stance in the form of plantarflexion push-off work. This energy recycling foot provides improved energy return as compared to conventional passive prostheses (Collins and Kuo, 2010).

All subjects were tested while walking on three different CESR energy-recycling springs. Spring specifications (Table 3.1) were selected in an attempt to systematically vary energy storage and return in the prosthetic foot. These springs were termed according to their stiffness as Hard (stiffest), Medium, and Soft-PC (softest, pre-compressed) springs. We found the Soft spring to have insufficient stiffness to prevent the heel component from reaching its maximum displacement limit during walking. We therefore found it helpful to pre-compress (PC) the Soft spring beneath the heel (see Table 3.1 for details), so that the Soft-PC spring exhibited increased spring force for a given heel displacement and thus higher energy storage capacity (Figure S6). Amputees walked on a slightly less stiff Hard spring compared to Non-Amputees due to comfort-related issues (see Supplementary Material). During testing, Amputee subjects walked at 1.14 m/s and Non-Amputees at 1.25 m/s to approximate typical self-selected walking speeds. We measured ground reaction forces, full-body kinematics, and oxygen consumption and carbon dioxide production. These tests were conducted on the three springs, applied in random order. Walking on a conventional prosthesis was also collected as a control condition. Non-Amputees wore the prostheses unilaterally on a prosthesis simulator boot, with a lift shoe on the contralateral foot, and also performed an additional Shod condition with normal street shoes on both feet. All Amputee prosthesis alignments were performed by the same experienced prosthetist. Typical clinical practice would involve re-aligning the foot for each new component in order to maximize the functional benefit to the user. However, in order to preserve the systematic nature of our study, springs were exchanged without disturbing prosthetic alignment, thus avoiding effects due to variation of an addition parameter. CESR alignments were performed with the Medium spring. Prior to the testing day, subjects were given a brief acclimation period, in which they walked on the treadmill and overground while wearing the CESR prototype in each of the three spring configurations.

We compared mechanical and metabolic results across spring conditions, and looked for similarities and differences in trends between Amputee and Non-Amputee subjects. We computed the following mechanical estimates: (1) rate of work performed on the body center-of-mass (COM) by the individual limbs [80], (2) ankle/foot power using an inter-segmental energy balance calculation between the ankle/foot and rest of body [81], [82], [83], and (3) knee and hip joint power using standard inverse dynamics (e.g., [84]). Ankle/foot power was estimated using the inter-segmental power method rather than traditional inverse dynamics because elastic portions of prosthetic feet are difficult to model as rigid bodies [83]. The inter-segmental approach treats the entire foot and ankle as a deformable body and measures the instantaneous forces and velocities of its contact with the ground and leg to estimate mechanical power due to both rotation and translation. We additionally used displacement sensors on the CESR foot to estimate elastic energy storage in the energy-recycling spring. We computed summary work measures by integrating the power estimates over phases of the gait cycle – Collision, Rebound, Preload, Push-off, Swing – as defined by fluctuating regions of positive and negative

COM work rate [5], [4]. We also estimated metabolic energy consumption using indirect calorimetry [85]. Within a subject group and across spring conditions, statistical analysis was performed using a repeat measures ANOVA with a Holm-Sidak correction. No statistical tests were performed comparing the conventional foot to any other condition. Mechanical and metabolic measures for Amputee vs. Non-Amputee subjects were not tested statistically either due to differences in experimental protocols, such as walking speed, average age, and leg length due to the simulator boot height (see complete methodological details in supplementary material). Inter-group similarities and differences may, however, still provide useful insight and therefore are compared qualitatively in terms of work magnitudes and trends across spring conditions.

Results

The results are reported separately for Amputees and Non-Amputees, followed by qualitative observations comparing both groups. In all figures and tables, the results represent the means and standard deviations computed across all subjects.

Amputees

Prosthetic foot mechanics varied with spring stiffness (Figure 3.2). Greater energy storage in the spring during Collision translated into more work performed in the subsequent ankle/foot Push-off. The amount of energy stored and returned by the prosthesis increased with decreasing stiffness. The Soft-PC energy-recycling spring stored the most energy (16.6 (1.0) J), followed by the Medium spring (8.1 (2.8) J), and then the Hard spring (6.5 (1.0) J; Table 3.1). Energy stored in the forefoot keel was of similar magnitude for all CESR conditions, about 8.5 J for Amputees, based on ankle/foot power estimates (see Figure 3.2, negative work during 40-80% of stance phase). Therefore, prosthetic energy return (from both the energy-recycling spring and the elastic forefoot) followed the same trend as energy storage, with Soft-PC performing the most prosthetic Push-off work (24.9 (1.5) J), Medium in the middle

(20.0 (1.3) J) and the Hard spring performing the least (18.8 (1.9) J). The Soft-PC spring stored and returned significantly more energy than the Medium and Hard springs (P < 0.008; Table 3.2). On average, the Medium spring stored and returned more energy than the Hard spring, but this difference did not achieve statistical significance in Amputees (P = 0.17).

Table 3.1. Energy Captured by CESR Spring (J)

	Amputees	Non-Amputees
Hard	-6.5 (1.0)[a]	-6.2 (1.3)[c,d]
Medium	-8.1 (2.8)[b]	-8.9 (1.2)[c,e]
Soft-PC	-16.6 (1.0)[a,b]	-13.3 (3.3)[d,e]
P-values (alpha=0.017)	[a]6.9e-4 [b]0 .0077	[c]2.9e-5 [d]4.1e-6 [e]6.9e-5

Table 3.2. Ankle Push-off (J). †*Italicized conditions are provided for reference, but were not tested statistically.*

	Prosthetic Limb		Intact Limb	
	Amputees	Non-Amputees	Amputees	Non-Amputees
Hard	18.8 (1.9)[h]	22.1 (2.7)[j,k]	18.3 (2.6)	16.5 (4.0)
Medium	20.0 (1.3)[i]	24.1 (2.4)[j,l]	17.8 (2.2)	17.6 (4.7)
Soft-PC	24.9 (1.5)[h,i]	26.7 (3.5)[k,l]	19.1 (3.0)	17.5 (5.2)
Conventional†	*12.2 (2.8)*	*10.2 (1.8)*	*20.4 (3.1)*	*15.1 (3.7)*
Shod†	-	-	-	*16.8 (4.4)*
P-values (alpha=0.017)	[h]0.0048 [i]0.0012	[j]1.4e-4 [k]1.1e-5 [l]0.0076	All > 0.18	All > 0.051

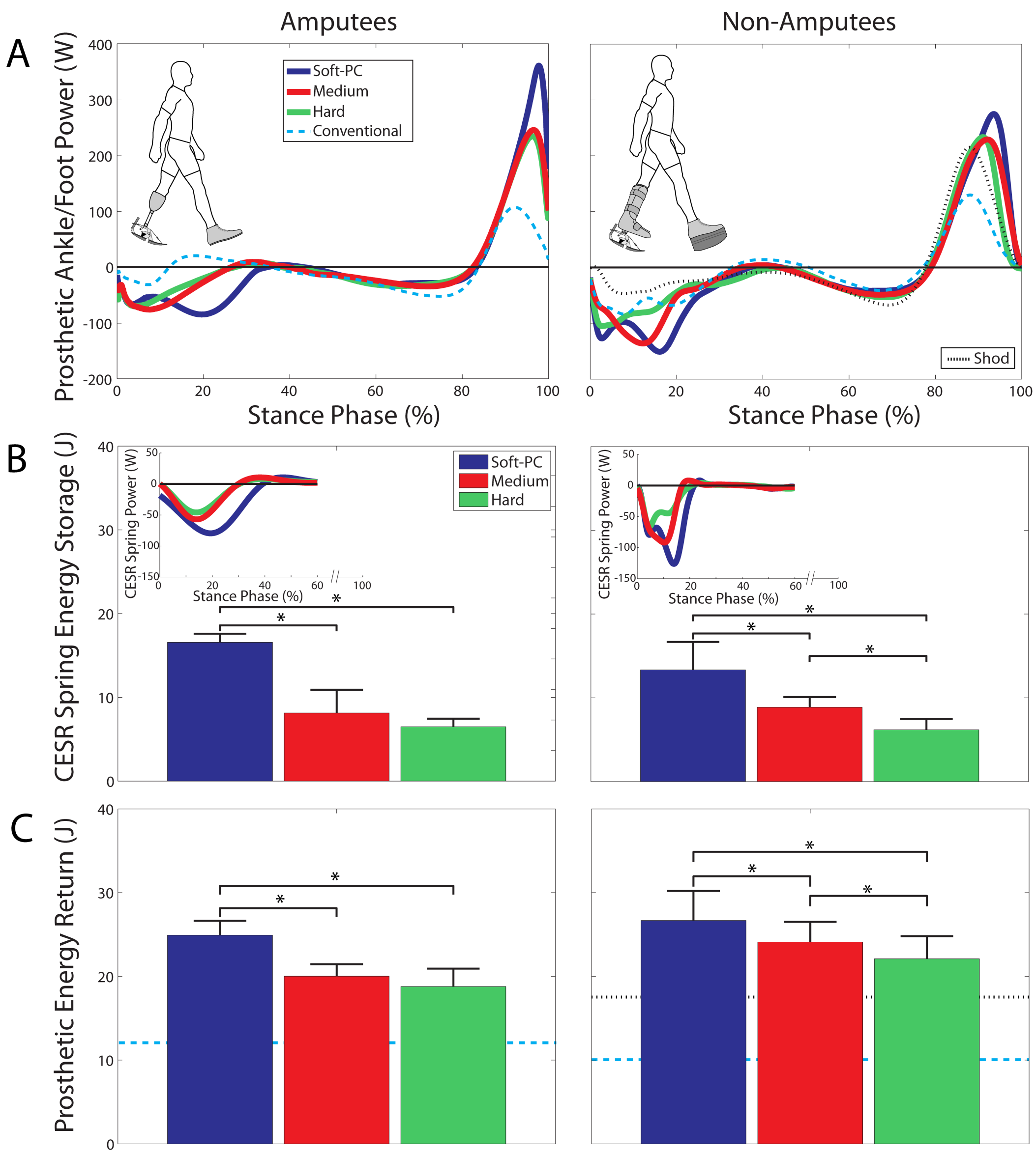


Figure 3.2. Prosthesis power, energy storage and return. Average (A) ankle/foot power, (B) CESR spring energy storage and (B) prosthetic energy return are shown for Amputees (left column) and Non-Amputees wearing simulator boots (right column). The (A) ankle/foot power estimates all energy flow into and out of the foot, and the inset in (B) CESR spring power, represents energy storage contributions of just the spring. Power is plotted across stance phase from heelstrike to toe-off. Results from a Conventional passive prosthesis (dashed line) and typical Shod Non-Amputee gait (dotted line) are provided for reference. Prosthetic ankle/foot mechanics varied with spring stiffness in both Amputees and Non-Amputees: the softest, pre-compressed (Soft-PC) spring stored and returned the most energy, while the stiffest (Hard) spring stored and returned the least. In all figures, the results represent the means and standard deviations computed across all subjects.

Spring stiffness also affected COM mechanics (Figure 3.3). Push-off work performed on the COM followed the same trend as ankle/foot work, tending to increase with decreasing spring stiffness. The Soft-PC spring resulted in the most COM Push-off work (17.7 (2.6) J), followed by the Medium spring (15.3 (1.8) J). Similar to ankle Push-off, COM Push-off was slightly lower with the Hard spring (14.9 (2.3) J) than Medium, but the difference again did not reach statistical significance. In Amputees, varying from the Hard to Soft-PC spring led to a 32% increase in ankle/foot Push-off work and a 19% increase in COM Push-off work.

Other mechanical measures showed little variation with spring stiffness. Firstly, spring choice had little effect on COM work rate of the intact limb (Figure 3.3). For instance, intact limb COM work measures were not found to differ across spring conditions during Collision ($P > 0.46$) or Push-off ($P > 0.09$) phases of gait (Table 3.3, Table 3.4). Secondly, prosthetic limb knee and hip kinetics (Figure 3.5, Figure 3.8, Figure 3.10) and intact limb ankle, knee and hip kinetics (Figure 3.7, Figure 3.9) exhibited little change with spring stiffness.

Metabolic energy expenditure was also affected by spring choice (Figure 3.4). On average, the Medium spring yielded the lowest net metabolic rate, 4% lower than the Soft-PC spring and 5% lower than the Hard spring. This result reached statistical significance for the Medium vs. Soft-PC comparison in Amputees ($P = 0.0025$). A strong trend was also observed in comparing Medium vs. Hard, but it did not reach statistical significance ($P = 0.076$).

Amputee gait kinetics while walking on the Conventional foot appeared qualitatively consistent with prior literature on below-knee amputee gait (e.g., [59]).

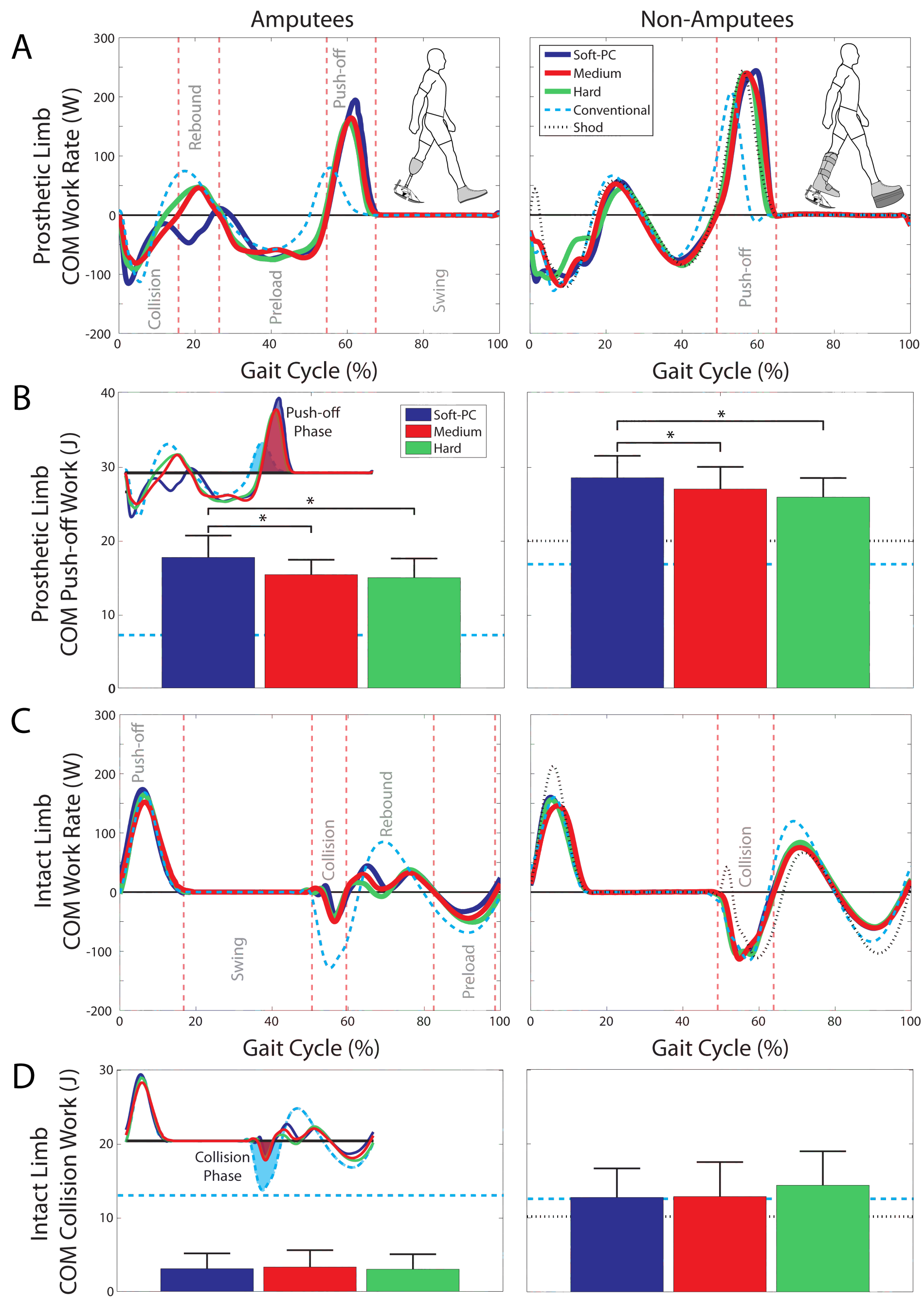


Figure 3.3. COM work rate, prosthetic limb Push-off and intact limb Collision work. Average COM work rate for (A) prosthetic and (C) intact limbs are plotted across a full gait cycle (beginning with prosthetic heelstrike) for Amputees (left column) and Non-Amputees (right column). Phases of gait – Collision, Rebound, Preload, Push-off, Swing – are defined for each limb based on alternating regions of positive and negative COM work, and approximate regions are shown for the Medium spring by vertical lines. The (B) prosthetic limb Push-off work and (D) intact limb Collision work were integrated from the shaded phases shown, which were defined independently for each subject and condition. Softer spring stiffness led to increased prosthetic limb COM Push-off, but stiffness did not have a significant effect on intact limb Collision work. Also, amputees appeared to perform less prosthetic limb COM Push-off work and less intact limb COM Collision work than Non-Amputees. Reference lines are shown for walking on a Conventional prosthesis (dashed line) and in street shoes (dotted line) for Non-Amputees.

Table 3.3. COM Collision Work (J). †*Italicized conditions are provided for reference, but were not tested statistically.*

	Prosthetic Limb		Intact Limb	
	Amputees	Non-Amputees	Amputees	Non-Amputees
Hard	18.8 (1.9)[h]	22.1 (2.7)[j,k]	18.3 (2.6)	16.5 (4.0)
Medium	20.0 (1.3)[i]	24.1 (2.4)[j,l]	17.8 (2.2)	17.6 (4.7)
Soft-PC	24.9 (1.5)[h,i]	26.7 (3.5)[k,l]	19.1 (3.0)	17.5 (5.2)
Conventional†	*12.2 (2.8)*	*10.2 (1.8)*	*20.4 (3.1)*	*15.1 (3.7)*
Shod†	-	-	-	*16.8 (4.4)*
P-values (alpha=0.017)	[h]0.0048 [i]0.0012	[j]1.4e-4 [k]1.1e-5 [l]0.0076	All > 0.18	All > 0.051

Table 3.4. COM Push-off Work (J). †*Italicized conditions are provided for reference, but were not tested statistically.*

	Prosthetic Limb		Intact Limb	
	Amputees	Non-Amputees	Amputees	Non-Amputees
Hard	14.9 (2.3)[m]	25.8 (2.6)[o]	16.9 (4.2)	18.2 (5.7)
Medium	15.3 (1.8)[m]	26.9 (3.0)[p]	17.4 (4.8)	18.9 (5.8)
Soft-PC	17.7 (2.6)[m,n]	28.4 (3.0)[o,p]	19.1 (5.2)	18.9 (7.1)
Conventional†	*7.3 (1.4)*	*17.1 (2.9)*	*15.3 (3.1)*	*16.9 (5.1)*
Shod†	*N/A*	*N/A*	*N/A*	*16.7 (4.9)*
P-values (alpha=0.017)	[m]0.0028 [n]0.014	[o]4.7e-4 [p]6.9e-4	All > 0.095	All > 0.2

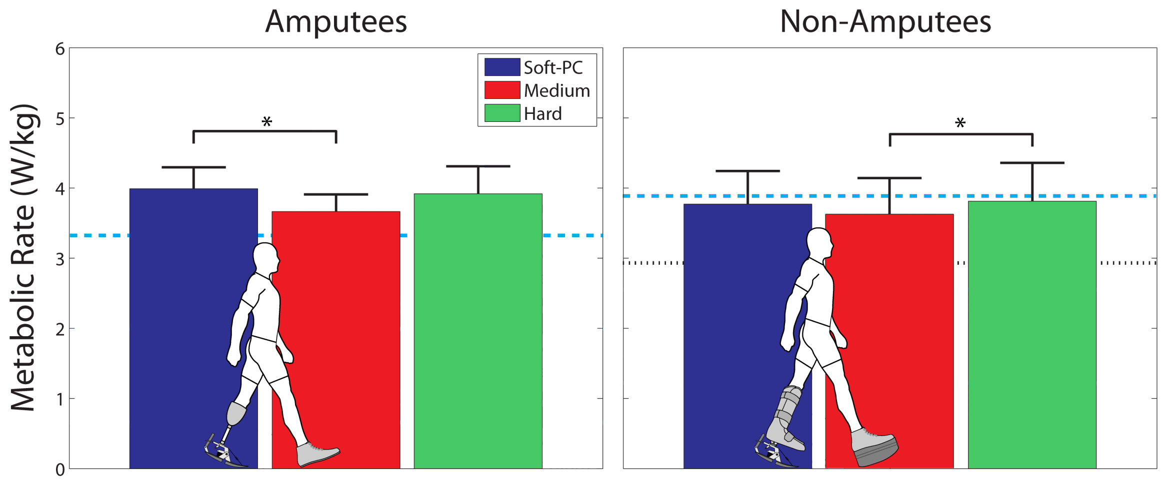


Figure 3.4. Average metabolic rate for Amputees (left) and Non-Amputees (right). Results for a Conventional passive prosthesis (dashed line) and shod Non-Amputee gait (dotted line) are shown for reference. We observed mixed metabolic results that were qualitatively similar in Amputee and Non-Amputee subjects across spring conditions. The Medium spring was significantly lower than Soft-PC spring in Amputees, and lower than the Hard spring in Non-Amputees subjects. Although metabolic energy expenditures were found to be of similar magnitudes in both Amputees and Non-Amputees, we consider this coincidental and attribute no special significance to the absolute comparison. There was no reason a priori to expect similar energetic magnitudes given differences between groups and experimental protocols. Qualitatively, Amputee subjects tended to expend more energy walking on the CESR foot than on the Conventional prosthesis, whereas Non-Amputees tended to expend less energy while walking on the CESR.

Non-Amputees

CESR energy storage and return increased with softening spring stiffness in Non-Amputees (Figure 3.2). The Soft-PC spring stored the most energy (13.3 (3.3) J), followed by the Medium (8.9 (1.2) J) and Hard (6.2 (1.3) J) springs (all differences statistically significant, $P < 7e\text{-}5$; Table 3.1). Prosthetic ankle Push-off work was highest with the Soft-PC spring (26.7 (3.5) J), followed by Medium (24.1 (2.4) J) and Hard (22.1 (2.7) J) springs (all differences statistically significant, $P < 0.008$; Table 3.2).

In Non-Amputees, prosthetic limb COM Push-off work also tended to increase with softening spring stiffness (Figure 3.3). COM Push-off work was 28.4 (3.0) J, 26.9 (3.0) J, and 25.8 (2.6) J for Soft-PC, Medium and Hard springs, respectively. Soft-PC vs. Medium ($P = 5e\text{-}4$) and Soft-PC vs. Hard ($P = 7e\text{-}4$) comparisons showed statistically significant differences. In Non-Amputees, varying spring properties yielded a 21% increase in ankle/foot Push-off work and a 10% increase in COM Push-off work.

Some other mechanical measures showed little variation across the gait cycle as a function of spring choice. These included intact limb sagittal plane joint kinetics and kinematics (Figure 3.7, Figure 3.8), intact limb COM work rate (Figure 3.3) and prosthetic limb sagittal plane knee and hip kinetics (Figure 3.5, Figure 3.9, Figure 3.10).

In Non-Amputees, the average metabolic rate was lowest while walking on the Medium spring by 7-8% as compared to the other spring conditions (Figure 3.4). This result was statistically significant for the Medium vs. Hard comparison (P = 0.0045), but did not quite reach significance for the Medium vs. Soft-PC comparison (P = 0.063).

