PRELIMINARY INVESTIGATION OF STATIC AND DYNAMIC HYSTERESIS OF DMSP-5 FLUIDIC MUSCLE

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Pneumatic artificial muscles are pneumatic actuators with elastic structure distinctive for their nonlinear characteristics and hysteresis. In the paper we concentrate on the investigation of hysteresis in one of the recent types of fluidic muscles (DMSP-5). The main objective is to examine the hysteresis with respect to its rate-independence as well as its nonlocality. We used two types of excitation signal (triangular with linearly decreasing amplitude and triangular with randomly chosen amplitude) and two types of tests (quasi-static and dynamic with two different frequencies) to observe the overall quality of hysteresis. It was found out that the hysteresis show intricate behavior in response to the random excitation and well-defined formation of major and minor loops in response to linearly decreasing signal. It can also be described as rate-independent, with the width of hysteresis major loop being maximal for quasi-static measurement (5.26 mm) but with small differences between static and dynamic tests.

KEYWORDS

non-local hysteresis, muscle pressure, muscle displacement, pneumatic artificial muscle, loop formation

1 INTRODUCTION

Pneumatic artificial muscles (PAMs) are pneumatic devices with a special construction consisting of a rubber tube and inextensible braiding and capable of contracting when filled with pressurized air. They have been the object of intensifying research since late 90s due to their remarkable power/weight ratio, inherent compliance and safety and clean operation. As such they are suitable for applications in industrial area, biorobotics or rehabilitation ([Robinson 2014], [Liu 2015], [Sekine 2016]). Since the very beginning of PAMs research, their specific properties naturally associated with unusual construction have been viewed as a challenge for modeling and control. Of those, hysteresis has occupied important place as a typical trait of PAMs that needs to be addressed if high control performance is to be achieved. Significant number of works have been devoted to the study of PAM modeling, with some of these dedicated particularly to modeling of hysteresis. An important update on the state of PAM modeling was presented by one of the pioneers in the field in [Tondu 2012]. In the time of publishing, this article presented then-current state-of-the-art in static and dynamic modeling of McKibben pneumatic muscles with one part dedicated to a hysteresis phenomenon in PAMs, particularly with emphasis on identification of possible sources of hysteresis in this type of pneumatic muscles. In [Antonelli 2017], finite element method was used for modeling of PAM with inner tube material nonlinearity taken into account yet with friction between muscle strands neglected. Both isotonic and isometric tests were carried out to validate the developed models. Researchers in [Doumit 2017] proposed the dynamic model of PAM where muscle geometry, force, inertia, fluid dynamics, static and dynamic friction, heat transfer and valve flow were all included. The model presented in this work did not explicitly included hysteresis modeling. In contrast to usual function of PAMs as active devices, the work [Doumit 2017] presents the development of stiffness model where its use as a passive device is assumed. A modified Prandtl-Ishlinskii hysteresis model was used in [Liu 2017] to model a hysteresis in length-pressure relationship of PAM using two asymmetric operators for ascending and descending branches. Similarly to [Nozaki 2014], finite element method was used for the development of PAM model with a separate simple model without strand friction developed first and then followed by the development of friction model. Another approach to PAM modeling is presented in [Sarosi 2015a], where the analytical model of muscle force is developed with its parameters determined using optimization algorithm. The resulting model is then validated on an experimental rig consisting of a muscle place horizontally, slider and screw spindle. The hysteresis phenomenon in PAM is specifically treated in [Sarosi 2015b], where the author uses an analytical expression for muscle force with experimentally determined coefficient to approximate hysteresis in force/contraction relationship. A more theoretical approach to PAM modeling can be found in [Sorge 2013] where researchers took into account the stress field inside the rubber tube regarded as Mooney-Rivlin hyperelastic material. Preisach hysteresis model was used in [Van Damme 2008] to model a hysteresis in special type of muscles called pleated pneumatic muscles. Other type of hysteresis model appeared [Vo-Minh 2011], where Maxwell-slip model as a lumped-parametric quasi-static was used to model force/length hysteresis in PAM. The return to the first effort in PAM modeling can be seen in [Wang 2015], but this with improved non-linear quasi-static model based on finite strain theory.

