

# A Model of Muscle-Tendon Function in Human Walking

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*Abstract*— In this paper, we study the mechanical behavior of leg muscles and tendons during human walking in order to motivate the design of economical robotic legs. We hypothesize that quasi-passive, series-elastic clutch units spanning the knee joint in a musculoskeletal arrangement can capture the dominant mechanical behaviors of the human knee in level-ground walking. Since the mechanical work done by the knee joint throughout a level-ground self-selected-speed walking cycle is negative, and since there is no element capable of dissipating mechanical energy in the musculoskeletal model, biarticular elements would necessarily need to transfer energy from the knee joint to hip and/or ankle joints. This mechanism would reduce the necessary actuator work and improve the mechanical economy of a human-like walking robot. As a preliminary evaluation of these hypotheses, we vary model parameters, or spring constants and clutch engagement times, using an optimization scheme that minimizes ankle and hip actuator positive work while still maintaining human-like knee mechanics. For model evaluation, kinetic and kinematic gait data were employed from one study participant walking across a level-ground surface at a self-selected gait speed. With this under-actuated leg model, we find good agreement between the model's quasi-passive knee torque and experimental knee values, suggesting that a knee actuator is not necessary for level-ground robotic ambulation at self-selected gait speeds.

## I. INTRODUCTION

In order to design mechanically economical, low-mass leg structures for robotic, exoskeletal and prosthetic systems, designers have often employed passive and quasi-passive components [1-5, 8-11]. In this paper, a quasi-passive device refers to any controllable element that cannot apply a non-conservative, motive force. Quasi-passive devices include, but are not limited to, variable-dampers, clutches, and combinations of variable-dampers/clutches that work in conjunction with other passive components such as springs.

The use of quasi-passive devices in leg prostheses has been the design paradigm for over three decades, resulting in leg systems that are lightweight, energy efficient, and operationally quiet. In the 1970's, Professor Woodie Flowers at MIT conducted research to advance the prosthetic knee joint from a passive, non-adaptive mechanism to an active

device with variable-damping capabilities [1]. Using Flowers' knee, the amputee experienced a wide range of knee damping values throughout a single walking step. During ground contact, high knee damping inhibited knee buckling, and variable damping throughout the swing phase. This allowed the prosthesis to swing before smoothly decelerating prior to heel strike. Motivated by Flowers' research, several research groups developed computer-controlled, variable-damper knee prostheses that ultimately led to commercial products [2-5]. Actively controlled knee dampers offer clinical advantages over mechanically passive knee designs. Most notably, transfemoral amputees walk across level ground surfaces and descend inclines/stairs with greater ease and stability [6,7].

Quasi-passive devices have also been employed in the design of economical bipedal walking machines and legged exoskeletons. Passive dynamic walkers [8] have been constructed to show that bipedal locomotion can be energetically economical. In such a device, a human-like pair of legs settles into a natural gait pattern generated by the interaction of gravity and inertia. Although a purely passive walker requires a modest incline to power its movements, researchers have enabled robots to walk across level ground surfaces by adding just a small amount of energy at the hip or the ankle joint [9, 10]. In the area of legged exoskeletal design, passive and quasi-passive elements have been employed to lower exoskeletal weight and to lower system energy usage. In numerical simulation, Bogert [11] showed that an exoskeleton using passive elastic devices can, in principle, substantially reduce muscle force and metabolic energy usage in walking. Walsh et al. [12] built an under-actuated, quasi-passive exoskeleton designed for load-carrying augmentation. During level-ground walking, the exoskeleton only required 2 Watts of electrical power for its operation with on average 80% load transmission through the robotic legs.