Amputees vs. Non-Amputees

We observed similar trends in gait mechanics and metabolic cost as a function of spring stiffness for both Amputee and Non-Amputee subjects walking on the CESR foot. In both groups, varying spring stiffness led to similar systematic changes in foot function, joint kinetics and whole-body gait mechanics. Softer springs led to more energy storage and return in the CESR foot, which in turn led to more Push-off work performed by the prosthetic ankle and more Push-off work performed on the body COM. Intact limb ankle, knee and hip kinetics and prosthetic limb knee and hip kinetics showed little variation as a function of CESR spring stiffness. Additionally, metabolic energy expenditure was 4-8% lower for both groups walking with the Medium spring than the Soft-PC or Hard springs.

We observed similar mechanical function of the CESR foot in both groups, as indicated by similar magnitudes of energy storage and return performed by the prosthesis (Figure 3.2). Prosthetic energy storage for the Hard, Medium and Soft-PC springs varied from about 7-17 J in Amputees and from about 6-13 J in Non-Amputees. Prosthetic ankle Push-off work varied from about 19-25 J in Amputees and from about 22-27 J in Non-Amputees.

However, between groups there were also some substantial differences in COM and joint work magnitudes, specifically during the Push-off phase of gait. Prosthetic limb COM Push-off work appeared substantially lower (by about 10 J) in Amputees (15-18 J) than Non-Amputees (26-28 J) walking on the CESR foot (Figure 3.3). A similar difference was observed between the two groups with the Conventional prosthesis (7 vs. 17 J). Prosthetic limb knee and hip kinetics also differed between groups. Inverse dynamics estimates indicated that Amputees performed substantial prosthetic-side negative knee work (about 11 J) during Push-off (Figure 3.5). In contrast, Non-Amputees exhibited prosthetic-side positive knee work (about 1.5 J) during Push-off. Amputees performed about five times more prosthetic side hip work during Push-off (approximately 13 J vs. 2.5 J) compared to Non-Amputees (Figure 3.5). With the Conventional prosthesis, we made the same observation: prosthetic side knee and hip Push-off work magnitudes were substantially larger in Amputees than Non-Amputees.

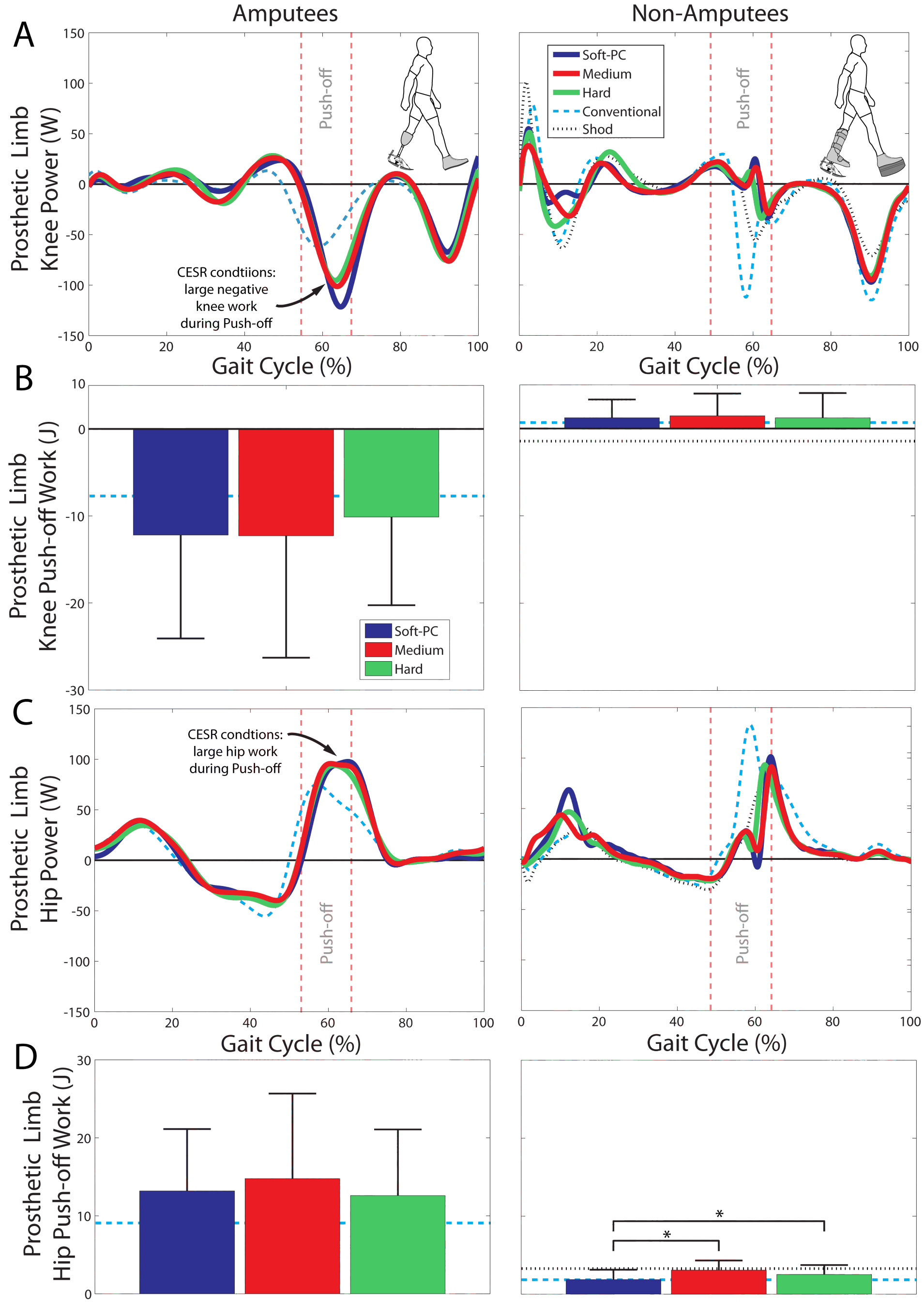


Figure 3.5. Prosthetic limb knee and hip powers, work during Push-off phase. Average prosthetic (A) knee and (C) hip are plotted across the full gait cycle (beginning with prosthetic heelstrike) for Amputees (left) and Non-Amputees (right). The prosthetic limb (B) knee work and (D) hip work during Push-off phase were defined separately for each subject and condition (see supplemental material, Figure 3.6). Vertical lines show approximate Push-off phase for Medium spring. Knee and hip power showed little variation within a group due to varying CESR spring stiffness. Amputees, however, seemed to exhibit large negative knee work (i.e., absorption) near terminal stance, which was not observed in Non-Amputees subjects. Amputees also appeared to perform more hip work during Push-off as compared to Non-Amputees. Reference lines are shown for walking on a Conventional prosthesis (dashed line) and in street shoes (dotted line) for Non-Amputees.

Discussion

We studied how systematic changes in energy-storing spring properties affected Amputees and Non-Amputees walking on a CESR prosthetic foot prototype. We found that several features of gait changed as a function of spring stiffness, including COM Push-off and Collision mechanics, and that Amputee and Non-Amputee groups both shared many similar trends in mechanical and metabolic measures. Softer springs stored and returned more energy, leading to higher prosthetic ankle and COM Push-off work. But there appeared to be biomechanical disadvantages to the softest spring (Soft-PC), as intermediate stiffness (Medium) yielded the best energetic economy. We also observed substantial differences in joint and COM work magnitudes between Amputees and Non-Amputees, most notably during prosthetic limb Push-off. These differences may provide insight into the effects of transtibial amputation on gait.

Regarding the energy-recycling spring, the general trend in both Amputees and Non-Amputees was that the lower the spring stiffness, the greater the energy captured. Softer springs therefore led to more energy return, yielding higher prosthetic ankle Push-off work, and ultimately to increased prosthetic limb COM Push-off work. Pre-compression was necessary for the Soft-PC spring due to the limited displacement range of the heel, but still allowed substantial energy storage (Figure 3.11), notably more than the Medium or Hard spring. Stiffer springs store and return less energy because they undergo less displacement for a given force. These trends do not, however, mean that the lowest stiffness is clinically optimal or yields the best overall function. In fact, the Medium spring yielded the best energetic walking economy by 4-8% in both groups, which may be relevant for Amputees who typically expend about 10-20% more energy than Non-Amputees [19], [20], [86] [87].

There are a number of possible disadvantages to the Soft-PC and Hard springs. First, storing too much energy in the Soft-PC spring may have led to increased metabolic cost due to abnormally high COM energy losses during Collision or due to interference with the natural Rebound of the body from early- to mid- stance (Figure 3.3). Meanwhile, the Hard spring may not have stored and returned enough energy. A second disadvantage may be associated with high initial loading rates at heelstrike, due to the stiffness of the Hard spring and pre-compression of the Soft-PC spring. High loading rates may have caused discomfort or instability, leading subjects to walk with greater muscle co-contraction or other compensations that were metabolically costly. Further study may be needed to understand the effects of high initial loading rates due to pre-compression. However, the alternative was to not pre-compress the Soft-PC spring, which we found to cause an even more uncomfortable collision when the spring fully compressed and the heel component bottomed out during gait. An alternative to pre-compression would be to increase the foot's internal range of motion to allow the soft spring to displace without bottoming out, but altering the range would affect the rolling shape of the foot and too large a range would lead to scuffing of the heel on the ground during leg swing. For the three springs tested in this study, rolling shapes were found to be relatively similar from mid- to late-stance, but did exhibit some qualitative differences during heelstrike collisions (Figure 3.12). Practical trade-offs between range of motion, energy storage and rolling shape are difficult to discern from gait analysis of three conditions, but this limitation does not change the main result: energy storage and return within the CESR foot increased with softening spring stiffness, but there were disadvantages to excessively low or high stiffness, such that some intermediate stiffness was optimal.

Although the CESR prototype was found to successfully return captured energy during Push-off, the work may not have been performed optimally. Dynamic walking models predict that Push-off work of appropriate timing and magnitude can reduce Collision work of the contralateral limb [49], [36] and thus reduce overall work on the COM. Such an effect was previously observed when comparing CESR with a conventional foot in Non-Amputees wearing a prosthetic simulator [43]. The negative trend between Push-off and Collision work magnitudes was also observed in other studies on Amputees wearing conventional prostheses [69] and on Non-Amputees wearing shoes with different rolling shapes [88]. The trend was not, however, observed in this study when comparing across the three CESR springs, which yielded a smaller range of Push-off magnitudes than the previous study [43]. Additionally, there are other studies that did not find that greater COM Push-off work reduced Collision work, such as when comparing typical non-amputee gait to gait with artificially fixed ankles [89] or when comparing impaired and unimpaired limbs [88]. In these cases, other factors may have played a more important role. For example, concomitant changes to limb mechanics may have dominated over the broad effect of Push-off work on Collision. Confounding factors such as timing of Push-off, fluctuations in COM velocity, the peak power, or altered patterns of work production may have adversely affected the overall COM work magnitudes. In particular, we suspect that Push-off should begin earlier in the gait cycle relative to Collision, which is theoretically advantageous [49], [36] and for unilateral amputees, perhaps with magnitude more comparable to intact limb Push-off.

Although the CESR foot performed similar amounts of energy storage and return in both groups, some COM and joint work measures varied substantially. Due to the various methodological differences, we had no a priori expectation that absolute COM and joint work magnitudes should be similar in Amputees and Non-Amputees. However, since the CESR foot stored and returned approximately the same amount of energy in both groups, we were able observe how this similar magnitude of prosthesis work affected joint and COM mechanics. Notably, we found dramatic differences in prosthetic limb knee, hip and COM work during Push-off. Despite a similar amount of CESR Push-off work, Amputees produced much less prosthetic limb COM Push-off work than Non-Amputees, and much more knee and hip work during Push-off (Figure 3.5). In Non-Amputees, the magnitudes of prosthetic side knee and hip work were small and slightly positive during Push-off. But in Amputees, the substantial negative knee work appeared to partially absorb the CESR Push-off work, reducing energy transfer from the prosthesis to the body COM and negating some of the benefits of prosthetic ankle Push-off. Simultaneous positive power at the hip might then be a compensation for knee absorption in Amputees and may lead to an increase in metabolic cost. Since inverse dynamics estimates cannot differentiate biarticular energy transfer from monarticular muscle work, we cannot directly estimate a metabolic penalty associated with this increased biological joint work. Overall, substantial kinetic differences between groups suggest Amputees and Non-Amputees used prosthetic Push-off differently.

In addition to Push-off, we also observed qualitative differences in the Collision phases of the intact and prosthetic limbs. Again, these differences are noteworthy given similar CESR foot function and energy storage in both groups (Figure 3.2; Table 3.1). Firstly, COM Collision work done by both limbs appeared substantially lower in Amputees than Non-Amputees (Figure 3.3; Table 3.3). Secondly, Amputees performed little to no negative knee work during Collision phase, while Non-Amputees performed significant negative knee work (Figure 3.5, Figure 3.8, Figure 3.10). Thirdly, the rate of CESR spring loading after heelstrike appeared higher in Non-Amputees than Amputees (Figure 3.2). Collectively, these observations also suggest some differences in gait strategies used by Amputees vs. Non-Amputees. In general, Amputees appeared to adapt Collision strategies that resulted in smaller COM work magnitudes and less knee work. While these differences may be indicative of socket interface or comfort-related issues, it is interesting that they were also observed in the intact limb.

Although we observed similar trends in metabolic energy expenditure between the two groups, we do not quantitatively compare the absolute magnitudes. As with mechanical work, there was no reason to expect similar energetic magnitudes across groups given the differences between experimental protocols. Consequently, metabolic comparisons were only made within each group. Within group comparisons did suggest that metabolic rates for Non-Amputees were lower while walking on this CESR prototype than on the Conventional foot [43], but this relative trend was not observed in Amputees. The CESR principle may still benefit Amputee gait, although it may require slightly different design parameters or further user acclimation.

We propose several possible explanations for the observed mechanical and metabolic differences between Amputees and Non-Amputees. The CESR foot may have been returning too much energy for the walking speed tested, such that Amputees used knee absorption to regulate the excess power. This CESR prototype was originally designed for walking speeds faster than those studied in these groups (e.g., 1.5 m/s), when higher ankle/foot Push-off is more desirable. Since Amputees were walking more slowly, they may have adapted by over-compensating with negative knee work, effectively negating mechanical Push-off benefits of the CESR foot. Another factor may have been that the CESR foot length was too long for the Amputees. The prosthetic foot length relative to leg length was 11% longer in Amputees than Non-Amputees due to the added height of the simulator boots. Longer feet tend to cause later Push-off [37], which was observed in comparing Amputees vs. Non-Amputees (Figure 3.3), and may have led to associated changes in knee and hip kinetics. Another possibility may be that slower CESR spring loading (Figure 3.2) in Amputees led to a sub-optimal foot shape during rolling of the foot during single support. Rolling foot shape has been shown to affect metabolic cost substantially [37], [88]. Or Amputees may adopt gait mechanics that are more heavily influenced by factors other than mechanical foot function. For example, the interface between the socket and residual limb may affect how comfortable it is for Amputees to transmit forces between the prosthesis and the rest of the body, which may partially explain differences in knee and hip kinetics. Or the length of the residual limb may limit the ability to control energy storage and return in the foot, since shorter residual limbs have smaller moment arms to generate torques and less surface area over which to distribute forces. Finally, a possible factor in this study could be related to insufficient acclimation time to the new CESR prosthesis. Despite similar adaptation times for Amputees and Non-Amputees walking on the CESR foot (about 30 min), Amputees might have developed habits based on their prescribed passive prostheses that made it more difficult for them to adapt to a prosthesis with more energy storage and return. Or perhaps years of experience walking on Conventional feet biased Amputees, who needed more time to adapt to the CESR foot. Such a bias would not be expected in naïve Non-Amputees, which may partially explain improved metabolic cost in this group.

We acknowledge several methodological limitations in directly comparing Amputees to Non-Amputees. We tested Amputees at slightly slower speed (1.14 vs. 1.25 m/s) and with a less stiff Hard spring (262 vs. 324 N/mm, Table 3.5). Experimental protocols were designed to be as identical as practical, but there were necessary differences because some Amputee subjects had a slower comfortable walking speed on the treadmill and reported discomfort in the residual limb if the energy-recycling spring was too stiff. Additionally, Amputees were tested at self-selected step frequency and Non-Amputees at a fixed metronome frequency, although resulting stride times were found to be nearly identical (Table 3.6). We studied gait mechanics for Amputees during overground walking, but for Non-Amputees during treadmill walking. Although kinematics and kinetics have been observed to be quite similar in overground vs. treadmill walking in healthy individuals [90], [91], walking on a prosthesis may be more affected, perhaps due to balance-related issues that might be exacerbated on a treadmill. Given limited Amputee subject availability and study time before fatigue onset, there were also practical limitations in how many different spring conditions could be tested. Three springs were selected to vary energy storage and return based on pilot studies performed on Non-Amputees.

Although prosthetic technologies are far from mature, it appears that specific mechanical parameters have deterministic effects on gait biomechanics. To understand these effects, it is necessary to conduct systematic studies that control or isolate parameters of interest. We found that systematically varying CESR spring stiffness led to altered prosthetic ankle mechanics, which in turn affected COM mechanics and metabolic energy expenditure. When measuring the same type of prosthesis with varying spring parameters, we found very similar trends in many mechanical and metabolic measures for Amputees and Non-Amputees. We also found it helpful to perform comparisons with Non-Amputees wearing prosthetic simulators, since studying this group eliminates variability due to residual limb length, socket interface and secondary disabilities. Inter-group comparisons may provide insight regarding the disadvantages associated with the loss of an active ankle. Non-Amputees exhibited substantial differences in work magnitudes for some COM and joint measures, suggesting that additional considerations may be important for optimizing prosthetic design parameters for Amputees. Ultimately, more systematic studies of prosthetic design parameters could help inform fundamental design principles for feet.

Supplementary Material

We used the Controlled Energy Storage and Return (CESR) prosthetic foot as a tool to explore the effects of altering ankle/foot mechanics, specifically Push-off work, on whole-body walking mechanics and metabolic cost. We were able to systematically vary energy storage and return in the foot by testing subjects walking on the CESR prosthesis with different energy-recycling springs (Table 3.5). Function of the CESR foot and experimental protocols for Amputee and Non-Amputee subjects are detailed below.

CESR Prosthetic Foot

The CESR foot is an energy recycling prosthesis capable of capturing energy elastically after heelstrike, storing the energy through mid-stance and returning the energy at terminal stance in the form of plantarflexion push-off work (Figure 3.1). The energy recycling concept is based on dynamic walking principles [49], [36], [80] and human gait experiments [43] that suggest Push-off is the most economical time in the gait cycle to perform positive work because it minimizes mechanical Collision losses. Compared to conventional passive prosthetic feet, increased prosthetic ankle/foot Push-off work of the CESR foot more closely emulates the function of the natural ankle during walking. Energy recycling is accomplished by loading a compression die spring under the heel during early stance, locking the spring energy into place with a one-way clutch, then releasing a second latch that returns the spring energy to the forefoot for Push-off during the end of stance (Figure 3.1). The mechanism then resets during Swing phase. In addition to energy recycled from the CESR spring compression, Pre-load energy is also stored and returned in the forefoot, a cantilevered carbon fiber keel similar to those used in conventional dynamic elastic response prostheses. The prototype used in these studies was designed with separate toe and heel sections, each of which can articulate about a main mediolateral axis of the foot. A low-power microcontroller (worn on a small backpack with battery) used information from on-board angular displacement sensors at the heel and toe to determine activation timing of small motors that controlled energy storage and return. The mechanism and control design of the CESR foot are fully detailed in Collins and Kuo (2010) [43].

Amputee Protocol

Amputee subjects were recruited through the Seattle Veteran Affairs Hospital based on the following inclusion criteria: unilateral, transtibial amputation, age 18-80, weighing 70-100 kg, prosthesis user for minimum of 2 years, active and independent ambulator (i.e., no upper-limb aids), no history of injurious falls within the previous 6 months and free from neurological deficits or other musculoskeletal disorders. Seven subjects were initially recruited, but two were excluded because they were unable to complete the full protocol. This study was approved by the Veterans Affairs' Institutional Review Board and all subjects gave informed consent prior to participation.

Amputee subjects (N=5, age 50 ± 13 years, 76.7 ± 3.3 kg, leg length 0.97 ± 0.02 m) were tested at the Center of Excellence for Limb Loss Prevention and Prosthetic Engineering in Seattle, WA. Each subject performed acclimation and testing protocols on separate days, with the first day serving solely as an acclimation and training period. Treadmill acclimation was accomplished by 5 minutes of walking (1.14 m/s) on the subject's own prescribed foot. Next, about 5 minutes of overground and 5 minutes of treadmill walking were performed on the CESR foot for each of the three spring conditions and on the Conventional foot. The Conventional foot was worn and aligned inside a shoe, but since it was weight-matched to the CESR foot it was not intended to be a precise clinical baseline, only a qualitative control. All prosthetic alignment was performed by the same experienced prosthetist. In addition to the prosthesis, during all conditions subjects wore a small backpack (0.80 kg) containing a battery and microcontroller, which was connected to the prosthesis via ribbon cable and to the analog data acquisition via coaxial cable. To ensure full recovery after acclimation, at least 1 day separated training and collection sessions for all subjects. To ensure full retention of training, this recovery period was typically also fewer than 3 days, though in some cases was as high as 7 days due to Amputee subject availability.

During the data collection session, we recorded kinematic, kinetic and metabolic data for subjects walking on the CESR foot at 1.14 m/s at self-selected step frequency. Testing of the three spring conditions– Hard, Medium and Soft-PC – was randomized. For each condition we collected metabolic data, immediately followed by kinematic and kinetic measurements. Initially, a resting metabolic baseline was collected for 6 minutes during quiet standing. Next, treadmill metabolic testing lasted 10 minutes, the last 3 minutes of which were analyzed as steady state. Metabolic energy expenditure was approximated from oxygen and carbon dioxide exchange rates (i.e., indirect calorimetry; [85]) collected by the Oxycon Mobile wireless ergospirometry system (Viasys Healthcare, Yorba Linda, CA). In addition, a manual harness system was used during treadmill walking for safety reasons; however, the cable remained slack as not to interfere with gait. No adverse events that required engagement of the harness occurred for any subjects.

Kinematic and ground reaction force data were measured while subjects walked overground across force plates embedded along a 10 m walkway. Gait kinematics were collected with a 12-camera Vicon motion capture system (Oxford Metrics, Oxford, England) sampled at 120 Hz. We placed thirty-five 14 mm reflective markers on each subject at locations consistent with Vicon Plug-in-Gait full-body model (Oxford Metrics; Oxford, England). We placed an additional 4 markers on the CESR foot (heel, toe, medial/lateral articulating axis) to track motion of the foot segments. Ground reaction forces were collected with 2 Bertec force plates (Columbus, Ohio) and 2 AMTI force plates (Arlington, VA) sampled at 1200 Hz. A minimum of 6 successful overground trials were collected for each condition. We defined successful trials by the following two criteria: (1) walking speed was within range 1.14 ± 0.11 m/s as measured by photo gates and (2) at least two sequential foot strikes occurred on separate force plates.

Gait data were filtered (Woltring with mean-square-error value of 20), then standard 3D inverse dynamics were calculated using Vicon Plug-In-Gait dynamic model (Oxford Metrics; Oxford, England). A stride was defined from heelstrike to subsequent ipsilateral heelstrike based on gait events determined from Vicon's event detection algorithm. Inter-segmental power and COM work rate were calculated from forces and kinematics filtered at 25 Hz (Butterworth, 3rd order).

