Influence of Parallel Spring-Loaded Exoskeleton on Ankle Muscle-Tendon Dynamics During Simulated Human Hopping

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Abstract—**Robotic assistance for rehabilitation and enhancement of human locomotion has become a major goal of biomedical engineers in recent years. While significant progress to this end has been made in the fields of neural interfacing and control systems, little has been done to examine the effects of mechanical assistance on the biomechanics of underlying muscle-tendon systems. Here, we model the effects of mechanical assistance via a passive spring acting in parallel with the triceps surae-Achilles tendon complex during cyclic hopping in humans. We examine system dynamics over a range of biological muscle activation and exoskeleton spring stiffness. We find that, in most cases, uniform cyclic mechanical power production of the coupled system is achieved. Furthermore, unassisted power production can be reproduced throughout parameter space by trading off decreases in muscle activation with increases in ankle exoskeleton spring stiffness. In addition, we show that as mechanical assistance increases the biological muscle-tendon unit becomes less 'tuned,' resulting in higher mechanical power output from active components of muscle despite large reductions in required force output.**

INTRODUCTION

Years of research on the mechanics and energetics of locomotion have established that compliant tissues (i.e. tendon and aponeurosis) are crucial in shaping efficient and stable locomotion [1]. Achieving a 'tuned' state in a compliant muscle-tendon (MT) system depends on optimal interaction between the frequency and amplitude of muscle activation, the material properties of series elastic tissues, and the fundamental properties of the contractile element (activation dynamics, force-length, and force-velocity). When MT interaction dynamics are optimally 'tuned', series elastic tissues stretch and recoil accounting for much of the overall MT length change. Maximizing elastic energy storage and return allows in-series muscles to remain nearly isometric, reducing metabolic demand with little effect on overall MT power output [2]. It is not surprising, then, that humans prefer to use neuromuscular control strategies during cyclic movements that allow their muscles to operate in a nearly isometric state when coupled with a series elastic tendon [3].

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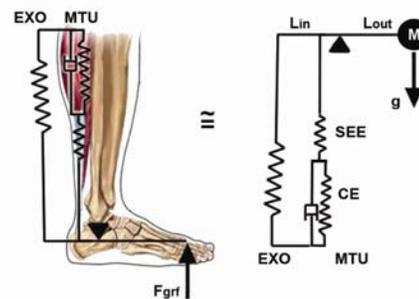
Figure 1. The basic model used for MTU-Exo simulations. It consists of a Hill-type muscle (CE) in series with a Hookean spring (SEE) and in parallel with a second Hookean spring (Exo) operating across a lever arm on the center of mass. This is meant to reflect triceps surae-Achilles tendon interaction mechanics in the presence of passive mechanical assistance from a spring-loaded ankle exoskeleton.

In recent years, there has been rapid progress in the development of wearable robotics designed to assist/enhance human movement [4, 5]. Despite recent technological advances, few studies have examined the effects of parallel mechanical assistance on underlying MT interaction dynamics. Because it is difficult to make direct measurements of in-vivo MT behavior, a modeling approach holds great promise in addressing key questions regarding human physiological response to wearable robotic assistance. The purpose of this study was two-fold: 1) To develop a modeling framework to study assisted human hopping and 2) To employ the model to examine how the presence of parallel mechanical assistance influences MT interaction dynamics.

We hypothesized that: 1) The coupled spring-loaded exoskeleton (Exo)-compliant MT system can produce periodic mechanical power output by trading off increased Exo stiffness with decreased muscle activation and 2) Muscle contractions will remain nearly isometric (i.e. 'tuned').

