Soft artificial muscle for wearables assistive jumping

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Declaration of own work

I declare that the work in this MSc dissertation was carried out in accordance with the requirements of the University's Regulations and Code of Practice for Research Degree Programmes and that it has not been submitted for any other academic award. Except where indicated by specific reference in the text, the work is the candidate's own work. Work done in collaboration with, or with the assistance of, others, is indicated as such. Any views expressed in the dissertation are those of the author.

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Ethics statement

To fill in according to the Dissertation Handbook Section 3.2.

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Abstract— **This paper proposes the jumping soft actautor desgin and characterises the actuator performance. This actuator can be used for rehabilitaiton of patients or in rescue operations. Jumping take-off velocity of 2m/s and the displacement of about 3cm were observed from a scaled leg model with femur size of 20cm.**

I. INTRODUCTION

Jumping is a powerful locomotion technique, used by many animals to overcome obstacles, escape predators, and traverse complex uneven terrain. Assistive jump systems can help disabled people in rehabilitation and improve the mobility of healthy people in rescue and military operations. However, many conventional robotics solutions use rigid components that may compromise comfort and even safety. To address these limitations, we propose a soft artificial muscle that is compliant and can store and release energy like muscles.

Recent research on artificial muscles for assistive applications has explored a variety of soft actuation methods to overcome performance and safety challenges. For example, as discussed by [1] and [2], numerous actuator types, including piezoelectric actuators and dielectric elastomer actuators, have been investigated for soft robotics. While these actuators offer impressive responsiveness and precision, their need for high operating voltages raises significant safety concerns for wearable and human-assistive devices.

In contrast, McKibben artificial muscles [3] (also known as pneumatic artificial muscles, or PAMs) operate via a biomimetic contraction mechanism that closely resembles natural muscle behavior. However, instead of storing and releasing elastic energy for rapid motion, McKibben muscles depend on pressurized air for actuation. This reliance not only limits their speed and efficiency in high-speed tasks like jumping, but also complicates the scaling process. When scaled to human-sized applications, these muscles require significantly higher air volumes and pressures, resulting in increased energy consumption and notable hysteresis losses that compromise precision and force output.

Similarly, studies on fluidic actuators, such as those used in small-scale legless robots [4], have demonstrated rapid and continuous motion. However, these systems have yet to be scaled effectively for human applications. In light of these limitations, there is growing interest in alternative actuation methods that can deliver high performance without compromising safety or efficiency, particularly for assistive technologies.

In parallel, considerable progress has been achieved in the development of exoskeletal systems aimed at enhancing human mobility. For example, exoskeletons designed for walking assistance have demonstrated a reduction in metabolic cost by approximately 15% [5], while those aimed at enhancing jumping performance have achieved increases in jump height of around 6.4% [6]. These systems typically incorporate spring-based mechanisms to store mechanical energy and facilitate load transfer through rigid skeletal support structures, with springs in some designs working in parallel with the quadriceps. However, achieving optimal energy storage remains a challenge. In this regard, [7] provides evidence that under conditions of short-duration muscle contractions, reducing the stiffness of the spring can significantly improve energy storage, suggesting that current designs could be further optimized for efficiency.

The ability to store and release elastic energy efficiently is a key feature of jumping systems. The study [8] shows a mechanism that utilizes antisymmetric equilibrium states to enable controlled energy storage and release, achieving efficient power amplification. The bistable jumper demonstrates multimodal locomotion, switching seamlessly between vertical height jumps (12.7 body lengths) and distance jumps (20 body lengths), such adaptability can be critical in certain emergency applications.

Furthermore, the concept of power amplification through latch mechanisms has been explored to achieve high-impact movements. In the work presented by [9], a motor is used to compress a spring, and the rapid release of a latch yields a significant burst of energy. This technique not only underscores the potential for achieving explosive force generation but also emphasizes the need for accurate timing and control to safely manage the sudden energy discharge during actuation.

