Design and preliminary testing of a pneumatic exoskeleton for walking assistance

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Abstract

Tripping is one of the major causes of falls in elderly people. Injuries that originate from falls usually have severe consequences on their quality of life. The best solution for this problem is prevention of such traumatic events which proves to be a challenge in the current state of elderly care. This is why we are in an urgent need of developing new solutions that could help in the prevention of falls and fall related injuries in this ever increasing part of our entire population. One possible preventive solution could be the use of assistive devices such as exoskeletons. In this work we present a design of a pneumatic exoskeleton for walking assistance primarily designed as a platform to study active control solutions for preventing falls caused by tripping. We also present a preliminary evaluation of the exoskeleton capabilities on affecting the gait characteristics of a subject during walking.

1 Introduction

Population ageing is one of the many global issues we are facing. In 2012, 11.5 % of the global population was older than 60 years and by 2050 this percentage is projected to increase to 21.8 % [1]. This is a problem because elderly people usually require more assistance for even simple everyday tasks and they are more prone to injuries. One of the big causes of injuries in elderly people are falls which usually have severe consequences on their quality of life [2]. And one of the main causes of falls is tripping [3, 4, 5]. The mechanisms of tripping have been widely explored in the past [6, 7]. In order to prevent a fall caused by a trip the support limbs have to counteract the angular momentum gained by the body during impact with the obstacle. This can be achieved with the adaptation of different recovery strategies. In the work of Eng et al. [8] two such strategies were described which depended on the part of the swing cycle the perturbation occurred. If the perturbation occurred in late swing, a lowering strategy of the swing leg was adopted followed by a step of the contralateral limb to overtake the obstacle. When perturbations occurred in early swing phase (at approximately 20 % of the swing phase) an elevating strategy was adopted by the subjects. This comprised of a flexion of the swing limb and extension of the stance limb. This strategy was further investigated in the works of Pijnappels et. al. [9, 10]. Here more focus was put towards the support limb and its contribution in the control of angular momentum after tripping. It was shown that the support limb provided subjects with more time and clearance for proper positioning of the recovery limb. Additionally the support limb also restrained or reduced the angular momentum ob the body acquired when tripping. Further work with elderly people showed that some individuals cannot properly reduce the angular momentum after a trip [11, 12]. This was due to a lower rate of change of moment generation in all support limb joints. One possible solution for this could be the use of an assistive exoskeleton to provide additional torque to counteract the uncompensated forward momentum. Only one similar device was successfully presented in the work of Monaco et. al. [13] where an assistive exoskeleton was used to help regain stability of a human after a slip. In this work we present the design and preliminary testing of an exoskeleton that will provide a platform to explore possibilities of assistance in tripping scenarios. Additionally we present an early investigation in the effects of the exoskeleton on the kinematics of human gait.

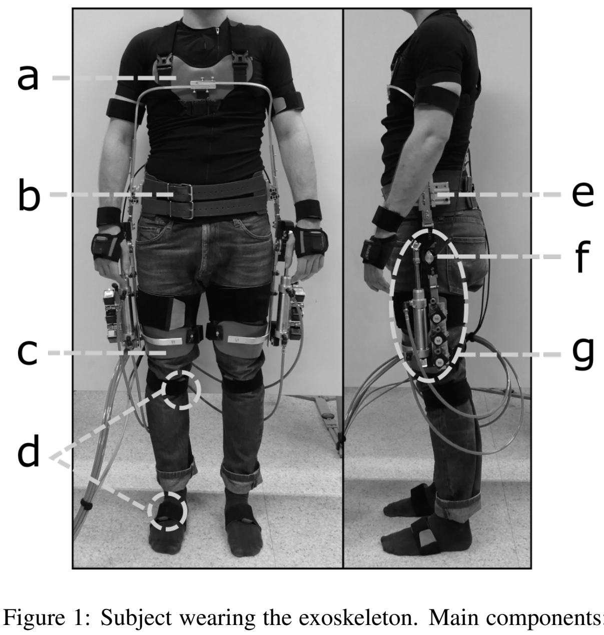

Figure 1: Subject wearing the exoskeleton. Main components: a) chest strap, b) hip strap, c) leg strap, d) Xsens sensors, e) adjustable hip mount, f) encoder, g) pneumatic actuator.

