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Modelling and Design of a Synergy-based Actuator for a Tendon-driven Soft Robotic Glove

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Abstract—The need for a means of assistance in human grasping, to compensate for weakness or to augment performance, is well documented. An appealing new way of doing so is through soft, wearable robots that work in parallel with the human muscles. In this paper we present the design and modelling of a tendon-driving unit that empowers a wearable, soft glove. Being portability one of our main objectives, we use only 1 motor to move 8 degrees of freedom of the hand. To achieve this we use an underactuation strategy based on the human hand's first postural synergy, which explains alone $\approx 60\%$ of activities of daily living. The constraints imposed by the underactuation strategy are softened, to allow adaptability during grasping, by placing elastic elements in series with the tendons. A simulation of the dynamic behaviour of the glove on a human hand allows us to quantify the magnitude and distribution of the forces involved during usage. These results are used to guide design choices such as the power of the motor and the stiffness of the springs. The designed tendon-driving unit comprises a DC motor which drives an array of spools dimensioned according to the first postural synergy, an electromechanical clutch to hold the hand in position during static posture and a feeder mechanism to avoid slacking of the tendons around the spool. Finally, the tendon-driving unit is tested to verify that it satisfies motion and force characteristics required to assist its wearer in activities of daily living.

I. INTRODUCTION

The elegant and unmatched abilities of the human hand in grasping, communicating and gesturing make its deterioration of functionality one of the most debilitating experiences that one could live. It is thus not surprising that recovering its functionality has been identified as a high-priority need in all those subjects affected by functional limitations caused by neuromuscular disorders such as tetraplegia [1], spinal cord injury [2] or stroke [3]. Alongside clinical cases, a large number of elderly people experiences age-related muscle weakness that hinders the accomplishment of simple activities of daily living (ADLs) [4]. Lastly, repetitive and forceful tasks, such as those performed in construction, military and logistics have been shown to increase the likelihood of musculoskeletal disorders [5], which may ultimately lead to hand functional limitations.

This impelling need for a means of assistance in human

Fig. 1: Prototype of the soft robotic glove for grasping assistance.

grasping promoted the development of several hand exoskeletons, most of them relying on the use of a rigid, multi-degree of freedom (DOF) linkage structures attached to the hand dorsal side. Whilst being reliable and fully controllable, rigid structures are bulky and heavy and face some still unsolved challenges. Amongst them the most significant is misalignment of the robots' joints with the biological ones: various solutions have been proposed such as redundant DOFs, self-aligning mechanisms, or remote centers of rotation [6], all of them adding complexity and hence undesired weight to the systems.

Alternative actuation methods include flexible pneumatic actuators [7] and tendon-driven systems that mimic the way natural muscles transmit forces to the joints [8]. The latter strategy has been successfully employed in lower-limb assistive exoskeletons: in [9] a bowden-cable transmission allows to locate the actuation unit away from the end-effector and torques are applied to the joints by retracting a set of tendons. The result is a wearable, clothing-like design that works in parallel with the human muscles to reduce the metabolic cost of walking.

This last solution exploits the user's own skeleton as the architecture devoted to transmit forces and torques through the joints, avoiding joint misalignment. Additionally, the

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absence of a rigid mechanical link both helps to keep the overall weight and size of the system low and entails a less invasive cosmetic appearance, making the device a more likely candidate for everyday use.

The device described in [8] is an example of the application of these same principles to the hand. Moreover, it deals with the notorious complexity of the hand by employing a differential mechanism for underactuation.

A. Underactuation and Hand Synergies

Anatomically, the human hand has 15 joints and 20 DOF. This high dimensionality is partly responsible for its unparalleled dexterity but poses unsolved control challenges. As a consequence, underactuation is an inevitable path when emulating or assisting the human hand.

The most common solutions for underactuation include differential drives, mechanical linkage coupling and variable compliance coupling [10] but the results found in [11] have inspired novel methods. Santello *et al.* noticed that a remarkably small number of kinematic coordination patterns are able to explain most of the variance in everyday hand movements. The high variance explained by a small number of these components (also known as postural synergies) suggests that everyday activities can be explained by linearly combining very few hand postures.

The first application of this principle to underactuate a robotic hand is described in [12] where a set of pulleys dimensioned according to the first two synergies actuate a tendon-driven hand. Adding compliance relaxes the hard constraints imposed by the synergy model and yields to a *softly underactuated* model, capable of adapting to the environment during grasping [13].