Although passive and quasi-passive devices have been exploited to improve overall system economy in legged systems, the resulting structures failed to truly mimic human-like mechanics in level-ground ambulation. In this paper, we seek to understand how leg muscles and tendons work mechanically during walking in order to motivate the design of economical, low-mass robotic legs. We hypothesize that quasi-passive, series-elastic clutch units spanning the knee joint in a musculoskeletal arrangement can capture the dominant mechanical behaviors of the human knee in level-ground walking. Since the human knee performs net negative work throughout a level-ground walking cycle [13], and since a series-elastic clutch is incapable of dissipating mechanical energy as heat, a corollary to this hypothesis is

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that such a quasi-passive robotic knee would necessarily have to transfer energy via elastic biarticular mechanisms to hip and/or ankle joints. Such a transfer of energy would reduce the necessary actuator work to track human torque profiles at the hip and ankle, improving the mechanical economy of a human-like walking robot. As a preliminary evaluation of these hypotheses, we put forth an under-actuated leg model that captures the gross features of the human leg musculoskeletal architecture. We vary model parameters, or spring constants and clutch engagement times, using an optimization scheme that minimizes ankle and hip actuator positive work while still maintaining human-like knee mechanics.

## II. METHODOLOGY

### A. Leg Model

Figure 1 shows a two-dimensional musculoskeletal model of the human leg comprising nine series-elastic clutch/actuator mechanisms. The model was derived by inspection of the human musculoskeletal architecture. Our previous study [14] revealed that, for a self-selected walking speed, hip and ankle actuators are required to provide enough positive mechanical work at the hip and ankle joints, while the quasi-passive elements capture the dominant self-selected walking behavior at the knee joint. Therefore, the hip monoarticular, extensor/flexor units and the ankle monoarticular, plantar flexor unit are the only active components of the leg model with unidirectional actuation capable of producing net work. The remaining muscle-tendon units are modeled as series-elastic clutches. For each of these quasi-passive units, when a clutch is disengaged, joints rotate without any resistance from the series spring. Once a clutch is engaged, it holds the series elastic spring at its current position, and the spring begins to store energy as the joint rotates, in a manner comparable to a muscle-tendon unit where the muscle generates force isometrically. The model comprises six monoarticular and three biarticular tendon-like springs with series clutches or actuators. It is noted that the ankle-knee posterior unit and the ankle plantar flexor both share the same distal tendon spring (See Fig. 1). Hip, knee and ankle joints have agonist/antagonist pairs of monoarticular springs with series clutches or actuators. Further, the leg model includes two knee-hip and one ankle-knee biarticular units. The knee-hip anterior unit works as an extensor at the knee joint and as a flexor at the hip, while the knee-hip posterior unit works as a flexor at the knee joint and as an extensor at the hip. We assume that all monoarticular units are rotational springs and clutches, while all biarticular units act around attached pulleys with fixed moment arm lengths. The moment arms for biarticular units are taken from the literature. [15, 16]

### B. Optimization

The model has a total of 15 series-elastic clutch parameters:

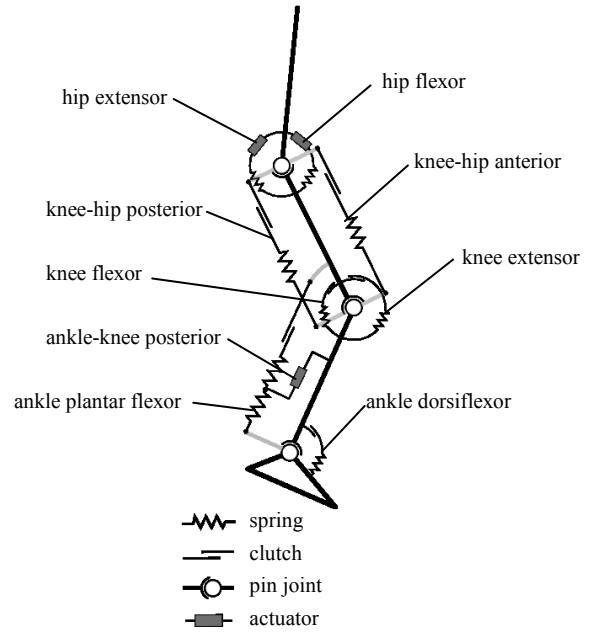


Fig. 1. Three-actuator leg model. Only three actuators act about the model's ankle and hip joints. Series-elastic clutch units span the model's ankle and knee in a musculoskeletal arrangement.

nine spring constants and six distinct times when clutches are engaged and disengaged. Fitting this model to biomechanical data involved determining the spring rates of the model's nine springs and the engagement and disengagement times of the six associated clutches.

