Effects of Passive Ankle Exoskeleton on Human Energy Expenditure: Pilot Evaluation

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Abstract. Exoskeletons can be utilized for rehabilitation purposes as well as for assistance and augmentation of motion of patients with disabilities, workers, the elderly and even healthy people. Compared to powered solutions, unpowered passive exoskeletons have been shown to have significantly higher chances of end user acceptance, because of simpler design, no complex electronics and potentially lower cost. In this paper we present the results of a flat walking test using an unpowered passive ankle exoskeleton. Important exoskeleton aspects such as ergonomics, comfort, and robust design are outlined and areas for improvement are highlighted. The paper also presents the results of the evaluation of the exoskeleton device in a pilot study, where its physiological effects are assessed for four participants via measurements of oxygen consumption and EMG muscle activity during five 10-min walking sessions under different conditions. Results show that significant metabolic cost reduction can only be achieved with a proper mechanism spring selection.

Keywords: passive exoskeleton, passive orthosis, metabolic cost, efficient walking, energy cost reduction

1 Introduction

Exoskeleton technology has been, since its beginnings roughly 50 years ago [1], witnessing intense development [2]. Recent advances in actuators, sensors, materials, batteries and computer processors further intensified the development of wearable technology [3]. Furthermore, the increasing longevity and declining birthrate in post-industrial societies will in the future affect health spending, retirement policies, use of long-term care services, work-flow policies, and income [4]. This has led to an increase in the research of exoskeletons.

An important part of everyday human activities is walking, that apart from providing individual social independence, also has significant effects on human near-term and long-term health [5]. Consequently, a considerable amount of exoskeleton development is now focused on the augmentation of lower extremities to assist in walking. One of the aims of lower extremities exoskeleton research is the reduction of metabolic expenditure, which can be achieved in different ways.

Ferris et al. [6] have shown that muscle effort reduction of the hip and in the ankle can be achieved by providing assistance to the human hip only. Gams et al. [7]

showed that statistically significant decrease in metabolic cost can be observed when using a robotic knee exoskeleton during periodic squats, if a proper control method is applied. Periodic squatting was used to roughly approximate human walking over a rough terrain. In a study by Galle et al. [8] it was shown that the metabolic reduction gets better with practice, because of neuromotor adaptation. Humans adapt other muscles on other joints, apart from the one that is actuated with the use of an exoskeleton, which then significantly reduces muscular activity in all leg muscles.

Jimenez-Fabian et al. [9] and Gams et al. [7] have shown that besides the exoskeleton design, the chosen control strategy can have a big impact on the end resulting metabolic reduction. Proportional myoelectric control shows some distinctive advantages over other types of control [10]. To accommodate uncertain dynamics of the human-robot system, iterative learning can prove to be very effective in learning of control of cyclic tasks [11]. The chosen control strategy can also have an effect on the human adaptation to the exoskeletons [12].

Another way to achieve metabolic reduction is with the use of passive solutions, that do not use motors, batteries or controllers. These solutions can prove to be superior in some aspects to actuated solutions, because their weight, complexity and price makes them simpler, more affordable and perhaps easier to introduce into everyday lives. Sawicki et al. [13] outlined the importance of tendons in human walking that enhance the efficiency of the ankle joint work, by storing and releasing elastic energy in the Achilles tendon during each step.

Collins et al. [14] presented an unpowered exoskeleton that achieved metabolic reduction of human walking with a completely passive device. The exoskeleton they presented used a spring between the insole and the frame around the calf. The spring is attached to a mechanical clutch, which allows the spring to store energy at some parts of the gait, release it at an appropriate time, and not hinder the motion while the leg is in the swing phase.

In this paper we present our implementation of such an unpowered exoskeleton. The goal of our research was to: 1) get insight into the workings of the unpowered exoskeleton; 2) to confirm if the exoskeleton reduces metabolic cost as intended; and 3) to narrow the range of the appropriate spring stiffness that can achieve the reduction of metabolic cost.

