Controlling Pneumatic Artificial Muscles in Exoskeletons with Surface Electromyography

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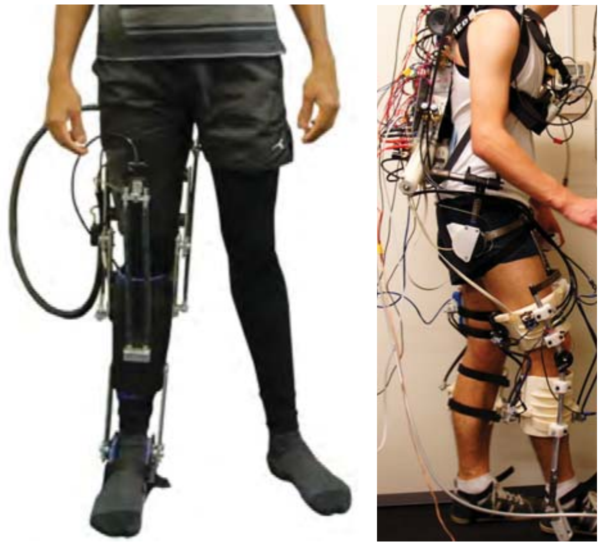
***Abstract*—Powered exoskeletons are gaining more interest in the last few years, as useful devices to provide assistance to elderly and disabled people. Many different types of powered exoskeletons have been studied in the past.**

In this research paper, a soft lower limb exoskeleton driven by pneumatic artificial muscles (PAMs) is presented. It features a lightweight, flexible solution to attach actuators to human limbs without imposing restrictions on limb movements. Ankle plantarflexion and knee extension by both legs are assisted.

Surface electromyography (sEMG) sensors are used to measure muscle activity, which are amplified and used as input to control the pressure in the corresponding PAMs proportionally to sEMG activity level. No force or torque measurements/estimations are performed, this is not strictly necessary because of the similarity between human muscles and PAMs. Employing the principle of assistance-as-needed, allows for a range of activities to be supported, such as walking, standing up and climbing stairs.

Metabolic measurements in different scenarios did not register a significant change in energy expenditure when the (passive) exoskeleton is worn, even not when it is activated although the assistance can be experienced physically. More research is needed to optimize the whole system in different aspects, in order to obtain a significant reduction in energy expenditure and make the system practical.



(a) First prototype (b) Second prototype

Fig. 1: Earlier exoskeleton prototypes at University of Canterbury.

I. Introduction

Powered exoskeletons are useful in giving assistance to those who need extra support in the limbs. Example users are eldery people and patients recovering from a stroke or other injuries.

Many different kinds of powered exoskeletons have been proposed, designed and built. At the University of Canterbury, research focuses on lower limb assistance at (combinations of) the knee, hip and ankle joints, where pneumatic artificial muscles (PAMs) are being used as actuators.

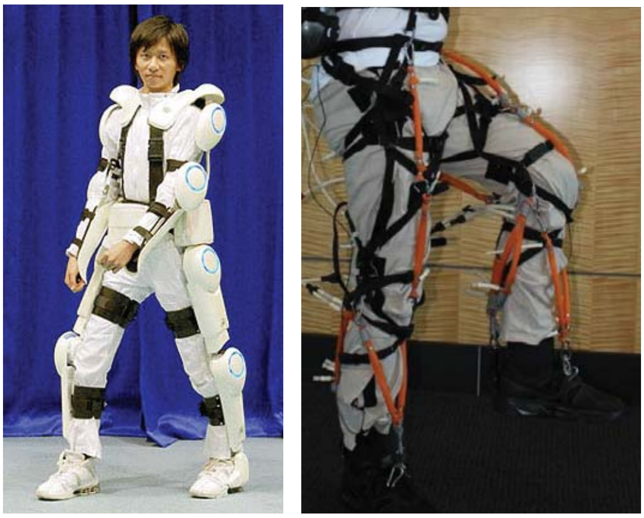
One key problem in designing a powered exoskeleton is the transfer of forces from actuators to the human body. Most often, rigid exoskeletons with mechanical joints are used, but these tend to be very heavy. More recently, soft exoskeletons with no rigid structure have been investigated with some positive results.