These actuators must work in conjunction with human muscles, so we need to examine the biomechanics of jumping. The biarticular muscles present near the calf region transport mechanical energy to the distal joints of the body [10]. Study shows that the ground reaction force for a vertical jump can be between 1000N and 2500N [11]. The jump torque at the knee is between 80Nm and 100Nm for basketball players [12]. The optimal knee angle for jumping actuation also affects the maximum jump height, which was found to be between 87 degrees and 107 degrees.

Not just the mechanisms, but the material properties also affect jumping performance. A study by [13] reveals that the anisotropic stiffness properties of spider membranes—characterized by regions of higher modulus—play a crucial role in controlling the folding mechanics of spider leg joints . This natural strategy for adjusting mechanical properties dynamically provides a valuable model for developing artificial muscles that are both lightweight and capable of precise actuation. By mimicking such biological systems, it may be possible to design actuators with tunable stiffness properties that better accommodate rapid, energy-efficient motions.[14][15][16][17]

I scale down the forces design for experiment using the femur bone to height esimation [18]. Using normal bmi value we can get the weight to be about 19kg. The grf are about 1.5 times the body weight from [11]. Which means force of about 280N force. For the experiment, the assitive force have been kept lower.

In this paper, single-sided TPU coated Nylon is used to fabricate inflatable wrinkle bending actuator with multiple layers. This material is anisotropic. A single leg-like setup was used and the actuator was symmetrically mounted behind knee joint. The concept behind jumping is that the bending actuator can inflate and produce required torque, but it requires fact actuation, so it is coupled with a latch mechanism using weights which help in storing the elastic energy. The material stiffness also non-linearly increases under high strain rate. The main contribution of this paper is utilising the energy storage mechanism and simplification of bending actuator while ensuring high force output.

II. RESEARCH METHODOLOGY

A. Implementation

In this paper, inflatable soft bending actuator is studied. It is made up of Nylon with TPU coating on one side. This was chosen as it can produce high torque and can be used for in energy storage mechanism. For the experiment studied in this paper, the bending actuator was inflated while being loaded with known weights and deflection from the rest position and the jump height was studied. After the removal of loads, the actuator remains in a quasi-static phase after a fast dynamic jump, that reading is noted. While being loaded, the stiffness of soft material like TPU changes, which affects the jump. These measurements act as baseline for dynamic systems where impulse is added to the system instead of steady load.

- The initial calculation for the actuator was done using [19]. The actuator geometry is derived based on (eq (1)–(12)) as shown in the appendix. However, modifications were made to simplify the fabrication. Instead of fusing the fabric layers as done in the original method, the pouches were fabricated using a heat sealing machine and sewn 3 mm apart onto a fabric base. This spacing made it easier to attach and align the layers during assembly. The equations used for designig are:

  Actuator Geometry and Force Modeling

$$\delta = \frac{\alpha}{2(n-1)} \tag{1}$$

  where α is the bending angle and n is the number of layers.

$$a_1 = cs_1 \sin\delta \tag{2}$$

$$a_2 = Cs_2 - cs_1 \tag{3}$$

$$a_3 = Cs_2 \sin\delta \tag{4}$$

where

$$s_1 = (1 - \sin\delta)^{-1}, \quad s_2 = (1 + \sin\delta)^{-1} \tag{5}$$

To quantify the surface areas relevant to deformation and interface interactions, we define:

$$S_{\text{Def}} = 2wh \tag{6}$$

$$S_{\text{Inf}} = 2w(b_1 + b_2 + b_3) + S_{\text{side}} \tag{7}$$

where w and h denote the width and height of the deforming region, b_1, b_2, and b_3 represent the interface segment lengths, and S_{side} accounts for the additional side surface contributions.

The side surface area contribution is defined as:

$$S_{\text{side}} = 2\left(a_1^2\varphi_1 + a_3^2\varphi_2 + (a_1 + a_3)b_2\right) \tag{8}$$

where a_1 and a_3 denote the respective layer amplitudes, φ_1 and φ_2 are angular coefficients associated with each layer, and b_2 represents the intermediate interface length.

