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Novel Compliant Actuator for Wearable Robotics Applications*

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***Abstract*— In the growing fields of wearable robotics, rehabilitation robotics, prosthetics, and walking robots, variable impedance and force actuators are being designed and implemented because of their ability to dynamically modulate the intrinsic viscoelastic properties such as stiffness and damping. This modulation is crucial to achieve an efficient and safe human-robot interaction that could lead to electronically generate useful emergent dynamical behaviors. In this work we propose a novel actuation system in which is implemented a control scheme based on equilibrium forces for an active joint capable to provide assistance/resistance as needed and also achieve minimal mechanical impedance when tracking the movement of the user limbs. The actuation system comprises a DC motor with a built in speed reducer, two force-sensing resistors (FSR), a mechanism which transmits to the FSRs the torque developed in the joint and a controller which regulate the amount of energy that is delivered to the DC motor. The proposed system showed more impedance reduction, by the effect of the controlled contact forces, compared with the ones in the reviewed literature.**

I. INTRODUCTION

A novel compliant actuation system is proposed for use in wearable robotics applications where large forces are required to support and assist the user movements (i.e. a gait rehabilitation robot). In case of post-stroke patients only the affected leg has to be supported or assisted while the movement of the unaffected leg should not be hindered. Not hindering the motions of one of the legs means that mechanical impedance of the robot should be minimal. Another application for compliant actuators can be found in walking assistive exoskeletons, which are designed to relief a considerable amount of the user bodyweight by identifying certain walking stages, in order to develop torque in the specific actuators when assistance is needed and also to track the user limbs with minimal impedance when no assistance or resistance is needed.

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In [1] a compliant actuator is defined as a device that will allow deviations from its own equilibrium position, depending on the applied external force. The equilibrium position of a compliant actuator is defined as the position of the actuator where the actuator generates zero force or zero torque. The compliance adaptability can be achieved by two methods: passive and active compliance; the first one uses a passive elastic element (spring) to store energy, which is not the case for actuators with active compliance, where the controller of a stiff actuator mimics the behavior of a spring. Many issues arising in the design of interactive wearable robots are similar to those appearing in the field of haptic, or more precise: kinaesthetic robotics [2].

The characteristic feature of these robots is the bi-directionality of being able to both 'read from' and 'write to' a human user. In [3] are proposed two basic control schemes to achieve bi-directional interaction between human and robot: impedance-control-based that 'measure position and display force', and admittance-control-based that 'measure force and display position'. In this work we propose an alternative way to achieve the bi-directionality of an interactive wearable actuator in which we measure contact forces in the actuator and display torque at the DC motor. Further details of our approach will be given in the next sections.

There have been different approaches that combine the passive and active methods with any of the control schemes, Impedance or Admittance, to produce a bi-directional adaptable compliance system. In [4] is presented a rotational actuator with adaptable compliance, (MACCEPA for Mechanically Adjustable Compliance and Controllable Equilibrium Position Actuator), achieved by a spring attached between a lever arm and the output element of the actuator sharing both the same rotational axis. The lever arm is in charge to set the equilibrium point while the output element is connected to the load so that external forces tending to move the equilibrium point can be absorbed by the spring. Compliance adaptability is achieved by stretching and relaxing the spring using a servomotor placed at the output element to do the task.

In [5], they propose an actuator with adjustable stiffness (AwAS) that can independently control equilibrium position and stiffness by two motors. The mechanism designed to achieve compliance adaptability is quite interesting because they move the attaching points of the springs in a perpendicular direction to the force exerted by them, which requires less energy and a smaller actuator can be used to do the job.

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We propose a *Rotational Interactive Actuator* with *"Active Compliance"* RIAwAC flexible enough to be attached in active orthoses for upper and lower limbs as well. As an application example, Fig. 1 shows a knee orthosis with the actuator attached to it. The actuation system uses an electromagnetic actuator as the torque source, more specifically a DC motor with a built-in speed reducer. The total weight of the actuation system with the electronics is relatively low (0.50 kg).

Figure 1. The RIAwAC attached to a mechanical orthosis designed to provide assistance/resistance in the knee joint.

II. DESCRIPTION OF THE RIAWAC SYSTEM

The entire actuation system is composed of 10 basic elements which are listed as follows:

1) Geared DC motor (RX-64 Dynamixel, 7.5 Nm)
2) Output Torque Transmission Element
3) Load Link (free rotational movement)
4) Big supporting arm
5) Small supporting arm
6) Stationary Link
7) Force Sensors left and right (FSR Tekscan)
8) Damping Silicon Discs
9) Bearings
10) Fasteners

Fig. 2 shows the disassembled elements of the actuator.