The exoskeleton presented here uses the same principle as the original [14]. We replicated the exoskeleton and performed tests in order learn more about the concept and to find ways to improve the current design. We used different materials and introduced some minor changes in the design to make production cheaper and assembling more convenient. A small study was performed to test the exoskeleton and to see if a reduction in metabolic cost of walking can be observed.

2 Exoskeleton Design & Mechanics

We designed a prototype passive ankle exoskeleton based on the work of Collins et al. [14]. Our goal was to achieve a working prototype in as few iterations as possible. With this in mind, we incorporated some changes and used different materials to make the production faster, cheaper and more convenient. The working principle, however, stayed the same as in the original.

The exoskeleton comprises of an insole part, a simple hinge joint, a stainless steel frame, an attachment strap, a mechanical clutch and a linear tension spring, as shown in Fig. 1. The frame of the exoskeleton is connected to the insole with a simple hinge joint. The user attaches it to the leg with a strap.

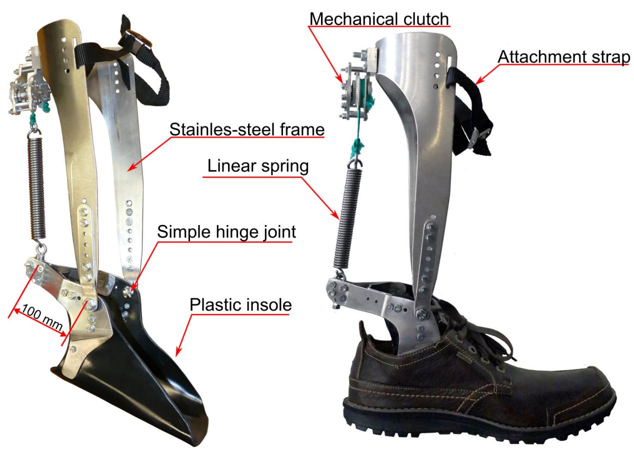

Fig. 1: Our prototype version of the unpowered passive ankle exoskeleton.

A separate mirrored exoskeleton was made for the other leg. Plastic material was used to construct the insole part using a heat-forming process and reinforced with a stainless steel insole frame. The insole shape was designed to be comfortable and to still fit into an average adult male shoe. Thus, the users could use their own shoes.

Stainless steel, which is rigid and light enough, was used to construct the leg frame. A material with a larger strength/weight ratio could also be used. Due to the relatively high strength/weight ratio and accessibility of this material, we performed no weight/strength optimization. The exoskeleton presented in this paper weights 0.8 kg. We believe, however, that the weight could be significantly reduced in next design iteration.

The mechanical clutch design is largely based on the original [14], with some modifications to make the production and assembly more convenient. Its components were manufactured out of aluminium, stainless steel and plastic, using laser-cutting process and 3D-addition technologies. The working principle of the clutch stayed the same as in the original.

Throughout the construction process we found out that the following aspects of exoskeleton design have to be taken into consideration. To maximize the effectiveness, commercial value and potential user acceptance, the exoskeleton should be as simple as possible. The possibility to use his own shoes and donning the exoskeleton over clothes makes an impression that the exoskeleton is very simple and easy to use. The exoskeleton should be easy to don-on or don-off. Ergonomic design is an important factor of comfort and safety. The exoskeleton should not impede on the user, which means that its kinematic obstruction should be minimal. From where follows, that the exoskeleton should have an appropriate size, movement range and enough kinematic degrees of freedom. It should also have minimal weight and minimal inertia, whereas it should still be sturdy enough for the intended task.

