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Mechatronic Design of a Sit-to-Stance Exoskeleton.

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Abstract— **This paper describes the design and development of an exoskeleton that can deliver assistance-as-needed to patients or elderly with muscle weakness. Since the proof-of-concept is a first step towards the development of a final commercial prototype, the design had to be adaptable for patients with different heights, be comfortable for the patients, safe in use, energy-efficient and affordable in production. For this reason a modular system was built, using the same compliant actuator system in all joints. This paper describes the global design decisions made and the construction of the actual prototype.**

I. INTRODUCTION

Mobility is one of the most important things in life, but in order to be mobile a person needs to be capable of standing up first. Several devices have been designed to aid a person with a sit to stand motion. Hirata et al. [1] suggest an electric bed that moves up and down to help people get up. Chugo et al. [2] added an additional feature in the form of an actuated support bar that emulates the assistance of a nursing specialist. While these devices require an actuated sitting surface and are thus not applicable in every day life, [3] describes an active mobile walker with two handles that pulls the patient up.

When the mobility issues extend further than difficulties with standing up, a more versatile aid is required. Wearable devices such as exoskeletons provide a solution to this as they can assist the wearer during a range of motions such as standing up, walking, climbings stairs etc. The RoboKnee by Yobotics, Inc [4] and the exoskeleton by Karavas et al. [5] are examples of single joint exoskeletons. Both consist of a knee brace actuated by a compliant actuator, the RoboKnee by a series elastic actuator [6] and the Karavas exoskeleton by a CompAct-ARS actuator [7]. The knee joint is a logical choice because in standing up it is the most demanding joint [5].

The Vanderbilt exoskeleton [8], marketed as Indego, actuates the knee as well as the hip of the wearer during standing up and walking. It is meant to be worn with an ankle-foot orthosis. The ReWalk [9], Ekso [10] and HAL [11] exoskeletons are all full lower limb devices. The hip and knee joints of the devices are actuated, the ankle joints are passive. While the knee is the major source of power during sit to stand transition, that task is transferred to the ankle in level walking. Therefore, researchers of the University of Alabama [12] implemented an active ankle joint in their exoskeleton. The hip joint however has been excluded in the design to reduce the structural complexity of the device. The joints of the knee-ankle-foot orthosis are driven by two pneumatic cilinders.

Clearly there is a growing need for mechatronic devices that dynamically interact with humans, such as orthoses, prostheses and rehabilitation equipment. A lot of researchers recognize this need, as is shown by the multitude of devices that are being developed. For an overview of exoskeletons and active orthoses with applications ranging from military operations to rehabilitation, the authors refer to [13] and [4]. As these human-worn robotic devices interact with the patient on a high level, additional aspects become very important: safety, wearability, energy autonomy and intelligent interaction, along with psychological aspects. Researchers from different fields will attempt to address these aspects in the framework of the MIRAD project (http://www.mirad-sbo.be/). The exoskeleton discussed in this paper is the first prototype within this project.

The prototype is meant to provide assistance to its wearer during a sit to stand motion. The required assistance is determined following an assistance-as-needed strategy where active participation of the user is promoted. The use of compliant actuation in this prototype decouples the inertia between different links of the exoskeleton and connects them through the use of an energy storing element. This is one of the key factors to improve safety in this application of human-robot interaction [14] and minimize the energy consumption to enhance human performance to the required level. For an overview of different types of compliant actuators and their advantages, the authors refer to [15]. The prototype makes use of existing braces in order to obtain a wearable device and to keep the expenses low. The cost is further reduced by using identical modular actuators rather than diversifying the actuator for every joint. Progress beyond the extensive state of the art follows from the combination of implementing compliant actuators with passive compliance into a full lower body exoskeleton, an assistance-as-needed strategy and the awareness of cost-effectiveness.

Section 2 describes the process that was followed for the design of the exoskeleton. It accounts for all the decisions that were made concerning the different aspects of the design. In section 3 an overview of the global build-up of the prototype is provided. Furthermore the device's wearability has been assessed during an unpowered sit to stand transition.

