

# Spring Uses in Exoskeleton Actuation Design

Shiqian Wang, Wietse van Dijk  
Biomechanical Engineering Dept.  
Delft University of Technology (TU Delft)  
Delft, the Netherlands  
shiqian.wang@tudelft.nl

Herman van der Kooij  
Biomechatronics and Rehabilitation Technology  
University of Twente  
Enschede, the Netherlands  
Biomechanical Engineering Dept.  
Delft University of Technology,  
Delft, the Netherlands  
h.vanderkooij@utwente.nl

**Abstract**— An exoskeleton has to be lightweight, compliant, yet powerful to fulfill the demanding task of walking. This imposes a great challenge for the actuator design. Electric motors, by far the most common actuator in robotic, orthotic, and prosthetic devices, cannot provide sufficiently high peak and average power and force/torque output, and they normally require high-ratio, heavy reducer to produce the speeds and high torques needed for human locomotion. Studies on the human muscle-tendon system have shown that muscles (including tendons and ligaments) function as a spring, and by storing energy and releasing it at a proper moment, locomotion becomes more energy efficient. Inspired by the muscle behavior, we propose a novel actuation strategy for exoskeleton design. In this paper, the collected gait data are analyzed to identify the spring property of the human muscle-tendon system. Theoretical optimization results show that adding parallel springs can reduce the peak torque by 66%, 53%, and 48% for hip flexion/extension (F/E), hip abduction/adduction (A/A), and ankle dorsi/plantar flexion (D/PF), respectively, and the rms power by 50%, 45%, and 61%, respectively. Adding a series spring (forming a Series Elastic Actuator, SEA) reduces the peak power by 79% for ankle D/PF, and by 60% for hip A/A. A SEA does not reduce the peak power demand at other joints. The optimization approach can be used for designing other wearable robots as well.

**Keywords**—exoskeleton; actuation; spring; SEA; PEA; wearable robots

## I. INTRODUCTION

By 2005, more than 250,000 people in the United States and about 2.5 million people all over the world, according to a conservative estimate, were living with spinal cord injury (SCI) induced paralysis [1]. And the number increases every year. Research has been conducted to mitigate the negative effects on the patient's personal and social life.

Some exoskeletons have been designed to help the SCI patients or stroke survivors walk again. ReWalk, claims to be the first commercially viable upright walking assistance tool, and enables wheelchair users with lower-limb disabilities to stand, walk, and even climb stairs [2]. Berkeley Bionics claims that with their newly developed eLEGS system, a walking speed of 0.89 m/s is attainable [3]. Both ReWalk and eLEGS weigh about 20 kg, which is relatively lightweight, but for both systems the users cannot free their hands and have to use crutches to maintain stability and compensate for the power

insufficiency at the actuators. The REX robotic exoskeleton from REX bionics has five actuated degrees of freedom (DoFs) at each leg, and enables the user to walk, make turns, and even climb stairs and slopes [4]. Based on observation, the walking speed of the REX exoskeleton is slower than eLEGS and its weight is about 38 kg. All three exoskeletons use electric motors due to their excellent controllability and ease of supplying power, compared with other actuators such as hydraulic and pneumatic actuators. However, in most of the current designs, the relative low power and torque density of the electric motor result in either a heavy or a functionally limited system. Other actuators with high power and force densities either are difficult to control or too noisy to wear, or require auxiliary energy supply [5].

The goal of this paper is to investigate the potential reduction in power and torque demands on actuators (mainly electric motors) by using series elastic actuation (SEA) and parallel elastic actuation (PEA) in the exoskeleton (Fig. 1). Only linear spring will be considered at this stage.

