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Development of a Soft-Inflatable Exosuit for Knee Rehabilitation

Saivimal Sridar, *Student Member, IEEE,* Pham H. Nguyen, *Student Member, IEEE,* Mengjia Zhu, *Student Member, IEEE,* Quoc P. Lam, and Panagiotis Polygerinos*, *Member, IEEE*

***Abstract*— This paper presents the design, development and preliminary evaluation of a soft-inflatable exosuit for knee rehabilitation. Soft-inflatable actuators made of heat-sealable thermoplastic polyurethane (TPU) materials are fabricated in two different beam-like structures (I and O cross-section actuators) and mechanically characterized for their torque performance in knee-extension assistance. The fabrication procedure of both types of actuators is presented as well as their integration into a lightweight, low-cost and body-conforming interface. To detect the activation duration of the device during the gait cycle, a soft-silicone insole with embedded force-sensitive resistors (FSRs) is used. In evaluation studies, the soft-inflatable exosuit device is tested for its ability to reduce muscle activity during the swing phase of the knee. Using sEMG (surface electromyography) sensors, the rectus femoris muscle group of a healthy individual is recorded while walking on a treadmill at a constant speed, with and without the soft device. Preliminary testing presents a promising 7% reduction in muscle activity and demonstrates the applicability of the soft-inflatable exosuit in knee rehabilitation scenarios.**

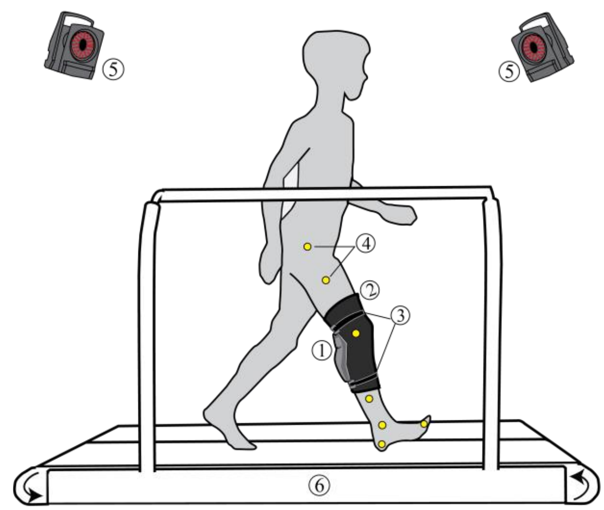


Figure 1: Illustration of the concept of the rehabilitative test setup done with the soft-inflatable knee exosuit. Key design elements include: (1) Two pockets of inflatable actuators. (2) Elastic knee-sleeve to attach exosuit to user's legs. (3) Straps to tighten and loosen the device. (4) Passive reflective markers represented by the dots at the hip, thigh, knee, shin, ankle, heel, and toe. (5) Motion capture cameras. (6) Treadmill to run tests.

I. Introduction

Stroke has become one of the leading causes of disability in the world that results in paralysis, loss of motor function, and muscle atrophy; usually requiring physical therapy and rehabilitation [1], [2]. Stroke can affect parts of the upper or lower limbs, or both, depending on the severity of the incident. With the rapid increase in the number of stroke patients, the demand for well-trained physical therapists and new methods for more effective physical rehabilitation has exponentially increased [3].

In particular, reports indicate that stroke patients with affected lower limbs can experience weakness and loss of control in their quadriceps muscles [4], [5]. In healthy individuals, quadriceps act as dampers which stabilize the knee joint and also produce extension motion of the leg during walking. However, in the case of individuals with paresis after a stroke, generation of the required knee-extension moment, so as to perform common mobility tasks, is partially lost. Therefore, to compensate for quadriceps weakness, patients resort to the adoption of new types of gait patterns for walking that deviate from the standard and can cause serious health and social repercussions [6]. Potential solutions in aiding therapy of stroke afflicted patients is the use of robotic rehabilitation devices. Many attempts for physical rehabilitation of paretic limbs have been made in the form of rigid exoskeletons and assistive devices [7]–[11]. The primary concern about these devices is that they are heavy, bulky, and not portable. The weight of these devices coupled with weakness in patients may lead to discomfort and increased dependency on supervised physical therapy.

