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Coupled, closed-system fluidic actuators for use in wearable rehabilitation devices

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Abstract— **This paper presents a novel closed-system, coupled soft actuator that aims to increase the applied bending moment that can be powered by a single pneumatic pump. The actuator incorporates both positive pressure and vacuum actuators of established design. The purpose of this development is to enable the design of an effective soft robotic wearable device for the rehabilitation of the revolute joints in post-stroke individuals. The design of a test rig to provide consistent, quantitative data on the output of the soft actuators is presented, allowing a comparison of the positive pressure, vacuum and combined (positive and vacuum) actuators. This combination demonstrates the ability to significantly increase the torque output when compared to a single actuator using the same pump for input, potentially reducing the weight of a wearable device. The closed-system, coupled soft actuator system shows opportunity for use in a wide range of applications due to this reduction in pump weight and isolation from environmental conditions.**

I. BACKGROUND

A. Stroke rehabilitation

Stroke is a medical condition which results from a loss of blood supply to the brain. The effects of stroke are wide-ranging, depending on the part of the brain affected, but often include problems with limb strength and coordination, which can be debilitating [1]. For a person recovering from stroke, getting immediate and frequent ongoing rehabilitation is key to recovery, though availability of one-to-one treatment with physiotherapists may limit this. For this reason, research has been conducted into robotic rehabilitation and many devices now exist which can assist physiotherapists. The InMotion ARM [2], MIT Manus [3] and iPAM [4] are examples of robotic devices designed to assist in stroke rehabilitation of the upper limb. Unfortunately, due to their large size, high cost and requirement for precise fitting and calibration, they are not suitable for home use and still require significant one-to-one time with a physiotherapist.

B. Soft actuators

A potential solution for reducing cost, size and to increase user comfort is the use of wearable rehabilitation devices based on soft actuators. In this paper we use soft actuators that are made of compliant materials and remain relatively compliant throughout their actuation. A class of soft actuators which lends itself to use in wearable devices is fluid-driven actuators, which are light and constructed from relatively cheap materials. These typically use pressurised air to produce movement.

Both the McKibben Actuator [5] and Bubble Artificial Muscles [6], examples of fluidic actuators, have been shown to produce forces comparable to those of human muscle for the same size and have been incorporated into rehabilitative and assistive devices. They produce a high force output, particularly in proportion to their weight, though their limitation is the low stroke length which can be problematic in high range-of-motion joints such as the elbow, which is the focus of this paper.

Another potential solution is the use of soft actuators which produce a bending motion, typically due to differential strain within the actuator as pressure is adjusted. There is a multitude of designs, several of which have been used in similar rehabilitation devices for the hand [7]. To determine whether this class of actuators would be suitable for the mobilisation of other joints in the upper limb, an understanding of the requirements for bending moment and range of motion of these in typical rehabilitation activities must first be achieved.

C. Requirements

Rosen et al. [8] conducted a study to determine these requirements whereby a participant conducted 24 activities of daily living (ADLs) whilst kinematic data of the arm was measured using a motion capture system. This data was then used to calculate the angle, velocity, acceleration, and torque based on a seven degree-of-freedom model. For the elbow, the maximum flexion angle was 162.9 degrees, and the maximum torque was 3.76 Nm. The maximum torque at the wrist was significantly lower, at 0.37 Nm in radial deviation and 0.25 Nm in wrist flexion. In a similar scoping study, Xiloyannis et al. give a figure of 146 degrees for the range of motion and 4.45 Nm for maximum torque at the elbow [9]. In a 1996 review of multiple studies into ranges of motion during ADLs [10], using a telephone required the maximum elbow flexion, at 160 degrees.