2 Design and experiment

We developed a bilateral powered pneumatic exoskeleton. The main mechanical components are presented in Figure 1. They include a belt, chest strap, leg strap, adjustable hip mount and an actuation mechanism on each side. The actuation mechanism consists of a bidirectional pneumatic cylinder and a rotational joint (This is more clearly visible in the schematic diagram in Figure 3). The Xsens sensors (Xsens, Enschede, Netherlands) are not an integral part of the exoskeleton but were used in this work to evaluate the effects of the exoskeleton on the human kinematics.

2.1 Actuation mechanism

The applied pressure in the cylinder chambers produces torque at the hip to aid in the flexion or extension of the legs. Due to the constraints of the mechanism's kinematic chain the torque in the hip joints can be calculated based on the angle of the rotational joint and the applied pressure. The angle is measured by rotational encoders mounted in these joints. Whereas the pressure in the cylinders is measured with the feedback loop provided by the valves used for controlling the cylinders.

2.1.1 Calculations of insertion points

One of the most important design aspects of such an actuation mechanism are the insertion points of the cylinder. These define the maximum torque produced at the rotational joint and the range of motion of the mechanism. An analytical formulation difficult to achieve for such a problem. This is why a numerical approach was used to calculate the optimal solution. At first, the requirements for the range of motion were set to $-20°$ to $110°$. Then different cylinders were taken into account from a list of possible configurations of length and cylinder width. Cylinders of same lengths with different diameters provide more force, but are also heavier. Our optimization procedure included weight as a parameter for finding the best possible solution. After the mathematical calculations we designed a 3D model of the actuation mechanism to check for mechanical constraints that would limit the construction of the mechanism (thickness of the rotational joint, encoder mount, etc.). The model is presented in Figure 2. After minor modifications of the insertion points placement in the modelling environment, the parts for the fixation of the cylinders were 3d printed. This enabled us to achieve the desired mechanism characteristics.

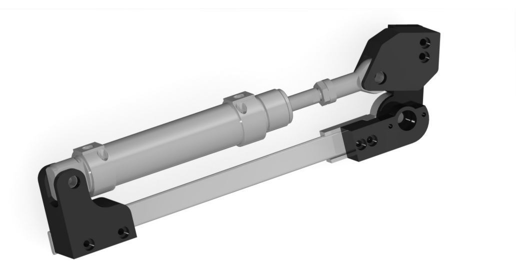

Figure 2: 3D model of the actuation mechanism.

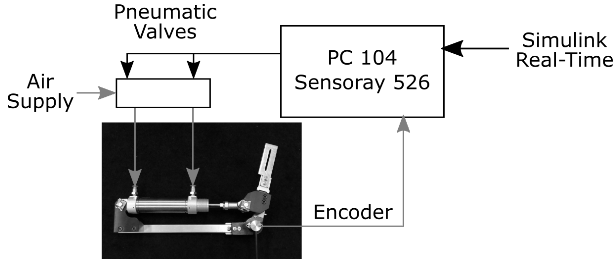

Figure 3: Schematic diagram for the exoskeleton control.

2.2 Control

The schematic diagram for the exoskeleton control is presented in Figure 3. Each actuation mechanism is controlled with two pneumatic valves (Norgren VP50-10SBJ1-11H00, Norgren Ltd, Blenheim Way, Fradley Park, Lichfield) which enable precise control of the pressure in both cylinder chambers. These valves are controlled via analog signals (0-10 V) and provide an analog output (0-10 V) for feedback of the current pressure. The controller was developed in Simulink Real-Time (Mathworks, Natick, MA, United States) and runs on a PC 104 with added connectivity of a Sensoray 526 card (SENSORAY, 7313 SW Tech Center Dr., Tigard, OR 97223).