Similarly, in the case of walking assistance during rehabilitation, bulky devices may inhibit human motion causing an unnatural gait, which is detrimental for recovery. The issue of joint misalignment found in most rigid exoskeletons can cause further complications. Therefore, there is a need for lightweight devices that offer a higher degree of compliance to the user and can successfully assist in rehabilitation scenarios [12]–[14]. The use of soft orthotic devices aiding the rehabilitation of disabled limbs could minimize or eliminate these issues. There are many advantages to soft robotics, such as compliance, high power-to-weight ratio, and low fabrication costs. Typically, fluidic actuators [15]–[17], or cable driven mechanisms are used to provide joint torque support and minimize musculoskeletal system loading [18]–[20]. Despite the recent advancements of inflatable actuators [21]–[23], enough literature is not published on the modeling and development of inflatable actuators for rehabilitative applications.

In this paper, we present a modular, light-weight, low-cost, soft knee exosuit that utilizes a new design of inflatable actuators to provide assistance to knee-extension motion during rehabilitation. Figure 1 shows an illustration of the

*Corresponding author.

Saivimal Sridar, Pham H. Nguyen, Quoc P. Lam and Panagiotis Polygerinos are with The Polytechnic School, Ira A. Fulton Schools of Engineering, Arizona State University, Mesa, AZ 85212, USA (email: ssridar@asu.edsu; nhpham2@asu.edu; qlam1@asu.edu; polygerinos@asu.edu).

Mengjia Zhu is with the School for Engineering of Matter, Transport and Energy, Ira A. Fulton Schools of Engineering, Arizona State University, Tempe, AZ 85281, USA (mzhu48@asu.edu).

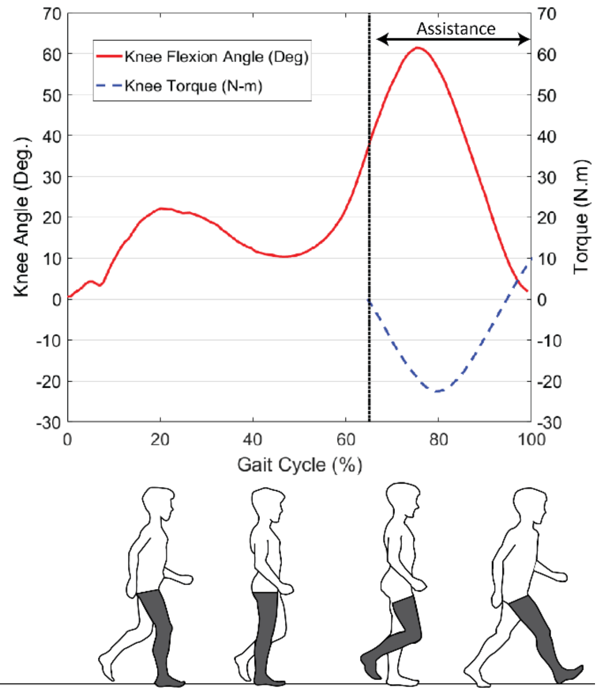


Figure 2: Knee flexion angle and torques during the swing phase of walking for shaded leg at a speed of $1\,m/s$.

proposed concept where a user wears the soft-inflatable exosuit to perform rehabilitation exercises on a treadmill, while motion capture cameras and force sensors monitor knee-extension angles and ground reaction forces respectively. The remainder of the paper is organized as follows. Section II, briefly discusses the biomechanical modeling of the knee joint during walking, and covers the design and characterization of the novel inflatable actuators. The integration of these inflatable actuators into an exosuit interface is discussed in Section III. Finally, Section IV, presents preliminary evaluation and performance results of the soft exosuit prototype.