D. Existing Devices

Many stroke rehabilitation exercises consist of performing specific tasks with varying levels of assistance from a physiotherapist [11], therefore the above values would be a suitable target for elbow flexion in a rehabilitation device. Some muscle strength remains, meaning the required assistance would be a proportion of this maximum torque. Another consideration is the provision of elbow extension,

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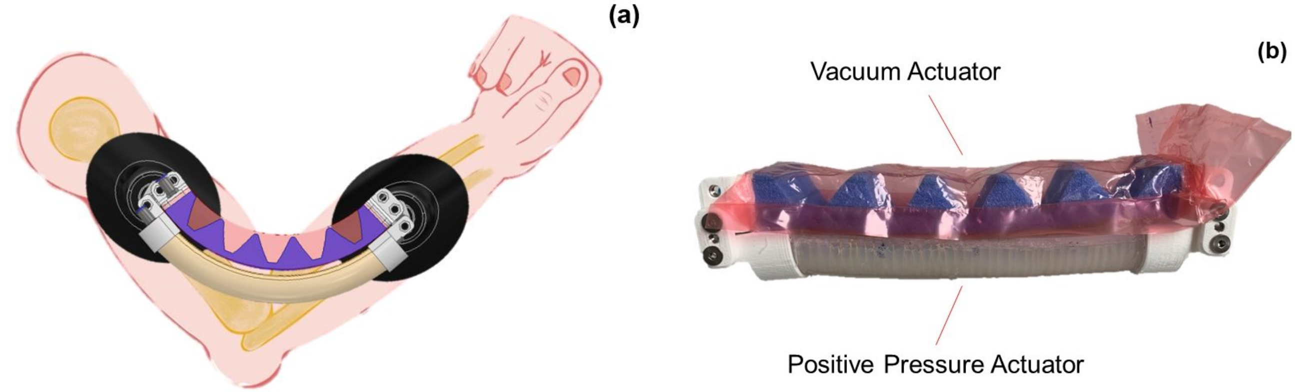

Fig. 1. (a) Conceptual drawing showing potential application of coupled actuators in a wearable device (b) Combined Actuator

the straightening of the arm. Nam et al. [1] have demonstrated the ability of a simple pneumatic chamber to assist with elbow extension, by inflating an inelastic chamber which straightens on pressurisation. Elbow extension has the added complication of spasticity, the involuntary contraction of muscles. Typically, this occurs in the biceps post-stroke and causes the lower arm to be lifted to the shoulder [1]. As it has been shown that simple pneumatic actuators can be used for extension, this study will focus on flexion of the joint.

A wearable device is presented in [12], which provides both flexion and extension assistance at the elbow using soft robotic actuators. Whilst the tip force of both the flexion and extension actuators is provided, this does not allow for comparison with similar devices, nor the requirements for torque at the elbow. The device was tested on humans and provided only 50% of the required range of motion. Wearable devices intended for healthy humans to assist with manual tasks are presented in [13] and [14]. Both use thermoplastic polyurethane (TPU) bladders which are able to provide significant bending moments of up to 27 Nm, well in excess of the levels determined above. However, the downsides of these systems are firstly a requirement for large volumes of high-pressure gas, and secondly that they cannot provide the full range of motion of the elbow joint.

E. A novel, closed-system design

In the combined actuator system proposed in this paper (see Fig. 1b), fluid is removed from the vacuum actuator and placed within the positive pressure actuator, creating a closed system with no requirement for an external stored volume. This builds on previous work, where it has been shown that both positive pressure and vacuum actuators can be used together to increase the force output and performance of soft actuators [15].

The novel aspects of this work are the use of a single pump and controller for both positive pressure and vacuum and the use of a closed system. For wearable devices, it is typically desirable to minimise the size, weight and power consumption of the pump. By using a closed system, a single pump can be used to drive both the positive and negative actuators, reducing the overall size and weight of the device. In addition, the actuator control is simpler in coupled, closed-systems than it is for actuators with separate controllers.

The primary aim of this paper is to characterise the performance of the coupled, closed-system design and compare this to the performance of the individual actuators using a novel method of measuring the maximum bending moment over a certain range of angles. A secondary aim is to assess the potential for use in a wearable device for stroke rehabilitation, by comparing the bending moment and range of motion with those of the elbow during activities of daily living.