Useful assistance can only be given when the actuators operate in accordance with human movements and intentions. One way to register these intentions, is to measure muscle activity with surface electromyography (sEMG) sensors.

A. Earlier research at University of Canterbury

Research at University of Canterbury about sEMG-controlled exoskeletons started in 2011. The first published paper [1] focused mainly on the task of estimating knee joint torque from sEMG measurements. A first prototype of a powered exoskeleton (Figure 1a) was demonstrated by end-2012. It involved assisting the knee joint both in extension and flexion direction, but it was limited to one leg only [2].

In 2013, a second prototype (Figure 1b) was designed and constructed [3]. This one features a rigid structure with mechanical joints, and involved assistance for the knee and hip joints for both legs using PAMs attached to the structure. Rotary encoders measured the angles and angular velocity of the hip and knee joints, which was used to drive the PAMs in



(a) HAL-5, Japan (b) Harvard's exosuit

Fig. 2: Exoskeletons from Japan and Harvard.

Fig. 3: Exoskeleton developed at University of Twente.

the gait cycle.

B. Research elsewhere

In Japan, the Hybrid Assistive Limb (HAL) exoskeleton has been developed by Tsukuba University and Cyberdyne [4]. The last prototype (HAL-5, Figure 2a) consists of a full-body rigid structure in which motors drive the joints. sEMG sensors are used to register muscle activity signals; these and other sensors control the motors such that human motions are amplified. Assistance level has proved to be very effective; when wearing the powered exoskeleton, one can carry much heavier weights than without exoskeleton. The product is mature and already being used in some hospitals.

At Harvard University's Wyss Institute, a soft exosuit has been designed (Figure 2b). It does not use a rigid structure, but constructs anchor points on the body using a web of straps over the body. The exosuit is only intended to support the human gait cycle; sensors register the current phase in the gait and assistance is given at certain moments for fixed durations. Currently, the active exosuit does not increase or reduce energy expenditure compared to not wearing the suit [5].

At University of Twente, the Mindwalker has been developed (see Figure 3), a lower limb exoskeleton providing active gait assistance. Steps are initiated by the user by shifting body weight to the left or right, after which the exoskeleton performs a single step in the gait [6].

Other examples of exoskeleton studies are the quasi-passive knee exoskeleton [7], and Berkeley's lower extremity exoskeleton [8].

II. Exoskeleton design

At start of the project at University of Canterbury, the second prototype [3] was about to be finished. The original plan was to upgrade this one by equipping sEMG sensors and redesigning the software.

A. Redesign

The second prototype gave many useful insights, but it did not provide enough assistance. The excessive weight and certain design issues severely limited the effectiveness of the PAMs, which could not be solved. So a new exoskeleton had to be designed. Part of the pneumatics and sEMG sensors of the first and second prototypes were re-used, the rest was largely redesigned from scratch.

