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A quasi-passive model of human leg function in level-ground walking

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Abstract - In this paper, we seek to understand how leg muscles and tendons work mechanically during walking in order to motivate the design of efficient robotic legs. We hypothesize that a robotic leg comprising only knee and ankle passive and quasi-passive elements, including springs, clutches and variable-damping components, can capture the dominant mechanical behavior of the human knee and ankle during level-ground walking at self-selected speeds. As a preliminary evaluation of this hypothesis, we put forth a simple leg model that captures the gross features of the human leg musculoskeletal architecture. We vary model parameters, or spring constants, damping levels and times when clutches are engaged, using an optimization scheme where errors between model joint behaviours and biological joint mechanics are minimized. For model evaluation, kinetic and kinematic gait data are employed from a single participant walking across a level-ground surface at a self-selected gait speed (1.3m/sec). With only a single hip actuator, we find good agreement between model predictions and experimental gait data, suggesting that knee and ankle actuators are not necessary for level-ground robotic ambulation at self-selected gait speeds. This result is in support of the idea that muscles that span the human knee and ankle mainly operate eccentrically or isometrically, affording the relatively high metabolic walking economy of humans.

Index Terms – biomechanics, biomimetics, quasi-passive, robotic leg design.

I. INTRODUCTION

When designing efficient, 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. Thus, 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 the 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 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 efficient bipedal walking machines and legged exoskeletons. Passive dynamic walkers [8] have been constructed to show that bipedal locomotion can be very energy efficient. 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]. In the area of legged exoskeleton design, passive and quasi-passive elements have been employed to lower exoskeletal weight and to improve system energy efficiency. In numerical simulation, Bogert [10] showed that an exoskeleton using passive elastic devices can, in principle, substantially reduce muscle force and metabolic energy in walking. Walsh et al. [11] built an under-actuated, quasi-passive exoskeleton designed for load-carrying augmentation. During level-ground walking, the exoskeleton only required two Watts of electrical power for its operation with 90% load transmission through the robotic legs.

Although passive and quasi-passive devices have been exploited to improve overall system efficiency 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 efficient, low-mass robotic legs. We hypothesize that a robotic leg comprising only knee and ankle passive and quasi-passive elements, including springs, clutches and variable-damper components, can capture the dominant mechanical behaviours of the human knee and ankle for level-ground walking at self-selected speeds. As a preliminary evaluation of this hypothesis, we put forth a simple leg model that captures the gross features of the human leg musculoskeletal architecture. We vary model parameters, or spring constants, damping levels and times when clutches are engaged, using an optimization scheme where errors between model joint behaviour and biological joint mechanics are minimized.

II. METHODOLOGY

A. *Quasi-passive leg model*

Figure 1 shows the two-dimensional musculoskeletal model employed in this investigation. The model was derived by inspection of the human musculoskeletal structure. The model comprises a hip actuator, a knee variable damper, and six monoarticular and three biarticular tendon-like springs with series clutches. The hip actuator is the only source of nonconservative, positive work in the leg model. The variable-damper at the knee is used to continuously modulate damping levels, dissipating mechanical energy where needed. Model ankle, knee and hip have agonist/antagonist pairs of monoarticular springs with series clutches. Further, the leg model includes two knee-hip biarticular elements and one ankle-knee element. The knee-hip biarticular elements act about attached pulleys, and thus their knee and hip moment arms are invariant to joint rotation. The ankle-knee biarticular element also attaches at the ankle joint with a pulley, but attaches to the thigh such that the knee moment arm rapidly goes to zero just as the knee reaches full extension. The ankle-knee component also comprises a clutch attached to the shank. This second clutch is used to convert the ankle-knee biarticular element into an ankle monoarticular element. This conversion occurs when the knee reaches full extension at which time the clutch connected to the shank is engaged and the thigh clutch is disengaged. Since the knee moment arm is zero when the knee is fully extended, a discontinuity in knee torque does not occur during the conversion process.

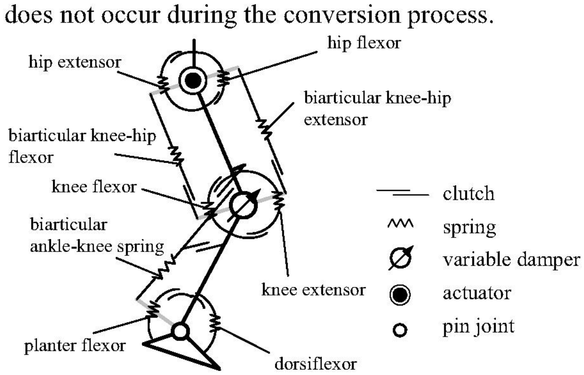

Figure 1. The quasi-passive leg model. The model comprises a variable-damping device, six monoarticular series-elastic clutches and three biarticular series elastic clutches. Only a single actuator acts at the model's hip joint. The ankle-knee biarticular element also includes a clutch connected to the shank.

