NEUROMECHANICS AND AUGMENTATION OF MUSCLE-TENDON ACTUATORS IN UNSTEADY CYCLIC TASKS

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“It is not the critic who counts; not the man who points out how the strong man stumbles, or where the doer of deeds could have done them better. The credit belongs to the man who is actually in the arena, whose face is marred by dust and sweat and blood; who strives valiantly; who errs, who comes short again and again... who at the best knows in the end the triumph of high achievement, and who at the worst, if he fails, at least fails while daring greatly.”

Teddy Roosevelt

For my late father J Punith Kumar

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SUMMARY

Legged animals traverse complex terrestrial environments with greater economy, stability, and agility than their legged robotic counterparts despite having larger (neural) delays in their controllers and larger series elasticity in their actuators – conditions that arguably make generating forces on the ground less controllable when compared to legged robots. Our central hypothesis is that achieve this partly by shifting computation from their nervous systems to the mechanical properties of their actuators.

To investigate the hypothesis stated above, we ask three central questions :

1. CH 1: Assuming you had zero control of your nervous system when faced with a perturbation, can biological actuators maintain stability by themselves?
2. CH 2: If a particular morphology is self-stable, does it mean it lacks agility? And how does it relate to metabolic cost? Or to state it more clearly, How do stability, agility, and economy tradeoff for a particular actuator morphology?
3. CH 3: How does modifying morphology with a parallel elastic exoskeleton affect stability, agility, and economy?

To answer these questions, typical animal biomechanics experiments have critical drawbacks. Animals in in-vivo experiments have multiple joints and multiple muscles in each joint, and thus render directly linking individual actuator's mechanics to whole body behavior impossible. In addition, the underlying neural control of the muscles and mechanical properties of the body and environment are out of the experimenter's control to test hypotheses about how this self-correcting behavior changes as a function of them. While in-vitro workloop experiments mitigate the problems mentioned above, the trajectory of the biological is pre-determined – and not a function of interaction with the physics of the environment. Therefore, there is a critical need to develop a framework that marries the strengths of in-vivo biomechanics experiments – namely, unconstrained physically realistic

movements in complex terrain – with the strengths of in-vitro biomechanics experiments – namely, easy input output measurements and experimental control of neural stimulation and environmental properties (like morphology and terrain). Thus, to answer the questions above we developed a framework where a single muscle-tendon unit is interacting with a mass in gravity through a lever arm in closed loop to generate cyclic movement in variable terrain (both in simulation and in-vitro closed-loop experiments), variable morphology (in simulation) and variable nervous system control (in simulation).

Through our work, we show :

1. CH 1: Biological actuators can maintain stability by themselves despite zero control of your nervous system when faced with a perturbation.
2. CH 2: Series elastic springs influence stability, agility, and economy of movement such that animals need to pick one to two of the three
3. CH 3: Parallel elastic actuators are able to augment stability, agility (in terms of positive power) and economy of movement. They do however reduce the capacity of negative power generation at high exoskeleton stiffness values.

Payoff : Through our research, : (1) we gain fundamental neuromechanical understanding of how animals enable stable movement in uneven terrain and (2) we generate a template for the design of a new generation of bioinspired robotic actuators to enable legged and wearable robots to navigate the world as it is built.

CHAPTER 1

STABILITY OF ISOLATED MUSCLE-TENDONS AGAINST UNEXPECTED CHANGES IN GROUND HEIGHT

1.1 Abstract

Legged animals navigate complex and unpredictable terrestrial environments with incredible stability. They do so despite large neural and inertial delays that render high-bandwidth feedback control of their kinetics and kinematics infeasible. In fact, running, and hopping animals stably navigate unexpected changes in ground height without significant changes to the neural activation of their muscles. Thus, we ask in the extreme case, is it possible for muscle-tendon units to reject a broad range of terrain perturbations without any change to neural control? To answer this question, we emulated ankle-based hopping on a movable ground using a novel biorobotic interface. We attached the plantaris longus muscle-tendon unit of 5 American Bullfrogs to a servo motor programmed to behave like a mass in gravity through a lever arm. We cyclically stimulated the muscle-tendon unit to generate hopping, and once the system reached steady state, we changed the height of the virtual ground and measured the amount of work done on the mass in the 4 steps after the perturbation and compare it to an 'ideal actuator' that perfectly dissipates the applied perturbation and a Hill-type model of the muscle. We show that muscle-tendon units dissipate 95% of energy injected by the perturbation in the first hop, and after 3 hops dissipate all the energy in the system. Furthermore, we show that Hill-type models underestimate the force/work done by the real muscle-tendon unit – thus suggesting that non-quasistatic muscle properties significantly contribute to the rapid stabilization observed above. Future studies should investigate what mechanical and cellular property of muscle-tendon units enable this behaviour and what the role and effect of reflex feedback is on top of this be-

haviour. Leveraging the results of our study and suggested future studies, roboticists of the future may be able to build legged systems with cheap, robust low bandwidth sensors and controllers by embedding self-stabilizing properties in the mechanics of their actuators.

Keywords : Muscle, Tendon, Muscle-Tendon Unit, Stability, Preflex, Uneven Terrain

1.2 Background

1.2.1 Legged animals maintain stability in uneven terrain despite motor control constraints

Legged animals traverse unpredictable terrain environments with incredible stability [1, 2, 3, 4, 5]. They do so by (a) setting the initial angle of their limbs with the ground in a narrow range when they are in the aerial phase [6, 7] and (b) rapidly modulating the impedance of their limbs in a narrow range [6, 7, 4] in the stance phase by modulating the impedance of their distal joints and actuators [8, 5]. However, unlike legged robots, they achieve this feat despite large neuromechanical constraints in their controllers and actuators that make regulating limb/joint function difficult.

For starters, animals have neural [9] and inertial delays [10] that are on the order of the time constants of their gait. This means, unlike legged robots, animals are unable to regulate joint function in a high-bandwidth closed loop manner. In addition, their actuators (i.e. muscles) function in a highly context dependent manner - where the forces they produce for the same activation can be different if their instantaneous [11] and historical [12] lengths and velocities are/were different. This is further complicated in the distal joints, where muscle function is buffered by highly compliant in-series tendons [13] that affect muscle state in a load-dependent manner. So the question arises, how do animals rapidly modulate joint/limb function despite these aforementioned constraints?

Surprisingly, when we look at the muscle level, both guinea fowl [14] and cockroaches [15] activate their distal muscles with the same timing and magnitude while navigating uneven terrain as compared to level ground - thus suggesting the adaptive energy dissipation/generation is happening automatically due to underlying context dependent changes in muscle function. While human runners [16] and hoppers [17] do modulate the magnitude of their muscle activation when their limb is in contact with the ground, the percentage increase in their stimulation is relatively smaller than the percentage change in joint kinetics - thus suggesting humans also partly rely on underlying actuator dynamics to stabilize themselves.

1.2.2 Can muscle-tendon units maintain stability in uneven terrain without neural control?

Thus, we're interested in the question : Can muscle-tendon units generate stable movement in uneven terrain with a purely feedforward neural control? Isolated muscle-tendon work-loop experiments, while not done in the context of a mass in gravity, have shown that the work produced by the muscle-tendon unit is a function of the timing of activation of the muscle-tendon unit relative to the timing of stretch [18]. Interestingly, the magnitude and sign of the work done switches based on whether the relative timing of the activation increases or decreases. This is a promising result for our question since in feedforward driven movement in uneven terrain environments, even though the timing of the stimulation does not change in absolute time (due to lack of change in neural control), the relative timing of muscle stimulation to muscle-tendon stretch at ground contact is increased when the height of the ground is dropped and decreased when the height of the ground is raised – thus leading to automatic and context-dependent phase shift in the stimulation of the muscle-tendon unit that could lead to adaptive dissipation of the perturbation. However, since the experiments above had a fixed length trajectory of the muscle, it can only say what the work done by the muscle-tendon unit is under various stimulation timing would be for fixed length trajectories and not the question we are interested in - which is can muscle-tendon units functionally stabilize a mass hopping on uneven terrain? This requires the muscle-tendon unit to be dynamically coupled with a mass, and the emergent mechanics of the two to be allowed to act.

One way to do so is mathematical modeling. Previous modelling studies where a mass in gravity was actuated by a Hill-type muscle with no series elasticity to generate hopping on a movable ground [19]. They showed that muscles were stable in cyclic hopping, and that this intrinsic stability of muscles was determined by non-linear shape of the muscle's force-velocity property. However, (a) Hill-type models mainly capture quasi-static muscle behavior and are particularly bad at capturing dissipative tasks and therefore may not fully capture the behavior of real muscle-tendon units in dynamic tasks like navigating uneven terrain, (b) the model lacked a series elastic element whereas distal joints in animals have large series elasticity that buffer muscle action and (c) the stimulation pattern was phase-locked to time of the mass reaching apex height, thus requiring neural feedback and changing neural control timing when height of the mass changes in subsequent steps after the perturbation. Thus, in this study, we ask: Can isolated 'real' muscle-tendon units cyclically actuating a mass in gravity to generate hopping reject a broad range of terrain perturbations without 'any' change in neural control?.

To test this question, we emulated hopping with a single cyclically stimulated muscle-tendon unit using a biorotobtic interface that emulated a mass in gravity through a lever arm. We cyclically stimulated the muscle-tendon unit at 2Hz to generate steady state hopping, and once the model achieved stable hopping, we perturbed the height of the ground while the mass is in the air to simulate uneven terrain. We investigated the energy generated/dissipated in each hop and how many hops the mass takes to recover from the perturbation, in addition to how safely it recovers from it. We hypothesize that real muscle-tendon units can reject a broad range of perturbations without a change in neural control. In particular, we predict that the energy dissipated in the first hop is proportional to the energy injected. Secondly, we predict that the energy dissipated after 4 consecutive hops is statistically equal to the energy injected.

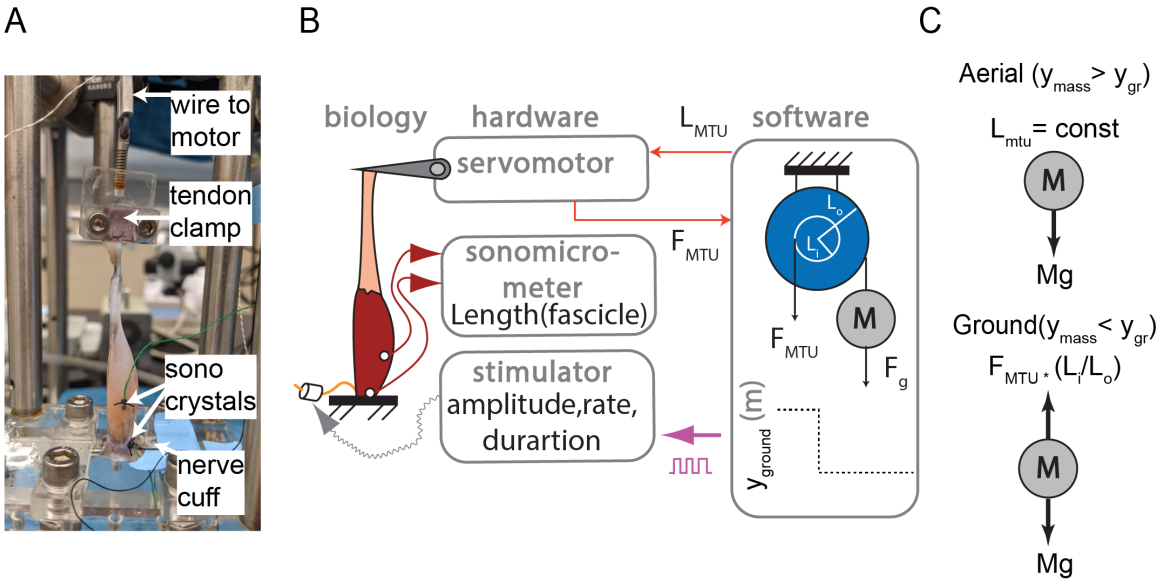

Figure 1.1: A) Plantaris Longus muscle-tendon unit of an American bullfrog instrumented with sonomicrometry crystals, nerve cuff and tendon clamp. (B) Schematic of physics emulator attached to muscle-tendon unit. (C) Definitions of aerial and ground phase. In aerial phase, the length of the muscle-tendon unit is kept constant and only gravity acts on the mass. In stance phase, the length of the MTU is proportional to the change in mass position and the both the force of the MTU and gravity apply to the mass.

1.3 Methods

1.3.1 Muscle-Tendon Unit Extraction

All experiments shown here were approved by the Georgia Institute of Technology Institutional Animal Care and Use Committee. The frog was anesthetized with vaporized isoflurane (5% at 0.5L/min flow rate) for approximately 30 minutes to achieve an adequate plane of anesthesia . In particular, under adequate anesthesia, the frog does not exhibit a righting reflex when laid in a supine position. Other indications may be loss of blinking when eye is pupated or depression of buccal breathing. Once anesthetized, frogs were euthanized with a guillotine followed by adjunctive pithing. To pith the frogs, a metal probe was used to disrupt the brain anteriorly and the hindbrain and spinal cord posteriorly. Once the frog is euthanized, we isolated the plantaris longus muscle attached to the knee joint, its tendon and the sciatic nerve from surrounding tissue and placed it in an aerated bath of amphibian Ringers solution.

1.3.2 Muscle-Tendon Unit Setup and Instrumentation

The muscles were instrumented with two sonomicrometry crystals (1mm, Sonometrics) implanted along a proximal muscle fascicle. A bipolar stimulating electrode cuff was placed around the sciatic nerve and connected to an Aurora 701C Stimulator (Aurora Scientific). A metal rod was inserted through the knee joint and clamped to a Plexiglass plate to fix the muscle-tendon unit at one end. On the other end, the tendon was clamped to a custom friction-based tendon clamp that is attached to a wire with a hook. The hook of the wire was attached to the end of the servo motor's arm. Thus, when the servo motor moves, it stretched or shortened the muscle-tendon unit based on the direction of movement.