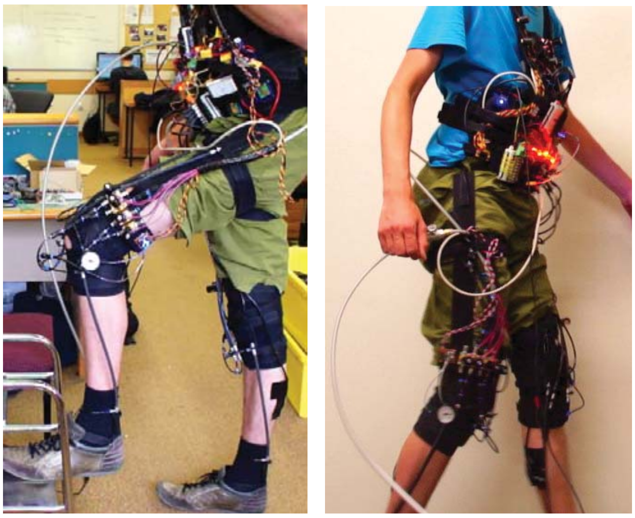
B. Supported joints

Because the first author was the only developer and tester of the system, it would be most practical if the exoskeleton could be put on without assistance. The easily accessible joints (ankle, knee) were thus favoured over the difficult ones (hip, upper limbs). Ankle plantarflexion and knee extension were seen as the most useful joint motions to provide assistive support for, because these require relatively large forces in certain motions involving lifting the body's centre of mass, such as the sit-to-stand motion. The antagonist motions were thus excluded from support.

C. Choice of PAMs

For each of the four actuated joints, a pair of either 285mm or 330mm long, and 5mm (inside) diameter PAMs was chosen. When fully contracted, its unloaded length is about 50mm shorter and the theoretical maximum force (per pair) is 270N. More powerful PAMs were only available in lengths over 500mm, so these were not an option.

If a PAM shortens by 50mm while the insertion point is mounted at 50mm from the joint axle (a typical knee joint radius), then the joint is turned by approximately 60 degrees



(a) Stepping on a stairs (b) Walking on treadmill

Fig. 4: Photos of new exoskeleton.

(depending on the actual geometrical layout). A maximum theoretical torque of 270N · 0.05m = 14Nm can be delivered if the PAM is at full length, the torque decreases approximately lineary with the contraction percentage.

D. Exoskeleton structure

Good attachment points (origin and insertion) for the PAMs are essential. For the ankle joint, the origin of the PAMs are placed at either side of the knee, and the insertions at the heel; this can be observed in Figure 4a. For the knee joint, origin of the PAMs are placed at the side of the hip and the insertion just below the patella.

To reduce weight, no rigid structure was used. Instead, anchor points for the PAMs were constructed around the feet, knees and hips using double-sided velcro tape of different lengths and widths, with loops consistently facing inwards and hooks outwards. Velcro tape was also attached to the threaded end of all PAMs. Strong connections were established by adding multiple layers, if needed.

Knee pads were used to distribute the pulling force of the PAMs over a large area, and to protect the knee skin from discomfort. The velcro tape network around the upper legs transferred part of the downwards pulling forces to the waist, more specifically the ilium parts of the pelvis. Figure 4a and 4b show two images of the assembled exoskeleton.

Comfort level was found to be fine everywhere, except around the ilium. It is possible to relieve this discomfort by distributing PAM forces to other parts of the upper body as well, as has been done in [5]. This was not implemented, because the discomfort was not a problem in the tests and because the upper body was already carrying the electronics.

E. sEMG sensors

To measure muscle activity, active sEMG sensors were placed on certain muscles. Every sensor has two nickel-plated contacts, each 9mm in diameter and its centers spaced 20mm apart. The sensor houses a two-stage 4000x differential amplifier, of which the output signals are anti-aliased, acquired by A/D convertors at a rate of 2000Hz and then recorded by a laptop. The amplifier circuit is protected by a layer of epoxy, and the whole sensor is attached to the prepared skin with adhesive tape.

One sensor was placed at the rectus femoris muscle, to record knee extension effort. Another was placed at the gastrocnemius, which records ankle plantarflexion muscle activity. A ground reference electrode was placed at the skin over the knee bone. In [1] the reference electrode was set to the average sEMG electrode potential, but in this study no advantage was found of this method over simply connecting to ground potential.

The disadvantage of using only one sEMG sensor per muscle is that the error level is relatively high. sEMG data is stochastic in nature, and smoothing operations by means of a low-pass filter introduces a phase shift and thus a delay in the activation signal. The error level can be reduced by combining information from multiple, independent sEMG sensors. But when the exoskeleton is worn, available skin surface over relevant muscles is restricted, making it difficult to position multiple sEMG sensors.

