Investigation on a bipedal robot: Why do humans need both Soleus and Gastrocnemius muscles for ankle push-off during walking?

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Abstract—Legged locomotion in humans is influenced by mechanics and neural control. One mechanism assumed to contribute to the high efficiency of human walking is the impulsive ankle push-off, which potentially powers the human swing leg catapult. However, the mechanics of the human’s lower leg with its complex muscle-tendon units spanning over single and multiple joints is not yet understood. Legged robots allow testing the interaction between complex leg mechanics, control, and environment in real-world walking gait. We custom developed a small, 2.2 kg human-like bipedal robot with soleus and gastrocnemius muscles represented by linear springs, acting as mono- and biarticular elasticities around the robot’s ankle and knee joints. We tested the influence of three soleus and gastrocnemius spring configurations on the ankle power curves, on the synchronization of the ankle and knee joint movements, on the total cost of transport, and on walking speed. We controlled the robot with a feed-forward central pattern generator, leading to walking speeds between 0.35 m/s and 0.57 m/s at 1.0 Hz locomotion frequency, at 0.35 m leg length. We found differences between all three configurations; the soleus spring supports the robot’s speed and energy efficiency by ankle power amplification, while the GAS spring facilitates the synchronization between knee and ankle joints during push-off.

I. INTRODUCTION

How complex the human bipedal walking is becomes apparent when attempting to technically replicate or restore the human lower limb’s musculoskeletal system, e.g., with humanoid robots or lower limb prostheses. Robots and prostheses do not yet reach human performance in terms of efficiency, mobility, and robustness. The current gap implies a missing understanding of biomechanics and control of human locomotion. One mechanism assumed to contribute to the high efficiency of human walking is the impulsive ankle push-off, which potentially powers the human swing leg catapult (SLC) [1]. A catapult’s function is physically characterized by slow storage of elastic energy, followed by a rapid release of stored energy, with a substantially higher output power that accelerates a projectile. Hof et al. described first how the ankle power burst in the late stance phase is preceded by a slower energy storage phase in human walking [2]. The observed ankle power burst is hereby higher than the plantar flexor muscles’ peak power [1], [3], [4]. This indicates that additional passive structures store elastic energy as kinetic energy [2], [5]. The rapid release of the stored elastic energy at the end of stance is available to accelerate the stance leg into swing like a catapult firing its projectile—the swing leg [1]. A catapult has three main mechanical components: an elastic element, a block, and a catch with or without escapement. In the human lower leg, the complex interplay between the thigh-shank-foot segment chain and muscle-tendon-units (MTU) spanning the ankle joint make it hard to identify the exact swing leg catapult mechanics and its functionality. Elastic energy storage in stance and rapid recoil during push-off is facilitated by the soleus (SOL) and gastrocnemius (GAS) muscles [3], [4], [6], [7]. The GAS muscle works mostly isometrically during stance, and potentially facilitates energy efficient loading of its muscle tendon unit. In contrast, the SOL muscle seems to contribute to ankle coordination by active contraction [8]. Up to 91% of the power output during push-off is provided by elastic energy [9]. Changes in the effective ankle stiffness influence the energy efficiency of walking, as studies with passive ankle foot orthosis showed [10], [11]. The combination of ground and foot is the human leg catapult’s block, against which the swing leg catapult unloads when shifting the body weight from the stance to the trailing leg. This happens in the very brief double support phase of walking, during late stance. Interestingly, the catch of the swing leg catapult has not been identified in the human leg yet. Previous investigations into multi-articulate actuators and spring-tendons in robots showed several advantages. The BioBiped3 robot with biarticular muscles showed improved balance control during upright standing and locomotion, and axial leg function during bouncing [12]. CARL robot improved its hopping efficiency by 16% by using multiarticular actuators [13]. A simulation study shows that combining mono- and bi-articular foot prosthesis actuation reduces peak power requirements [14]. So far, the exact role and function of plantarflextor spring-tendons (GAS and SOL) as part of the swing leg catapult during walking has not been studied in robots.

In this work we show the influence of SOL and GAS muscles on a human-like bipedal robot’s ankle power curve, ankle and knee joint synchronization, energy efficiency and walking speed. As testing platform the developed robot features both ankle muscles as linear springs. We analyze the robot’s energy consumption for one gait cycle and report the
observed ankle power amplification of the swing leg catapult, inspired by human locomotion research like [1], [3], [4], [6]. Developing a deeper understanding of the swing leg catapult and its functional components will help to increase the efficiency of walking for legged robots and bears great potential to improve gait rehabilitation devices and prosthesis toward a more natural gait pattern.

