A Simulation-Based Study on a Clutch-Spring Mechanism Reducing Human Walking Metabolic Cost

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***Abstract*—Reducing the metabolic cost of walking has long been a challenge for exoskeleton researchers. In a recent breakthrough, a passive exoskeleton was reported, which yielded significant energy saving with a clutch-spring mechanism working in parallel with calf muscles. In this study, we investigate whether the same biomechanism exists in the hip and knee joints. We employed OpenSim, an open-source platform, and the MATLAB Optimization Toolbox to determine the engaging and disengaging timings and optimize the stiffness of the springs for walking energetic efficiency. When applied to the ankle joint, the proposed approach yielded results that agreed with the reported ones. We then extended this method to the hip and knee joints. The simulations showed that the springs could save up to 6.38%, 4.85%, and 7.63% for the ankle, knee, and hip joints at the optimal stiffness of 8.20 kN/m, 7.35 kN/m, and 4.15 kN/m, respectively.**

***Index Terms*—metabolic cost, simulation, elastic elements, human walking.**

I. Introduction

Exoskeletons are wearable robots that work in parallel with the human body to provide assistance or augmentation for human activities. They are designed to combine human intelligence and robotic power. The initiative of exoskeletons dates back to 1965, when General Electric and the US military coproduced the first known exoskeleton: Hardiman [1]. Since then, researchers have broadened and deepened their research on wearable robots, and a great number of exoskeletons have been developed [2–4]. Leg exoskeletons, as one major category of exoskeletons, also witnessed substantial progress. Various devices have been built to serve different purposes, such as walking assistance, rehabilitation, and strength augmentation [4].

Walking assistance devices are designed to offload the muscles' burden during walking. This is especially of significance for long-distance walking as well as for locomotion with weak legs. They exist in two groups: active and passive/quasi-passive. Active exoskeletons use motors to offload joint torque to assist human movements, whereas passive ones exert no active torque. Scientists usually assess the performance of such devices by measuring the metabolic cost. An ideal device will result in a drop in the metabolic cost.

Active exoskeletons use external actuations to unload joint torque for metabolic cost saving. As the ankle joint constitutes a large proportion of the energy consumed during walking, it has been the focus of assistive exoskeletons. A number of studies have reported success in metabolic cost saving using powered exoskeletons. Asbeck, De Rossi, Holt, and Walsh [5] presented a soft exosuit to provide assistive torque at the hip and ankle joints during walking. They reported a metabolic cost reduction of 6.4% when walking with the exosuit actuated in comparison with the unactuated condition. Malcolm, Derave, Galle, and De Clercq [6] used a pneumatic driven exoskeleton to assist plantar flexion motion at the ankle joint. The study reported an energy saving of 6% compared with natural walking. In [7], the authors designed a soft suit powered by pneumatic actuators to

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power the hip, knee, and ankle joints. A pilot study on the metabolic cost, only with ankle joint actuation, resulted in a small increase compared with natural walking and a considerable decrease compared with walking with the exosuit unactuated. Using external torque to offload the gross torque generated by muscles can save energy for locomotion. However, the great mass and inertia of active devices impose burdens on subjects, which often result in higher metabolic costs in comparison with natural walking.

To reduce the mass and inertia issues of active exoskeletons, researchers also strived for achievements in passive exoskeletons [8–10]. The absence of actuation components makes passive exoskeletons simpler, lighter in weight, and energy-efficient. Passive devices usually employ elastic elements to supplement the biomechanism of human walking in order to make walking more efficient.

