Design and development of customized physical interfaces to reduce relative motion between the user and a powered ankle foot exoskeleton

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Abstract—Exoskeletons have shown their ability to assist locomotion and augment human performances. However, the benefits of wearing these devices depend on how effectively power can be transmitted from the device to the user’s biological structures. Recent studies have shown evidence of inefficient power transmission, with losses of up to 50%. The problem of power transmission can be mitigated by designing interfaces that increase contact stiffness and reduce relative motion between the limb and the robot. In this contribution, the design and development of physical interfaces for rigid lower limb exoskeletons is presented. The relative motion between the human and the developed interface is evaluated using a motion capture system and compared to the performances of a commercially available interface. Results indicate a clear reduction in relative motion between the user and the exoskeleton when the customized interface is worn.

I. INTRODUCTION

Robotic exoskeletons have undergone a rapid expansion over the last decade, showing promising results in areas such as rehabilitation engineering, assistance for daily activities and power augmentation for industrial applications [1]. However, the benefits of this new generation of devices are limited by how effectively power can be transmitted to the user’s biological structures. It is known that compliance at the physical interface can constitute a considerable energy sink [2]. And evidence of this inefficient energy transmission is growing, with reports of up to 50% of the mechanical power lost in the interaction due to soft tissue compression and harness compliance [3]. The problem of power transmission can be mitigated by designing interfaces that increase contact stiffness and reduce relative motion between the limb and the robot [4],[5], which we will further denote as interface migration. The main difficulty in developing interfaces with high contact stiffness lies in the fact that humans are composed of soft tissues, and consequently, braces tend to compress tissues and migrate rather than achieve leg motion [4].

One proposed solution is to compress soft tissues, as this will increase the stiffness of the limb [4]. Considering the fact that users may have to wear the devices for substantial amounts of time, and that comfort of these devices is still an issue [1], this design approach was not considered in this contribution. Another solution is to interact with bony prominences of the biological structures. These anatomical regions are stiffer but often sensitive to pressure. Although this principle was applied on a soft exosuit [6], the question for rigid lower limb exoskeletons remains.

The main objective of the contribution is to show the design and development of physical interfaces that reduce interface migration in comparison with a commercially available interface. On that regard, few research studies have successfully eval-

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Fig. 1. The customized physical interface, designed to increase contact stiffness and reduce relative motion between the user’s skin and the lower limb exoskeleton. The interface was 3D printed and reinforced with a carbon fibre composite to achieve a lightweight yet rigid structure.
The paper is structured as follows; in Sec.II, the different design criteria for the development of the customized interface are explained, as well as the realization or manufacturing of the prototype. After this, the experiment to evaluate the interface migration of other wearable robots, such as lower limb prosthetics [7], soft exosuits [6] and hand exoskeletons [8] by means of motion capturing techniques. In this study, a similar evaluation will be performed except the fact that, the developed interface is compared to a generic one, i.e. a commercially available interface which was previously used inside the MIRAD project [9].

The design of the interface was based on three design criteria; evaluated the interface migration of other wearable robots, such as lower limb prosthetics [7], soft exosuits [6] and hand exoskeletons [8] by means of motion capturing techniques. In this study, a similar evaluation will be performed except the fact that, the developed interface is compared to a generic one, i.e. a commercially available interface which was previously used inside the MIRAD project [9].

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II. METHODS

The methods section is divided in two parts. In the first part, we discuss the design criteria for the interface; the requirements for a successful interface implementation. In the second part, the methods for the evaluation of the interface’s performance are explained.

A. Design requirements and assumptions

The design of the interface was based on three design criteria;

1) Biological torque: Figure 3 shows the torque curves of the ankle and the knee in the sagittal plane, of healthy subjects with a mass of 75.0 kg at natural gait speed [10]. The sum of these torques is the resulting torque on the shank when both the knee and ankle are assisted. This torque is transferred to the leg through interaction forces (such as depicted on figure 2) and these interaction forces generate pressure on the soft tissues of the leg. Excessive pressure is one of the main concerns when applying loads to the human body [4]. Thus, interaction forces should be minimized and therefore, interaction surfaces should be distanced as far as anatomically possible. Contact area should be maximized in order to reduce the pressure magnitude and ensure a safe interaction.