In most of the works mentioned above researchers used PAMs of McKibben type with a rubber tube enclosed by a braided mesh. Different construction with strands being embedded into the tube is used in muscles made by Festo, which are termed fluidic muscles. We used the latest type of these muscles (DMSP-5) with 5mm diameter, which are intended for special applications where a powerful yet extremely light actuator is needed. Our objective is here to investigate the basic properties of hysteresis in this type of fluidic muscle under isotonic conditions, which is important for its compensation using an inverse hysteresis model in position control. We examine the hysteresis present in fluidic muscle quantitatively (in terms of hysteresis width) and qualitatively (in terms of its rate-independence and non-locality).

2 HYSTERESIS AND ITS PRESENCE IN FLUIDIC MUSCLE

Hysteresis can be most fundamentally described as a dependence of the state of a system on its history. Despite being most naturally associated with magnetism, it is found in a number of other areas including unconventional actuators (piezoelectric, shape memory alloys or elastic). If we disregard the internal structure of a system and only account for the dependence of continuous output $y(t)$ on continuous input $u(t)$, we can observe the possible evolution of $(u(t),y(t))$ dependence in a hypothetical hysteretic system in Fig.1. If the input is increased from u_1 to u_2, the output follows ABC curve that is monotonically increasing. If the input starts to decrease from $u_2$to u_1, the output starts to follow different curve (CDA) that is monotically decreasing. In the example shown, the curve ABCDA forms so called major loop of the hysteresis. If, on the other hand, direction of the input is change within the interval $u_1 < u < u_2$, the output follows smaller loop bounded by the boundaries of major loop [Bertotti 1998].

According to [Bertotti 2005], at any time instant *t* depends *y(t)* on previous evolution of input *u* and on the initial output y_0

$$y(t) = [\Psi(u, y_0)](t),\ \forall t \in [0, T] \qquad (1)$$

while

$$(u(0), y_0) \in \Theta,\ [\Psi(u, y_0)](0) = y_0 \qquad (2)$$

where $\Psi(\cdot, y_0)$ - operator that acts among suitable spaces of time-dependent functions for any fixed y_0. It is also of note that the rate-independence is considered the main characteristic of hysteresis. This characteristic is related to the fact that at any time instant, *y(t)* depends only on $u([0,t])$ and on the order in which these values were attained. If a periodic signal is considered, then the values of output should not depend on the frequency of input signal.

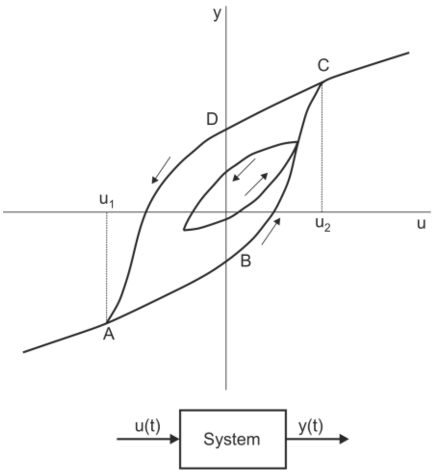

Figure 1. Hysteresis in a system (u(t) – time dependent input, y(t) – time dependent output)

Hysteresis with local and nonlocal memory

In systems with local hysteresis, the actual values of *(u,y)* are enough to identify the state and each point in *(u,y)*-plane is uniquely associated with one and only one state (Fig.2 left) [Mayergoyz 2003]. In a simple case as shown in this figure, the subsequent evolution of the state from given point will depend only on direction of input variation, i.e. if it increasing or decreasing. It is also obvious that branching in that type of hysteresis occurs only at the points of major loop.