Different schemes have been investigated for passive exoskeleton design. Based on the hypothesis proposed in [11], XPED [9] employed artificial tendons spanning the hip, knee, and ankle joints to restore and redistribute energy during walking. However, the latest version does not end up with energy saving. The MIT exoskeleton [12] is based on an analysis of clinical gait data, especially on positive and negative power patterns of human walking. Gait phases that are potential for energy storage and release are identified and integrated into the exoskeleton design. The exoskeleton focuses on the hip and ankle joints for energy efficiency. A study on the exoskeleton reported a 10% increase in metabolic cost, whereas there was a 23% increase without energy saving components. This proved the effectiveness of using elastic elements to improve energy efficiency. Collins, Wiggin, and Sawicki [10] produced an unpowered exoskeleton that reduces the metabolic cost of walking. This exoskeleton consists of a simple linear spring and a mechanical clutch placed parallel to the calf muscles. The clutch is designed to activate and deactivate the spring at certain angles, which enables the spring to restore energy after heel contact and release energy to assist push-off. The study also obtained the optimal stiffness that results in the highest energy efficiency. The MIT exoskeleton and the Collins exoskeleton indicate that using elastic elements with proper active time can improve walking energy efficiency. Collins' study also indicated the importance of the right stiffness for maximum energy saving. However, the study only focuses on the ankle joint; such a concept for hip and knee joints has not been explored. Virtual modeling has considerable advantages in device prototyping, especially in reducing the design time and cost. In the field of biomechanics, many programs currently exist that support comprehensive musculoskeletal models [13–16]. These programs are able to measure and compute key biomechanical quantities ranging from limb position and tendon extension to joint torque and power. Some of the programs also support the addition of external structures, which allow researchers and engineers to efficiently test their devices in virtual simulations rather than constructing a physical prototype.

A desired software package should meet the following requirements to study metabolic cost: a close musculoskeletal simulation of human walking, modeling of an exoskeleton, and metabolic consumption calculation. OpenSim [16] readily provides human gait models built as per accurate experimental observations, and it features tools for metabolic cost calculation. Therefore, we employed OpenSim to conduct simulations and combined it with the MATLAB Optimization Toolbox to optimize the engaging and disengaging timings as well as the stiffness of the springs to improve walking energetic efficiency.

This paper is organized as follows. Section II details the methodology that we adopt to solve for optimal timings and stiffness. Section III gives the simulation results and discusses the energy saving performance of different conditions. Section IV concludes the paper.

II. Clutch-Spring Mechanism Investigation

A. Simulation Environment

OpenSim is a piece of software that is capable of accurate musculoskeletal locomotion simulation, featuring powerful tools such as inverse and forward dynamics, computed muscle control (CMC), and metabolic probe [17]. Based on the captured kinematic data and ground reaction forces on a subject, the inverse dynamics tool can reveal the torque at each joint. The CMC tool replaces the joint torque by muscle actuations. Thus, a musculoskeletal model actuated by muscles can be obtained. The metabolic probe enables OpenSim to compute the energy consumption.

A standard procedure for studying the metabolic cost of walking in OpenSim can be given as follows:

- Import a musculoskeletal model into the virtual environment, and modify its parameters to suit the object.
- Import the exoskeleton model, and integrate it into the human model.
- Import motion capture data, and integrate the kinematic observation into the model to generate a gait cycle.
- Import external forces (ground reaction forces), and conduct inverse dynamics to compute the torque at each coordinate.
- Adopt CMC to generate a muscle-driven walking simulation.
- Adopt metabolic probe to gain an insight into energy consumption.

It is notable that, during the entire procedure of gait modeling, the kinematics remain unchanged, while the whole model strives to track the kinematic trajectories. OpenSim also supports the implementation of exoskeleton models constructed in CAD software. According to the aforementioned modeling procedure, adding masses to the model would violate the dynamic simulation. Therefore, in this study, we made the exoskeleton massless. In practice, a physical realization using carbon fibers does not introduce too much mass either.

Overall, OpenSim can generate a gait simulation for a human-exoskeleton system and can return metabolic information. However, OpenSim does not come with functions for direct exploration in active and inactive time intervals and spring stiffness. An approach is developed and tested in this study.

B. *Human Walking Model*

An accurate human musculoskeletal walking simulation for level ground locomotion is the foundation of this study. The OpenSim community readily developed a number of walking models for research purposes. In this study, we used a model focusing on the lower extremities. This model was from a male subject 75 kg in weight and 1.8 m in height, walking at 1.2 m/s on a horizontal treadmill. The kinematic data are from [18]. It is notable that a number of simplifications have been made in the model:

- The upper body is concentrated at the torso.
- Only movements in sagittal plane are considered.
- Muscles are modeled into key flexor/extensor muscle groups.

In the simulation of the human-exoskeleton system, we assume that when the subject walks with the exoskeleton, his gait kinematics do not change.