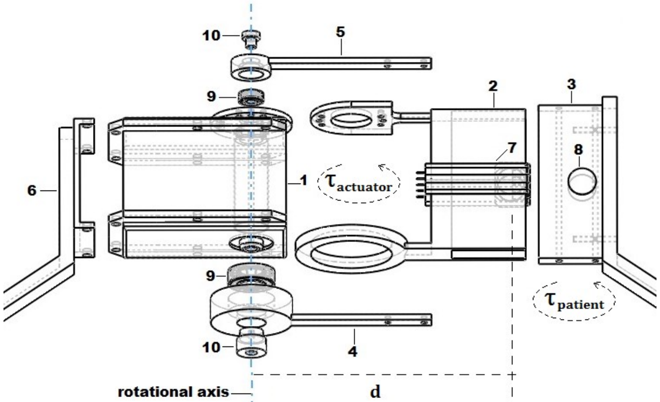

Figure 2. Components of the RIAwAC (disassembled view)

The main idea of this mechanical configuration is that the load torque applied by the patient between the Stationary Link 6 and the Load Link 3 around the rotational axis can be measured by the Force Sensors 7 placed in both sides (Left and Right) of the Output Torque Transmission Element 2 to measure force in both directions.

Measuring interactive forces developed inside the joint is achieved by both Force Sensors 7 placed inside the walls of the Output Torque Transmission Element 2 which is attached directly to the output shaft of the geared DC motor 1 and is in charge of transmitting the generated torque $\tau_{actuator}$ to the Load Link 3. This way, it can be possible to measure the interaction between the user and the RIAwAC actuator in terms of contact forces F_c.

Only one force sensor (Left or Right) can be activated at the same time depending on the direction and the magnitude of the torque provided by the patient, $\tau_{patient}$ and the torque developed by the actuator, $\tau_{actuator}$. When the relative rotating directions between the Load Link 3 and the DC motor 1 are opposite then the contact force magnitude is:

$$\mathrm{F_c} = \frac{(\tau_{\mathrm{patient}} + \tau_{\mathrm{actuator}})}{\mathrm{d}} \qquad (1)$$

The force sensor activated is the left one 7L when the direction of the Load Link torque is in the CCW and the motor torque is in the CW direction. The activation of the right sensor 7R can be obtained with the same reasoning.

When the relative rotating directions between the Load Link 3 and the DC motor 1 are equal then the contact force magnitude is:

$$\mathrm{F_c} = \frac{(\max[\tau_{\mathrm{patient}}, \tau_{\mathrm{actuator}}] - \min[\tau_{\mathrm{patient}}, \tau_{\mathrm{actuator}}])}{\mathrm{d}} \qquad (2)$$

Then, the force sensor activated is described in the following table:

TABLE I. ACTIVATION OF FORCE SENSORS DEPENDING ON THE MAGNITUDE AND DIRECTION OF THE USER AND MOTOR TORQUE

	CW rotation	**CCW rotation**
$\boldsymbol{\tau_{patient} > \tau_{actuator}}$	7R	7L
$\boldsymbol{\tau_{patient} < \tau_{actuator}}$	7L	7R

The chosen sensors to measure forces are Force Sensor Resistor (FSR) from FlexiForce® brand. The selection of this kind of sensor is due to its high linearity (±5%), high repeatability (±2.5%), low hysteresis (4.5%), low noise, low power consumption and low purchasing cost.

III. CONTROL SYSTEM BASED ON FORCE EQUILIBRIUM

The block diagram representation of the control system implemented to achieve force tracking and minimal mechanical impedance is depicted in Fig. 3. The amount of voltage and polarity delivered to the DC motor by the controller are calculated based on the deviation in the FSR's signals. The controller input is the difference between the signals coming from the force sensors as expressed in (3).

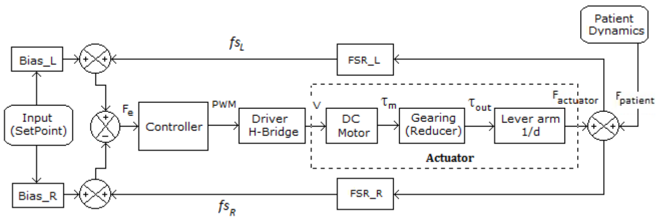

Figure 3. Block diagram of the force control system for the RIAwAC

$$F_e(t) = fs_L(t) - fs_R(t) \qquad (3)$$

The controller output is the duty cycle of a Pulse Width Modulated signal (PWM) which activates the H-Bridge (driver) to feed the DC motor. The amount of voltage calculated by the controller is proportional to the difference existing between the FSR's signals, the bigger the difference the greater the voltage needed to equilibrate the system and return to zero the contact forces F_c. Reducing to zero the contact forces means that the actuator is perfectly compensating the input load force applied by the patient at the Load Link (Fig. 2) and thus reducing to zero the mechanical impedance of the geared DC motor.