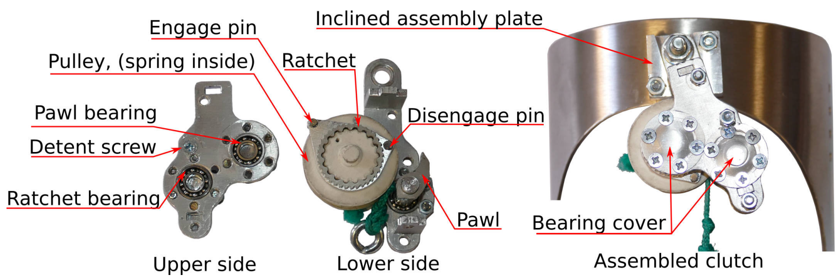

Fig. 2: Our mechanical clutch assembly.

In the current design of the passive exoskeleton we found out that the pawl engaging and disengaging is an important factor of proper operation. During the gait phase, locking of the pawl relies only on gravity. Because the pawl is very small, the force of gravity is not very large. As a result, the working of the clutch is not very robust. Simply the friction in the pawl bearing can be large enough to prevent the pawl from engaging correctly. To accommodate this, we designed the shaft to only loosely fit the bearing. This makes the use of a rolling bearing unnecessary.

To magnify the effect of gravity on the movement of the pawl, we added an inclined assembly plate, that keeps the mechanical clutch at an angle to the exoskeleton frame. This increased the reliability of the clutch. Using a different design approach, the usage of this plate could be prevented. The size of the ratchet teeth is an open issue. The one in our prototype should have larger teeth. This would prevent unwanted disengagement of the pawl during the walk, which proves to be problematic, since the disengagement of the clutch affects the kinematics of the walk. The user, relying on the pawl to engage, can stumble because he does not expect the sudden release of exoskeleton assistance.

Another issue is timing, which is determined by the positions of the engage pin and disengage pin, and depends also on the length of the rope. The problem is that the exchange of the spring with a spring of different stiffness can compromise the timing, if the spring does not have an appropriate length, keeping in mind that springs of different characteristics normally have a different length, if the outer diameter of the spring and the thickness of the wire stays the same. The original exoskeleton from [14] had a system that could adjust the length of the rope with a screw. This is not very practical, especially if the exchange needs to be very fast. The use of a rope is also problematic since it is hard to tie it at a specific length.

3 Metabolic Cost Study Design

A randomized cross-over study was employed in order to evaluate the efficiency of the passive ankle exoskeleton. Four healthy males volunteered to participate in the study. They were informed about the procedures beforehand and gave a free informed consent. The baseline of participant characteristics are outlined in Table 1. The participants were asked to perform five 10-min series of walks on a horizontal treadmill at a constant speed of 4 km/h. Between the walking sessions, the subjects rested for at least 15-min, sitting on a chair.

The five series consisted of a referential exoskeleton free walk; a walk with a springless exoskeleton; and three walks with the exoskeleton on both legs, but with spring of different characteristics (5 N/mm, 12 N/mm and 20 N/mm). The order of sessions was randomized for each participant.

Table 1: BASELINE PARTICIPANT CHARACTERISTICS

	Mean	Standard Deviation
Age (years)	30	7
Weight (kg)	76	5
Height (cm)	178	3
Body Mass Index (kg/m^2)	24	1

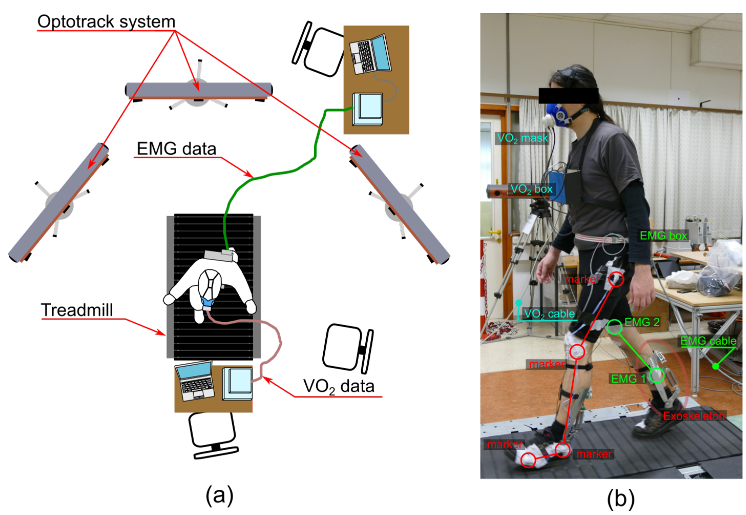

Fig. 3: (a) Measurement system, (b) participant during the test.