1.3.3 Determining Muscle Properties

To determine the stimulation voltage, we twitched the muscle at increasing voltages till the peak twitch force is reached. Next, keeping various passive tensions on the muscle-tendon unit, we stimulated the muscle at an action potential frequency of 100Hz for 300ms and recorded the initial and steady-state force the muscle-tendon unit reaches for each condition. We fit an exponential to the initial force-length curve to get the passive force-length relationship. By subtracting the passive forces at the final muscle lengths from the final force values and fitting a second order relationship to the resulting points, we got the active force length relationship. The force at peak force is determined to be Fmax and the length of the muscle-fascicle at that point is the optimal length of the muscle Lopt. We then assume that the maximum shortening velocity of the muscle is a fixed multiple of the optimal muscle length Lopt. We then fit a Hill-type model to the trials above to estimate the activation and deactivation time constants of the muscle (Tact and Tdeact).

1.3.4 Biorobotic Interface

We developed a biorobotic interface [20, 21] that emulates hopping using only one joint (i.e., the ankle). The biorobotic interface consists of an Aurora Scientific 310C-LR Motor and a dSpace Microlabbox Real-Time Controller. At each time point, dSpace multiplies the muscle-tendon unit force (Fmtu) by the effective mechanincal advantage (EMA) (Lin/Lout) to get the force experienced by the mass (Fmass). This in turn moves a simulated mass in gravity. The dspace model consists of a human body that is modelled as a lumped point mass in a gravitational field. The mass is modelled as a hybrid dynamical system with two states: (1) aerial and (2) stance. In the aerial phase, the body is in the air and has no contact with the ground. Thus, the mass only force experienced by the mass is the force of gravity. When the body is in contact with the ground, it is in the stance phase. During the stance phase, the mass is actuated by both the gravitational field and muscle-tendon unit force. The position of the mass where the body loses contact with the ground and the state transitions to aerial from stance is designated as $y = 0$.

$$\ddot{y} = -g + \frac{F_{mass}}{m} \qquad \text{(stance)} \qquad (1.1)$$

$$\ddot{y} = -g \qquad \text{(aerial)} \qquad (1.2)$$

dSpace then converts the change in mass position (y) to change in MTU length (by multiplying it with EMA and reversing the sign) and sends it as an input to the servo motor and the motor stretches or shortens the MTU accordingly.

1.3.5 Determining Cycle Time

We aimed to stimulate the muscle-tendon unit at the resonant frequency of the passive muscle-tendon unit [20]. First, we allowed the inactive muscle-tendon unit to freely oscillate against the mass. We used this data to estimate the linearized stiffness of the muscle-tendon unit at 0.3*Fmax. Given the stiffness, we estimate the mass and effective mechanical advantage (EMA) of the system by satisfying the following constraints:

$$f = \frac{\text{EMA}}{2\pi}\sqrt{\frac{k}{M}} \tag{1.3}$$

$$\frac{(M * g)}{\text{EMA}} < 0.3 * F_{max} \tag{1.4}$$

1.3.6 Experimental Protocol

Once we determined the mass and effective mechanical advantage of the system, we then proceeded to the biology-in-the-loop experiments. We stimulated the muscle-tendon unit at 100 Hz using an Aurora Scientific Stimulator. The stimulation pattern consists of a 2Hz square wave with a 10% duty cycle. This generated force from the muscle-tendon unit and moved the emulated mass and created a hopping pattern. After 5-7 hops, we suddenly dropped/raised the height of the ground (i.e. change the reference of ground position from y=0 to $y = H_{perturb}$ and observed the emergent muscle-tendon mechanics for 5 more hops and calculate the following outcome measures of interest.

1.3.7 Measurements

1. Energy Change (per step): We measure this by measuring the difference between the total energy of the system in the previous step and the current step. We do this by measuring the total energy when the mass is in peak height in the air and kinetic energy is zero. So, the energy change in a particular hop ‘i’ is calculated as:

$$\Delta E_i = mg(y_i - y_{i-1}) \tag{1.5}$$

2. Cumulative Energy Change : We measure this by cumulatively summing the energy change described above.

$$E_i^N = \sum_{i=1}^{N} E_i \tag{1.6}$$

3. Settling time: Defined as the hop at which the slope of the line that is fit to cumulative work given input energy is statistically similar to 1.

$$T_{settling} = min(N) \text{ where slope}(E_i^N \; vs \; mgH_{perturb}) = 1 \tag{1.7}$$

1.3.8 Normalization

To give context to the numbers and make them interpretable/relatable, we normalized the numbers in our results to those of a representative spring-mass system that resonantly hops at the frequency of 2Hz. We thus normalized time by $T_0 = 0.5s$, mass by M_0, and acceleration by $A_0 = 9.8m/s$. Thus, we normalized forces by $F_0 = A_0M_0$, stiffness by $K_0 = M_0W_0^2$ where $W_0 = 2\pi/T_0$ is the resonant frequency of the system. Subsequently, we normalized lengths by $L_0 = F_0/K_0$ which represent the length that the representative spring compresses when loaded by the mass and $V_0 = L_0W_0$. We also normalized energy by $E_0 = 0.5K_0L_0^2$ which is the energy stored in the spring when loaded by the mass and power by $P_0 = 0.5F_0V_0$.

1.4 Results

In this paper, we investigated whether real muscle-tendon units can reject a broad range of perturbations without a change in neural control. To test this question, we emulated hopping with a single cyclically stimulated muscle-tendon unit using a biorotobtic interface that emulated a mass in gravity through a lever arm. We cyclically stimulated the muscle-tendon unit at 2Hz to generate steady state hopping, and once the model achieved stable hopping, we perturbed the height of the ground while the mass is in the air to simulate uneven terrain. We investigated the energy generated/dissipated in each hop and how many hops the mass takes to recover from the perturbation in addition to how safely it recovers from it.

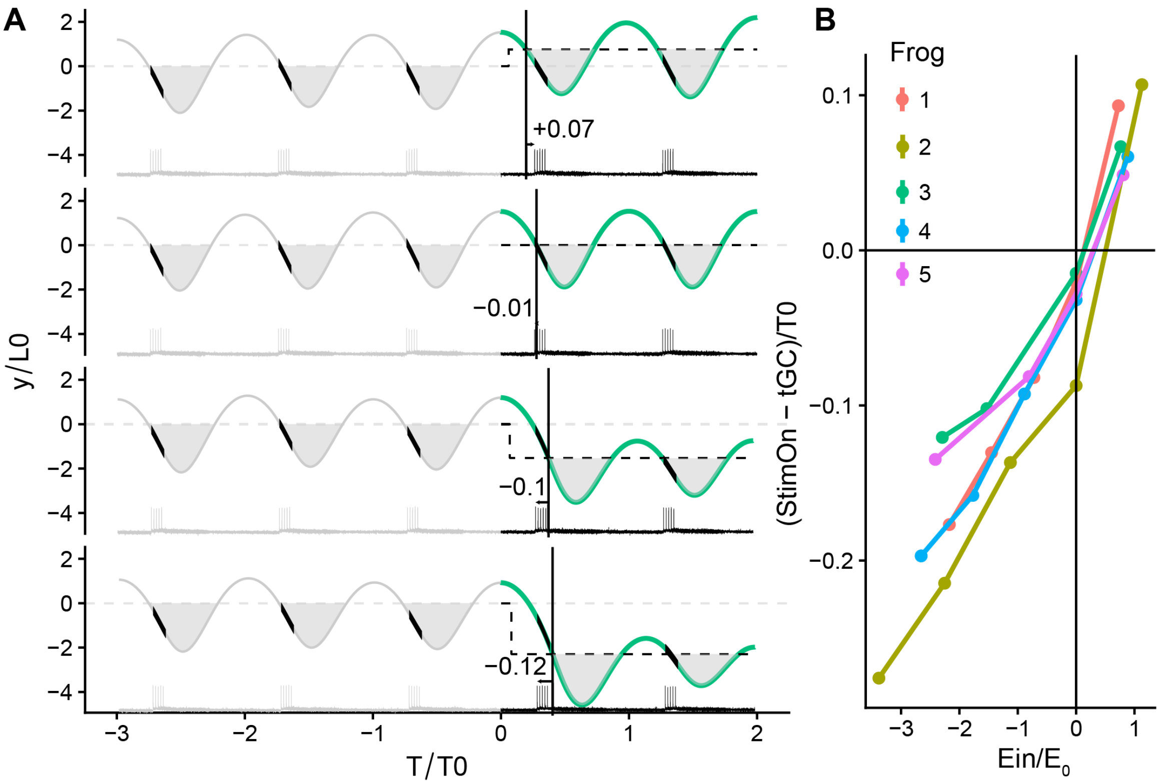

Figure 1.2: A) Position of the mass and nerve stimulation across time for different heights of a drop from one experiment. Note that stimulation is sent at a constant frequency, and in steady state results in constant phase with respect to mass position. When the height of the ground is changed, however, the time of stimulation with respect to ground contact is shifted earlier for a drop in ground height and later for raised ground height (B) Phase of stimulation with respect to ground contact time for all trials. Note how phase of stimulation is earlier (more negative) for drops in ground height and later (more positive) for raised ground heights

1.4.1 Cyclically Stimulated Muscle-Tendon Units Generate Hopping

In Figure 1.2, we see example data from one experimental session. We can see that when the muscle-tendon unit is stimulated with a constant frequency, steady state hopping is achieved. During steady state hopping, the mass cyclically oscillates between its maximum height in the aerial phase and its minimum height in the stance phase. In the aerial phase, the mass follows a parabolic path and gravity is the only force being exerted on the mass. Therefore, the mass maintains constant total energy by exchanging its kinetic and potential energy. During the stance phase, however, the muscle-tendon unit also produces force on

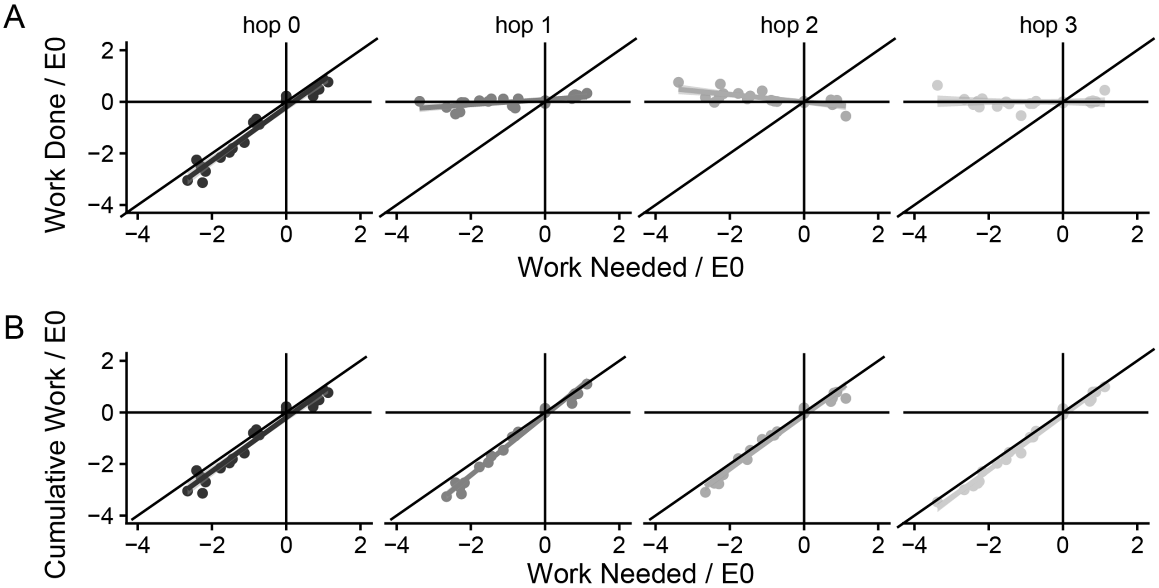

Figure 1.3: A) Work done on the mass during the first, second, third and fourth hop after the perturbation for each drop height. Note the X axis is the energy injected for each drop height and the Y axis is the work done on the mass. So the ideal scenario is shown by the diagonal line, where the work done is exactly equal and opposite to the energy injected. (B) Cumulative work done on the mass done after the first, second, third and fourth hop after the perturbation for each drop height. See table for R-squared, slopes and p-values.

the mass. Therefore, the muscle-tendon unit initially does negative work on the mass as the direction of ground reaction force generated by the muscle-tendon unit is opposite the direction of movement of the mass till the minimum height is reached. This is followed by positive work done by the muscle-tendon unit on the mass as the force and direction of movement is aligned. In a steady state cycle, the total work done on the mass is zero, and therefore it maintains the same apex height.

1.4.2 Muscle-Tendon Units Generate/Dissipate Energy After Change in Ground Height

When the height of the ground is changed, work is done on the mass by gravity and therefore the velocity when the mass reaches touchdown is higher/lower when the height of the ground drops/increases. When the mass is in the stance phase after the perturbation, the muscle-tendon unit does work on the mass and the relative difference between the work done by the muscle-tendon unit and gravity determines how close the system is to steady state (Figure 1.3). During the first hop after the perturbation (i=0), we see that the energy generated/dissipated by the MTU is directly proportional to the energy needed to be generated/dissipated (i.e. the slope of 1.118 is significantly different from 0 p=7.53e-15 and the best fit line has an R2 of 0.9467). In addition, we fail to reject the hypothesis that the line of best fit is statistically different from a line with slope 1 (p=0.054), but the intercept is statistically different from 0. Over the next hop (i=1), we see that further energy is dissipated/added and slope of cumulative work deviates further to 1.21 and is statistically different from a slope of 0(p=2e-16) and 1 (p=2e-5). In the next hop (i=2), the system returns to steady state where the slope is significantly different from 0, but statistically similar to 1. In addition, the intercept at hop (i=2) is also statistically similar to 0.

1.4.3 Phase Of Stimulation Shifts When Height Of Ground Changed

To explain how muscle-tendon units generate/dissipate work in this context, we look at the underyling neuromechanics of the task. While the absolute timing of stimulation is kept constant, the phase of stimulation changes when the height of the ground is changed. This is due to the timing of ground contact being earlier/later when the height of the ground is made higher/lower (Figure 1.2). Thus, the phase of stimulation is made more negative with height drops and more positive by height increases.