II. Soft-Inflatable Actuator Design & Characterization

A. Biomechanics of Walking

To set the functional requirements for the soft knee exosuit and the soft-inflatable actuators, the behavior of the knee joint and the muscles associated with the extension and flexion of the knee were investigated in existing literature. Additionally, an experimental study of the knee angles and torque generated during walking, was performed on an instrumented treadmill (side-by-side belt instrumented treadmill, Bertec Inc., Columbus, OH.) with ground reaction force recording capabilities. A motion capture system (T40s, VICON Inc., Los Angeles, CA.) with ten high-speed infrared cameras were also utilized to create a complete kinematic model of the lower body. Passive reflective markers were placed on the ankle, hip and knee joints along with a set of markers on the thigh, shin, foot and toe of a healthy test participant as illustrated in Fig. 1.

The knee joint angles at all points and torques during the swing phase of walking were computed using inverse kinematics and dynamics following the methods described in [24]. From the computed knee angles and the ground-reaction forces obtained from the treadmill, the stance and swing phases of the gait cycle were established. The swing phase of walking was identified as ideal for providing knee joint extension assistance by aiding the quadriceps to complete the swing, as shown in Fig. 2.

Also, it should be noted that the knee joint in itself does not produce the torque, as in the case of motor driven exoskeletons [25]. The forces generated in the joint are due to the action of the muscles contracting and relaxing, during motion. In a walking cycle, the peak torque generated during the swing phase of the leg was determined to be $22\,N.m$ at a walking speed of $3\,m/s$, and was also verified in other studies conducted by [24]. It should be noted that our exosuit needs to aid the user during rehabilitation and not completely assist the knee joint to perform rehabilitative activities. Therefore, having rehabilitation of the quadriceps and the limitations of soft robotics in mind, a partial assistance of 20% ($4.4\,N.m$) was set as the required torque to be provided to the knee joint during the swing phase.

B. Theoretical Modeling of Soft-Inflatable Actuators

Inspiration was drawn from the Euler-Bernoulli beam theory to design the soft-inflatable actuators [26] that power the exosuit. Based on beam deflection theory we have:

$$\frac{d^2}{dx^2}\left(EI\frac{d^2\omega}{dx^2}\right) = q \tag{1}$$

where, E is the elastic modulus and I is the second moment of area of the cross-section of the beam, the curve $\omega(x)$ describes the deflection of a beam in a direction perpendicular to the axis parallel to the length of the structure at position x, and q is the force per unit length. As per (1), for a constant deflection, the elastic modulus E, and the moment of inertia I are directly proportional to the force causing the deflection. Therefore, for structures made from the same material, the deflecting force would depend on the moment of inertia, I.

Two types of structures - with I and O cross-sections were investigated for comparison between the bending forces that can be resisted by the structures at the same internal pressure. It was hypothesized that the soft-inflatable design with I cross-section would be able to withstand higher forces

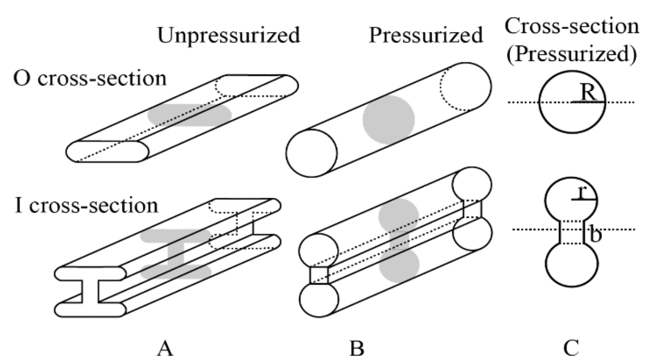


Figure 3. The structural differences between O and I cross-section. (A): Isometric view of the actuators when it is unpressurized. (B): Isometric view of the beam when it is pressurized. (C): Inflated cross-section of the two types of actuators.

before buckling. Figure 3A, shows the cross-section of the inflatable actuators in their uninflated state. However, the cross sectional profile in soft structures change when they are pressurized pneumatically (Fig. 3B). It should be noted that when unpressurized, the actuators cannot resist bending forces; but when pressurized, the cross-section and stiffness drastically change. Figure 3C shows the inflated cross-sections for both the actuators where the O cross-section is approximated to be circular while the I cross-section is a combination of two circles and a square.