Non-Amputee Protocol

The Non-Amputee study was performed using similar methods to the Amputee study, but with some methodological differences due to different subject groups and equipment available at each testing site. The notable methodological differences are that in the Non-Amputee study:

(1) kinematics, kinetics and metabolic cost were recorded simultaneously because of availability of an instrumented force treadmill
(2) an additional control condition, shod walking in street shoes, was performed
(3) prosthetic simulator boot and lift shoe were worn by Non-Amputee subjects
(4) subjects walked at slightly faster speed (1.25 vs. 1.14 m/s)
(5) the Medium and Soft-PC springs were identical, but the Hard spring was stiffer (324 vs. 262 N/mm due to comfort-related issues in the Amputee group)
(6) the conventional prosthetic foot (control condition) was worn without a shoe
(7) the subject group was considerably younger (24 ± 3 vs. 50 ± 13 years)
(8) subjects walked at fixed metronome frequency in all CESR spring condition trials

Non-Amputee subjects were recruited at the University of Michigan based on the following inclusion criteria: age 18-50, weighing 70-100 kg and with no known gait or balance impairments. The study was approved by University of Michigan Institutional Review Board and all subjects gave informed consent prior to participation.

All Non-Amputee subjects (N = 11, age 24 ± 3 years, mass 79.6 ± 7.1 kg nominally, 82.6 ± 7.1 kg while wearing prosthesis and accessories, leg length 0.97 ± 0.04 m nominally, leg length 1.08 +/- 0.04 m while wearing simulator boot) were tested in the Human Neuromechanics Laboratory at the University of Michigan. Subjects underwent training that was similar to the Amputee protocol. Training occurred 2 days prior to data collection. Subjects wore a prosthetic simulator boot (1.30 kg) with CESR foot (1.37 kg) attached unilaterally on the right leg and wore a lift shoe (1.42 kg) on the left leg to account for additional height (0.13 m) of the prosthesis. The simulator boots were modified AirCast© boots that immobilize the ankle and provide prosthetic attachment beneath the foot [43], [37]. Subjects wore a microcontroller and battery backpack identical to that described in the Amputee protocol. Spring conditions were changed without removing the prosthesis from the simulator boot in order to achieve consistent prosthetic alignment across conditions.

Kinematic, kinetic and metabolic data were measured simultaneously while subjects walked at 1.25 m/s on a custom-built instrumented treadmill. Conditions included baseline resting during quiet standing (6 minutes), walking in normal street shoes (10 minutes) and energy recycling spring condition trials (10 minutes). All walking conditions were randomized. Oxygen consumption and carbon dioxide production rates were measured using an open-circuit respirometry system (Physio-Dyne Instrument, Quogue, NY). Kinematic data were recorded at 120 Hz using an 8-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA). Markers were placed bilaterally according to a modified Helen Hayes marker set. Markers were placed on the simulator boot in locations approximating the anatomical bony landmarks and markers were rigidly attached to the prosthesis as described in the Amputee protocol. Ground reaction force data were recorded at 1200 Hz using a custom-built instrumented split-belt treadmill [92]. Kinematic and force data were filtered at 25 Hz. Standard 3D inverse dynamics were computed using custom software.

Data Analysis

We estimated mechanical power from measured kinematics and forces, and estimated metabolic energy consumption from oxygen consumption and carbon dioxide production. We made comparisons between spring conditions within a single group. Trends from the Amputee vs. Non-Amputee studies were qualitatively compared given the confounding methodological differences discussed above. We used the following mechanic and metabolic metrics:

1. *Center-of-mass work rate* was calculated independently for each limb. The work rate was computed from the 3D dot product of each limb's ground reaction force with COM velocity [80]. COM velocity was calculated from integration of ground reaction forces, assuming steady-state, periodic strides. This assumption was particularly strong for trials using an instrumented treadmill.

2. *Inter-segmental power* between foot and shank, i.e. power attributed to the prosthesis, was calculated based on summing translational and rotational work rates at the distal shank [81–83]. Translational power was calculated as the dot product of ankle joint force and translation velocity of the ankle marker (lateral malleolus). Rotational power was calculated as the dot product of the ankle moment (about the axis at the lateral malleolus) and angular velocity of the shank. This inter-segmental power method was used to estimate ankle/foot power in place of inverse dynamics because the former makes no rigid-body assumptions about the prosthetic feet, whereas

inverse dynamics is poorly suited to capture the unmodeled degrees of freedom in prosthetic feet (e.g., heel and toe keel deformations). In particular, standard inverse dynamic models fail to capture work performed about the independently articulating heel in the CESR prosthesis.

3. *CESR energy-recycling spring power (or rotational heel power)* was a supplementary metric used to approximate energy stored in the compression spring. This heel power was calculated from multiplying ankle moments with heel angular velocities, the latter derived from the on-board angular displacement sensor (a non-contact potentiometer).

4. *Joint power* about the ankle, knee and hip were computed from standard inverse dynamics as described previously in the Amputee and Non-Amputee protocols. Anthropometric data were modified to reflect changes in foot mass due to the CESR foot, simulator boot, and lift shoe, as appropriate for each subject and condition. Prosthetic ankle joint power from inverse dynamics was not reported (see #2 above on inter-segmental power)

5. *Metabolic power* was estimated from oxygen consumption and carbon dioxide production during steady-state gait, which we chose as the last 3 minutes of each walking trial. Power calculations were based on standard indirect calorimetry equations relating substrate metabolism to energy production [85]. Net metabolic results are presented, meaning that the resting metabolic rate was subtracted from metabolic power for each walking condition. As verification, we confirmed that respiratory exchange ratios were less than one for all subjects and conditions, indicating that energy was supplied primarily by aerobic metabolic pathways.

For each mechanical estimate, work summary measures were calculated by integrating under the power curves during specific phases of the gait cycle (Figure

3.6). These phases or integration regions – Collision, Rebound, Pre-load, Push-off and Swing – were based on alternating regions of positive and negative COM work (Figure 3.3). At the level of the prosthesis, energy-recycling spring energy storage was integrated over the negative power region in early stance, which in some cases was longer than the COM-defined Collision phase (Figure 3.2). Energy return was defined as the integral over the positive power region preceding toe-off. All statistical comparisons were performed using a repeat measures ANOVA with Holm-Sidak correction. Nominally, P-values less than 0.05 were considered statistically significant, but with correction alpha = 0.017.

All mechanical measures were non-dimensionalized, averaged across subjects, and then re-dimensionalized for reporting purposes. Normalization constants were based on units of mass M, leg length L, and gravitational acceleration g. Average power and work normalization constants for Amputees were $Mg^{3/2}L^{1/2}$ = 2319 W and MgL = 729 J, and for Non-Amputee subjects were 2632 W and 874 J.

Table 3.5. CESR energy-recycling spring stiffness (N/mm). * Soft-PC spring was also pre-compressed by 25 mm.

	Amputees	Non-Amputees
Hard	262	324
Medium	157	157
Soft-PC*	42	42

Table 3.6. Stride Time (sec). †*Italicized conditions are provided for reference, but were not tested statistically.*

	Amputees	Non-Amputees
Hard	1.28 (0.05)	1.30 (0.03)
Medium	1.28 (0.05)	1.30 (0.04)
Soft-PC	1.31 (0.02)	1.30 (0.04)
Conventional†	*1.26 (0.06)*	*1.19 (0.06)*
Shod†	*N/A*	*1.17 (0.06)*
P-values (alpha=0.017)	All > 0.17	All > 0.028

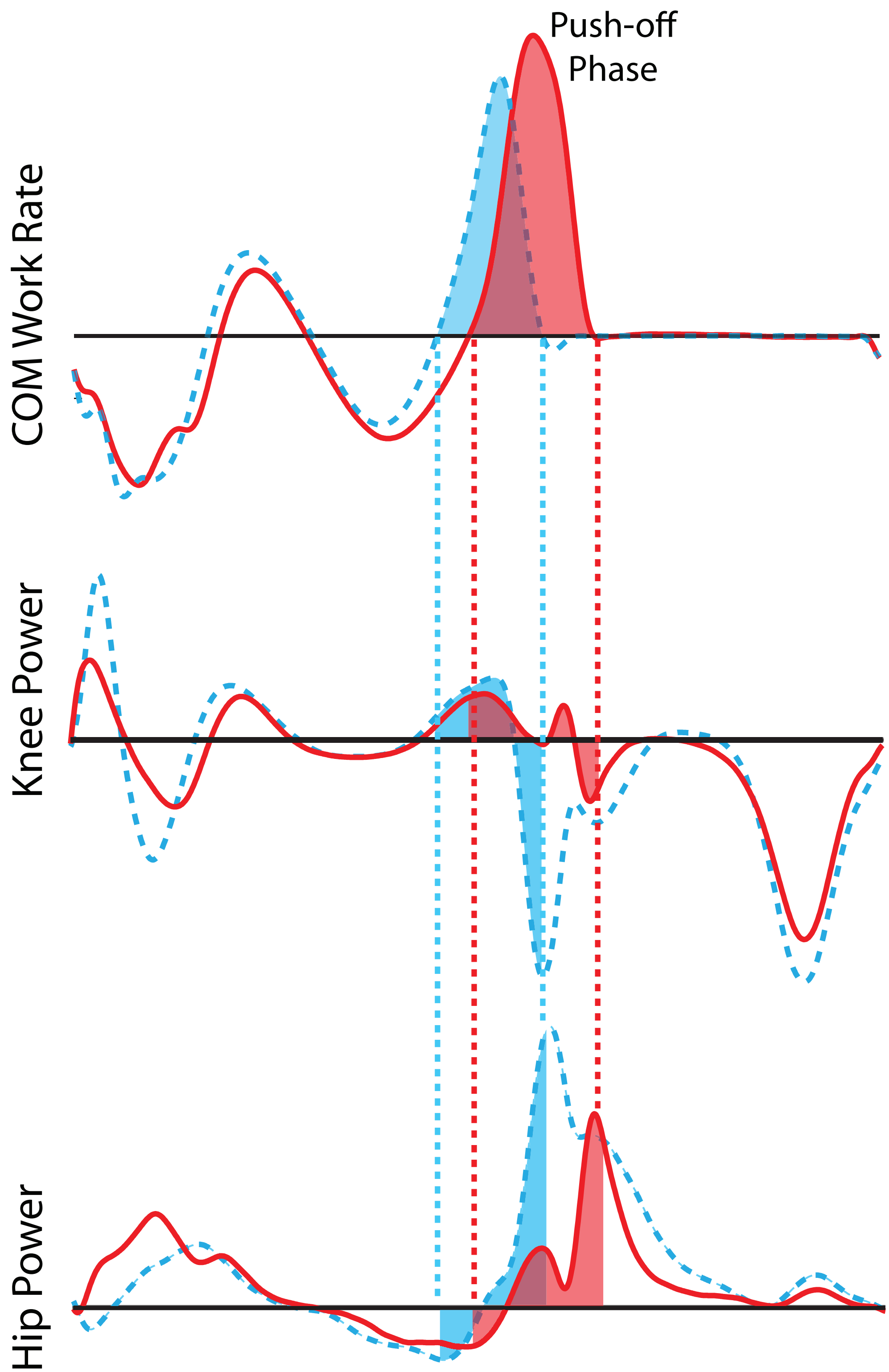


Figure 3.6. COM work rate and joint power over a stride: calculating work during Push-off. Phases of gait were defined independently for each subject and condition, based on alternating regions of positive and negative COM work rate. Knee and hip powers were integrated over shaded regions to compute joint work during Push-off phase. Plots are shown over a full stride from prosthetic limb heelstrike to heelstrike.

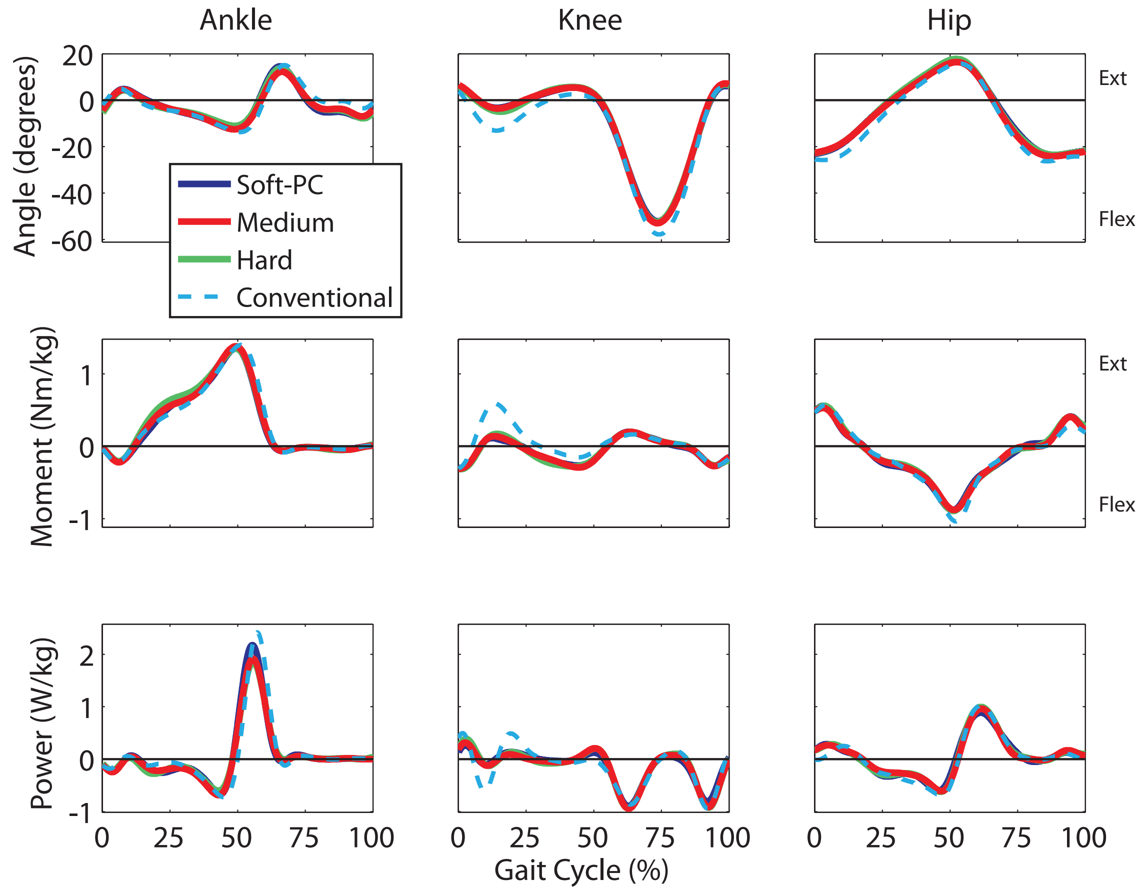


Figure 3.7. Average sagittal plane joint angles, moments and powers for ***Amputees' intact limb***. Plots are from intact limb heelstrike to heelstrike.

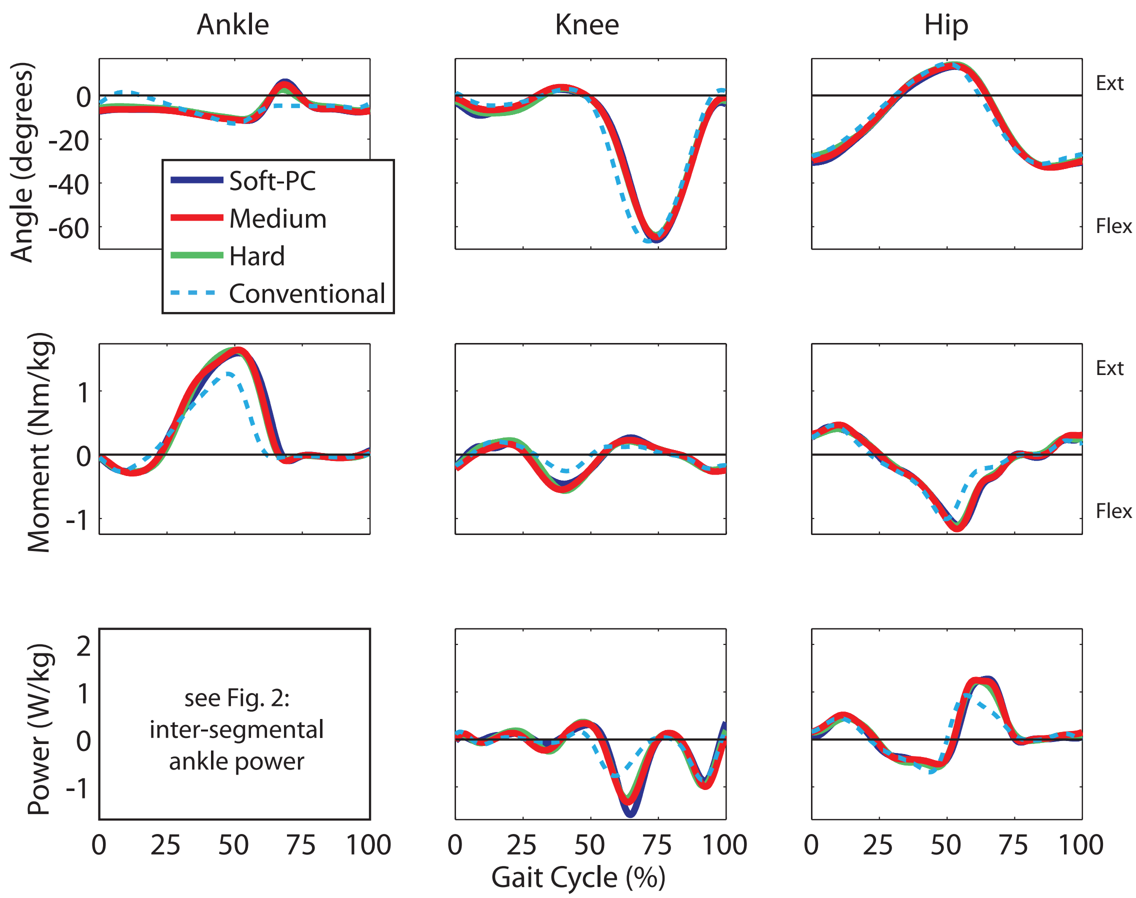


Figure 3.8. Average sagittal plane joint angles, moments and powers for ***Amputees' prosthetic limb***. Plots are from prosthetic limb heelstrike to heelstrike.

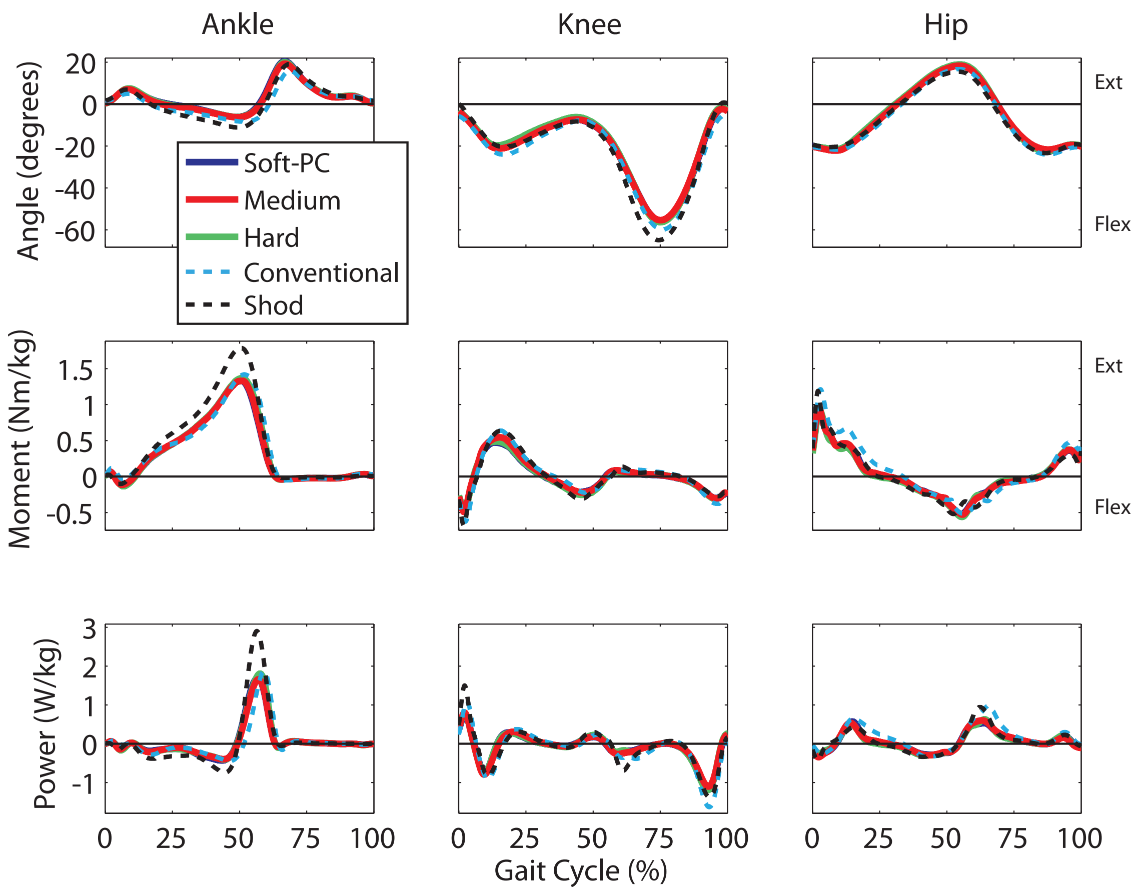


Figure 3.9. Average sagittal plane joint angles, moments and powers for ***Non-Amputees' intact limb***. Plots are from intact limb heelstrike to heelstrike.

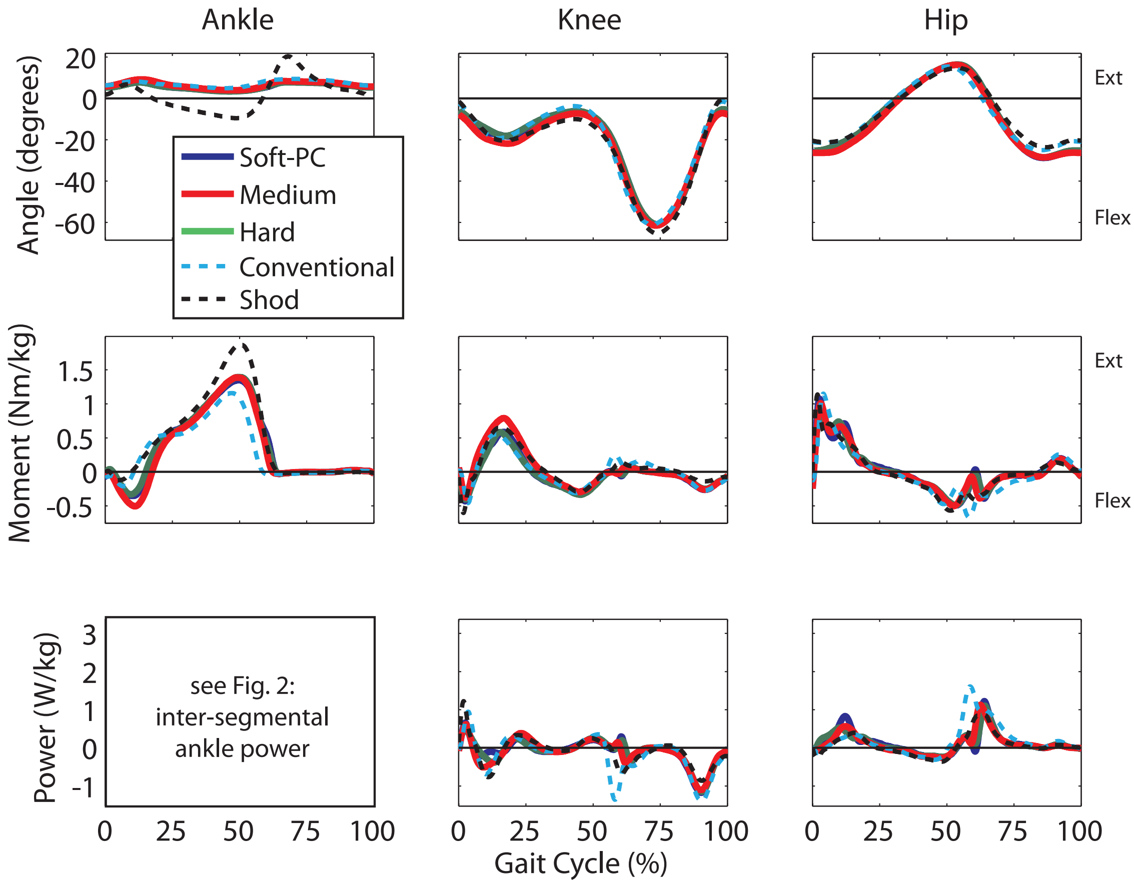


Figure 3.10. Average sagittal plane joint angles, moments and powers for ***Non-Amputees' prosthetic limb***. Plots are from prosthetic limb heelstrike to heelstrike.

