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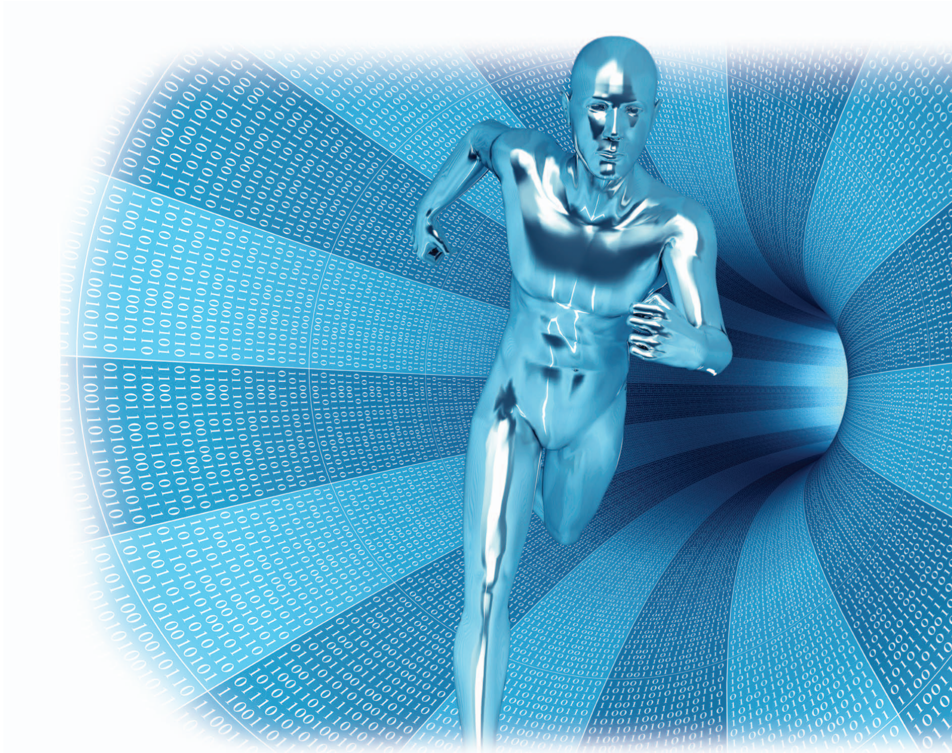
XPED2

A Passive Exoskeleton with Artificial Tendons



Wearable exoskeletons might reduce human effort during walking. However, many of the current exoskeletons rely on heavy actuators and/or external power supplies; this has a negative impact on their efficiency and operation range. As an alternative, (quasi)passive exoskeletons have been developed. One of the proposed passive exoskeleton concepts is the exotendon concept of van den Bogert [1]. In this concept, long elastic cables span multiple joints. The cables can temporarily store and transfer energy between joints. In simulation, the average absolute joint torque can be reduced by 71%. The simulations are based on the hypotheses 1) that the exoskeleton does not influence the joint angles and 2) the total joint torques and a reduction in the human joint torques results in a reduction in the metabolic cost of walking. The goal of this article is to experimentally evaluate the exotendon concept and test the hypotheses underlying it. We implemented the exotendon concept in a lightweight exoskeleton. Experimental results show that the exotendons indeed reduced the average absolute joint torques. However, the exotendons also influenced the joint kinematics, and the metabolic cost of walking did not decrease. Therefore, the underlying assumptions of the exotendon concept are invalid. We also found that, in practice, the amount of support given by the exotendons is limited to about 35% of the theoretical optimal support. For higher levels of support, the motion is hindered and the support is experienced as uncomfortable by the users of the exoskeleton.

Hypothesis

It has recently been shown that walking with an exoskeleton can reduce the metabolic cost of walking [2]. For exoskeletons to become useful in daily life, their power consumption is a key factor. The required power is often provided by batteries, which limits the operating time of the exoskeletons. For the HAL exoskeleton (Cyberdyne, Tsukuba, Japan) and Ekso (Ekso Bionics, Richmond, California), the operating time is approximately 3 h [3], [4].