To compare the bending load of the two actuators, the total cross-sectional area and the length of each actuator was designed to be the same. Equation (2) shows the relation between the cross-sectional areas of the two actuators where R is the radius of the inflated O cross-section, r is the radius of the circles of the inflated I cross-section, and b is the side of the square (Fig. 3C).

$$\pi R^2 = 2\pi r^2 + b^2 \tag{2}$$

$$R = \sqrt{\frac{(2\pi r^2+b^2)}{\pi}} \tag{3}$$

The radius of the O cross-section, R was written in terms of the parameters of the I cross-section, r and b to set a common ground between the two different cross-sections as shown in (3). The moment of inertia, for both the cross-sections were computed using the parallel axis theorem and are as follows:

$$I_O = \frac{\pi}{4}R^4 = \frac{\pi}{4}\left[\frac{2\pi r^2+b^2}{\pi}\right]^2 \tag{4}$$

$$I_I = \frac{5\pi}{2}r^4 + 2\pi b r^3 + \frac{\pi b^2}{2}r^2 + \frac{b^4}{12} \tag{5}$$

where, I_O and I_I are the moment of inertia for the O cross-section and I cross-section, respectively. When compared, I_I was found to be greater than I_O for all possible values of R, r, and, b as illustrated in Fig. 4.

C. Fabrication of Soft-Inflatable Actuators

Figure 5A shows the prototypes of the inflatable actuators with the O and I cross-sections. For the O cross-section, two pieces of thermoplastic polyurethane (TPU) layers are heat sealed together on each of the four sides. The fabrication of the I cross-section requires additional steps where the material has to be folded and sealed multiple times to create an I cross-section and the center seams need to be heat-sealed to keep the structure intact when inflated, as seen in Fig. 5B. Also, provision for uniform distribution of pressurized air is provided. Both the ends of the actuator are heat-sealed where tubing fittings are attached 2.5cm from the sealed ends.

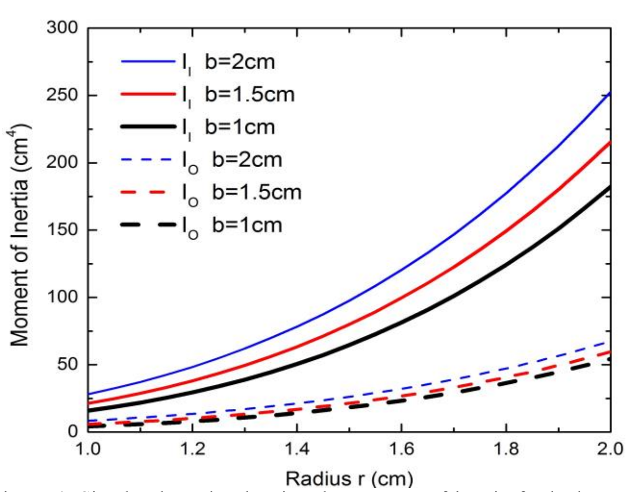


Figure 4. Simulated results showing the moment of inertia for both cross sections using effective radius of the I and O cross-sections computed for possible values of r, b, and R, where R is related to r and b as shown in (3).

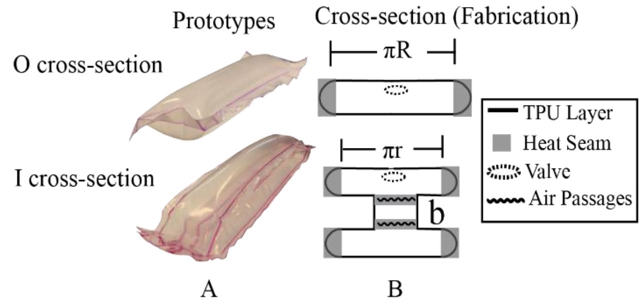


Figure 5. The prototypes and fabrication of O and I cross-section of the inflatable actuators. (A): Corresponding prototypes of each type of actuator using the DT 2001 TPU material. (B): Cross-section of both actuators and the fabrication process for each type of actuator.