METHODS

To investigate the effects of mechanical assistance on compliant muscle-tendon interaction dynamics, we developed a simple mathematical model of the human triceps surae muscle-tendon unit (MTU) working in parallel with a passive Hookean spring (Fig. 1). The biological MTU in this model is comprised of a Hill-type muscle, or contractile element (CE) in series with a Hookean tendon-spring, or series elastic element (SEE) [6] operating across a lever on a mass under constant gravitational load. We based our muscle (maximum active force F_{max} = 6000 N, maximum CE velocity V_{max} = .45 m/s, L_0 = .055 m, k_{CE} = 90,000 N/m) and tendon (k_{SEE} = 180,000 N/m, slack length l_{slack} = .237 m)

properties on data documented for the human triceps surae muscle-tendon complex [7]. We chose parameters for the load (M = 35 kg, in/out lever arm length ratio ~.33) to reflect realistic body weight and mechanical advantage seen at the ankle joint of a single limb during two-legged hopping. To drive muscle force generation, we used a feed-forward neural control signal with no reflex feedback. This was achieved by stimulating the modeled muscle-tendon system over a range of muscle activations at a single driving frequency (2.2 Hz). We chose this driving frequency because it "tuned" the muscle (i.e. caused it to contract nearly isometrically) at full activation in the absence of the parallel ankle Exo. This parameter set and driving frequency are consistent with preferred behavior observed in human hopping studies [8, 9].

In order to model the effects of a wearable passive Exo at the ankle during hopping, we provide parallel assistance with a simple linear spring (k_{Exo}) that has a slack length equal to MTU length at touchdown in the aforementioned tuned state ($l_{slack\ Exo}$ = .306 m). To explore the interplay between muscle activation and parallel spring stiffness, we ran 10 second (22 period) simulations. The range of assistance (k_{Exo}) provided spanned 0-100% of the effective stiffness for the purely passive MTU (k_{MTU} = 60,000 N/m), and muscle activation ranged from 10-100%. By varying both activation amplitude and the level of mechanical assistance in our coupled Exo-MTU system, we were able to explore how the addition of a passive linear spring in parallel with the human triceps surae-Achilles tendon complex affected the naturally efficient mechanics of the MTU.

<div align="center">RESULTS</div>

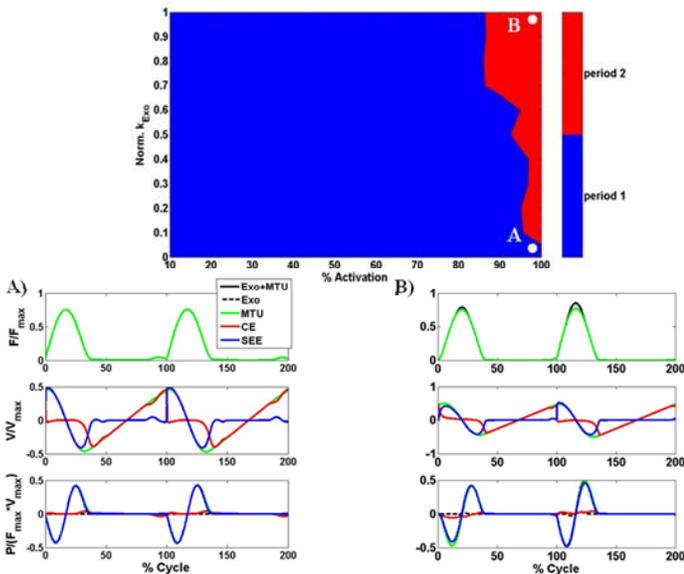For every simulation we ran, uniform cyclic behavior was established within ~12-15 cycles of simulated hopping. The

