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A Passive Exoskeleton with Artificial Tendons

Design and experimental evaluation

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Abstract— **We developed a passive exoskeleton that was designed to minimize joint work during walking. The exoskeleton makes use of passive structures, called artificial tendons, acting in parallel with the leg. Artificial tendons are elastic elements that are able to store and redistribute energy over the human leg joints. The elastic characteristics of the tendons have been optimized to minimize the mechanical work of the human leg joints. In simulation the maximal reduction was 40 percent. The performance of the exoskeleton was evaluated in an experiment in which nine subjects participated. Energy expenditure and muscle activation were measured during three conditions: Normal walking, walking with the exoskeleton without artificial tendons, and walking with the exoskeleton with the artificial tendons. Normal walking was the most energy efficient. While walking with the exoskeleton, the artificial tendons only resulted in a negligibly small decrease in energy expenditure.**

Passive exoskeleton; walking; spring; artificial tendon; EMG

I. INTRODUCTION

Different exoskeletons have been developed for medical or military use like the IHMC exoskeleton or the BLEEX [1, 2]. The high power requirements of these exoskeletons negatively influences their operation radius and weight (the IHMC exoskeleton requires an external power supply; the BLEEX has a 30kg on board power supply). Although other, lighter, exoskeletons have been developed, like ReWalk and eLegs [3, 4], the problem of limited energy resources still persists. Alternatively passive exoskeletons have been developed that aim for energy efficient walking. By using only passive elements, the power requirement reduces to zero. For passive exoskeletons lightweight design is even more important. The mass and inertia of the exoskeleton results in a higher energy expenditure [5] that can only be compensated by the passive elements of the exoskeleton. Some of these passive exoskeletons aim at walking in reduced gravity, like the Gravity-Balancing Leg Orthosis and the MoonWalker [6, 7]. Another approach is the use of a mechanism that minimizes the mechanical work at the joints. It is assumed that that by minimizing mechanical joint work the metabolic energy expenditure decreases. During a typical human gait cycle the leg joints perform positive as well as negative work. If the energy dissipated due to the negative work is stored, transferred, and reused, a more efficient gait cycle is possible. In humans and animals multiarticular tendons act as an elastic energy buffer and link between the joints [8, 9]. This mechanism lowers the net joint work. However, the tendons are acting in series with a muscle. Muscular effort is required to tension the tendon. This means that the system can not be used without energy expenditure. The use of elastic elements (artificial tendons) in parallel with the muscle tendon system circumvents this problem. Such a mechanism has been theoretically studied for human walking [10] and robotic walking [11], but the principle has not been applied in an exoskeleton. Goal of this study was to investigate if it is possible to lower the energy expenditure of walking by applying artificial tendons. This is done by designing and evaluating an exoskeleton with artificial tendons.

Figure 1. Left: Schematic drawing of the exoskeleton concept with artificial tendons (black). Right: A photo of the exoskeleton

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II. DESIGN CONCEPT

A. Working principle

The model for the artificial tendons is similar to [10, 12]. The artificial tendon is an elastic cable that spans multiple joints. In this study a configuration was chosen where the artificial tendon spans the hip, knee, and ankle joint. This particular configuration was chosen as a tradeoff between efficiency and complexity [12]. Additionally this configuration is interesting since it has no equivalent muscle tendon combination in the human leg. The artificial tendon is on one end attached to the foot and at the other end to the pelvis (fig. 1). At the joints in between, the artificial tendon has an offset by a lever at the ankle (d_{ankle}) and the hip (d_{hip}), and by a lever with an attached pulley at the knee (d_{knee}, r_{knee}). These offsets cause the artificial tendon to change length when the joint angles change. The elongation and the stiffness (k) of the tendon introduce a force in the artificial tendon. A torque is introduced by the force and the offset from the joint rotation centre. For some joint angles the artificial tendon is shorter than its slack length (l_{slack}) in which case no torques are exerted. The human effort of walking is influenced by the artificial tendons, since the joint torques the human has to provide equal the joint torques required for walking minus the joint torques provided by the artificial tendons. The torque characteristics of the artificial tendon can be changed by altering the joint offsets and the spring characteristics (spring stiffness and slack length).