To evaluate the exoskeleton operation and its effect on the user, we measured the average *oxygen consumption* (*VO_2*) of each user using a portable metabolic cart (*$K4b^2$*, Cosmed Italy). The whole measurement system is presented in Fig. 3(a). To asses the overall muscle effort during walking, we also measured the EMG activity of the Soleus muscle on the right calf of each participant. The signals were rectified and integrated for a time of 7-min, out of a 10-min walking session, to exclude the starting and ending discrepancies, and normalized to the normal walking value, as shown in equation (1).

$$E_{\mathrm{ki}} = \int_{1.5\,\mathrm{min}}^{8.5\,\mathrm{min}} |U_{\mathrm{ki}}(t)|\,dt \Big/ \int_{1.5\,\mathrm{min}}^{8.5\,\mathrm{min}} |U_{\mathrm{kn}}(t)|\,dt \cdot 100\%, \tag{1}$$

where E_{ki} is the integrated EMG response. $U_{\mathrm{kn}}(t)$ is the EMG voltage with respect to time, where n denotes normal walking session for a specific participant k and i denotes the walking session for five different boundary conditions. The EMG signals were recorded using the surface electrodes and the DataLog (Biometrics Ltd.) data acquisition device. The oxygen consumption and EMG activity were then compared between groups of participants and different boundary conditions using average value (2) and sample standard deviation (3) for four participants:

$$\bar{E}_{\mathrm{i}} = \frac{1}{4}\sum_{\mathrm{k}=1}^{4} E_{\mathrm{ki}}, \tag{2}$$

$$S_{\mathrm{Ei}} = \sqrt{\frac{1}{3}((E_{\mathrm{ki}}) - \bar{E}_{\mathrm{i}})^2}, \tag{3}$$

where again the k represents a participant and i the chosen boundary condition.

4 Results

Figs. 4, 5 show the graph of measured VO_2 response and EMG activity of muscles depending on the boundary conditions. The measurement is divided into the average values for the first pair, then for the second pair and then for all four participants.

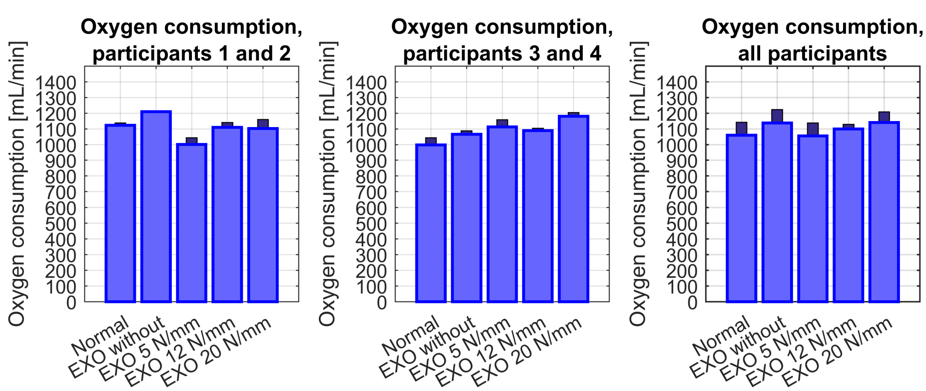

Fig. 4: Average Oxygen consumption and sample standard deviation for the first and second pair, and all of participants.