To determine the critical parameter c governing the system behaviour, we solve the quadratic equation:

$$P_2c^2 + P_1c + P_0 = 0 \tag{9}$$

where P_2, P_1, and P_0 are coefficients derived from the actuator geometry and material properties.

Force is given by:

$$F = Pw\left[(a_1 + a_3)\sec\delta + (a_1 - a_3)\tan\delta\right] \tag{10}$$

Torque provided by drive to knee:

$$\tau = MA \cdot F\cos\alpha \tag{11}$$

where MA is the distance between the force provided by the driver and the axis of rotation of the knee, defined as:

$$MA = C\tan\left(\frac{\alpha}{2}\right) - \sin\left(\frac{\alpha}{2}\right) + t_{\text{leg}} \tag{12}$$

- To prevent lateral distortion or slipping of the actuating layers, the fabric base was bonded to a thin plastic sheet.
- The base fabric was then mounted onto a T-shaped acrylic plate, which helped transmit force more effectively and provided a stable interface to connect the actuator to the mechanical leg.
- The actuator is attached to the back of the knee. It can be seen in fig 1.

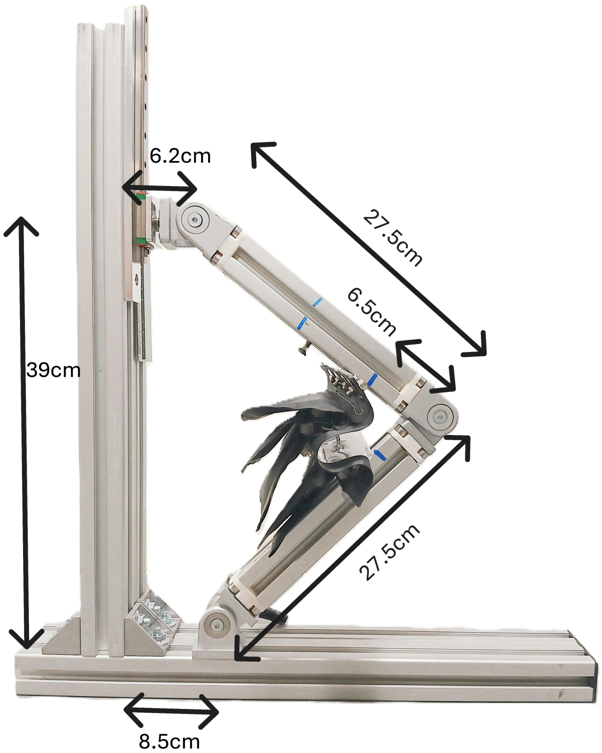

Fig. 1. Base setup without actuation

Fig. 2. dynamic loading setup

B. Experiment Methodology

- In the initial phase, the bending actuator is not inflated.
- The bending actuator is inflated at a certain pressures.
- This causes the leg to rise vertically. This is the maximum the leg can rise in unloaded case due to bending actuator and without any latch snap mechanism. Note this value as h_i.
- Repeat the process from initial phase only this time load the leg with known weight before inflating the bending actuator.
- This time the leg rises less than the hi due to additional weight. Note this height as h_f.
- Remove the weight and put some latch to keep the leg in the same height level. The same compression and the same stored elastic energy.
- Remove the latch. This should cause the leg to perform a jump motion.
- The leg will settle soon at a quasi-static phase after bump. We call this bump and not jump because the leg is fixed to the base in the experimental setup so it cannot jump but just provide upwards bump. This bump height is higher than hi. Note this value as h_b.
- repeat this process for varying weights
- repeat the whole process at different pressures.

There is also a dynamic experimental setup (see 2) where a laser sensor and load cell are mounted, and the force is applied dynamically with varying stroke lengths, with varying actuation pressures and with some additional static loading.

- Inflate the bending actuator at no static load.
- After the actuator ascension stops, shake it so that it gets settled on its stable height quickly.
- Now start sensor reading and data collection.
- Press the actuator to the required stroke length while observing the laser sensor reading.
- Suddenly remove force.
- Stop data collection.
- Depressurize the actuator.
- Repeat the same for required load condition of 400g
- Repeat this for the different pressure conditions
- Repeat this for different stroke lengths.