The controller is designed to always reduce the input error. When a given load is applied to the actuator $F_{patient}$, it activates one of the two force sensors (depending on the relative directions of the load applied by the patient and the force developed by the actuator, $F_{actuator}$). If the right sensor is activated the input error F_e is negative and the controller must compute a CW manipulation with magnitude proportional to the error in order to reduce the contact force perceived by the FSR. If the left sensor is activated the error F_e is positive and then the controller must generate a CCW manipulation to reduce the input error. The result of this approach based on equilibrium forces is a natural-efficient-low cost way to achieve Minimal Mechanical Impedance. Fig. 4 shows the experimental results when testing the mechanical impedance in controlled and uncontrolled situations. We applied by hand a 3Hz quasi-sinusoidal input load with the controller online, a few cycles after we turned it off and kept applying the load. The results show an average force reduction of 78 N in the mechanical impedance (15 dB) using a Proportional Fuzzy controller.

To achieve force tracking with the proposed control scheme in Fig. 3, we introduce two variables in the system Bias_L and Bias_R. These variables are used to add a deviation in the sensors signals fs_L and fs_R respectively. The main idea is to "deceive" the controller by adding a deviation in any of the sensors signals to "unbalance" the system even if no external load is applied to the actuator. The result is a virtual input error that doesn't exist but will force the controller to generate a manipulation that reduce that virtual error, in such a way that the controller will tend to track the deviation placed in any of the two variables to balance the system. The PID parameters are: $K_p = 6.7$, $K_i = 1.8$ and $K_d = 0.9$. The Proportional fuzzy controller uses 7 symmetrical membership functions for the error and manipulation variables, Mamdani inference product and centroid method for defuzzification.

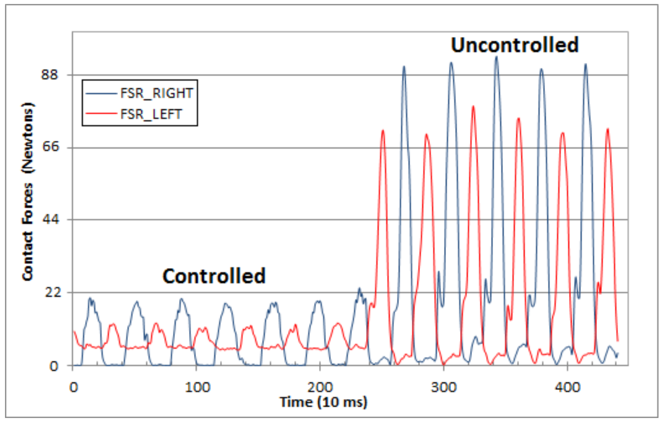

Figure 4. Controlled and uncontrolled response of the RIAwAC in terms of contact forces @3Hz

Fig. 5 shows the force tracking results using both a PID controller and a Proportional Fuzzy controller.

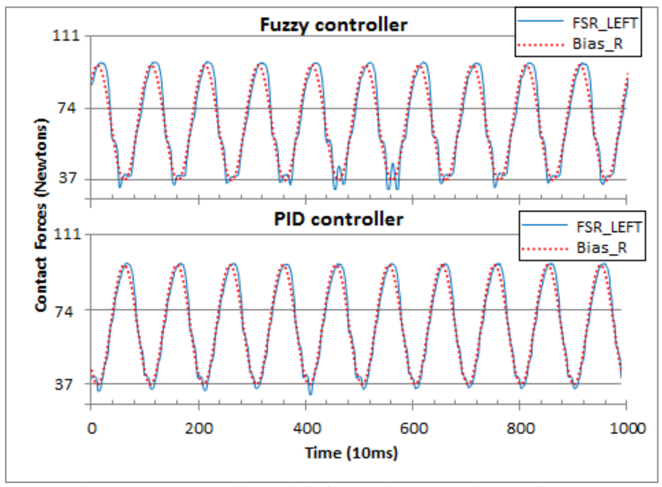

Figure 5. Force tracking with @1Hz sine as the input reference

The experimental results show a better tracking performance using the PID controller achieving 99.91% of fidelity with a reference sinusoidal signal of 1Hz compared to 94.62% achieved by the Proportional fuzzy controller. A variance-accounted-for factor based measure is used to quantify the corresponding fidelity, defined as

$$\text{Fidelity} = \left(1 - \frac{\text{var}(y - Y)}{\text{var}(y)}\right) 100\% \qquad (4)$$

Where y is the vector of the sampled measurement and Y the vector of the reference sine.