The chosen dimensions for the actuators should be based on the proportions of the femur and tibia for accurate fitting, and the number of required actuators. For the following tests, values of R = 1.94cm, r = 1.27cm, b = 1.27cm, and 25.4cm actuator length are used in the construction of the inflatable actuator. The corresponding second moment of inertia, I_O and I_I , are 2.94 cm^4 and 36.99 cm^4, respectively. It is noted that different values of R, r, and b can be used for modeling, as long as they satisfy the relationship in (2).

Inflatable actuators fabricated from two types of TPU materials are tested – DT2001 with thickness of 0.1524 mm (American Polyfilm Inc., Branford, CT.) and Stretchlon 200 with thickness of 0.0381 mm (Fibre Glast Dev. Corps, Brookville, OH.).

D. Evaluation of Soft-Inflatable Actuators

With physical rehabilitation assistance in mind, the soft-inflatable exosuit is conceptualized to assist the user by providing knee-extension moment through attaching the inflatable actuators behind the user's knee (the popliteal fossa or kneepit), as seen in Fig. 6. The hypothesis is that by

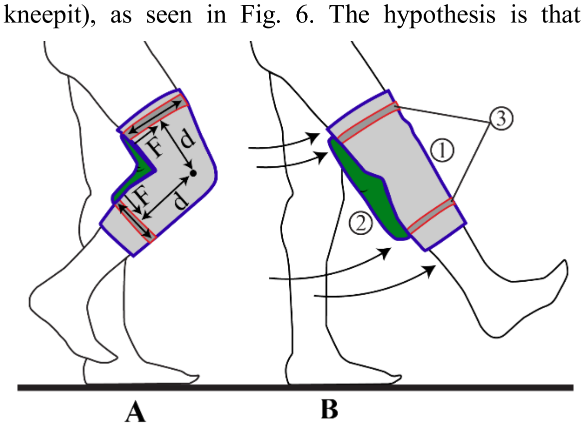


Figure 6: Illustration of force and torque relationship in the exosuit. (1) The knee-sleeve. (2) The inflatable actuators. (3) The hook and loop straps. (A): User in mid-swing gait phase and start of actuator inflation. (B): User at the end of the swing gait phase and completely inflated actuators.

strapping the actuators at a fixed length (d) along the calves and the hamstring, they could provide the forces (F) required to assist with the knee-extension motion. Thus, helping the user from the mid-swing to the end of the swing phase of their gait cycle.

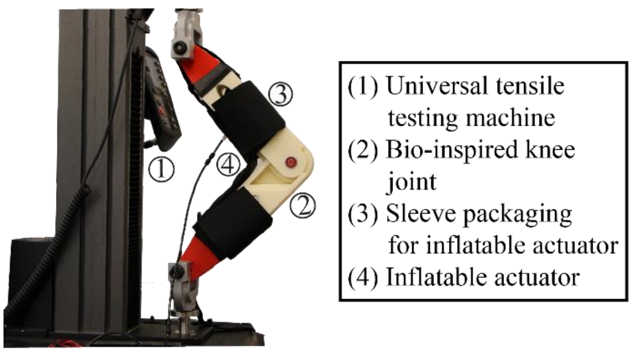
A biologically-inspired joint was first fabricated to assist with simulating the biological knee joint. This knee joint was assembled using two pieces made of acrylonitrile butadiene styrene (ABS) plastic (Fortus 450mc, Stratasys, Eden Prairie, MN) to mimic the femur and tibia of a human leg. The rotating joint of the test apparatus is designed to allow 135° of rotation, with ball bearings at the center of the joint for frictionless motion.