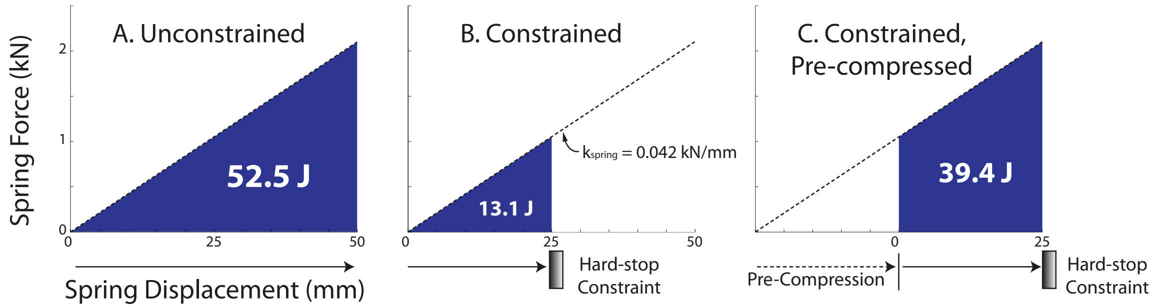


Figure 3.11. Effect of pre-compression on energy storage capacity of a spring. Example force-displacement curves are shown for a single spring under 3 different conditions: (A) unconstrained, (B) constrained and (C) pre-compressed and constrained. For (A) unconstrained springs (i.e., unlimited range over which they can compress), energy storage capacity is inversely proportional to spring stiffness (k_{spring}), so the softer the spring the more energy stored for a given load. An unconstrained spring of stiffness 0.042 kN/mm will deflect 50 mm when loaded with 2.1 kN, storing 52.5 J of energy. Practical limitations, however, such as the finite range of heel rotation in the CESR, often introduce a (B) hard-stop constraint. Assuming a maximum spring deflection constraint of 25 mm, the same spring would only be capable of storing 13.1 J (25% of the unconstrained spring capacity). By performing a (C) pre-compression on the spring, while keeping the maximum deflection constraint, the spring can store 39.4 J (75%). In summary, pre-compression allows a constrained spring to store more energy than it would otherwise be able to store.

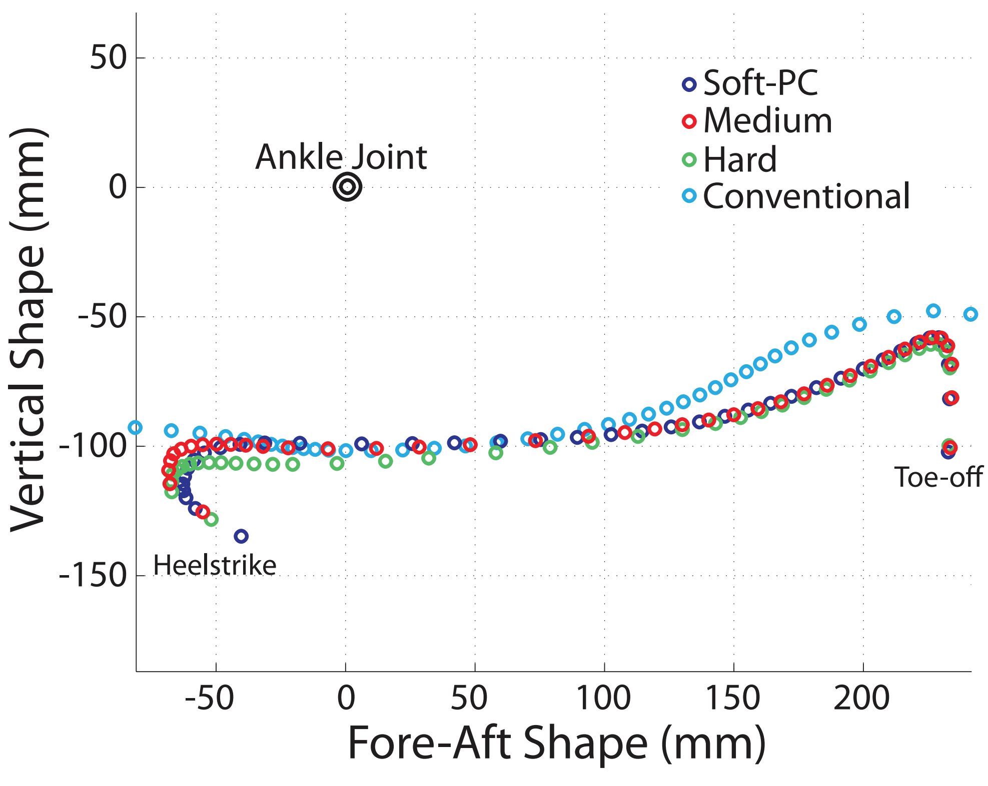


Figure 3.12. Rolling shape of the prosthesis. Average rolling foot shape (center of pressure plotted with respect to the ankle in the shank reference frame [78]) is presented for Amputee subjects. All CESR conditions appear similar during mid- to late-stance, but as expected some differences are observed during heel loading in early stance. We had no specific hypotheses regarding rolling shape. Implications of these differences are unclear and may require further study. Circles in plot represent samples at about every 0.02 seconds

Chapter 4.

Human Walking Isn't All Hard Work: Evidence of Soft Tissue Contributions to Energy Dissipation and Return

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Abstract

The muscles and tendons of the lower extremity are generally considered the dominant producers of positive and negative work during gait. But soft tissue deformations not captured by joint rotations might also dissipate, store, and even return substantial energy to the body. A key locomotion event is the collision of the leg with ground, which deforms soft tissues appreciably in running. Significant deformation might also result from the impulsive ground collision in walking. In a study of normal human walking (N = 10; 0.7 – 2.0 m/s speeds), we show indirect evidence for both negative and positive work performed by soft tissue, consistent with a damped elastic collision and rebound. We used the difference between measured joint work and another quantity—the work performed on the body center of mass—to indicate possible work performed by soft tissue. At 1.25 m/s speed, we estimated that soft tissue performs about 7.5 J of negative work per collision. This constitutes about 60% of the total negative collision work and 31% of the total negative work per stride. The amount of soft tissue work during collision increases sharply with speed. Each collision is followed by 4 J of soft tissue rebound also not captured by joint work measures. Soft tissue deformation may save muscles the effort of actively dissipating energy, and its elastic rebound could save up to 14% of the total positive work per stride. Soft tissues not only cushion impacts but also appear to perform substantial work.

Introduction

Locomotion is accomplished by performing positive and negative mechanical work with the body. During steady gait on level ground, no net work is performed on the environment, so that mechanical work of the body must sum to zero over a stride. Muscles, in series with tendons, are recognized to provide most of the positive work through rotations of the joints. But it is less appreciated that soft tissues, such as plantar fascia, cartilage and the viscera, may deform and perform significant negative work without necessarily rotating the joints. Although much of this work may be dissipative, some may be elastic, implying the possibility for energy return. Work by soft tissue deformation may be helpful for locomotion if it reduces the negative work needed from active muscle, or if it performs some of the positive work. How and where this work occurs may influence the likelihood of injuries and degenerative damage to tissues. We therefore seek to quantify the contribution of work from soft tissue deformations to human walking.

Soft tissues certainly deform during human walking [93–95]. For example, empirical data show substantial deformations of the heel pad [96–99] and foot arch [100], [101]. Forces are transmitted through the rest of the body in a traveling wave [102–104], and “wobbling mass” models show that soft tissue motion can explain the forces transmitted due to impacts from running [105–107] and jumping [108–111]. Similar effects may apply to walking [112], where the relevant soft tissue work may be performed by motion of the viscera, compression of the intervertebral discs, heel pads or joint cartilage, or even transverse muscle motion (their deformation due to inertia as opposed to active shortening). The prior literature primarily focuses on the effect of soft tissues on vibrations and joint forces and torques. There is, however, little experimental evidence regarding the work performed by soft tissue deformations during walking.

One reason why evidence is limited is that soft tissue work is difficult to measure. In human studies, the standard method of quantifying work is inverse dynamics analysis (e.g., [113], [114]), which estimates the joint torques and powers. The integrated power, or *joint work,* is the result of both concentric and eccentric muscle actions as well as passive tendon elasticity, acting to rotate the joints. Inverse dynamics is based on an assumption of rigid bodies and does not quantify soft tissue deformations between or within them. Previous studies have noted how force and torque errors may result from incorrect rigid body assumptions [110], [111], [115], but few have examined the effect on the mechanical energetics of walking. The unmodeled soft tissue dynamics mean that joint work estimates from rigid body models may be insufficient to summarize the work performed by the entire body. For the purposes of this study, we define *soft tissue work* as that not performed by lower-limb joint rotations, and therefore not captured by rigid-body inverse dynamics in traditional gait analysis. An example of such work is that performed by passive dynamic walking machines that can descend a gentle slope with freely swinging joints [116]. Inverse dynamics would be expected to yield practically no joint work, even though there is clearly energy lost in each leg's collision with ground, and even though the legs appear to be rigid.

Indirect evidence suggests that joint work fails to capture significant work performed elsewhere in the body. Using inverse dynamics, DeVita et al. (2007) [117] found that the negative work estimated for the lower extremity joints during stance phase was 32% lower than the positive work (-34 vs. 50 J per step, not including swing phase) in subjects walking at 1.5 m/s. We have hypothesized that substantial negative work is performed by soft tissue and cannot be captured by conventional inverse dynamics [112], potentially explaining this work inconsistency. In order to test this hypothesis and study the energetic role of soft tissue, additional methods of quantifying human locomotion are needed to complement inverse dynamics.

As a point of comparison, we propose using a second measure, of the work performed on the body center-of-mass (COM). The *COM work* is defined as the vector dot product of each limb's ground reaction force with the COM velocity

(Figure 4.1) obtained by integrating the ground reaction forces [118]. We have used this method to show that the collision of the leg with the ground performs negative work on the body's COM [119], [120] in the first 15% of a stride (beginning with heel-strike). The collision work is about 14 J at 1.25 m/s, and increases sharply with walking speed [118], [119]. The COM work analysis makes no assumptions about rigid bodies and therefore captures both joint and soft tissue work. However, it does not estimate individual joint contributions or work performed relative to the COM, the latter generally considered small during stance phase [121–123]. Despite these limitations, comparison of COM and joint work may provide insight into the nature of soft tissue work not captured by the lower extremity joints. We use the difference between these two measures, along with additional supporting evidence, as an indirect indicator of soft tissue work in human walking.

The purpose of this study was to quantify the contribution of soft tissue work to human walking. We propose that mechanical work captured by COM work, but not by inverse dynamics, is indicative of such soft tissue work. This comparison does not specify the location or type of tissue performing the work, but it does roughly indicate the magnitude and timing. Based on dynamic walking principles [112], [124], we hypothesize that (1) soft tissue performs significant negative work during the collision of the leg with the ground and (2) soft tissue dissipates more collision energy at faster walking speeds. These hypotheses were tested with measurements of steady walking performed by normal human participants, detailed as follows.

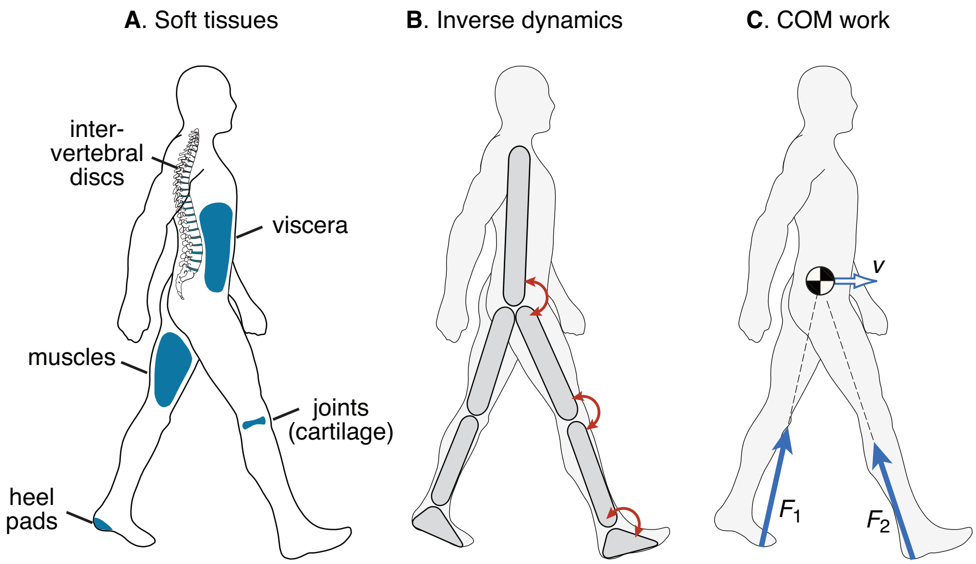

Figure 4.1. Soft tissues of the body and models for estimating work. (A) Deformation of tissues such as the heel pad, joint surfaces, muscles, viscera, and intervertebral discs may affect walking. (B) The standard inverse dynamics model for gait analysis includes the ankle, knee and hip joints of each leg, and computes joint work from force and torque balances between body segments that are assumed rigid. (C) Analysis of COM (center-of-mass) work does not assume rigid bodies, but quantifies the work performed by the two legs to move the COM, treated as a point mass. We compare estimates of joint work and COM work during normal human walking and propose that differences between these methods may be indicative of soft tissue contributions.

Methods

We compared mechanical work estimated from conventional inverse dynamics and COM work analysis for subjects walking across a range of speeds. We used the difference between rigid-body joint work from inverse dynamics and the whole-body COM work as an indicator of soft tissue deformations. These differences were examined in terms of individual phases of the gait cycle as well as over the entire gait cycle. We hypothesized that soft tissue work would increase with greater ground collisions at faster walking speeds. Consequently, we predicted that at higher speeds inverse dynamics would show greater net positive work over a stride, whereas COM work per stride would sum to zero regardless of speed. We measured kinematics and ground reaction forces for 10 subjects walking in normal street shoes on an instrumented treadmill at 8 speeds ranging from 0.7 to 2.0 m/s. All subjects (7 males, 3 females) were healthy and had no known gait impairments or abnormalities (24 ± 2.5 years old, 73.5 ± 15 kg, 1.76 ± 0.11 m). This study was approved by the University of Michigan Institutional Review Board and all subjects gave informed consent prior to participation in the experiment.

Ground reaction forces and lower-body kinematics were collected according to standard gait analysis procedures. Forces were recorded on a custom-built split-belt instrumented treadmill located in the Human Neuromechanics Laboratory at the University of Michigan. Separate force plates (Bertec, Columbus, OH) mounted beneath each belt of the treadmill independently measured reaction forces under each foot at 1200 Hz. Force plates were calibrated based on the methods described by Collins et al. (2009) [92]. Kinematic data were collected at 120 Hz via an 8-camera motion capture system and software (Motion Analysis Corp., Santa Rosa, CA). Passive, reflective markers were placed bilaterally on the ankle (lateral malleolus), knee (lateral epicondyle) and hip (greater trochanter). Additionally, we placed stiff marker triads on each thigh and shank, three markers on the pelvis (sacrum, left/right anterior superior iliac spine) and two markers on each foot (calcaneous, fifth metatarsal).

Randomized experimental trials consisted of subjects walking at self-selected stride frequency at each of the following 8 speeds: 0.7, 0.9, 1.1, 1.25, 1.4, 1.6, 1.8 and 2.0 m/s. Walking trials at each speed lasted 60 seconds, of which the middle 40 seconds were analyzed as representative of steady-state walking. The number of steps per trial varied based on subject and speed, but typically included at least 20 strides. Crossover steps on the split-belt treadmill, in which both feet simultaneously affected the same force plate, were omitted from analysis because of the need for limb-specific forces. Prior to the study, subjects were allowed a short acclimation period to adjust to treadmill walking. Of the 80 total trials (10 subjects, 8 trials each), 3 trials were excluded from analysis due to errors in data acquisition.

Inverse dynamics calculations (Figure 4.2) were performed using standard commercial software (Visual3D, C-Motion, Germantown, MD, USA) and its associated anthropomorphic model. We used a commercial package because it is representative of the procedure used by many laboratories, and because any standard method would be expected to yield similar trends. Analog force data was filtered at 25 Hz and marker motion was filtered at 6 Hz (Butterworth low-pass) prior to inverse dynamics calculations. Joint moments and powers were computed in all 3 dimensions. To facilitate comparison with COM work rate, we summed joint power in all planes and refer to this as summed ankle-knee-hip power, or total joint power. We produced summary measures of net work by integrating power over the entire gait cycle (defined as one stride, from heelstrike to subsequent heelstrike of the same limb), as well as over individual phases of gait as defined below.

We computed COM work rate independently for each limb (Figure 4.1). The work rate was calculated from the 3-dimensional dot product of each limb's ground reaction force with COM velocity [118]. COM velocity was determined from integration of ground reaction forces, assuming steady-state, periodic strides. We defined positive and negative COM work as the integrals over regions of positive and negative COM work rate, respectively. This work summarizes fluctuations in the energy of the COM, but not of motions relative to the COM which appear to contribute less to the overall energy of the body [121], [123]. From the beginning to end of a periodic stride of level walking, no net mechanical work is performed on the COM, assuming negligible air resistance and ground deformation. For many imperfectly periodic strides, we still expect the average summed positive and negative COM work for the body to be approximately zero.

One reason that net joint work may be non-zero is because soft tissue deformations may also perform work. We have previously hypothesized that this may occur during the collision of the leg with the ground following heelstrike, and have also speculated that there may be some passive elastic rebound following the collision [112]. Soft tissues may perform negative work and then return some fraction as positive work, and thus perform net negative work over an entire stride. Because joint work is predicted not to capture soft tissue work, we predict the total ankle-knee-hip work to be measured as net positive over a stride. To determine when the soft tissue work might occur within a gait cycle, we compare summed joint power against COM work rate. Even though the two are different measures of work, their difference may serve as a rough indicator of soft tissue work. To perform this comparison, we found it convenient to divide the gait cycle into five phases defined by major regions of positive and negative COM work (Figure 4.3): Collision (approximately 0-15% of stride), Rebound (15-30%), Pre-load (30-45%), Push-off (45-65%) and Swing (65-100%). We predicted that the greatest mismatch between joint work and COM work would occur during Collision, and this mismatch would increase with walking speed.

All analysis was performed on a stride-by-stride basis. For example, work values were computed for each stride in a trial, and these were averaged across strides to yield mean work for a given trial. All power and work analyses were performed with non-dimensionalized values to account for size differences between subjects, using body mass M, leg length L, and gravitational acceleration g as base units. Mean normalization constants were then used to re-dimensionalize values for reporting purposes. Average power and work normalization constants were $Mg^{3/2}L^{1/2}$ = 2357 W and MgL = 727 J, respectively.

Primary statistical analysis was performed using analysis of covariance to determine significance of work trends and offsets across speed. To examine work per stride and work per phase of gait trends across walking speed v, we performed a one-way analysis of covariance with $v^{2.8}$ as the predictor variable and work as the response. The 2.8 exponent was based on a prediction of collision work per step $W \propto v^2 l^2$ (with step length l) from dynamic walking models [120], [125], combined with the empirical relationship $l \propto v^{0.42}$ [126], [127]. We have previously found normal walking data to fit this relationship well [119], although for statistical comparisons we do not consider the particular exponent to be critical. Work trends were fit to $W = Cv^{2.8} + D$, where C is the coefficient and D is an offset. In some instances paired Student's t-tests were used as a secondary statistical means to compare COM and summed ankle-knee-hip work at each walking speed. In all analyses p-values less than 0.05 were considered statistically significant.

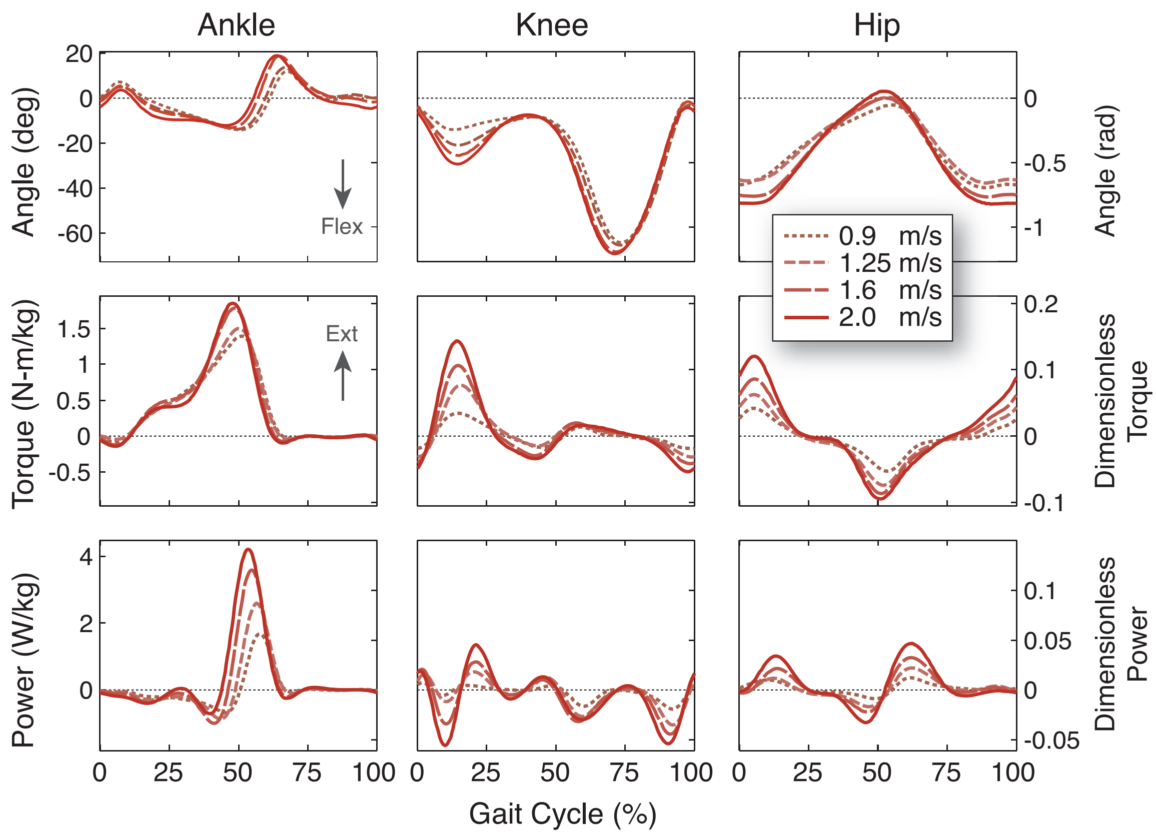

Figure 4.2. Average joint angle, torque, and power trajectories vs. time, as recorded for four walking speeds. Torques and powers were calculated from standard inverse dynamics methods, and were found to scale quite consistently with walking speed. Data shown are sagittal plane values, averaged across subjects ($N = 10$) and normalized to a gait cycle beginning with heel strike, although calculations of work were performed in all three dimensions. Angles and torques are defined as positive in extension. Standard gait analysis units are shown on the left-hand axes, and dimensionless scales are shown on the right-hand axes, using body mass, leg length, and gravitational acceleration as base units.