The biomechanical data for this procedure was obtained for one participant using a motion capture system and force platforms. The data collection procedure is described in the next section.

An optimization procedure was used to fit the parameters of the model to the biomechanical data. The cost function  $f(\mathbf{x})$  is as follows:

$$f(\mathbf{x}) = a \sum_j \left( \sum_{i=1}^{100} \left( \frac{\tau_{bio}^{ij} - \tau_{sim}^{ij}}{\tau_{j,max}^{bio}} \right)^2 + W_{act}^+ \right) \quad (1)$$

where  $\mathbf{x}$  is a vector comprising the 15 parameters mentioned previously;  $\tau_{bio}^{ij}$  and  $\tau_{sim}^{ij}$  are the angular torques applied around joint  $j$  at the  $i$ th percentage of the gait cycle from the biological data and the model, respectively; and  $\tau_{j,max}^{bio}$  is the maximum biological torque at joint  $j$ . Further,  $W_{act}^+$  is the total positive mechanical work performed by the three muscle actuators during the gait cycle. Finally,  $a$  is a constant weighting coefficient.

The cost function  $f(\mathbf{x})$  was minimized subject to the constraint that the net torque at the hip and ankle equaled the target biological torque determined through inverse dynamics. Namely, the net hip model torque is constrained to be equal to the biological hip torque derived from an inverse dynamics calculation. Similarly, the net ankle model torque is also constrained to be equal to the biological ankle torque. Since only one unidirectional muscle acts around the ankle joint, this constraint holds only when the net ankle torque  $\tau_{ankle}$  is positive (plantar flexion torque). In summary, the hip and

ankle constraints force the model to precisely track the biological hip torque throughout the gait cycle, as well as the biological ankle torque when the net ankle model torque is positive.

Since the model includes biarticular units capable of transferring energy between joints, the value for the weighting coefficient  $a$  in the cost function (1) influences the model's capacity to emulate the biological knee torque with quasi-passive elements. Therefore, we selected a weighting coefficient  $a$  (set equal to 50) that was sufficiently large such that further increases in  $a$  did not result in larger  $R^2$  values at the knee joint. Here  $R^2$  values were computed to quantify the level of agreement between model and biological data at the knee joint for torque and power curves (See equation 2).

The determination of the desired global minimum for this objective function was implemented by first using a genetic algorithm to find the region containing the global minimum, followed by the use of an unconstrained gradient optimizer (fminunc in Matlab) to determine the exact value of that global minimum.

### C. Data Collection and Analysis

Biological data, including kinetic and kinematic walking information, were collected at the Gait Laboratory of Spaulding Rehabilitation Hospital, Harvard Medical School, in a study approved by the Spaulding Committee on the Use of Humans as Experimental Participants. A healthy adult participant (male, age 22, body mass 73.9kg, self-selected walking speed 1.2m/s) volunteered for the study. The participant walked at a self-selected speed across a 10m walkway in the Motion Analysis Laboratory. The participant was timed between two fixed points to ensure that the same walking speed was used in experimental trials. Walking speeds within a  $\pm 5\%$  interval from the self-selected speed were accepted.