Both SEA and PEA have been used for walking robots and wearable robots. SEAs, due to their low output impedance, high force fidelity and excellent controllability, and energy storing capability [6], are the most common actuators in robotic, orthotic, and prosthetic devices. RoboKnee and Legged-Robot from Yobotics [7][8], the Robotic Tendon from Arizona State University (ASU) [11], and Powered Ankle-Foot Prosthesis from MIT [12], use linear series elastic actuators (L-SEA) to power the knee or the ankle. Rotary series elastic actuator (R-SEA) is used as well, e.g., the mobility assist exoskeleton from Florida Institute for Human and Machine Cognition (IHMC) [13] and eSEAJ from University of Twente [14]. In [11], researchers claimed that, in theory, adding a series spring to the ankle actuator could reduce the peak motor power from 250W to 77W for an 80 kg person and a much lighter motor could be used.

Fewer applications of PEA than SEA are employed. For

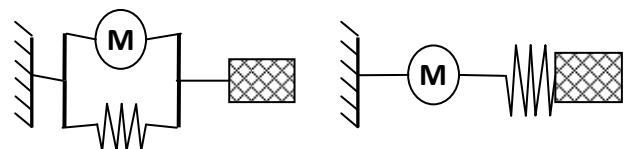


Figure 1. Parallel Elastic Actuation (PEA, left) and Series Elastic Actuation (SEA, right)

example, parallel elasticity is employed in the Powered Ankle-Foot Prosthesis from MIT [12] [16] and the ViSHaRD4 robotic arm from TU Munich [15]. In the Ankle-Foot Prosthesis, parallel elastic actuation is formed. The parallel spring reduces the load borne by the torque source (SEA, in this case) and improves the walking metabolic economy [16].

In general, adding a series elastic element reduces the peak power demand on the motor. Therefore the motor size can be reduced. The direct purpose of adding a parallel elastic element is to reduce the torque demand on torque source (e.g., the gear and motor system). In fact, both the motor size and gear ratio can be reduced. Reduced torque cuts down the input current and copper loss ( $I^2R$ ) and reduces energy consumption.

In this paper, we first present the gait data analysis in section II and then explain the optimization methods for SEA and PEA in section III. The optimization results are represented and discussed in section IV.

## II. GAIT DATA ANALYSIS AND RESULTS

### A. Experiment setup and data processing

Eight subjects (four females and four males), with no gait abnormality, were instructed to walk along a 15-m walkway at different speeds (slow 0.8 m/s and normal 1.2 m/s). The subjects are  $24.3 \pm 1.4$  year old,  $180 \pm 0.1$  cm tall, and weigh  $68.8 \pm 12.1$  kg (mean value  $\pm$  standard deviation). 3D kinematic data from the movements and ground reaction forces from force plates were recorded simultaneously, with the VICON MX system (Oxford metrics, Oxford, United Kingdom), and two force plates (AMTI, Watertown, Mass., United States). The raw data were first post-processed (e.g. filtering the high-frequency noise), and inverse dynamics was performed to obtain the joint torques, and finally the joint powers were calculated.