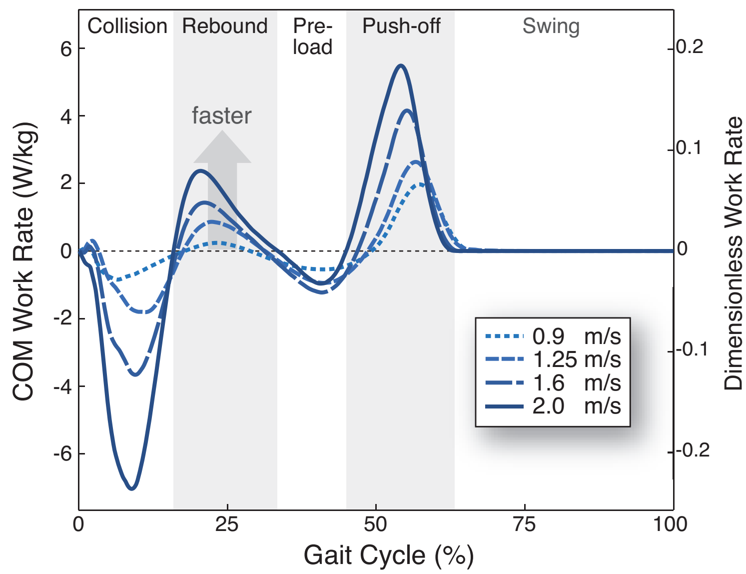

Figure 4.3. Average COM (center-of-mass) work rate for a single limb vs. time for a gait cycle, at four walking speeds. The COM work rate generally followed a consistent pattern of negative and positive work fluctuations, used to define five gait phases: Collision, Rebound, Pre-load, Push-off, and Swing. Data shown are average from all subjects (N = 10).

Results

We observed a qualitative correspondence between joint power and COM work rate (Figure 4.4). The summed ankle-knee-hip power generally displayed regions of negative work during Collision and Pre-load, and positive work during Rebound and Push-off, as is typical of COM work rate. The correspondence was less strong during Collision, where the summed ankle-knee-hip power was more positive than the COM work, indicating less overall negative joint work. Another difference was at the end of the swing phase, where the knee performs negative work over the final 20% of the stride. In contrast, the COM work rate is calculated through the stance leg and is not suitable for quantifying work of the swing limb. Therefore, COM and joint work are not directly compared during Swing. At the level of the joints, the COM work of Collision and Rebound could largely be attributed to the knee, and Pre-load and Push-off to the ankle, with less obvious correspondence at the hip.

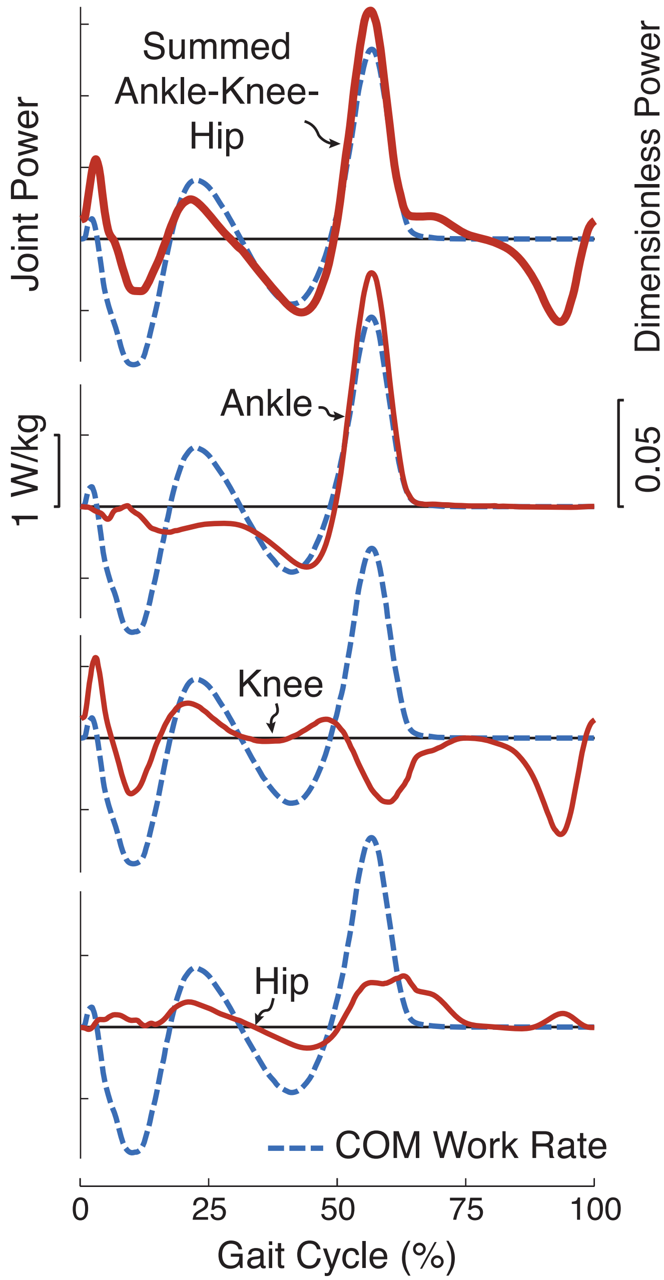

Figure 4.4. Single limb joint powers (solid lines) over a gait cycle, compared with COM work rate (dashed lines) for walking at 1.25 m/s. Ankle, knee, hip, and summed ankle-knee-hip powers for one stride are shown averaged across subjects. Summed ankle-knee-hip power was found to fluctuate between negative and positive work in rough correspondence with COM work rate.

A quantitative comparison of the work performed over each phase revealed notable trends with walking speed. The magnitudes of summed joint work and of COM work increased roughly with $v^{2.8}$ for all phases (Figure 4.5), except for Pre-load where the magnitudes actually decreased slightly. The largest difference between COM and ankle-knee-hip work was during Collision. This difference increased with walking speed, from 3.8 J at 0.7 m/s to 33.0 J at 2.0 m/s. The trends were significantly different, in both curve fit proportionality coefficient (P = 2e-25) and offset (P = 0.03). Furthermore, at all walking speeds the COM Collision work was significantly larger in magnitude than ankle-knee-hip work (paired t-tests, $P < 0.05$). These results are consistent with our expectations that joint work would not fully capture Collision work, by an amount increasing with gait speed.

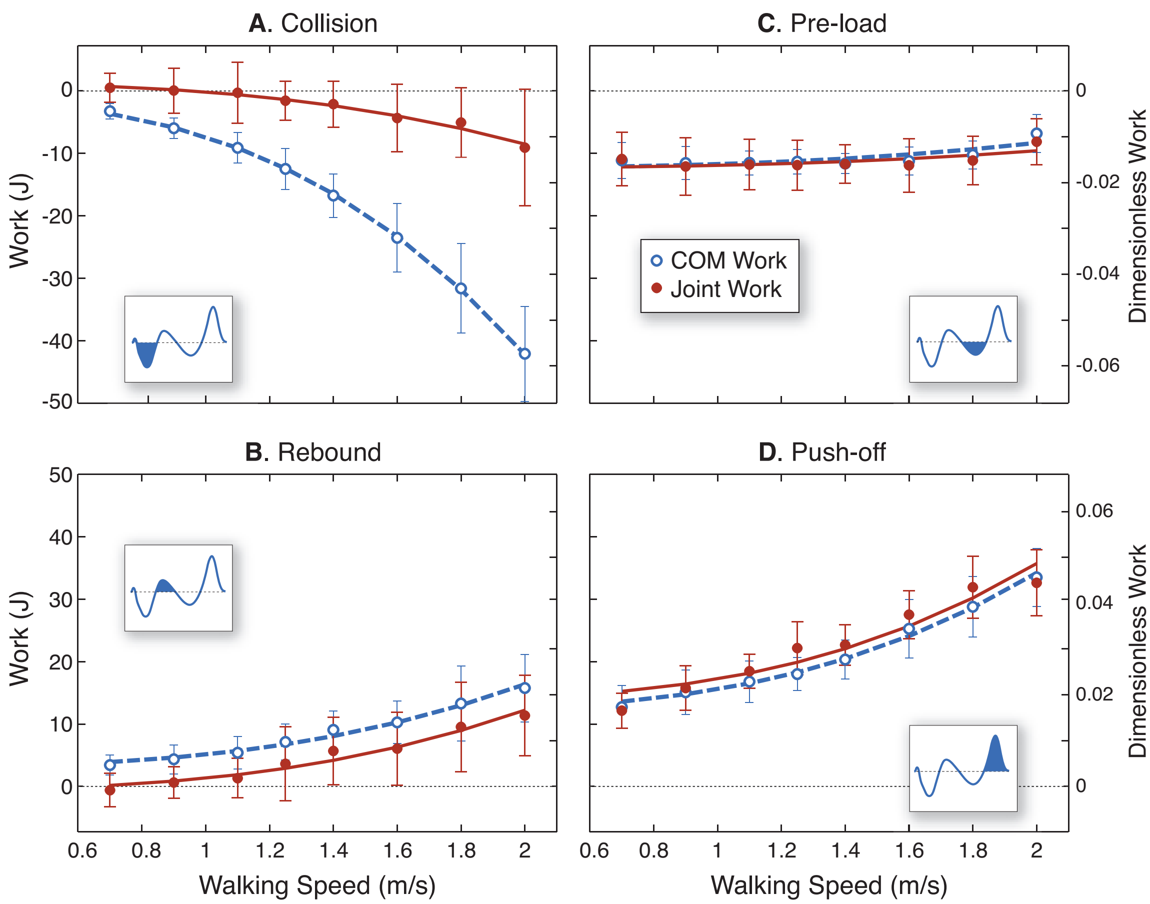

Figure 4.5. Average COM and summed ankle-knee-hip joint work for each phase of the gait cycle, plotted across walking speeds. Phases shown are (A) Collision, (B) Rebound, (C) Pre-load, and (D) Push-off. The summed joint work (filled circles) failed to capture significant negative work during Collision as compared to COM work (open circles), with a difference increasing significantly with walking speed v. An approximately constant but significant work difference was observed in Rebound. There was greater agreement during Pre-load and Push-off. Data were fitted with trends increasing with $v^{2.8}$ based on dynamic walking model predictions for Collision [119], [120], [127]. Data shown are means (filled and open circles) and standard deviations (error bars) across all subjects (N = 10). Curve fit R^2 values, fit coefficients and offsets are reported in the Appendix.

Another substantial difference was in the positive work of Rebound, with consistently less ankle-knee-hip joint work than COM work. This difference averaged 3.9 ± 0.4 J (mean ± s.d.), with maximum 4.4 J at 2 m/s and minimum 3.4 J at 1.4 m/s. Covariance analysis revealed no significant difference in fit coefficients (P = 0.85), but significantly different offsets (P = 0.001). Paired t-tests at each speed also showed significant differences (P < 0.05) at five of the eight walking speeds (0.7, 0.9, 1.1, 1.25, 1.6 m/s) and marginally significant differences (P < 0.08) at the remaining speeds. These results are consistent with a damped elastic Rebound of soft tissues that is not captured by joint work.

The observed COM and summed ankle-knee-hip work during the Pre-load and Push-off phases were in strong agreement. Neither phase showed a significant difference in fit coefficients or offsets (P > 0.05). Examining each speed separately, COM and summed ankle-knee-hip work magnitudes were not significantly different in Push-off and Pre-load across speeds (t-test, P > 0.20), with the single exception of Push-off work at 1.25 m/s (t-test, P = 0.04).

Net ankle-knee-hip work per stride was observed to increase with speed (Figure 4.6). On average, summed joint work for a single limb over the gait cycle was close to zero at slower walking speeds (e.g., -2.70 ± 7.38 J at 0.7 m/s), and increasingly net positive at faster speeds (e.g., 17.75 ± 16.63 J at 2.0 m/s). In contrast, net COM work was consistently small across walking speed, as expected. At 1.25 m/s, net ankle-knee-hip work was approximately five times greater than net COM work, 6.33 vs. 1.28 J, and at 2 m/s it was approximately fifty times greater, 17.75 vs. 0.35 J. Curve fits of $v^{2.8}$ revealed significant coefficient differences between COM and joint work (P = 3e-8). Meanwhile, regression offsets were not significantly different (P = 0.09).

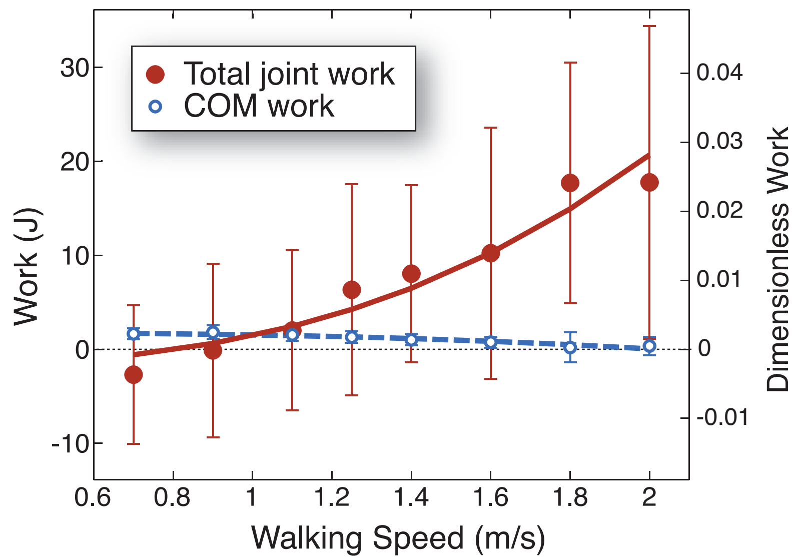

Figure 4.6. Average summed ankle-knee-hip work and COM work per stride across walking speeds. Net COM work (open circles) was close to zero at all speeds, as is required for steady gait. In contrast, summed ankle-knee-hip work (filled circles) showed a strong increase with speed, with less negative work captured than positive work at most speeds, indicating an inconsistency in joint work estimates. Data shown are means (filled and open circles) and standard deviations (error bars) across all subjects (N = 10). Curve fit R^2 values, fit coefficients and offsets are reported in the Appendix.

Discussion

We had sought to determine whether soft tissues contribute significant work to human walking. We use non-joint work—that not captured by inverse dynamics—as an estimate of soft tissue contributions. We tested for evidence of such work and whether its magnitude increased with walking speed. Our results suggest that negative work is indeed performed by soft tissues, to a degree perhaps comparable to the joints themselves. This dissipative soft tissue work occurs primarily during Collision and increases with gait speed. The joint work captured by rigid-body inverse dynamics may therefore seriously underestimate the total negative work performed by the body, and perhaps even some of the positive work as well. We next examine these findings in detail, along with their underlying assumptions.

We found two indicators of soft tissue work, the first coming from the joint work results alone. Net ankle-knee-hip work over a stride was measured as positive for most speeds (Figure 4.6). Negative ankle-knee-hip work over a stride was 6.3 J or about 18.6% less than the positive work at 1.25 m/s. This represents a self-inconsistency in joint work measurements, because the net mechanical work performed over a stride of steady walking must be zero. Inverse dynamics consistently fails to capture a significant percentage of work, especially negative work, performed by the body during gait.

The second indicator comes from the difference between joint work and COM work, which indicates when in the gait cycle soft tissue work might be performed. Results suggest that substantial negative soft tissue work is performed during Collision. At the nominal 1.25 m/s, ankle-knee-hip negative work (ignoring the early positive transient, see Figure 4.2) failed to capture about 7.5 J during Collision, which amounts to about 31% of the negative work per stride, using COM work for comparison. Across all walking speeds, this soft tissue work appears to constitute about 60% of the negative Collision work. The high forces and rate of work associated with Collision phase appear well-suited for deforming soft tissues in human walking. As a point of comparison, a passive dynamic walking machine descending a 2.3% slope with step lengths similar to humans would perform an equivalent amount of negative Collision work (about 12.5 J per step at 1.25 m/s), even though it performs no work through joint rotations. Our present results suggest that soft tissue deformation in humans may account for most of the negative work following heelstrike, with joint work of the ankle, knee, and hip capturing only a fraction of the total energy dissipated in Collision.

The COM versus joint work comparison also indicates that some positive soft tissue work is performed during Rebound. The observed difference in positive Rebound work was less substantial, approximately 4 J, and varied little with speed. At 1.25 m/s, this difference constituted about 10% of the positive work per stride performed by the lower extremity joints, and 14% of the positive COM work. Soft tissue contributions to Rebound were not proportional to the difference in the Collision, but might nonetheless represent a damped elastic recoil not attributable to ankle-knee-hip joint rotations. Elastic energy return by soft tissue could perform 10 – 14% of the positive work otherwise required of active muscle, perhaps saving a roughly proportionate amount of metabolic energy.

Several trends were observed in mechanical work as a function of walking speed. The magnitude of COM Collision work increased approximately with speed raised to the 2.8 power (R^2 = 0.89, Figure 4.5), as predicted by dynamic walking models [127], [128]. The positive COM work during Push-off also increased at that rate (R^2 = 0.77), as it largely offsets the negative Collision work [119], [120]. The slight Pre-load work trend, for which there was no prediction, was decreasing in magnitude with speed. There was also no explicit trend predicted for net ankle-knee-hip work (Figure 4.6) other than an increase with speed. The measured net joint work over a stride did in fact increase with speed, and its difference with COM work also increased during Collision. These findings are consistent with our hypothesis that substantial negative work is performed by soft tissue deformations during Collision.

Our conclusions are based on experimental estimates for mechanical work. To minimize methodological errors, we followed standard gait analysis procedures for motion capture and inverse dynamics, and found our results in good agreement with prior joint kinetics literature [114]. One area of sensitivity is to joint center location [114], [129], for example at the knee [130]. We therefore performed a sensitivity analysis in which the knee joint center was artificially translated fore-aft by ±3 cm from the nominal rigid-body model. This changed the summed ankle-knee-hip work results by a substantial offset of 12-20 J across all speeds, but had virtually no effect on the trend with walking speed (see Figure 4.7). Our results also appear consistent with prior estimates using independent measurement and filtering methodologies to estimate joint work [114], [117] and COM work (using overground force plates [118]). Similar trends regarding soft tissue work are also reported by Soo and Donelan (2010) [131], who applied analogous comparisons to a task that isolates step-to-step transitions from human walking.

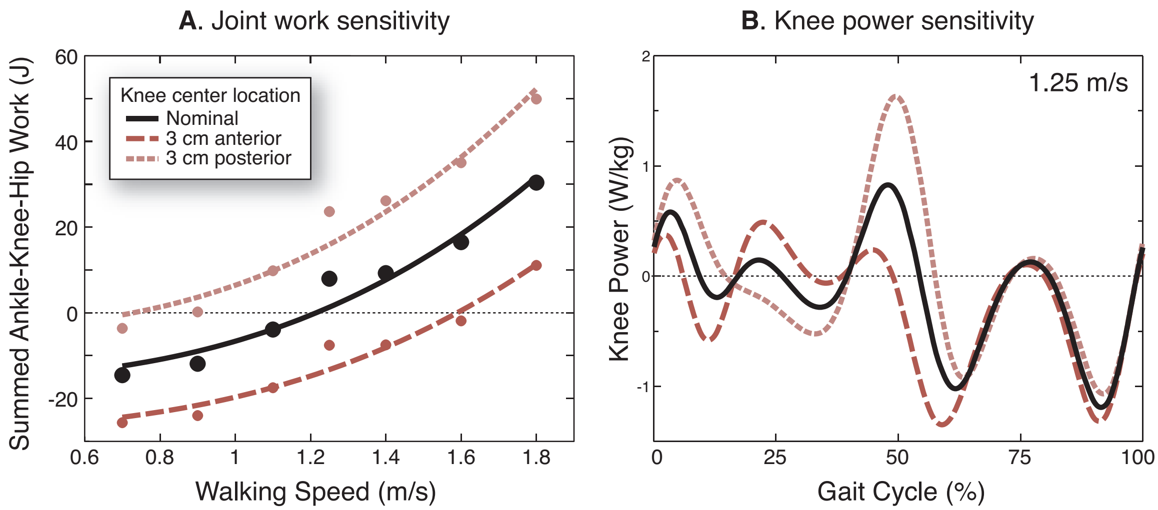

Figure 4.7. Sensitivity of inverse dynamics to knee joint center location, examined with data from a single subject. (A) Perturbation of the observed knee joint center by 3 cm in anterior and posterior directions caused offsets in the magnitude of net ankle-knee-hip work per stride, but did not affect the speed-dependent trend. (B) Perturbations caused substantial changes in knee power trajectory, but the overall discrepancies in joint work are not explained by possible methodological errors in marker placement. The nominal knee joint center was defined by motion capture markers on the medial and lateral epicondyles of the femur.

There are limitations to directly comparing COM and joint work. COM analysis quantifies only the work performed on the COM and assumes most of it to be performed by the legs. Large rotational motions such as pitching of the trunk or swinging of the arms could therefore potentially cause misattribution of work to the legs. These motions are, however, typically assumed to contribute little to the joint work of normal walking and are often not included in inverse dynamics measurements [114], as was the case here. COM work also does not capture the work performed to move body segments relative to the COM (sometimes referred to as "internal work" as opposed to "external work," e.g., [132]). There is substantial work performed relative to the COM during the Swing phase, especially to slow the swing knee [123], which is one reason why we did not directly compare that phase. As for the stance phase, estimates from the literature show positive work performed relative to the COM during Push-off as the trailing limb accelerates rotationally [121–123], in amount perhaps sufficient to explain the (statistically not significant) difference we observed between COM and joint work (Figure 4.5). Similarly, Collision also shows a relatively small amount of positive work relative to the COM (less than 3 J) that offsets the negative COM work slightly, but explains at most a third of the difference between COM and joint work (Figure 4.5). It therefore appears that COM work might be an overestimate, and joint work an underestimate, of the actual negative work of the entire body during Collision.

Rigid-body assumptions made in standard inverse dynamics also complicate the definition and interpretation of soft tissue work. We have treated a substantial difference between COM and joint work as indirect evidence of soft tissue work, but it is possible that work is performed by joints whose associated bodies might be rigid (but perhaps difficult to measure), as opposed to "soft." It is also possible that the work estimates for each individual joint are simply inaccurate. However, soft tissue deformations are a likely explanation for these joint work inaccuracies, because of well-recognized force and torque errors induced by wobbling mass and rigid body assumptions [109], [111], [115]. All these issues could potentially be addressed by measuring additional rigid body segments not conventionally captured, or by modeling a wobbling mass with additional rigid bodies, albeit with limits to practicality. In gait analysis, much effort is devoted to reducing "skin marker artifact" or "soft tissue motion artifact" (e.g., [110], [113]). Our results suggest that the complete elimination of such artifacts would still leave a "rigid tissue motion artifact," since the skeleton accounts for well less than 20% of total body mass in humans and most other animals [133]. To be truly accurate, inverse dynamics would require the correct displacements and inertias of all moving body parts, which are distributed and continuous as opposed to lumped and discrete. Given these limitations, we interpret our results conservatively by observing work trends across a range of walking speeds, which we believe to be relatively insensitive to errors in absolute work estimates.

