Abstract—This paper introduces the design, modeling and control of a novel inherently compliant actuator with the ability to manually reconfigure the level of stiffness (CompAct-ARS). CompAct-ARS is intended to be utilized in assistive devices for legs, which will aid users in different motion tasks such as walking, sit-to-stand movement and squatting and will demonstrate the benefits of passive compliance. The actuator modeling design and mechanics are introduced. Experimental and simulation data are used in order to determine the design specifications of the actuator for its use in a knee exoskeleton system. Finally an assistive control strategy is introduced that will allow the proposed system to provide motion assistance to the subject knee.

I. INTRODUCTION

Lower limb exoskeletons can be used for human performance augmentation, assistive locomotion and rehabilitation purposes. Human performance augmenting exoskeletons can enhance the user’s capability to carry heavy loads and/or increase the muscle activation during different motion tasks [1], [2], [3]. Assistive exoskeletons are aimed at assistive ambulation for individuals with impaired legs [4], [5], although they can also be used to assist sound function. On the other hand the goal of rehabilitation exoskeletons [6], [7] is to provide a gait rehabilitation solution to patients preferably when residing outside the clinic and without the support of a physiotherapist.

These robotic devices have mutual requirements regarding their actuation performance, the weight and size of both actuator and structure, the wearability of their design and the safety of the user. Researchers who are focused on the development of exoskeletons have made efforts to address these aspects. However, there still remain limitations to be overcome before this technology can be used effectively [8]. As exoskeletons are devices wearable by humans, safety of interaction is crucial requirement in preventing both injury and discomfort [9]. The enabling technology and the control strategy are the keys to achieve safety and adaptability of physical human-robot interaction (pHRI). To date the majority of the presented exoskeletons employ stiff actuators, which impose on the exoskeletons significant limitations related to safety issues and the ability to interact with human. This substantially relies on the fact that the high frequency output impedance of a stiff actuator is dominated by the reflected inertia of the rotor. For this purpose, safety-oriented control techniques have been proposed to improve the dynamic behavior of stiff actuation during pHRI [10], however there are ineffective for frequencies above the closed loop bandwidth of the control system.

Introducing inherent compliance to the actuation system decouples the inertia of the motor drive from the output link and renders the output mechanical impedance lower across the frequency spectrum [11]. Thus, the robot improves its ability to intrinsically absorb impacts and enhances physical human-robot interaction. In addition, due to the presence of the employed passive elements which can store and release energy during one cycle of a periodical motion, compliant robots can be more energy efficient as compared to traditional stiff actuated systems. This feature is critical in the creation of energetically autonomous exoskeletons for locomotion.

In order to achieve the features described above, our primary goal is to build an inherently compliant lower body exoskeleton to be used as a lower limb assistive device. First step along this path is the development of a compliant knee exoskeleton, which will provide motion assistance both to patients with impaired legs and healthy individuals for different motion tasks such as walking, standing up-sitting down and squatting. The realization of such a compliant exoskeleton remains a challenging task requiring more compact, modular and high performance actuation units. This paper presents the design, modeling and control of CompAct-ARS (Actuator with Reconfigurable Stiffness). CompAct-ARS is an intrinsic compliant actuator with manually reconfigurable stiffness which has the ability to provide the required assistive torques to the knee joints of the wearer.

II. COMPACT-ARS

A. Functional Principle of CompAct-ARS

CompAct-ARS as mentioned above is a series elastic actuator with the ability to regulate off-line the level of stiffness in as wide a range as needed. This feature permits the experimentation with different compliance levels and the adaptation of the joint to fit specific task requirements. The elimination of active tuning of the spring stiffness through a second motor was performed to reduce the weight and dimensions of the unit. The working principle of the CompAct-ARS is based on the CompAct-VSA (Variable Stiffness Actuator) [12], which uses a lever arm mechanism with a variable pivot axis. As it is shown in Fig. 1 the springs are kept fixed while the position of the pivot is changing.
Thus, the apparent stiffness at the lever arm is adjusted theoretically from zero to infinite by moving the pivot point. If the pivot moves to the end of the lever which is connected to the springs the stiffness turns to zero, whereas if it moves to the other end the stiffness turns to infinite and the lever becomes rigid.

**B. Mechatronic Design**

As series elastic actuator, CompAct-ARS is consisted by two subassemblies: the elastic module and the motor housing. The motor housing subassembly includes among others a Kollmorgen RBE1810 frameless brushless motor and a harmonic drive CSD 25 with gear ratio 100:1. The elastic module contains the compliance elements and the lever arm mechanism with the reconfigurable pivot point (Fig. 2(b), 2(c), 2(d)). As it is shown in Fig. 2(a), the location of the pivot can be altered by tuning two set screws and so the apparent stiffness at the output load can be reconfigured according the application.