C. *Exoskeleton Implementation and Spring Investigation*

In order to mount the springs in parallel with human muscle groups, a wearable exoskeleton has been built in SolidWorks on the basis of the morphology of the subject. The model is given in Fig. 1. Then, the 3D model was imported and attached to the human model. A walking cycle for the human-exoskeleton model can be generated in OpenSim, as given in Fig. 2.

Figure 1. Exoskeleton 3D model.

To solve for optimal spring stiffness, we first used path actuators in OpenSim for tension force generation in order to replace the clutch-spring mechanisms. Then, using the CMC tool, OpenSim can automatically return optimal tension forces for minimal energy consumption for the human-exoskeleton system. Then, we employed optimization techniques to fit the spring force curves into the optimal tension forces, aiming at an optimal stiffness value for each spring. We repeated this method for each joint for appropriate engagement and stiffness.

The CMC tool is a static optimization process to minimize a cost function, which is a sum of squared muscle controls plus a weighted sum of squared acceleration errors; it is given as follows:

$$J = \sum_{i=1}^{n_x} x_i^{\,2} + \sum_{j=1}^{n_q} w_j (\ddot{q}_{tj} - \ddot{q}_{mj})^2 \,, \tag{1}$$

where n_x is the number of muscles/actuators; x_i is the *i*th muscle control, $0 \le x_i \le 1$; n_q is the number of coordinates to be tracked; w_j is the weight assigned to each acceleration error; $\ddot{q}_{tj}$ is the tracked acceleration of the *j*th coordinate; and $\ddot{q}_{mj}$ is the acceleration of the *j*th coordinate generated by muscles.

It is notable that the forces generated by the muscles/actuators comply with the following equation:

$$F = F_{max} x \,, \tag{2}$$

where x is the muscle control and $F_{\max}$ is a scaling factor provided to the muscles/actuators, which can be changed.

When the path actuators are added to the model, the controls for path actuators are embedded in the total muscle controls. The CMC tool will solve for a set of optimal controls, including the path actuators, to regenerate a dynamic walking model. To maximize the participation of the actuators, it is reasonable to set a high $F_{\max}$ to increase its weight in the optimization. Consequently, this will result in relatively high actuation forces for the path actuators, requiring high stiffness for substituting springs. A trial in OpenSim indicated that an $F_{\max}$ of 1000 N would require a spring of stiffness 30 kN/m, which is not realistic. In order to ensure that the actuator forces are suitable to be realized by springs, $F_{\max}$ was set to be the maximum force that a 10 kN/m spring could exert during a cycle, which is 200 N. This value is applied to all the three joints.

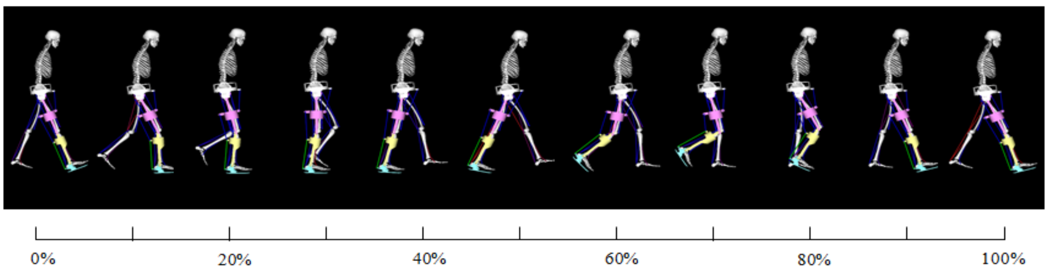

Figure 2. A CMC generated gait cycle of the human-exoskeleton model.

Implementation of the actuators can optimally offload the burden on human muscles; therefore, the force profiles of the actuators provide references for the best energy saving performance by external tension forces. Rerunning the CMC tool, we can get the optimal tension force curves. Then, the results are used as references for optimization of the spring stiffness values.