This study is not intended to indict inverse dynamics as a method. It is well recognized that rigid body assumptions lead to errors in torques and forces, and our results suggest that these may in turn cause substantial errors in mechanical work estimates during walking. Most studies using inverse dynamics draw conclusions based on controlled comparisons that require precision but not absolute accuracy. We believe inverse dynamics to be quite consistent and to provide good precision, despite limitations in absolute accuracy. Of course, COM work also has several limitations as discussed above, and both methods are indirect indicators of mechanical work performed on the body, let alone by muscles and tendons. Overall, the two methods have different limitations, and should be treated as imperfect but complementary indicators of mechanical work.

We have presented preliminary evidence of soft tissue energy absorption and return. The difference between COM and ankle-knee-hip work provides indirect evidence of soft tissue work, roughly indicating when, but not where in the body it is performed. The greatest impacts are experienced near the ground, and so the heel pad [98], [109], [111], [115], plantar fascia [101], [110], [113], and other tissues of the shank might dissipate substantial energy. They might also provide some damped elastic recoil, but other possible contributors include intervertebral discs [134], articular cartilage [98], [135], [136], and the viscera [137], [138], supported by the elasticity of the peritoneum. Further research is needed to understand the distribution of soft tissue work throughout the body. Our findings also require corroboration, perhaps with more direct observational techniques such as imaging (e.g., [139–141]) and direct strain or force measurements [139]. A challenge in most estimates of soft tissue work is the need for material and other parameters that are difficult to identify from independent experiments, and internal forces and displacements that are difficult to measure. It is therefore helpful to study soft tissue work using multiple approaches.

We believe that soft tissues play an underappreciated role in walking. Not only do they reduce peak impact loads, but they also dissipate, store, and even return energy. Their deformation is well recognized at the level of localized tissues, but the associated work is not considered in most studies of overall gait. The total amount of Collisional negative work is largely dictated by the pendulum-like walking motion [112], and may be distributed between muscle fibers, tendon, and soft tissue deformations [100], [101]. Soft tissue deformation may in fact account for much of the Collisional work, and thus reduce the proportion of negative work performed by muscle and perhaps even the subsequent positive work if there is appreciable elastic Rebound. Also perhaps underappreciated is negative work as a whole, since its existence is the reason why positive work must be performed at all. We propose that negative work is equal to positive work, not only in quantity, but also in scientific importance.

Supplementary Material

Appendix

The quantitative results summarized in the Results are tabulated here. Table 4.1 reports curve fit parameters for COM work, and Table 4.2 reports parameters for ankle-knee-hip joint work. Table 4.3 reports the average stride parameters for the various walking speeds.

Table 4.1. Nonlinear regression parameters for COM work, reported as means across subjects, with 95% confidence interval in parentheses. Coefficients are reported in dimensionless form.

COM Work	**Coefficient *C***	**Offset *D***	R^2
Collision	-0.178 (0.014)	-0.002 (0.002)	0.89
Rebound	0.058 (0.011)	0.004 (0.002)	0.57
Pre-load	0.017 (0.008)	-0.017 (0.001)	0.20
Push-off	0.095 (0.012)	0.017 (0.002)	0.77
Swing	-1.3e-4 (1.3e-4)	4.2e-5 (2.1e-5)	0.05
Net Work (Stride)	-0.008 (0.003)	0.003 (4e-4)	0.30

Table 4.2. Nonlinear regression parameters for ankle-knee-hip joint work, reported as means across subjects, with 95% confidence interval in parentheses. Coefficients are reported in dimensionless form.

Ankle-Knee-Hip Work	**Coefficient *C***	**Offset *D***	R^2
Collision	-0.043 (0.016)	0.002 (0.002)	0.27
Rebound	0.056 (0.016)	-7.e-4 (0.003)	0.39
Pre-load	0.012 (0.013)	-0.017 (0.002)	0.04
Push-off	0.095 (0.014)	0.019 (0.002)	0.71
Swing	-0.022 (0.004)	-0.006 (6e-4)	0.61
Net Work (Stride)	0.098 (0.036)	-0.002 (0.006)	0.28

Table 4.3. Stride parameters for each walking speed, reported as means across subjects, with standard deviations in parentheses.

Speed (m/s)	**Stride Length (m)**	**Stride Time (sec)**
0.7	1.06 (0.12)	1.51 (0.15)
0.9	1.22 (0.11)	1.35 (0.11)
1.1	1.31 (0.10)	1.19 (0.08)
1.25	1.37 (0.08)	1.10 (0.06)
1.4	1.46 (0.10)	1.04 (0.06)
1.6	1.61 (0.12)	1.00 (0.07)
1.8	1.72 (0.13)	0.95 (0.06)
2	1.81 (0.08)	0.90 (0.03)

Chapter 5.

Mechanical Work as an Indirect Measure of Subjective Costs Influencing Human Movement

Published in PLoS ONE (2012).

Abstract

To descend a flight of stairs, would you rather walk or fall? Falling seems to have some obvious disadvantages such as the risk of pain or injury. But the preferred strategy of walking also entails a cost for the use of active muscles to perform negative work. The amount and distribution of work a person chooses to perform may, therefore, reflect a subjective valuation of the trade-offs between active muscle effort and other costs, such as pain. Here we use a simple jump landing experiment to quantify the work humans prefer to perform to dissipate the energy of landing. We found that healthy normal subjects (N = 8) preferred a strategy that involved performing 37% more negative work than minimally necessary (P < 0.001) across a range of landing heights. This then required additional positive work to return to standing rest posture, highlighting the cost of this preference. Subjects were also able to modulate the amount of landing work, and its distribution between active and passive tissues. When instructed to land softly, they performed 76% more work than necessary (P < 0.001), with a higher proportion from active muscles (89% vs. 84%, P < 0.001). Stiff-legged landings, performed by one subject for demonstration, exhibited close to the minimum of work, with more of it performed passively through soft tissue deformations (at least 30% in stiff landings vs. 16% preferred). During jump landings, humans appear not to minimize muscle work, but instead choose to perform a consistent amount of extra work, presumably to avoid other

subjective costs. The degree to which work is not minimized may indirectly quantify the relative valuation of costs that are otherwise difficult to measure.

Introduction

Humans appear to value economy of movement [142–148], leading to the expectation that the muscles will not usually perform more mechanical work than necessary to complete a motor task. While this general observation seems applicable to costly locomotor tasks such as walking, economy of work may also be relevant to other tasks, such as those primarily involving energy dissipation. In these cases, factors other than work and energy expenditure also clearly influence the preferred movement strategy. For example, humans usually prefer to walk down a long flight of stairs, when it might require less muscular effort simply to fall down them, allowing the work to be performed passively, through soft tissue deformations. Falling might save the energy expended to perform active negative work, but perhaps at the expense of other costs, such as pain or risk of injury. It is difficult to quantify other unknown factors such as pain. But a person's own valuation of their relative costs may be indicated behaviorally by how he or she chooses to perform a task, for example actively vs. passively. The amount and distribution of negative work humans choose to perform may therefore indicate a trade-off between work and other, less easily quantified costs, which may explain why some tasks are performed uneconomically.

A task particularly suited for this inquiry is landing from a jump. Landing collisions dissipate the kinetic energy gained from the descent, largely through negative work performed actively by lower extremity muscles [149–152]. The work not due to active muscle is presumably performed passively [153–156], through the deformation of soft tissues such as the heel pad [96–99], viscera [137], [157], [158], and vertebral discs [134]. The proportion of work performed actively vs. passively can be modulated, for example, humans can perform "softer" landings, involving greater flexion of the knees, less passive work, and more muscle work [155]. Greater amounts of active negative work might reduce concentrated strain energy that could cause injury to soft tissues, but such landings may also be more metabolically costly.

Alternatively, humans can perform “stiffer” landings, which are more economical from a mechanical work perspective. After all, even a small amount of joint flexion during jump landing could be considered uneconomical, because it entails doing more than the minimum amount of negative work, which may also require subsequent positive work to compensate. One need only practice a few stiff landings to surmise that pain and discomfort are among the countervailing costs. These and other subjective costs almost certainly play a role in many movement strategies, but they are difficult to quantify and compare against each other. It is here that biomechanical measures may be helpful, because they facilitate quantification of the opportunity cost—in terms of work or energy—of a person’s preferred movement strategy. In terms of jump landing, a preference to perform more than the minimum amount of work may indicate the relative weighting of other costs on the preferred movement strategy.

The purpose of the present study was to quantify the preferred jump landing strategy of humans in the context of work-like costs. We hypothesized that the preferred landing strategy is a compromise between different costs for both stiffer and softer landings. While stiffer landings may reduce active work, they may entail other costs such as pain or discomfort, perhaps related to excessive passive work performed by soft tissue deformations. Softer landings may reduce passive deformations, but at the cost of increased work overall. Therefore, we tested whether the preferred strategy entails performing more negative work than necessary, and investigated how this work is distributed between active and passive tissues.

Methods

To test our hypothesis, we measured the work performed by healthy human subjects when landing from jumps. We compared that work against the minimum necessary to land and return to the same final posture, and tested whether the preferred strategy entailed excess work. We also estimated the contributions of active muscles from joint work computed from standard inverse dynamics, and the work of soft tissue deformations based on the total mechanical work performed on the entire body. We measured landings from 8 healthy adult subjects (77.5 ± 14.4 kg, 0.94 ± 0.05 m leg length, 6 male and 2 female) performing vertical jumping over a range of heights.

Ethics Statement

This study was approved by the University of Michigan Institutional Review Board. All subjects gave written informed consent prior to the experiment.

Subjects performed jumping trials with two landing strategies, treated as separate conditions. In the Preferred condition, subjects were given no landing instructions, whereas in the Soft condition, they were instructed to land as quietly as possible. To avoid affecting subjective preferences, the Soft condition was always tested after Preferred. As a demonstration, one subject also performed an additional condition, Stiff landing, in which he landed flatfooted with his knees fully extended. A trial consisted of standing at rest with one foot on each force plate and with arms crossed, jumping vertically, landing back on the same force plates, and finally returning to the original rest posture (Figure 5.1). We defined *net landing height* as the difference between the maximum height of the body center of mass (COM) and the final standing rest posture (Figure 5.1). We defined the theoretical minimum amount of work necessary for landing as the gravitational potential energy associated with this displacement. This assumes a person could land with all of the joints vertically aligned so that no joint rotations would occur in landing.

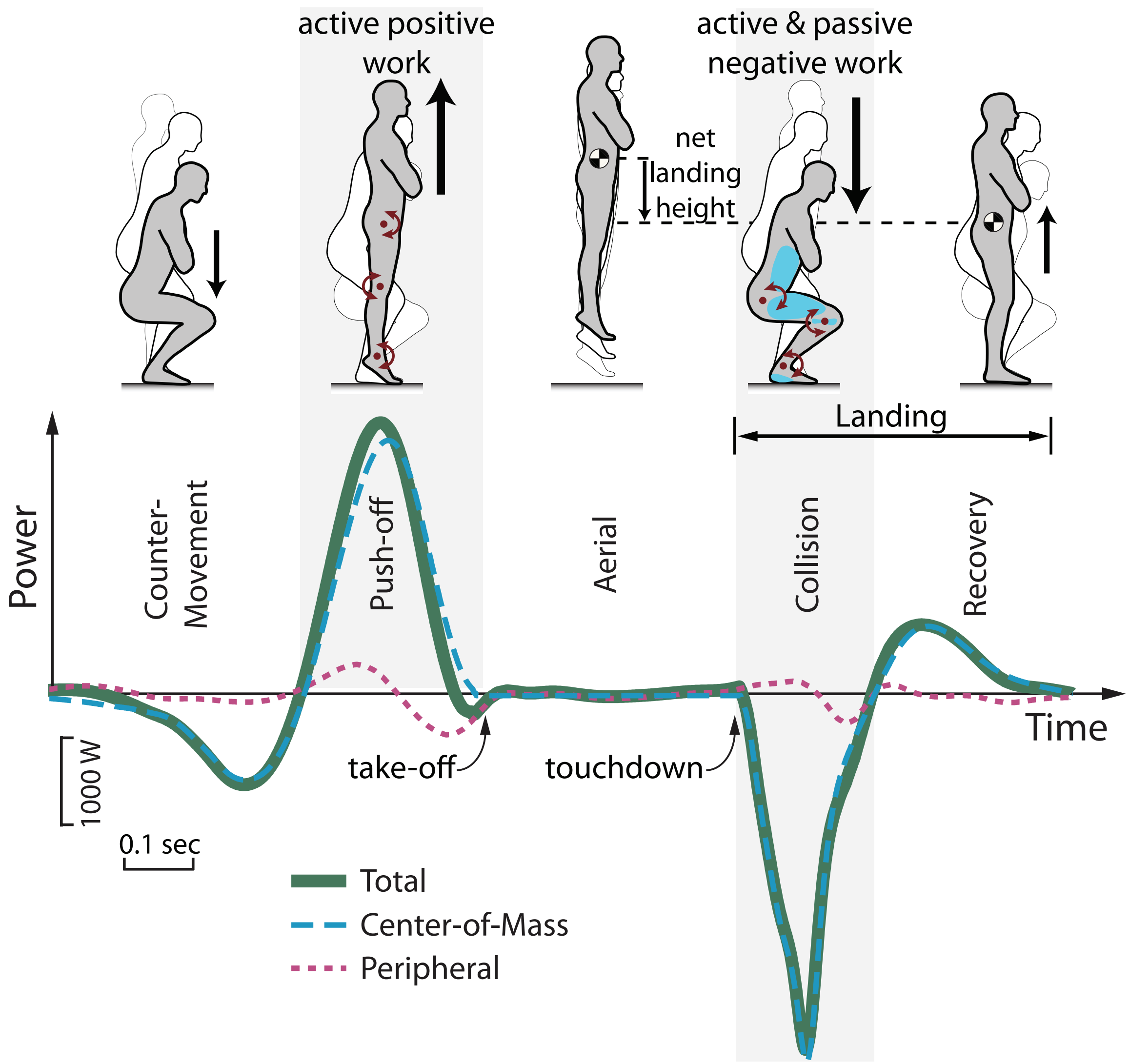

Figure 5.1. Total mechanical power vs. time for vertical jumping and landing. Subjects jumped vertically, landed and returned to rest. Representative trial demonstrates phases of a jump: Counter-Movement, Push-off, Aerial, Collision and Recovery, defined by zero-crossings of center-of-mass (COM) power. Landing is represented by the Collision and Recovery phases. Total power was estimated as the sum of COM power (due to motion of the COM) and Peripheral power (due to motion relative to the COM). Net landing height was defined as the displacement between maximum aerial height of the COM and standing rest posture.

Subjects performed forty jumping trials for each condition, over a range of heights. Subjects received verbal instructions to jump 10 times at each of four different subjective heights between "very low" and "high," to yield a range of landing displacements. Subjects kept their arms crossed throughout the duration of each trial in order to avoid work done by the arms. A trial was performed again if the subject's feet did not land back on their original force plates.

Ground reaction forces and full-body kinematics were collected according to standard biomechanical procedures for inverse dynamics analysis. Forces were recorded under each foot independently using two in-ground force plates (Advanced Mechanical Technologies Inc., Newton, MA, USA) at 1000 Hz. Kinematic data were collected at 100 Hz *via* an eight-camera infrared motion capture system (Vicon Motion Systems, Los Angeles, CA, USA) and software (Vicon Nexus v1.5.0). Passive, reflective markers were placed bilaterally on the ankle (lateral and medial malleoli), knee (lateral and medial femoral epicondyles) and hip (greater trochanter). Additionally, we placed four segmental markers on each thigh and shank, and on the pelvis (left/right anterior and posterior superior iliac spines). Three additional markers were placed on each foot (calcaneous, first and fifth metatarsals). Upper-body markers were placed on the neck (at the level of C7), the shoulders (acromion) and the elbows (olecranon bursa). Joint locations for the ankle, knee and hip were computed based on a functional joint center algorithm [129] in commercial software (Visual3D, C-Motion, Germantown, MD, USA). Prior to analysis, forces were filtered at 25 Hz and marker positions at 6 Hz using a 4th order low-pass Butterworth filter.

We estimated 3-D mechanical work for different net landing heights and conditions and the distribution of work between active and passive tissues. We defined Total Mechanical power as that due to motion of the body center-of-mass (COM work rate) plus that due to motion relative to the body COM (Peripheral power). We computed COM work rate based on the dot product of ground reaction force with COM velocity, also derived from forces [118], [159]. We estimated Peripheral power as the time derivative of changes in translational and rotational energy relative to the COM, assuming rigid-body segments. This is also sometimes referred to as "internal work," (e.g., [160]). As another quantification, we defined Summed Joint power (or simply Joint power) as the combined power from the ankle, knee and hip of both limbs plus the power due to rotation of the trunk about the lumbosacral joint, all using standard rigid-body inverse dynamics methods. We used the Total mechanical work performed on the body, but not captured by rigid-body Joint work estimates as an indicator of soft tissue deformations (similar to [153], [154]). We estimated the Soft Tissue power contribution as the difference between Total Mechanical power and Summed Joint power (Figure 5.2). This assumes that most of the soft tissue deformations were captured by COM work rate, which captures both rigid and soft bodies, as opposed to the Peripheral Power contribution, which only quantifies rigid-body motions (see further details in Supplementary Material).

Work summary measures were integrated from the power estimates over various jump phases. We divided each trial into phases – Counter-Movement, Push-Off, Aerial, Collision, Recovery – defined by separate regions of positive and negative COM work (Figure 5.1). The Collision and Recovery phases together account for the work of landing. Power and work measures were non-dimensionalized before regression analysis to account for size differences between subjects, using body mass (M), leg length (L) and gravitational acceleration (g) as base units. All results are presented in dimensionless units. Mean power and work normalization constants were $Mg^{3/2}L^{1/2}$ = 2302 W and MgL = 712 J, respectively. We computed work measures for each trial individually, and then performed linear regressions on Total and Soft Tissue work with respect to net landing height. Student's t-tests were used to compare regression coefficients and determine statistical significance at a level of $P < 0.001$. We performed fits to $W = Bh$, where W is work, h is net landing height, and B is the slope coefficient, with an assumed offset of zero. We defined the proportion of work done passively during each phase of the jump as the ratio of the Soft Tissue work coefficient (B_{SoftT}) divided by the Total work coefficient (B_{Tot}).

Finally, we performed methodological control tests to validate the novel Soft Tissue work estimates. We asked each subject to perform 10 squatting trials, which involved squatting down slowly, then returning to resting posture. This yielded measurements of joint rotations similar to the jumping trials, but without the aerial phase or jarring landing collision.

Results

Mechanical work varied consistently with net landing height and landing strategy. We generally observed the Preferred landings to involve more negative Collision work overall than the minimum theoretically possible. And Soft landings tended to involve more work than Preferred. When negative landing work was performed in excess of the minimum possible, extra positive work followed in order to return to standing rest. The Total Collision work was distributed between a combination of Joint and Soft Tissue contributions, with the amount and distribution also varying systematically with landing height and strategy. The contribution of Soft Tissue work to the Collision was highest at low landing heights, and approached an approximately constant percentage for higher jumps. During Soft landings we observed more work overall, especially from Joint contributions, and during Stiff landings we observed less work, but with increased Soft Tissue contributions.

During Preferred landings, subjects performed more Total negative work than necessary. At the greatest net landing height of about 42 cm, subjects performed about -477 J of negative Collision work, and then 145 J of positive Recovery work. The theoretical minimum Collision would be about -319 J from the potential energy of body weight (759.4 N) at that height, followed by 0 J of Recovery. The amount of negative Collision work changed approximately linearly with net landing height (Figure 5.2). This work is described by the work coefficient B (total landing work per unit landing height), which in dimensionless units may be interpreted as a relative amount of Collision work compared to the theoretical minimum. A work coefficient of -1 therefore corresponds to that minimum. In Preferred landings, the relative Collision work was -1.37 ± 0.01 (mean ± 95% confidence interval) from a linear fit (R^2 = 0.96), meaning that subjects performed 37% more negative work than minimally necessary ($P < 0.001$; Figure 5.2). This was then followed by a similar amount of positive Recovery work to return to standing rest, with slope 0.339 ± 0.007 (R^2 = 0.76), which was significantly different from zero.

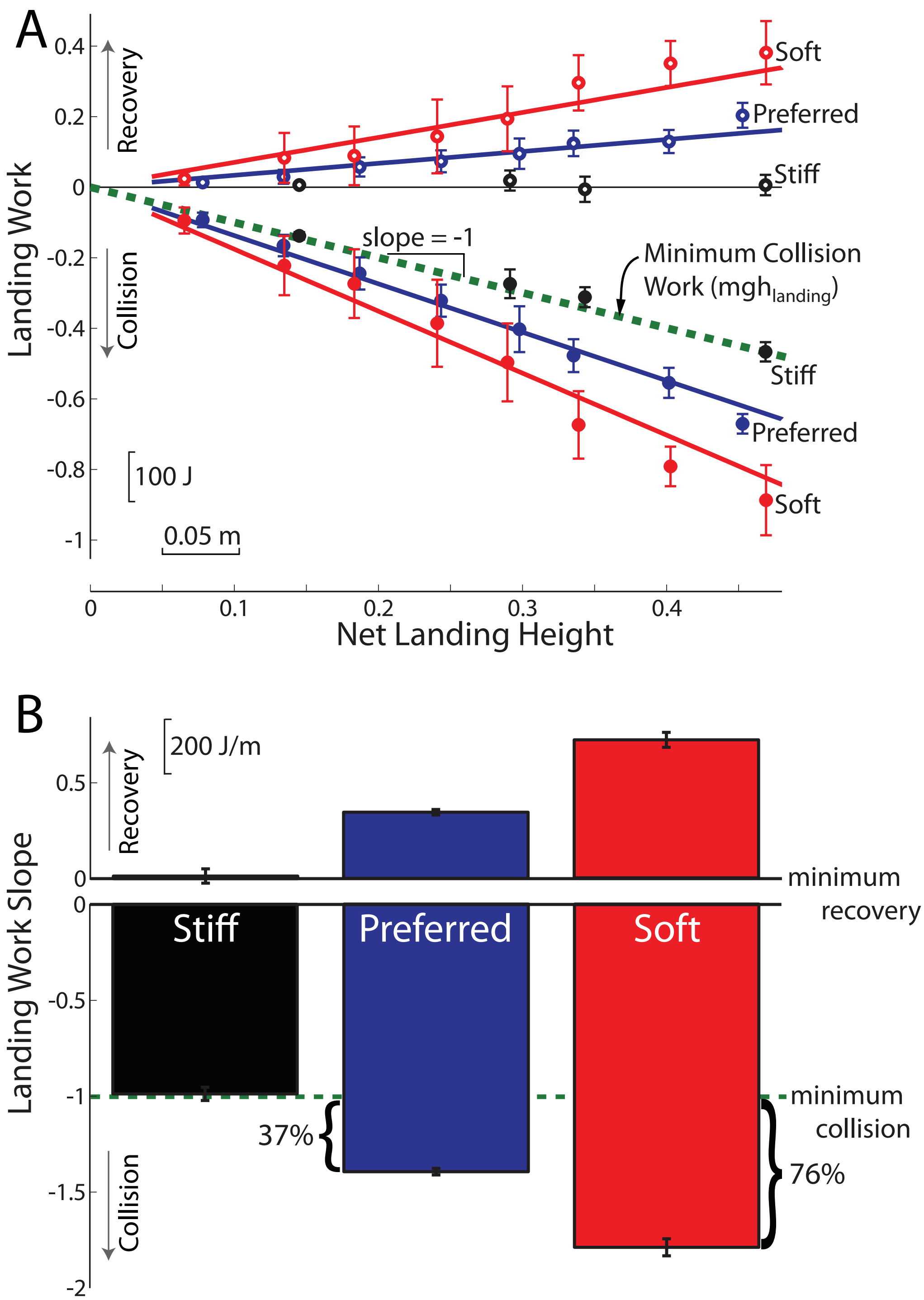

Figure 5.2. Total landing work: plotted (A) as a function of net landing height and (B) relative to minimum possible work for each landing phase and strategy. The theoretically minimum amount of landing work possible was defined as the change in potential energy from peak aerial height to standing rest (dashed line). Stiff landings achieved very close to this theoretical minimum. During Preferred landings, subjects performed 37% more Collision work than the minimum, necessitating additional positive Recovery work. Soft landings were even more extreme, with subjects performing about 76% more negative work than necessary. All relative work amounts were significantly different from each other ($P < 0.001$).