B. *Optimization Strategy*

The model has a total of 19 series-elastic clutch parameters: 9 spring constants and 10 distinct times when clutches are engaged and disengaged. To optimize these model parameters, human gait data comprising sagittal plane joint angles, moments and powers for hip, knee and ankle were obtained from the literature [12]. These gait data are for a 70kg study participant with a 0.9 meter leg length and a walking speed of 1.3 m/s.

In the optimization strategy, the simplex search method was employed to find optimal series-elastic clutch parameters. Optimization was conducted so as to minimize the following fitness function:

$$\begin{aligned} fitness &= \sum_{joint} \int_t (P_{bj}(t) - P_{sj}(t))^2 dt \\ &= \sum_{joint} \int_t (P_{bj}(t) - \tau_j \dot{\theta}_j(t))^2 dt \end{aligned} \quad (1)$$

where P_{bj} and P_{sj} denote the biological and simulated power of joint j, respectively. τ_j and $\dot{\theta}_j$ denote the simulated torque and biological joint velocity of joint j. In the optimization, joint position and velocity were determined from the biological gait data, and then the integral of the square error of power during one step was minimized.

In the second stage of the optimization process, knee damping and hip actuator power were included. At the knee joint, P_{sj} was initially calculated without the effect of the variable damper so as to maximize the energy exchange between the model's springs and to minimize mechanical energy dissipation from the variable damper. Similarly, P_{sj} at the hip joint was obtained without the effect of the hip actuator to minimize nonconservative actuator work. In this two-stage optimization strategy, we minimized the use of the damper and actuator to produce the most energy efficient walking solution.

III. RESULTS AND DISCUSSION

A. *Ankle Joint*

Ankle mechanics from the model are generated by the ankle dorsi and plantar flexion elements and the ankle-knee biarticular element. Figure 2 shows the ankle moment (upper left) and power (upper right) from both the model and biological data. The model's moment and power curves are in good agreement with biological data except for the early dorsiflexion phase from 10% to 35% gait cycle, and at terminal stance, about 62% gait cycle. Here good agreement is defined as a model prediction within plus or minus one standard deviation about the biological data mean.

Based upon the biological data presented here, the ankle performs more positive than negative work throughout the stance period. Thus, during the powered plantar flexion phase, more positive work is performed by the ankle than can be produced with only stored elastic energies in monoarticular components. Consequently, for the quasi-passive model mechanical work performed by the knee has to be transferred to the ankle. Essential for this energy transfer is the ankle-knee biarticular component. As shown in Figure 3, during early stance knee extension, the clutch attached above the knee in the ankle-knee biarticular element engages. Once the knee reaches full extension, the moment arm at the knee becomes zero, and the clutch below the knee engages followed by the disengagement of the above-knee clutch. These control actions ensure that all the energy stored in the ankle-knee spring is employed to power the ankle. For the model results shown in Figure 2, 4 joules are transferred from knee to ankle.

The lower plot in Figure 2 shows the power output of each component that spans the model's ankle joint. Throughout the gait cycle, the plantar flexor and ankle-knee biarticular components are dominant with a very small contribution from the dorsiflexor component. During stance, the plantar flexor and ankle-knee biarticular elements do negative work, storing energy during the controlled dorsiflexion phase. That stored energy is then released during powered plantar flexion at terminal stance. After toe-off, no spring force is generated. Table 1 shows the stiffness of each series spring spanning the ankle joint. The ankle and knee moment arms for the ankle-knee component are both 5cm. As noted earlier, the knee moment arm rapidly goes to zero just before the knee reaches full extension.