A total of six walking trials were conducted. The data collection procedures were based on standard techniques [13, 17-19]. Prior to modeling and analysis, all marker position data were low pass filtered using a 4<sup>th</sup> order digital Butterworth filter at a cutoff frequency of 8Hz. The filter frequency was based on the maximum frequency obtained from a residual analysis of all marker position data, and processed as one whole gait cycle with 100 discrete data points from the heel strike to the next heel strike of the same leg.

### D. Model Evaluation

#### 1) Model Joint Torque and Power Agreement

To evaluate the level of agreement between experimental data and optimization results, we used the coefficient of determination,  $R^2$ , where  $R^2=1$  only if there is a perfect fit and  $R^2=0$  indicates that the model's estimate is worse than using the mean experimental value as an estimate. More specifically,  $R^2$  was defined as

$$R^2 = 1 - \frac{\sum_{i=1}^{100} (x_{bio}^i - x_{sim}^i)^2}{\sum_{i=1}^{100} (x_{bio}^i - \bar{x}_{bio})^2} \quad (2)$$

where  $x_{bio}^i$  and  $x_{sim}^i$  are the biological and simulated data at the  $i$ th percentage gait cycle. Here  $\bar{x}_{bio}$  is the grand mean over all walking trials and gait percentage times analyzed, or

$$\bar{x}_{bio} = \frac{1}{N_{data} N_{trial}} \sum_{j=1}^{N_{trial}} \sum_{i=1}^{N_{data}} x_{bio}^{ij} \quad (3)$$

where  $N_{data}$  and  $N_{trial}$  are the number of data points and trials, respectively, and  $x_{bio}^{ij}$  is the biological data at the  $i$ th percentage gait cycle for the  $j$ th trial.

#### 2) Mechanical Economy

In this investigation, we used a metric called the mechanical economy as a measure of actuator work requirements in walking. The mechanical economy  $c_{mt}$  [19] is a dimensionless number defined as

$$c_{mt} = \frac{W_{act}^+}{MgL_{leg}} \quad (4)$$

where  $M$  is the total mass of the participant,  $g$  is the acceleration due to gravity, and  $L_{leg}$  is the participant's leg length. Further,  $W_{act}^+$  is the nonconservative, total positive mechanical work performed by the three muscle actuators during the gait cycle.

#### 3) Whole-body Mechanical Energetics

Willems et al. [20] reported that the total mechanical energy of the body can be estimated by subdividing the body into multiple rigid segments and computing their gravitational potential and kinetic energies. Such an assertion assumes, of course, that little energy is stored elastically within muscle, tendon and ligaments. However, there is mounting in vivo evidence (e.g. see Ishikawa et al. [21]) that considerable amounts of elastic energy are stored during walking. In this investigation, the total whole body mechanical energy  $E_{tot,wb}$  was calculated from the gravitational potential energy  $E_{pot}$ , the kinetic energy  $E_k$  and the elastic energy storage  $E_e$  as predicted from the leg model, or

$$E_{tot,wb} = E_{pot} + E_k + E_e \\ = Mg h_{cm} + \frac{1}{2} M v_{cm}^2 + \sum_{i=1}^n \left( \frac{1}{2} m_i v_{r,i}^2 + \frac{1}{2} m_i K_i^2 w_i^2 \right) + \frac{1}{2} \sum_{i=1}^9 k_i \Delta x_i^2 \quad (5)$$

The quantities in equation 5 are defined as follows:  $h_{cm}$  and  $v_{cm}$  are the height and linear velocity of the whole-body center of mass (CM);  $v_{r,i}$  is the linear velocity of the CM of the  $i$ th segment relative to the whole-body CM;  $K_i$  and  $w_i$  are the radius of gyration and the angular velocity of the  $i$ th segment around its CM;  $m_i$  is the mass of  $i$ th segment; and  $k_i$  and  $\Delta x_i$  are the spring stiffness and displacement of the  $i$ th model spring, respectively. Here  $h_{cm}$  and  $v_{cm}$  are calculated from the position of the whole-body CM. The position of the whole-body CM was estimated using

$$\vec{r}_{cm} = \frac{\sum_{i=1}^n m_i \vec{r}_{cm}^i}{M} \quad (6)$$

where  $\vec{r}_{cm}^i$  is the position vector to the CM of  $i$ th segment relative to the whole-body CM. In equation 5,  $h_{cm}$  is the vertical component of  $\vec{r}_{cm}$ , and  $v_{cm}$  is the Euclidean norm of the differentiated  $\vec{r}_{cm}$ .