Due to the repetitive and cyclical property of walking

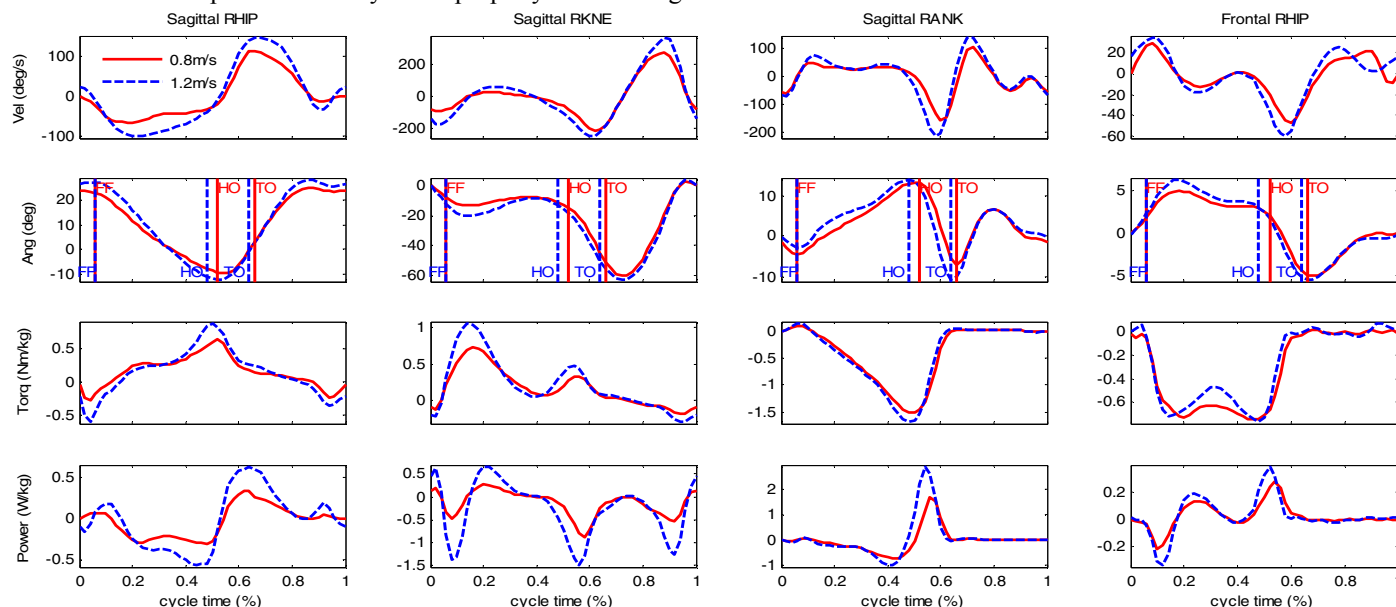


Figure 2. Inter-subject averaged gait data. Torques and powers are normalized to body mass; the time is normalized to gait cycle time; *Sagittal* and *Frontal* indicate human anatomical planes; RHIP = right hip; RKNE = right knee; RANK = right ankle; the same in other figures.

movement, some commonly used techniques were employed to process the data [17]. Intra- and inter-subject averaged profiles have been used. The logic behind this is that measurements from different subjects show very similar trends and patterns.

In the following section, only averaged joint angle, velocity, torque, and power are shown. By assuming the gait data can be linearly scaled according to subject's body weight, torques and powers were normalized by the subject's body weight (mass). The time has been normalized to the gait cycle time. A gait cycle is divided into phases according to the placement of a foot and the interaction between foot and ground. Four events, heel strike (HS), foot flat (FF), heel off (HO), and toe off (TO) divide the gait cycle into four consecutive phases accordingly, early stance, mid-stance, late-stance, and swing phase.

In the exoskeleton we are developing, four degrees of freedom (DOFs) will be actuated: the hip flexion/extension (F/E), hip abduction/adduction (A/A), knee flexion/extension (F/E), and ankle dorsi/plantar flexion (D/PF).

### B. Temporal profiles

The processed gait data for two walking speeds (slow walking 0.8 m/s, solid line, and normal walking 1.2 m/s, dash line) are shown in Fig. 2. Similar gait profiles (i.e., kinematic and kinetic patterns) were observed for both slow and normal walking. Peak powers and torques increase as walking speed increases. The late-stance time (interval between HO and TO) increases with speed. Gait data ranges at 0.8 m/s (TABLE I), were scaled to a body weight of 130 kg. 130 kg reflects the maximal weight for subjects who will be included and the maximal weight of the exoskeleton (30 kg). The peak torque and power, especially those at the ankle, indicate strong demand on the motor power and the torque capacity of the gear and motor.

Slow/normal walking gait data in sagittal plane were compared to the gait data collected by D. Winter [17]. Similar

patterns were observed.