For participants 1 and 2, as shown in the Fig. 4, a decrease of average metabolic cost (-10%) can be observed compared to normal walking, when using the exoskeleton with a spring of stiffness 5 N/mm. The decrease of average metabolic cost using springs of other stiffness can also be observed, but is not as apparent. The range of appropriate spring stiffness can thus be narrowed to an area lower than 12 N/mm and near 5 N/mm. The exoskeleton without the spring produced some load or constriction on the user, since the metabolic cost has risen significantly (+8%), compared to normal walking. The main cause could lie in the added weight (0.8 kg) or in the introduced kinematic constraints.

For participant 3 and 4, as seen in Fig. 4, the mechanism shows less promising results. With the usage of the exoskeleton, the average metabolic cost has risen in every series. The main reason for a change in results between the first pair and the second pair could be the wrong timing setup. Between the second and third participant the rope on one of the exoskeletons snapped because of wear and was then exchanged. The clutch timing was then probably compromised. This points at the importance of the timing of the engaging and disengaging of the mechanical clutch.

The average metabolic cost for all participants is shown in the last graph in Fig. 4. A decrease of metabolic cost can be seen when walking with the exoskeleton with the 5 N/mm spring. However, the reduction is not great compared to normal walking.

The exoskeleton effect on the Soleus muscle was approximated by measuring its EMG activity. The results are shown in Fig. 5. Unexpectedly, the results do not directly coincide with the average oxygen consumption. For participants 1 and 2, as shown in the first graph of Fig. 5, the use of the exoskeleton increases the metabolic cost, which is expected. The Soleus EMG activity is smallest when the stiffness is 20 N/mm, and not 5 N/mm, as in the measurement of the average oxygen consumption. This indicates that some other muscles are the source for the metabolic reduction, and not the measured Soleus muscle. Results for participants 3 and 4, shown in the second graph in Fig. 5 are different. This supports the assertion that the exoskeleton effect did change because of compromised clutch timing. Some decrease of EMG activity can be seen, however, that effect is postponed to springs of greater stiffness. The third graph, depicting the average for all the participants, again shows the decrease of Soleus muscle activity.

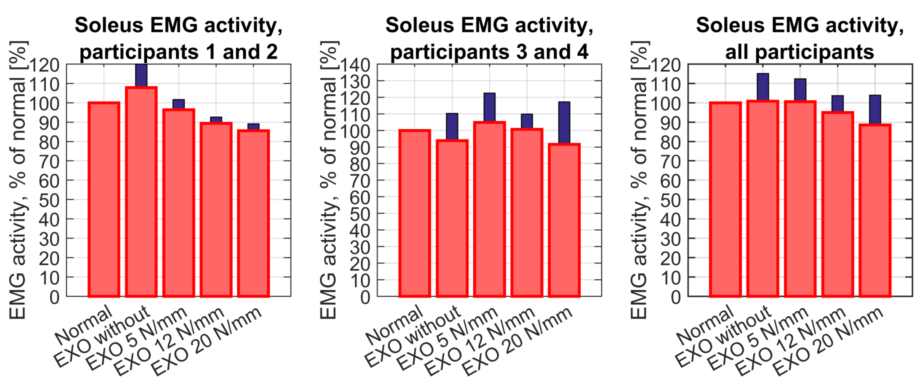

Fig. 5: Average Soleus muscle EMG activity and sample standard deviation for first and second pair and all of participants.

5 Conclusion

The collected data shows that the exoskeleton can reduce the metabolic cost of walking, if the clutch timing is correct and if an appropriate spring stiffness is chosen.

The exoskeleton ergonomic design seems to be important, since it can influence the metabolic cost reduction greatly. An important aspect are also the introduced kinematic constraints, that can increase the metabolic load on the user. Therefore, special attention is needed in the light-weight exoskeleton design and weight/strength optimization.

The clutch mechanism has some weak points that reduce the overall robustness of the clutch operation. When designing the clutch, focus at these areas is recommended.

Based on the results, the appropriate spring stiffness seems to be smaller than 12 N/mm and near 5 N/mm. The pilot study confirmed the potential of the unpowered ankle exoskeleton for reducing the energy cost of walking.

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