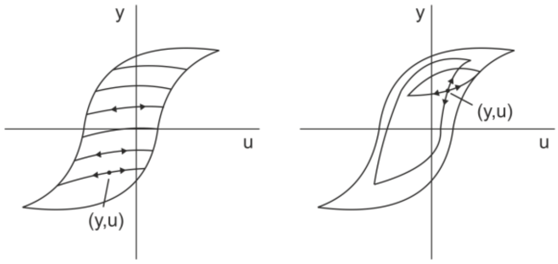

Figure 2. Local (left) and nonlocal (right) hysteretic memory

Much more complicated type of hysteresis is shown in Fig.2 on the right. In this case the knowledge of state *(y,u)* is insufficient for determining the subsequent evolution of the trajectory since an infinite number of trajectories can actually pass through a given point. The future evolution is thus dependent not only on given state but also on the previous history of inputs.

Pneumatic artificial muscles are well known for the presence of hysteresis, which makes the objective of achieving good tracking control performance very difficult. The existence of hysteresis in PAMs is associated predominantly with its unique construction composed of a rubber tube and a braided shell.

More specifically, four possible reasons for hysteresis in PAMs were identified in [Vo-Minh 2011] and later analyzed in [Tondu 2012]:

I. Friction between strands
II. Friction between strands and rubber tube (bladder)
III. Conical deformation of the bladder at its ends
IV. Stretching of the bladder due to increased volume

It was pointed out in [Tondu 2012] that the first reason should be considered the most probable as a source of hysteresis in PAMs due to others being either unable to cause it or too insignificant to have noticeable effect. On the other hand, the analysis presented in [Tondu 2012] is clearly related to PAMs whose structure is formed by an internal bladder surrounded with a braided sheath. Fluidic muscle made by Festo (used in our experiments) uses a different structure with the strands being embedded within the rubber tube itself in two layers placed a certain distance apart (Fig.3) [Bergemann 2002].

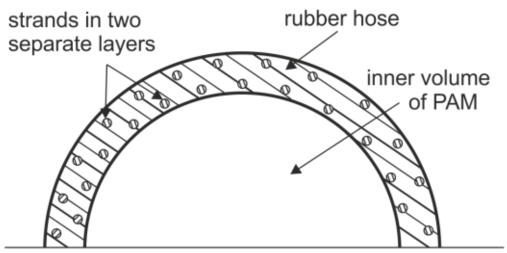

Figure 3. Inner structure of Fluidic muscle according to [Bergemann 2002]

The material between the layers is yielding, meaning that it does not interfere with the movement of strands, yet it is able to maintain their separation. This difference in construction puts the first reason (i.e. friction between the strands) for hysteresis in Fluidic muscle out of question and makes other reasons (possibly related to the yielding material) more favorable. Currently, we do not attempt to identify the reasons for hysteresis in fluidic muscle but only concentrate on measuring its extent and character (both static and dynamic) in displacement under isotonic (constant load) conditions.

Since hysteresis represents a nonunique mapping between the input and output variables dependent on the history of inputs, its presence adversely affects the control performance. One of the effective control strategies to tackle this problem is to use an inverse hysteresis block with the desired value of control variable $y_d(t)$ at its input (Fig.4).

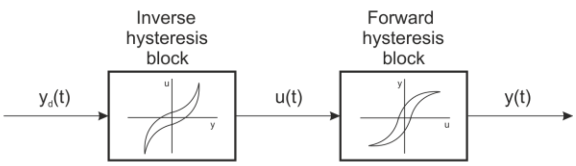

Figure 4. Compensation of hysteresis using inverse hysteresis block used in control of hysteretic systems

Due to the need for inversion of hysteresis, it is important whether given hysteresis operator is in analytical form, which allows for its inversion. For certain types of operators, approximate data-based methods (e.g. in the form of look-up tables can be viable alternative [Esbrook 2013]. Using only feedforward model-based hysteresis compensation might be insufficient, as the errors in inversion (for instance, due to uncertainties in parameters) are always present. Their effect on control performance can be effectively reduced by introducing a feedback loop into the control system.