Figure 2. Top: A contour plot of period 1 and 2 behavior throughout the muscle activation-Exo spring stiffness parameter space. **Bottom**: MTU-Exo period 1 and 2 dynamics for points A (left) and B (right) in the top panel plotted for two consecutive hopping cycles.
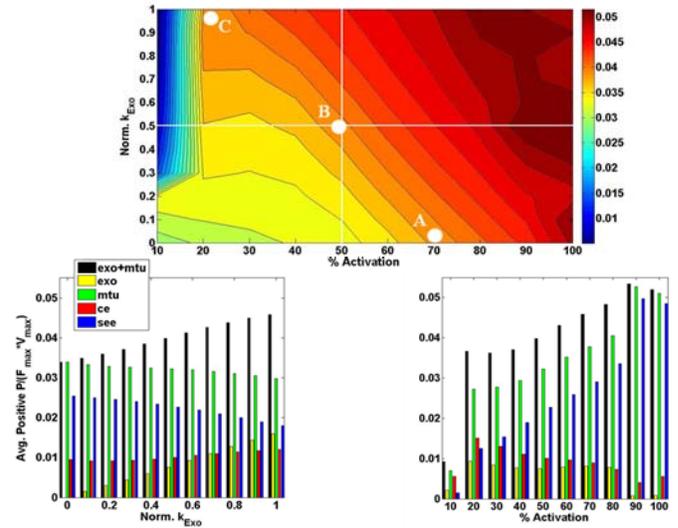
Figure 3. Top: Contour plot of MTU-Exo average positive mechanical power production over a single hopping cycle. **Bottom:** Mechanical power is broken down into components along the lines k_{Exo} = 30,000 N/m = .5*k_{MTU} and muscle activation = 50%. Average power values are normalized to $F_{max}*V_{max}$ of the CE.

overwhelming majority of the muscle activation-k_{Exo} parameter space exhibited period 1(repeated each activation cycle) behavior. We observed some period 2 (repeated behavior every 2 cycles) behavior in regions of high muscle activation with k_{Exo} greater than zero (Fig. 2). Period 2 behavior is most clearly exhibited in patterns of CE and SEE mechanical power output. For example, the CE absorbs net energy in one cycle, and generates it in the next. This in turn influences the amount of energy stored and returned by the SEE in consecutive hopping cycles. Assuming humans prefer to hop with period 1 behavior, we chose not to examine any aspects of the period 2 Exo-MTU dynamics.

Baseline (i.e. unassisted, k_{Exo}=0) average mechanical power production of the coupled Exo-MTU system over the course of a hopping cycle can be maintained by reducing magnitude of muscle activation while increasing k_{Exo} (Fig. 3, top). This invariant behavior is exhibited along contours moving from the bottom right hand corner (high muscle activation, low k_{Exo}) to the top left corner (low muscle activation, high k_{Exo}) of parameter space. These contours end abruptly at 20% muscle activation and k_{Exo} = .2k_{MTU}, indicating that there is a minimum threshold of activation required to effectively utilize a passive ankle exoskeleton. Contours of coupled Exo-MTU mechanical power output are also non-uniform in the case of high activation and k_{Exo} greater than zero. This region has already been rejected as a viable operating point for the Exo-MTU system because of its period 2 dynamics.

As k_{Exo} increases, so does the overall mechanical power output of the coupled Exo-MTU system (Fig. 3, bottom left). Additionally, as k_{Exo} increases, the average positive mechanical power contributed by both the MTU and SEE decreases, while that of the CE *increases* slightly. In other words, as k_{Exo} increases and muscle activation is held constant, the mechanical demands on the CE are elevated
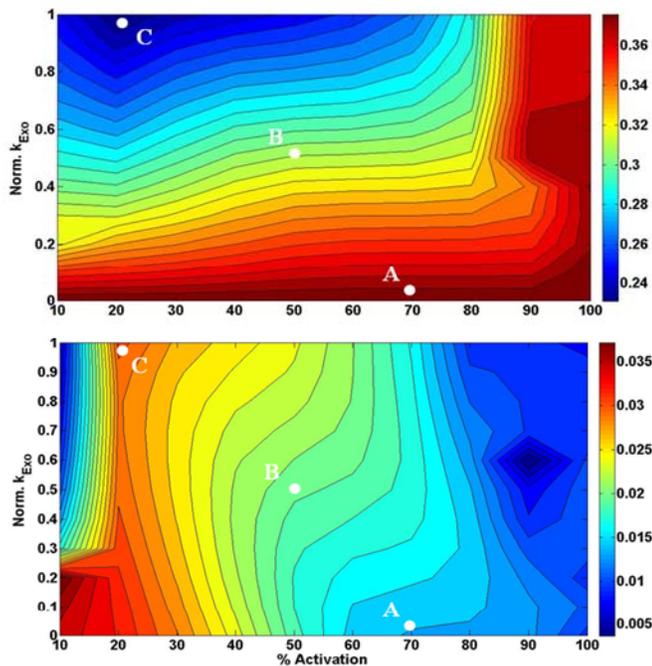
Figure 4. Top: Average force-impulse produced by the MTU over a single hopping cycle. Values are normalized to F_{max} of the CE. **Bottom:** Average positive CE mechanical power over a single hopping cycle. Values are normalized to $F_{max}*V_{max}$ of the CE.