The leg model presented here enables an estimate of elastic energy storage and the percentage recovery between mechanical energies (elastic, potential, kinetic) of the body in walking. Percentage recovery (*REC*) between elastic energy and potential/kinetic energies is defined as

$$REC = \frac{|W_{pk}| + |W_e| - |W_{tot}|}{|W_{pk}| + |W_e|} \times 100 \quad (7)$$

where  $W_{pk}$  is the sum of the positive increments of potential plus kinetic energy curve,  $W_e$  is the sum of the positive increments of the elastic energy curve, and  $W_{tot}$  is the sum of positive increments of both curves in one complete walking cycle. The percentage recovery is 100 percent when the two energy curves are exactly equal in shape and amplitude, but opposite in phase, or when the total mechanical energy of the body does not vary in time.

### III. RESULTS

#### 1) Model Joint Torque and Power Agreement

Figure 2 shows biological data and simulation results of ankle and knee joints: (a) ankle torque, (b) ankle power, (c) knee torque, and (d) knee power. Since the model includes an agonist/antagonist pair of actuators spanning the hip joint, hip torque and power curves precisely track biological data. Although there is only one unidirectional actuator about the ankle joint, both  $R^2$  values for ankle torque and power are equal to one. This is because, during most of the stance phase, ankle torque is positive (20% to 100% of the stance phase), where the ankle plantar flexor is dominant. For the knee joint, the simulation result shows a  $R^2$  value of 0.98 for both knee torque and power curves, even though there are only quasi-passive units spanning the joint.

#### 2) Biarticular Energy Transfer

Since the mechanical work done by the knee joint throughout a level-ground self-selected-speed walking cycle is negative [13], and since there is no unit capable of dissipating mechanical energy in the musculoskeletal model, the biarticular elements of the model would necessarily need to transfer energy from the knee joint to hip and/or ankle joints. In the model, a small amount of energy was transferred from knee to ankle (2.6 Joules) by the biarticular ankle-knee posterior unit, but a much greater amount of energy was transferred from knee to hip (10 Joules) by the biarticular knee-hip anterior and posterior units. In the leg model these biarticular units that transfer energy from the knee are critical for reducing actuator work to track human torque profiles at the hip and ankle, lowering the mechanical economy predictions of the leg model.