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The paper ends with a short discussion and a summary of the future work.

II. DESIGN PROCES

A. Decisions made

Wearability is one of the key properties that make an exoskeleton useful or useless for support of daily life activities, which is exactly why it influences a lot of the decisions that were made regarding the design. Wearability is influenced by the weight of the device, and requires the structure, actuators and controllers to be as light as possible. Another aspect of wearability is that patients need to accept the device and be willing to wear it. To limit the weight of the structure two strategies have been followed. First of all a lightweight, commercial, passive exoskeleton serves as a base for the prototype. Note that starting from a commercial device also gives a headstart towards patient acceptance. Secondly the device contains only single degree of freedom (DOF) joints. Given that the device will only be used to test sit and stand actions and that the range of motion (ROM) of the non-sagittal DOFs is very limited in that case (measurements of lower limbs kinetics during sit to stand were performed by Afschrift et al.), it was investigated that 6 active flexion/extension joints are sufficient to execute a sit to stand activity. As an extra effort towards wearability, the adaptability of the commercial device to a patient's height will be kept intact to the extent possible.

Because of the advantages for the wearability, safety and energy consumption of exoskeletons, a compliant actuation strategy was followed. The actuators are based on the MACCEPA drive concept [16]. The advantages are that a MACCEPA actuator can be built with standard off-the-shelf components, that it has a linear angle-torque characteristic and that the control of compliance and equilibrium position is fully independent. Finally it was also decided to use identical actuators for all the joints. At this stage of the project, developing specific actuators for each joint would hardly lead to a diminishment of the weight of the final prototype, while it would considerably increase the effort of the design and construction. As the protoype will only be used to perform sit to stand tests in a lab setting, the incorporation of an insole with pressure sensors is not necessary. The ground reaction forces will be determined with external force plates. The built-in sensor set consists of two encoders per joint. Furthermore, the device is not autonomous. A communication and power link to the outside world is present. In fact the system operates according to the same principles as a servomotor does: one needs only to supply the power and a control signal for the system to operate as the internal control and sensors do the rest.

As for the amount of assistance, the authors have based their decision on the data provided by Afschrift et al. [17]. As mentioned earlier, one of the innovative aspects of this work is the use of an assistance-as-needed strategy to determine the support provided by the exoskeleton. Assistance-as-needed means that although the exoskeleton supports the sit to stand activity, it still promotes active participation of the user by only bridging the capability gap, i.e. the gap between the capabilities of the user and the task requirements. This is essential to maintain neuromotor function and prevent disuse [17]. In Fig. 1, the blue bars represent the peak torques for a constant assistance level of 30%, meaning that during the entire test the actuators have to provide 30% of the biological torque that is necessary to perform the sit to stand movement. This is plotted for the ankle, knee and hip joint. The cyan, yellow and red bars represent the peak torques that are required to bridge the capability gap of patients with a muscle weakness of resp. 70, 80 en 90%. Muscle weakness is defined as a decrease in terms of percentage of the maximal isometric force of all muscles. The muscles of a person with a weakness of 70%, can still generate 30% of the maximal isometric force of healthy muscles. It is clear

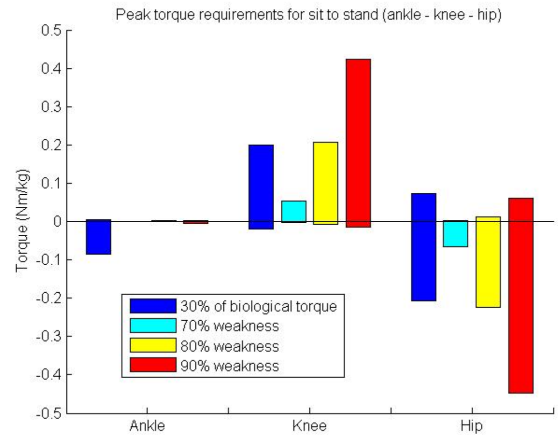


Fig. 1: Peak assistive torques required during sit to stand for a constant assistance level of 30% and assistance-as-needed for patients with a muscle weakness of resp. 70, 80 and 90% (Data from [17]).