II. MATERIALS AND METHODS

A. Actuators

To understand the characteristics of the soft bending actuators, tests were performed to compare the bending moment from 0 to 150 degrees in flexion, which is the typical range of motion of the elbow joint identified above. The soft actuators we used were a positive-pressure fibre-reinforced silicone based on the instructions of the Soft Robotics Toolkit [16] and a vacuum actuator based on the design by Li et al. [17]. These were chosen as they are both established, simple and lightweight designs with potential for use in wearable devices.

The positive pressure actuator consists of a moulded silicone bladder (Dragon Skin 10, Smooth On) with strain-limiting textiles wrapped around cylindrically and fixed on one side. The wrapped fibre constrains the radial expansion of the bladder, and the additional layer causes a strain differential between the upper and lower faces, causing bending [18]. The actuator used in this testing was scaled, such that the length and wall thickness remained the same as the Soft Robotics Toolkit example, but the internal diameter was doubled to 25.4 mm.

The vacuum actuator uses the principle of a shaped foam structure encased within an air-holding bladder. Glued to either end of the foam structure is a 3D-printed end cap with M4 nuts encased. The bladder (Aerofilm VB160) was impulse sealed using an RS Pro Type G 400mm impulse sealer to form an airtight chamber. An additional 3D-printed

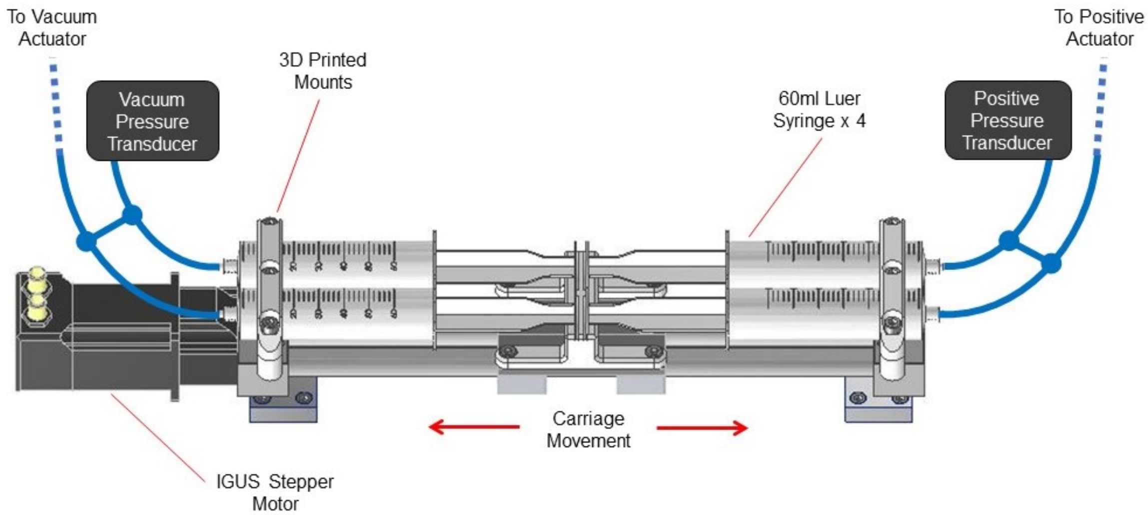

Fig. 2. Pump used for testing, consisting of IGUS Linear Actuator and four Luer 60 ml syringes

bracket is then screwed to the end cap and the holes sealed with silicone. When a vacuum is applied, the bladder conforms to the structure, causing a differential strain within the top and bottom layers of the bladder which causes bending. The two actuators combine effectively as the two strain-limiting layers can be connected, forming a single strain-limiting layer within the actuator.