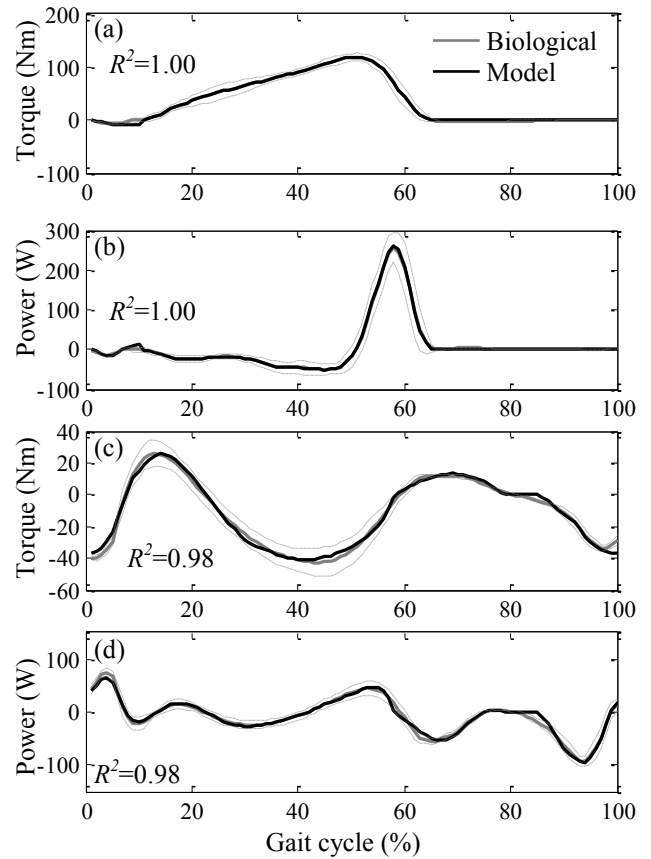


Fig. 2. Biological data and simulation results of (a) ankle torque, (b) ankle power, (c) knee torque, and (d) knee power. All curves start at heel strike and end with the heel strike of the same foot. Here  $R^2$  is calculated between biological data and simulation results during the stance phase (0-62%) for the ankle joint and during the whole gait cycle (0-100%) for the knee joint.

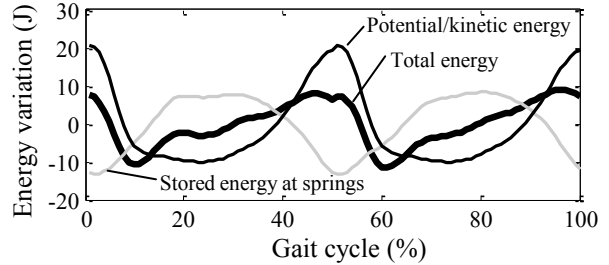


Fig. 3. Mechanical energy variations during the walking cycle. Thin black and gray curves are the potential plus kinetic energy curve and the elastic energy curve stored within all model springs at each percent gait cycle, respectively, and the thick black curve is the sum of all three mechanical energies. All curves begin at a heel strike (0%) and end at the next heel strike of the same foot (100%).

#### 3) Mechanical Economy

The non-conservative, total positive mechanical work is generated only from the three actuators in the model since the series-elastic clutch units can only apply conservative spring forces. At a self-selected walking speed, the mechanical economy of the study participant as predicted by the musculoskeletal model is 0.03. This value was computed using equation (4) where  $W_{act}^+$  was calculated by computing the area under the positive power curves for the three model actuators. The mechanical economy for human walking has