(Figs. 3, 4, and 5).

When muscle activation increases so do average positive power contributions from the MTU and SEE. Meanwhile, CE and Exo power outputs show a generally decreasing trend with increases in activation (Fig. 3, bottom right). In other words, the greater the activation, the less the CE and Exo contribute to average positive mechanical work, and the more 'tuned' the MTU system becomes (Fig. 3, 4, and 5).

DISCUSSION

The total mechanical power output of the coupled Exo-MTU system depends heavily on both k_{Exo} and CE activation. Notably, system mechanical power output is nearly twice as sensitive to muscle activation as it is to Exo stiffness, as indicated by slope of the contour passing through points A,

B and C (Fig. 3). Thus, small changes in motor control (i.e. CE activation amplitude) could have large effects on total power output of the coupled Exo-MTU system.

The relative sensitivity of the coupled Exo-MTU system to CE activation amplitude has clear benefits that can be exploited to achieve optimal mechanical assistance. As hypothesized, by simply decreasing CE activation, it is possible to maintain uniform, cyclic positive mechanical power output of the coupled Exo-MTU system in the face increasing Exo spring stiffness (k_{Exo}).

Trading off decreased CE activation for increased k_{Exo} not only results in invariant mechanical power production, but also appreciable sharing of force generation between the artificial (i.e. the ankle Exo springs) and biological (the ankle MTU) components of the system. This force sharing is significant, reducing loads on the MTU by up to 33% (Fig. 4, bottom, point A vs. C). Furthermore, over a large range of CE activations (~20%-80%), even small increases in Exo stiffness (k_{Exo}) can significantly reduce force requirements on the MTU (Fig. 4, top).

At face value, the reduced force requirements on biological MTUs afforded by a parallel ankle Exo ($k_{Exo} >0$) would seem to have clear performance benefits (e.g. reduced fatigue and metabolic energy expenditure). While a passive ankle Exo reduces CE activation and the force requirements of the MTU during hopping, it also drastically increases the CE mechanical power output (Fig. 4, bottom).

In order to more closely examine this phenomenon, we selected three points along a contour of equal Exo-MTU mechanical power output ranging from a CE activation of 70% and k_{Exo} of zero to CE activation of 20% and k_{Exo} equal to k_{MTU} (Figs. 3, 4, and 5). As Exo mechanical assistance increases and CE activation decreases, we observe a significant shift in phase between onset of activation and time of peak force (Fig. 5). For point A activation is nearly coincident with force onset. At point C, however, the onset of activation occurs much later with respect to force onset; and is in fact nearly coincident with peak force. This altered phasing disrupts the optimal tuning of the MTU. As a result, the CE no longer contracts isometrically, and elastic energy cycled in the SEE is significantly reduced (Fig. 5).