### C. Joint torque-angle curves

To obtain the stiffness characteristics of individual joints, joint torques were plotted against joint angles (Fig. 3). The stiffness  $K$  is the tangent of the torque-angle curve.

$$K = dT/d\theta_j \quad (1)$$

where  $T$  is the joint torque and  $\theta_j$  is the joint angle.

TABLE I. GAIT DATA RANGES FOR A 130 KG SUBJECT WALKING AT 0.8 M/S

Quantities	Sagittal plane						Frontal plane	
	HIP (F/E)		Knee (F/E)		Ankle (D/PF)		Hip (A/A)	
	Min	Max	Min	Max	Min	Max	Min	Max
Angle Velocity [deg./s]	-66	114	-218	275	-154	104	-48	29
Joint Angle [deg.]	-9.5	24.6	-60.3	2.6	-7.0	13.5	-5.0	4.9
Joint Torque [Nm]	-37.1	80.6	-24.2	94.4	-196	13.3	-97.0	2.3
Joint Power [W]	-39.3	43.9	-114	34.3	-95	217.2	-28.0	36.4

During walking, the hip functions as a nonlinear spring in sagittal plane. From HS to HO (early stance and mid-stance phase) the hip spring is loaded, and from HO to end of the stride (late-stance and swing phase), the hip is unloaded. The loading and unloading curves overlap each other. The knee functions as a linear spring during early- and mid-stance phase. And the spring characteristic is altered after the heel is lifted. The ankle works as a linear (or very close to) spring during stance phase. After the heel is lifted (HO), the spring stiffness is changed slightly.

The torque-angle relation suggests that parallel springs (either linear or nonlinear) could be used to store the energy.

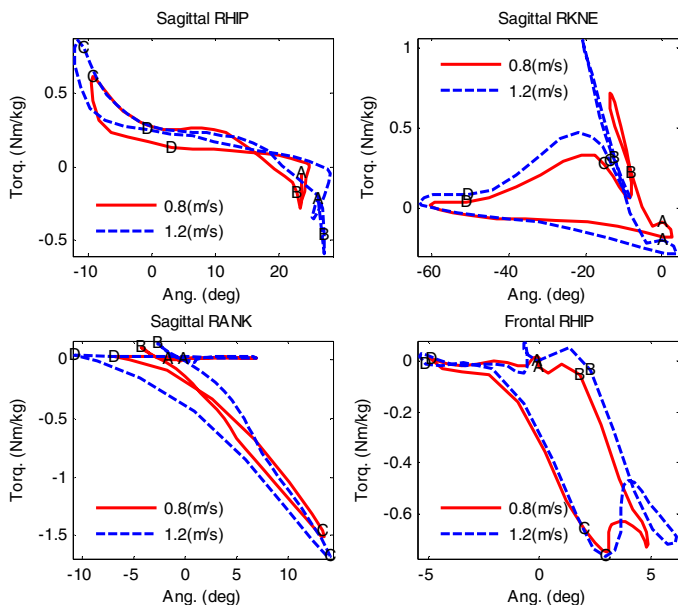


Figure 3. Torque-angle curves. At different walking speeds (0.8 m/s and 1.2 m/s), torque-angle curves differ slightly, but exhibit similar patterns.

By comparing the torque-angle curves at these two walking speeds, we can make a preliminary conclusion that, when walking speed varies in the range of 0.8~1.2 m/s, speed influences on joint stiffness are relative small. The spring stiffness is not necessarily to be adaptive to speed for slow and normal walking.

## III. OPTIMIZATION METHOD

Before explaining the optimization method, the assumptions we made are discussed.

### A. Assumptions

It is assumed that the exoskeleton will totally mimic the human natural gait, so the gait data when wearing the exoskeleton can be obtained by multiplying the normalized joint torques and powers with the total weight of the wearer and the exoskeleton. Furthermore, the exoskeleton is to help spinal cord injured (SCI) patients to walk. To avoid introducing too much psychological burden to the patients, we are aiming at relative low speed, that is, 0.8 m/s. In this paper we first focus on the speed of 0.8 m/s, and then we explain the speed dependency in section IV.B.