II. METHODS

We designed a human-like bipedal robot at the size of a small child. The robot’s hip and knee joints are actuated and controlled by an open-loop central pattern generator [15]. The control patterns follow human-like joint trajectories. The robot’s ankle joints are not actively actuated. To study the effect of SOL and GAS muscles acting at the knee and ankle joints, SOL and GAS muscle tendon units are represented by linear springs connected to tendons acting on pulleys of fixed radii around the joints (Fig. 1). We tested three spring configurations: SOL + GAS, SOL and GAS.

A. Bipedal robot design

Our humanoid robot features passive toe and ankle joints, actuated knee joints with parallel elasticity, and actuated hip joints (Fig. 1). The robot’s actuated segments for hip and knee are adapted from the SOLO robot module which is described in detail in [16]. Segment length ratios and joint range of motions of the robot (Tab. II) were designed according to 2D neuromuscular simulation studies of human walking [17].

The robot’s elasticities represent muscles in the human leg and foot. Spring stiffness values were based on initial approximations; we tuned the stiffness before experiments leading to the robot’s stiffness values (Tab. II). SOL and GAS springs spanning the ankle joint are loaded from the maximum ankle plantarflexion (PF) angle of 22° until the maximum ankle dorsiflexion (DF) angle of 15°. At peak dorsiflexion angle both ankle springs produce a torque leading to reaction forces equivalent to one third of the robot’s weight acting at the hip joint. The VAS spring spans the knee joint acting on its knee cam radius \( r_{VAS} \), and induces knee torque in parallel to the knee motor, similar to [18]. The VAS spring is loaded starting from 0° knee joint angle. The VAS spring produces a torque of 10% of the robot’s weight at 20° knee flexion angle and 0° ankle angle. Rotational toe springs are loaded in stance from an initial toe angle of 15° up to 45°. At its maximum angle, the toe spring produces a torque of one third of the final robot weight acting at the toe’s tip. Additionally, a toe tendon (without spring) was attached between the thigh and toe segments which facilitated toe clearance in swing as the knee was flexed more than 20° (Fig. 1).

B. Robot control

We implemented Central Pattern Generators (CPGs) that controls the bipedal walking gait. The CPG is a modified Hopf oscillator modeled in phase space [19], and reference
trajectories are calculated by oscillator nodes. With a half-cycle desired phase shift between left and right oscillator nodes, a walking gait is implemented.

CPG joint trajectories are controlled with eight parameters. The hip offset ($\Theta_{\text{hipOffset}}$) and amplitude ($\Theta_{\text{hipAmplitude}}$) describe the hip trajectory. The knee offset ($\Theta_{\text{kneeOffset}}$) and amplitude ($\Theta_{\text{kneeAmplitude}}$) define the knee trajectory. The frequency ($f$) describes the robot’s overall speed, and duty factors ($D_{\text{hip}}$ and $D_{\text{knee}}$) define the ratio between stance phase and swing phase for hip and knee separately. The hip swing steady parameter ($\varphi_{\text{hip}}$) defines the hip’s steady duration, at the end of the swing phase, as a fraction of a gait cycle.

The CPG is described by a set of coupled differential equations:

$$\dot{\Phi}_i = \Omega_i$$

where $\Phi$ is the oscillatory phase vector and $\Omega$ is the angular velocity vector for leg $i$.

$$\dot{\Phi}_i = 2\pi f + \sum_{j=1}^{N} \alpha_{\text{dyn},ij} \cdot C_{ij} \cdot \sin \left( \Phi_j - \Phi_i - \Phi_{d,ij} \right)$$

where $f$ is the frequency, $\alpha_{\text{dyn},ij}$ is the conversion constant of the network dynamics between nodes $i$ and $j$ (we use $\alpha_{\text{dyn},ij} = 5$), $C_{ij}$ is the coupling matrix weight between nodes $i$ and $j$, $\Phi_{d,ij}$ is the desired phase difference matrix value between nodes $i$ and $j$.

$$\phi_i = \begin{cases} \frac{\Phi_i}{2D_{\text{hip}}} & \Phi_i < 2\pi D_{\text{hip}} \\ \frac{\Phi_i + 2\pi(1 - 2D_{\text{hip}}) - 2\pi \varphi_{\text{hip}}}{2(1 - D_{\text{hip}} - \varphi_{\text{hip}})} & 2\pi D_{\text{hip}} \leq \Phi_i \text{ and} \\ \frac{\Phi_i - 2\pi D_{\text{knee}}}{2(1 - D_{\text{knee}})} & \Phi_i < 2\pi(1 - \varphi_{\text{hip}}) \\ 0 & \text{else} \end{cases}$$

where $\phi_i$ is the $i$th end-effector phase, $D_{\text{hip}}$ is the duty factor for the hip. The desired hip trajectory is calculated from the end-effector phase of Eqn. (3).