1.4.4 Change Of Phase Leads To Work Done

When the height of the ground is raised, the time of stimulation is later than steady state.Figure 1.4. Thus, the muscle-tendon unit is passively stretched, followed by activation of the muscle, thus leading to shortening of the muscle and lengthening of the tendon followed by muscle-lengthening - i.e. a stretch shorten cycle. This, in conjunction with the work added by the motor, leads to overall positive work. When the height of the ground is dropped, the muscle-tendon unit is activated earlier than ground contact. This leads to internal shortening of the muscle against the tendon in the aerial phase, followed by rapid absorption of

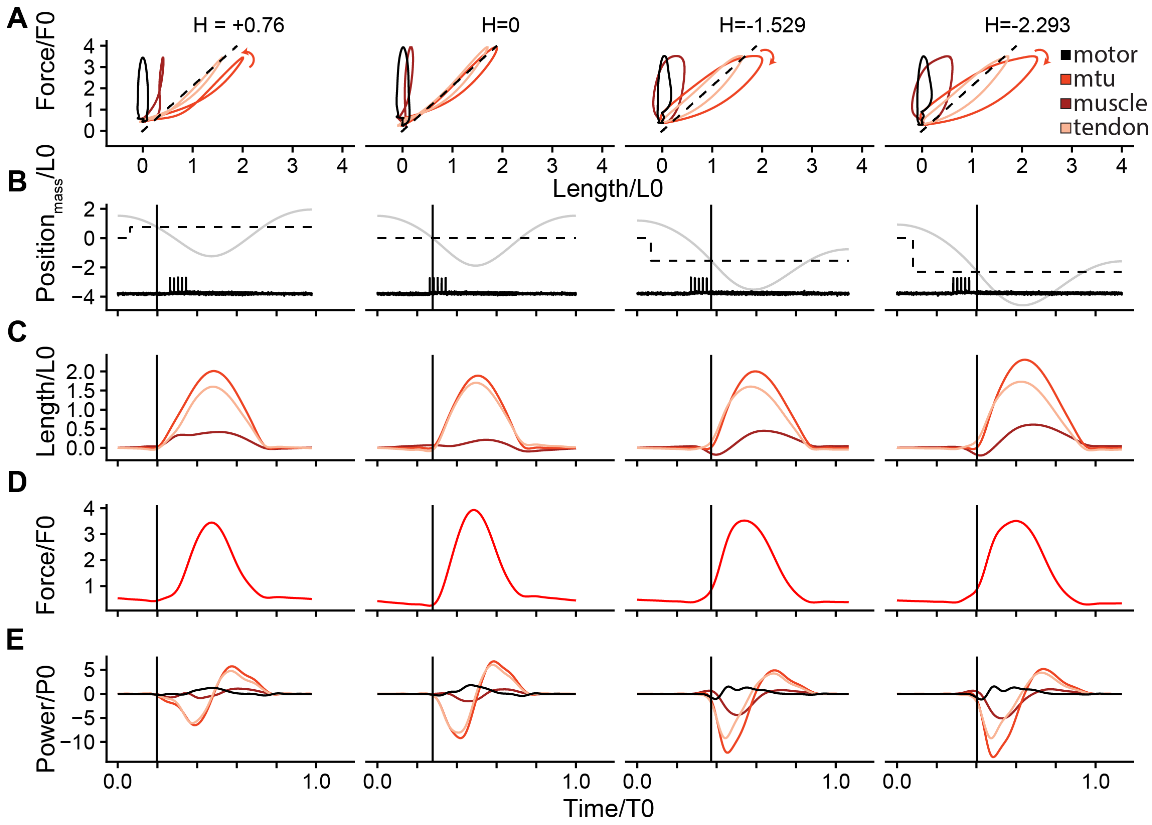

Figure 1.4: A representative dataset from a single preparation showing A) workloops (B) mass position, ground height and stimulation (C) length (D) forces (E) powers for the motor (black), muscle-tendon unit (red), muscle (maroon) and tendon (pink) at ground height changes of +0.75, 0, -1.529 and -2.293 L0. Note how change in ground height causes change in phase of stimulation with respect to the ground, which leads to change in direction of workloop from a passive stretch-active shorten positive workloop for ground height raise and active shorten-passive stretch negative workloop for ground height drops. Also note how tendon buffers muscle powers and lengths.

negative work by the tendon followed by relengthening of the muscle - i.e. a shorten-stretch cycle. This, in conjunction with negative work by the tendon, leads to negative work by the muscle-tendon unit.

1.4.5 Hill Models Underestimate Force And Negative Power

In Figure 1.5, we can see time series data of the real muscle versus its Hill-type model for one frog. In D, We can see that for this experiment, Hill model forces initially match real muscle tendon forces close and as the muscle begins stretching, we see that the forces

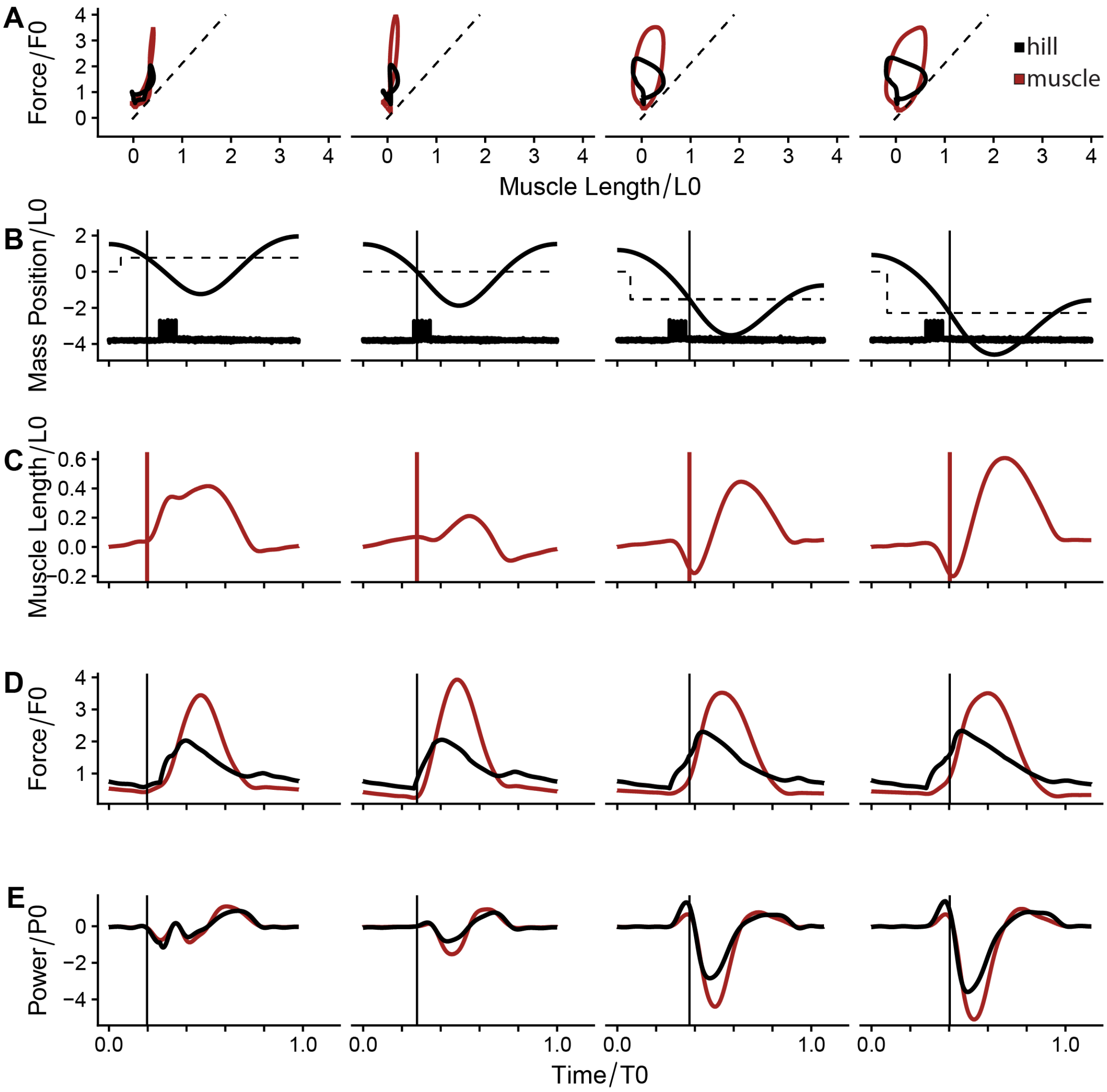

Figure 1.5: A representative dataset from a single preparation showing A) workloops (B) mass position, ground height and stimulation (C) length (D) forces (E) powers for the muscle (maroon) and hill model (black) at ground height changes of +0.75, 0, -1.529 and -2.293 L0. Note how muscle forces are larger than hill model forces. Also note how hill model powers deviate from real muscles, mainly during active stretch.

deviate. This is shown even more clearly in E, where the positive powers closely match each other, however the negative powers for the Hill-model are lower. In Figure 1.6, we show the above for all trials. We see that the hill model predictions of force are proportional to actual muscle force. However, they are lower in magnitude. Similarly, we see that hill model predictions of muscle work are proportional to actual muscle work, however are

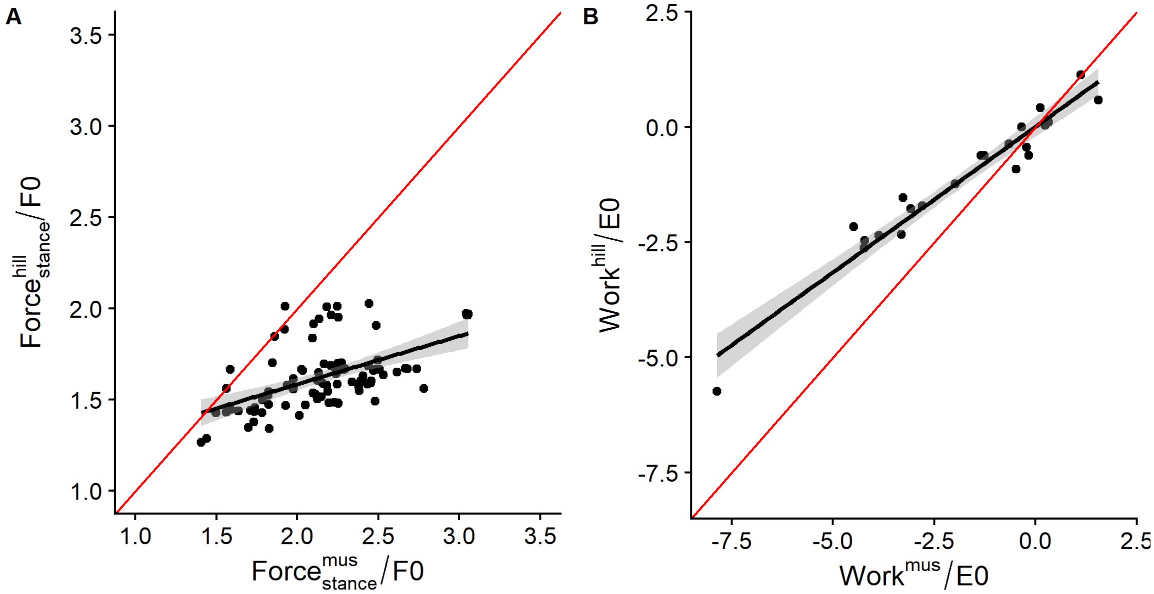

Figure 1.6: (A) Force produced by hill model compared to force produced by real muscle. (B) Work produced by hill model compared to force produced by real muscle. Note how the hill model predicts significantly lower forces and works than real muscles.

lower in magnitude.

1.5 Discussion

In this paper, we were interested in the broad question : Can muscle-tendon units reject a broad range of perturbations without change in neural control? In particular, we hypothesized that muscle-tendon units would dissipate/generate energy proportional and opposite to the energy generated/dissipated by a perturbation in ground height during a hopping task. We also hypothesized that the muscle-tendon unit would dissipate all the energy within 5 hops. In this study, we were able to confirm both hypotheses. In fact, we show that the energy generated by the muscle is proportional to the energy injected by the perturbation during the first hop after the perturbation with a slope that is statistically similar to 1, however has an intercept statistically different from 0. This shows that muscle-tendon units adaptively dissipate perturbations even without neural control, even in one cycle. In addition, we show that the energy dissipated in the third hop after the perturbation has a slope and intercept that is statistically similar to 1, thus showing that muscle-tendon units are able to dissipate perturbations over multiple hops completely without change in neural control.

While our results are the first demonstration of muscle-tendon units 'generating' stable locomotion in uneven terrain without change in neural control, this hypothesis is supported by animal experiments where neural control was statistically similar during a perturbation while the animal changed the work their limbs did to dissipate the perturbation [3, 14]. However, the studies above had multiple limbs, joints and muscles - thus leading us to not be able to directly link a muscle-tendon units' action to the center of mass. In addition, even though the neural stimulation pattern had similar statistics during perturbations and steady state, that does not necessarily mean that the trajectory/pattern of stimulation is the same. This is especially important because even if you have two stimulation patterns with the same firing rate, if the timing of one of the action potential varies even in the order of milliseconds, we can get significantly different forces and work [22]. We were able to sidestep these problems by directly controlling the stimulation pattern to be identical for each step, and by having the dynamics of the muscle-tendon unit directly actuate the center of mass of the body without any other contributor.