When two sensors were placed adjacent longitudinally, then the sEMG signals turned out to be highly correlated. This may be caused by propagation of motor unit action potentials (MUAPs) through muscle fibers, resulting in similar signals to be measured at different locations along the muscle surface. Because of these complications, only one sEMG sensor was used per supported joint.

F. Driving the PAMs

The system is powered by an external air supply of 6 bar, which is commonly available in hospital settings and considering that the exoskeleton is primarily used for rehabilitation training purposes. A more mobile system would be possible by using a portable pressurized air tank, although limited to a few minutes of walking [5], unless some form of pressure generation or conservation technique(s) are employed.

For each pair of PAMs, inlet and outlet valves (type PVQ33-6G-16-01, max. flow rate 100ℓ/min) were used to adjust the pressure inside the PAMs. The valves are electrically operated and airflow varies approximately lineary with both the pressure drop over the valve, and the electrical current.

The pulling force in a PAM depends on both the length of the PAM, and the pressure of the air inside it. To control the force (and respond quickly to changing demands), some form of feedback is necessary. There are two main possibilities:

- **Force feedback** A load cell in series with the PAM can measure the force, and airflow can be applied to increase or decrease this force, without having to know the pressure itself. This approach is useful when the forces applied by the human in the joint are being estimated, and assistive force derived from it.

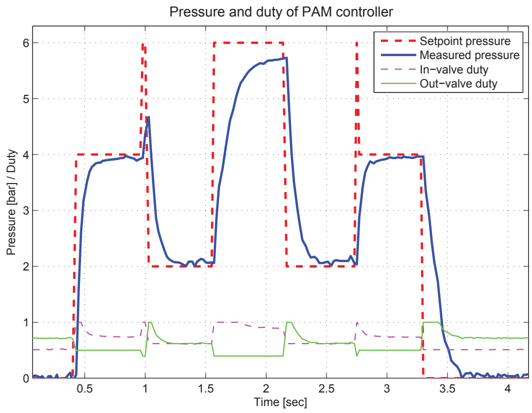


Fig. 5: Pressure controller performance test.

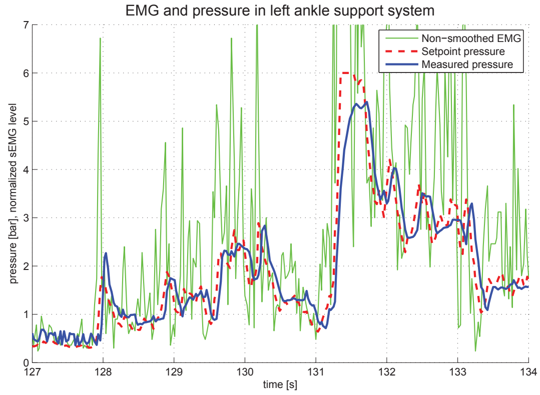


Fig. 6: Example sEMG and pressure graphs.

- **Pressure feedback** A sensor measuring air pressure inside the PAMs makes it possible to bring the PAM at the desired pressure quickly. If the PAM length would be known (e.g. from the joint angle), then the pulling force could be derived as well.

In this study, pressure feedback is used, because the goal is to let the PAMs behave as close to the human muscles as possible. The sEMG signal is a measure of the muscle activity which translates to PAM pressure, independent of the joint angle or forces involved.

G. Connecting sEMG to PAMs

PAM pressure was set proportional to sEMG activity: $P = \alpha V_{emg}$, where P is the PAM pressure in bar, V_{emg} is the (filtered, not amplified) sEMG voltage level in volt (or mV), and α the gain factor in bar/mV. The set of gain values for each joint was experimentally derived, by adjusting these parameters with turn knobs while performing various motions with the exoskeleton powered on, aiming for maximal effective assistance.

Joint angles were not measured and no torque estimation was performed. By placing each PAM approximately parallel to the muscle it supports, and controlling pressure proportional to muscle activation level, the natural behaviour of the muscle is mimicked as close as possible. The idea is that the human body adapts to the presence of the additional (artificial) muscles, and adjust EMG signals accordingly.