The employed springs are pre-compressed half of the maximum compression which is 20 mm. The selection of the springs’ constant $k_s$ and their maximum displacement is described in subsection III-B. In addition, the actuator contains four position sensors and one torque sensor. One 16 bit optical encoder (Avago Technologies) mounted on the elastic module measures the angular deflection of the link while a potentiometer (Austria Microsystems) measures the pivot position. On the other hand, the motor housing subassembly employs two encoders and one torque sensor. The torque sensor, which utilizes semiconductor strain gauges, measures the torque provided by the motor. One 12 bit optical incremental encoder (MicroE Systems) monitors the position of the motor and one magnetic absolute encoder (Austria Microsystems with 12 bit of resolution) measures the position of the motor after the harmonic drive.

**C. Stiffness Modeling**

The stiffness model of the CompAct-ARS is derived from the lever arm mechanism which was described in subsection II-A and is demonstrated with more details in Fig. 3. Suppose that the lever arm is at its equilibrium position and we apply an external torque $\tau_{ext}$ which rotates the lever arm with an angle $\phi$ with respect to the pivot ($P$). As a result, the springs are displaced by $x_s$ and consequently generate a force $F_s$ applied in ($\Gamma$) given by

$$F_s = 2k_s l_1 \tan \phi$$  \hspace{1cm} (1)

where $l_1$ is the distance form the pivot ($P$) to point ($\Delta$) which is the connection point between the springs and the lever at lever’s equilibrium position. The perpendicular force to the

![Fig. 1. Concept of working principle of CompAct-ARS; Stiffness is adjusted by varying the location of the pivot.](image1)

![Fig. 2. a) Section of the CAD assembly of CompAct-ARS. b) Horizontal section view of the elastic module. c), d) Views of the first prototype’s elastic module. e) Overall view of CompAct-ARS.](image2)
lever arm is $F_s \cos \phi$. Due to the lever mechanism, a force

$$F_l = F_s \cos \phi \frac{PT}{AP} = 2k_s l_2^2 \tan \phi \frac{1}{AP}$$

(2)

is therefore applied in (A), where $\frac{PT}{AP}$ is the amplification ratio and

$$AP = \sqrt{R^2 + d^2 + 2Rd \cos \theta_s}$$

(3)

where $R = AO$, $d$ is the distance from the center of rotation of the joint ($O$) to the pivot ($P$) and $\theta_s = q - \theta_e$ is the angular deflection of the elastic transmission defined as the difference between the angle of the output link $q$ and the angle of the motor $\theta_e$ after the reduction drive. The resultant force $F_E$ along the perpendicular to $AO$ is

$$F_E = \frac{F_l}{\cos \epsilon} = \frac{F_l}{\cos (\theta_s - \phi)} = 2k_s l_2^2 \tan \phi \frac{1}{AP \cos (\theta_s - \phi)}$$

(4)

As $\sin \phi = \frac{R}{AP} \sin \theta_s$, the (4) becomes

$$F_E = \frac{2k_s l_2^2 \sin \phi}{AP \cos \phi \cos (\theta_s - \phi)} = \frac{2k_s R^2 l_2^2 \sin \theta_s}{AP^2 \cos \phi \cos (\theta_s - \phi)}$$

(5)

Thus, the elastic torque $\tau_E = F_E R$ which at the equilibrium counterbalances the external torque is

$$\tau_E = \frac{2k_s R^2 l_2^2 \sin \theta_s}{AP^2 \cos \phi \cos (\theta_s - \phi)}$$

(6)

The allowable range of the deflection of the elastic transmission $\theta_s$ is from 0 to 0.2 rad. For this small range of deflection the elastic torque expression can be simplified as

$$\tau_E = \frac{2k_s R^2 l_2^2 \theta_s}{l_2^2}$$

(7)

Finally the stiffness of the elastic mechanism $K_s = \frac{\partial \tau_E}{\partial \theta_s}$ is formulated such as

$$K_s = \frac{2k_s R^2 l_2^2}{l_2^2} = 2k_s \alpha^2 R^2$$

(8)

where $\alpha = \frac{l_1}{l_2}$ is the ratio of the lever arm.