$$\Theta_{\text{hip},i} = \Theta_{\text{hipAmplitude},i} \cos \phi_i + \Theta_{\text{hipOffset},i}$$

$\Theta_{\text{hip},i}$ is the hip trajectory for the $i$th end-effector, $\Theta_{\text{hipAmplitude},i}$ is the hip amplitude and $\Theta_{\text{hipOffset},i}$ is the hip offset. Similarly to hips, the desired knee trajectory is shaped by its duty factor. With the end-effector knee phase of Eqn. (4), we use the phase to calculate the sine-shape for the knee trajectories in Eqn. (5).

$$\phi_i = \begin{cases} \frac{\Phi_i - 2\pi D_{\text{knee}}}{2(1 - D_{\text{knee}})} & \Phi_i < 2\pi D_{\text{knee}} \\ 0 & \text{else} \end{cases}$$

$$\Theta_{\text{knee},i} = \Theta_{\text{kneeAmplitude},i} \sin \phi_i + \Theta_{\text{kneeOffset},i}$$

The reference trajectories for hip and knee, used to design the CPG control, were extracted from a 2D neuromuscular simulation of human walking [17]. We initially set the CPG parameters based on this reference, and found the parameters during the experiments tuning toward the most successful walking pattern, as presented in Tab. I. We ran the controller feed-forward, i.e., the controller was streaming a position signal to the brushless actuator module adapted from [16], without incorporating external feedback.

| TABLE I: CPG control parameters. |
|----------------------------------|
| Parameter                        | (GAS+SOL) | Value     |
| Frequency                        | $f$       | $f$       | $f$       |
| Hip duty factor                  | $D_{\text{hip}}$ | 0.65      | 0.65      | 0.65      |
| Knee duty factor                 | $D_{\text{knee}}$ | 0.65      | 0.65      | 0.60      |
| Hip amplitude                    | $\Theta_{\text{hipAmplitude}}$ | 30°       | 30°       | 27.5°     |
| Knee amplitude                   | $\Theta_{\text{kneeAmplitude}}$ | 70°       | 70°       | 70°       |
| Hip offset                       | $\Theta_{\text{hipOffset}}$ | 5°        | 2.5°      | 2.5°      |
| Knee offset                      | $\Theta_{\text{kneeOffset}}$ | 2°        | 2°        | 2°        |
| Hip swing steady                 | $\varphi_{\text{hip}}$ | 0.05      | 0.05      | 0.05      |
| Knee swing steady                | $\varphi_{\text{knee}}$ | 0.05      | 0.05      | 0.05      |

CPG trajectories are generated by Eqns. (4) and (6). The CPG output forms a position reference for open-loop PD controllers to calculate desired joint motor currents:

$$i_{\text{hipMotor}} = k_p,\text{hip} \Delta \Theta_{\text{hip}} + k_d,\text{hip} \frac{d}{dt} \Delta \Theta_{\text{hip}}$$

$$i_{\text{kneeMotor}} = k_p,\text{knee} \Delta \Theta_{\text{knee}} + k_d,\text{knee} \frac{d}{dt} \Delta \Theta_{\text{knee}}$$

where $i_{\text{hipMotor}}$ and $i_{\text{kneeMotor}}$ are the desired motor current values for hip and knee, respectively. Controller gains are $k_p$ and $k_d$ for the PD controller. For all of experiments, $k_p,\text{hip} = 30$ and $k_d,\text{hip} = 0.2$ are used for hip modules while $k_p,\text{knee} = 15$ and $k_d,\text{knee} = 0.2$ for knee modules. $\Delta \Theta$ is the error between the desired angle value and the current angle.

C. Experimental setup

We tested three robot ankle configurations: SOL only, GAS only, and SOL + GAS. In the first experimental configuration we used the physiological ratio between SOL and GAS maximum isometric force, 3 : 1 [21] (GAS + SOL). In the second (SOL) and third configurations (GAS) 1 : 0 and 0 : 1 spring stiffness ratios were used to test the influence of SOL and GAS separately. We slightly adapted CPG parameters for each setup to ensure stable robot gaits (Tab. I). The robot walked on a treadmill with a gait cycle frequency of 1.0 Hz in all three configurations. At its best possible performance, a gait with a 25° hip joint amplitude and 0.35 m leg length will lead to a locomotion speed of 0.6 m/s, at 1.0 Hz locomotion frequency. This is 50% of the preferred transition speed (PTS) between walking and running. We calculated the robot’s preferred transition speed as two thirds of the theoretically maximum walking speed at Froude number $Fr = 1$ [22], [23]. The robot’s torso was kept vertical with a four-bar mechanism moving on a horizontal slider for medio-lateral and antero-posterior torso stability in all experiments (Figs. 1 and 5).