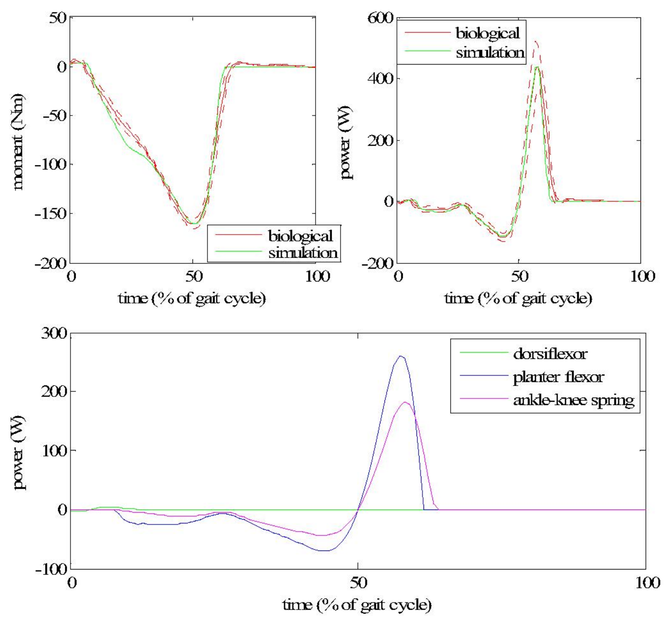

Figure 2. Ankle moment and power from both the model and biological data. The upper left and right plots show ankle moment and power, respectively. Solid red line is the mean of the human gait data (n= 7 walking trials), and dashed lines are one standard deviation about the mean. Solid green line is the model or simulation curve. All curves start at heel strike (0% gait cycle) and end with the heel strike of the same leg (100% gait cycle). In the lower plot, mechanical power of each model component is shown.

Table 1: Stiffness of ankle springs

Ankle dorsiflexor spring stiffness	25.7 Nm/rad
Ankle plantar flexor spring stiffness	360.5 Nm/rad
Ankle-knee biarticular spring stiffness	60,010 N/m

B. *Knee joint*

Knee mechanics from the model are generated by the flexor/extensor monoarticular components, the ankle-knee and knee-hip biarticular components, and the knee variable-damper (see Figure 1). Figure 3 shows moment (upper left) and power (upper right) for both the model and biological data. The model's moment and power curves are in good agreement with biological knee data except around late knee extension at ~35% gait cycle and at toe-off, or ~62% gait cycle.

The lower plot in Figure 3 shows the power output for each model component that spans the knee. During early stance knee flexion from 0% to 12% gait cycle, the monoarticular extensor component stores elastic energy and the knee-hip flexor releases energy previously stored during the swing extension phase. As the knee extends from 12% to 35%, the knee extensor releases its energy and the ankle-knee biarticular and the flexor monoarticular components store energy. The energy stored in the ankle-knee spring powers ankle movements, whereas the energy stored in the knee flexor powers terminal stance knee flexion. The knee moment and power curves are completely smooth because the above-knee clutch of the ankle-knee component is disengaged only when its knee moment arm is zero at full knee extension. During swing phase flexion, knee-hip biarticular extensor stores energy in order to power knee extension in preparation for heel strike. The knee-hip biarticular flexor stores energy during late swing extension and that energy is then used for buckling the knee just after heel strike. The variable damper is mainly used during terminal stance and the swing phase. Table 2 shows the stiffness of each model spring that spans the knee joint, except the ankle-knee spring stiffness which is listed in Table 1. During the early stance phase, high stiffness springs are required to inhibit knee flexion/extension immediately following heel strike. The knee and hip moment arms for the knee-hip flexor component are both 3cm. Further, the knee and hip moment arms for the knee-hip extensor component are 5cm and 2cm, respectively.

Table 2: Stiffness of knee springs

monoarticular knee extensor	206.3 Nm/rad
monoarticular knee flexor	190.2 Nm/rad
knee-hip biarticular extensor	2,483.2 N/m
knee-hip biarticular flexor	103,060 N/m

C. *Hip Joint*

Hip mechanics from the model are generated by the flexor/extensor monoarticular components, the knee-hip biarticular components, and the actuator (see Figure 1). Figure 4 shows moment (upper left) and power (upper right) for both the model and biological data. Because of the hip actuator, the model's moment and power curves perfectly match the biological mean curve.

