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Soft artificial tactile sensors for the measurement of human-robot interaction in the rehabilitation of the lower limb

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Abstract— **A new and alternative method to measure the interaction force between the user and a lower-limb gait rehabilitation exoskeleton is presented. Instead of using a load cell to measure the resulting interaction force, we propose a distributed measure of the normal interaction pressure over the whole contact area between the user and the machine. To obtain this measurement, a soft silicone tactile sensor is inserted between the limb and commonly used connection cuffs. The advantage of this approach is that it allows for a distributed measure of the interaction pressure, which could be useful for rehabilitation therapy assessment purposes, or for control. Moreover, the proposed solution does not change the comfort of the interaction; can be applied to connection cuffs of different shapes and sizes; and can be manufactured at a low cost. Preliminary results during gait assistance tasks show that this approach can precisely detect changes in the pressure distribution during a gait cycle.**

I. Introduction

In the last decades, robot-aided rehabilitation was widely investigated and proved to be as good as or even better than conventional therapy ([1], [2]). Several different lower-limb rehabilitation robots have been proposed in the literature, belonging to two main architectures: end-point machines and multiple-point machines. In the first, the robot is connected to the patient only at the foot, which is then moved by the machine to perform the gait training task ([3], [4]). With this kind of devices, the imposed position or applied torque on each single joint cannot be controlled directly. On the other hand, multiple-point machines (exoskeletons) are connected to the user at several points, one or more for each limb ([5], [6], [7], [8]). This way, the exoskeleton can perform precise rehabilitation tasks by applying torques or predefined angular positions to each joint.

The most widespread way to interface the user with an exoskeleton is through connection cuffs: soft belts of adjustable size that are fastened to the user's leg. An example of this solution is the connection cuff commercialized by

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Fig. 1. The Lopes lower-limb rehabilitation exoskeleton, together with the sensorized fastening belts. In this example, only the right hip fastening point is sensorized with six pads (three in the front, and three in the rear of the belt).

Hocoma[1], and adopted in the Lokomat® exoskeleton ([8]), as well as in the LOPES lower-limb exoskeleton ([6]), shown in Figure 1. This solution is preferable for several reasons: it increases the contact area, thus reducing the interaction pressure and the stress on the user limb; it adapts to the shape of the leg (which, due to the contraction of muscles, changes during the gait) and it requires a small number of different sizes to fit most of the users.

The measurement of the interaction force between the user and the robot is commonly retrieved through a load cell placed at the connection between the cuff and the exoskeleton link ([7], [5]). In this work, we propose an alternative method to sense the interaction, based on a distributed measure of the normal pressure over the whole user-belt contact area. In Section II, we give the motivation and a rationale for our proposal. In Section III, we introduce the tactile sensing technology we will use in Section IV to implement in our sensorized connection cuff. In Section V we will present some preliminary results obtained during a gait assistance task performed with the LOPES on a healthy subject, and

[1]Hocoma AG, Industriestrasse 4 CH-8604 Volketswil, Switzerland.

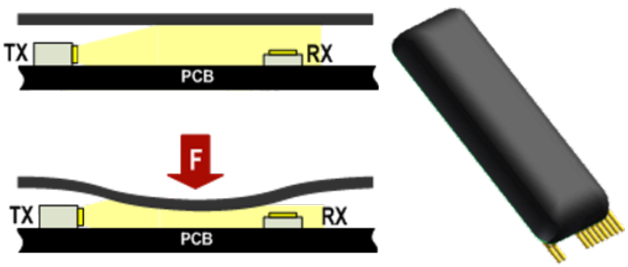

Fig. 2. Schematic representation of the Skilsens sensitive pad. On the right, the full pad is represented, along with its eight voltage output channels (two additional ones to power the device). On the left, a transversal section of the pad is shown. Labelled with “TX”, the LEDs, whose light is emitted along the horizontal axis. Labelled with “RX”, the light receivers. When a pressure is applied to the silicone bulk, the resulting deformation allows less light emitted by the LEDs to reach the receivers, resulting in a change in the voltage output.

will draw some conclusions.

II. Motivation and Rationale

In exoskeletal robots the force is transmitted by a pressure distribution over the whole contact area (i.e. over the whole connection cuff). This pressure is not uniform across the belt (higher pressure is likely to be found in the central part of the belt, along the direction of motion), and can potentially vary with time. How this pressure varies with time and space during a gait training task, however, is not known.