The knee joint was designed with provisions to attach the inflatable actuators in different orientations for more exhaustive testing. To securely mount the actuator to the device, a fabric sleeve for the actuator was fitted with straps to secure the actuator at equal distances from the kneepit.

To measure the force output from the inflatable actuators, the knee joint was mounted securely on a universal tensile testing machine equipped with a load cell (Instron 5944, Instron Corp., High Wycombe, United Kingdom) to capture the force data, as shown in Fig. 7A. The knee-extension angle on the test apparatus was set at 60° as per the maximum knee-flexion angle during the swing phase of walking. For the tests, multiple force output readings were collected as the actuators were inflated at intervals of 3.45 kPa.

From Fig. 7B, it is observed that the DT2001 material provides better force output with increasing pressures and the actuator with an I cross-section generates, as hypothesized, higher forces than the one with an O cross-section for the same material, at the same pressure. It is also observed that the Stretchlon 200 material generates significant amounts of creep with increasing pressure, hence making the material unsuitable for this application. Therefore, the inflatable actuators with an I cross-section fabricated with DT2001 are selected to be implemented in the exosuit.

The torque exerted by the inflatable actuator about the knee is computed by resolving the obtained forces from the universal testing machine, perpendicular to the surface of the test apparatus and multiplying it with the moment arm of the force. It is determined that to produce a torque of 4.4$N.m$, two actuators supplied with 27.57kPa would be required.



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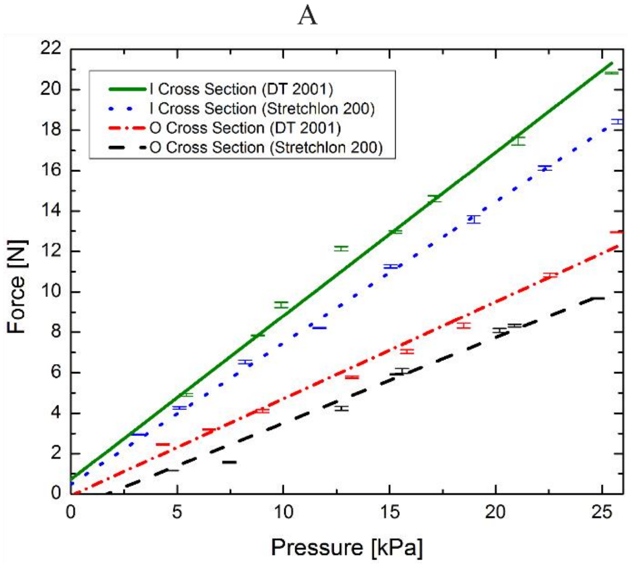

B

Figure 7. (A) Testing equipment and setup for the characterization of the soft-inflatable actuator. (B) Force (N) versus Pressure (kPa) graph of both I and O cross-sections made of two different types of materials. The first type is DT2001 and the second Stretchlon 200. It is noted that the I cross-section actuators show higher performances than the O cross-section actuators and that the DT2001 material demonstrates better performances than the Stretchlon 200 material.

III. Design of the Soft-Inflatable Exosuit

The knee-exosuit proposed in this work, Fig. 8, is designed to be light-weight (weighs 160 g with the onboard electronics) and body-conforming to the user's knee. The material used for its manufacture is an elastic fiber (neoprene) that allows for maximum body conformity when flexing the knee. Along the neoprene knee sleeve, an inelastic fabric pocket is sewn and aligned at a fixed length from the kneepit, to house separately, two inflatable actuators. These pockets allow for rapid switching of the actuators in cases of failure due to permanent deformation or rupture. The exosuit is additionally secured to the wearer's leg with hook and loop straps allowing the wearer to adjust the tightness around the thigh and calves to eliminate unwanted slippage of the

Figure 8: The prototype of the soft-inflatable exosuit worn by user around the knee joint. Top-left: Two inflatable actuators embedded inside a knee-sleeve pouch are used to assist with joint extension. Top-right: Electronics pouch attached onto knee-sleeve. Bottom-Right: Two FSR sensors embedded in soft-silicone insoles, placed inside the user's shoes.

exosuit while walking.