The frequency response of the closed loop system was evaluated using a sinusoidal input with amplitude of 67 N and a frequency from 0 to 5 Hz. The attenuation breaking point starts around 2.5Hz for the Fuzzy controller and around

3Hz for the PID controller. At 4 Hz (which is the max walking frequency for an average person) the phase angles and amplitude attenuations are 70º and 20% respectively for the proportional fuzzy controller and 35º and 30% for the PID controller as shown in Fig. 6.

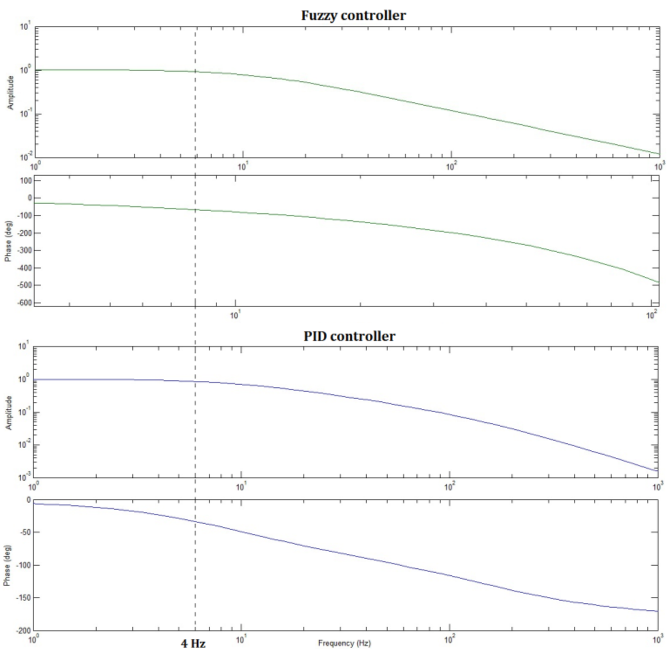

Figure 6. Bode plots of the closed-loop response tracking a sinusoidal reference with amplitude of 67 N.

IV. RIAwAC IN ACTIVE ORTHOSES FOR REHABILITATION AND ACCURATE DIAGNOSIS

The proposed control system shown in Fig. 3 provides the flexibility and autonomy for the RIAwAC to be part of a centralized system (i.e. powered exoskeleton) composed of multiple actuators and one main controller in charge to determine the assistance, minimal impedance or resistance needed in each joint, depending on the walking stage in course and the pathology presented by the patient. The flexibility/autonomy of the actuator is attributed to the fact that it only needs to receive a *SetPoint* from the main controller to perform assistance, resistance or minimal impedance. The RIAwAC can be very useful in an active orthosis to assist in the rehabilitation process. On one hand the therapist could evaluate online and more precisely the patient mobility range of a given articulation (i.e. the knee) and muscular strength using the information from the force sensors and potentiometer placed in the actuator, in order to diagnose the patient evolution. In the other hand the therapist could use the information gathered from the sensors in the evaluation process and then program the RIAwAC to assist or resist to specific movements of the patient with a controllable force depending on the rehabilitation stage of the patient. In this way, the rehabilitation process could be more effective and achieved in a shorter period of time.

V. CONCLUSION

The RIAwAC showed a reduction in the mechanical impedance of 15dB compared to 11dB from MACCEPA [6] and 13dB from de Distributed Series Elastic Actuator [3].
This significant reduction is due to the combination between the fast response of the DC motor in transmitting torque to the load joint and the control scheme based on equilibrium forces, although the sampling time of the control system is crucial to respond as fast as the highest load frequencies introduced by the user.

In tracking forces the RIAwAC and MACCEPA achieved similar results of fidelity using a sinusoidal reference, 99.91% @1Hz and 98.9% @1.5Hz respectively. The performance of the proposed actuator was better compared to the Distributed Series Elastic Actuator which showed 40º of phase angle and 74% of amplitude attenuation @4Hz using 2 Nm amplitude of the sinusoidal reference, RIAwAC showed 35º phase angle and attenuation of 30% @4Hz and 2 Nm.

In general the proposed actuation system showed encouraging results in tracking forces and indirectly reducing the mechanical output impedance of the actuator. Also the actuation system met the specifications in frequency (4Hz) although maximum torque developed by the geared DC motor is almost one third of the average torque developed by a knee in a person (20Nm), but still these results can be extrapolated to a bigger motor.
The advantage of an active compliant actuator can be better appreciated from a design point of view; we were able to reduce the weight, housing space, mechanical complexity, implementation cost and at the same time we improve the performance needed in an interactive-flexible actuator for wearable robots applications with no sensors attached directly to the user to detect the intention. Clinical tests are the next step to check the reproducibility and also the effect of sensors deterioration in rehabilitation.

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