A widespread way to measure the interaction force is to apply a load cell at the connection point between the cuff and the exoskeleton link ([7], [5]). While this solution, though expensive, is practical and can give precise measurements, it does not allow to have a quantification of the whole pressure distribution on the fastening belt. This information could be useful both for assessment and control purposes. For example, it could be used to detect the user's motion intentions (e.g. change in the gait cadence or in the step amplitude) or to close a loop for interaction force control (e.g. to replace a series-elastic force sensor). These new tools could be useful to increase the performance of robotic trainers. On the assessment side this measure could allow to monitor the interaction comfort or to assess a robotic rehabilitation therapy. Moreover, with this solution the measure is the closest possible to the user, and all the dynamics of the interaction can be sensed. The potential benefit of this additional information could be huge, and still needs to be investigated.

For these reasons, we propose a distributed pressure measurement system to be applied to the whole user-robot contact area. To obtain this, we will sensorize the inside of a commonly used fastening belt with a soft artificial tactile sensor. In this way, we will not change the comfort for the user and will not change the pressure distribution, while getting the desired pressure distribution measure.

III. The soft tactile sensing technology

The sensitive element used in our solution is the Skilsens artificial tactile sensor ([9], [10]). This sensor can measure normal forces/pressures, while it is insensitive to shear forces. Figure 2 shows a sketch of the sensitive pad, along with a schematic representation of its working principle. This sensor is made of an external shell of soft silicone[2], covering a PCB which houses eight sensitive elements disposed along the sensors' longitudinal axis. Each sensitive element is made of a LED light source, and of a light receiver. Being made of widely spread electronic components, the Skilsens sensor can be manufactured at a very low cost. The application of this technology is not limited to the one presented here: other applications, such as a sensorized insole, or a sensorized fingertip, are currently being investigated.

The sensor is based on a double transduction principle. The silicone shell acts as a mechanical force/deformation transducer: in this prototype, a maximum load force of 6N can be supported by the bulk, when applied uniformly. [3] At the same time, an opto-electronic deformation/voltage transduction is performed. As depicted in Figure 2, which shows a transversal section of the sensor, the LED emits light across the longitudinal axis, which is received on the opposite side by a light sensor. The deformation of the silicone, which can be considered opaque, reduces the light received on the opposite side, so that the sensor acts overall as an inversely proportional pressure/voltage transducer. The spatial resolution in the pressure detection is determined by the number of light receivers and by the size of the silicone bulk. In the version used in this work, eight light sensors are used, and the overall size is of 60x20x4 mm. Note that in order to estimate the force distribution on the pad from the voltage outputs, a characterization of the sensor is necessary. This calibration ought to be done under different conditions in terms of pressure distribution, and is not detailed in this work.

IV. Sensing human-robot interaction in lower-limb rehabilitation tasks

As mentioned above, a widespread connection solution is that commercialized by Hocoma. This solution, shown in Figure 3a, is quite general and is used in other robotic platforms, among which the LOPES rehabilitation exoskeleton. We used this solution as a platform for our sensor system. The cuff is made of a rigid carbon fibre frame, connected to the exoskeleton by an aluminium bar. This frame houses an adaptable and flexible belt to be fastened on the user's leg. Our proposed solution, shown in Figure 3b and Figure 3c, is to put a number of sensory elements in between the user's leg and the belt. In order to house the Skilsens pads, and keep them fixed on the belt, we designed a rigid plastic

[2]In this prototype, *SORTA-Clear 40* Shore A 40 Silicone, Smooth-On, Inc.

[3]This value was chosen to have maximum sensitivity over the range of forces involved in the specific task detailed in Section IV. It is worth to note that under different situations (control algorithm, rehabilitation exoskeleton) the interaction forces may vary significantly from these.