The amount of work could also be modulated by different landing strategies (Figure 5.2). In the Soft landing strategy, subjects performed 76% more Collision work than necessary (slope = -1.76 ± 0.02; R^2 = 0.84), followed by a similar amount of positive Recovery work (0.707 ± 0.020; R^2 = 0.59). The single subject who performed the Stiff landing was able to achieve Collision work magnitudes within 4% of the theoretical minimum, -0.969 ± 0.017 (-735.7 ± 13.1 J/m; R^2 = 0.97), followed by an insignificant amount of positive work after landing, 0.013 ± 0.017 (P = 0.37; R^2 = 0.01). The Collision work of Soft and Preferred landings were significantly different from each other and from the theoretical minimum. For example, for the highest net landing heights (approximately 42-44 cm), the Soft landing Collision work was about -630 J, compared to -480 J for Preferred and -330 J for Stiff landings.

We observed indications of substantial Soft Tissue work, specifically during Collision (Figure 5.3, Figure 5.4). Total Mechanical power exhibited a large positive region of Push-off before take-off, followed by a large negative region of Collision immediately after touchdown. Summed Joint power followed a similar pattern, but with a lower magnitude of negative work during Collision, indicating contributions from Soft Tissue work (Figure 5.3). Representative Joint powers are shown in the Supplemental Material (Figure 5.8, Figure 5.9, Figure 5.10), as are two other independent indicators of Soft Tissue work (Figure 5.7), further demonstrating that Joint work estimates fails to account for the Total work performed during landing.

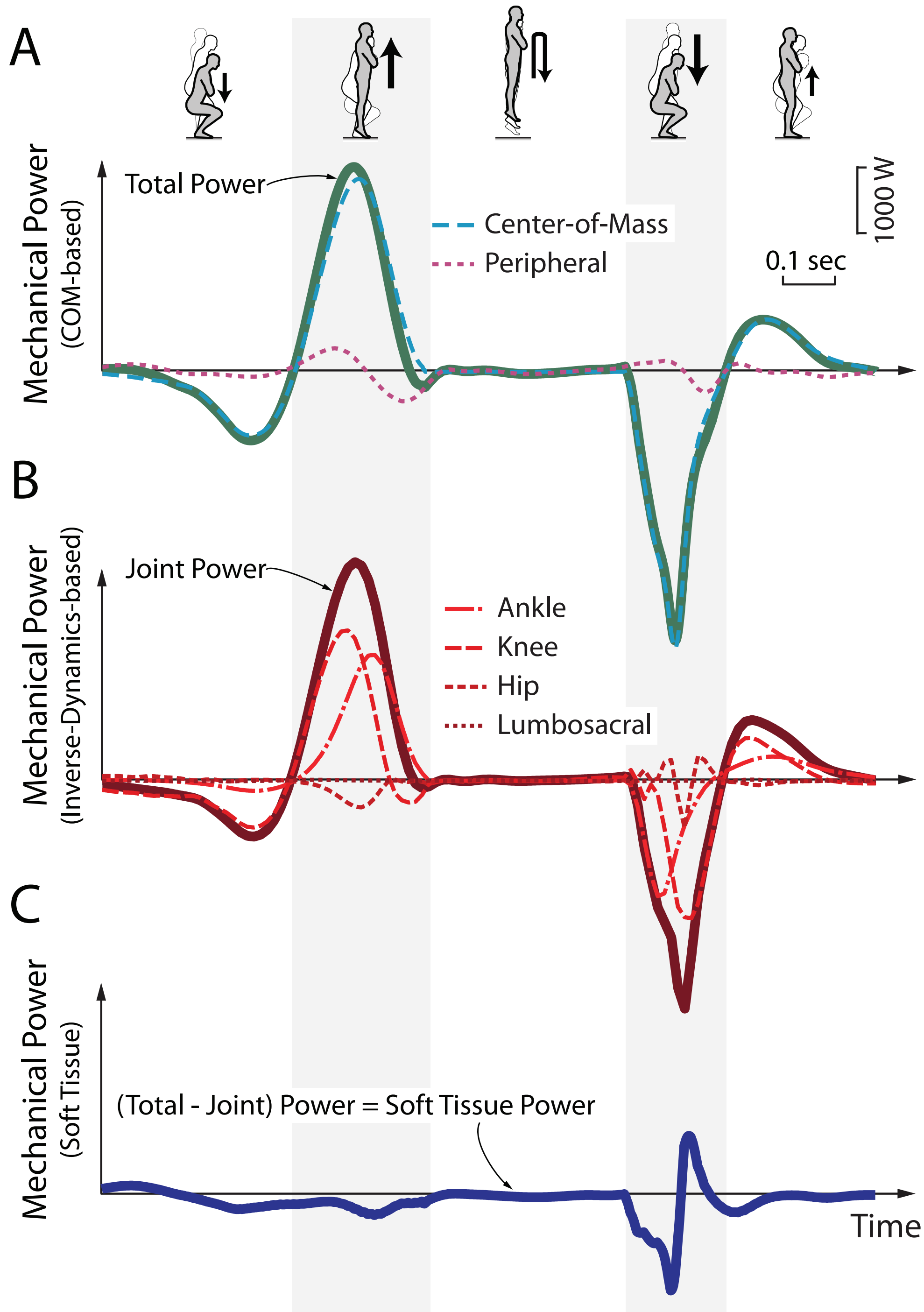

Figure 5.3. Mechanical power estimates. (A) Total power was estimated as the sum of center-of-mass (COM) power (due to motion of the COM) and Peripheral power (due to motion relative to the COM). (B) Joint power represents net contributions from muscle-tendon acting about the joints (ankle, knee, hip, lumbosacral), based on standard inverse dynamics. (C) Soft Tissue power is defined as the Total power minus the Joint power (see Supplementary Material for more details on calculations). Soft Tissue power was close to zero in most phases of the jump, except during Collision (immediately after touchdown), when it exhibited regions of negative, then positive power.

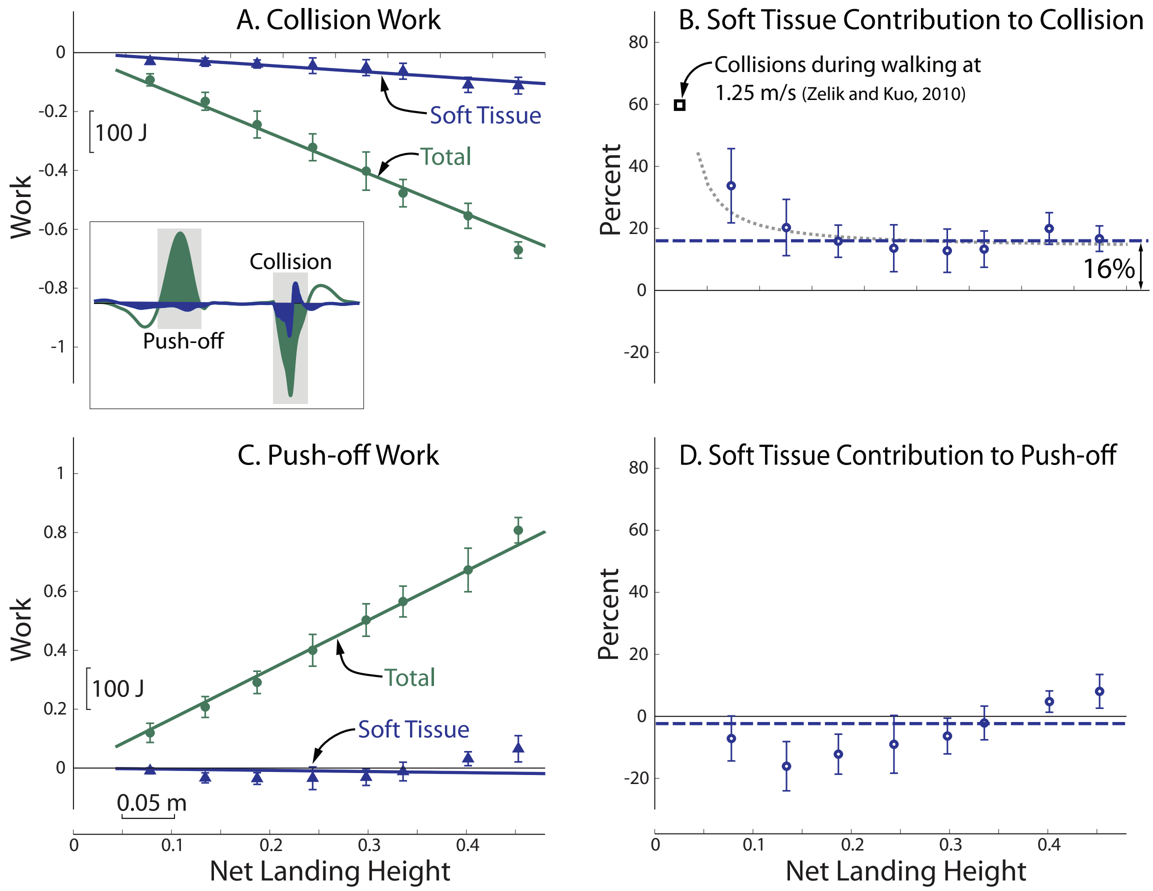

Figure 5.4. Collision and Push-off work, and contributions from passive soft tissues to Total work during Preferred landing. Total and Soft Tissue work are plotted as a function of net landing height for (A) Collision and (C) Push-off (N=8). Passive contributions were computed as the ratio of linear regression slopes (BSoftT/BTot) for (B) Collision and (D) Push-off, yielding asymptotic proportions of 16% and -2%, respectively (shown as dashed lines). Soft Tissues therefore contributed substantially to Collision, but not to Push-off. For low net landing heights, the Soft Tissue contribution to Collision appeared to be greater than the asymptote (deviation shown as gray dotted line). That deviation appears consistent with heelstrike Collisions during walking, which are similar in magnitude to landings of about 3 cm and have Soft Tissue contributions of about 60%, as estimated previously [153].

Joint and Soft Tissue Collision work both increased with net landing height (Figure 5.4). For Preferred landings, the Soft Tissue work of Collision was about -79.8 J for net landing heights of 42 cm (Figure 5.4). Soft Tissue Collision work exhibited an approximately linear relationship with net landing height, with work coefficient -0.215 ± 0.005 (R^2 = 0.50). Soft Tissue work was therefore about 16% of Total Collision work, a proportion approached at greater landing heights (Figure 5.4). But the ratio was generally higher for low landing heights, for example 34% at 7.3 cm, with some subjects exhibiting percentages as high as 50-70% on individual trials.

Subjects were able to modulate the amount and distribution of work during landing Collisions as a function of landing strategy. During Soft landings, subjects performed more Collision work through Joint rotations, and thus less through Soft Tissue than during Preferred landings (Figure 5.5, 10.6% vs. 16.0%). This difference was primarily due to a significant increase in the magnitude of Total Collision work, and to a lesser extent to a significant decrease in the magnitude of Soft Tissue work, -0.186 ± 0.007 vs. -0.220 ± 0.005 (Soft vs. Preferred, Figure 5.5), with the relative differences indicating an increase in Joint work. The opposite occurred with the single subject's data for Stiff landings, where Soft Tissue work constituted as much as 60-80% of the Total Collision work, a percentage that decreased with net landing height.

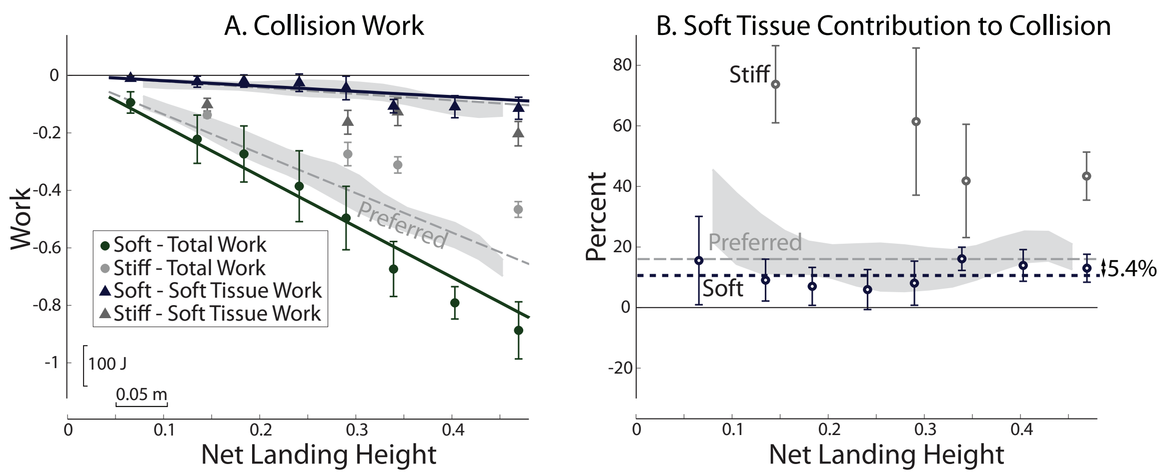

Figure 5.5. Collision work, contributions from passive soft tissues during Soft and Stiff landing. (A) Soft landings exhibited a significant increase in magnitude of Total work and a significant decrease in magnitude of Soft Tissue work during Collision, causing (B) a reduction in Soft Tissue contributions from the Preferred 16% to 10.6%. Stiff landings had the opposite effect, reducing Total Collision work and increasing Soft Tissue Collision work, with the overall effect of substantially increasing the proportion of Collision performed passively.

Following the initial region of negative Soft Tissue work after touchdown, we observed a region of positive Soft Tissue work (Figure 5.6). This was evident for all landing conditions and typically occurred during the Collision phase (when the body overall was performing negative work). In some cases it also continued into Recovery (when overall positive work was performed). The positive Soft Tissue work increased with Collision magnitude, and was typically about 20-30% of the magnitude of the negative Soft Tissue work (Figure 5.6).

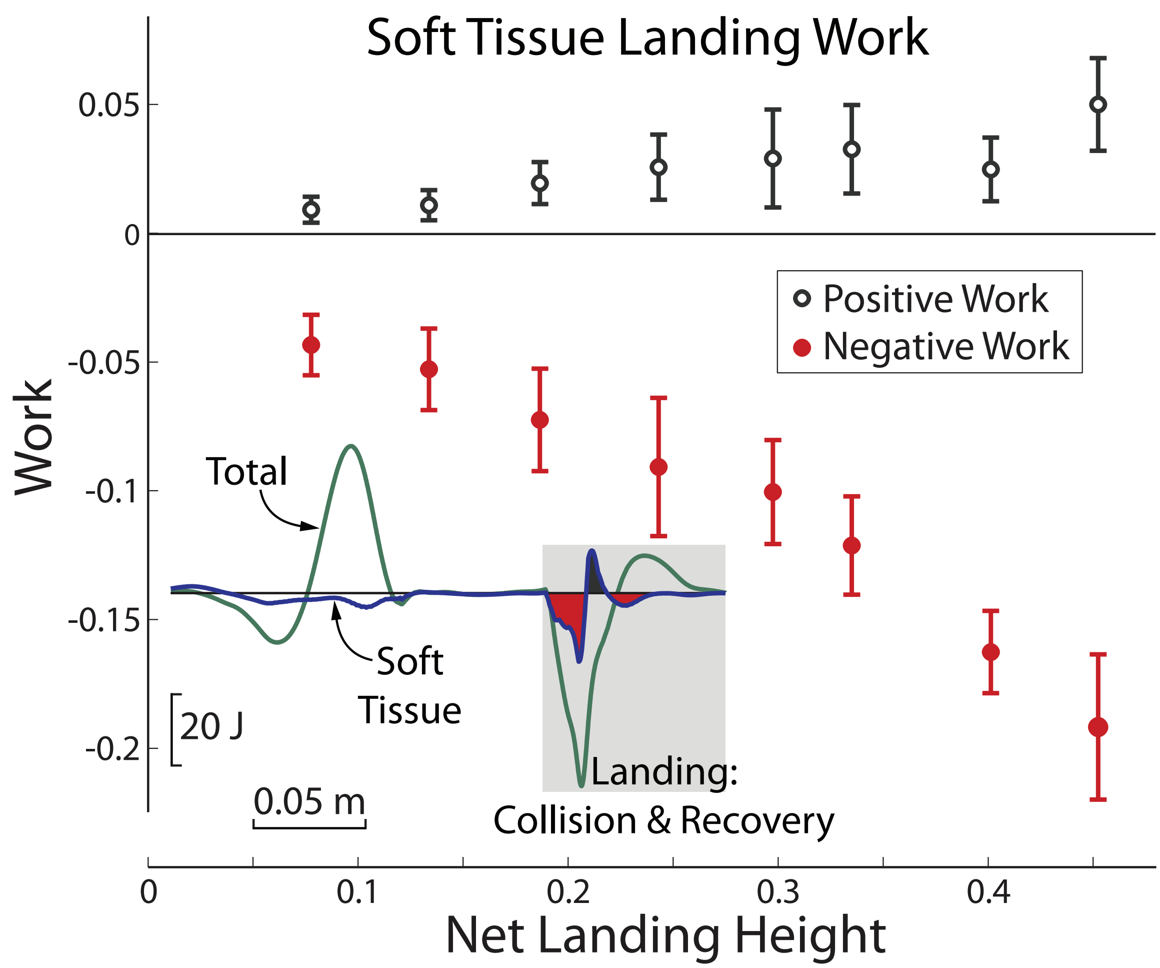

Figure 5.6. Soft Tissue work after touchdown in Preferred landing. Positive Soft Tissue work immediately following the negative Collision work suggests an elastic rebound of passive tissues. For all landing conditions, the magnitude of positive Soft Tissue work after landing was about 20-30% of the negative work.

As a methodological test of the quantification of Soft Tissue work, we also examined other phases of the jump and the control squatting trials. Push-off and Counter-Movement phases would be expected to be less impulsive and more active, and therefore should be well explained by the measured Joint work. Indeed, Soft Tissue work, accounted for only -2.3% of the Total work during Push-off (slopes -0.039 ± 0.010 and 1.674 ± 0.009 for Soft Tissue and Total, respectively). And Soft Tissue work (slope -0.034 ± 0.003) accounted for less than 6% of the Total Counter-Movement work (slope -0.588 ± 0.008) across net landing. Similarly, during the squatting trials, Joint work was within 6% of Total work. Positive work amounts were 0.249 ± 0.063 (Joint) vs. 0.263 ± 0.066 (Total), mean ± standard deviation, and negative work amounts were -0.254 ± 0.064 (Joint) vs. -0.270 ± 0.068 (Total). This was equivalent to averages of 177 J of Total positive work, 187.1 J of positive Joint work, -180.7 J of Total negative work, and -192.1 J of negative Joint work. We also tested for internal consistency of each of these work measures by summing positive and negative work over the full squatting trial. As expected, both measures summed close to zero, 0.005 ± 0.007 and 0.007 ± 0.005 for Total and Joint work, respectively (equivalent to 3.7 J and 5.0 J).

Discussion

Although humans seem to value economy of movement [142–148], they might prefer to perform extra muscle work to avoid other subjective costs, such as excessive soft tissue deformations during large Collisions that could cause pain or risk of injury. We found that subjects normally preferred a landing strategy that involved performing 37% more negative Collision work than necessary (Figure 5.2). This extra work then required an equal amount of positive Recovery work to return to standing rest. This consequence was highlighted by Soft landings that entailed 76% more negative work than necessary, followed by a similar amount of positive Recovery work. The preferred amount of extra work performed, and its associated energetic cost, may reflect how the subjects value the trade-off between economy and subjective costs such as pain. Humans appear to be willing to consistently exchange an extra amount of mechanical work to avoid costs associated with landing too stiffly. We propose that quantification of this work may thus serve as an indicator of a person's subjective valuation of that preference.

Subjects preferred a landing strategy that involved a combination of active and passive Collision work. The proportion of work performed passively by Soft Tissues was typically 16% for moderate to high jumps (Figure 5.4). This amount may seem modest, but is actually comparable to the contribution from any individual joint during landing [149–152]. Thus, the error from not accounting for Soft Tissue work is about the same as that from failing to include one of the leg joints. The proportion may even be greater for relatively small Collisions, where in some cases Soft Tissue contributions exceeded the combined contributions from all the joints. Landing Collisions from the lowest heights were similar in magnitude to those observed in the Collision following heelstrike in walking, and the relative passive contributions were in relatively good agreement with previous estimates of 60% for walking [153], [154]. The preference to perform work more passively during small collisions and more actively during larger Collisions is consistent with the proposed trade-off between economy and other costs, since we expect humans would be willing to exert more muscle effort to avoid costs associated with larger and potentially more damaging Collisions.

Subjects were able to modulate the distribution of work between active and passive tissues. When instructed to land softly, subjects performed a higher percentage of the Collision work actively, and when instructed to land stiff-legged the subject performed more passively, consistent with some preliminary quantifications [155]. Soft landing trials were subjectively reported to be more fatiguing than Preferred, while the subject who demonstrated Stiff landings reported substantial discomfort in his knees and lower back. During Soft landings, subjects performed on average 89.4% of the Collision work actively. We suspect that the proportion could be increased in actual practice, when humans may also use their arms and other joints to perform negative work, or perform more complicated land-and-roll maneuvers as in martial arts. However, since more than 80% of the body is comprised of “soft” (i.e., non-skeletal) tissues [133], there may be some practical limit to the maximum percentage of Collision work that can be done by active joint rotations when landing from a given height. Strength and flexibility may also be factors limiting an individual's ability to absorb Collision over an extended duration (Figure 5.11), as previously observed in comparing athletes vs. sedentary individuals in a drop landing task [161]. Alternatively, the Stiff landing condition demonstrated that humans are capable of performing most of the Collision work passively. In fact, landings could potentially be performed completely passively if a subject were to simply fall limply onto the ground, although that could be painful and would also require more positive Recovery work to return to standing rest. Of course, there are practical limitations to testing this empirically, which is why we did not test the hypothetical example of walking vs. falling down stairs. It nevertheless appears that humans can choose a wide variety of landing strategies, of which their preference is quite consistent and involves more work than necessary.

Soft tissues also appear to perform some positive work during a damped-elastic rebound after landing. We observed fluctuations in Soft Tissue power after touchdown, initially negative, then followed by a region of positive power. While passive tissues can only perform net negative work, elasticity of these tissues could allow them to store and return some energy, for example in the bouncing of viscera [112], [153], [162]. Independent of the landing condition, the energy returned by this damped-elastic rebound appeared to be about 20-30% (Figure 5.6). Similar evidence for a damped rebound of soft tissues has previously been observed in walking, with energy returns of about 10% at 1.25 m/s [153]. These are, however, very rough estimates because soft tissues could elastically rebound on a variety of time scales. The Soft Tissue measure only captures the net power resulting from the simultaneous deformations of many distributed soft tissues in the body. Nevertheless, this passive elastic rebound could affect the energetic economy or preferred frequency of cyclic movements, such as walking, running or hopping.