The high power requirements of exoskeletons contrast with the efficient locomotion found in nature. Human and animal legs possess mechanisms that save energy while in motion. Elastic tendons can temporarily store energy, and multiarticular tendons and muscles can transfer energy between joints [5]–[7].

The model optimizations of van den Bogert suggest that human joint torque and power can be reduced by placing elastic structures, called exotendons, parallel to the leg [1]. These exotendons have a similar function to biological uni- and multiarticular tendons. Simulations suggest that the human joint torque, the torque provided by the leg muscles, can be reduced by 21% with uniarticular exotendons at the ankle. This reduction increases to 46% if triarticular exotendons are used that span the hip, knee, and ankle. For more complex configurations with multiple exotendons per leg, the predicted reduction increases to 71%. The hypotheses underlying the exotendon concept are: 1) the exotendons do not influence the joint angles and total joint torques (the sum of the human joint torques and the exoskeleton joint torques), and 2) a reduction in the human joint torques results in a reduction in the metabolic cost of walking.

The first hypothesis has been tested for the hip and ankle separately with uniarticular powered exoskeletons [8], [9]. These studies show that joint total torque patterns do not change when an external support is provided but joint angle patterns do change. Two exoskeletons with uniarticular exotendons demonstrated a relative reduction in metabolic cost.

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Wearing these exoskeletons without the elastic elements increased the walking metabolism, which was only partially compensated when the elastic elements were added to the exoskeleton [10], [11].

The goal of this article is to experimentally evaluate the exotendon concept of [1]. In our study, we use triarticular exotendons as a compromise between predicted reductions and complexity. Our experiment is built to test the previously mentioned hypotheses underlying the exotendon concept. Based on these hypotheses, we expect that our exoskeleton reduces the metabolic cost of walking. This article first describes the design of the exoskeleton and then describes the experimental evaluation of the exoskeleton.

Design

Working Principle

The exoskeleton uses a mechanism of springs, cables, lever arms, and pulleys to temporarily store and transfer energy between joints. The working mechanism is similar to that of [1], and it is shown in Figure 1. The characteristics of the support given by the exotendon can be varied. The slack length, lever arm lengths, and spring stiffness of the exotendon were optimized so that the average absolute human joint torque was minimal. In the optimizations, the human joint torque was the torque observed in a typical gait pattern minus the exotendon torque summed over the hip, knee, and ankle (Figure 2).