It should be noted that increasing contact area to reduce pressure does not necessarily result in a more comfortable interaction, this is known as the concept of spatial summation [11].

2) Contact stiffness and migration reductive features:

The interface should keep the exoskeleton in place, i.e. it should prevent interface migration. This will prevent misalignments which can induce the generation of shear forces. Shear forces can induce friction which can cause discomfort to the user.

Additionally, the interface should be designed to increase the stiffness of contact between the user and the exoskeleton to increase power transmission [5],[4]. Increasing the stiffness of contact between the user and the robot can be achieved by interacting with stiffer anatomical regions such as bony prominences. On the other hand, interacting with bony prominences requires careful design. Anatomical regions such as the ankle malleolus and the knee condyles require protection. Previous designs have incorporated this design criteria by interacting with the shin [12],[13],[14]. Another proposal to increase contact stiffness is to compress soft tissues, effectively increasing the stiffness of the limb [4].
3) Lightweight & compact design: an important design factor is to reduce the added weight of the exoskeleton to the user. The first reason being that, added weight increases metabolic cost of wearing the device, especially at distal regions of the lower limbs [15]. And secondly, the added inertia will increase the power demands for the actuation system and reduce the amount of assistance the device can provide.

The frames are the mechanical linkages between the actuator and the interface and most exoskeleton designs employ metals for their frames [1]. An alternative is to use fiber-reinforced plastics in combination with additive manufacturing techniques to create hollow and lightweight, yet rigid structures.

Another aspect is the compactness of the device. Bulky exoskeletons can limit portability and form a practical barrier to their everyday use as a mobility device, limiting the user acceptance [16]. Frames and interfaces could be integrated into one single component to reduce the complexity and bulk of the device.

B. Interface realization

The first design criteria, which requires the interface to transfer a predominantly plantarflexor torque, resulted in the implementation of the two plantarflexor interaction surfaces, shown in figure 2. These are distanced as far as anatomically possible; on the upper side the knee cap is a region we do not wish to interact with, as it deforms during walking. And on the lower side, the lower end of the achilles tendon is a vulnerable region [4], and should thus be avoided.

Contact area of the plantarflexor surfaces were maximized to reduce pressure. The constraint for the surface is anatomical, as the foot should fit through the opening, between the two plantarflexor surfaces, to don the interface.

To increase contact stiffness between the interface and the user (second criteria) the author opted for a customized interface, i.e. a patient-specific design. This type of design is the status quo for the design of prosthetic sockets [17], where achieving stiff contact is critical [18].

To clarify, a generic interface that intends to accommodate a wider set of people, will necessarily add compliance in the system to compensate for anatomical differences between the subjects. This compliance is counteracting our design criterium, hence the proposed customization. Therefore, a digital description of one healthy subject’s lower limb was acquired with a structured light three-dimensional scanner (Artec Eva, Artec 3D [3]). This design method allows to interact with specific anatomical regions, such as the malleoli, while keeping a stiff interface structure. The aim of the design being to reduce migration from the user’s leg during load transfer and improve transmission of mechanical energy from the actuator to the user’s leg.

Fig. 4. A digital description of the shank and foot of a healthy subject was acquired with a structured light three-dimensional scanner (Artec Eva, Artec 3D [3]). This design method allows to interact with specific anatomical regions, such as the malleoli, while keeping a stiff interface structure. The aim of the design being to reduce migration from the user’s leg during load transfer and improve transmission of mechanical energy from the actuator to the user’s leg.