Keeping in mind that a wearable device for human grasp assistance needs to be lightweight, low-power consuming and cosmetically acceptable, we hereby present the design and characterization of an actuator intended to drive the soft glove illustrated in Figure 1. We exploit the concept of hand postural synergies to use only one motor to actuate 8 DOF (flexion/extension of the thumb, index and middle fingers) and use a bowden cable transmission to locate the motor on a belt or in a backpack. The characteristics listed in Table I, obtained from existing studies of the human hand, are combined with a dynamic model of the human fingers, driven by artificial tendons, to identify the force and motion requirements of the actuator. We then present the first prototype of the actuator and evaluate its bandwidth and maximum applicable force.

II. MODELLING

In order to evaluate the actuator's power requirements and to choose a reasonable value of stiffness for the springs in series with the tendons we used a bi-dimensional model of the fingers on the sagittal plane. The index and middle fingers are modelled as a 3-link kinematic chain, as shown in Figure 2, while the thumb is modelled as a 2-link chain. The metacarpalpalangeal (MCP), peripheral interphalangeal (PIP) and distal interphalangeal (DIP) joints

TABLE I: System requirements

Characteristics	Requirements	
Force/Motion: [14], [15], [16]		
	Fingers	Thumb
Range of motion (per joint) [°]	90	80
DOF	MCP ^a -PIP ^b -DIP ^c	MCP-PIP
Grasping force [N]	10	
Bandwidth [Hz]	1.2-1.6	
Practical considerations: [17]		
Distal frame weight [kg]	≤0.5	
Proximal pack weight [kg]	≤2.5	

^aMetacarpophalangeal

^bProximal interphalangeal

^cDistal interphalangeal

are modeled as revolute joints.

Starting from the Lagrangian form one can derive the dynamics of the fingers in joint space θ :

$$M(\theta)\ddot{\theta} + C(\theta, \dot{\theta})\dot{\theta} + N(\theta) = \tau \quad (1)$$

where $M(\theta) \in \mathbb{R}^{n \times n}$ is the *Inertia* matrix, $C(\theta) \in \mathbb{R}^{n \times n}$ is defined as the *Coriolis* matrix and $N(\theta) \in \mathbb{R}^n$ is a column vector that takes into account gravitational forces.

For the sake of accuracy we include in the model the natural elasticity and damping of human joints. We set joint damping to be a constant and joint stiffness to be a quadratic function of the joint angle, using the numeric values reported in [18]. The vector $\tau \in \mathbb{R}^n$, expressing the external torques on the joints, can thus be considered as the algebraic sum of two components: a passive one deriving from the joint's stiffness and an active one, i.e. the torque applied on the joints by the exoskeleton

$$\tau = \tau_A - \tau_P \quad (2)$$

$$\tau_P = D\dot{\theta} + E(\theta)(\theta - \theta_0) \quad (3)$$

with $\theta_0 \in \mathbb{R}^n$ being a vector of resting joint angles and $D, E(\theta) \in \mathbb{R}^{n \times n}$ diagonal matrices of joint damping and stiffness respectively.

To find an expression for the torque exerted by the exoskeleton on the human joints we define an *extension function* $h(\theta)$ that maps the joint angles to a corresponding displacement

Fig. 2: 3-link planar model of a human finger.

Fig. 3: Schematic drawing and geometrical parameters used to define the routing of the tendons along each joint of the fingers on the sagittal plane.

at the endpoint of tendon j :

$$h_f(\theta) = \sum_{i=1}^n 2\sqrt{a_i^2 + b_i^2} \cos\left(\phi_i + \frac{\theta_i}{2}\right) - h_{10} \quad (4)$$

$$h_e(\theta) = \sum_{i=1}^n R_i \theta_i - h_{20}. \quad (5)$$

Where h_f is the extension function for the flexor and h_e for the extensor. With reference to Figure 3, a_i is half the width of the i -th phalanx, b_i is the distance of the i -th anchor point from the adjacent joint's centre of rotation, $\phi_i = \arctan(a_i/b_i)$ and R_i is the radius of the i -th joint. The constants h_{10} and h_{20} assure that the extension functions are null for $\theta = \theta_0$. Defining $h \in \mathbb{R}^p$ to be a column vector containing the extension functions in a system of p tendons one can find:

$$\tau_A = P(\theta)f \quad (6)$$

$$P(\theta) = \frac{\partial h^T}{\partial \theta}(\theta) \quad (7)$$

where $f \in \mathbb{R}^p$ is the force applied on the tendons by a motor unit and $P(\theta) \in \mathbb{R}^{n \times p}$ a coupling matrix between the tendon forces and the joint torques. If we assume the tendons to have negligible mass, the expansion to the case with elastic elements in series with the tendons is straightforward:

$$\tau_A = P(\theta)K(e - h(\theta)) \quad (8)$$

with $e \in \mathbb{R}^p$ being the displacement of the tendons as governed by the actuator and $K \in \mathbb{R}^{p \times p}$ a diagonal matrix of tendons' stiffness. Note that for $e = 0$ and $\theta = \theta_0$ the tension in the cables is null.