To maximize energy saving, the springs use the same engaging timings of the actuators. They are disengaged when returning to their engaging lengths. The stiffness values are optimized for the entire active periods. The cost function for optimization of each spring is defined as the sum of squared errors between the actuator force and the spring generated force in the reference points

$$f=\sum_{t=t_{en}}^{t_{disen}}(F_{act}(t)-k\Delta l(t))^2 \quad , \tag{3}$$

where k is the spring stiffness; t_{disen} is the disengaging time of a spring; t_{en} is the engaging time of a spring; t is the time from t_{en} to t_{disen} (it is given discrete values); $F_{\mathrm{act}}(t)$ is the actuator force at time t; $\Delta l(t)$ is the spring stretch after the engaging timing. The spring disengages when $\Delta l(t)$ = 0.

The parameters of the exoskeleton are given in Table I. A schematic drawing of the exoskeleton is illustrated in Fig. 3.

TABLE I. EXOSKELETON PARAMETERS.

Angle	Value (degrees)	Length	Value (mm)
α_1	80.00	l_1	172.62
α_2	20.51	l_2	244.00
α_3	29.05	l_3	242.00
α_4	33.14	l_4	150.91
α_5	15.44	l_5	309.82
α_6	58.00	l_6	94.34

The spring stretch $\Delta l(t)$ is given by the following equations:

$$d_{BC}(t)=l_5^{\,2}+l_6^{\,2}-2l_5l_6\cos[\pi+\theta_{ankle}(t)-\alpha_5-\alpha_6],$$

$$\Delta l_{ankle}(t)=d_{BC}(t)-d_{BC}(t_{en}), \tag{4}$$

$$d_{CD}(t)=l_3^{\,2}+l_4^{\,2}-2l_3l_4\cos[\pi+\theta_{knee}(t)-\alpha_3-\alpha_4],$$

$$\Delta l_{knee}(t)=d_{CD}(t)-d_{CD}(t_{en}), \tag{5}$$

$$d_{EF}(t)=l_1^{\,2}+l_2^{\,2}-2l_1l_2\cos[\pi-\theta_{hip}(t)-\alpha_1-\alpha_2],$$

$$\Delta l_{hip}(t)=d_{EF}(t)-d_{EF}(t_{en}), \tag{6}$$

where d_{BC} is the distance between points B and C at time t; d_{CD} is the distance between points C and D at time t; d_{EF} is the distance between points E and F at time t; $\Delta_{\mathrm{ankle}}(t)$ is the ankle spring stretch after engagement at time t; $\Delta_{\mathrm{knee}}(t)$ is the knee spring stretch after engagement at time t; $\Delta_{\mathrm{hip}}(t)$ is the hip spring stretch after engagement at time t; $\theta_{\mathrm{ankle}}(t)$, $\theta_{\mathrm{knee}}(t)$, and $\theta_{\mathrm{hip}}(t)$ are the angular displacement values for the ankle, knee, and hip joints, respectively, at time t.

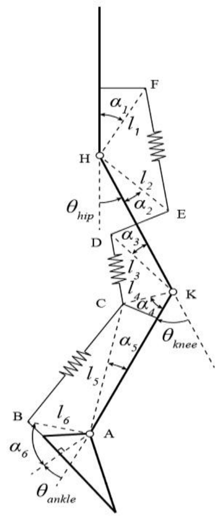

Figure 3. Schematic of a human leg and the exoskeleton.

Using the function "fminsearch" in the MATLAB Optimization Toolbox, we obtained the optimal stiffness for each spring. To evaluate the effectiveness of the optimization results, the metabolic cost of walking with optimal stiffness and other spring stiffness values are compared.