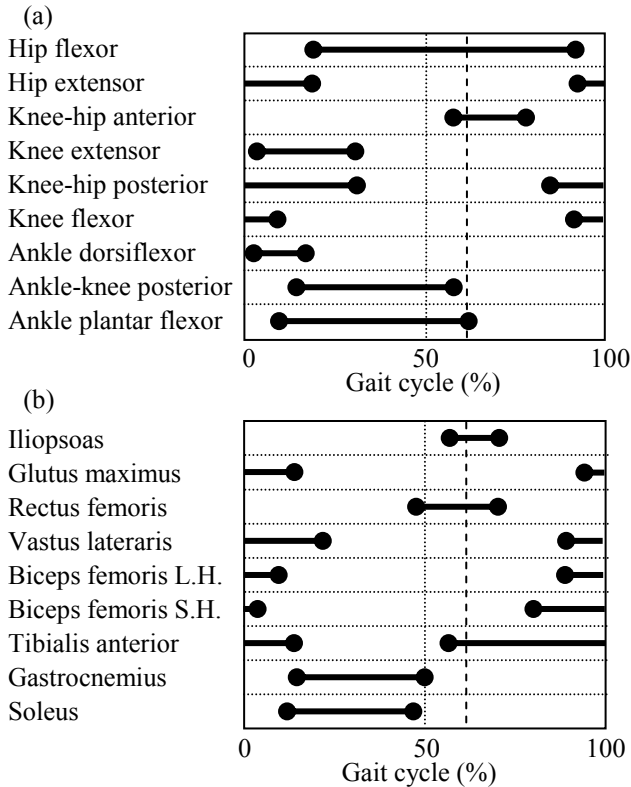


Fig. 4. Model agreement with electromyography. In (a), clutch engagement and actuator activation time spans are shown for each model unit. In (b), muscle electromyography from the literature [24] is shown for corresponding muscles. As before, the gait cycle begins with a heel strike and ends with the next heel strike of the same foot. The vertical dashed line denotes 62% gait cycle, which is the moment when the foot lifts from the ground surface.

been estimated at 0.05 [23]. Perhaps the mechanical economy is lower for the model because the energetic requirements of other body parts, e.g. the trunk, neck, head and arms, were not considered. Moreover, each spring-clutch unit does not generate any nonconservative work which may be necessary to stabilize walking.

#### 4) Whole-body Mechanical Energetics

Figure 3 shows potential plus kinetic energy variations, as well as variations in elastic energy storage from both legs as estimated from the model, versus percent gait cycle. The potential plus kinetic energy curve and elastic energy curve are generally out of phase and thus energy exchange is therefore substantial between these mechanical energy forms. Cavagna et al. [22] reported that the percentage recovery between potential and kinetic energy is 65% on average and is maximized at normal walking speeds. Similarly, recovery between elastic, potential, and kinetic energies was estimated to be 65% using equation (7).

The model results suggest that substantial exchange occurs between all mechanical energies, and this mechanism enables humans to walk economically at normal self-selected speeds.

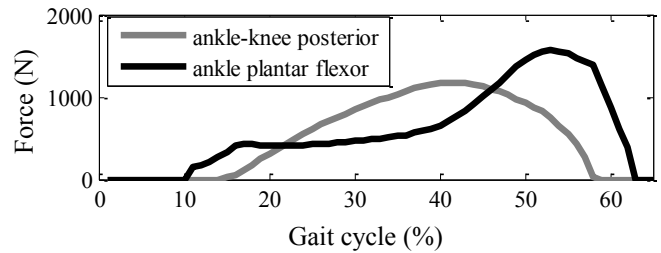


Fig. 5. Forces generated by the ankle-knee posterior and ankle plantar flexor units. Both of these model elements are activated from the early stance phase until toe off at 62% gait cycle.

#### 5) Model Agreement with Electromyography

The leg model was derived by inspection of the human musculoskeletal structure, and thus it is reasonable to compare muscle electromyography (EMG) data with clutch engagement or actuator activation patterns.

Figure 4(a) and (b) show clutch engagement/actuator activation time spans from the musculoskeletal model as well as muscle EMG patterns from the literature [24]. The black horizontal bar in Fig. 4(a) shows the duration when clutches are engaged or actuators perform negative or positive work. Muscle EMG patterns and the corresponding clutch engagements/actuator activations are qualitatively similar between model and biological data, except for the model's hip flexor and ankle dorsiflexor units. The hip flexor unit is activated during a much longer duration than the Iliopsoas EMG signal. This difficulty may be due to the fact that the model did not include any hip ligament compliance. Neptune [24] reported that a more compliant ligament that engaged during hip flexion caused an increase in hip flexor contribution. Furthermore, as shown in Fig. 4(b), the tibialis anterior is activated during the swing phase to apparently keep the ankle adequately dorsiflexed for foot clearance and to orient the foot for the next ground contact. As shown in Fig. 4(a), the model's ankle dorsiflexor clutch is not engaged during the swing phase. This is the case simply because we assumed zero ankle torque during this phase of gait.