For both PEA and SEA, linear springs are considered. The benefit of using linear springs is investigated. To increase the controllability of individual joints, bi-articular and multi-articular actuation like in [9][10], are not considered. The other assumptions on the actuation are

- Frictions due to motor and transmission (e.g., gears) are neglected. Since motor copper and iron losses are not included, efficiencies of the motor and gears are not included in the optimization process;
- Parallel springs can be engaged and disengaged when necessary; to avoid chattering, engagement and disengagement occurs when the spring is at rest (zero spring force);
- The inertias of the motor and gears are not included. Thus the power required for accelerating the motor and gear rotating parts is not included in the optimization;
- The motor requires (the same amount of) energy for both motoring and braking modes.

### B. Optimization method for SEA

The main purpose of using a series spring is to store the energy and reuse it during power-demanding moments during gait, for example, during push off at the ankle, so that the peak power demand on the motor can be reduced. In a SEA, the motor power  $P_m$  can be calculated using the following formulas [11],

$$P_m = \left| T \cdot \dot{\theta}_J + \frac{T \cdot \dot{T}}{K_S} \right|, \quad (2)$$

where  $T$  is the joint torque and  $\dot{T}$  is the time derivative of joint torque;  $\dot{\theta}_j$  is the joint velocity;  $K_S$  is the spring stiffness of the series spring;  $\dot{\theta}_m$  is the motor velocity (input side of the spring, output side of the gears).

The objective is to find a spring stiffness that minimizes the peak motor power

$$f(K_S) = \max(P_m), \quad (3)$$

$$\min_{K_S \in (0, +\infty)} f(K_S).$$

### C. Optimization method for PEA

The primary goal of adding a linear parallel spring is to reduce the peak torque demand on the motor and gears during gait. To reduce the peak torque by fitting a line (linear spring), the least absolute deviations (LAD) method would be the right choice. However, considering the possibly unstable or multiple solutions of LAD, we choose the least squares (LS) method to fit the torque-angle curve to find the optimal spring stiffness. The LS method tries to distribute the torque evenly around the spring torque-angle curve. Further, for most problems (with no outliers in the data set), the solutions of LAD and LS are very close. The objective function is

$$f(K_P) = \sum_{i=1}^N [T_i - K_P(\theta_{J,i} - \theta_E)]^2, \quad (4)$$

$$\min_{K_P \in (0, +\infty)} f(K_P),$$

where  $T_i$  is the joint torque at the  $i$ th time instant;  $\theta_{J,i}$  is the joint angle;  $K_P$  is the parallel spring stiffness;  $\theta_E$  is the spring engagement angle. When the joint reaches this angle, the spring torque is zero, that is, the spring is at rest.

## IV. RESULTS AND DISCUSSION

This section shows the optimization results, i.e. the torque reduction, power or energy reduction. The discussion will be carried out joint by joint.

### A. Optimization results and discussions

#### 1) Series elastic actuation (SEA)

At the ankle, when the spring stiffness increases from zero to infinity, the peak motor power first drops rapidly and then rises relative slowly to gait power 217 W (Fig. 4 upper left plot). The minimum indicates that the optimal spring (stiffness 782 Nm/rad) reduces the peak motor power from 217 to 45 W (by 79%), and the rms power from 61.7 to 13.3 W (by 78%). The close-to-zero motor speed during mid-stance (Fig. 4 upper right plot) suggests that the motor tries to hold the position of the spring and the spring is absorbing energy. The motor torque is exactly the same as the ankle torque during gait, under the assumption given in section III.A. Although series elasticity