### III. Knee Exoskeleton Application

#### A. Knee Exoskeleton Specifications

Acquiring the knowledge behind the biomechanics of human lower limbs plays crucial role in the design of exoskeletons and orthotic devices for locomotion. Many researchers have used Clinical Gait Analysis (CGA) data as a primary basis for the design specifications of the exoskeletons [1], [13]. From open literature it is possible to perceive that running and standing up motion are the most demanding tasks in terms of torque, power and range of motion of the human knee joint [14], [15], [16]. Therefore, some of the specifications for the proposed actuation unit have been obtained from simulation and experimental data of the sit-stand-sit movement cycle.

For this purpose, a simulation model of human has been developed. This 6 DOF (hip, knee, ankle) planar model can replicate the sit-stand-sit motion of the human in the sagittal plane (Fig. 4). For the simulations a male of 82.5 kg with 1.85 m height has been considered according to the anthropometric data in [17]. The lumped mass models of the shank, thigh, pelvis and torso have also been considered according to the anthropometric data in [17]. To derive the torque characteristics of each joint of the human leg two different types of joint trajectories were used. Firstly, parametric trajectories based on literature biomechanics data [16], [18], [19], [20] were used as input to the human model. Fig. 5 demonstrates the parametric trajectories and the corresponding torque characteristics for the standing up motion. For evaluation and comparison the following experiment took place: a healthy subject with a weight of 81 kg (this is close to the simulated model weight of 82.5 kg) was instructed to repeatedly stand up and sit down without hand assistance (worst case/highest torques). A Vicon motion capture system with 6 cameras operating at 250 Hz was used to monitor the trackers’ position and reconstruct the trajectories of their hip, knee, and ankle joints of both legs. These trajectories were used as input to the simulation model in order to obtain the torque characteristics and correspond to the continuous motion of standing up, sitting down and standing up (Fig. 6). Note that Fig. 5 and Fig. 6 demonstrate also the ankle and hip trajectories to allow full comparison of the simulation results. However, for the actuator design
only the knee trajectories have been considered.

In Fig. 5 and Fig. 6 it is possible to notice that the task of standing up is performed approximately over the same duration both in parametric and Vicon trajectories. Thus, it is reasonable to compare the two knee torque characteristics obtained with different joint input trajectories fact which results that the maximum torque, which the knee joint requires, is around 100 Nm (omitting the high peaks in Fig. 6). This torque value, which can also be confirmed by biomechanics data in the literature [16], [18], [20], was used as torque specification for the knee actuator design. The spikes observed in Fig. 6, which were amplified by the numerical differentiation used for obtaining the motion acceleration, are due to the noise of the Vicon system’s tracking. In addition, analysis of the torque-velocity requirements using these simulation results was conducted in order to determine the suitable selection of the pair DC motor-harmonic drive but this analysis is not presented due to the paper length constraint.

**B. Selection of Actuator Design Variables and Design Analysis**

In this subsection, the selection of the design variables of the actuator and its performance analysis are described with particular focus on the application of a knee exoskeleton device. Both are based on simulation of the stiffness model introduced in subsection II-C. The fundamental design parameters which affect the behavior of the actuator in terms of stiffness, elastic torque and stored potential energy are the spring rate $k_s$, the length of the lever arm $L = l_1 + l_2$ and the maximum displacement of the springs $x_{fmax}$. As it is described in subsection III-A the torque specification for the actuator is 100 Nm. At this point authors are not aware of any study regarding the stiffness range of human knee joint during walking or sit-stand-sit movement. In [21] there is an attempt to specify the characteristic stiffness of the knee in flexion and extension modes during gait cycle. According to that, we adopt a suitable range of stiffness from 300 Nm/rad to 600 Nm/rad. The passive deflection of the elastic mechanism $\theta_s$ is given by

$$\theta_s = \frac{\tau E}{K_S}$$

and thus the required angular deflection for the actuator is shown in Fig. 8(a). However, one feature of the functional principle of the actuator is that the maximum angular deflection $\theta_s$ is constrained by the maximum allowable...
displacement $x_{max}$ of the pre-compressed springs and varies as function of the pivot position $l_1$. Fig. 8(a) depicts the achievable deflection $\theta_s$ for different values of maximum spring displacement $x_{max}$. It is obvious that by increasing the maximum displacement of the springs the angular deflection increases as well. However, the free length of the springs is increasing which leads to size increment and also for a given spring diameter the constant of the spring $k_s$ is decreasing. This will have implications to the size of the actuator assembly. Thus, a proper trade off should be made according to the application of the actuator in order to determine the design parameters of the springs. In Fig. 8(b) it is shown the utilizable range of $l_1$ for different values of the lever arm length $L$. By selecting greater lever length the distance that the pivot needs to cover for stiffness range 300 Nm/rad - 600 Nm/rad increases as well which improves accuracy but on the other hand increases the dimensions of the actuator. In Table I the selected design variables for the realization of the first prototype of CompAct-ARS to be used in the knee exoskeleton device are presented.