III. RESULTS AND DISCUSSION

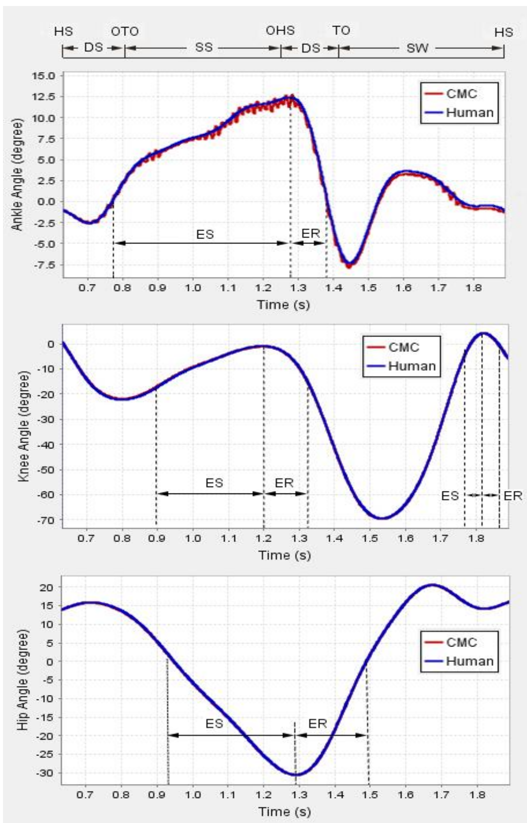

Figure 4. Comparison of angular displacement values. Gait events and phases are indicated at the top of the plots: HS, heel strike; OTO, opposite toe-off; OHS, opposite heel strike; TO, toe-off; DS, double support phase; SS, single support phase; SW, swing phase. Energy storage and return phases are marked in the plots: ES, energy storage; ER, energy release.

The CMC generated kinematics are compared with the captured human gait, as in Fig. 4. The joint angles at hip and knee generated by CMC are identical to the recorded human data. Differences occur in the ankle joint. As the tracking errors are within 3%, we conclude that the CMC is able to reproduce the provided human gait.

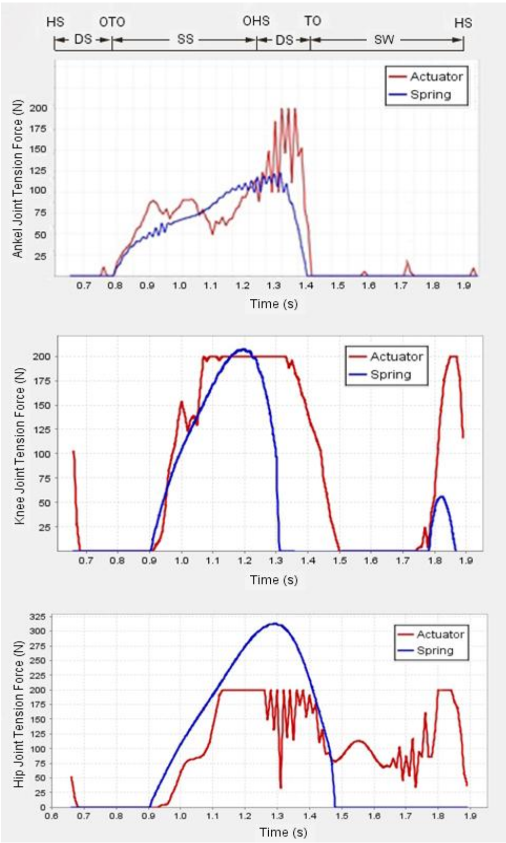

Figure 5. Tension force comparison

The optimal path actuator force for each joint is given in Fig. 5. The ankle actuator is engaged at 0.78 s, slightly after the ankle angle reaches the first minimum. This slightly differs from the unpowered exoskeleton in [10] whose engaging timing is at the minimum. According to [6, 19], humans seek a similar torque pattern when walking with assistive devices at the ankle joint. A typical ankle torque profile indicates that, at the first minimal ankle angle, the torque is provided for dorsiflexion [20]. If the actuator is engaged during 0.71–0.78 s, it works opposite to the gross muscle torque. Inferentially, this would result in higher energy consumption. Therefore, a slightly late engagement would save more energy. The actuator disengages at 1.38 s after toe-off; this is due to the fact that no torque is needed at the ankle joint in the swing phase. This is in line with the disengaging time in [10]. Between the engaging and disengaging times, the actuator exerts forces with similar patterns to the human ankle joint torque. This indicates its capability to offload the muscles' work. Overall, the optimal force provides a good match with a previous study, which verifies the proposed method.

The knee actuator is engaged in two periods. The first period starts at 0.90 s and ends at 1.33 s, occupying a single stance phase, push-off phase, and early swing phase. The second period occurs at 1.77 s and ends with heal strike, to aid deceleration and flexion in the late swing. The hip actuator is engaged at around 0.9 s, and it is disengaged after heel strike. It assists the single stance, push-off, and entire swing phase.