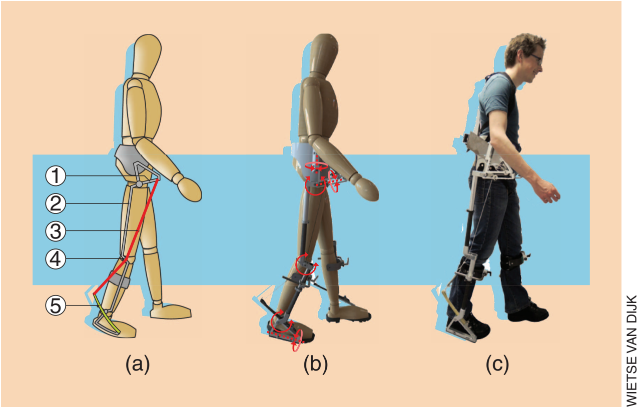

Figure 1. The XPED2 exoskeleton. (a) The working principle: The main functional part of the exoskeleton is the exotendon 3), a cable that spans between a lever at the pelvis 1), via a pulley at the knee 4), to a leaf spring at the foot 5). The leaf spring at the foot accounts for the elasticity of the mechanism. Since the cable has an offset from the joint centers, the deformation of the spring, and thereby the force in the cable, depends on the joint angles. In this configuration, hip extension and ankle dorsiflexion will tighten the cable and hip flexion and ankle plantarflexion will loosen the cable. The movement of the knee has almost no effect on the tension of the cable. For some combinations of joint angles, the cable is slack, and there will be no force in the cable. The force in the cable multiplied by the offset from the joint gives the moment that the exotendon exerts around that joint. The exoskeleton is connected to the human via a rigid frame 2) with connections at the pelvis, shank, and foot segments. (b) The computer-aided design model of the XPED2 exoskeleton. The XPED2 has six degrees of freedom (DoF) per leg: flexion/extension, ab/adduction, and endo/exorotation at the hip; flexion/extension at the knee; and plantar/dorsiflexion and pronation/supination at the ankle. (c) One of the developers wearing the XPED2 exoskeleton.

Exoskeleton

The proposed mechanism was realized in the XPED2 exoskeleton (Figure 1). The design is anthropomorphic and has six degrees of freedom (DoF) per leg. These DoF were possible trough serial hinge joints from the pelvis attachment to the foot attachment. An additional attachment to the human body was made at the shank. The total mass of the exoskeleton is 6.91 kg and is distributed over the pelvis (3.57 kg), the thighs (2×0.40 kg), the shanks (2×0.72 kg), and the feet

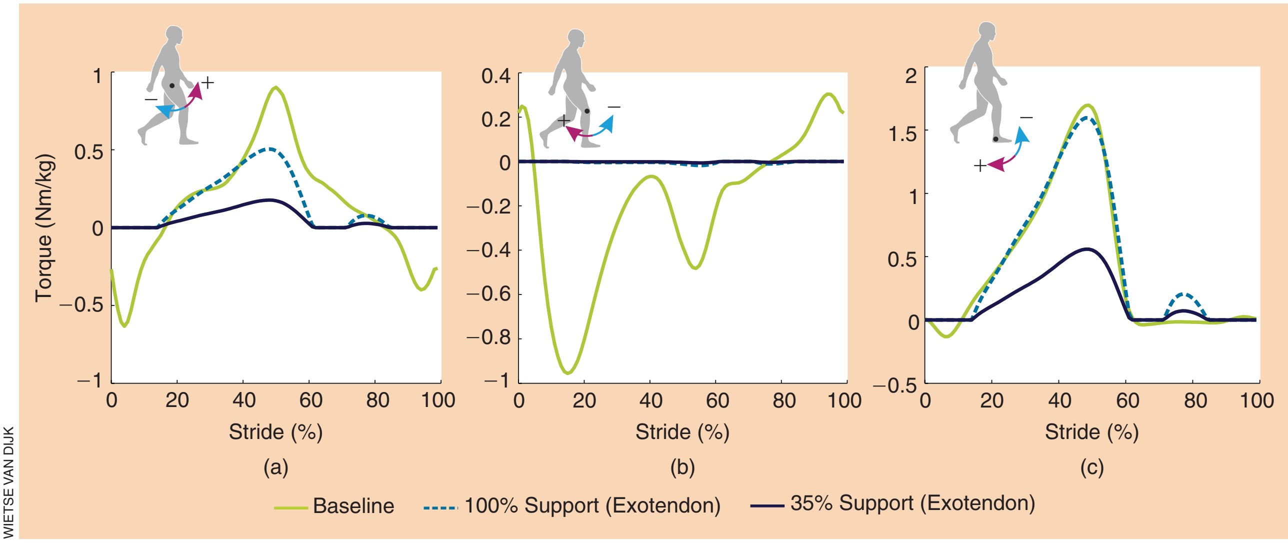

Figure 2. The exotendon optimization results for (a) the hip, (b) the knee, and (c) the ankle. The torques exerted by the exoskeleton are optimized to match those normally observed in human gait (green lines). The optimization assumes that total torque (human + exotendon) is constant across the different conditions. The optimal (100%) exotendon torques are shown with the dashed blue lines. Preliminary tests have revealed that users are uncomfortable if very high torques are exerted on the body. The settings used for the XPED2 exert 35% of the optimal torques (solid blue lines) to ensure that users can walk comfortably with the exoskeleton. The optimizations are done in the sagittal plane since this is the dominant plane for walking.