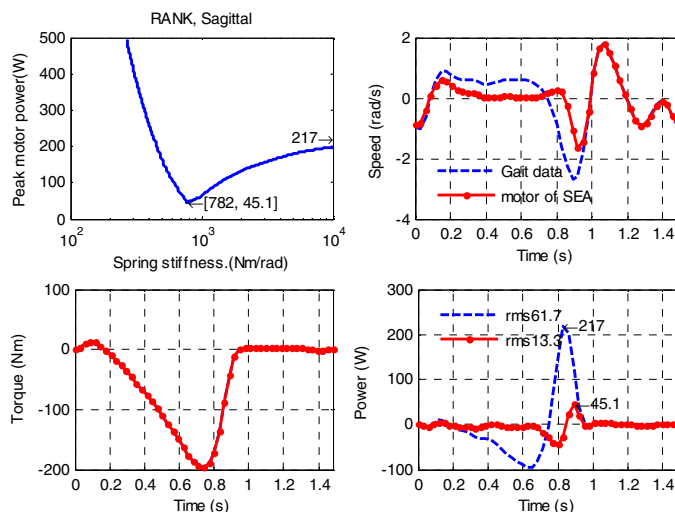


Figure 4. SEA for the ankle. Dash line: collected gait data scaled to a total weight 130 kg; solid-dot-line: the motor speed, torque and power. Optimal spring reduces the motor power by reducing the motor speed, and the motor torque is the same as joint torque

can reduce the power and energy demands, it cannot reduce the high demand on motor torque. It requires a relatively heavy torque motor or a large gearbox (spindle, harmonic drive, gear sets, etc.) with a high gear ratio to generate and strong structure to bear the 200 Nm peak torque.

The same optimization method is applied to other joints. For hip F/E and Knee F/E, no obvious reduction in peak power and rms power is observed. At the hip in frontal plane, adding a series spring can reduce the peak motor power from 36.3 to 14.5W (by 60%) and the rms motor power from 12.2 to 6.18W (by 49%).

#### 2) Parallel elastic actuation (PEA)

When a parallel spring is added to the joint, the motor only needs to deliver part of the joint torque, that is, the *residual torque*, which is the difference between the required joint torque and spring torque. The torque and power profiles at the hip are shown in Fig. 5.

By fitting a line to the torque-angle curve, we find the optimal (in a least-square sense) spring stiffness. At the hip, the parallel spring reduces the peak motor torque from 80.6 to 27.2Nm (by 66%) and from 97.1 to 45.9 Nm (by 53%), and the rms motor power from 24.8 to 12.5W (by 50%) and from 12.2 to 6.73W (by 45%), in sagittal plane and frontal plane, respectively.

At the knee, the spring stiffness changes dramatically after heel off (HO). When implementing a spring with fixed spring stiffness, the motor has to produce a large torque in swing phase (larger than natural gait torque) to counteract the spring to move the lower leg. There is no benefit of adding a parallel spring during swing. Similarly, if a parallel spring is added to the ankle, the ankle motor has to counteract the spring during late stance and swing phase (Fig. 6 middle plot), since the spring stiffness varies and even becomes zero during swing. The conflict can be solved by adding multiple springs with different stiffness values, which work alternately. That will introduce complex and heavy mechanism (see e.g., [18]). For