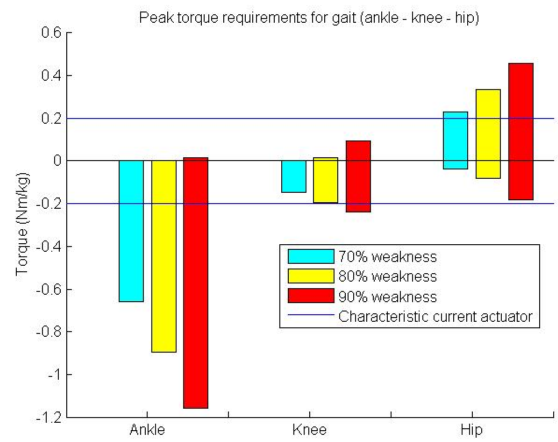


Fig. 2: Peak assistive torques required during gait for assistance-as-needed for patients with a muscle weakness of resp. 70, 80 and 90% (Data from [17]).

(a) HIPO (b) KAFO

(c) R.O.M. walker

Fig. 3: Commercial passive exoskeleton (Picture credits: http://www.orthoservice.com/download/catalogue/Mac_OSX_Start.htm).

from the bar plot that the peak torques required at a constant assistance level of 30% are similar to those required to bridge the capability gap of patients with a muscle weakness up to 80%. For this prototype the required level of assistance was thus set on a constant assistance of 30%. Given that the wearer will weigh about 80kg, this boils down to a peak torque of $15Nm$ that needs to be provided by the actuators. Note that this amount of torque only applies for the hip and knee joint. For the ankle a peak torque of $7Nm$ is required, following the constant 30% assistance strategy. Since it was decided to use identical actuators for all the joints, this means that the actuator at the ankle will be overdimensioned.

It needs to be noted that since wearability and energy consumption are such important aspects in the design of exoskeletons, a diligent reader might suggest (based on Fig. 1) to omit the ankle actuator. As the ankle of weakened persons requires little to no assistance during a sit to stand activity in an assistance-as-needed approach, the advantages of having an actuator at the ankle might not be enough to balance the negative effect that it has on wearability and energy consumption. However, in Fig. 2 the reader can clearly see that during level walking the assistive torque that is required at the ankle is significantly larger than the torques needed at the other joints. This difference becomes as large as a factor 5 between the required knee and ankle torques in the case of a muscle weakness of 90%. Because the final goal of the project is to build and test a prototype for sit to stand and walking, the ankle actuator has been already implemented.

B. Design approach

The commercial device that serves as a base for the exoskeleton is the hip brace 'HIPO' from Orthoservice (Fig. 3a). The HIPO brace is combined with the KAFO extension to create a full lower limb exoskeleton (Fig. 3b). However the foot part of the KAFO serves only for better attaching the device to the wearer and is not fit to transfer an assistive torque to the wearer. This part is thus replaced by Orthoservice's R.O.M. walker (Fig. 3c). The R.O.M. walker was chosen because of its design that is completely focused on walking. The curved sole was designed in such a way to promote a natural gait pattern. Since the commercial device is completely passive, actuators need to be mounted onto it to provide the wearer with the necessary assistance.

As stated earlier, the actuators are based on the MACCEPA principle [16]. Fig. 4 schematically shows the working principle of the actuators. The upper leg (1), lower leg (2) and the large pulley with lever arm (6) are hinged together by the joint axis (3). The position of the lever arm is set by the motor (4) on the upper leg, via a belt transmission between the large pulley (6) and the small pulley (5) that is mounted onto the driver axle of the motor. A cable (7) connects the lever arm, with the end of the compressive spring (8). When the angle between the lever arm and the lower leg increases, the cable pulls at the spring tail (9), compressing it. The compliance of the actuator can be changed by precompressing the spring. This is done by manually moving the spring head (10) down the lower leg, towards the tail.