Fig. 5. Left: The customized interface, 3D printed with PLA (Polylactic acid) and reinforced with a carbon fiber and epoxy resin composite. Right: In this design Plastazote LD33 with a thickness of 3mm was used to manufacture the padding, as prescribed by Ottobock for AAFO applications [20]. The shape of the padding was derived from the digital description of the interface, and lasercutted to fit the interface precisely.

5. To limit the mass of the interface but also allow for rapid iterations of the interface, it was 3D printed in PLA using conventional FDM printing technology. The process took approximately 15 hours (Ultimaker 3, 60 mm/s print speed, 30% infill). The 3D-printed part was reinforced with carbon fibers using an artisanal method (marouflage), in which carbon fibers and epoxy resin are manually layed on top of the 3D printed structure.

C. Description of the experiment

In order to evaluate the performances of the customized interface, it was deemed interesting to compare it with a generic interface, depicted in figure 8. This type of generic
The subject’s left foot and shank were connected to the shank interface and footplate, which were attached to a series elastic actuator. The ankle actuator described in [21] was used in both trials to provide torque in the sagittal plane. The subject stood straight wearing the AAFO (Active Ankle Foot Orthosis) on one leg. The torque applied by the actuator was sinusoidal, with a frequency of 0.5 Hz and the subject was asked to resist the motion of the actuator. The torque amplitude was increased until the subject felt uncomfortable.

Fig. 6. The experiment was divided into two trials: one with the generic interface and one with the customized interface. It consisted of a quasi static setup, where a healthy subject stood straight in a controlled environment. The subject’s left foot and shank were connected to the shank interface and footplate, which were connected to a series elastic actuator. The ankle actuator described in [21] provided a sinusoidal torque reference while the subject was asked to resist the motion of the actuator. Reflective markers were placed on the shank of the subject and on the interface, depicted in figure 8. The position of the markers were measured using a Vicon motion capture system. A vector was generated between those markers, by subtracting their coordinate values. The sagittal- (blue) and transverse (purple) components of this vector were constructed and analyzed.

Fig. 7. The ankle foot actuator, based on MACCEPA principle used for the experiments. It is a series elastic actuator that generates a torque around the ankle joint when a differential angle $\alpha$ is created between the output link and the lever arm. The red dashed line represents the connection between the lever arm and the output link through the spring. When $\alpha$ increases, the red dashed connection pulls on the spring and a torque is generated. Energy is transferred to the shank through the interface and to the foot through the footplates. A detailed description of the actuator can be found in [21].

Fig. 8. The subject is wearing the customized interface (left), and the generic interface (right) connected to the ankle actuator. The reflective markers used to assess the relative motion between the interface and the leg are highlighted. The vector constructed with these markers is drawn in red. It is projected onto the sagittal and transverse planes to assess the relative motion between the interface and the leg, in the respective plane of motion.

Reflective markers were placed on the user’s leg and on the AAFO (17 for the generic interface trial and 19 for the customized interface trial), of which 4 were placed on the shank, 4 on the thigh, 3 on the actuator, 6 on the generic interface and 8 on the customized interface. The position of the markers were recorded using a Vicon motion capture system.
system.

Kinematics of the markers were measured using a Vicon motion capture system with a sampling rate of 100 Hz. Torques transferred to the interface are derived by measuring the deflection angle $\alpha$ of the actuator, by means of a magnetic encoder. The output angle of the actuator is measured with an optical encoder. All the actuator’s angles measurement were performed at a 1kHz sampling rate. The signals of the motion capture system and the actuator were synchronized using a triggering signal from the motion capture system. The participant gave written consent for the experiment.

D. Data analysis

The relative motion between the interfaces and the user’s shank were calculated as the difference in the position vector between two markers, the actuator’s side interface marker and the marker placed on the tibia bone of the shank. This vector was chosen as it best reflects the motion of the two bodies. This vector was then projected onto the sagittal and transverse planes of reference for evaluation. Marker data was filtered using a 4th order Butterworth filter with a cutoff frequency of 8 Hz. Actuator data was filtered with a 1st order Butterworth filter with a cutoff frequency of 15 Hz.