B. Optimization

The artificial tendons are tuned to minimize the effort of walking. This is done by optimization of the spring characteristics. To perform the optimization the assumptions are made that: 1.) A cost function can be formulated that scales with the energy expenditure of walking, and 2.) The gait kinematics do not change under influence of the artificial tendons. The chosen cost function to minimize is the absolute residual human joint work during one gait cycle:

$$J = \sum_{i=hip,knee,ankle} \int_{t=0}^{T} \left|P_i(t)\right| dt \, .$$

The interval [0, T] is one gait cycle and P_i indicates the power at joint i. The efficiency (η) of the artificial tendons based on the cost function with no support (J_0) and with support (J_s) has been defined as:

$$\eta = 1 - \frac{J_s}{J_0} \, .$$

The optimization was done by a genetic algorithm derived from [13]. For the optimization gait data from an internal gait database was used. The database includes overground walking data of eight subjects (4 male, 4 female, age 24±1) walking at 1.2 m/s. Joint torques and work from the database are normalized to the subject's weight (m [kg]). From this data the average torque profile during one step was calculated. An optimization of the parameters gave a maximum theoretical efficiency of 40.9% (fig. 2). If 100% of these optimized torques were to be provided, the ankle torque would change sign. As a result, the human might need to recruit different, antagonistic, muscle groups while walking with assistance from the artificial tendons if compared to normal walking. Therefore the exoskeleton torques were reduced to 66% of the optimal torques, so this problem no longer occurred. Additionally for the experiments the knee offset was fixed at zero to prevent locking of the exoskeleton in slight hyperextension. Maximum efficiency given these two constraints was 29.9% (fig. 2**).** During all optimizations the spring stiffness was fixed since all joint torques can already be individually adapted by changing the lever arms.

C. Exoskeleton Design

The artificial tendons were implemented in a lower extremity exoskeleton (fig. 1). The joint offsets and the slack length of the artificial tendons are tunable. The exoskeleton is anthropomorphic so that the movements of the exoskeleton pelvis, thigh, shank, and foot segments match their human equivalents. The exoskeleton was attached to the wearer by a backpack resting on the pelvis, straps at the thigh and shank, and shoes at the feet. The thigh and shank are adjustable in length and different straps can be fitted to accommodate users of different sizes. The exoskeleton provides flexion/extension at the hip and knee, dorsi-/plantarflexion at the ankle, and hip ab/adduction. The range of motion of these joints is sufficient to provide walking. The weight of the exoskeleton is approximately 12kg.

III. DATA COLLECTION

The performance of the exoskeleton was evaluated by experimental testing on human subjects. Goal of the experiment was: 1.) To evaluate the effect of the mass and inertia of the exoskeleton; 2.) To evaluate the effect the artificial tendons. This was done by comparing different walking conditions with and without the exoskeleton and artificial tendons. The comparison was made based on energy expenditure and muscle activation. Additionally the force in the tendons was measured and the feet of the subjects were tracked with optical markers.