Two force-sensitive resistors (FSRs) are placed and casted into a thin (4.5 mm), soft-silicone insole (Ecoflex 30, Smooth-On Inc., Macungie, PA) that is inserted inside the wearer's shoe. One sensor is located at the ball of the foot to measure the toe-off forces while the other is at the heel to measure heel-strike forces.

An additional small fabric pocket is sewn on the knee sleeve to incorporate all the required monitoring electronics. The electronics include a microcontroller with links to a custom board that facilitates connections to the insole sensors, valves controller, and fluidic pressure sensor.

The electro-pneumatics of the systems include three pneumatic valves (MHE3-MS1H valves, Festo, Hauppauge, NY) that are placed in series to control venting of air pressure during pressurization or depressurization. A single fluidic pressure sensor (ASDXAVX100PGAA5, Honeywell International Inc., Morris Plains, NJ) is added to the system to monitor the internal pressure of the inflatable actuators. The actuators are pressurized using a pneumatic line that is connected to a pneumatic supply source, as well as a vacuum pump (DV-85N-250 pump, JB Industries, Aurora, IL), which facilitates faster depressurization rates (at 0.00142 m^3/s).

The soft-inflatable exosuit is designed to provide assistance to the wearer's knee joint starting from the mid-swing (beginning of the inflation of the actuators) to the end of the swing phase of the gait cycle. At toe-off, a binary controller actuates (inflates) the exosuit aiding in the extension motion during swing phase. Reversely, the exosuit would deflate at heel strike, allowing the user to perform an unrestricted knee flexion motion.

IV. Evaluation of the Soft-Inflatable Exosuit

To validate the effectiveness of the inflatable exosuit to assist with rehabilitation exercises, surface electromyographic (sEMG) sensors (Delsys® Trigno®, Delsys, Natick, MA) are placed at the quadriceps of a healthy participant, after the skin was treated with rubbing alcohol solution. The sensors are placed at the rectus femoris, vastus lateralis, and vastus medialis, of the quadriceps to record muscle activation with and without the exosuit. The vastus intermedius muscle of the quadriceps is not considered in the testing as it lies closer to the skeletal structure and its operation cannot be captured using sEMG sensors.

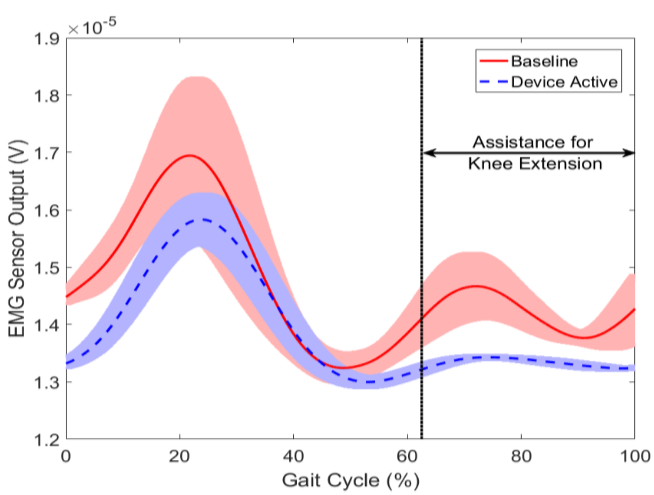

Figure 9: Processed sEMG data for the rectus femoris muscle for human walking at 0.5 m/s averaged for five gait cycles without exosuit assistance (Baseline) and with exosuit assistance (Device Active).

A test protocol where the participant would walk on an instrumented treadmill for three minutes per trial, at a set speed of 0.5m/s, was applied. Post the completion of a trial, the participant is allowed to rest for five minutes to recover from any fatigue that might have occurred in the muscles. Safety measures such as emergency stops and quick deflation of the exosuit are also incorporated in case of any discomfort caused to the test subject.