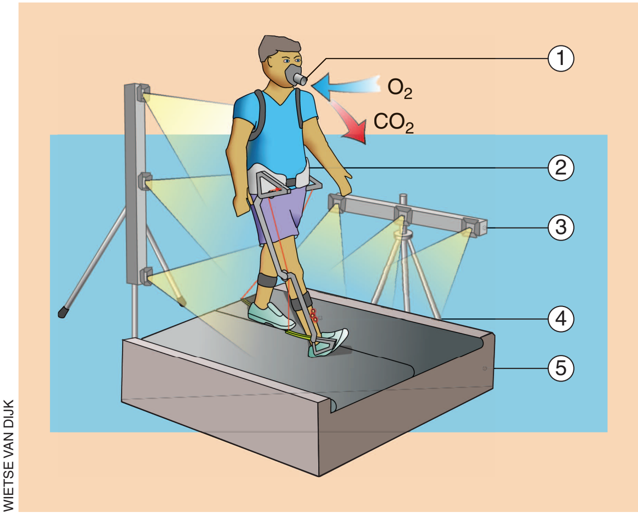

Figure 3. During the experiments, the following measurements were taken. The metabolic power is measured by a respiratory measurement system 1) two subjects were measured using a K4B2 system, Cosmed, Pavona, Italy; four subjects were measured using a Jaeger Oxycon Pro system, Viasys Health Care, Warwick, United Kingdom that measures the gas flow and composition. The force in the cable was measured by load cells at the end of the exotendon 2) LCM200, Futek, Irvine, California, United States. The kinematics were measured by four trackers 3) VZ4000, Visualeyez, Burnaby, British Colombia, Canada tracking light-emitting markers 4) placed on the exoskeleton and the human. Ground reaction forces were measured with a dual-belt instrumented treadmill 5) Y-mill, Forcelink B.V., Culemborg, The Netherlands.

(2×0.55 kg). The added mass will increase the energy consumption. Given the mass distribution and the empirical relations of [12], the estimated increase will be about 14%. The lever arms at the pelvis and the foot and the slack length were adjustable so that the exoskeleton characteristics could be matched to the optimization results. The length of the shank and thigh segments was adjustable, and different shoes were available to adjust for the subject's size. The tendons were made from Dyneema cable. The elasticity of the system was achieved by making the ankle levers elastic. The levers are of a custom design glass fiber leaf spring with unidirectional fibers.

Experiment

Subjects

Six subjects volunteered and gave a signed informed consent before participating in the study. Five were male and one female. The mean age was 21 years, three months (standard deviation 11 months), the mean height was 1.80 (standard deviation 0.04) m, and the mean weight was 72.9 (standard deviation 7.3) kg in weight. The subjects were recruited from the Dutch student population. The subjects were selected if they were in good physical condition without gait abnormalities and if they could fit the exoskeleton.

Data Recordings

The measurement setup is shown in Figure 3. The metabolic power was recorded using a respiratory measurement system. The motions of the lower body were recorded with markers bilaterally on the foot, shank, thigh, and pelvis and on the right side of the exoskeleton. Ground reaction forces were recorded by a dual-belt instrumented treadmill. The forces in the exotendons were recorded with load cells.