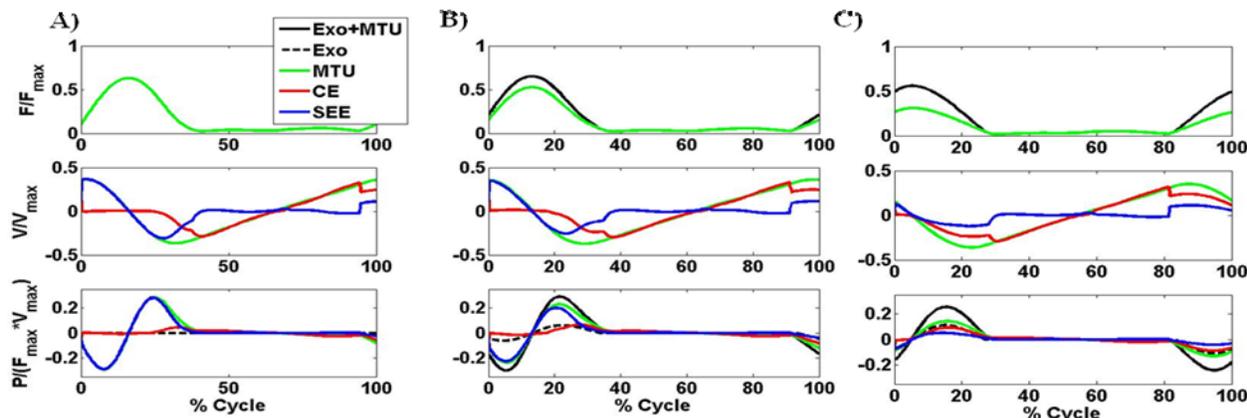The increase in CE positive work with increasing Exo

Figure 5. Force, velocity, and mechanical power for each component of the MTU-Exo system plotted over the course of a single hopping cycle beginning with activation onset. Plots A, B, and C correspond to the points A, B, and C spanning parameter space highlighted in Figures 3 and 4. Force is normalized to F_{max}, velocity to V_{max}, and power to $F_{max}*V_{max}$ of the CE.

mechanical assistance may at first seem counter-intuitive. There are two factors that can influence the amount of positive mechanical work done by muscle during active force production- magnitude of force and overall shortening. We can rule out an increase in force as a potential mechanism, as MTU force requirements are reduced by 30% moving from point A to C in parameter space (Fig. 4, top). That leaves length change as the underlying cause of increased mechanical work requirements on the CE. Increases in CE excursion during active force production can either be the result of greater velocities during shortening, or an earlier onset of shortening for a given velocity. We observed no shift in the operating point on the force-velocity curve as exoskeleton assistance increased (Fig. 6, bottom). We do, however, observe that with higher levels of exoskeleton mechanical assistance there is an earlier onset of CE shortening, resulting in longer shortening times, increased excursion, and significant increases in CE mechanical work (Figs. 5 and 6, top). In summary, despite a 30% reduction in MTU force-impulse, there is a 260% *increase* in CE excursion that leads to an almost 100% increase in CE positive mechanical work as Exo spring stiffness increases from zero to $1.0*k_{MTU}$.

CONCLUSIONS

To summarize, it is possible to maintain a baseline level of mechanical power production with the addition of a passive spring in parallel with the human ankle. This can be achieved by reducing muscle activation in conjunction with increased Exo spring stiffness (k_{Exo}). This leads to three key alterations in the dynamics of the biological MTU: 1) A decrease in force production for the MTU 2) An increase in the operating length and positive mechanical work for the CE and 3) A decrease in the amount of elastic energy cycled in the SEE. In essence, though mechanical assistance reduces force requirements, it disrupts optimal 'tuning' within the MTU. Disrupted 'tuning' may have deleterious consequences for metabolic cost of mechanically assisted hopping. Although their relative contributions are unclear, both force and work demands influence metabolic energy consumed by the CE. The uncertain relationship between CE power/force production and metabolic cost makes it difficult to predict which combination of CE activation and k_{Exo} will minimize metabolic cost.