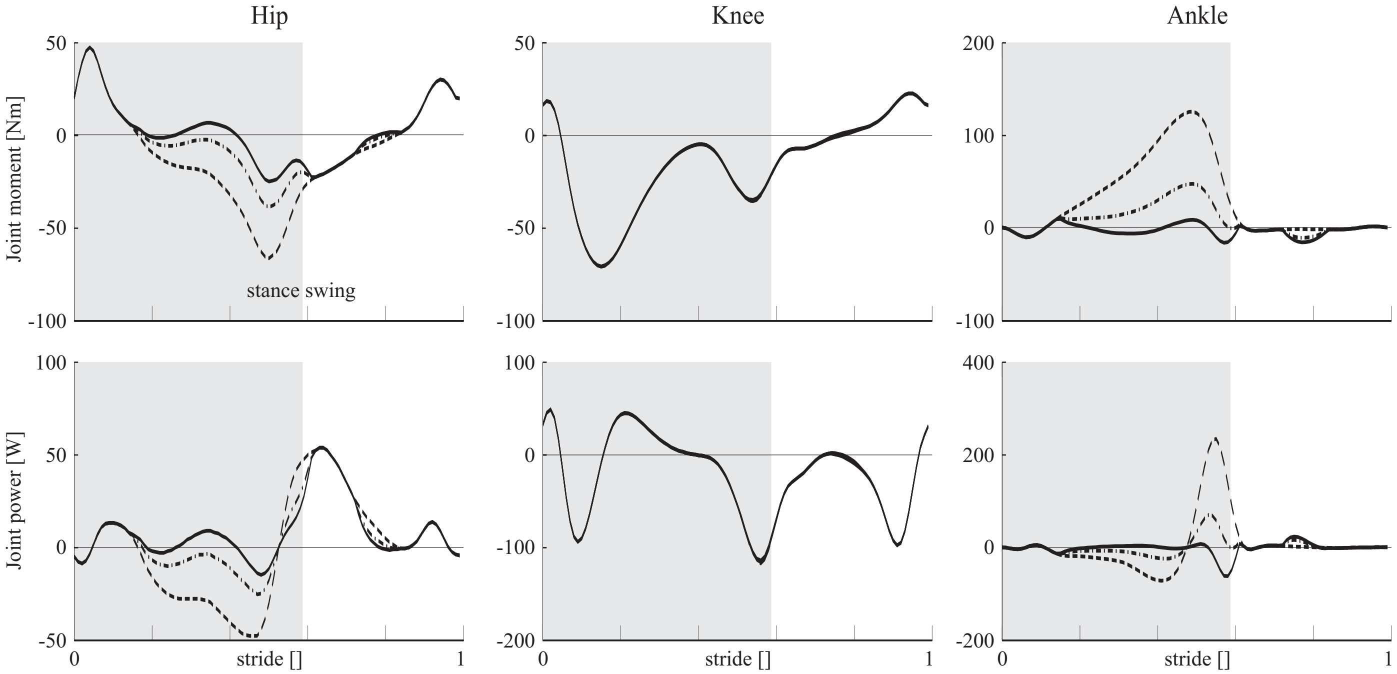
A. Subjects

Nine healthy subjects (8 male, 1 female) between the age of 23 and 64 (mean 31±13) participated in this study. The weight of the subjects was 75.1±6.5kg and their length was 1.79±0.04. All subjects had no symptoms of orthopedic or neurological disorders and gave informed consent before participating in the experiments. Subjects were selected based on length and shoe size since the exoskeleton fitted only a part of the population.

B. Experimental Aparatus and Recordings

1) Artificial tendon force: The tension force in the artificial tendons was measured by two load cells (Futek LTH 300, Irvine, CA).

2) Energy expenditure measurement: A open cirquit respirometry system (Jaeger Oxycon Pro, Viasys Health Care, Warwick, UK) was used to measure the oxygen consumption ($\dot{V}_{O2}$ [ml·s^{-1}]) and carbon dioxide production ($\dot{V}_{CO2}$ [ml·s^{-1}]).