III. RESULTS

From fig 3, we can see that the actuator exhibits linear displacement behaviour under different loading conditions. With increasing pressure the displacement height lines get closer. fig 3 shows that the jump height also follows the same pattern and gets closer with increasing pressure. We can see from the graphs in fig 3 that after certain final height, the bump height plateaus in case of static loading. We can also see that the stable height is more for higher operating pressure. But the difference between stable heights decrease as pressure increases. This is likely because this material exhibits increased stiffness at high pressure conditions so even increasing pressure gives less strain. Bump height increases with pressures and is stable to higher loads than in less pressure condition. Bump height increases with pressure because more pressure can balance more weight and then removing latch can convert more of that elastic potential

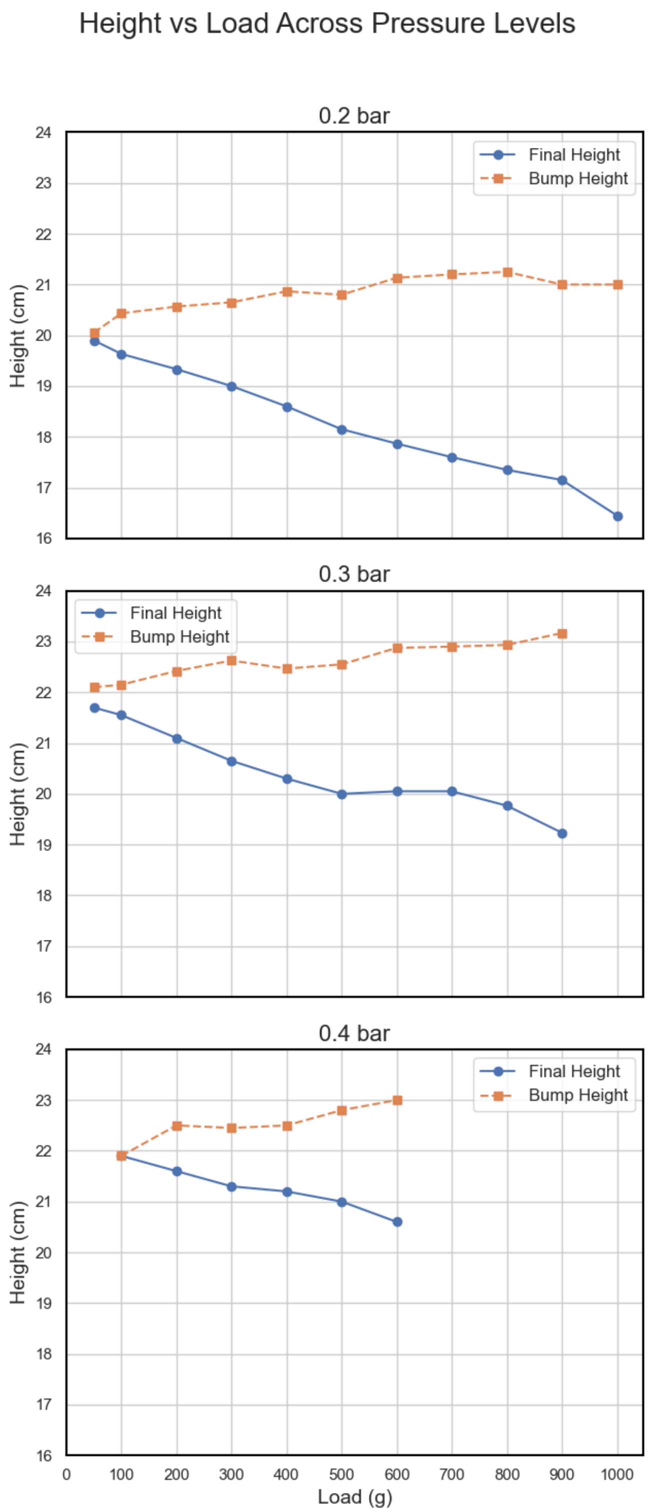

Fig. 3. Averaged Bump height from final height in loading at different pressures in static loading