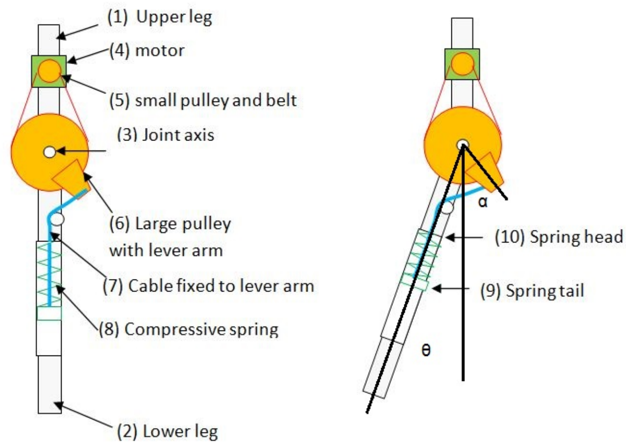


Fig. 4: Schematic representation of the working principle of the actuator.

Theoretically, the above described design would result in the characteristic that is shown in Fig. 5. The inputs that are required to construct the characteristic are: the spring constant (k), the length of the lever arm (D), the distance of the joint axis to the center of the pulley on the lower leg (L) and the torque angle necessary for an output torque of $15Nm$. The torque angle is defined as that angle that causes the spring to compress and as a result causes a torque output of the actuator. Physically it is the angle between the lever arm (6) and the lower leg (2) in Fig. 4. Earlier experience with MACCEPA actuation taught us that a torque angle of 20° for a peak torque of $15Nm$ leads to good results for the actuator's torque resolution. The values of k, D and L should the be chosen to reach this goal of $15Nm$ *per* 20°. The value of L and D is also limited by the dimensional constraints, i.e. the actuator has to fit onto the wearer's limbs. On the graph, the torque output of the actuator is shown as a function of the torque angle for different precompression rates. The graph clearly shows the linear behaviour of the actuator for a precompression rate of around 50% but more importantly it shows that theoretically the actuator should be capable of providing the required peak torques.

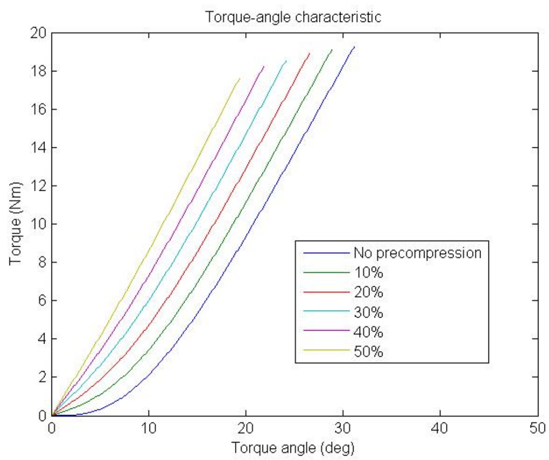


Fig. 5: Theoretical torque-angle characteristic of the actuator as a function of the precompression of the spring. Precompression is expressed as a percentage of the total compressable length of the spring.