Support ratio	Efficiency	Exoskeleton parameters					
	η[%]	**d_{hip}[mm]**	**d_{knee}[mm]**	**r_{knee}[mm]**	**d_{ankle}[mm]**	**k[Nm^{-1}]**	**l_{slack}[mm]**
66% support	29.9%	50.9	-20	20	143	15000	-862.3
100% support	40.9%	50.9	-20	20	143	10000	-862.3

Figure 2. Optmization results. The upper graphs shows the joint moments for one gait cycle, the lower graph shows the joint powers for one gait cycle. The table gives the values for the exoskeleton parameters. The gait cycle starts at heelstrike and ends at the next heelstrike of the same leg. The data is scaled to a subject mass of 80kg. The knee offset was fixed at zero. The dashed line (— —) shows normal gait data. The solid line line (——) shows residual joint moments and powers when the artificial tendons offer optimal (100%) support. The dot-dashed line (• —) shows residual joint moments and powers with partial (66%) support .

3) Muscle activation measurement: During all trials EMG from eight muscle groups of the left leg was recorded. These muscle groups were: gluteus maximus, gluteus medius, biceps femoris, gastrocnemius medialis, rectus femoris, adductor longus, vastus lateralis, and tibialis anterior. The electrodes were placed acording to the Seniam guidelines [14]. The data were recorded using a Delsys Bagnoli EMG system (Delsys Inc., Boston, MA).

4) Heel position: The position of both heels was recorded using reflective markers. The position of the markers was tracked using a motion capture system (Vicon, Oxford Metric Group, Oxford, UK)

C. Experimental protocol

First the subject's energy expenditure in rest was determined. This was done in a five-minute trial where the subject had to stand still. After the standing trial the exoskeleton was adjusted to the subject size. The artificial tendons were attached to the exoskeleton and the lever arms and slack lengths were adjusted to the values determined in advance by the optimization procedure, based on physical properties of the individual test person. After that, the subject walked two or three practice trials to make sure the exoskeleton had a comfortable fit to the body and the artificial tendons were tuned as planned. During these practice trials small adjustments could still be made. Next, the different walking trials were recorded. The duration of each trial was ten minutes. The last five minutes of the trial were used to determine the median energy expenditure and the last 30 seconds to determine the mean muscle activation. After each trial the subject had a five minute break. In between the trials the exoskeleton was not removed. All trials were performed on a treadmill at a fixed speed of 1.11m/s (4km/h). Three different walking conditions were evaluated during five trials. The different trials are summarized in Table 1.

D. Data processing

1) Energy expenditure: The energy expenditure per kg mass ($\dot{E}$ [W/kg]) for each condition is estimated by a formulae derived from [15]:

$$\dot{E} = \frac{16.48V_{O2} + 4.48V_{O2}}{m} .$$

To estimate the metabolic cost of walking the median value of a condition is taken and the rest rate is substracted.

TABLE I. WALKING CONDITIONS

Acronym	Order	Description
EA1,EA2	1st, 3rd	walking with the exoskeleton and the artificial tendons attached to the exoskeleton.
E1,E2	2nd, 4th	walking with the exoskeleton without the artificial tendons attached to the exoskeleton
NW5	5th	walking without the exoskeleton (normal walking)

2) Muscle Activation: The EMG signal is filtered. First notch filters are applied to remove grid noise (50, 150, 250, 350Hz). Subsequently, the signal is band pass filtered with a second order Butterworth filter between 10 and 400Hz to remove movement artifacts. After that the signal is rectified and low pass filtered (zero phase) at 4Hz. From the last 30 seconds of each condition for each subject an average step is calculated. For each subject the EMG data is normalized to the normal walking condition.

3) Average step: For the analog signals (EMG and tendon force) an average step cycle is calculated. The cycle starts at heelstrike and ends at the subsequent heelstrike of the same leg. Heelstrikes are detected using the data from the optical markers according to the method proposed by [16].

E. Statistical analysis

1) Energy expenditure: Different conditions are compared by performing a paired t-test on the median of the energy expenditure values for each subject.

IV. RESULTS

A. Artificial tendon force

The tensioning of the cable starts and ends at the instants which were well predicted by the optimization. The measured tendon force is lower than the value predicted by the optimization (fig. 3).