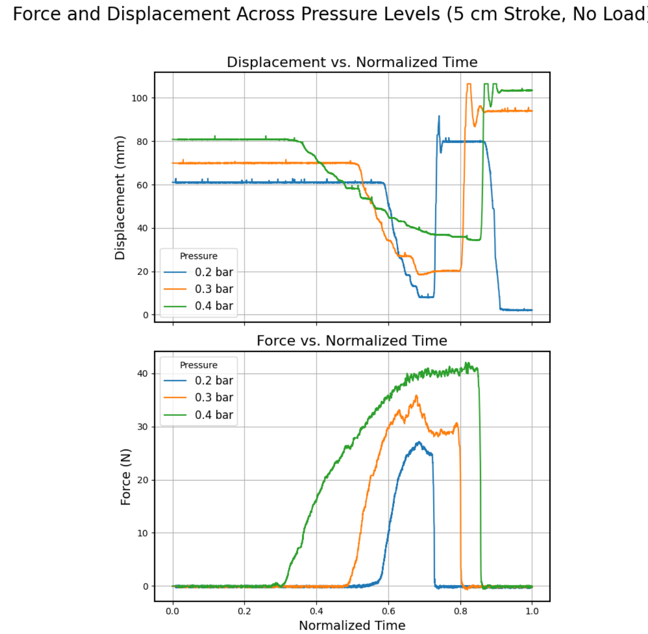

Fig. 4. Force and displacement in dynamic loading

energy to kinetic energy required for bump. Even after increasing loads the bump height stablizes, can be because at higher loads at a constant pressure, more force might be getting cancelled by the increasing stiffness of the actuator.

The force shown in fig 4 is applied on the actuator after it is inflated but in 3 the actuator inflated while in the loaded state. The downward displacement in 4 shows the dynamic loading phase, which is also accomapied by a corresponding rise in force. After the phase, the latch is removed and the actuator leads to vertical displacement shown by the sudden increase in height and drop in force. We can note that there is difference between initial stable height after inflation of bending actuator and the final height, this difference in height is the kinetic energy delivered by the actuator. The derivative of the displacement vs time in 4 gives a maximum take-off velocity of about 2m/s on average at 0.4bar pressure actuation. Average humans jump with a vertical velocity of about 3m/s [10].

We can observe the difference between static 3 and dynamic loading 4 that the bump height stablizes after certain loading and does not increase even with an increase in loading due to increase in stiffness, and we can see similar behaviour in dynamic loading (see 6) as the force rises non-linearly but the change in displacement is less. But in case of dynamic actuation the there shoudld not be sufficient time for stiffening, but the displacement is still less with increasing force, this might be due to viscoelastic damping losses resulting in internal heat losses.

In fig 5 and fig 6, the experiment is performed with a 400g static load on the actuator to measure the impact from no static load on from fig 4. It is observed that the increase in pressure leads to more displacement as compared to the increase in stroke length. This can be because pressure