By substituting (3) and (8) in (1) we obtain:

$$M\ddot{\theta} + C\dot{\theta} + N + D\dot{\theta} + E(\theta - \theta_0) + Qh = Qe, \quad (9)$$

where we have removed the dependency on θ for the sake of conciseness and defined a new coupling matrix $Q := PK$ between the tendon displacement and the joint's torques. The term Qh defines the restoring torque of the soft exoskeleton as a result of moving the fingers away from the resting position, i.e. the stiffness of the tendon-network perceived on the human joints. Figure 4b shows a comparison of this term, normalised by θ , for various spring constants,

with the stiffness of biological joints. Subsequently, we add an external force acting on the center of mass of the distal phalanx that models fingertip forces during ADLs. We assume the contact force to be 10N, acting on the direction normal to the main axis of the distal phalanx and we use a point-contact with friction model with a coefficient of static friction of 0.79N/m [19].

The contribution of the resulting external force $F \in \mathbb{R}^2$ can be mapped to the joint space using the geometrical Jacobian $J(\theta) \in \mathbb{R}^{2 \times n}$, (9) thus becomes

$$M\ddot{\theta} + C\dot{\theta} + N + D\dot{\theta} + E(\theta - \theta_0) + Qh = Qe + J^T F. \quad (10)$$

Finally we include a term modelling losses due to friction since they are often not negligible in bowden cable systems. The main parameter affecting the entity of such losses is the bending angle of the outer housing. Assuming that the contact between the inner cable and the outer sheet can be modelled as the sliding of a cable over a fixed cylinder, the force transmission efficiency becomes:

$$f_{in}/f_{out} \approx e^{-\mu\phi}, \quad (11)$$

with μ being the friction coefficient between the cable and the outer sheet, ϕ the total wrap angle in the transmission, between the motor unit and the glove, and f_{in}, f_{out} the tension in the cable before and after the transmission respectively. Assuming the actuator to be mounted on a belt on the waist, a reasonable and abundant estimate of the wrap angle is π . Using a teflon-teflon static friction constant yields to a loss in efficiency of $\approx 20\%$.

A numerical solution to equation (10), implemented in Simulink (The MathWorks, Inc., Natick, Massachusetts, United States), was used to assess the forces in the tendons, the required motor torque and the stiffness of the tendon network perceived on the human joints. We used anthropometric data from [20] to choose the geometrical dimensions of the human hand. The input for the model was, in both cases, a rotation of the motor spool between 0 and $2/3\pi$ for a duration of 2s, following a smooth trajectory. The results, dependent on the tendons' stiffness, are shown in Figure 4. Moving three fingers with a single motor unit according to the first postural synergy simply means coupling the displacement vectors e for the thumb, index and middle fingers with a synergy vector $S \in \mathbb{R}^{3p}$ which we extracted using Principal Component Analysis on the kinematic data in [11] (see Appendix for details). The resulting displacement δe for the three pairs of tendons can be expressed as $\delta e = S\delta\sigma$, where $\sigma \in \mathbb{R}$ is scalar controlling the degree of activation of the synergy. If we use a rotary motor to actuate an array of spools around which the tendons are wrapped, S could be the ratio between the spools' diameters and $\delta\sigma$ a displacement of the motor.

Based on the obtained results, shown in Figure 4, we roughly need the motor to achieve at least 2Nm torque, 3N/mm series elastic elements and tendons able to withstand at east 100N of tension.

Fig. 4: Results of the dynamic model of the human fingers, used to guide the design choices of our tendon-driving unit. The plots display the results obtained by solving the dynamic equation under typical finger loading condition. (a) Resulting tension in the tendons (top) and required motor torque (bottom) for varying tendon stiffness. (b) Perceived stiffness of the tendon-network on the index MCP joint for three values of spring constant. Results are compared with the natural stiffness of the joint $E(\theta)$, as defined in [18]. These results suggest that we need at least 2Nm of torque from the motor, a stiffness of ≈ 3 N/mm on the tendons and tendons that withstand 100N of tension.