A total of six trials are performed on a single participant – three with the exosuit (Device active) and three without the exosuit (Baseline) following the aforementioned study protocol. The sEMG data collected during trials are processed to compare the effect of the exosuit on the gait cycle. Five gait cycles are averaged for both the Baseline and the Device active test for the rectus femoris muscle group and plotted along with their standard deviation as seen in Fig. 9.

Processing of the sEMG data shows a reduction in muscle activity of 7% in the rectus femoris muscle group during testing with the exosuit active, while the other muscle groups show reductions with lower percentages. The three sets of data pertaining to one single participant demonstrate consistent results with all sets of data having similar amounts of reductions. A promising reduction in the muscle activity of the rectus femoris muscle group was observed but further investigation into the other muscle groups of the quadriceps need to be performed for more conclusive results.

V. Conclusions and Future Work

In this paper, we developed and evaluated a soft-knee exosuit powered by inflatable actuators to provide knee-extension moment, for the rehabilitation tasks, such as controlled walking. We introduced the designs of new inflatable actuators with two different cross-sections and types of materials, which were inspired by structural mechanics and beam theories. These actuators were tested and analyzed for their force output on our experimental test setup. It was determined that the inflatable actuator with an I cross-section produced a larger force output at same pressures when compared to an actuator with O cross-section of equivalent area and length.

Utilizing the selected I cross-section, we integrated two inflatable actuators, into a light-weight, body-conforming, exosuit that is worn around the user's knee to provide an estimated 20% of torque assistance. The soft-inflatable exosuit included onboard electronics and off-board electro-pneumatics. To set the activation phase of the inflated actuators, which was during the mid-swing to the end of the swing gait phase of the wearer, a custom insole-embedded force sensor was placed in the shoe.

To evaluate the exosuit and demonstrate its applicability in rehabilitation scenarios by reducing muscle effort to the wearer, we allowed a healthy individual to walk on an instrumented treadmill for three minutes at a set speed of 0.5m/s, while collecting sEMG signals for the rectus femoris, vastus medialis, and vastus lateralis muscle groups on the quadriceps. From the results, a promising reduction in muscle activity of 7% in the rectus femoris muscle was recorded. A reduction in muscle effort during the stance phase was also observed while the exosuit was active. This was due to a latency in deflation of the exosuit which allowed for some muscle support to be offered during the leg impact with the ground. The effect of the exosuit on the other muscle groups need to be analyzed extensively to determine the overall effect of the assistance provided. Also, a study walking with the exosuit, inactive and active needs to be performed in order for a more comprehensive biomechanics analysis to be provided. Effects of the inaccuracies in force transfer due to the uneven surface area in the kneepit will also be analyzed. Additionally, tests with an increased number of participants will provide more comprehensive results.

In the future, we look to further understand the design of the inflatable actuators and their performance variations (by varying the type of materials, beam shapes, and sizes) through computational modeling. We plan to implement force and pressure control along with more advanced user intent detection to allow for a versatile exosuit. Also, by improving the rate of inflation and deflation of the inflatable actuators, precise and timely assistance could be achieved.

We also noticed slow signal drifts in FSR sensors, which did not affect our detection of required gait phases (beginning and end of swing phase) because of the placement of the sensors. However, if more gait events are required to be detected for device activation, future calibration and hysteresis compensation of the sensors might be able to improve sensors' inaccuracies.

Future efforts will also be made to improve the functionality of the exosuit for a wider variety of tasks such as sit to stand, stair ascent, seated rehabilitative exercises, walking up or down a ramp. Steps to manufacture a new prototype with onboard pneumatic system and untethered communication to the insole sensors will be explored for better functionality and capabilities of providing assisted living. Finally, trials with impaired participants will be pursued in collaboration with our clinical partners at Barrow Neurological Institute in Phoenix, Arizona.

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