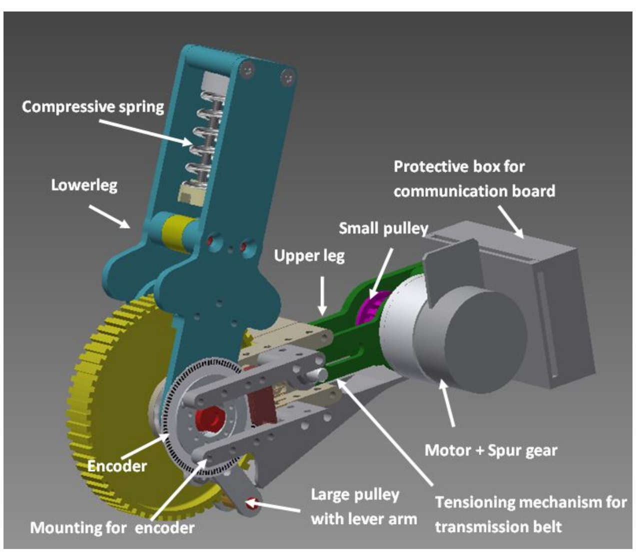


Fig. 6: CAD drawing of the actuator.

In Fig. 6, a 3D drawing of the actuator is shown. It weighs $1.4kg$, is about $90mm$ wide and $300mm$ long. The latter dimension was determined to be the maximum allowable length of the actuator when the exoskeleton was fitted onto a test subject of $1m65$ tall. Because of the belt transmission it is easily demountable since alignment is not as big of an issue as it would be with gears. To be able to make use of this belt transmission but not to excessively enlarge the width of the actuator, the authors have implemented an EC45 flat, outerrunner Maxon motor because of its limited dimensions.

Two encoders are mounted on each actuator: one incremental optical encoder and one absolute magnetic encoder. The optical encoder measures the angle between the upper link of the exoskeleton joint and the lower link (angle θ in Fig. 4). This encoder will also be used to derive an accurate velocity signal for the high level control that will be developed in the future. The absolute magnetic encoder determines the angle between the lever arm and the link onto which the motor is mounted (angle α in Fig. 4). The torque angle, which is the angle of the lever arm with respect to the other link, can then be obtained by calculating the difference between both measured angles.

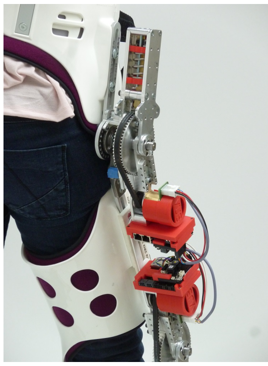
The communication with the actuators is done using the EtherCAT protocol via one hip joint that is linked to an extern computer. All other electronics boards are daisy chained. It is also noteworthy that each electronics board has its own CPU. As a result of this, not much external electronics hardware is necessary, which makes it a lot easier to implement the rest of the hardware on the back of the exoskeleton in a later stage. The power is supplied via one power cable per leg that is splitted towards the three actuators.

For more detailed information concerning the actuator design, recorded characteristics and performance, the authors refer to [18].

III. PROTOTYPE

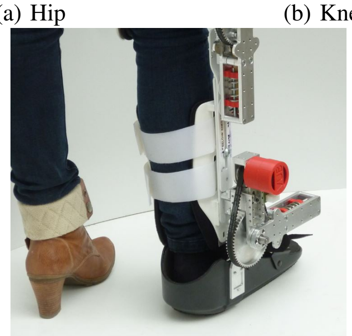
In order to eliminate alignment problems, the original joints of the exoskeleton are demounted and replaced by the actuator joints. This is shown in Fig. 7b for the knee joint and in Fig. 7c for the ankle joint. However, if the joint is loaded in non-saggital DOF, this could damage to the actuator. More specifically in the hip, where the biological range of motion and the torques during walking in the non-saggital DOF are relatively large [19], this is an actual risk. One option would be to redesign the actuator at the hip, making it able to withstand the load in the frontal plane. A less complex solution is to mount the actuator in parallel with the original hip joint as it was designed to operate smoothly during level walking. This set-up is clearly visible in Fig. 7a.