A Windows 7 laptop is used to process raw sensor input data and calculate valve control signals. sEMG signal acquisition was performed with a NI USB-6009 device, other sensors were read with A/D converters connected to a NI USB-8451 device, to which the valve driver circuit was connected as well.

H. Timing

To provide effective assistance, each PAM must contract and extend simultaneously with the muscle it supports. If the exoskeleton lags behind, then the assistance is less effective or may even impede human movements. So a correct timing is crucial.

There is a delay between firing of the EMG signal and the corresponding (start of) muscle contraction. Earlier research found this delay to be about 20ms to 80ms [2]. Ideally, the assistance system should have about the same delay to keep on par with the human muscles.

EMG smoothing involves a delay which depends on the cutoff frequency of the final low-pass filter. Values of 1Hz to 5Hz have been suggested [2]; a frequency of 2.5 Hz corresponds with a time constant of 64ms, which is the time to rise to 63% of the final value after a step input.

The periodic task of the control software operated at 30Hz; higher rates were found to cause irregularities in the task execution time, likely caused by USB and/or OS real-time characteristics. At 30Hz, software latency can be as high as 33ms, an average latency of 20ms is assumed.

The pneumatic valves use a linear solenoid actuator. Performance tests showed that when setpoint pressure steps from 2.0 to 4.0 bar, the measured pressure starts rising within 10ms and reaches 3.8 bar in approximately 80ms, for one of the pair of PAMs used in the exoskeleton (see Figure 5). The limiting factors here are the electromechanical valves, for which the maximum airflow is limited.

A measure of the latency from sEMG signal activity to the corresponding rise of PAM pressure can now be estimated to be 64ms+20ms+10ms = 94ms. It is higher than the delay in muscle contraction, so the user may wish to use a higher sEMG smoothing cut-off frequency to reduce the latency, at the cost of a less smooth assistance behaviour.

III. Test Method

To determine the effectiveness of the exoskeleton, oxygen consumption measurements were performed while the test subject worked through certain scenarios under different situations. A COSMED K4b^2 breath analyzer system was used for this.

When wearing the powered exoskeleton, the test subject was allowed to adjust parameters for assistance level, left/right assistance balance and sEMG signal smoothing cutoff frequency while exercising before the test.

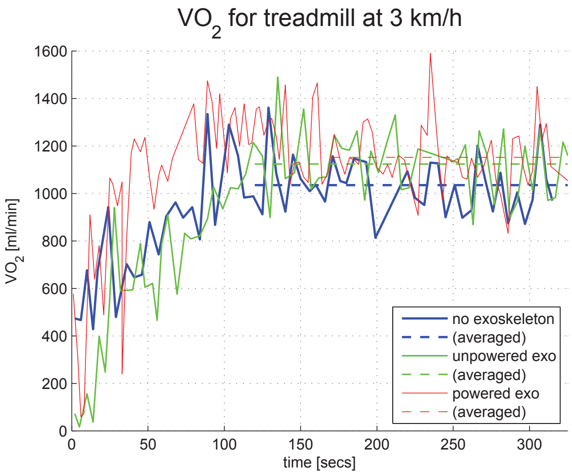


Fig. 7: VO_2 consumption at 3 km/h.

scenario	no exoskeleton	unpowered exo	powered exo
walk 3 km/h	1035 (100%)	1124 (108.6%)	1152 (111.3%)
walk 5 km/h	1466 (100%)	1534 (104.7%)	(skipped)
stairs (1st run)	2316 (100%)	2222 (95.9%)	in 2nd run only
stairs (2nd run)	in 1st run only	1917 (100%)	1999 (104.3%)

TABLE I: Average VO_2 consumption for different scenarios and setups, in ml/min and indexed.