2.2.1 Modes of operation

Our exoskeleton actuation mechanism and controller enables us to use different modes of operation. The exoskeleton can be operated in a torque control mode or position control mode. In torque mode the pressure in the cylinders is controlled so that for the given position of the mechanism the torque produced by the mechanism remains constant. A position control scheme was also tested where the pressure in the cylinders was controlled to position the mechanism at a given angle. However, this mode of operation was not used in this work.

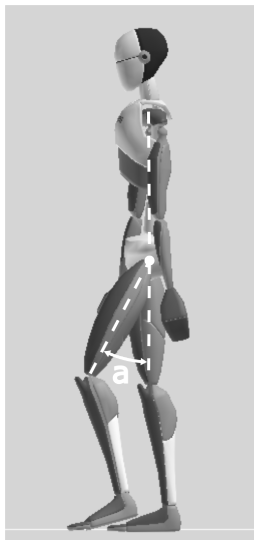

Figure 4: Human model in MVN Analyze software. a) hip angle of left leg

2.3 Experiment

For preliminary testing of the exoskeleton capabilities we tested the system on one subject. During normal walking with a constant speed on a level floor the left actuator was activated to aid in the flexion of the left leg (4 bar input pressure in the bottom chamber). The activation occurred in mid swing phase at a 10 degree flexion of the left leg (measured with the encoder) and was randomly enabled or disabled. The duration of the activation was set to 200 ms. During the experiment, kinematic data of the subject were recorded using the Xsens Awinda motion capture system (Xsens, Enschede, Netherlands). Figure 4 shows the human model during the experiment.

Ad-hoc post-processing of the acquired data was performed in Matlab (Mathworks, Natick, MA, United States). Data from the controller (encoder angles, pressure in cylinder chambers) and data exported from MVN Analyze (human joint angles) was synchronized and combined.

3 Results

One of the most critical aspects of control is timing. In our set-up we analysed the weakest link, to asses the capabilities of the exoskeleton. One of the most important characteristics for the actuation mechanism is the delay of the pneumatic valves after an input signal. There are two key temporal characteristics that are of interest in our system. The delay time (the time it takes for the output to reach 10 % of the end value) and the response time (the time it takes for the output to reach 90 % of the end value) of the pneumatic valves. We measured this based on the mean and standard deviation of 20 responses to the 4 bar step input signal used in our experiment. For the pneumatic actuation mechanism used in this experiment we measured a delay time t_d of 35 ms and a response time t_r of 100 ms. The graph is presented in Figure 6.

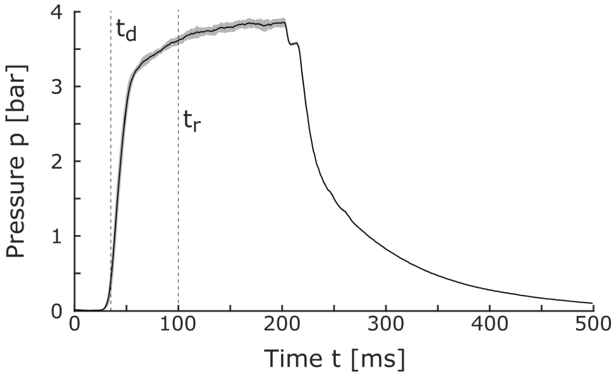

Figure 6: Mean and standard deviation of 20 responses to the 4 bar step signal. t_d marks the delay and t_r marks the response time.

Data from the Xsens MVN Analyze software was used to evaluate the effect of the exoskeleton on gait kinemat-