Figure 5 shows the forces generated by the ankle-knee posterior clutch and the ankle plantar flexor. The shape of the model's force curves resemble EMG patterns reported in [25, 26]. Model agreement is especially good between the ankle plantar flexor and the soleus muscle. As reported in [25,26], the soleus EMG signal increases after heel strike and is kept nearly constant from early to mid stance (10% to 40% gait cycle). The soleus EMG signal then sharply increases again in late stance during powered plantar flexion. As shown in Fig. 5, the model's plantar flexor force output exhibits qualitatively similar patterns to these soleus EMG signals.

## IV. DISCUSSION

### 1) The Importance of Tendon Stiffness Tuning

Humans walk economically with a mechanical economy equal to 0.05, a value that is significantly lower than current

humanoid robots with an actuator at each leg degree of freedom (6 DOF per leg). How do humans walk so economically? It is believed that the morphology and control of the human leg allows for a significant energy exchange between kinetic, gravitational and elastic energy domains throughout the walking cycle, thereby minimizing the required muscle fascicle work and mechanical economy. This particular issue is difficult to evaluate since elastic energy storage in a tendon is difficult to measure experimentally in walking humans. Due to limitations of *in vivo* experimental techniques, it is difficult to measure muscle state, tendon elongation, and muscle-tendon force during walking. As a solution to this difficulty, walking models have been employed to estimate tendon energy storage and muscle work. For example, Anderson and Pandy [27] employed a three-dimensional human musculoskeletal model with a Hill-type muscle model and optimized muscle activations to minimize metabolic cost. However, their model prediction overestimates metabolic cost by 46%. In more recent work, Neptune et al. [24] used a two-dimensional human musculoskeletal model with a Hill-type muscle model, and optimized muscle activations such that the error between the simulation result and kinetic and kinematic human gait data were minimized. However, their muscle fascicle mechanical work estimation was approximately twice as large as would be expected from metabolic measurements<sup>1</sup>. In these earlier investigations, tendon stiffness values from the literature were used as model inputs. However, tendon stiffness measured at one specific location across the tendoneous sheet does not comprise the total muscle series compliance. The total passive series compliance comprises both proximal and distal tendons often including aponeurosis tissue. Thus, the stiffness of the total series-elastic element is generally unknown. Since fascicle work is directly dependent on how much elastic energy is stored and released from muscle-series compliant structures, we feel in the development of walking models it is critically important to optimize both muscle activations and tendon stiffnesses simultaneously.

Perhaps humans maximize elastic energy usage while walking in order to minimize concentric muscle action. In this investigation, we represented all muscle-tendon units spanning the knee joint using a series-elastic clutch model. The model predicted knee torque, mechanical economy, and EMG values. The success of the model is in support of the hypothesis that muscle-tendon units spanning the human knee joint mainly operate as series-elastic clutch elements, affording the relatively low mechanical economy of humans.

<sup>1</sup> For a 75Kg walking person, the metabolic cost is approximately 185 Watts [28]. With a cycle time equal to approximately 1.2 second, 222 Joules of metabolic energy are required per walking cycle. Since the efficiency of positive muscle work is approximately 25%, one would expect positive muscle work not to exceed 56 Joules per walking cycle. Neptune et al. [24] estimated positive muscle work at 100 Joules per walking cycle.

## V. CONCLUSION

In this paper, we present a two-dimensional leg model that captures the gross features of the human musculoskeletal architecture. The model is under-actuated; three actuators act at the hip and ankle with only passive and quasi-passive components spanning the knee. Using the leg model, we evaluate the hypothesis that a robotic leg can capture the dominant mechanical behavior of the human knee during level-ground ambulation at self-selected speeds. Upon optimizing model spring stiffnesses and actuator contributions, we find good agreement between knee biological and model data throughout the walking cycle. The model also predicts a low mechanical economy for walking. The success of the model is provocative as it suggests that walking robots can achieve human-like leg mechanics and energetics without knee actuators. In the development of low-mass, highly economical prostheses, orthoses, exoskeletons, and humanoid robots, elastic energy storage and human-like leg musculoskeletal architecture are design features of critical importance.

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