B. Test Arrangement

Tests were run to characterise the bending moment produced by the actuator at angles of 0 degrees (straight) to 150 degrees (fully-flexed), at 10-degree increments. Understanding these characteristics is necessary to design an effective wearable system and provide an indication of the behaviour in its end-use, attached to the the arm of the patient undergoing rehabilitation.

To ensure that the volume of air applied to both the positive pressure and vacuum actuators was consistent, a pump was built using a linear actuator (IGUS SLW1040) and four 60ml Luer syringes (Cole Parmer 07945-42) connected by 4mm OD rigid tubing (Festo PEN-4X0,75-BL). By programming the stroke of the linear actuator (using an IGUS D1 controller), a known volume could be added and/or removed from the two actuators (see Fig. 2). A volume change of 100ml was effected in the syringes, with the aim of producing a 100 kPa change in the positive pressure actuator. This pressure was chosen to maintain a safety factor of two within the actuator, and similar actuators have been shown to burst around 200 kPa. When data collection was complete, the test actuator was tested and found to burst at a pressure of 230 kPa.

To reproduce the loading conditions of use in upper limb rehabilitation, the actuators were connected to a single degree-of-freedom revolute joint, constructed from 3D-printed parts and 12mm OD carbon fibre tube. These lightweight materials were used to minimise the mass of the test equipment (lower arm mass was 7 grams). Attached at the joint axis was a wheel, which rotated with the distal portion of the arm. One end of an inextensible aramid thread was attached to the circumference of the wheel, with the other end attached to a load cell. As the radius of the wheel is known, by measuring the force at the load cell, the moment at the elbow could be calculated (see Fig. 3). The force output was measured using a 3kg load cell (OBUG-3kg-0.5-000, Applied Measurements). Actuator pressure was measured using two differential pressure gauges (Honeywell SSCSNBN030PDAC5) and the angular position of the joint was measured using a rotary hall-effect sensor (Piher PSC360G2-F1A-C0000-ERA360-05K). All sensors provided a voltage output, read by the National Instruments Data Acquisition (NI-USB-6341 DAQ) and logged using Matlab.

C. Work input

By using a single pump to drive both actuators, the mass of the wearable device could be reduced. However, the additional force of two actuators does not come for free, and more work must be done by the pump to pressurise both the positive pressure and vacuum actuators. As the Luer syringe has a known cross-sectional area and the stroke of the IGUS linear actuator is also known, we can calculate the work done (W) by the linear actuator as follows:

$$W = \frac{(P_2 - P_1) \cdot (x_2 - x_1)}{2} \cdot A_p \tag{5}$$

Where x_1 and x_2 are the start and finish positions of the linear actuator and P_1 and P_2 are the measured pressures at the start and finish for both the vacuum and positive pressure systems, A_P is the area of the piston. This is calculated from the point the actuator begins moving to the point of maximum pressure. By comparing the moment generated by