Overall, the path actuators are engaged mainly for the single stance phase, push-off phase, and early swing phase, where pushing off and leg swing cost most of the energy during a walking cycle. The actuator force profiles for hip and knee joint indicate possible energy saving at these 2 joints, which can be estimated by the areas under the curves. However, the knee and hip joints have not been fully explored in previous studies.

The optimization results of spring stiffness are also illustrated in Fig. 5, in comparison with the optimal actuator forces. The springs are restricted by the geometry of the exoskeleton and angular displacement of each joint. When the springs return to their rest length after engagement, they can no longer provide tension force. Differences between optimal actuator forces and spring provided forces are expected. Overall, the ankle spring has the best match. Engaging and disengaging timings coincide with the optimal force. Both hip and knee joints have an early disengaging timing, which reduces the effect for energy saving.

The energy storage and return phases of each spring are illustrated in Fig. 4. In the ES phase, the springs are stretched to store energy. In the ER phase the springs release their energy to assist walking. It is obvious that all the springs store energy during the single support phase and return energy to assist push-off. Additionally, the hip spring also assists early swing.

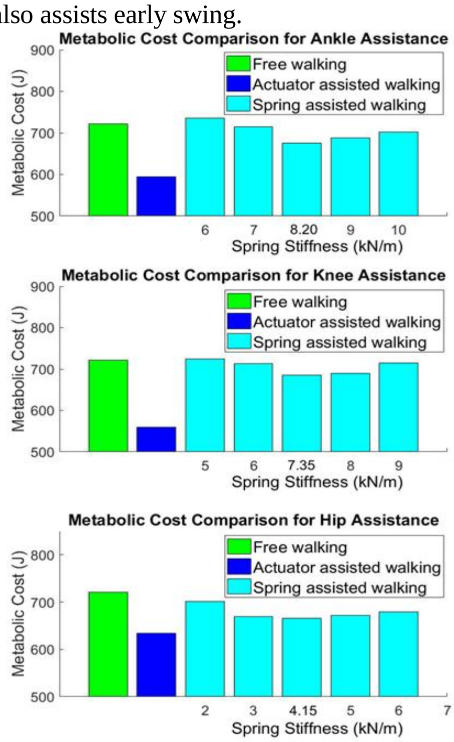

Figure 6. Optimal stiffness evaluation.

The metabolic costs for three conditions—free walking, walking with actuator assistance, and walking with spring assistance—are compared in Fig. 6.

Optimal actuator forces end up with the maximum energy saving, compared with free walking, of 17.48%, 22.33%, and 12.07% for the hip, knee, and ankle joints, respectively. The hip and knee joints save more energy than the ankle joint, with the knee joint holding the most. This hints that assistance at knee joint may result in considerable energy saving, while current exoskeletons mostly concerns hip and ankle joints [5–7].

Springs are not equivalent to active tension forces at energy saving because of the restricted stretching of the springs. Optimal stiffness values bring about the highest energy efficiency in the three joints. These values were found to be 8.20 kN/m, 7.35 kN/m, and 4.15 kN/m for the ankle, knee, and hip joints, with metabolic cost saving of 6.38%, 4.85%, and 7.63%, respectively. The metabolic cost versus spring stiffness pattern in the ankle joint agrees with the study in [10]. Compliant springs save less energy because of their inability to generate sufficient forces. Stiff springs are likely to cause muscles to balance the overgenerated spring force, resulting in increases of metabolic cost. This validates the simulation-based method for investigating the clutch-spring mechanism in energy saving. Similar patterns are also found in the hip and knee joints, which suggests the same biomechanism in the hip and knee joints.

IV. Conclusions

We extended the concept of clutch-spring-based exoskeletons assisting ankle movements into the hip and knee joints using a spring-loaded exoskeleton. The simulation result of the ankle joint matched that in the previous studies in engaging and disengaging timings and stiffness pattern. This validated the effectiveness of our simulation-based method. The clutch-spring mechanism was also found to be energetically beneficial for the hip and knee joints. Optimal engaging and disengaging timings and stiffness values were also obtained for these two joints. The methodology involved in this study is applicable to any subject with customized exoskeletons to save walking energy.

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