Our experimental estimates of mechanical work are subject to a number of limitations. We assumed that work done about the ankle, knee, hip and lumbosacral joints accounted for most of the active work of jump landing. It is also possible that substantial work is performed about other, unmodeled joints, or that some joint work is performed passively by series elastic elements, or that our mechanical work estimates are simply inaccurate. However, our methodological tests suggest that the Joint estimates are reasonably accurate since they yielded approximately zero net work as required for full-cycle squatting trials, and agreed well with Total work during the non-Collision phases of the jump cycle. We also found that the indirect estimate of Soft Tissue work during landing was supported by two other independent findings (Figure 5.7), which both suggested that the magnitude of negative work done by passive tissues increased with net landing height. One indicator was the imbalance between positive and negative Joint work over the entire jumping and landing task, which should sum to zero. The other was that the Joint work magnitude after touchdown was less than the potential energy change from peak aerial height. This agreed with the observed temporal aspects of Soft Tissue work, specifically that Joint work measures failed to capture substantial negative work after touchdown, work we attribute to passive tissue deformations. Collectively, these separate yet corroborating indicators suggest that passive tissues do indeed perform substantial work during jump landings.

A different limitation is that our estimates do not indicate where in the body the passive work is performed. Other techniques, perhaps using imaging or direct strain measurement (e.g., [139–141]), may provide more detail regarding passive dissipation and damped-elastic rebound. Another limitation is that unlike COM work, the Peripheral work estimates only capture "rigid" work, and not that due to passive tissue within a segment relative to that segment's COM. Errors from that assumption might be expected to cause an imbalance in Total work over a full jump cycle; however, Total work summed close to zero (Figure 5.7), providing some support for this assumption. A fourth limitation is related to the challenge of selecting filter parameters for force and motion data, which have been shown to affect joint moment impact peaks, typically within the first 30 ms of touchdown [163]. Therefore, we checked if our conclusions were sensitive to these parameters. Informal experimentation with different filter cut-off frequencies had little effect on the magnitude of negative Soft Tissue work estimates during Collision, although higher motion cut-off frequencies did tend to increase the subsequent burst of positive work, which we have speculated might represent a damped elastic rebound of passive tissues. Therefore, the estimated 20-30% efficiency of the damped passive rebound may actually be an underestimate. Finally, in this study we only consider the work performed by muscles, and not other contributors to energy expenditure such as force or rate of force production. Literature suggests that the muscle work is more metabolically costly than producing the same force isometrically [164], but there may nonetheless also be costs associated with producing force. These and other costs are best addressed through separate experiments designed to reveal their nature (e.g., [165], [166]).

We have alluded to subjective costs that might influence how humans prefer to land from a jump. Subject feedback suggested that pain-like factors might be one of the prevailing costs associated with preferred landing strategy, but with our current methods we have no way to extract the dominant factors from the pool of possible subjective costs. A variety of other subjective factors such as balance [167], safety, gracefulness, societal expectations, and even fun might also be relevant for motor tasks in general. If it were possible to hold all these other factors constant, previous findings [142–148] suggest that humans would choose to minimize active muscle work, and thus metabolic cost. Minimizing energy expenditure has been implicated as one factor influencing continuous, cyclic motions, such as walking [142–148], and while its relative importance may be diminished in some discrete movements, such as jump landings, it is still expected to be an influential factor since continuous motions could simply be thought of as a sequence of discrete actions.

We propose that mechanical work or energy may serve as a common currency for evaluating trade-offs that are otherwise difficult to quantify. If work were the only factor influencing preference, we would expect it to be minimized, but the degree to which it is not minimized in certain tasks may provide insight into the relative importance of other factors. This idea of work as a common currency seems to be consistent with other landing studies, which indicate that people choose to perform more work when landing on stiff surfaces than on compliant, and presumably more comfortable, surfaces [168–170]. We actually would prefer metabolic energy as a currency for its greater physiological relevance, but it is less amenable to measurement in discrete tasks such as considered here. Unlike force or kinematic measures, work and energy are both objective, scalar variables, regardless of how many and which joints are involved in a task. Work is part of most motor tasks, making it an appropriate currency for evaluating subjective factors that might not be as common to movement strategies in general. While this study provides an initial impetus for the use of mechanical work as a means of indirectly quantifying subject costs, further research is needed to validate work as a common currency. Future studies might explore, for instance, how the partitioning of work changes with fatigue or discomfort. Ultimately, by manipulating subjective factors that influence preference, it may be possible to alter the distribution of work and economy of movement. For instance, by changing characteristics of biomechanical aids (e.g., shoes), we may be able to alter the importance of other subjective factors (e.g., comfort) such that humans prefer to perform a task more or less economically. Mechanical work would serve as an indirect measure of the relative weighting of these other factors, and could therefore be a useful metric for selecting components or properties for such aids.

The preferred strategy for a motor task is presumably a balance between competing trade-offs. Some may be quantifiable like work, and others may be difficult to define, let alone quantify, like pain. Biomechanical measures usually focus on active motions performed by muscle, but it appears that passive deformation of soft tissues may also be important. The work they perform can affect what is needed from active muscle, as well as the pain that is felt by both active and passive tissues. It might seem obvious why humans prefer to walk rather than fall down a flight of stairs. But less obvious is that the work performed by doing so may also be an indirect valuation of the relative costs of falling or other alternatives.

Supplementary Material

Appendix

Detail regarding calculation of work measures is given here. We defined Soft Tissue power (P_{SoftT}) as the difference between Total mechanical power (P_{Tot}) of the body and inverse dynamics estimated Joint power (P_{Joint}),

$$P_{\mathrm{SoftT}} = P_{\mathrm{Tot}} - P_{Joint}$$

where Total power was defined as the sum of power due to motion of the COM (P_{COM}) plus power due to motion relative to the COM ($P_{\mathrm{Peripheral}}$), according to König's Theorem [171]:

$$P_{\mathrm{Tot}} = P_{\mathrm{COM}} + P_{\mathrm{Peripheral}}$$

COM power (or work rate) was defined as the dot produce of the ground reaction forces (F_{i}) with the velocity of the COM (v_{COM}). Forces were measured under each foot *via* force plates. The velocity of the COM was integrated from the acceleration of the COM (a_{COM}, equal to the sum of the forces divided by the subject mass, m), assuming zero average velocity over each jumping trial.

$$P_{\mathrm{COM}} = \sum_{legs} F_i \cdot v_{\mathrm{COM}}$$

$$v_{\mathrm{COM}} = \int a_{\mathrm{COM}} dt = \int \frac{\sum F_i}{m} dt$$

Peripheral power was estimated as the time derivative of the translational kinetic energy of each segment with respect to the COM plus the and derivative of segmental rotational kinetic energy. Segmental velocities (v_{s}) and angular velocities (w_{s}) were based on filtered marker motions, assuming rigid-bodies (with a minimum of four markers on each segment). Inertial properties (I_{s}, m_{s}) of each segment (foot, shank, thigh, pelvis, head-arms-trunk) were estimated based on anthropomorphic regression tables [172].

$$P_{Peripheral} = \sum_{segments} \frac{d}{dt}\left(\frac{1}{2} m_s (v_s - v_{COM})^2 + \frac{1}{2} I_s \omega_s^2\right)$$

This estimate fails to capture some of the motion of tissue within a segment relative to that segment's COM. This contrasts with COM work rate, which does include the contributions of soft tissues to overall COM motion. Our measure of Peripheral power therefore assumes that most of the work performed relative to the COM is quantified by the motion of rigid body segments. Methodological tests detailed in the discussion suggest that the error from this assumption is not large.

Standard 3-D inverse dynamics were used to estimate power (the dot product between joint moment, M_j, and joint angular velocity, w_j) about the ankle, knee, hip and lumbosacral joints, then summed to obtain net Joint power.

$$P_{Joint} = \sum_{joints} M_j \cdot \omega_j$$

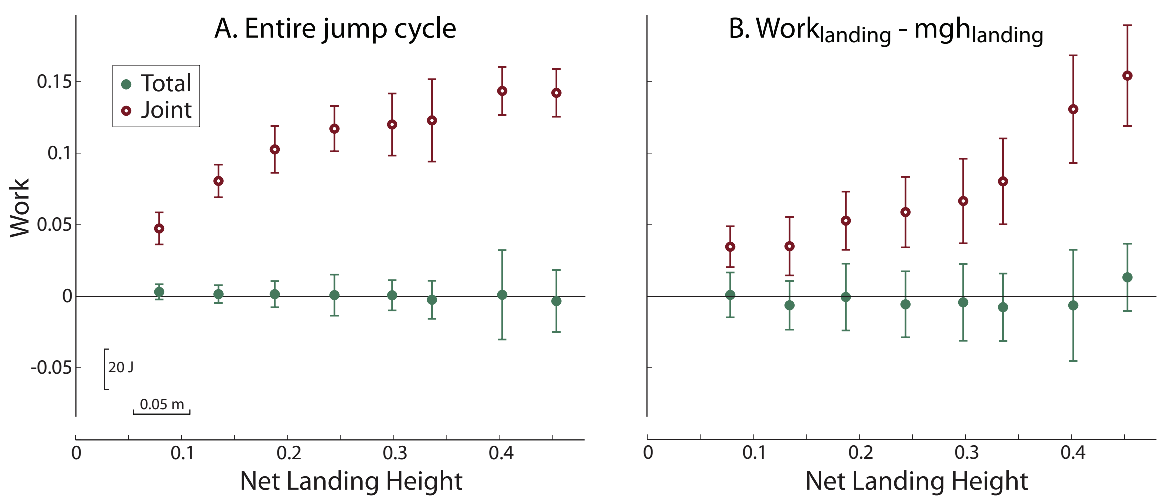

Figure 5.7. Additional indicators of Soft Tissue work. (A) Net work done over the entire jump-landing cycle should sum to zero. This was observed for Total mechanical work, but not for summed Joint work. (B) Net work done during landing minus the change in potential energy during aerial descent should also sum to zero. Similarly, Total mechanical work sums to zero, as expected, but Joint work does not. Both independent methods indicate that Joint work estimates fail to capture substantial work, increasing with net landing height.

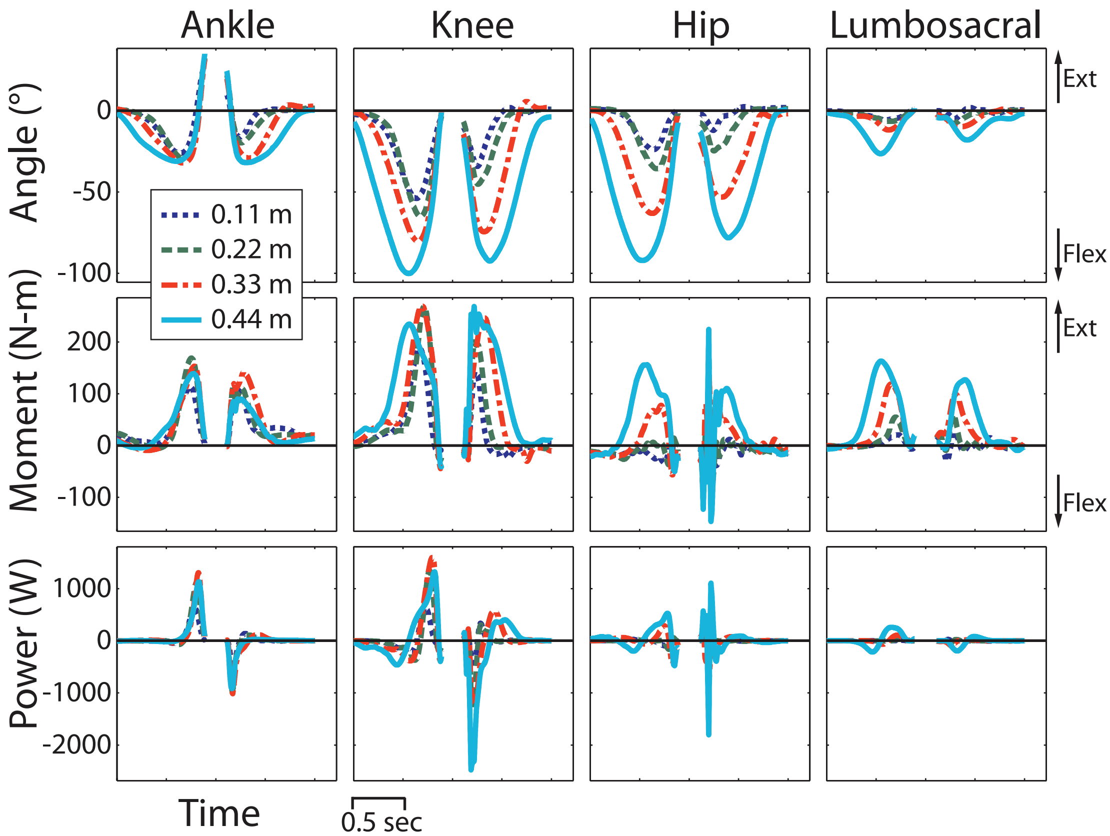

Figure 5.8. Joint angles, moments and powers for a range of net landing heights (representative trials from a single subject). Data are only plotted for periods of ground contact when the primary work was performed. Plots are aligned at times of take-off and touchdown, so that differing time durations of aerial phases are not shown; a typical duration of 200 ms is shown for reference.

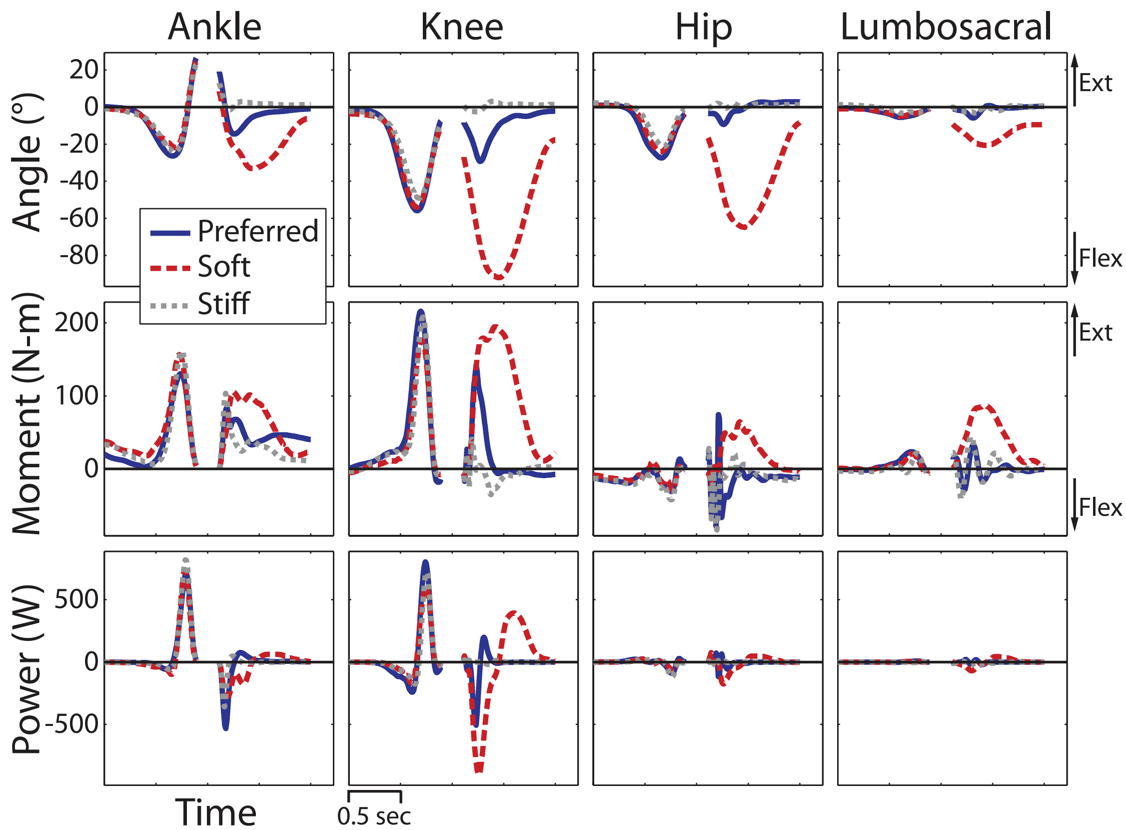

Figure 5.9. Joint angles, moments and powers for Preferred, Soft and Stiff landings conditions with net landing height of 0.12 m (representative trials from a single subject). Data are only plotted for periods of ground contact when the primary work was performed. Plots are aligned at times of take-off and touchdown, so that differing time durations of aerial phases are not shown; a typical duration of 200 ms is shown for reference.

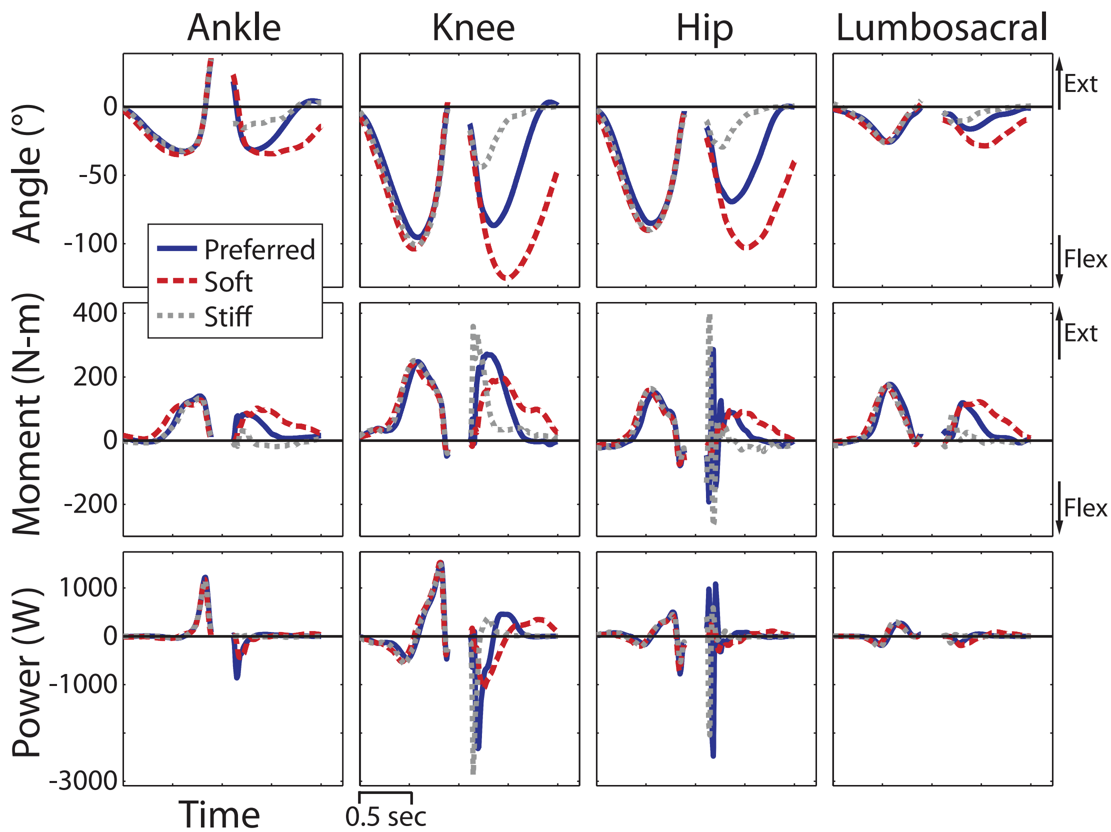

Figure 5.10. Joint angles, moments and powers for Preferred, Soft and Stiff landings conditions with net landing height of 0.46 m (representative trials from a single subject). Data are only plotted for periods of ground contact when the primary work was performed. Plots are aligned at times of take-off and touchdown, so that differing time durations of aerial phases are not shown; a typical duration of 200 ms is shown for reference.

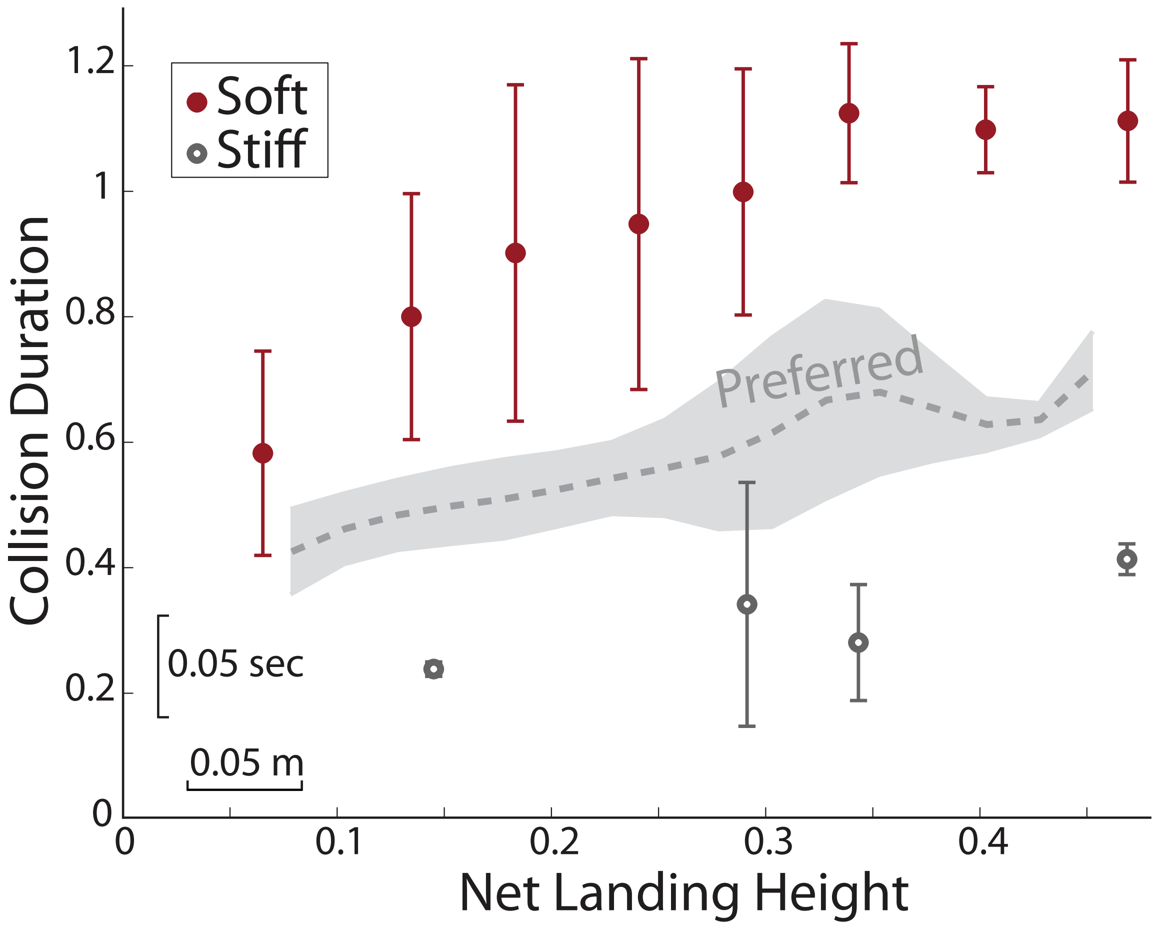

Figure 5.11. Time duration of Collision phase. Soft landings, which were associated with increased active muscle work, exhibited extended Collision durations ($P < 0.001$). Stiff landings, which were associated with increased passive soft tissue work, exhibited shortened Collision duratio

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