III. RESULTS

This occurred at 7.5 Nm amplitude for the generic interface. Although with the customized interface, an amplitude of 20 Nm was applied, only the 7.5 Nm region will be shown for the sake of comparison with the generic interface.

The relative motions in the sagittal and transverse planes as a function of the torque provided by the actuator are depicted in figure 9 and figure 10, respectively. The data for 50 sinusoidal torque cycles are reported.

It can be observed from the data that for both planes of motion (sagittal and transverse) the relative motion between the customized interface and the leg is considerably lower in comparison with the generic interface. Also, the hysteresis is significantly lower for the customized interface, which means less energy was dissipated. It should also be stated at this point that the user reported an uncomfortable sensation during the experiment with the generic interface, caused by the friction between the leg and the interface. This was the reason for which the torque applied was no longer increased for the generic interface (maximal torque amplitude of 7.5 Nm).

In the sagittal plane the relative motion curves are asymmetrical for both interfaces, i.e. the relative motion values have a bigger amplitude in the negative region. This means both interfaces allow more relative motion in the dorsiflexor phase than in the plantarflexor phase. On the other hand, in the transverse plane of motion, only the generic interface showed significant asymmetrical relative motion. In the low torque region reported here, the customized interface showed little to no relative motion in the transverse plane.

IV. DISCUSSION

In this experiment, both interfaces showed considerable asymmetrical relative motion in the sagittal plane of motion. The relative motion was higher in the dorsiflexor phase than in the plantarflexor phase. This phenomenon was expected considering that the dominant torque at the ankle joint in human gait is in the plantarflexion direction.

During the plantarflexor phase, energy is transferred to the tibia bone (rigid) through the main tibial interfacing surface (rigid). In the dorsiflexor phase, the energy is transferred to the gastrocnemius (soft) through straps (soft). Thus resulting
in a stiffer connection during plantarflexor phase compared to dorsiflexor phase. Additionally, the footplates of the actuator provide a series stiffness to take into account. The behaviour of the footplates were not recorded in this experiment, but a non-linear behaviour could be observed during the experiment. During dorsiflexion the footplates were bending, while this was not the case during plantarflexion, since the ground was providing resistance. This can be noticed in the attached videographical evidence [22].

V. CONCLUSIONS & FUTURE DIRECTIONS

An individualized physical interface of an active ankle foot orthosis was developed for a healthy test subject. Results indicate a clear advantage on the reduction of migration of the customized interface compared to a commercially available one. Future assisted walking trials on a larger group of people should indicate whether the proposed solution is promising. Subjective assessments of people suffering from different trauma might also reveal the true benefits of fully customized interfaces, i.e. the ability to interact with different regions of the body to achieve similar effects.

The intention of this contribution is to pave the way towards the design of more efficient and comfortable physical interfaces for wearable robots. An important aspect that was not assessed in this study, is the pressure on the user’s skin. First of all, as this prototype interacts with bony prominences of the lower limb, which are vulnerable areas, the integration of pressure sensors will be a key feature to ensure a safe and comfortable interaction in future designs. Additionally, misalignments between the exoskeleton and the human joint can be compensated for by allowing for relative motion. Preventing this relative motion might increase pressure on the user, unless self-aligning or misalignment compensation mechanisms are integrated in the kinematical chain of the robot.

Moreover, comfort is a vital aspect of physical interfaces that will enable the potential mass adoption of the technology. And in that regard, customization based on 3D scanning and 3D printing technology is a promising design framework. As discussed by Moerman et al. [17], interfaces for wearable interfaces such as sockets, are mostly non-data-driven designs and thus, we find a discrepancy in the quality of the interface. On the other hand, the integration of sensors to measure amplitude as well as spatial and temporal distributions of pressure along with a fully digitized description of the interface and subject, will allow for data-driven designs of physical interfaces. This will result in truly customized solutions which could have important implications on the benefit and mass adoption of wearable robotic technologies.

References