Two scenarios were set up for quantitative testing: walking on a treadmill at a constant velocity (3 km/h for five minutes and 5 km/h for three minutes), and ascending/descending an improvised flight of stairs consisting of three steps (plus ground level) for three minutes, going up and down once every eight seconds.

In each of the two scenarios, three situations were tested: without exoskeleton, with passive (unpowered) exoskeleton and with powered exoskeleton.

Only one full day was available for testing with the breath analyzer system. This was unfortunate, because the six tests could not all be taken with the same initial fitness level, and repeating tests was not feasible.

Because of the time involved to put the exoskeleton on and take it off, first the tests without exoskeleton were performed in all scenarios. After this, the exoskeleton was put on and testing continued with unpowered and powered exoskeleton in all scenarios.

IV. Results and Discussion

Typical values for the gain factor α are 3.0 bar/mV and 4.6 bar/mV for the left and right ankle joints, and 9.6 bar/mV and 9.2 bar/mV for the left and right knee joints, respectively. A gain factor of 3.0 bar/mV means that PAM pressure is set to 3.0 bar when the average (unamplified) sEMG level is 1.0mV. The gain is higher for knee joints, because sEMG signal levels were found to be lower for the rectus femoris muscle (acting on knee joint) than for the gastrocnemius (acting on ankle joint).

Figure 6 shows the rectified and scaled sEMG level (green), setpoint pressure (red) and measured pressure (blue) of the left ankle (plantarflexion) during 7 seconds, while the test subject was performing unspecified motions. It can be seen that the non-smoothed sEMG signal is stochastic; the setpoint pressure is much smoother with some delay (50-100ms typically), and the measured pressure in turn follows the setpoint with minimal delay (50ms typically), except in the lowest and highest pressure ranges.

A. Treadmill scenario

Oxygen consumption measurements for the 3 km/h treadmill scenario are graphed in Figure 7. In the first minute of the test, the oxygen consumption rises slowly to some value from where it stabilizes. Comparing the stable levels of the three situations (averaged over the last three minutes), we can observe that oxygen consumption level differences are minimal.

Walking at 5 km/h turned out to be too difficult to assist effectively. With typical gain settings, the exoskeleton was impeding certain body movements, rather than supporting them. Lowering the gain factors or changing the time constant of the sEMG low-pass filter did not help; optimal gain values were found to be just zero, so this scenario was skipped in the powered exoskeleton tests.

B. Stairs scenario

The stairs scenario was the one where the assistive force of the exoskeleton could be experienced best. When ascending a stairs, the knee joint slowly goes from 90° flexed to fully extended position and the ankle joint performs about 30° plantarflexion, while the human body is lifting itself. During this motion, it was experienced that the PAMs gave significant support. As no load cells were installed, the actual assistive force is unknown, but the maximum force is known to be 270N. So at a pressure of 6.0 bar and 50% of maximum contraction, the assistive force would be approximately 135N.

The test with the unpowered exoskeleton in the stairs scenario was repeated before performing the final test with powered exoskeleton, because the fitness levels were no longer comparable which influenced the energy expenditure. So the scenario is split in two runs.

In the first run, wearing the (unpowered) exoskeleton dropped the energy expenditure by 4%. There is no good explaination for this, other than statistical variation. The second run shows that powering the exoskeleton increases the energy expenditure by 4%. Again, this difference is too small to be statistically significant.

Figure 8 shows the pressure graphs of all four PAMs. Climb is started with the right leg; we can observe that at t=1s, right knee support is maximally activated, helping to extend the knee. Left ankle, and then right ankle supports with stepping up. One second later, the pattern is repeated with the left leg, and then with right again after which the top of the stairs has been reached.