Protocol

Three walking conditions were evaluated during the experiments: 1) baseline: the subject walks without the exoskeleton, 2) no support: the subject walks with the exoskeleton, but the exotendons are not tensioned, and 3) support: the subject walks with the exoskeleton and the exotendons are tensioned, so the user gets support from the exoskeleton. All measurements were performed at a fixed treadmill speed of 1 m/s. The subject walked with the exoskeleton during three sessions on separate days. The first two sessions were practice sessions. These sessions were held to minimize the effect of learning that was noted in evaluation of a previous version of this exoskeleton [13]. On these days, the subject could practice for 45 min at a self-selected speed and at least 10 min at the speed of 1 m/s. During the first sessions, subjects were allowed to take breaks at random intervals. Measurements were taken during the third session. Each session started with fitting the XPED2 to the body and adjusting the settings of the lever arms and slack lengths to their desired values.

During the last sessions, five walking trials were conducted: 1) two no support, 2) two support, and 3) one baseline trial. The order of the trials was quasi-random. The baseline trial was randomly assigned to the start or the end of the walking trials. The order of the support and no support trials was alternating, with the first trial randomly a support or no support trial. Each walking trial lasted 10 min. Additional trials were needed for the inverse dynamics analysis: the recording of a standing pose and identification of anatomical landmarks with a probe. In addition, the metabolic power at rest was recorded for 5 min while the subject was sitting in a chair.

Data Analysis

Kinematics and Kinetics

The kinematic and kinetic analysis was performed for four of the six subjects. Joint kinematics were calculated from the marker data. Joint kinetics were obtained from the marker data, ground reaction forces, and exotendon forces with inverse dynamics [14] using BodyMech (VU Medical Center Amsterdam, The Netherlands). The XPED2 is anthropomorphic, and we assumed that the mass of the segments of the exoskeleton was rigidly connected to the body parts to which they were parallel. A median step was calculated from a 2-min sample taken from the end of each trial. The data were split into individual strides based on the heel-strike events. The heel-strike events were derived from the vertical ground reaction force, measured by the treadmill. The median stride time was calculated for all six subjects. The differences between conditions were compared by the average over the subjects. A statistical analysis was done by a paired t-test.