| TABLE II: Robot configurations. |
|----------------------------------|
| Parameter                        | GAS | SOL | GAS |
| SOL Stiffness [N/mm]             | 4.5 | 6.1 | 0   |
| GAS Stiffness [N/mm]             | 1.4 | 0   | 6.1 |
| Robot mass [kg]                  | 2.22| 2.05| 2.05|
Fig. 2: Hip, knee and ankle joint angle curves during the three experimental configuration and human reference data (from [20]). Shading is the standard deviation of the curves. Positive direction corresponds to flexion for hip and knee angles, and to DF for the ankle angle. The vertical dashed lines indicate toe-off. The joint angles show no substantial changes between GAS + SOL and SOL configurations, while all three joint curves change in the GAS configuration.

D. Data acquisition and analysis

Our robot is instrumented for kinematics and power measurement. We use rotary encoders for hip and knee (AEDT-9810, Broadcom, 5000 CPR), four-bar mechanism and ankle (AMT-102, CUI, 4096 CPR), and treadmill (AEAT8800, Broadcom, 4096 CPR). The power consumption for hip and knee motor is measured by current sensors (ACHS-7121, Broadcom). All sensors are sampled at 600 Hz by a single board computer (4B, Raspberry Pi Foundation), and interpolated to 1000 Hz before analysis.

For each robot configuration (Tab. II), we commanded the robot to walk on the treadmill for 120 s. The collected data were trimmed to the 100 s (100 cycles) where the robot’s gait is in steady state. We averaged the gait cycles starting from touch-down without any filtering. Touch-down was defined as the moment where the ankle joint starts dorsiflexing after swing.

III. RESULTS

Joint angles show no large differences between the GAS + SOL and SOL configurations. For the GAS configuration the ankle joint dorsiflexes sooner than in the other configurations and the ankle plantarflexes less during push-off (Fig. 2). The main differences between the robot and human data are visible in ankle joint and knee angles, during stance. Similarly, the ankle joint power curve does not change between the GAS + SOL and SOL configurations, while for the GAS configuration the negative power peaks are larger and the positive power peak is lower than in the other two configurations (Fig. 3). Power amplification values of the ankle joint power (ratio of positive and negative power peaks Fig. 3) are 3.3, 5.1 and 1.2 in the GAS + SOL, SOL and GAS configuration, respectively. Positive peaks occur at 55% of the gait cycle in all three robot configurations. Negative peaks occur at 50% in GAS + SOL and SOL configurations, and at 40% in the GAS configuration. Compared to the human data, the positive power peaks in the GAS + SOL and SOL configurations are almost twice as large as in the humans but in the GAS configuration the positive peak height is close to identical to the human. A large negative peak appears in stance before push-off when GAS is in the configurations, while this peak does not show in human data (Fig. 3).

The segment synchronization (Fig. 4) shows differences between all three configurations. In the SOL configuration the ankle starts plantarflexing in late stance when the knee starts flexing while in the other two configurations the ankle dorsiflexes 5° and 10° more, respectively, after the knee starts flexing. The ankle push-off starts at 5°, 15° and 20° knee flexion in the SOL, GAS + SOL and GAS configurations, respectively. In the GAS configuration the knee starts flexing in late stance at 8° ankle dorsiflexion while the knee flexes only at 20° ankle dorsiflexion by the other two configurations (Fig. 4).

We recorded the robot’s total power consumption of 20 W, 17 W, and 18 W, and walking speed of 0.55 m/s, 0.57 m/s and 0.35 m/s, for configurations GAS, SOL, and GAS + SOL, respectively. These power values include the robot’s motor drivers and its robot-mounted master control board, which draw in sum about 9 W standby power. We calculate the robot’s total cost of transport (COT) as \(\text{COT} = P/(mgv)\), with \(P\) as the average total positive power (we remove negative power values) measured including all actuators and
Fig. 3: Ankle joint power curves of three experimental configurations and human reference data (from [1]). Shading is the standard deviation of the curves. Positive values indicate positive mechanical joint power. Power amplification ratios are 3.3, 5.1 and 1.2 in configuration GAS + SOL, SOL, and GAS, respectively.

motor drivers [24], [25], m as the robot’s mass, and \( v \) as the average horizontal humanoid robot speed over 100 gait cycles. Therefore, the humanoid robot’s total, positive-power COT values are 2.9 J/J, 1.5 J/J, and 1.5 J/J for configurations GAS, SOL, and GAS + SOL, respectively. We find no major difference between configuration SOL and GAS + SOL, but a 90% larger COT value for the GAS-only, compared to both other configurations.