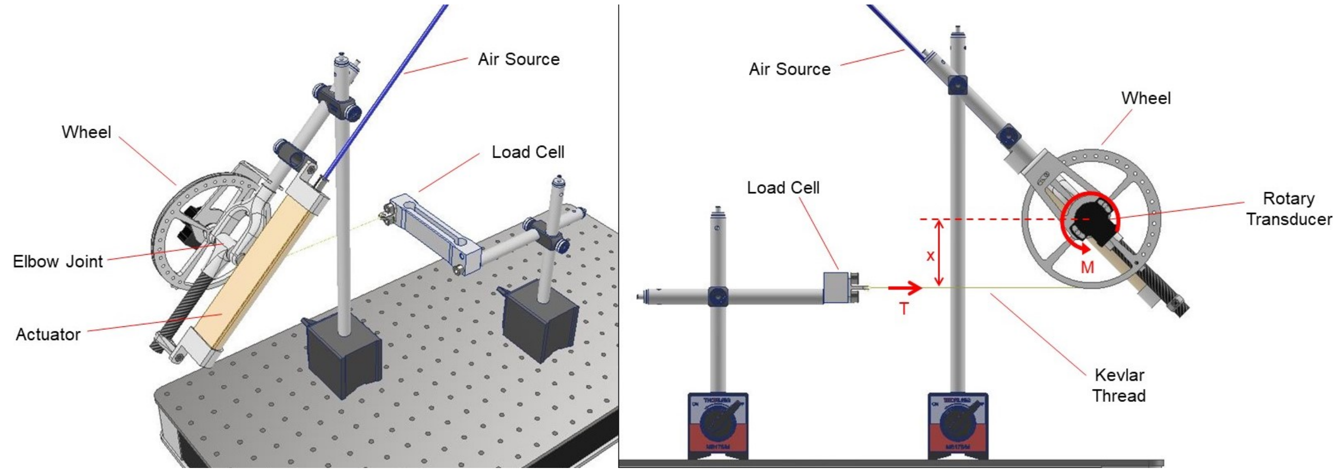

Fig. 3. Test arrangement for the positive actuator, showing the attachment method, elbow and wheel setup and the force applied to the load cell

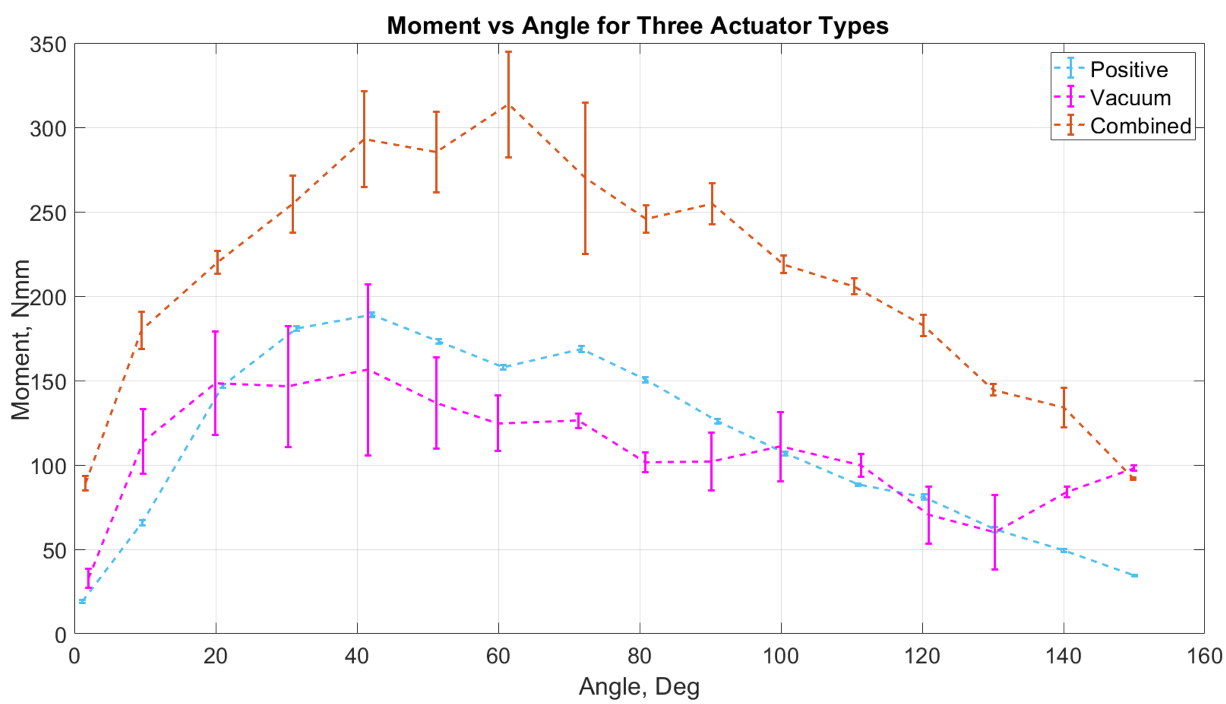

Fig. 4. A plot of bending moment vs. angle for the three actuator types

the actuator at various angles with the work done by the pump to generate it, a form of efficiency can be calculated. This allows a comparison between the actuators tested, but cannot be related to true values of efficiency (work in vs. work out) calculated elsewhere. Note that this approximation assumes that the pressure differential increases linearly.