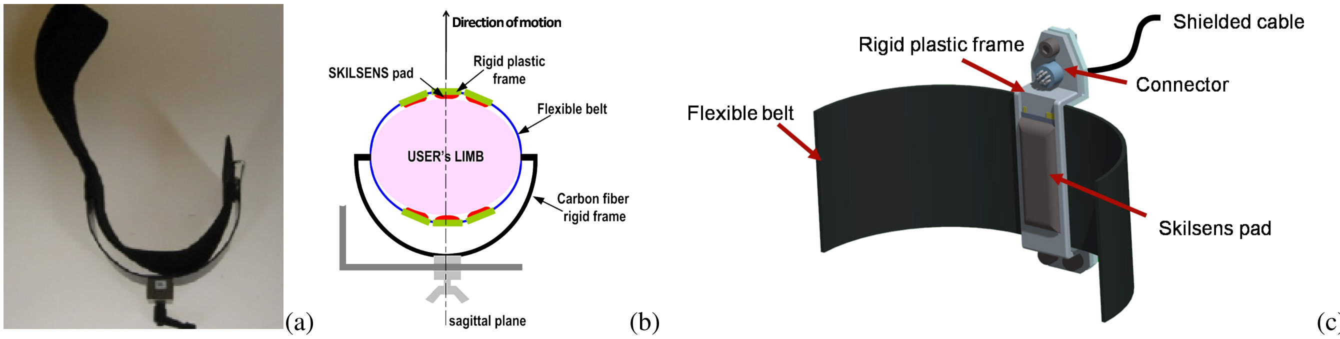

Fig. 3. (a) The Hocoma fastening belt. (b) Sketch of the sensorization of the fastening belt. The belt is connected to the exoskeleton through a rigid carbon fibre frame, and an aluminium bar. Some Skilsens sensorized pads are added in the inside of the belt, in order to measure the normal pressure distribution over the interaction area. Depending on the number and position of the sensors, it is possible to increase or decrease the spatial resolution in different parts of the contact area. (c) 3D sketch of the sensorized belt. For the sake of clarity, only a single sensorized element is depicted. The Skilsens pad is attached to the inside of the belt through a rigid plastic frame, which houses also the connector for the signal cable. This solution will allow to easily move the pads along the belt, as well as to quickly add or remove them. Note that only the silicone part of the sensor is in contact with the skin of the user, without changes in the comfort of the interaction.

frame, whose bulk is entirely on the outside part of the cuff. This way, only the silicone structure of the sensor is in contact with the limb, with no resulting changes in the interaction comfort. This frame, which also houses the connector for the compound signal/power cable, allows to easily increase or decrease the number of sensors distributed over the belt, as well as to quickly change their position. Due to encumbrance limits of the pad itself, a maximum number of 10 to 12 sensors can be applied per connection cuff. In order to preliminarily validate our approach, we sensorized a cuff with six Skilsens pads, three in the front and three in the back side, and mounted it on one of the six connections of the LOPES gait rehabilitation exoskeleton. Figure 1 shows the complete system, while Figure 4 shows the cuff being fastened on a healthy subject. The subject was requested to walk on a treadmill at a forced constant velocity of 4 Km/h. Three different working conditions were tested: a "no-assistance" condition, a "low-assistance" condition and a "high-assistance" condition. During all conditions the robot control provided a torque compensation for the gravity forces due to the exoskeleton masses. In the "no-assistance" condition, the robot was controlled in zero-torque mode. In the "low-assistance" and "high-assistance" case, an additional assistive torque was provided to the hip joint. While the description of the assistance algorithm is beyond the scope of this article, it is worth to say that the applied torque depended solely on the measured joint angle, and that the difference between the "low-" and "high-" assistance conditions was only due to a stiffness gain factor of a virtual impedance field imposed to the joint. For each of these conditions, the subject walked for a period of about one minute. In this preliminary analysis, we decided to ignore transient effects involved in the change between different assistance levels, and considered only the last 10 gait cycles of each condition, corresponding to a "steady-state" behaviour. During the whole task, we acquired the hip joint angle, the assistive torque exerted to the hip, and the voltage output of each of the 6 Skilsens pads (for each pad, eight voltage outputs

Fig. 4. First prototype of the sensorized belt during a gait rehabilitation task performed by the LOPES. In this example, the pressure is measured using six elements on the upper leg connection cuff (only three can be seen in this picture), and four (two visible) for the lower leg.

were recorded). All the data, acquired for the right leg only, were low-pass filtered and averaged over the steady-state gait cycles.