IV. DISCUSSION

We show the differences in the function of SOL and GAS muscles, represented by linear springs, in the ankle push-off during walking gait using three experimental setups on a human-like bipedal robot. Ankle power amplification is visible in all three experimental configurations Fig. 3. We see clear differences between the setups not only for the ankle power curves but also in the synchronization of the ankle and knee joints Fig. 4. The SOL spring provides ankle power amplification while the GAS spring facilitates the synchronization between knee and ankle joints during push-off.

Walking speed and COT did not change substantially between the GAS + SOL and SOL configurations, the latter performed slightly better. Even though the power amplification was higher in the SOL configuration and the ankle-knee synchronization was different, the energetic consumption did not change substantially. In the GAS configuration, the power amplification (ratio of positive and negative power peaks) decreased Fig. 3 and in accordance speed decreased and COT increased. Our robotic results support the understanding that the SOL muscle is the major elastic element providing energy for the ankle push-off during walking as shown in human subject studies [3], [4], [6], [7] and studies that emulate SOL function by springs with clutches in ankle-foot orthoses [11], [10].

While trying to reach a stable gait pattern with the robot in the different setups, the CPG parameters had to be minimally re-tuned and the ankle-knee joint angle synchronization changed as the main difference between the SOL and GAS springs is that one spans only the ankle joint while the other spans both the ankle and knee joints. To reach a stable walking pattern with the robot in SOL configuration only the hip offset had to be decreased by 2.5° compared to the GAS + SOL configuration because the stride length of the robot became too large in the SOL configuration, the contralateral leg hit the ground as it was going into swing. This effect of decreasing GAS stiffness on the stride length was also shown in a simulation study [26]. To reach a stable gait in the GAS configuration the duty factor of the knee was decreased by 2.5% of the gait cycle, in order to start knee flexion at the end of stance earlier than...
Fig. 5: Snapshots of GAS + SOL configuration during one gait cycle. In-air swing leg retraction is visible in snapshot 2 and 6. 167 ms time passed between snapshots, one entire gait cycle lasts 1.0 s.

the moment of maximum hip extension. Additionally, we decreased hip amplitude and offset both by 2.5° compared to the GAS + SOL configuration as without a SOL spring, as the ankle joint dorsiflexed at the end of stance the knee buckled too early. The motors could not keep the knee straight for larger ankle dorsiflexion angles. The changes of the CPG parameters are apparent in the ankle-knee joint angle synchronization (Fig. 4) as well. The knee starts flexing at 8° ankle DF angle in the GAS configuration instead of 18° DF in the GAS + SOL configuration.

The biarticular GAS spring apparently changes the ankle-knee joint synchronization which is an important aspect of human ankle push-off during walking [27] because it affects the energy transfer between ankle and knee joints.

The robot’s toe clearance in the swing was a limiting factor, which made larger-than-human hip amplitudes necessary, and hip and knee joints were CPG position-controlled during the whole gait cycle. Consequently, the natural springy knee flexion of the human is missing in stance, and the robot’s ankle joint angles show clear differences in all three configurations compared to humans'. That is, the active control of the knee and hip joints affects the natural ankle, knee and hip dynamics induced by the energy release during ankle push-off. In the future, switching knee and hip actuators to low gain control in selected parts of the gait cycle could potentially lead to more natural swing leg dynamics.

Video footage of the human-like bipedal robot’s locomotion on the treadmill in all three configurations is available on YouTube [https://youtu.be/CbJadoKjSX8] and through the submitted supplementary video file.

V. CONCLUSION

In conclusion, differences are apparent between the robot configurations using both SOL and GAS springs, only SOL spring and only GAS spring in the ankle power curves, in the synchronization of the ankle and knee joints, in cost of transport, and in walking speed. The SOL spring provides ankle power amplification seen in the ankle power curve, allowing lower cost of transport and higher walking speed, while the GAS spring facilitates the synchronization between knee and ankle joints during push-off. In the future, more natural swing leg dynamics should be investigated by switching between low and high gain control of the actuators during a gait cycle. We are further interested in actuating the currently passive ankle mechanism by mechanically efficient, series elastic actuation [28] to adjust push-off timing.

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