D. Testing

The procedure for testing was as follows. The elbow joint was set to the required angle (between 0 and 150 degrees) by rotating the wheel and holding it in place with a pin, as shown in Fig. 3). The distance between the load cell and actuator mounts was then adjusted such that the aramid thread was tight but under minimal tension. The actuator was then pressurised, held for a period of 5 seconds and then returned to ambient pressure. A slot for the pin allowed the actuator to flex an additional 10 degrees before being stopped, meaning that upon pressurisation, only the aramid thread attached to the load cell was inhibiting further movement. This was repeated five times for each angle and each actuator (a total of 225 operations).

Following the tests using the syringe-based pump, additional testing was carried out on the combined actuators using a 12 V vacuum pump (Airpo D2028B), with both actuators connected such that air was drawn from the vacuum actuator into the positive actuator. This followed the same test regime at angles of 0 to 150 degrees, to demonstrate the feasibility of the concept with a more realistic power source.

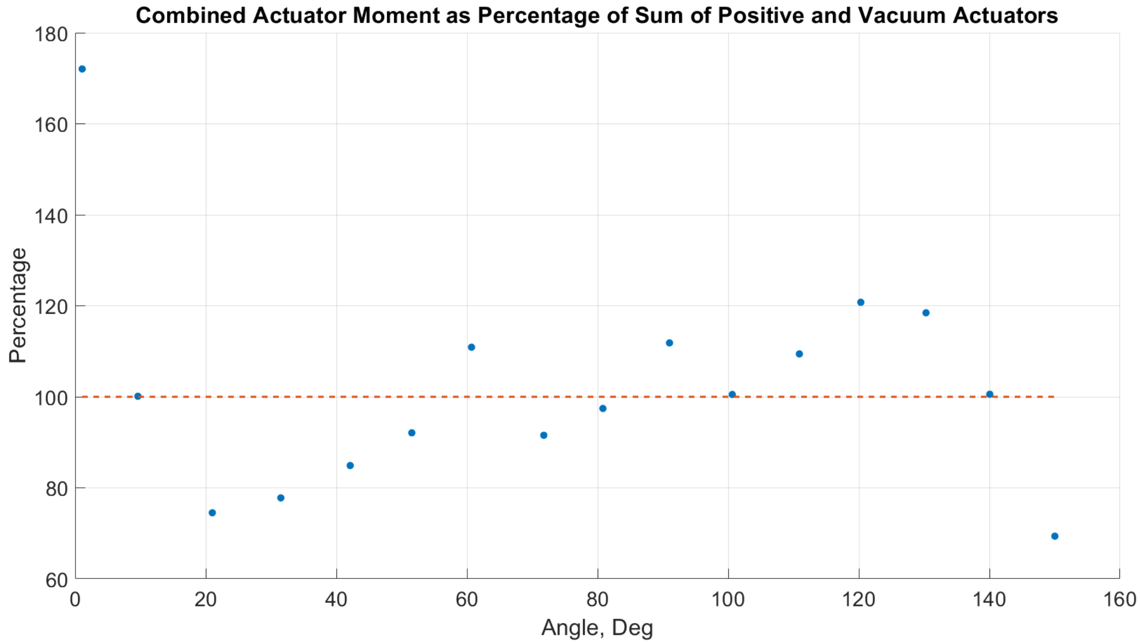

Fig. 5. A plot of the combined moment as a percentage of the sum of positive pressure and vacuum actuators

III. RESULTS

A. Bending Moment

Fig. 4 shows the maximum bending moment at each angle for the positive, vacuum and combined actuators, plus the standard deviation of the 5 measurements for each data point. The positive pressure actuator produced a moment which increased with flexion angle until a peak of 189 Nmm at 40 degrees, after which it gradually reduced. The maximum standard deviation of these measurements was low (±1.7 Nmm), showing good repeatability of results.