1.5.1 Role Of Contractile Properties In Stability

In addition, we ask what the role of contractile mechanics is on the behavior we observe. We show that Hill-type models predict lower forces and negative work during perturbed trials, thus showing the importance of non-steady state muscle properties in unsteady movement. In fact, previous studies have shown that even if muscles follow the same Hill-type model states, they produce drastically different forces and work if their history of movement was different [23]. However, which property may underly the behavior we see in our study is unclear from our study. While previous studies in unsteady tasks like standing balance have shown the importance of short range stiffness [24], we hypothesize that this is not at play here due to the cyclic nature of movement. Instead, we posit that force enhancement might be at play since the main discrepancy between the hill model and real muscle seems to occur when the height of the ground is dropped and the Hill-type model maintains similar force as the real muscle during the initial shortening of the muscle, however loses force after the stretch and end of stimulation - similar to residual force enhancement trials [25]. However, this is hard to conclude without explicit modeling of the same either mechanistically via cross-bridge or phenomenological models as an addition to the Hill-type model. Future studies should try to model the behavior we see in our study to posit which property of muscles may be involved in self-stable behavior of muscles.

Interestingly however, we also show that Hill-type muscles do generate work proportionally to what is needed - thus they do not completely produce the wrong trends or behavior. This is similar to previous studies in humans [26] and goats [27] that show that Hill-type models are able to capture muscle dynamics in a gross manner, while perhaps missing the details. Thus, the self stabilizing property of muscle-tendon units is a combination of Hill-type steady state properties and dynamic muscle properties. This is supported by previous work in the literature that shows that Hill-type muscles are able to generate stable movement in uneven terrain by leveraging their force-velocity property [19]. However, the key difference in our study and the study above was that their study lacked series elasticity and had a phase-locked neural control, not time locked.

1.5.2 Role Of Series Elasticity In Stability

One of the open questions left from our study however is the role of series elasticity in enabling the stable movement. In our study, we see how series elasticity enabled the shifting of muscle behavior from a shorten-stretch negative work cycle to a stretch-shorten stretch positive work cycle due to the change in stimulation time. This is because, if it weren't for series elastic elements, muscles would always have to stretch first and shorten later - while series elastic elements allow the overall kinematics to be decoupled from that of the muscle. In fact, the length and power patterns that we saw the muscle-tendon unit produce when the height of the ground was dropped were similar to those of turkeys undergoing drop landing [28, 29] - thus pointing at the potentially important role of series elasticity in both allowing the muscle-tendon unit to be stable and to buffer demands on muscle power. However, we were unable to test the role of series elasticity directly, as we were unable to change series elastic properties to see their effect in a differential manner. Future studies should use mathematical modeling or biorobotic tools to modify the series elasticity of muscle-tendon units and study their effect directly on stability.

1.5.3 Limitations

Our study has several limitations in terms of answering the broader question of self-stability of muscle-tendon generated movement. One potential limitation in our study is inaccuracy of the motor in following the commanded lengths of the mass. This tends to add positive work to the mass in each cycle. However, since motor work was not adaptive to the task, we are encouraged that the adaptive nature of the system is derived from the muscle-tendon unit. In addition, our study only evaluated a single muscle-tendon unit actuating a mass in one dimensional movement. How this generalizes to a whole limb/animal is an open question. However, since animals use a proximo-distal gradient [8] in joint function during unsteady tasks like uneven terrain with the proximal joints doing the main adaptive work, and since distal joints directly contact and interact with the ground, we are encouraged that our results are broadly applicable. In addition, we are further encouraged by human hopping studies where the task is dominated by distal joints [5] that show similar trends in muscle-tendon dynamics when people are hopping on a platform that is suddenly removed [17]. Yet, our conclusions are still in one dimension, whereas a lot of important instabilities in locomotion occur in 2D or 3D. Future studies should use biorobotic tools [30] or build robots that are actuated by muscles [31] and can move in 2D or 3D to further test the self-stable property of muscles.

1.6 Conclusion

In this study, we asked: is it possible for muscle-tendon units to reject a broad range of terrain perturbations without any change to neural control? By using a novel biorobotic interface, we show that muscle-tendon units dissipate energy proportionally to the energy injected in the first hop and are able to dissipate all the energy within 3 hops. Furthermore, we show Hill-type models underestimate the force of the real muscle-tendon unit – thus suggesting the importance of dynamic muscle properties. Future studies should investigate what mechanical and cellular property of muscle-tendon units enable this behaviour and what the role and effect of reflex feedback is on top of this behaviour. Leveraging the results of our study and suggested future studies, roboticists of the future may be able to build legged systems with cheap robust low bandwidth sensors and controllers by embedding self-stabilizing properties in the mechanics of their actuators.

CHAPTER 2

TRADEOFFS BETWEEN STABILITY, AGILITY, AND ECONOMY IN MUSCLE-TENDON ACTUATED CYCLIC TASKS

2.1 Abstract

Animals tune the mechanical properties of their actuators to the primary task that they care about ethologically. However, they have to perform other tasks as well, and it is unclear how generalizable the benefits of morphological properties are to different tasks. For example, human tendons are tuned for mechanical energy storage and return to enable metabolically cheap locomotion, however it is unclear if that morphology makes humans less stable or less agile due to ’spring-like’ tuning. Thus, here we ask : How does muscle-tendon morphological tuning affect stability, agility, and economy of cyclic movement? In particular, we take a look at the effect of series elasticity on agility, stability, and economy of hopping using a mathematical modeling approach. We built a model of hopping where a mass was actuated by a cyclically stimulated muscle-tendon unit across a lever arm to generate hopping. We then studied the metabolic cost of hopping per hop (metabolic power) for one steady state cycle, the average energy dissipated per hop (stability power) when the system is perturbed by changing the ground, and the maximum work done per hop (agility power) when the timing of stimulation is changed during a hop. In this study, we found that while Achilles muscle-tendon morphology may be tuned to reduce metabolic cost, it may cause lower stability and agility. In addition, we this tendon stiffness value lies on a pareto front where deviating from it causes a tradeoff between metabolic power, stability, and agility. This implies that animals need to tune their series elasticity to the task they care most about. Future studies building on this work should look at how animal tendon stiffness determines their function - both in comparative and rehabilitative settings.

2.2 Background

Animals heavily rely on the mechanical properties of their morphology to move in the world. So much so, that there is plenty of evidence suggesting that animals infact optimize the appropriate morphological variable that best aids their main tasks. For example, tendons stiffness in bullfrogs are optimized for energy storage during jumping ([32]), while tendon stiffness in the calf muscles of humans are potentially optimized for energetic cost of steady movement ([33]). Similarly, with regard to muscles, experimental studies show that muscles in swimming frogs are thought to be optimized for rowing [34] and mathematical modeling has shown that muscle-tendon design is tuned to the type of oscillatory movement the animal performs [35]. Yet, a few questions arise from this - (1) what is the form-function space that muscle-tendon units reside in across various tasks? and (2) since animals tune their muscle-tendon units to a particular task, does it mean that they lose performance on other tasks? For example, by leveraging a tendon stiffness that optimizes for metabolic cost using energy storage and return in the tendon, does it mean that animals may be unstable because springs cannot dissipate energy? Alternatively, animals are stable due to morphological properties, does it mean that they won't be able to voluntarily change their movement and be agile and be sucked back into their stable limit cycle? And lastly, if animals choose morphological properties that make them agile and does it mean they have large metabolic penalties using muscle states that prone to generate work?

One way to study this question is to study related but different animals/athletes to tease apart their differences and functional outcomes because of it. Previous studies have shown that running dogs have leaner limbs and slower oxidative fibers, whereas fighting dogs are bulkier and have fast fibers [36]. Also, looking at endurance trained vs power trained athletes, we know that longer fascicles allow for greater power but have a higher metabolic cost[37]. However, the problem with such studies are that they are primarily correlational, and animals differ in multiple properties - in addition to the property of interest. For example, we know that sprinters have higher tendon stiffness values than distance runners, however they also had higher gastrocnemius forces - thus it wasn't possible to determine if this was an athletic advantage or not [38]. Another way to answer the same question would be to use mathematical modeling. This allows us to control all the variables and inputs, thereby allowing precise and causal relationships being drawn. Previous modeling work has shown that tendon dynamics dictates locomotor performance on the hydrodynamic work vs economy spectrum [39]. However, this study was not done in the context of cyclic movement.

Therefore, in this paper, we ask : given a muscle-tendon morphology, what does it mean for stability, agility, and economy during cyclic terrestrial locomotion? In particular, we strive to find how stability, agility, and economy of hopping change with changing tendon stiffness. To answer this question, we built a model of human hopping with a single lumped calf muscle-tendon unit. We do so because in this context, whole body mechanics and energetics is directly linked to muscle-tendon unit body mechanics and energetics, and thus allows us to directly infer how muscle-tendon properties determine whole body level performance tradeoffs across tasks [40]. We hypothesize that there exists a tendon stiffness that is resonant with the task that leads to minimum metabolic cost, and that higher and lower tendon stiffnesses would lead to higher metabolic cost [33]. Secondly, since locomotion that is energetically optimal is potentially spring-like and resonant to the task [20], we hypothesize that tendon stiffnesses that lead to minimum metabolic cost would lead to minimizing stability. Lastly, since agility requires high sensitivity to neural control and stability requires minimum sensitivity, we hypothesize that agility and stability are inversely related. In addition, we hypothesize that agility and economy are inversely related, since high agility potentially requires the muscle to be in a primarily active state that can respond with both active positive and negative work.

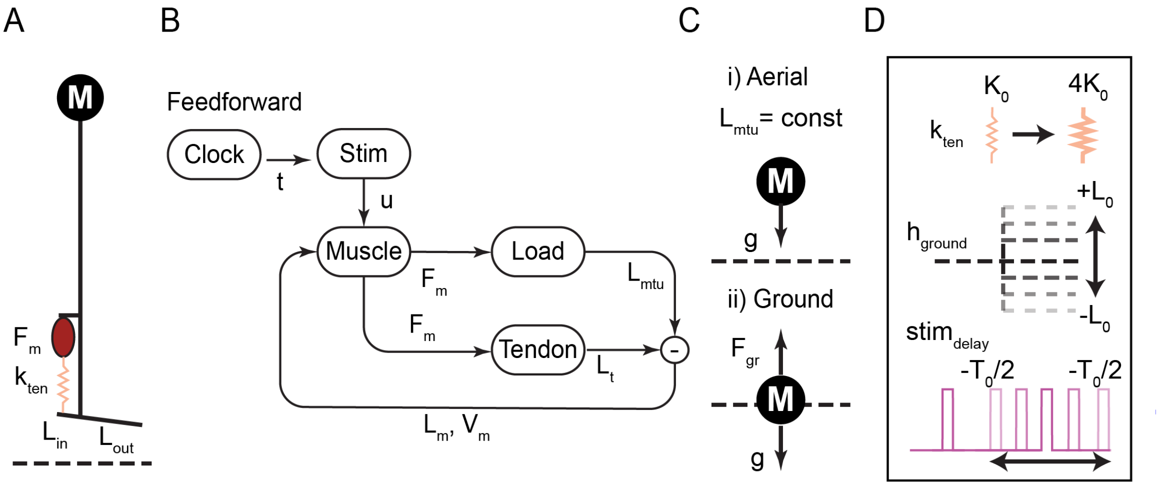

Figure 2.1: A) Diagram of mathematical model (B) Schematic of mathematical model. (C) Definitions of aerial and ground phase. In aerial phase, the length of the muscle-tendon unit is kept constant and only gravity acts on the mass. In stance phase, the length of the MTU is proportional to the change in mass position and the both the force of the MTU and gravity apply to the mass. (D) Parameter sweeps of interest. We vary tendon stiffness from K_0 to 4 K_0, height from -L_0 to L_0 and change the timing of stimulation from -T_0/2 to T_0/2

2.3 Methods

Previously, simple models of human hopping have been used to investigate the effect of the underlying neuromechanics [40] and passive exoskeletons [41] on the mechanics and energetics of hopping. We use a modified version of the model above to investigate its stability in uneven terrain. The model consists of a mass in gravity that is cyclically actuated by a compliant muscle-tendon unit. Once the model achieves stable hopping, we either (a) perturb the height of the ground or (b) perturb the timing of stimulation while the mass is in the air. We investigate the metabolic cost of the hop before the perturbation, the dissipative power of the muscle-tendon unit to reject perturbations and the range of works/powers the muscle-tendon unit can generate by shifting the timing of stimulation.