Fig. 8 represents a schematic front view of one leg of the exoskeleton and the mounting of the actuators. The exoskeletal support structure is drawn on the right, the braces are shaded. The 3 actuators are depicted in black on the left, with the shaded part representing the electronics board on top. The grey mounting pieces have been custom designed for each joint in such a way that only minor adaptations to the exoskeleton were necessary. The positioning of the mounting pieces as close to the joint axes as shown, allowed to reuse a lot of the mounting holes of the original joint hinges. More importantly, it has made it possible to preserve the exoskeleton's adaptability to a patient's height. The location



(a) Hip

(b) Knee



(c) Ankle

Fig. 7: Joints of the sit-to-stance prototype

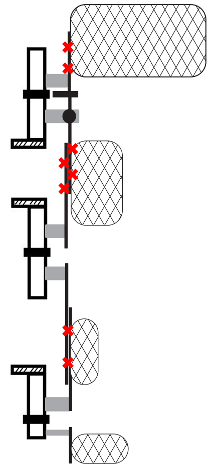


Fig. 8: Schematic view of the connections between the actuators and the exoskeleton.

Fig. 9: Unilateral version of the sit-to-stand prototype.

of the adjustment screws are marked by a red cross. They can all be accessed by means of a hexagonal wrench. Note that aside from the adaptability to the height of different subjects, the adaptability of the hip abduction/adduction has been preserved as well. The adduction joint is marked just below the hip flexion joint by a black circle. It suffices to demount the top mounting piece of the hip actuator to be able to acces the fixed adduction joint. The adduction joint is visible on the photo in Fig. 7a. It is covered in blue tape to damp the noise that results from the actuator executing a force onto the exoskeleton at that location.

In Fig. 9, you see the right leg of the MIRAD sit-to-stand prototype. The assembly of the actuators and the mounting onto the bilateral device took a grand total of two days. The entire exoskeleton weighs $13kg$, which consists of 6 times $1.4kg$ for the actuators and $4.6kg$ for the braces and the support structure. It can be adapted in height for wearers ranging between $1m65$ and $1m90$ and is easy to put on. Sit to stand actions were already performed while the device was not powered. Fig. 10 shows several stills taken from the movie of an unpowered sit to stand transition. From left to right several phases in the movement are shown. The first 3 shots represent: sitting, bending over to initiate standing and lift-off from the chair. In the last 3 shots all joint are extending, pushing the body upright. As the wearer experienced no difficulty in standing up, the implemented DOFs were considered sufficient to allow for a comfortable movement. Because of the flexibility of the braces and such, non-sagittal DOFs are not completely blocked in the prototype. Additionally, the sole of the R.O.M. walker was specifically designed for walking. Both facts make that it is still fairly easy to walk around wearing the device (unpowered). As such the current design might even serve as a suitable base for a gait-assisting prototype. Note that with this idea in mind, the actuator at the hip joint has been mounted up-side down. This way the electronics board is located at the thigh, where it least hinders the natural swinging motion of the arms during gait.

IV. CONCLUSION

In this paper the authors present an exoskeleton designed to assist the wearer during sit to stand activities. It consists of 6 active flexion/extension joints, that are driven by compliant MACCEPA-based actuators. Compliant actuation was chosen for its advantages in the area of wearability, safety and energy economy. The actuations follows an assistance-as-needed approach meaning that active cooperation of the wearer is strived for. This maintains neuromotor function and prevent disuse of the muscles. The actuators are mounted on an existing commercial orthosis to minimize expenses and to profit from the ergonomy and light weight of the orthosis.

Fig. 10: Stills taken from a video of an unpowered sit to stand transition.

The mounting of the actuators is done in such a way that the adaptability to a patient's height has been grossly preserved. Because of the high flexibility of the braces, the exoskeleton is also comfortable to walk in, which is the ultimate goal of the MIRAD project.

V. FUTURE WORK

The future work continues with getting the device fully operational. The exoskeleton will serve as a testbed for the high level control that is currently being developed by partners active in the MIRAD project. When the control is tested and approved, the exoskeleton will undergo sit to stand tests with healthy persons as well as clinical trials. These tests will provide us with valuable feedback concerning patient acceptance and the performance of the device. This will allow us to redesign the exoskeleton in a guided way, tackling the most crucial problems first.

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