The vacuum actuator also produced its peak bending moment of 156 Nmm at 40 degrees, though this showed less of a reduction with increasing angles, when compared to the positive pressure actuator. The maximum standard deviation of measurements was significantly higher (±50 Nmm). It was noticed that the behaviour of the actuator upon pressurisation depended on the relative movement of the film and skeleton, which varied with each iteration.

The combined actuator produced a peak bending moment of 314 Nmm at an angle of 60 degrees. The maximum standard deviation of ±45 Nmm was also much higher than that of the positive actuator, though this is to be expected as it also includes the highly-variable vacuum actuator.

As Fig. 5 shows, the combined moment varies between 70 and 180 percent of the sum of the two actuators on their own and the mean percentage is 102 over the full range. The ability to produce bending moments greater than both actuators individually could be due to the increased second moment of area and therefore stiffness from both actuators joined together. This demonstrates a significant advantage to using both actuators compared to using them individually where a single pump can be used.

B. Efficiency

As described in II-C, a form of efficiency was calculated for each actuator at all angles, allowing a comparison between them. Fig. 6 shows the efficiency of these from 0 to 150 degrees. As the pressure differential for the positive pressure actuator is much higher than that of the vacuum actuator, the work done by the pump is also much higher.

The vacuum actuator is by far the most efficient, especially at lower angles of up to 90 degrees where the bending moment generated per unit of work done is almost four times as much as the positive and combined actuators. If energy use were a key driver of the design, utilising solely vacuum actuators would most likely be best. However, if the weight of the device were the key driver, the addition of a positive pressure actuator could significantly increase the moment applied without additional pumps.

C. 12 V Pump

Testing of the bending moment achieved by the combined actuator during the 12 V pump testing described above gave the following results. The peak bending moment achieved occurred at 23 degrees of flexion and was 465 Nmm, which exceeds the maximum achieved using the syringe pump (313 Nmm). The reason for this higher result is that the pressure differential was higher for both the positive actuator (103 kPa vs 82 kPa) and the vacuum actuator (-67 kPa vs -33 kPa). There was a gradual decline in bending moment as the angle increases which can be explained by reviewing the pressure within the positive actuator, which reduces steeply from 103 to 45 kPa. Within this closed system, the only source of pressure for the positive actuator is the air removed from the vacuum actuator. As the flexion angle increases, the