B. Energy expenditure

Fig. 4 shows the energy expenditure during the different trials. The energy expenditure of all the trials with the exoskeleton was higher than that of the trials without the exoskeleton. Compared to normal walking, the average increase in energy expenditure of walking with the exoskeleton without the artificial tendons (E2) was 35.9±10.6%. When the conditions with the exoskeleton are compared with each other, the energy expenditure averaged over the subjects for every next trial is lower then the previous ones. This decreasing trend is significant for every combination of trials ($p < 0.05$). To partially eliminate this effect the average of the conditions without the artificial tendons E1 and E2 is taken, and compared to the second condition with the artificial tendons (EA2). Here the worst performing subject has been excluded. The performance of this subject was outside the 99% confidence interval and considered as an outlier. A small significant benefit of the artificial tendons is found (-2.14%, significance $p = 0.014$). For the best subject a higher benefit was measured (-7.12%).

C. Muscle activation

Generally, when walking with the exoskeleton, the EMG values increase when compared to normal walking. The largest difference is noticed in the activation of uniarticular muscles around the hip (gluteus medius, gluteus maximus, and adductor longus) during early stance. The different walking conditions with the exoskeleton are compared to determine the effect of the artificial tendons. Most noticeable is the decrease in the activation of the gastrocnemius muscle (fig. 5).

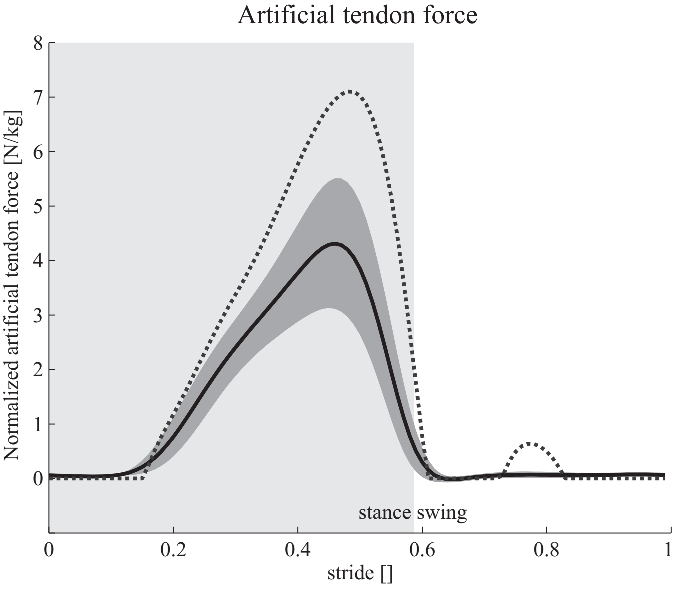


Figure 3. Artificial tendon force normalized to the weight of the subject. The dashed line showed the artificial tendon force predicted in simulation. The solid line shows the artificial tendon force averaged over the subjects measured in experiment. The dark grey area shows the standard deviation.