Fig. 5: Exploded view of the designed actuation stage. The tendons are wrapped around an array of three pairs of spools to actuate the thumb, index and middle finger in an agonist/antagonist fashion. A feeder mechanism allows to avoid the slacking of the cable around the spool. An electromechanical clutch can be used to lock the system and unload the motor, for power saving during static posture.

III. ACTUATOR DESIGN

The actuator empowering the tendon-based soft glove for grasping assistance is shown in Figure 5.

The motor driving the tendon's spool was chosen to provide the torque and bandwidth required for assisting the hand in ADLs, according to the requirements listed in Table I and obtained by the model. Furthermore, a clutch-based mechanism allows the hand posture to be firmly locked once the object has been grasped, releasing the motor from holding

the position and thus reducing the control effort and the associated power consumption.

The device consists of a DC brushless motor (Maxon EC-max, $\varnothing$ 22mm, 25Watt) equipped with a rotary encoder (Maxon Encoder MR, 512 CPT) and a planetary gearhead with a reduction of 23:1. A further 3:1 reduction stage between the motor and the spool shaft ensures the electromechanical clutch (Inertia Dynamics, SO11, $\tau_{max} = 0.68$ Nm) to withstand the required torque.

Fig. 6: Physical implementation of the first postural synergy of the human hand. The diameters of the spools driving the thumb, index and middle fingers were dimensioned according to the ratios found in the first hand postural synergy S (listed in Table II).

TABLE II: Pulley Dimensions

Finger	Synergy Ratio (S)	Pulley Diameter [cm]
Thumb	0.18	0.95
Index	0.36	1.93
Middle	0.42	2.2

An array of spools, consisting of three pairs of cylinders, drive the tendons routed through the thumb, index and middle fingers in an antagonistic fashion, such that retraction of the agonist causes release of the antagonist. We chose the maximum diameter to be 0.22m and defined the other diameters using the ratios corresponding to the 1st hand postural synergy (details in the Appendix). Numerical values for the three fingers are listed in Table II and shown in Figure 6.

The advantages of using a feeder mechanism are to prevent derailing of the tendons from the spools and eliminate the need of pre-tensioning the tendons [21]. This is achieved by routing each cable between a one-way clutch and an idle roller. The one-way clutches are oriented such that the free direction is the feeding while they are locked when the cables are coiling around the spool; this introduces a direction-dependent friction in the mechanism that prevents slacking of the cables around the spool. A 1mm-thick lining of urethane (Devcon Flexane 60) on the rollers and clutches ensures a higher adhesion.

Teflon-coated steel cables (Sava Industries, $\varnothing$ 0.686 mm) as tendons and Teflon-lined bowden cables were chosen to reduce the overall friction in the transmission and minimise stick-slip phenomena [22]. Finally, we enclosed the mechanism in a 3D-printed casing in ABS plastic, as shown in Figure 8a. A test-bench was used to verify the force and motion characteristics of the actuator.

IV. PERFORMANCE EVALUATION

The setup schematised in Figure 7 was used to verify that the tendon-driving unit can reach the velocities required for ADLs. The test-bench, shown in Figure 8b, consists of a single shaft housing an array of spools whose diameters ratios were dimensioned according to the first postural synergy.

Fig. 7: Model of the actuation stage and of the testbench used to test its bandwidth response. The clutch can be engaged, thereby locking the system in its current configuration. This allows to maintain a hand posture with very low power consumption. Elastic elements in series with the three pairs of tendons guarantee compliance and adaptability in grasping. The testbench used to test the actuator's bandwidth comprises a spool and an encoder to measure the angular displacement of the end-effector.

Each tendon, routed through bowden cables, was wrapped around the corresponding test-bench spool and placed in series with a compression spring ($k = 3.58\text{N/mm}$) on the end-effector; a screw mechanism was used to tune the pre-load of each spring.

The motor was then excited with a linear chirp position signal of the form:

$$s(t) = S_0 \sin(2\pi f(t)) \quad (12)$$

$$f(t) = f_0 + \frac{f_1 - f_0}{T}t, \quad (13)$$

with $f_0 = 1\text{Hz}$, $f_1 = 20\text{Hz}$ and $T = 120\text{s}$ and S_0 chosen so as to span half-ROM of a typical finger motion.

The resulting angular displacement was monitored using an encoder on the test-bench shaft (Omron E6C2, 512 CPT). Data acquisition was performed using a Quanser QPIDe control board at a sampling frequency of 1KHz; the low-level motor control was handled by a Maxon EPOS2 50/5 controller. Figure 8c shows the bode plot of the motor unit, extracted using a least square fitting in the Fourier domain. The system presents a bandwidth of 8 Hz, which translates into a maximum cable retraction speed from a minimum of 0.75m/s (thumb spool) to a maximum of 1.75m/s (middle-finger spool).