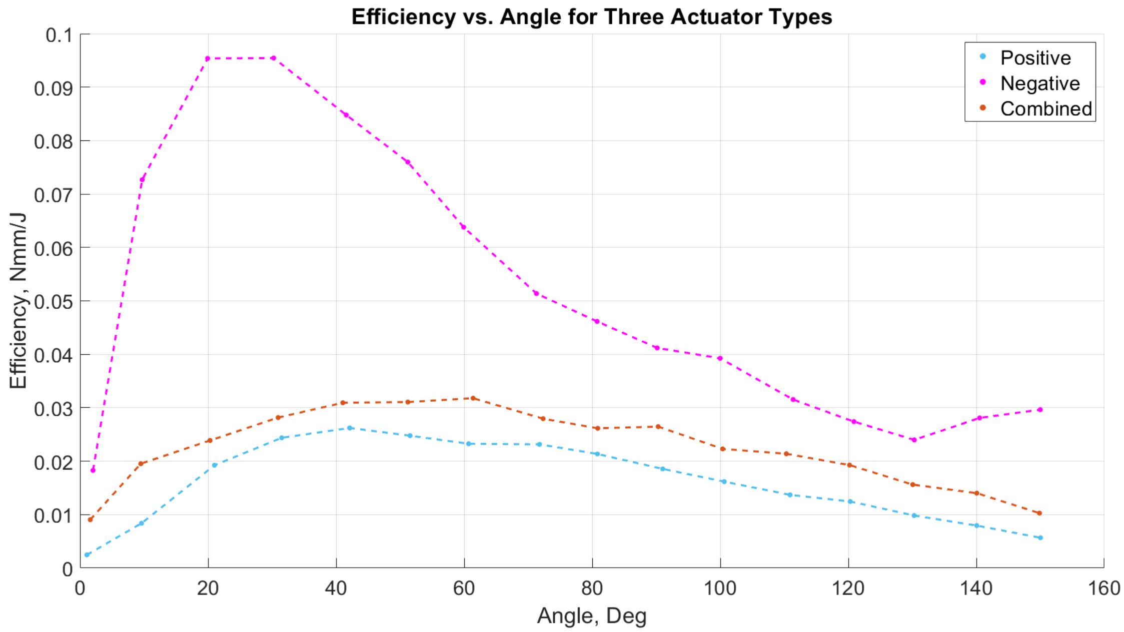

Fig. 6. A plot of efficiency vs. angle for the three actuator types

initial volume of the vacuum actuator decreases, reducing the available volume for the positive actuator. This highlights a potential drawback of closed systems such as this and the need for careful balancing of internal volumes and pressures for both actuators.

IV. Discussion

As discussed in Section I-D, several devices have been developed using soft robotic actuators to assist movement of joints within the upper limb. Whilst some of these devices are able to produce sufficient torque, their most significant drawback is a requirement for large volumes of high-pressure gas. Whilst the torque generated by the actuators tested in this paper is much lower than that of [13] for example, the principle of combining positive and vacuum actuators can be transferred to various actuator types and has the potential to increase performance, reduce pump weight and complexity.

The second drawback to existing devices is a range of motion which is less than the maximum required for rehabilitation. It has been shown within this paper that the actuators tested can provide 150 degrees of flexion. Whilst the test was designed to closely mimic the revolute joints of the upper limb, the attachment to a human arm will no doubt provide additional challenges. The device will require re-designing for an efficient transfer of torque from soft actuator to the musculoskeletal system whilst ensuring comfort and ease of attachment.

By using a closed system, additional benefits could also be sought in other applications. The use of a working fluid within the system which is different to the ambient fluid of the environment is often used in hydraulic systems, though these typically cause additional complications in the form of fluid storage. As this is a volume-matched closed system, no storage is required and the benefits of incompressibile fluids can be sought. For extremely cold environments, the system could be driven by a a non-freezing glycol to avoid icing.

V. Conclusions & Further Work

In this paper the performance of the coupled, closed-system design has been characterised and compared with the performance of the individual actuators using a purpose-built test rig. It has been shown that meaningful data can be derived from the described method, allowing a quantitative comparison between similar actuators. Through this characterisation, it has been demonstrated that by incorporating both positive pressure and vacuum actuators, the maximum bending moment generated by a single pump can be significantly increased. For an existing vacuum-only system, the addition of a positive pressure actuator could double the bending moment for an additional mass of 52 grams.

A secondary aim was to assess the potential for use in a wearable device for stroke rehabilitation, which influenced the test rig design and test parameters such as range of motion. From the work done by [8], for rehabilitation that consists of performing tasks similar to ADLs, a maximum torque of around 3.8 Nm is required at the elbow and 0.38 Nm at the wrist. However it is currently unclear what proportion of this moment would need to be generated by the device and further work is required to determine the true bending moment requirement. The results do show however, that a full range of motion of 150 degrees can be achieved by all actuator configurations. Future work will include testing additional pneumatic actuators (such as Pneu-Nets or Knit-Textile actuators), which would allow further quantitative comparison between these. Further investigation into the relative volume and pressure of the positive and vacuum actuators could also lead to increased performance.

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