<table>
<thead>
<tr>
<th>Description</th>
<th>Symbol</th>
<th>Value</th>
<th>Unit</th>
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<tbody>
<tr>
<td>Lever Arm Length</td>
<td>$L$</td>
<td>52</td>
<td>mm</td>
</tr>
<tr>
<td>Radius</td>
<td>$R$</td>
<td>26</td>
<td>mm</td>
</tr>
<tr>
<td>Utilizable Range of $l_1$</td>
<td>$l$</td>
<td>3.6</td>
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<tr>
<td>Spring rate</td>
<td>$k_s$</td>
<td>55</td>
<td>N/mm</td>
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<tr>
<td>Spring Diameter</td>
<td>$d_s$</td>
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<td>mm</td>
</tr>
<tr>
<td>Spring free length</td>
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<td>60</td>
<td>mm</td>
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<tr>
<td>Allowable Spring Displacement (Max)</td>
<td>$x_{max}$</td>
<td>10</td>
<td>mm</td>
</tr>
</tbody>
</table>

The stiffness $K_s$ of the elastic mechanism, as formulated in (8), is shown in Fig. 8(c) with respect to the variation of the pivot position $l_1$ for different values of spring’s rate $k_s$ while the desired stiffness range is marked. It is also worthy to mention that the mechanically allowable stiffness range at the elastic transmission of the prototype is from 200 Nm/rad to 800 Nm/rad (see Table II) and is limited due to the size of the actuator. Finally, in Fig. 8(e) is evaluated the performance of CompAct-ARS in terms of elastic torque as formulated in (6). It is possible to see that the maximum elastic torque $\tau_E$ which the elastic module can provide to the output link is 80 Nm (for spring’s rate $k_s = 55kN/m$). This torque value is close to the maximum torque that knee joint need for standing up and greater than the required torque during gait cycle. Finally, Table II depicts the specifications of the actuator.

**IV. ASSISTIVE CONTROL OF KNEE EXOSKELETON**

In this section the assistive control is introduced. Fig. 9 presents a schematic of a simplified human body model attached at the level of the knee with the compliant actuator CompAct-ARS. In comparison to Fig. 4 and to simplify the presentation of the assistive control concept the masses of the thigh and the torso have been combined into a single body of inertia $I_t$ and mass $m_t$. According to this representation the motion of the overall system (Human + Knee Exoskeleton) can be described by the following formula:

$$I \ddot{\Theta} + D \dot{\Theta} + K \Theta + G = \tau$$

(10)

where $I = diag(I_n, I_e)$ ∈ $R^{2 \times 2}$ with $I_n$ being the combined inertia of the human body, and $I_e$ is the inertia of the knee exoskeleton motor reflected after the reduction drive. In addition $D \in R^{2 \times 2}$ is the damping matrix, $K \in R^{2 \times 2}$ is a symmetric stiffness matrix, $G \in R^{2 \times 1}$ is the gravity matrix and $\tau \in R^{2 \times 1}$ represents the torque vector. In matrix form
Fig. 9. Schematic of simplified human body model attached with the knee exoskeleton.

(10) can be written as follows:

$$\begin{bmatrix} I_a & 0 \\ 0 & I_e \end{bmatrix} \begin{bmatrix} \ddot{q} \\ \dot{\theta}_e \end{bmatrix} + \begin{bmatrix} D_S & -D_S \\ -D_S & D_e + D_S \end{bmatrix} \begin{bmatrix} \dot{q} \\ \dot{\theta}_e \end{bmatrix} = \begin{bmatrix} \tau_h - mgpl_a \sin q \\ \tau_e \end{bmatrix}$$

where $\theta_e = \frac{\theta_m}{N}$, $\dot{\theta}_e = \frac{\dot{\theta}_m}{N}$ and $\tau_e = N\tau_m$ are the position, velocity and torque of the exoskeleton motor respectively reflected at the link side after the gear reduction ($N = 100 : 1$). In addition, $I_e = N^2 I_m$ is the reflected after the reduction drive motor inertia and $D_e = N^2 (D_m + \frac{K_S K_{bemf}}{R_m})$ is the damping of the exoskeleton motor reflected also after the reduction drive. $K_S$ and $K_{bemf}$ are the torque sensitivity and back EMF constant, $R_m$ is the stator resistance and $D_m$ is the physical damping of the motor. Finally, $q$, $\dot{q}$ are the position and velocity of the knee joint and $\tau_h$ is the torque applied by the human. Finally, $D_S$ represents the physical damping in parallel to the actuator spring $K_S$ and $I_a$ is the distance from the knee center to the center of mass of the combined human body. To realize the assistive during the sitting to standing pose transition we considered a scheme where the reference torque for the assistive exoskeleton $\tau_e$ is derived by the combination of two main components. The first component $\tau_{eq}$ is a model based component which is responsible to partially compensate the gravitational torques while the second term is an assistive term generated based on the user’s intended motion. We can write:

$$\tau_e = \tau_{eq} + \tau_u = \beta \tau_{eq} + \tau_u$$

where $\tau_{eq} = mgpl_a \sin q$ is the gravitational torque, $\beta$ is the gravity compensation scaling coefficient and $\tau_u$ is the assistive component. A block diagram of the control scheme is presented in Fig. 10. To generate the assistive torques we consider the following control law:

$$\tau_u = K_a(q_e - q) - D_a \dot{q}$$

The torque delivered to the human knee from the exoskeleton is equal to:

$$\tau_e = K_S(\theta_e - q)$$

where $q_e$ represents the equilibrium of the assistive spring damper network ($K_a, D_a$). To generate assistive forces towards the direction of motion here we select to update the equilibrium position $q_e$ in accordance to the user’s intended motion. The user’s intention in the real system can be obtained by monitoring the user applied torque. This can be done by using model based techniques as well as monitoring the interaction forces between the exoskeleton and user thigh or through the measurement of electromyographic (EMG) signals. Based on these an estimated user torque $\phi(\tau_h)$ which is a function of the real torque applied by the human $\tau_h$ can be obtained. Having derived the estimated user torque the following formula was used to derive the new desired position vector $q_e$ using the measured force signal.

$$\delta q_e = \begin{cases} 
\frac{k_f(\phi(\tau_h) - a)}{\tau_h - q_h} & \tau_h > a \\
0 & -a < \tau_h < a \\
\frac{k_f(\phi(\tau_h) + a)}{\tau_h + a} & \tau_h < a
\end{cases}$$

where $k_f$ and $a$ are the sensitivity constant and the noise dead band constant respectively. By injecting the desired equilibrium position $q_e$ derived from (15), into (13) assistive forces augmenting the user desired actions/motions can be generated that are governed by the stiffness $K_a$ of the spring damper network. Combining (13) and (12) the overall control law can be formulated for the sitting to standing transition task:

$$\tau_e = \beta \tau_{eq} + K_a(q_e - q) - D_a \dot{q}$$

Fig. 10. Block diagram of the system and control scheme.
shape user torque $\phi(\tau_h) = 10Nm$ applied for a period of 0.6 s and a sensitivity coefficient in (14), $k_f = 0.25$. The equilibrium position $q_e$ is derived by (14) and (15) and the desired position of the exoskeleton $\theta_e$ is estimated by (19). The trend of the knee angle $q$ towards the equilibrium position $q_e$ is depicted in Fig. 11(a). Additionally, in Fig. 11(b) are shown the triggering user torque $\phi(\tau_h)$ (in this simulated verification of the assistive control scheme a step type user torque input was considered) and the attracting torque $\tau_h$ generated by the assistive spring damper network ($K_a = 200N/m/rad$, $D_a = 600N\text{ms}ec/rad$). Finally, the gravity torque $\tau_g$ and the total reference force required by the exoskeleton are displayed considering the gravity compensation scaling coefficient $\beta = 1$, see (18). The human parameters used in the simulation study were $m_u = 30kg$ and $I_u = 0.4m$.

![Fig. 11. a) Trajectories of the equilibrium position of the assistive network, knee and exoskeleton. b) Torques generated by the device in response to the user applied torque.](image)

V. CONCLUSIONS AND FUTURE WORKS

In this work a new compliant actuator (CompAct-ARS) was developed and its application to a knee exoskeleton assistive device was presented. Fundamental starting point was to derive the mechanical requirements of the actuator from simulation and experimental data of the standing up and sitting down. Beneficial feature of the actuator is its ability to regulate off-line the level of stiffness in order to achieve suitable passive torque-to-angular displacement profiles for different users and/or different motion tasks. Finally, the knee exoskeleton associated with assistive control strategy will provide motion assistance to the knee of the user.

The principle of operation and design of the actuator were presented followed by the introduction of the assistive control scheme. Results from simulations demonstrated the functionality of the assistive control. Experimental trials will be carried out shortly using the assistive scheme presented. In particular, the work will focus on the experimental evaluation of the knee exoskeleton during sit-to-stand movement cycle for different stiffness level. Its performance in terms of power, torque and stored energy will be evaluated through trials.

REFERENCES