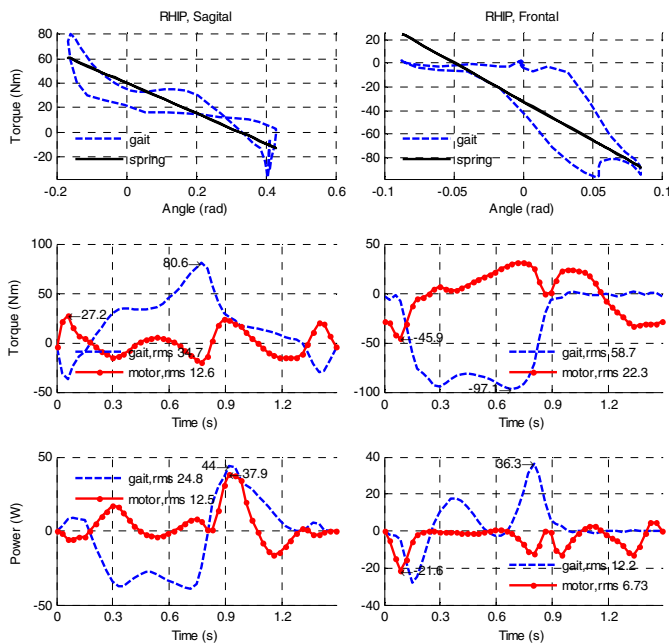


Figure 5. PEA for hip F/E (left column) and A/A (right column); dash line: collected gait data; solid-dot-line: motor torque and power. Both motor torques and powers are reduced by adding a linear parallel spring;

the above reason, we constrained the design to use only one spring. For the knee, no such single spring can achieve a large reduction in the motor torque and power. Hence, no spring is suggested for the knee. For the ankle, a unidirectional parallel spring is suggested. The spring engages when the ankle starts dorsiflexion.

The unidirectional parallel spring reduces the peak motor

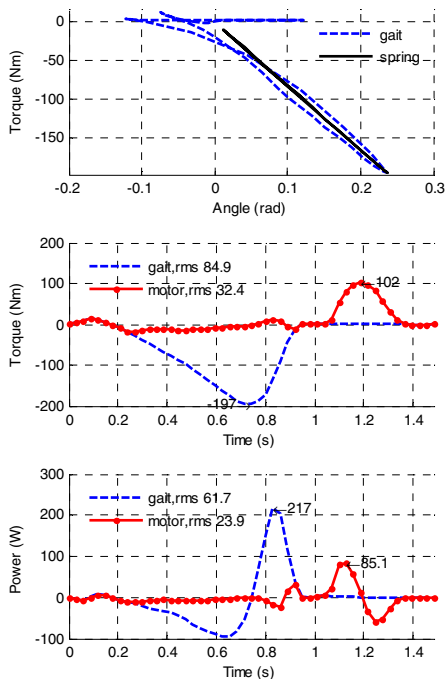


Figure 6. Unidirectional parallel spring for the ankle. PEA reduces the motor torque and power; after adding the spring, the peak motor torque occurs during swing, where the motor has to overcome the spring force.

torque from 197 to 102Nm (by 48%) and rms motor power from 61.7 to 23.9W (by 61%). When a spring is added, the peak motor torque occurs during swing, the motor has to fight against the spring to dorsiflex the foot and maintain foot clearance (Fig. 6).

### 3) Series elastic actuation vs. parallel elastic actuation

According to the above analysis of the results, at the hip, in sagittal plane, the use of a PEA is suggested, since parallel spring reduces both the torque and power demand on motor and gears. At the knee, no spring is suggested. At the hip in frontal plane and at the ankle, both SEA and PEA can achieve large power reduction, but parallel springs can reduce the torque, while series springs cannot. Hence PEA is more favorable for all these four powered DoFs.

### 4) Series elastic actuator for parallel elastic actuation

A parallel spring reduces the torque demand on the motor and gears. A series spring can reduce the peak motor power. We will investigate whether the use of a series elastic actuator to replace the motor in parallel elastic actuation can further improve the energy efficiency.

Using the optimal parallel spring stiffness and residual torques that the motor has to deliver, the optimization process for series elastic actuation is performed for all the joints. For example, the peak motor power for the hip is plotted against the series spring stiffness (Fig. 7). The series spring only achieves marginal reduction in the peak motor power (from 37.9 to 34.6W) at the hip in sagittal plane. At other joints, the series spring achieves either no or marginal improvement. In conclusion, the SEA cannot improve the energy economy further. However, the SEA can lower the mechanical output impedance at high frequency (by decoupling the motor mass from the load) and improve the performance of force control.