2.3.1 Center of Mass Dynamics

We developed a model of human hopping that simulates hopping using only one joint (i.e., the ankle) (see Figure 2.1). The human body is modelled as a lumped point mass in a gravitational field. The body is modelled as a hybrid dynamical system with two states: (1) aerial and (2) stance. In the aerial phase, the body is in the air and has no contact with the ground. Thus, the mass only force experienced by the mass is the force of gravity. When the body is in contact with the ground, it is in the stance phase. During the stance phase, the mass is actuated by both the gravitational field and the summed forces from muscle-tendon unit and passive exoskeleton. The position of the mass where the body loses contact with the ground and the state transitions to aerial from stance is designated as $y = 0$.

$$\ddot{y} = -g + \frac{F_{mass}}{m} \qquad \text{(stance)} \qquad (2.1)$$

$$\ddot{y} = -g \qquad \text{(aerial)} \qquad (2.2)$$

2.3.2 Muscle-Tendon Unit Dynamics

The force experienced by the mass and the mass' position is related to the force from the muscle-tendon unit + exoskeleton complex and their length by the effective mechanical advantage of the joint. The length of the muscle-tendon unit/exoskeleton is kept constant during the swing phase.

$$F_{mass} = \text{EMA} * F_{mtu} \qquad (2.3)$$

$$L_{mtu} = L^{0}_{mtu} - y * \text{EMA} \qquad \text{(stance)} \qquad (2.4)$$

$$L_{mtu} = L^{0}_{mtu} \qquad \text{(aerial)} \qquad (2.5)$$

The muscle-tendon unit is modelled as a Hill-type muscle in series with a nonlinear tendon. The force provided by the muscle is a function of the activation level of the muscle, the length of the muscle fascicle and the velocity of contraction of the fascicle. The length and velocity dependence are captured by the classical force-length and force-velocity properties of Hill-type models of muscle. The exoskeleton is modelled as a linear spring that produces force in proportion to the length of the muscle-tendon unit.

$$F_{mtu} = F_{max} * act * FL(L_{mus}) * FV(V_{mus}) \tag{2.6}$$

The length of the muscle is determined by the length of the muscle-tendon unit and the length of the tendon. And the length of the tendon depends on the force provided by the muscle and is characterized by a linear spring with a toe-region. The velocity is simply the differential of the lengths with respect to time.

$$L_{mus} = L_{mtu} - L_{ten} \tag{2.7}$$

$$L_{ten} = L^0_{ten} + \int_0^{F_{mtu}} \frac{dF}{k_{ten}(F)} \tag{2.8}$$

2.3.3 Activation Dynamics

The muscle is stimulated by a clock-based feedforward signal that consists of a square wave with 10% duty factor and a frequency of 2.5Hz and has a variable amplitude. The stimulation is then converted to an activation signal by the following activation-deactivation coupling equation.

$$a(t) = \frac{u(t)}{T_{act}} - (\beta + (1-\beta) * u(t))\frac{a(t)}{T_{act}} \text{ where } \beta = \frac{T_{act}}{T_{deact}} \tag{2.9}$$

2.3.4 Simulation Protocol

The mass is released from an arbitrary initial height and the stimulation is started with an arbitrary initial phase. The cyclic force generated by the muscle and exoskeleton in stance causes the mass to start hopping. The model is allowed to hop for 50 seconds to make sure that it has achieved steady hopping. After 50 seconds, when the mass is in the aerial state, we either (a) change the height of the ground by changing the threshold of where aerial phase leads to stance phase by a certain height h (i.e., y = -h) or (b) change the timing of stimulation by shifting the phase of the feedforward stimulation signal. The system is again allowed to settle into steady hopping for 50 seconds. This process is repeated for various drop heights (range: ± steady state hop height), tendon stiffnesses (range: 1-4 K_0, where K_0 is the stiffness of the spring-mass system of mass 35 kg and EMA 0.3 that would resonate at 2.5Hz) and timing of stimulation (range: ± 50% of cycle time) H is the height of the steady state hop. For each of these cases, we measure the metabolic cost of hopping during steady hopping, the mechanical energy generated/dissipated in each step after the perturbation, how long the system takes to dissipate 90% of the energy injected into the system and the peak strain of the muscle fascicle in each step.

2.3.5 Measurements

1. Work Done in A Step: Defined by the energy generated/dissipated in each hop. We measure this by measuring the difference between the total energy of the system in the previous step and the current step. We do this by measuring the total energy when the mass is in peak height in the air and kinetic energy is zero. So, the energy dissipated in a particular hop ‘i’ is calculated as:

$$W_i = mg(y_i - y_{i-1}) \tag{2.10}$$

2. Settling Time: Defined as the minimum time taken to dissipate 90% of the energy injected into the system.

$$T_{settling} = \sum_{i=1}^{N} T_i \text{ where} \tag{2.11}$$

$$N = \text{argmin}_N \sum_{i=1}^{N} (E_i - E_{perturb}) < 0.1 * E_{perturb} \tag{2.12}$$

3. Metabolic Power: This is the metabolic cost of steady hopping. The equation is

adapted from [35]

$$E_{met} = \frac{\int_{t_0}^{t_0+T0} P_{met}\, dt}{T0} \tag{2.13}$$

$$P_{met} = a(t) * F_{max} * V_{max} * \Phi(\frac{V_{mus}}{V_{max}}) \tag{2.14}$$

4. Agility Power: This is the range of powers possible in a single step.

$$E_{agility} = max(W_i) - min(W_i) \tag{2.15}$$

$$P_{agility} = max(W_i/t_i) - min(W_i/t_i) \tag{2.16}$$

5. Settling Power: Defined as the average perturbation dissipative power of the system.

$$P_{set} = \frac{\text{mean}(E_{perturb})}{\text{mean}(T_{settling})} \tag{2.17}$$

2.3.6 Normalization

To give context to the numbers and make them interpretable/relatable, we normalized the numbers in our results to those of a representative spring-mass system that resonantly hops at the frequency of 2.5 Hz. We thus normalized time by $T_0 = 0.4s$, mass by $M_0 = 35kg$, and acceleration by $A_0 = 9.8m/s$. Thus, we normalized forces by $F_0 = A_0M_0$, stiffness by $K_0 = M_0W_0^2$ where $W_0 = 2\pi/T_0$ is the resonant frequency of the system. Subsequently, we normalized lengths by $L_0 = F_0/K_0$ which represent the length that the representative spring compresses when loaded by the mass and $V_0 = L_0W_0$. We also normalized energy by $E_0 = 0.5K_0L_0^2$ which is the energy stored in the spring when loaded by the mass and power by $P_0 = 0.5F_0V_0$.

2.4 Results

2.4.1 Steady State Metabolics

During steady state hopping, the mass cyclically oscillates between its maximum height in the aerial phase and its minimum height in the stance phase (Figure 2.2). In the aerial phase, the mass follows a parabolic path and gravity is the only force being exerted on the mass. Therefore, the mass maintains constant total energy by exchanging its kinetic and potential energy. During the stance phase (denoted by shaded area) however, the muscle-tendon unit also produces force on the mass. Therefore, when the mass is in stance phase, is stimulated a little bit after ground contact (denoted by the thick black line). the muscle-tendon unit initially does negative work on the mass, as the direction of ground reaction force generated by the muscle-tendon unit is opposite the direction of movement of the mass till the minimum height is reached. This is followed by positive work done by the muscle-tendon unit on the mass as the force and direction of movement is aligned. During steady state hopping, the net work (positive and negative work combined) done by the muscle-tendon unit over a step is zero. This requires metabolic work. Thus, we can see the metabolic power over the hop cycle. We can integrate this and divide by hop time to get metabolic power. On the right panel, we see the metabolic power over multiple tendon stiffness values from 0-4 K_0. We see that metabolic cost is minimized at around 2 K_0 and is higher as we deviate from there. We also see that the metabolic power of Achilles tendon (denoted by black line) is close to the minimum.

2.4.2 Stability

In the left panel of Figure 2.3, we see what happens when we change the height of the ground suddenly. When we drop the height of the ground , the mass continues to move through its parabolic path in the air beyond the previous ground height if the height of the ground is dropped, and touches the ground before it if the height of the ground is raised.

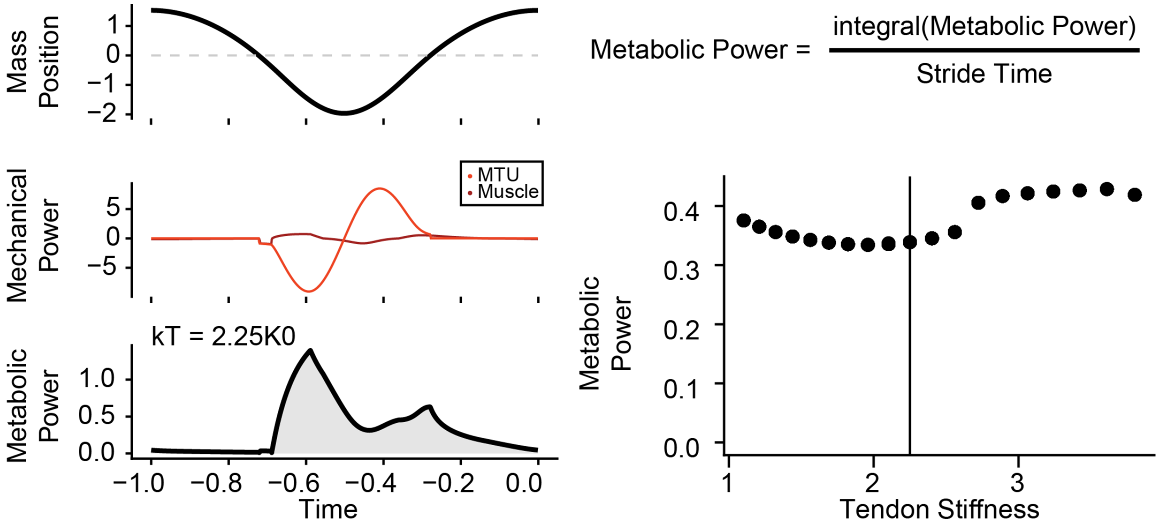

Figure 2.2: (left) Mass position, mechanical power of muscle-tendon unit and muscle, and metabolic power of the muscle for one steady-state cycle. We can integrate the metabolic power over time and divide by cycle time to get average metabolic power. This panel shows the above variables for a biological tendon stiffness of 2.25 K_0 (right) Here we show the average metabolic power for a steady state cycle for different tendon stiffnesses. The black line indicates the biological tendon stiffness value.

Therefore, energy is added to the system when height is dropped. When the mass is in the stance phase, the muscle-tendon unit does work on the mass, resulting in energy generation/dissipation in the first step. Over multiple hops, we see that the system eventually dissipates this energy away, and we can measure the settling time - which is the time taken to dissipate 90% of the perturbation. We can measure this for various hop heights as shown in the left bottom panel, and find the average settling power of the system for each tendon stiffness - where settling power is the average energy injected divided by the average settling time. When he look at the right panel of Figure 2.3, we see the average stability power of the muscle-tendon unit for each tendon stiffness. Here, we see that intermediate values of tendon stiffness have the lowest stability power and as we increase/decrease tendon stiffness from there, we see higher stability powers.

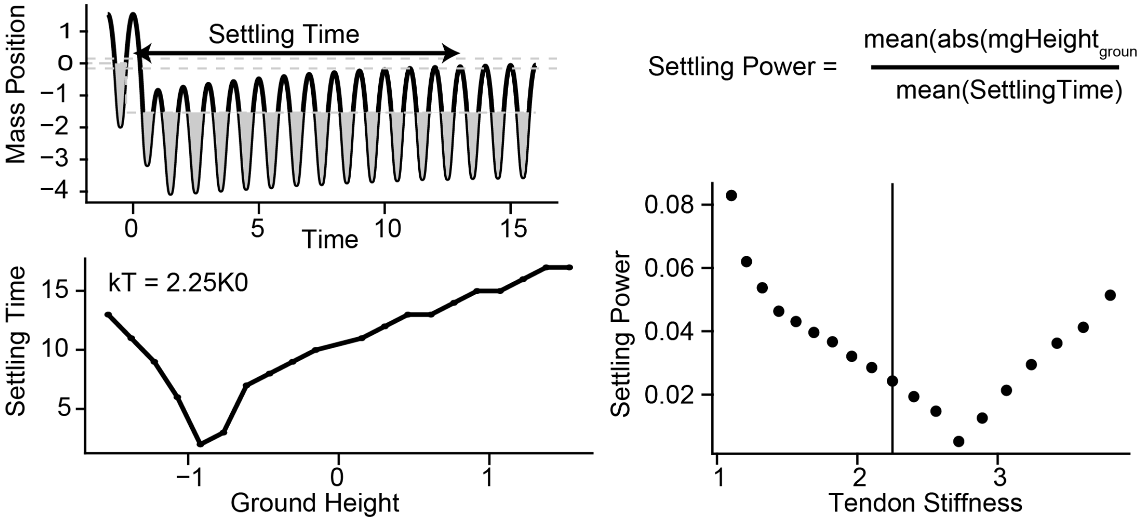

Figure 2.3: (left, top) Mass position and ground height over time. Note how when the height of the ground is changed, the mass overdissipates energy in the first hop and over 13 hops comes back to steady state. We define the time the mass takes to come back to steady state as settling time (left, bottom) Here we show settling time for different ground height changes for a biological tendon stiffness of 2.25 K_0. We take the average of energy injected by the height changes and divide it by the average settling time to get settling power. (right) Here we show settling power for various tendon stiffness values. Note, a settling power of 0 indicates the muscle-tendon unit is behaving spring-like. The black line indicates the biological tendon stiffness value.

2.4.3 Agility

In the left panel of Figure 2.4, we see what happens when we change the timing of stimulation suddenly. When we delay the timing of stimulation, the mass is actuated passively as it reaches the bottom dead center and is stimulated on the way up, thus generating positive work. In the bottom panel, we can see the work done on the mass for various stimulation delay and measure the maximum and minimum work done for each tendon stiffness. We call that the agility power of the system.

In the right panel of Figure 2.4, we look at the maximum negative power generated by the muscle-tendon unit as a function of tendon stiffness. We see that maximum negative power has a bowl shape, with maximum negative power near 2 K_0. The maximum negative power decreases for both increase and decrease of tendon stiffness from that midpoint. In addition, the positive agility power is zero around 2.5 K_0, and increases as we move away

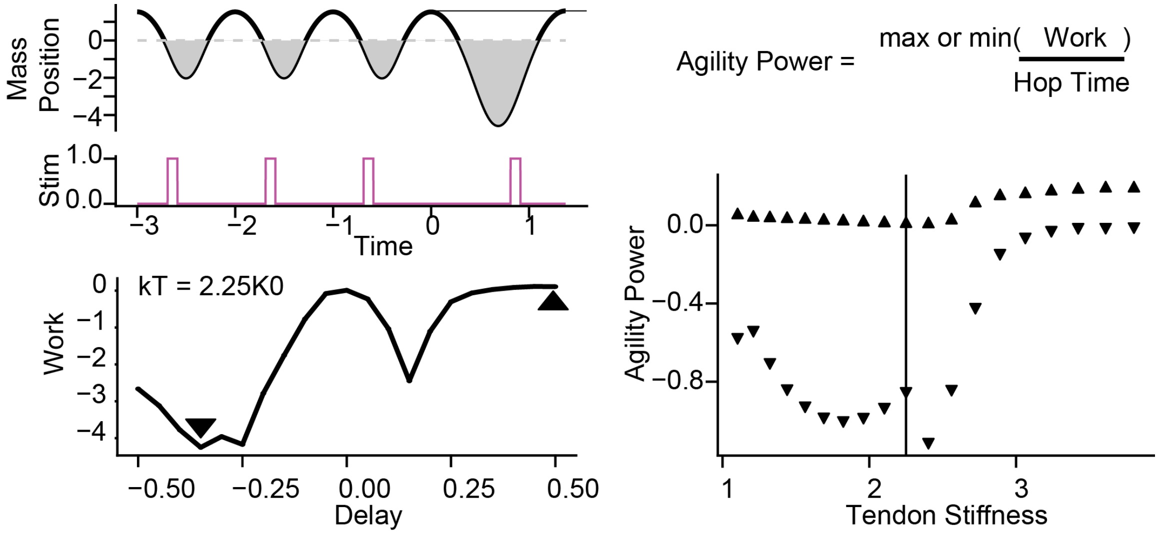

Figure 2.4: (left, top) Mass position and stimulation over time. Note how when the timing of stimulation is changed, the muscle-tendon unit passively does negative work on the mass followed by active positive work that leads to increase in hop height. (left, bottom) Here we show the work done on the mass for different delays in stimulation timing. We denote the maximum positive work with an upright triangle and maximum negative work with a downwards triangle (right) Here we show maximum positive power and maximum negative power for various tendon stiffness values. Note an agility power of 0 indicates the muscle-tendon unit is behaving spring-like and cannot generate positive work. The black line indicates the biological tendon stiffness value.

from that midpoint.