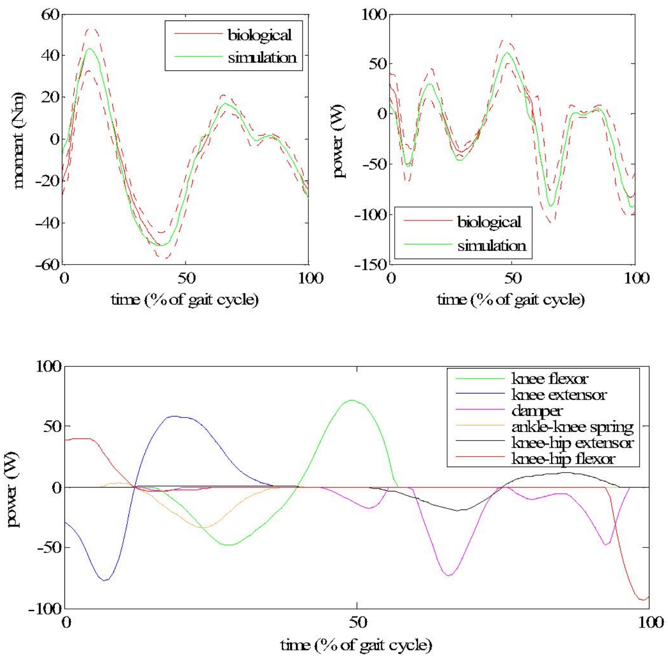

Figure 3. Knee moment and power from both the model and biological data. The upper left and right plots show knee moment and power, respectively. Solid red line is the mean of the human gait data (n= 7 walking trials), and dashed lines are one standard deviation about the mean. Solid green line is the model or simulation curve. As before, all curves start at heel strike (0% gait cycle) and end with the heel strike of the same leg (100% gait cycle). In the lower plot, mechanical power of each model component is shown.

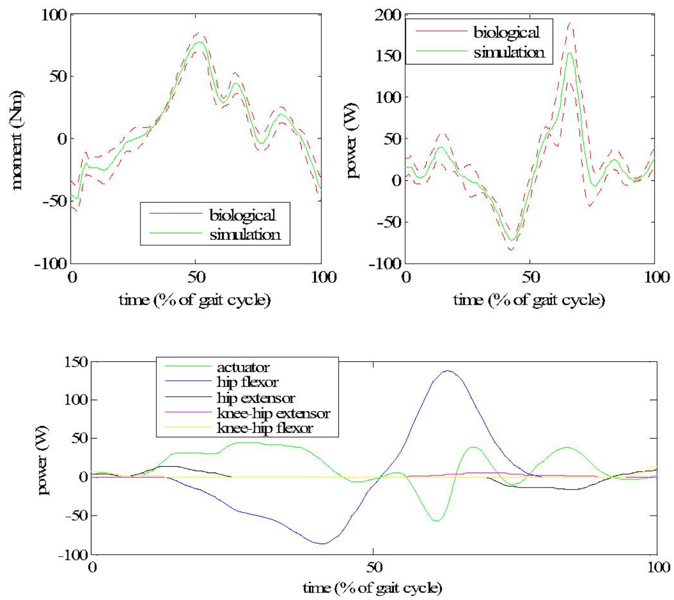

Figure 4. Hip moment and power from both the model and biological data. The upper left and right plots show hip moment and power, respectively. Solid red line is the mean of the human gait data (n= 7 walking trials), and dashed lines are one standard deviation about the mean. Solid green line is the model or simulation curve. As before, all curves start at heel strike (0% gait cycle) and end with the heel strike of the same leg (100% gait cycle). In the lower plot, mechanical power of each model component is shown.

The lower plot in Figure 4 shows the power output for each model component that spans the hip joint. During hip extension in stance, the monoarticular hip flexor stores energy that later powers hip flexion during terminal stance and early swing. As shown in Figure 4, the monoarticular extensor contributes little to the hip mechanics of the model. Further, since the hip moment arm for the knee-hip extensor (2cm) is relatively small in comparison to the knee moment arm (5cm), its contribution to overall hip mechanics is modest. Table 3 shows the stiffness of the flexor/extensor monoarticular springs. The hip extensor spring stiffness is relatively small given its negligible contribution to hip mechanics.

Table 3: Stiffness of hip springs

monoarticular hip flexor	103.1 Nm/rad
monoarticular hip extensor	53.5 Nm/rad

IV. 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; a single actuator acts at the hip with only passive and quasi-passive components spanning knee and ankle. Using the leg model, we evaluate the hypothesis that a robotic leg comprising knee and ankle passive and quasi-passive elements can capture the dominant mechanical behaviour of the human knee and ankle during level-ground ambulation at self-selected speeds. After model spring stiffnesses, damping levels and actuator contributions, we find good agreement between biological and simulated gait data for most regions of the walking cycle. The success of the model is provocative as it suggests that walking robots can achieve human-like leg mechanics without knee and ankle actuators. In the design of legged systems, elastic energy storage and human-like leg musculoskeletal architecture are design features of critical importance to the development of low-mass, highly economical walking machines.

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