FUTURE DIRECTIONS

The modeling framework presented here provides the foundation for future in-vivo studies of the human triceps surae-Achilles tendon complex during hopping with passive mechanical assistance. We will combine real-time ultrasound measurements of muscle fascicle length change with inverse dynamics analysis and indirect calorimetry to test our model's predictions. Comparing model predictions to experimental data will provide insights into two crucial questions in the field of wearable robotics: 1) How do relative contributions of force and work contribute to metabolic energy expenditure of the user? 2) What are the

relative roles of feed-forward vs. feedback mechanisms in shaping human neuromechanical adaptation to passive mechanical assistance?

Future work to extend the current modeling framework will include the addition of: 1) force, length, and velocity dependent reflex feedback pathways and 2) active modes of parallel mechanical assistance (e.g. a myoelectrically controlled pneumatic actuator).
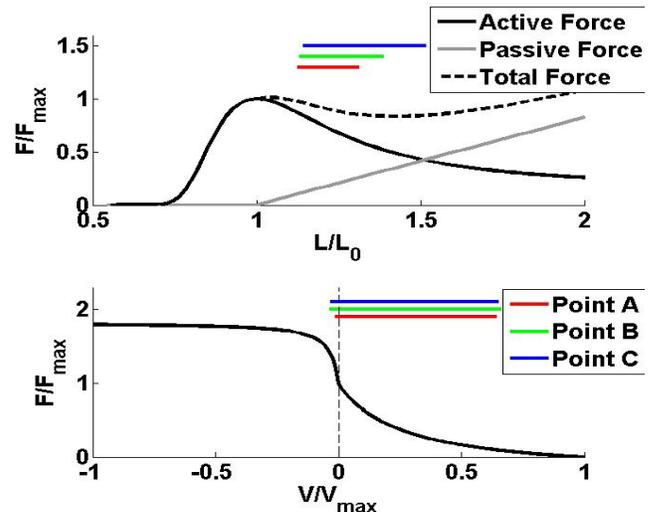

Figure 6. Top: Plot of the normalized force-length for the CE. Colored bars represent CE excursion during active force production for points A, B, and C in parameter space. **Bottom:** Plot of the normalized force-velocity for the CE. Colored bars indicate the range of operating velocities of the CE during active force production.

REFERENCES

[1]   R. M. Alexander, "Tendon elasticity and muscle function," *Comp Biochem Physiol A Mol Integr Physiol,* vol. 133, pp. 1001-11, Dec 2002.

[2]   T. J. Roberts and E. Azizi, "Flexible mechanisms: the diverse roles of biological springs in vertebrate movement," *J Exp Biol,* vol. 214, pp. 353-61, Feb 1 2011.

[3]   D. Takeshita, *et al.*, "Resonance in the human medial gastrocnemius muscle during cyclic ankle bending exercise," *J Appl Physiol,* vol. 101, pp. 111-8, Jul 2006.

[4]   D. P. Ferris, "The exoskeletons are here," *J Neuroeng Rehabil,* vol. 6, p. 17, 2009.

[5]   H. Herr, "Exoskeletons and orthoses: classification, design challenges and future directions," *J Neuroeng Rehabil,* vol. 6, p. 21, 2009.

[6]   F. E. Zajac, "Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control," *Crit Rev Biomed Eng,* vol. 17, pp. 359-411, 1989.

[7]   G. A. Lichtwark and A. M. Wilson, "Optimal muscle fascicle length and tendon stiffness for maximising gastrocnemius efficiency during human walking and running," *J Theor Biol,* vol. 252, pp. 662-73, Jun 21 2008.

[8]   D. P. Ferris, *et al.*, "Neuromechanical adaptation to hopping with an elastic ankle-foot orthosis," *J Appl Physiol,* vol. 100, pp. 163-70, Jan 2006.

[9]   A. M. Grabowski and H. M. Herr, "Leg exoskeleton reduces the metabolic cost of human hopping," *J Appl Physiol,* vol. 107, pp. 670-8, Sep 2009.