2.5 Tradeoffs In Agility, Stability, And Economy

In Figure 2.5 we see the tradeoffs in economy, stability and agility. In particular, there are two ways of looking at this. (a) We can look at what happens as we deviate from the value for Achilles tendon stiffness and (b) We can look at where the pareto front may lie for tradeoffs between economy, stability, and agility.

As we deviate from biological tendon stiffness by increasing it, we get increasing metabolic power, decreasing settling power, and decreasing agility power from the biological value. Thus, there is no local gradient for either of this what is useful. Similarly, as we reduce tendon stiffness, we see metabolic power decreases, settling power increases, agility positive power increases and agility negative power decreases. Thus, we can gain

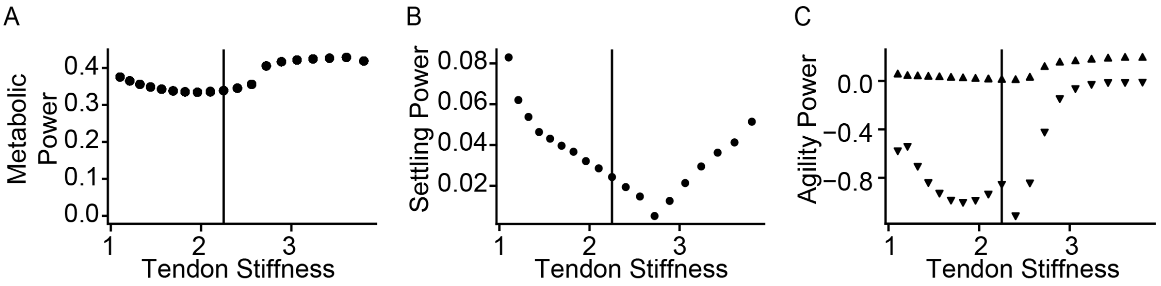

Figure 2.5: Metabolic power of a steady state cycle across tendon stiffnesses (B) Stability power across tendon stiffnesses (C) Positive and negative agility power. Note that metabolic power increases, settling power decreases, positive agility power increases and negative agility power decreases with increasing tendon stiffness from the biological value. Similarly, metabolic power decreases, settling power increases, agility positive power increases and agility negative power decreases with decreasing tendon stiffness from the biological tendon stiffness value.

economy, stability, and positive agility by decreasing tendon stiffness at the loss of negative agility.

By looking at it wholistically, we find two pareto fronts. The lowest tendon stiffness values have the highest stability, and intermediate metabolic cost. As we increase tendon stiffness from there, we gain economy negative agility and lose stability and positive agility - thus forming a pareto front for optimization. From there, we lose both metabolic cost, stability, and agility until we reach a tendon stiffness where metabolic cost is intermediate and stability is minimum and negative agility is maximum. From there, as we increase tendon stiffness, we gain stability and positive agility while losing economy and negative agility - thus forming another pareto front - albeit an inferior one to the previous case as it has both lower stability and economy.

2.6 Discussion

In this study, we aimed to understand how muscle-tendon morphology (series elastic specifically) affects stability, agility and metabolic cost of movement. In particular, since animals optimize tendon stiffness values for an optimal resonant spring-like gait that rely on their series elastic element, we hypothesized that biologoical tendon stiffness values would be optimal for metabolic cost and the worst for stability and agility. In addition, we hypothesized that there would be a pareto-front of tendon stiffness values that animals can use to tradeoff each of these outcome measures, as opposed to having one optimal tendon stiffness value that optimizes all of the above.

2.6.1 Role Of Tendon Stiffness In Determining Economy Of Movement

Previous mathematical modeling and experimental studies show that for a given tendon stiffness, there exists a stimulation frequency that minimizes metabolic cost, which relates to the system's resonant frequency [40, 20]. This is similar to our study, except we keep the frequency the same and vary the stimulation frequency. In addition, experimental studies of muscles interacting with variable series elasticity have shown that there exists a combination of compliant tendons with large strains (similar to the resonant case where the tendon is doing all the work) that lead to maximum efficiency [42]. Also, previous studies have found that biological tendon stiffness values maximize efficiency in hopping. While in our study, we didn't measure efficiency per se, we find that the same biological tendon stiffness value was close to the minimum of the metabolic cost bowl.

2.6.2 Role Of Tendon Stiffness In Determining Stability Of Movement

While no previous study has directly addressed the role of tendon stiffness in determining stability of movement, we find that biological tendon stiffness in our model produced behavior similar to real human experiments. In human hopping experiments, we found that participants underdissipated for small changes in ground height and overdissipated in large heights [17]. This is similar to what we see in our study, however humans tend to correct their errors relatively quickly (within 5 hops) whereas in our study, the model takes upto 15 hops to stabilize itself fully. This is likely due to humans being able to use reflexes and motor control strategies in the steps after the perturbation to stabilize themselves. We also know at the extreme case, Hill-type muscles are able to stabilize movement in uneven terrain (at least using a phase-locked neural control strategy) [19].

Another interesting result from our experiment stated that both large tendon stiffness values and small tendon stiffness values were good for stability, whereas intermediate values were the worst. However, an important point is that our definition of stability is a very particular one that relates to following the terrain after a step-down or step-up perturbation. Often in the real world where you have continuously varying uneven terrain, it is likely that the objective of animals is not to follow the terrain, and instead act like the active suspension in a car where the car body is blind to changes in terrain. In such cases, animals may want to choose tendon stiffness values that are closer to the lowest settling power instead.

2.6.3 Role Of Tendon Stiffness In Determining Agility Of Movement

In our study, we found that biological tendon stiffness values have minimal positive power capability and maximum negative power capability. This may be due to the fact that biological tendon stiffness values are close to resonant hopping with high hop heights. Thus, it may not be possible 'to improve upon' resonant hopping conditions by generating positive work and by changing stimulation timing we only generate large amounts of negative work.

We also found that agility is maximized at high tendon stiffnesses for positive work. In fact, This result is very counterintuitive since high tendon stiffnesses should lead to the muscle being directly coupled to the load, and thus lead to being muscle dominated. And assuming a task is muscle dominated, we would guess that the muscle would be better at dissipating energy due to the eccentric force velocity property. Instead, we see that large tendon stiffnesses have a lack of ability to generate negative work. Future analysis should look into the underlying muscle-tendon interaction at play here to determine why this is the case.

2.6.4 Economy, Agility, and Stability Tradeoffs In The Achilles Tendon

We find that human tendon stiffness lies on a pareto front of optimization, where humans prefer and optimize for economy. While there are no benefits of having a higher tendon stiffness, humans may gain stability and agility by having a lower tendon stiffness value while losing economy. This is an interesting result, as we know that several human populations have lower tendon stiffnesses. For instance, older adults have lower tendon stiffness values than young adults [43]. While previous work has shown that this leads to detriment in metabolic cost [43], here we show that this may be due to an increase in stability and agility gained by older adults - both important tasks as it relates to safety of movement being preferred over economy. However, that assumes older adults would need to move at speeds of younger adults for our results to still hold true.

2.6.5 Limitations

However, our work has several caveats. For one, we used Hill-type models that are known to be good at predicting steady state movement and not unsteady movement like uneven terrain or agility. However, previous modeling work has shown similar behavior in model as compared to experimental hopping [40] during steady state. In addition, we find that in our model, the trends of energy dissipation and generation are similar to those in human hopping even though the speed at which they occur is different [5]. Thus, while our studies may not predict exact values, we believe simple models likely show similar trends to experimental results. To test this experimentally, however, is very difficult. In in-vivo studies, we may be unable to modify morphology directly and precisely. For example, exercise studies seem to be able to change tendon morphology over a 14 week period, however they also change muscle properties. Another approach could be to compare different populations of animals [36] or humans [37] that have varying muscle-tendon properties and see their outcome measures. And while in-vitro experiments allow the variation of tendon stiffness directly, they only allow a very limited number of trials. We believe that we may be able to test the effect of tendon stiffness on stability for a fixed change in ground change using biorobotic tools [20] while still maintaining minimum trials. To study agility however across tendon stiffness value using an approach similar to ours would prove difficult. Thus, future studies should look for methods to maintain muscle life during in-vitro experiments over large number of trials [44] or build algorithms can optimize movement generated by a muscle-tendon unit in a sample efficient way.

Another limitation of our work is how we compute agility. In our study, we limited our search space along similar stimulation profiles that varied timing to span the space. However, animals may be able to use a more rich trajectory space of stimulation patterns. However, the question we are asking involves the trend across tendon stiffness values, and we believe the limitations and constraints on muscle function given a tendon stiffness will remain true even under different neural patterns. Thus, while magnitudes of maximum and minimum power may change, we believe that the trends across tendon stiffness will still hold true.

2.7 Conclusion

In this study, we aimed to answer the question : How do muscle-tendon morphological tuning affect stability, agility, and economy of cyclic movement? In particular, we take a look at the effect of tendon stiffness. We showed that minimum metabolic cost occurs at tendon stiffness equal to 2 times the stiffness of a spring-mass system that would resonate at 2.5Hz (the driving frequency of the model). However, the important phase transition of behavior occurs at tendon stiffnesses above 2.5 K_0. At this tendon stiffness value, settling power is minimized and agility transitions from being negative power dominated to positive power dominated. Biological tendon stiffness on the other hand had close to minimum metabolic cost, medium stability, low positive power agility and high negative power agility - thus showing that animals need to tradeoff stability, agility and economy, and humans potentially prioritize metabolic cost and voluntary energy dissipation over voluntary generation and stability. Future studies should try to determine this experimentally using biorobotic tools or building robots with muscle-tendon units grown in the lab with tailor made morphological features.

CHAPTER 3

TRADEOFFS BETWEEN AUGMENTATION OF STABILITY, AGILITY, AND ECONOMY WITH PASSIVE EXOSKELETONS IN MUSCLE-TENDON ACTUATED CYCLIC TASKS

3.1 Abstract

Passive exoskeletons augment steady-state movement by changing the morphology of human beings. The question we ask in this paper is, if exoskeletons affect morphology and morphology affects task performance, how do passive exoskeletons augment unsteady movement across various task objectives - namely stability, agility, and economy - and what are the tradeoffs among them? To answer this question, we developed a simple model of human hopping. The model consists of a mass in gravity that is cyclically actuated by a compliant muscle-tendon unit and a passive exoskeleton in parallel to produce hopping. Once the model achieves stable hopping, we either (1) perturb the height of the ground while the mass is in the air, or (b) change the timing of stimulation of the muscle-tendon unit. We then studied the metabolic cost of hopping per hop (metabolic power) for one steady state cycle, the average energy dissipated per hop (stability power) when the system is perturbed by changing the ground, and the maximum work done per hop (agility power) when the timing of stimulation is changed during a hop. We find that increasing exoskeleton stiffness decreases metabolic cost, increases stability and increases agility - thus suggesting that there are no tradeoffs in designing exoskeletons. Our results suggest that lightweight passive ankle exoskeletons have potential to move beyond the lab in both steady and unsteady conditions, and future studies should test human stability and agility with passive exoskeletons.

3.2 Background

3.2.1 Designing exoskeleton control policies over multiple tasks/objectives is a hard problem to solve

Exoskeletons are robotic manipulators that are worn by users in parallel to their own limbs. Thus, the total force applied on the environment by the user is the sum total of the action of the human and the exoskeleton, thereby reducing demands on the human for a given task or unlocking new tasks that were previously beyond the human's biological constraints. However, designing exoskeletons is a hard problem with several key challenges. First, we don't know what objectives humans are trying to solve for. For example, while energy minimization has been a leading principle that predicts human movement patterns and several devices exist that can reduce it, when humans are given the choice to pick exoskeleton parameters, they don't tend to pick the metabolically cheapest option[45]. Secondly, let's say we magically find out the objective function that humans are trying to solve for, we lack formalism that allow us to produce effective exoskeleton controllers directly from that objective function since (a) this is a shared control problem where humans and devices co-adapt[46], (b) we don't have a model of human adaptation and (c) neither can observe the full internal state of the other. Thus, current approaches typically rely on the rinse and repeat cycle of scientific or algorithmic hypothesis generation and human testing. The one case where we do have that information/formalism is in reducing metabolic cost of walking [47] - but it is a purely input-output model for a particular type of controller and task and was borne out of extensive experimentation and input-output data collection - simultaneously exhibiting how far exoskeletons have come, and yet how arduous the journey is and how far we have to go.

One general approach of solving shared-control problems with unknown objectives and dynamics is using the formalism of dynamical systems/Markov decision processes. These range from learning human dynamics and policies using EDMD/Koopman Operators when we know a goal[48], to POMDPs when we know the family of goals[49], to deep reinforcement learning when we don't know the goal[50]. In fact, these methods can also be used to augment humans when they are acting suboptimally [51]. However, all of these methods are designed to provide minimal interventions and don't deal with human adaptation over time that arises when we apply large interventions.