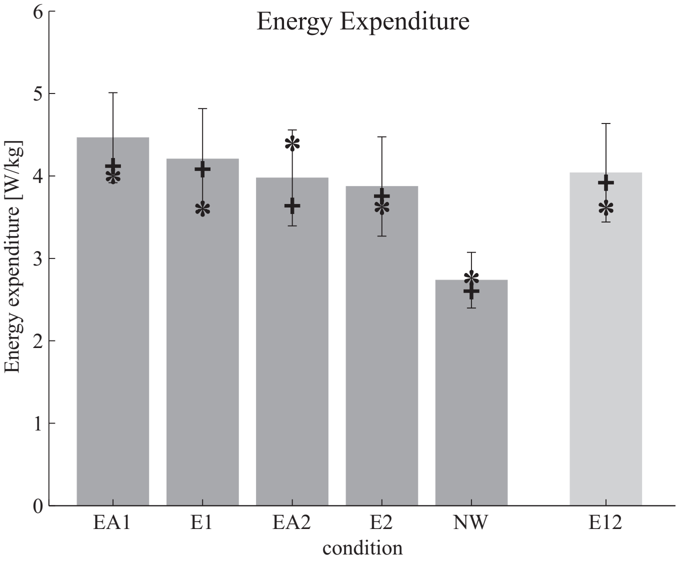


Figure 4. Energy expenditure. Five conditions are shown: Normal treadmill walking (NW), Walking with the exoskeleton and with the artifician tendons 1st and 2nd trial (EA1, EA2), and walking with the exoskeleton without the artificial tendons (E1 and E2). The rightmost column (E12) depicts the average value of E1 and E2. All differences are significant with a paired t-test (p<0.05). + indicates the best performing subject, * indicates the worst performing subject in terms of energy expenditure.

V. DISCUSSION

A. Artificial tendon force

The measured artificial tendon force differs with a roughly constant factor from the simulated artificial tendon force. The measured artificial tendon force is in all cases lower than the value calculated in the simulation. Apparently the subjects received a smaller support torque then expected. Possible