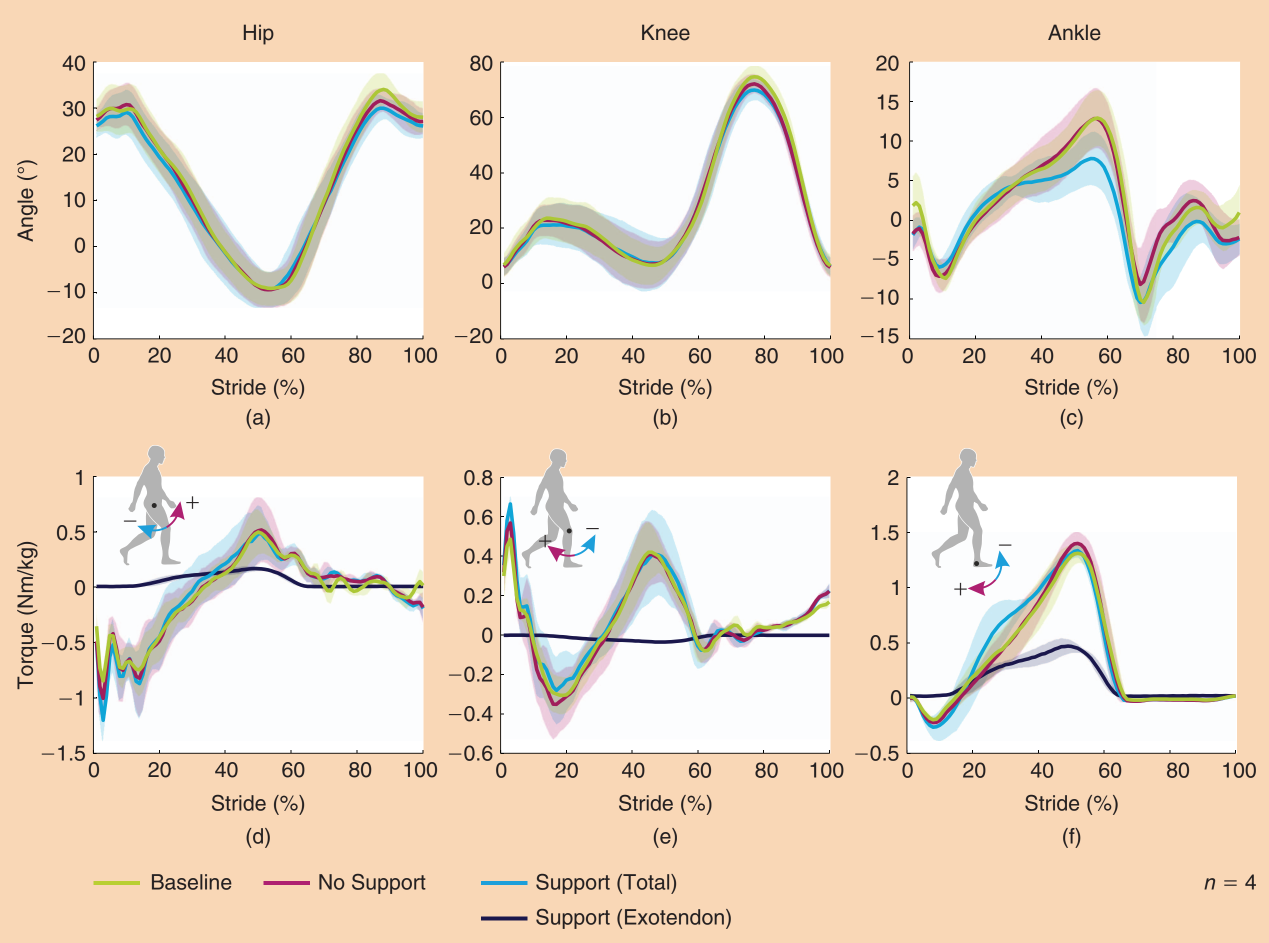

Figure 4. (a)–(f) Walking kinetics and kinematics. The lines represent the average over the subjects. The average strides start and end with a heel strike. The top row shows the joint angle for the (a) hip, (b) knee, and (c) ankle joint. For the hip and knee, no significant changes were found between the conditions. The maximal ankle dorsiflexion angle during support decreased by 5.0° compared with no support and 5.1° compared with the baseline condition. The bottom row shows the joint torque for the (d) hip, (e) knee, and (f) ankle joint. The shaded areas are the standard deviations.

Metabolic Cost

The metabolic power was calculated from the respiratory data. The following empirical relation for the metabolic power ($\dot{E}[\mathrm{W/kg}]$) was used [15]:

$$\dot{E} = \frac{16.48 \cdot 10^3\, \dot{V}_{O_2} + 4.48 \cdot 10^3\, \dot{V}_{CO_2}}{m}, \tag{1.1}$$

where $\dot{V}_{O_2}$ [L/s] and $\dot{V}_{CO_2}$ [L/s] are the oxygen uptake and carbon dioxide production, respectively, whereas m [kg] is the mass of the subject. For all reported metabolic powers, the metabolic power at rest has been subtracted. The differences between conditions were compared by the average over the subjects. A statistical analysis was done by a Wilcoxon signed rank test.

Results

Kinetics and Kinematics

Figure 4 shows the walking kinetics and kinematics. The exotendons changed the walking kinematics. The maximal ankle dorsiflexion angle during support decreased by 5° compared with the no support condition and 5.1° compared with the baseline condition ($p < 0.05$). The average ankle plantarflexion torque increased in the support condition by 0.025 Nm/kg compared with the no support condition and 0.022 Nm/kg compared with the baseline condition. This increase was not present for all subjects. The differences in kinematics and kinetics between the conditions were small for the hip and knee. Differences in stride time between the conditions were also small (maximal 1.36%) and not significant ($p > 0.1$).