3.2.2 Can we instead augment multiple/tasks by modifying the plant dynamics/morphology of the person?

So the question arises, is there a different approach we could take in augmenting human function across multiple tasks and objectives. To answer this question, we take a leaf out of how humans and animals control their movement. Unlike typical engineering systems, animals have large neuromechanical constraints like large neural[9] and inertial delays[10] and compliant actuators [52] that render a purely control centric approach infeasible. Thus, animals control their movement using a combination of morphological and neural computation. That is to say, they rely on the natural dynamics of their limbs and actuators in addition to their nervous system's computational power. In addition, as the task demands on animals change, they change their morphology appropriately via adaptation over short time scales and natural selection over long time scales[53, 54]. Thus, we ask, can we augment multiple objectives at the same time by modifying morphology? And if not, what are the functional tradeoffs of changing morphology?

One approaching of changing morphology and augmenting human movement is passive elastic exoskeletons. Recent advances in passive exoskeleton technology have shown that passive devices at the ankle can reduce the metabolic cost of walking[55] and passive devices at the hip can reduce the metabolic cost of walking[56] and running. Since these devices can't impart net energy to the body, they work by driving the underlying biological muscle-tendon units of the human to more energetically economical states. For example, while walking [57] and hopping [58] with passive exoskeletons, humans have

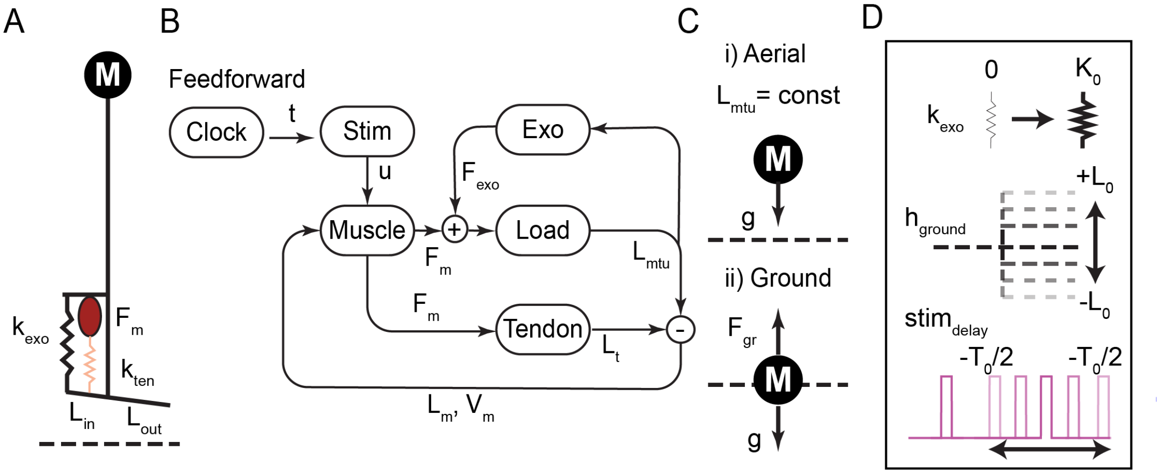

Figure 3.1: A) Diagram Of Model (B) Block Diagram of Model (C) Definition Of Aerial and Ground Phases

longer fascicle lengths, faster muscle velocity and lower muscle activations as compared to walking/hopping without an exoskeleton. Thus, even though elastic exoskeletons cannot impart net energy to the person, they change the underlying state of the muscle-tendon unit in the human – thereby changing the economy of movement. However, it is still unclear whether changing the morphology of human beings can augment function in unsteady environments like uneven terrain or improve agility and how this relates to the morphology of the humans themselves.

Thus, goal of this study is to investigate whether passive exoskeletons can augment the stability, agility, and economy of muscle-tendon units. Since passive exoskeletons lead to lower muscle activation during steady movement, we hypothesize that increased stiffness of passive exoskeletons will lead to lower stability and lower agility.

3.3 Methods

To test our hypothesis, we take a mathematical modeling approach. Previously, simple models of human hopping have been used to investigate the effect of the underlying neuromechanics [40] and passive exoskeletons[41] on the mechanics and energetics of hopping. We use a modified version of the above model to investigate its economy in steady state hopping, stability against unexpected perturbation and ability to voluntarily change movement by changing neural control. The model consists of a mass in gravity that is cyclically actuated by a compliant muscle-tendon unit and a passive exoskeleton in parallel to produce hopping. Once the model achieves stable hopping, we either (a) perturb the height of the ground while the mass is in the air to simulate uneven terrain (or) change the timing of stimulation and measure the work done by the combined exoskeleton-muscle-tendon system in the subsequent hops to determine stability and agility.

3.3.1 Center of Mass Dynamics

We developed a model of human hopping that simulates hopping using only one joint (i.e., the ankle) Figure 3.1. The human body is modelled as a lumped point mass in a gravitational field. The body is modelled as a hybrid dynamical system with two states: (1) aerial and (2) stance. In the aerial phase, the body is in the air and has no contact with the ground. Thus, the mass only force experienced by the mass is the force of gravity. When the body is in contact with the ground, it is in the stance phase. During the stance phase, the mass is actuated by both the gravitational field and the summed forces from muscle-tendon unit and passive exoskeleton. The position of the mass where the body loses contact with the ground and the state transitions to aerial from stance is designated as $y = 0$.

$$\ddot{y} = -g + \frac{F_{mass}}{m} \qquad \text{(stance)} \qquad (3.1)$$

$$\ddot{y} = -g \qquad \text{(aerial)} \qquad (3.2)$$

3.3.2 Muscle-Tendon Unit and Exoskeleton Dynamics

The force experienced by the mass and the mass' position is related to the force from the muscle-tendon unit + exoskeleton complex and their length by the effective mechanical advantage of the joint. The length of the muscle-tendon unit/exoskeleton is kept constant during the swing phase.

$$F_{mass} = \text{EMA} * (F_{mtu} + F_{exo}) \tag{3.3}$$

$$L_{mtu} = L^0_{mtu} - y * \text{EMA} \qquad \text{(stance)} \tag{3.4}$$

$$L_{mtu} = L^0_{mtu} \qquad \text{(aerial)} \tag{3.5}$$

The muscle-tendon unit is modelled as a Hill-type muscle in series with a nonlinear tendon. The force provided by the muscle is a function of the activation level of the muscle, the length of the muscle fascicle and the velocity of contraction of the fascicle. The length and velocity dependence are captured by the classical force-length and force-velocity properties of Hill-type models of muscle. The exoskeleton is modelled as a linear spring that produces force in proportion to the length of the muscle-tendon unit.

$$F_{mtu} = F_{max} * act * FL(L_{mus}) * FV(V_{mus}) \tag{3.6}$$

$$F_{exo} = [L_{mtu} - L^0_{mtu}] * k_{exo} \tag{3.7}$$

The length of the muscle is determined by the length of the muscle-tendon unit and the length of the tendon. And the length of the tendon depends on the force provided by the muscle and is characterized by a linear spring with a toe-region. The velocity is simply the differential of the lengths with respect to time.

$$L_{mus} = L_{mtu} - L_{ten} \tag{3.8}$$

$$L_{ten} = L^0_{ten} + \int_0^{F_{mtu}} \frac{dF}{k_{ten}(F)} \tag{3.9}$$

3.3.3 Activation Dynamics

The muscle is stimulated by a clock-based feedforward signal that consists of a square wave with 10% duty factor and a frequency of 2.5Hz and has a variable amplitude. The stimulation is then converted to an activation signal by the following activation-deactivation coupling equation.

$$a(t) = \frac{u(t)}{T_{act}} - (\beta + (1-\beta) * u(t))\frac{a(t)}{T_{act}} \text{ where } \beta = \frac{T_{act}}{T_{deact}} \tag{3.10}$$

3.3.4 Simulation Protocol

The mass is released from an arbitrary initial height and the stimulation is started with an arbitrary initial phase. The cyclic force generated by the muscle and exoskeleton in stance causes the mass to start hopping. The model is allowed to hop for 50 seconds to make sure that it has achieved steady hopping. After 50 seconds, when the mass is in the aerial state, we either (a) change the height of the ground by changing the threshold of where aerial phase leads to stance phase by a certain height h (i.e., y = -h) or (b) change the timing of stimulation by shifting the phase of the feedforward stimulation signal. The system is again allowed to settle into steady hopping for 50 seconds. This process is repeated for various drop heights (range: ± steady state hop height), exoskeleton (range: 0-1 K_0, where K_0 is the stiffness of the spring-mass system of mass 35 kg and EMA 0.3 that would resonate at 2.5Hz) and timing of stimulation (range: ± 50% of cycle time). For each of these cases, we measure the metabolic cost of hopping during steady hopping, the mechanical energy generated/dissipated in each step after the perturbation, how long the system takes to dissipate 90% of the energy injected into the system and the peak strain of the muscle fascicle in each step.

3.3.5 Measurements

1. Work Done in A Step: Defined by the energy generated/dissipated in each hop. We measure this by measuring the difference between the total energy of the system in the previous step and the current step. We do this by measuring the total energy when the mass is in peak height in the air and kinetic energy is zero. So, the energy dissipated in a particular hop 'i' is calculated as:

$$W_i = mg(y_i - y_{i-1}) \tag{3.11}$$

2. Settling Time: Defined as the minimum time taken to dissipate 90% of the energy injected into the system.

$$T_{settling} = \sum_{i=1}^{N} T_i \text{ where} \tag{3.12}$$

$$N = \text{argmin}_N \sum_{i=1}^{N} (E_i - E_{perturb}) < 0.1 * E_{perturb} \tag{3.13}$$

3. Metabolic Power: This is the metabolic cost of steady hopping. The equation is adapted from [35].

$$E_{met} = \frac{\int_{t_0}^{t_0+T0} P_{met}\, dt}{T0} \tag{3.14}$$

$$P_{met} = a(t) * F_{max} * V_{max} * \Phi(\frac{V_{mus}}{V_{max}}) \tag{3.15}$$

4. Agility Power: This is the range of powers possible in a single step.

$$E_{agility} = max(W_i) - min(W_i) \tag{3.16}$$

$$P_{agility} = max(W_i/t_i) - min(W_i/t_i) \tag{3.17}$$

5. Settling Power: Defined as the average perturbation dissipative power of the system.

$$P_{set} = \frac{\text{mean}(E_{perturb})}{\text{mean}(T_{settling})} \tag{3.18}$$

3.3.6 Normalization

To give context to the numbers and make them interpretable/relatable, we normalized the numbers in our results to those of a representative spring-mass system that resonantly hops at the frequency of 2.5 Hz. We thus normalized time by $T_0 = 0.4s$, mass by $M_0 = 35kg$, and acceleration by $A_0 = 9.8m/s$. Thus, we normalized forces by $F_0 = A_0M_0$, stiffness by $K_0 = M_0W_0^2$ where $W_0 = 2\pi/T_0$ is the resonant frequency of the system. Subsequently, we normalized lengths by $L_0 = F_0/K_0$ which represent the length that the representative spring compresses when loaded by the mass and $V_0 = L_0W_0$. We also normalized energy by $E_0 = 0.5K_0L_0^2$ which is the energy stored in the spring when loaded by the mass and power by $P_0 = 0.5F_0V_0$.

3.4 Results

In this paper, we investigate how modifying the morphology of a muscle-tendon unit using a passive exoskeleton affects stability, agility, and economy of movement. To answer this question, we use a mathematical modelling approach. We model the hopping human as a mass in gravity that is cyclically actuated by a compliant muscle-tendon unit and a passive exoskeleton in parallel. Once the model achieves stable hopping, we either (a) perturb the height of the ground while the mass is in the air to simulate uneven terrain or (b) perturb the timing of stimulation to simulate voluntary control. We investigate the energy lost/gained in each step, how many hops the mass takes to recover from the ground perturbation, and the metabolic cost of steady state hopping.

3.4.1 Steady State Metabolics

During steady state hopping the mass cyclically oscillates between its maximum height in the aerial phase and its minimum height in the stance phase (Figure 2.2). In the aerial phase, the mass follows a parabolic path and gravity is the only force being exerted on the mass. Therefore, the mass maintains constant total energy by exchanging its kinetic and potential energy. During the stance phase (denoted by shaded area) however, the muscle-tendon unit also produces force on the mass. Therefore, when the mass is in stance phase, is stimulated a little bit after ground contact (denoted by the thick black line). the muscle-tendon unit initially does negative work on the mass as the direction of ground reaction force generated by the muscle-tendon unit is opposite the direction of movement of the mass till the minimum height is reached. This is followed by positive work done by the muscle-tendon unit on the mass as the force and direction of movement is aligned. During steady state hopping the net work (positive and negative work combined) done by the muscle-tendon unit over a step is zero. This requires metabolic work. Thus, we can see the metabolic power over the hop cycle. We can integrate this and divide by hop time to get metabolic power. On the right panel, we see the metabolic power over multiple exoskeleton stiffness values from 0-K_0. We see that metabolic cost decreases for increasing exoskeleton stiffness.