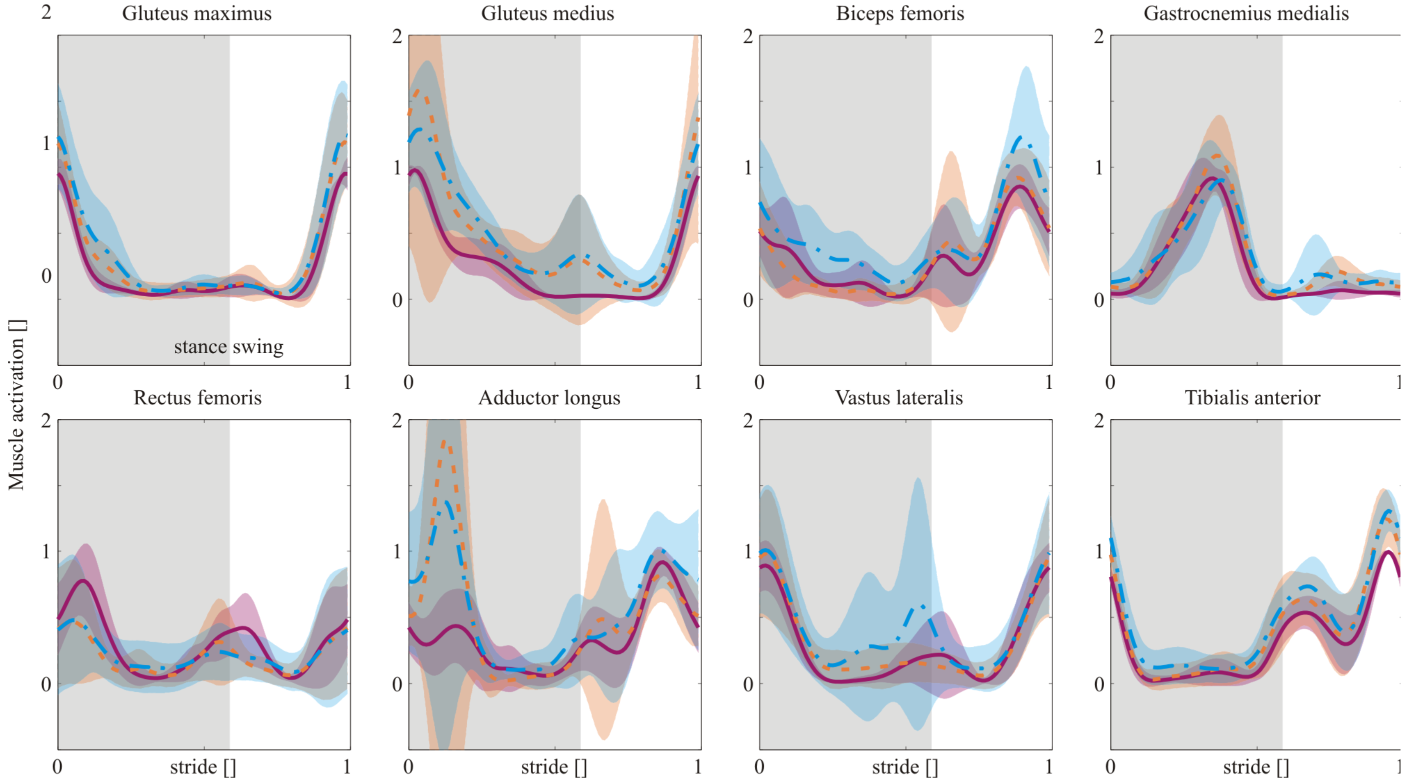


Figure 5. Averaged values and standard deviations across subjects (shaded areas) of the muscle activation patterns from eight subjects of eight muscle groups. The horizontal axis represents one gait cycle where the stance phase is markerd with grey. For each subject the EMG data is normalized between zero and one for the normal walking condition. The data is averaged over the subjects. Three conditions are depicted: Normal walking (purple, solid, ——), walking with the exoskeleton with the artificial tendons(orange, dashed, – –), walking with the exoskeleton without the artificial tendons (blue, dot-dached, • –).

explanations are: the subjects adapt their gait under influence of the exoskeleton with the artificial tendons, and/or the flexible or compressible parts of the exoskeleton (e.g. the foam in the backpack) or soft tissue deform under the loads and act as a serial spring.

B. *Energy expenditure*

The energy expenditure while walking with the exoskeleton (regardless of the artificial tendons) is generally higher than for normal walking. This is to be expected, since the exoskeleton adds mass and inertia to the legs. The subjects were also restrained in their motion since the exoskeleton has no mechanism for endo-/exorotation. When the conditions with exoskeleton are compared a strong effect of time was notable. This might indicate that there is a learning effect. This learning effect seems to overshadow the effect of the artificial tendons. For only one subject the energy expenditure while walking with the artificial tendons was lower than during each of the trials without the artificial tendons.

C. *Muscle activation*

Similar to the results found for the energy expenditure, walking with the exoskeleton has a large effect on muscle activation, when compared to normal walking. A large increase is seen at the uniarticular muscles around the thigh. This might be an indication of co-contraction. Co-contraction might indicate a stiffer walking pattern or a less adapted walking pattern. The large reduction in hip torque as expected from the simulation is not reflected in the muscle activation patterns around the hip. The largest positive effect of the artificial tendons is measured in the gastrocnemius muscle. This is what is expected since the moment arm around the ankle is the largest, and the artificial tendon acts parallel with the gastrocnemius muscle and is active during the same part of the gait cycle.

VI. CONCLUSION

The effect of the artificial tendons on the energy expenditure while walking is much lower than expected from the model optimizations. This limited effect could be caused by: 1.) The learning effect. It takes users more than the measurement time to adapt to walking with the exoskeleton. Thus the still decreasing energy expenditure hides the benefits of adding the artificial tendons; 2.) A significant effect on the gait pattern due to fixation of the exoskeleton to the test person. This could be explained by the increased mass and inertia of the leg as well as the implied constraints; 3.) The supportive torques being lower than expected; 4.) A nonlinear relationship between the reduction in the mechanical work and the reduction in energy expenditure and muscle activation.

VII. FUTURE WORK

Future research will focus on improving the results by improving the exoskeleton and the evaluation methods. Therefore the following steps will be taken:

- Longer testing that is specifically focused on the identification and minimization of the learning effect in order to (partially) remove it from the results.
- Iterative testing and evaluation of the gait kinematics and kinetics. During iterative steps spring stiffness, moment arms, and slack length of the tendon can be changed and better matched with the results from simulation.
- Minimizing the mass and inertia and improving the freedom of movement while walking with the exoskeleton. This will make it easier for the user to adapt to and walk with the exoskeleton, and decrease the effect of the added mass.

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