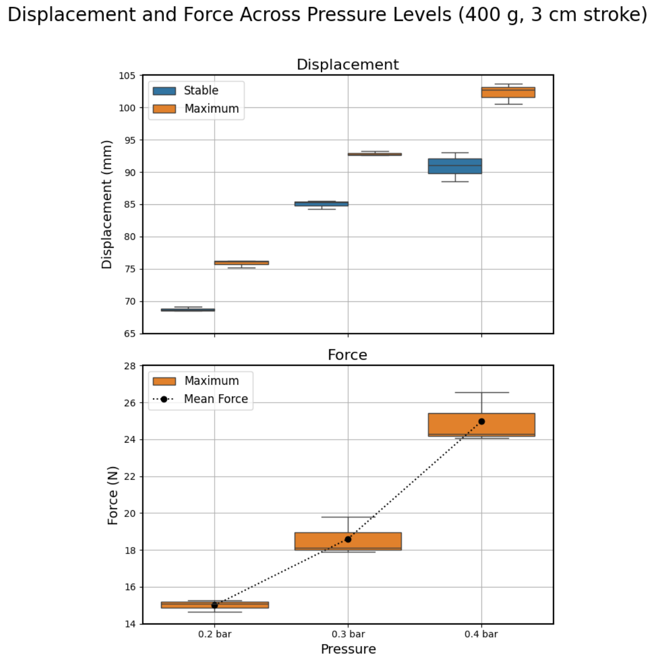

Fig. 5. Effect of pressure change in dynamic loading

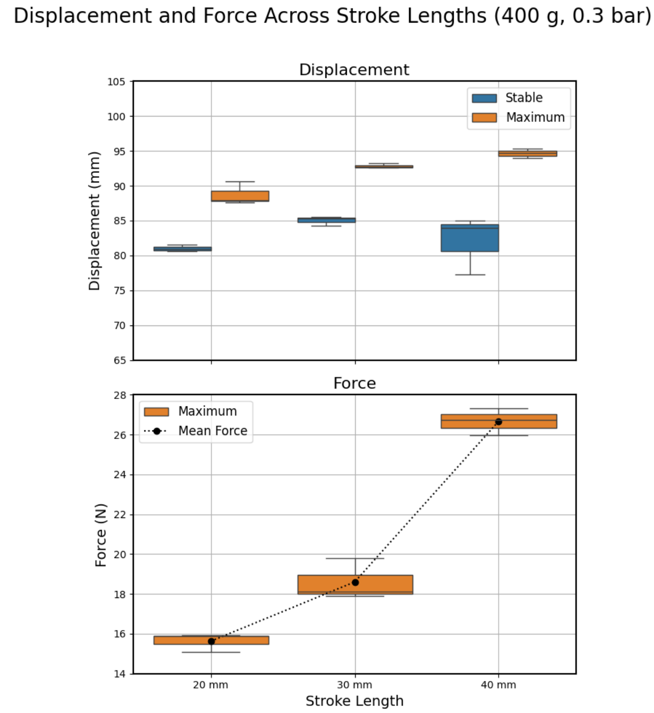

Fig. 6. Effect of stroke length change in dynamic loading

applies isotropically in actuating layer and it is internal so less losses but stroke length means applying force along vertical axis which can lead to some lateral dissipation of force. So, pressure leads to better transmission of force in this actuator.

IV. DISCUSSION AND CONCLUSIONS

The conclusion needs to provide

- In this paper, the energy storage capacity of the actuator in a static loading case with latch release was tested. This acts as a baseline for the dynamic loading system.
- Some of the limitations of this work are that since this was made using nylon fabric with heat sealing, there exists a limiting pressure where the heat sealing will break. This pressure is less than the pouch bursting pressure, so if a better fabrication method is chosen then the operating pressure can be increased. This placement of the actuator in this work was done by sewing the layers on the base, but this is not a precise way to do it. This work can be applicable to human jumping assistance if the person is injured or tired.
- This work helps to study the energy storage required for a jump from a crouched initial position. This helps in spring-like knee assistance [6] for jumping using soft and compatible materials with programmable stiffness properties dependent on load and pressure.
- there are limitations to achieving higher pressures by stroke length in both static and dynamic loading due to strain hardening during loading and viscoelastic damping respectively.These losses can be reduced by improving material such that they have higher linear region of operation and by improving the geometry of the actuator.
- There exist an optimal crouching force for humans after which the actuator produces less lift and at that point pressure control would be more feasible.

  Future work can also look at the better fabrication process for layer that can make the heat sealing stronger.This will help in increasing the operating pressures for higher jumps. Pressure sensor readings also be taken to know if operating pressure reduces during operation or if their is a need of a dynamic pressure control. The geometry of the actuator can also be improved. There should be a control system for actuation when there are actuators in both legs because even a slight delay in actuation can reduce jump height or even disbalance the person.

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