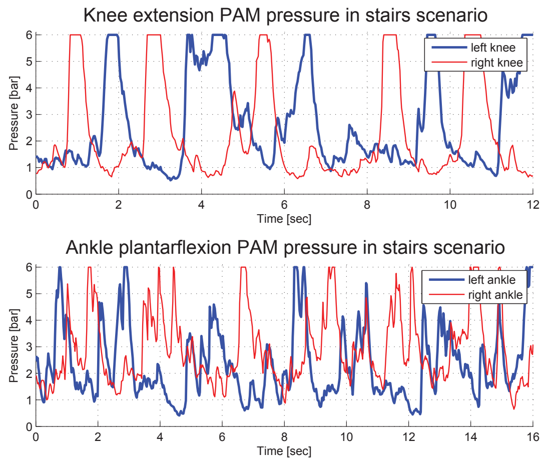


Fig. 8: Knee and ankle (setpoint) pressure graph for stairs scenario.

Around t=5s, descending starts; while right leg goes down, left knee extension support is maximally activated although the knee is being flexed, this is to oppose the gravity forces. At t=8s, ground level is reached after which the entire process is repeated. We can also observe that the ankle joint support is more random than knee joint support, this might be because ankle joint movement is used for keeping balance.

C. Possible reasons for ineffectiveness

An overview of measurement results is shown in Table I. The energy expenditure did not rise or lower significantly when the exoskeleton was worn, regardless whether it was powered on or off. For comparison, the second prototype (discussed in [3]) observed a 20% increase in energy expenditure when the exoskeleton was worn, and no significant additional change when it was powered on.

So, it can be said that the new powered exoskeleton does not add significant load to the person and it does not impede lower limb movements, but it is also unable to perform a positive, measurable net amount of work. There are various possible reasons for this:

- **Multisite sEMG** Only one sEMG sensor was used per joint. Combining information from multiple, independent sEMG sensors would give more accurate muscle activation information, because the signals are random in nature and need to be smoothed.
- **Timing** It is crucial that assistance is given at the same instant when the corresponding muscle contracts, but it turned out that it is difficult to optimize the time constant in the sEMG filter experimentally. If assistance leads or lags muscle activity, then the human body probably has to do extra effort to compensate for it.
- **Training** The human body has to adapt to the presence of artificial muscles. It may take a long time to get used to it and change the gait pattern accordingly to take optimal advantage of assistance.
- **PAM selection** Only the smallest PAMs (diameter 5mm) were used, because the available 10mm and 20mm PAMs were too long to install on the exoskeleton. Using bigger PAMs would result in more powerful assistance; the exoskeleton structure also needs to be reinforced in this case and valves with higher maximum airflow are needed.
- **PAM control strategy** In this research, PAM pressure was set proportional to muscle activity. This approach might be too simplistic. By measuring joint angles and estimating forces/torques, extra information would be available which might make a more effective support system possible.

V. Conclusion

When wearing the powered exoskeleton, the test subject experienced significant assistive forces in lower limb movements. The transfer of forces from PAMs to human body was better than in the second prototype discussed in [3].

Energy expenditure measurements did not show any significant change in energy consumption when the exoskeleton was worn, whether it was powered or unpowered. At least, it confirms that the exoskeleton is lightweight and easy to wear, which is an improvement over the second prototype which recorded a 20% increase in energy expenditure [3] when the exoskeleton was worn.

Some exoskeletons, such as HAL-5, are powerful enough to carry the extra weight and still deliver positive assistance. But exoskeletons involving PAMs might be better suited to soft exoskeletons.

The use of sEMG signals as input to the assistance control system gives the user full freedom in the motions. Unlike in other studies, it is not necessary to stick to a predefined gait cycle to receive assistance. Significant support was experienced when standing up and climbing a stairs, actions that involve lifting of the person's own body and thus require considerable effort.

The system can be improved in many ways. The choice and attachment of the PAMs could be improved; joint angles and force measurements could provide more insight in joint torques, power transfer and work; combining multiple (and smaller) sEMG sensors could should give smoother signals; a more detailed training system could result in a better optimized set of parameters for assistance control.

No earlier research involving a soft exoskeleton driven by sEMG signals could be found, so this research might be the first of this kind. One can expect that further research will result in better designs, getting closer to the goals of delivering significant, measurable assistance.

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