3.4.2 Effect Of Exoskeletons on Stability

In the left panel of Figure 3.3, we see what happens when we change the height of the ground suddenly. When we drop the height of the ground, the mass continues to move through its parabolic path in the air beyond the previous ground height if the height of the ground is dropped, and touches the ground before it if the height of the ground is raised. Therefore, energy is added to the system when height is dropped. When the mass is in the stance phase, the muscle-tendon unit does work on the mass resulting in energy generation/dissipation in the first step. Over multiple hops we see that the system eventually dissipates this energy away, and we can measure the settling time - which is the time taken

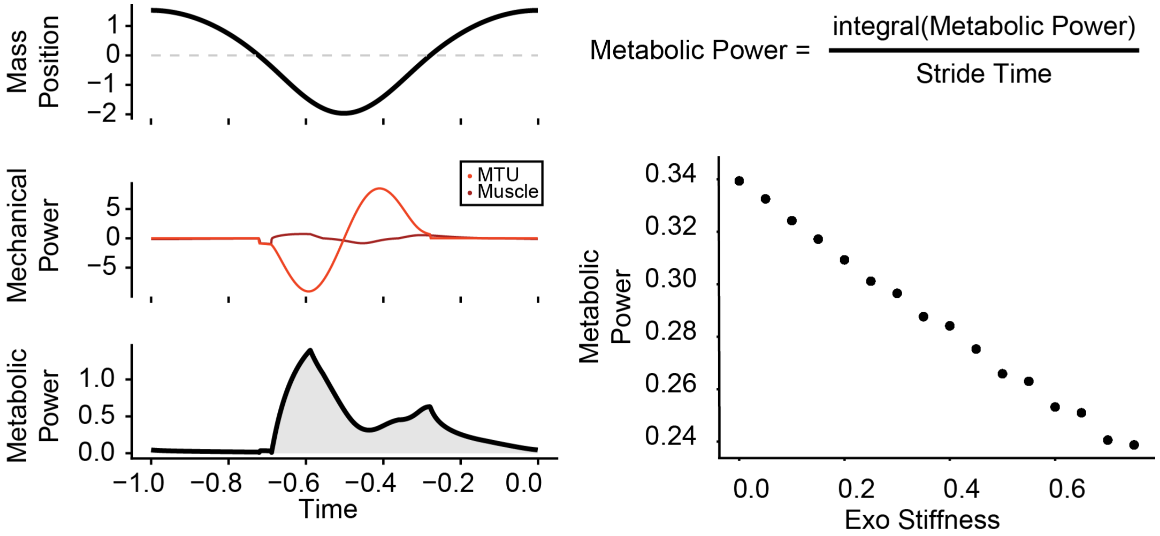

Figure 3.2: Mass position, mechanical power of muscle-tendon unit and muscle, and metabolic power of the muscle for one steady-state cycle. We can integrate the metabolic power over time and divide by cycle time to get average metabolic power. This panel shows the above variables for the no exoskeleton condition (right) Here we show the average metabolic power for a steady state cycle for different exoskeleton stiffnesses.

to dissipate 90% of the perturbation. We can measure this for various hop heights as shown in the left bottom panel, and find the average settling power of the system for each exoskeleton stiffness - where settling power is the average energy injected divided by the average settling time. When he look at the right panel of Figure 3.3, we see the average stability power of the muscle-tendon unit for each exoskeleton stiffness. Here, we see that increasing exoskeleton stiffness leads to increased settling power.

3.4.3 Effect Of Exoskeleton Stiffness On Agility

In the left panel of Figure 3.4, we see what happens when we change the timing of stimulation suddenly. When we delay the timing of stimulation, the mass is actuated passively as it reaches the bottom dead center and is stimulated on the way up, thus generating positive work. In the bottom panel, we can see the work done on the mass for various stimulation delay and measure the maximum and minimum work done for each exoskeleton stiffness. We call that the agility power of the system.

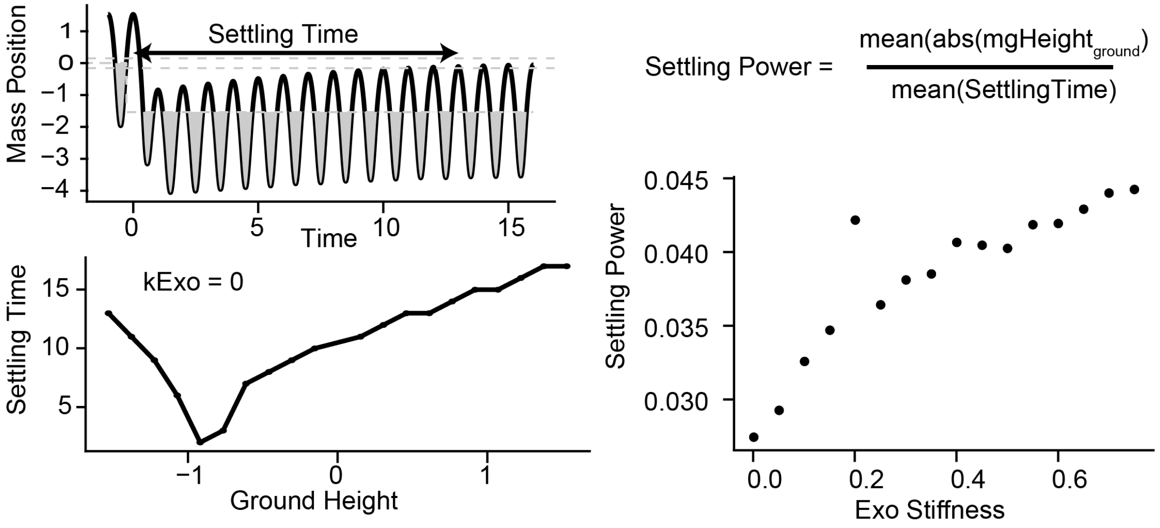

Figure 3.3: Mass position and ground height over time. Note how when the height of the ground is changed, the mass overdissipates energy in the first hop and over 13 hops comes back to steady state. We define the time the mass takes to come back to steady state as settling time (left, bottom) Here we show settling time for different ground height changes for the no exo condition. We take the average of energy injected by the height changes and divide it by the average settling time to get settling power. (right) Here we show settling power for various exoskeleton stiffness values. Note a settling power of 0 indicates the muscle-tendon unit is behaving spring-like

In the right panel of Figure 3.4, we look at the maximum negative power generated by the muscle-tendon unit as a function of exoskeleton stiffness. We see that increasing exoskeleton stiffness leads to higher positive and negative agility power.

3.4.4 Tradeoff Between Metabolic Cost, Agility and Stability

In Figure 3.5, we can see that there exists no tradeoffs for exoskeleton design. Increasing exoskeleton stiffness reduces metabolic power, increases settling power and increases agility power.

3.5 Discussion

In this study, we aimed to understand the tradeoffs in designing passive exoskeletons for augmenting stability, agility and metabolic cost of movement. We hypothesized that while passive exoskeletons would improve the metabolic cost of hopping due to lower demands

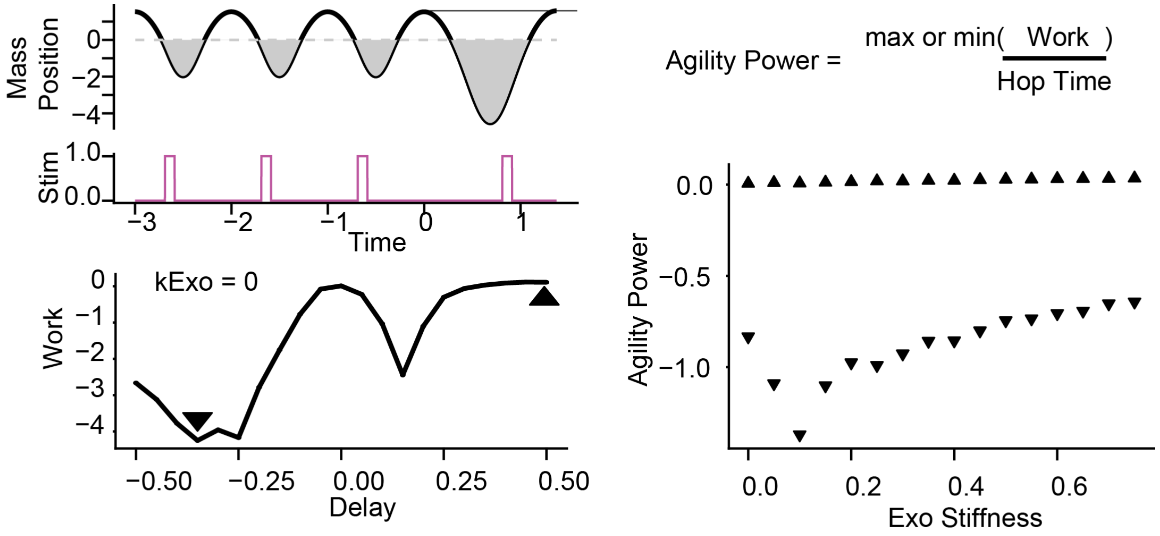

Figure 3.4: Mass position and stimulation over time. Note how when the timing of stimulation is changed, the muscle-tendon unit passively does negative work on the mass followed by active positive work that leads to increase in hop height. (left, bottom) Here we show the work done on the mass for different delays in stimulation timing. We denote the maximum positive work with an upright triangle and maximum negative work with a downwards triangle (right) Here we show maximum positive power and maximum negative power for various exoskeleton stiffness values. Note, an agility power of 0 indicates the muscle-tendon unit is behaving spring-like and cannot generate positive work.

on the muscle, they may also hamper agility and stability of movement due to increased contribution of a passive spring and reduced contribution of an active muscle.

3.5.1 Role Of Exoskeleton Stiffness In Determining Economy Of Movement

We found that increasing exoskeleton stiffness reduces metabolic cost. This is in agreement with previous studies that show exoskeleton stiffness reduces metabolic cost in both mathematical modeling [41] and experiments [58]. However, experiments in humans tends to show a U-shaped relationship with increasing exoskeleton stiffness for walking [55]. It is unclear if this is due to dynamics of hopping that is different from walking - especially in the hip joint that is not involved during hopping - or if this is a limitation with our model.

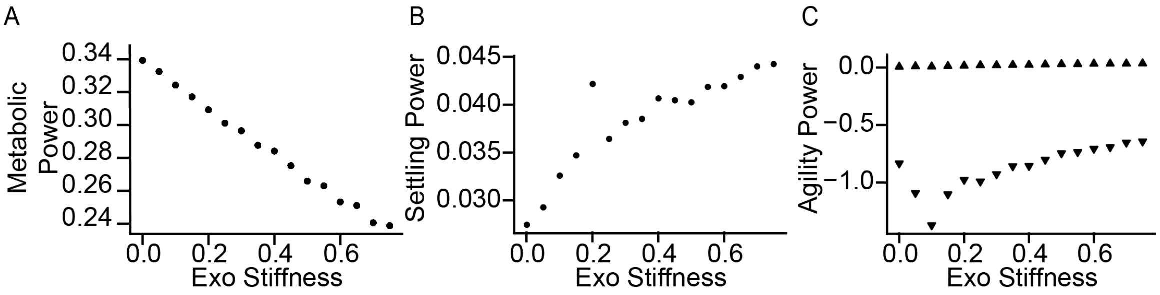

Figure 3.5: A) Metabolic power of a steady state cycle across exoskeleton stiffnesses (B) Stability power across exoskeleton stiffnesses (C) Positive and negative agility power. Note that metabolic power decreases, settling power decreases, positive agility power increases and negative agility power increases and then decreases with increasing exoskeleton stiffness.

3.5.2 Role Of Exoskeleton Stiffness In Determining Stability Of Movement

With regard to stability, we found that exoskeleton stiffness increases stability of hopping unilaterally - especially at large perturbations. This in interesting because although passive exoskeletons cannot themselves produce any work, they enable the underlying muscle-tendon unit to change behavior. While we are unaware of any studies that test this hypothesis directly, we think that this is because humans overdissipate energy when hopping on a movable subtract - especially under large perturbations [5]. Thus, adding an exoskeleton may reduce overdissipation by the muscle-tendon unit due to lower muscle activity, thus leading to higher stability.

3.5.3 Role Of Exoskeleton Stiffness In Determining Agility Of Movement

With regard to agility, we find that agility is higher with an exoskeleton than without is and is maximized by high exoskeleton stiffness for positive work and low exoskeleton stiffnesses for negative work. This is counter-intuitive because higher exoskeleton stiffnesses have lower muscle activations in our study, and we only changed the timing of muscle stimulation, not the magnitude. Thus, we would expect that higher exoskeleton stiffness would lead to lower agility due to lower force capacity in muscles. However, we do know that exoskeletons increase the range of muscle length and velocity during steady movement in hopping, thus perhaps allowing for a wider range of works and powers [41].

3.5.4 Economy, Agility, and Stability Tradeoffs In Passive Exoskeleton Design

However, the main question we were interested in was a question of tradeoff in exoskeleton design for the tree tasks. We find that passive exoskeletons not only reduce the metabolic cost of hopping, they also improve the stability and agility of movement. In addition, for metabolic cost, stability, and agility in terms of positive power, we find that higher the exoskeleton stiffness, higher the benefits. For agility in the negative power direction however, the benefit improves with increasing exoskeleton stiffness for small values of exoskeleton stiffness and subsequently reduces where the highest exoskeleton stiffness in our study had lesser agility in terms of negative power than the case without an exoskeleton. This means that exoskeleton designers may potentially face no trade-offs in designing passive exoskeletons. Thus, future studies should try to do experiments with passive exoskeletons in tasks other than economy and test our predictions.

3.5.5 Limitations

Our work has several potential caveats. For one, we used Hill-type models that are known to be good at predicting steady state movement and not unsteady movement like uneven terrain or agility. However, previous modeling work has shown similar behavior in Hill-type models as compared to experimental hopping [41] during steady state. In addition, we find that the trends of energy dissipation and generation for the no exoskeleton case are similar to those in human hopping, even though the speed at which they occur is different [5]. Thus, while our studies may not predict exact values, we believe simple models likely show similar trends to experimental results.

Another limitation of our work is how we compute agility. In our study, we limited our search space along similar stimulation profiles that varied timing to span the space. However, animals may be able to use a more rich trajectory space of stimulation patterns.

However, the question we are asking involves the trend across exoskeleton stiffness values, and we believe the limitations and constraints on muscle function given an exoskeleton stiffness will remain true even under different neural patterns. Thus while magnitudes of maximum and minimum power may change, we believe that the trends across exoskeleton stiffness will still hold true - and in fact may be further magnified as higher exoskeleton stiffnesses leads to larger amounts of untapped muscle force and activation that could be used.

3.6 Conclusion

In this study, we aimed to answer the question : How does exoskeleton stiffness tuning affect stability, agility, and economy of cyclic movement? We showed that passive exoskeletons are able to minimize metabolic cost, increase stability and increase agility despite being unable to generate power itself. It does so by influencing the underlying muscle-tendon behavior. Future studies should try to test these predictions experimentally in the lab. In addition, future studies should look at how variations in internal morphology caused by aging or different routines affects exoskeleton stiffness choices.

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