## B. Variable stiffness

Previous discussion focuses on one walking speed (0.8m/s). In this section, the sensitivity of optimal spring stiffness to speed and weight are briefly discussed. The dependencies will be illustrated using the ankle actuation.

### 1) Weight dependency

The optimal spring stiffness is linearly related to the total weight of the wearer and the exoskeleton, both for PEA and SEA (Fig. 8), under the assumption given in section III.A.

### 2) Speed dependency

Since we only obtained two sets of gait data, the conclusion we draw can only be based on these two data sets. At the ankle,

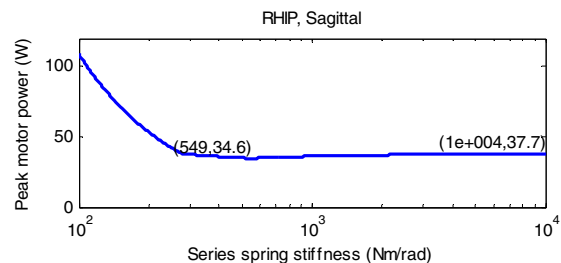


Figure 7. Using a series elastic actuator in parallel elastic actuation only achieves marginal benefit in terms of peak power reduction.

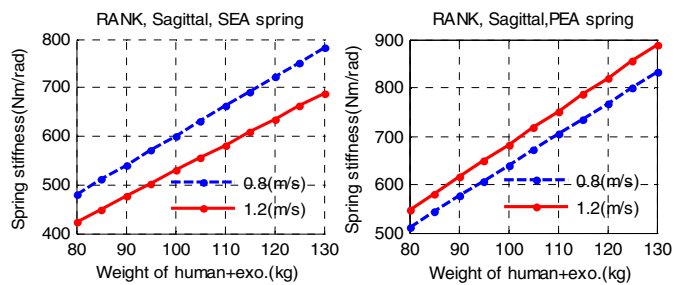


Figure 8. Optimal series and parallel spring stiffness is linearly related to total weight; walking speed affects the optima slightly.

the series spring stiffness reduces as the walking speed increases from 0.8 to 1.2 m/s. In contrast, the parallel spring stiffness increases as the walking speed increases (Fig. 8).

For a constant walking speed, the optimal spring stiffness can be tailored to a specific user (according to his/her body weight), which only requires offline (manual) adjustment of the spring. Increase/decrease in walking speed requires online (motorized) adjustment of the spring stiffness to achieve the maximal torque and power reduction. The implementation of variable stiffness spring likely ends up with complicated and heavy actuators (see e.g. [19]). For a total weight of 80kg, when the speed increases from 0.8 m/s to 1.2 m/s, the optimal parallel spring stiffness increases from 512Nm/rad to 547Nm/rad, that is, a sensitivity of  $(547-512)/512 = 6.8\%$  (Fig. 8 right plot). The optimal parallel spring stiffness is relatively insensitive to speed changes. One constant stiffness parallel spring for both speeds is not the optimal solution, but it can reduce the motor torque and power compared to a joint without any parallel spring.

## V. CONCLUSION

In this paper, the benefit of adding elasticity to exoskeleton actuation is investigated. Two types of elastic actuations, series elastic and parallel elastic actuation, are discussed. PEA reduces the peak torque by 66%, 53%, and 48% for hip flexion/extension (F/E), hip abduction/adduction (A/A), and ankle dorsi/plantar flexion (D/P), respectively, and the rms power by 50%, 45%, and 61%, respectively. SEA reduces the peak motor power by 79% for ankle dorsi/plantar flexion and by 60% for hip ab/adduction; but SEA doesn't reduce the joint torques. To summarize, we suggest using parallel elastic actuation to power the hip and the ankle. At the knee, no spring is suggested. Using a series elastic actuator as the torque source in parallel elastic actuation will not further improve the power efficiency. However, series elastic actuation can lower the mechanical output impedance at higher frequencies (e.g. due to impact) and improve the performance of force control.

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