The measured average absolute human torque was compared with the estimated value from the optimization (Figure 5). The optimization results predict a decrease in the average absolute joint torque of 17% for a subject of 70 kg (this value ranges between 16.8 and 18.3% for subjects between 60 and 100 kg). Experimentally, we found a reduction of 12.1% in the support condition relative to the no-support condition ($p = 0.089$), which could almost entirely be contributed to the ankle torque. The reduction for the ankle only was 29.0% ($p = 0.057$). Apart from the differences between conditions, the human torques in the experiment differ from the human torques in the optimization. The data for

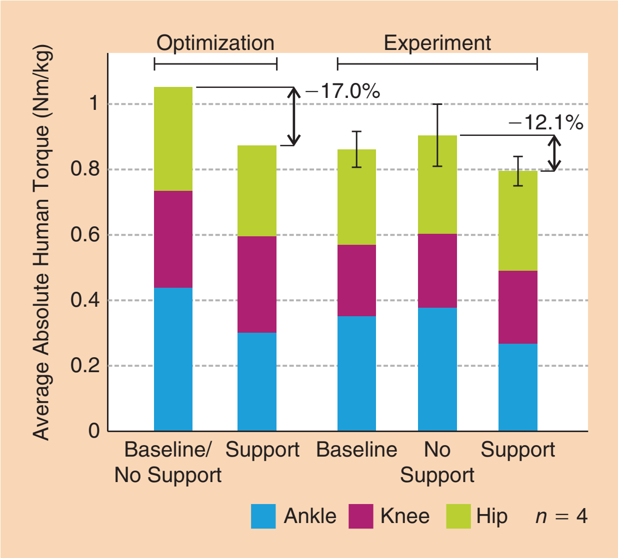

Figure 5. The mechanical performance: the average absolute human joint torque in different conditions. On the left, the optimization results. The optimization predicts a decrease in average absolute joint torque of 17%. Note that the effect of the added weight is not considered in the optimization, and, therefore, the baseline and no-support condition could not be discriminated. On the right, the experimental results. The difference between the support condition and no-support condition is 12.1%.

the optimization were obtained from different subjects in a different lab, causing intrasubject differences. A video of a subject walking in the support condition is available in the supplemental material in IEEE *Xplore*.

Metabolic Cost

The metabolic power of the various subjects is shown in Figure 6. For all subjects, the metabolic cost of walking during the baseline condition was the lowest. The metabolic cost in the no-support condition was significantly higher (21.2%) than in the baseline condition ($p < 0.05$). The added mass explains

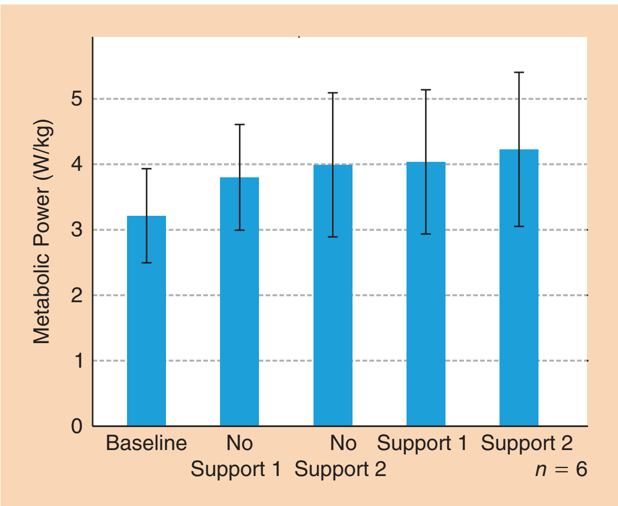

Figure 6. The metabolic cost: the metabolic power for the different walking conditions. The bars represent the average over the subjects with the standard deviations.

most of this increase since the estimated increase in metabolism due to the added mass was 14%. In contrast with our predictions, the average metabolic cost in the support condition was 6.1% higher than in the no-support condition ($p = 0.052$).