V. Results

Figure 5a shows the preliminary results we obtained on one subject. From the top to the bottom panel, the hip flexion/extension angle, hip assistive torque, upper leg front-central pad output, upper leg rear-central pad output. For the sake of simplicity, only the output of two pads is reported (the two central ones at the opposite sides of the upper leg). For each pad, the mean of the 8 channels output is shown. In this preliminary analysis, the raw voltage output of the pads is given, instead of an estimation of the pressure distribution along the sensor's bulk. Figure 5b shows the 8 voltage outputs of the front pad averaged over the steady state gait cycles. The x-axis represents the percentage of the gait cycle,

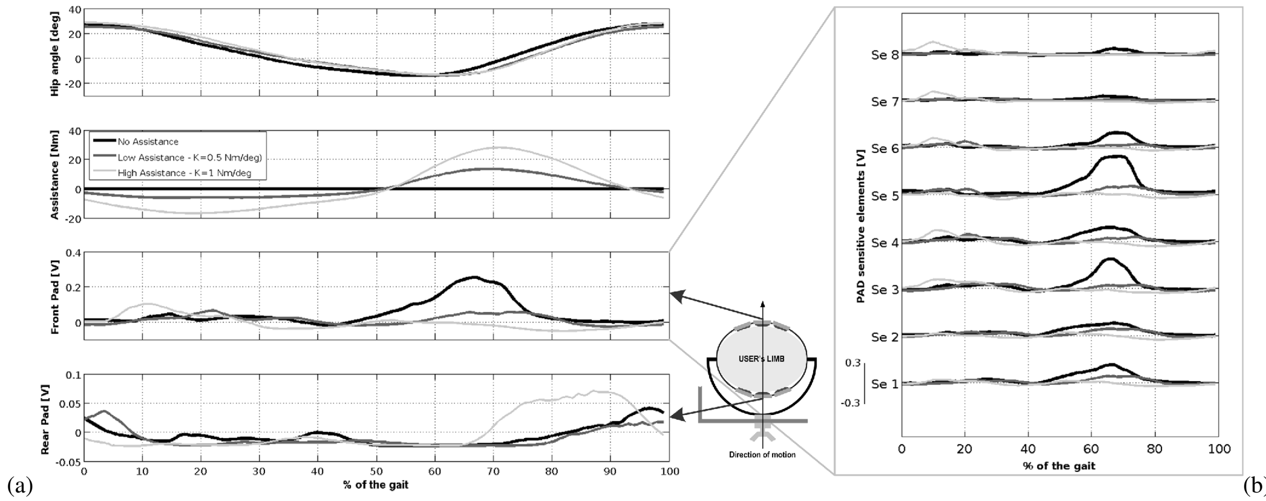

Fig. 5. (a) Preliminary results during a gait assistance task. From the top to the bottom panel: hip angle, hip torque assistance, upper leg front pad output, upper leg rear pad output. The eight voltage output of each pad are averaged to extract a single signal, and then averaged again over 10 gait cycles. (b) A detail on the 8 channels voltage output of the front pad. The outputs of the 8 sensitive elements are averaged over 10 gait cycles.

and a division is made between the stance phase (0-60%) and the swing phase (60-100%). At 0% of the gait cycle, the heel is touching the ground after landing, while at 60% the toe leaves the ground to begin the swing phase. The results are reported for the three different assistance conditions. Looking at the first two panels, one can see that while the gait shape does not change significantly, the assistive torque differs in the three conditions. Focusing on the pads output in the "no-assistance" case, one can see that the beginning of the swing phase is marked by a strong increase in the pressure on the front pad. This outlines the force transmitted from the user to the robot, in order to accelerate it. On the other hand, an increase in the rear pad pressure marks the beginning of the deceleration of the limb (at 90-100% of the gait), which is imposed by the user on the exoskeleton. The opposite condition, "high-assistance" shows a very different output of the two pads. The frontal pad does not show the force imposed by the user to accelerate the robot any more, thus the voltage increase in the 50-80% range is no longer visible. On the other hand, a strong force is sensed in the rear pad in the 70-95% range of the gait cycle, corresponding to the exoskeleton "pushing" the limb in the forward direction. The "low-assistance" situation shows an intermediate behaviour between the two aforementioned cases. Figure 5b shows the voltage output of the single sensitive elements of the frontal pad. One can see that the same pattern shown for the whole pad in Figure 5a can be found in each element, even though some quantitative differences exists between the 8 channels. This suggest a non-uniform pressure distribution over the interaction area.

VI. CONCLUSIONS

In this paper, we proposed a new method to measure the interaction force between the user and a lower-limb gait rehabilitation exoskeleton. Instead of using a load cell to measure the resulting interaction force, we proposed a distributed measure of the interaction pressure obtained by inserting soft silicone tactile sensors between the user's limb and the connection cuff. We preliminary validated our approach by applying it to the connection belt of the LOPES lower-limb exoskeleton. The approach proved to be effective to precisely detect changes in the pressure distribution during a gait cycle. Future work will include a full validation of our sensor on an increased number of subjects, possibly involving patients suffering motor impairments.

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