The maximum torque that the motor unit is able to apply was measured by attaching one of the tendons to a load cell (Futek LCB500) and measuring its tension upon the application of a smooth 5th order signal between 0 and the maximum rated torque for the DC motor. The resulting, filtered signal is shown in Figure 8d. Although not fully covering the maximum simulated required torque (Figure 4a),

(a)

(c)

(b)

(d)

Fig. 8: Evaluation of the performance of the designed actuator. (a) Actuator enclosed in a 3D printed ABS plastic casing. (b) Testbench used to measure the frequency response of the actuator. The tendons are attached to a set of pulleys on a shaft whose angular position is sensed by an absolute encoder. (c) Magnitude and phase of the Bode plot for the transfer function $\mathcal{H}$, between the end-effector angular position and the DC motor position, acquired using the set-up shown in Figure 7. The actuator has a bandwidth of 8 Hz, satisfying the requirements needed to emulate natural human hands kinematics. (d) Maximum applicable torque test.

we are confident that 2Nm is enough to provide a reasonable assistance in ADLs.

V. DISCUSSION

Loss of grasping force and dexterity massively impacts the quality of one's life, yet very few of the devices designed for grasping assistance fulfil the requirements for at-home, unsupervised use. We presented a model-based design of a tendon-driving unit for a soft, wearable glove.

Giving priority to the need to have a lightweight and low-power consuming actuator, we proposed an underactuated solution based on the statistical knowledge of human hand kinematics in daily life. The addition of series elastic elements was used to allow adaptability in the grasping phase and a safer interaction with the environment. Using a clutchable mechanism, our system can hold a static posture with minimal power consumption.

Most of our design choices were based on established knowledge of human's hand motion and force characteristics. The

proposed dynamic model was not intended to be realistically accurate, but served rather as a reference to guide us in the choice of the motor's characteristics, spring stiffness and the tendons' material.

There are a number of aspects and limitations to the proposed approach that will need to be addressed in future works. The effect of friction in the bowden cables, for example, was not evaluated in this paper but is likely to significantly affect the performance of the device. Most importantly, the use of only one closure pattern, even though being compliant, may obstruct the natural dexterity of human hands movement. Trials on healthy and impaired subjects, with a set-up to measure the distribution of forces during grasping would help to verify this point and assess the effectiveness of the chosen underactuation strategy.

APPENDIX

A. Extracting the First Postural Synergy of the Hand

This section details the procedure used to derive the synergy ratio S , whose values are listed in Table II.

We used a dataset of hand postures available on the Hand-Corpus online repository, consisting of 5 subjects performing 57 static grasp types, recorded by Santello *et al.* using a Cyberglove (details on the experimental set-up can be found in [11]).

The experimental data includes the joint angles of 15 DOF of the human hand; we only used the DOF on sagittal plane, i.e. the metacarpophalangeal (MCP) and inter phalangeal (PIP) joints of the fingers and the thumb. We then sum together joints belonging to the same finger; this was done because each spool drives a tendon that spans all of the finger's DOF on the sagittal plane. Lastly, we run Principal Component Analysis (PCA) on the dataset averaged over subjects to extract the postural synergies.

PCA can be defined as the projection of the data on a low-dimensional space such that the variance of the projected data is maximised. If the data consists of hand joint angles, the basis vectors of this subspace correspond to the hand postural synergies.

Consider, as in our case, the set of hand postures $\{x_n\} \in \mathbb{R}^{D \times N}$, where D is the number of DOF and N the number of postures. We want to project $\{x_n\}$ on a 1-D subspace spanned by the vector S , i.e. the first synergy. Each data point will thus become a scalar value $S^T x_n$ such that the variance of the projected data, defined as

$$\frac{1}{N} \sum_{n=1}^N \{S^T x_n - S^T \bar{x}\}^2 = S^T C S, \quad (14)$$

is maximised. Here $\bar{x}$ is the mean and C is the covariance matrix of the data. It can be shown that solving this optimisation problem using Lagrange multipliers is equivalent to finding the vector S such that:

$$S^T C S = \lambda. \quad (15)$$

This equation shows that finding the first principal component of $\{x_n\}$ corresponds to an eigendecomposition of the covariance matrix C . Its eigenvector with the largest eigenvalue λ is the first postural synergy S .

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