Discussion

We built the XPED2 to test the following hypotheses: 1) the exotendons do not influence joint angles and total joint torques and 2) a reduction in the human joint torques results in a reduction in the metabolic cost of walking. Although our experiment is based on data from a small number of subjects, we could identify some clear trends. Our experiments demonstrated that deviations from normal walking occur, and thereby we disproved our first hypothesis. The changes in ankle motion were also observed in [9]. Still, the exotendons did reduce the average absolute human joint torque. The reduction in average absolute human joint torque did not result in a measurable reduction in the metabolic cost—there was even an indication that the metabolic cost increased. This contradicts the second hypothesis. However, the effect could have been small and unnoticed. The exotendons did only provide a small amount of support (12.1%). When changing the exotendon parameters or configuration, this number might be increased. Given the changes observed in the walking pattern, it is unlikely that a 71% reduction in average absolute joint mentioned in [1] is feasible. The preliminary pilot trials taught us that subjects were uncomfortable with higher amounts of support.

Conclusions

Our exoskeleton has shown the contrast between the theory and the experiment and thereby stressed the importance of experimental evaluation of exoskeleton designs. We compared our results to experimental results obtained with other exoskeletons for walking augmentation (Table 1). Although some exoskeletons have shown a (relative) reduction in metabolic cost, there is no general consensus on how to reduce metabolic cost most effectively.

The hypothesis that the metabolic cost is reduced when the absolute joint torque is reduced is valid for isometric contractions, but in walking there are additional effects that might interfere with this relation. Energy storage and transfer between joints is already partly covered by the human tendons and biarticular muscles. Adding exotendons might interfere with these energy saving mechanisms. In humans, it has been shown that the Achilles tendon stiffness is optimal in the sense that muscle work during walking is minimal [5]. In hopping experiments, it has been shown that adding a parallel spring reduced muscle force but increased muscle work [16]. The observed changes in kinematics and kinetics might be explained by the fact that the support by the exoskeleton enforces a new equilibrium.

The metabolic cost of walking could be partially explained with the model used by [17]. The model predicts that the metabolic cost of walking emerges from the need to compensate for energy losses after impact. This is in line with experimental

Table 1. A comparison of the metabolic cost of walking for different exoskeletons.

Exoskeleton	Passive?	Number of Subjects	Weight and Constraints	Support
XPED2 (this article)	Yes	6	21.2%	6.1%
XPED1 [13]	Yes	9	35.9%	−2.1%
Wiggin et al. [10]	Yes	3	–	−10%
Walsh et al. [11]	Quasi*	1	24%	−12.9%
Malcolm et al. [2]	No	10	14.3%	−17.2%
Sawicki et al. [21]	No	9	8.3%	−11.5%
Norris et al. [22]	No	9**	16.2%	−13.9%
Wehner et al. [18]	No	1	12.8%	−10.2%

*Quasipassive means that energy is used for control, but there is no mechanical energy added to the system.
** The results for the young subjects are given
The column weight and constraints give the relative changes in metabolic cost due to the added weight and imposed constraints. The column support gives the relative changes due to the support (no support versus support condition). Positive values mean an increase in metabolic cost.

results of [2], where the highest reduction in metabolic cost was observed when the exoskeleton solely provided positive power. This result undermines the passive exoskeleton concept since passive exoskeletons are energy neutral at best. However, it has been shown that relative metabolic cost reductions with a passive exoskeleton are possible [10]. Different from our mechanism, this exoskeleton has a clutch that engages the elastic element. This allows for a better timing of the support, which might be essential to reduce the metabolic cost of walking [2], [18].

Despite the limited effect of exotendons on the metabolic cost of walking, we see potential for the application of exotendons in exoskeletons. Elastic elements in combination with actuators can lead to smaller power requirements on the actuation side [19] and give the opportunity to place the actuators on a more proximal, and thus more metabolically beneficial, place on the leg [20]. Combined with actuators, exotendons can contribute to elegant and lightweight exoskeleton designs.

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