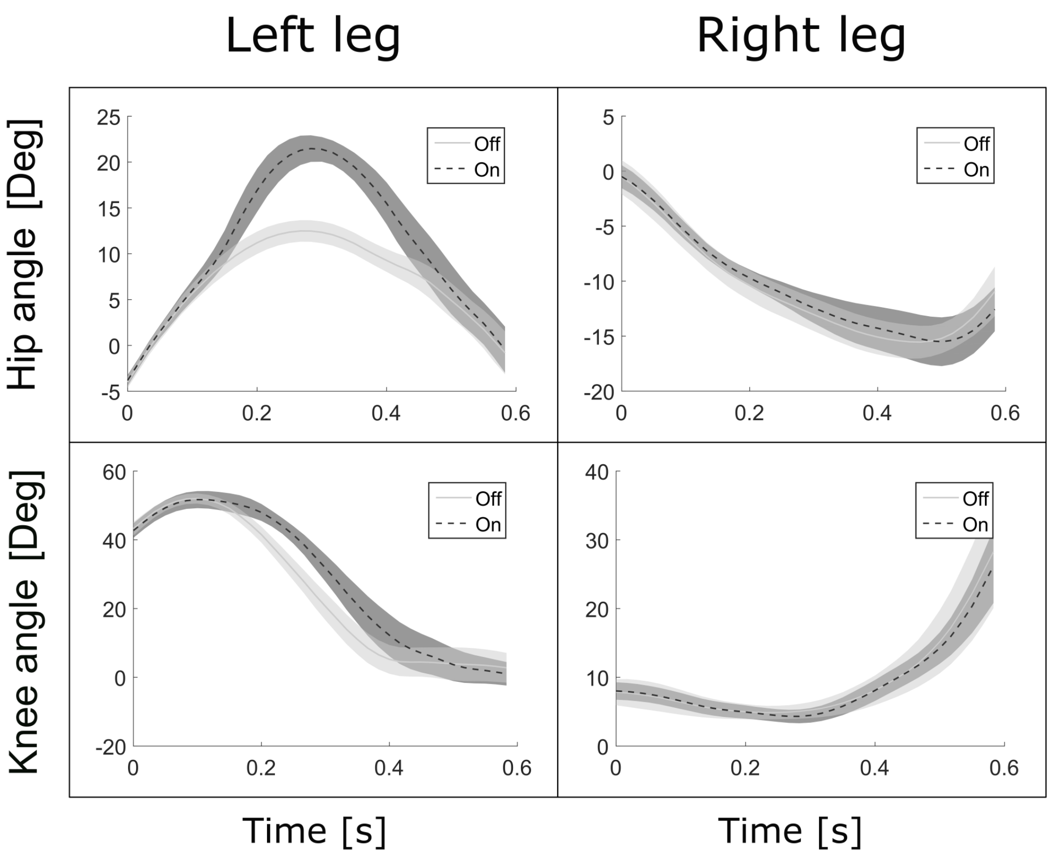

Figure 5: Joint angles for flexion/extension at the hip and knee joint.

ics. To achieve this we first had to synchronize various steps in time space. We did this by selecting points in the gait cycle when the value of the left leg hip angle crossed the threshold of -5 degrees with a positive gradient. This threshold corresponds to the activation of the exoskeleton but can be used to compare steps when the exoskeleton was disabled as it marks the same period of the gait cycle. This points established the start of a 600 ms time window for each step. We randomly selected 20 of this time windows for when the exoskeleton was activated or deactivated to compare the effects of the exoskeleton. For these two sets of steps we calculated the mean and standard deviation of the hip and knee angles for flexion/extension of both legs. We present this data graphically in Figure 5. Time zero represents the time when the exoskeleton was activated. As mentioned before, the activation occurred only for the left leg when it was in the swing phase. It can be clearly seen that there is a significant change in the hip joint angle for the left leg. A smaller change is also noticeable for the left knee angle. If we compare the delay from Figure 6 to the delay in Figure 5 the latter appears longer. This is due to the mass inertia of the legs as the applied torque in the joint needs to accelerate the leg in order for a change in the hip angle to be visible. On the right hand side graphs we observe that there was no significant effect on the hip and knee angles of the right leg. This was expected as the right exoskeleton actuator remained passive throughout the experiment. However this information is useful as it shows, that we can compare two separate sets of steps in the experiment.

4 Conclusion

The presented exoskeleton prototype is capable of inducing significant changes in the gait cycle in a controlled manner. This will enable further experiments and validation of various control algorithms for walking assistance or other tasks such as forward bending, squatting etc. A more precise measuring of the system parameters would be needed (delay and response times at different pressures and for different volumes) in order to conclude if the exoskeleton is viable for tripping prevention i.e. if it could react fast enough. This will be addressed in future work, where the exoskeleton design and experimental setup will be further optimized.

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