3 DESCRIPTION OF THE SYSTEM

The experimental setup used in measurements is shown in Fig.5. Its main component is FESTO DMSP-5 fluidic muscle with 5 mm diameter and length of 130 mm. This version of fluidic muscle can generate a maximum force of approximately 140 N at maximum operating pressure of 600 kPa. The displacement of the muscle was sensed using MSK1000 magnetic incremental sensor (accuracy $\pm$10 µm, resolution 1 µm) with MB100/1 magnetic band (resolution 0.01 mm). The flow of the pressurized air into the muscle was controlled through a pressure regulator (Matrix EPR50) using which it was possible to accurately set the desired muscle pressure. The version of pressure regulator used in experiments needed 4-20 mA current input signal for setting the desired muscle pressure, which was converted from 0-10V analog ouput. The actual value of pressure in the muscle was measured using the built-in pressure sensor in the pressure regulator with 10 V analog output. Both control and sensor signals were processed through I/O PCI card (Humusoft MF624) using one analog output (pressure regulator), one analog input (pressure sensor) and one encoder input (magnetic sensor). The picture of actual experimental set-up together with its main components is shown in Fig.6.

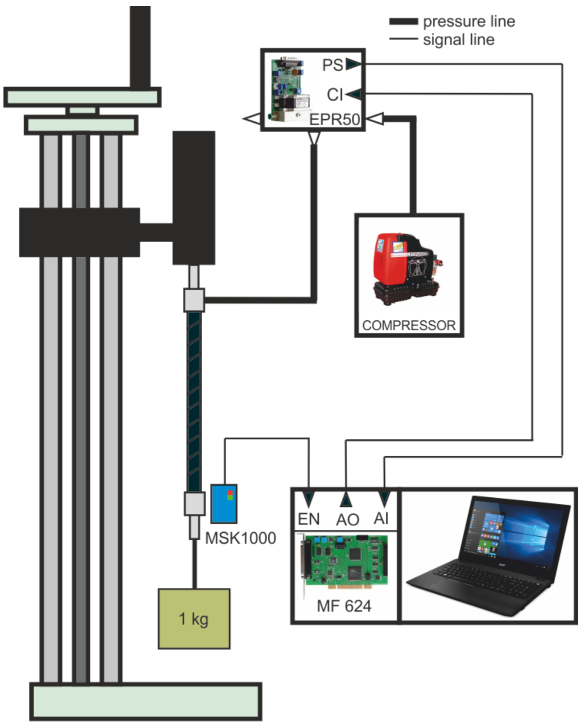

Figure 5. Schematic diagram of the experimental apparatus (MSK1000 – magnetic sensor, MF624 – I/O card, EPR50 – pressure regulator, PS – pressure sensor output, CI – 4-20 mA current input into pressure regulator, EN – encoder input, AI – analog input, AO – analog output)

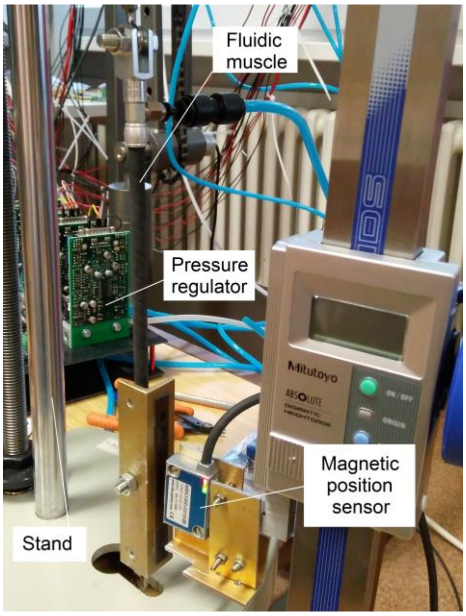

Figure 6. Test rig with Festo DMSP-5 used in experiments

4 EXPERIMENTAL RESULTS

In contrast to our previous work [Hosovsky 2016], we moved from using simple on-off valves for controlling the flow of pressurized air into the muscle to a pressure regulator capable of maintaining desired pressure at the output (i.e. in the muscle) through closed loop control. In this way we replaced the control of valves through a digital signal with direct control of muscle pressure through an analog signal. The control system for muscle displacement can be seen in Fig.7. With regard to fluidic muscle itself, the control (and also input) variable can be considered its internal pressure P_m. From control viewpoint, it is necessary to consider the pressure regulator, which works as a power modulating element. To investigate the phenomenon of hysteresis, the relationship between P_M and κ is of interest yet it may be important to look at the dynamics of pressure regulator as well.

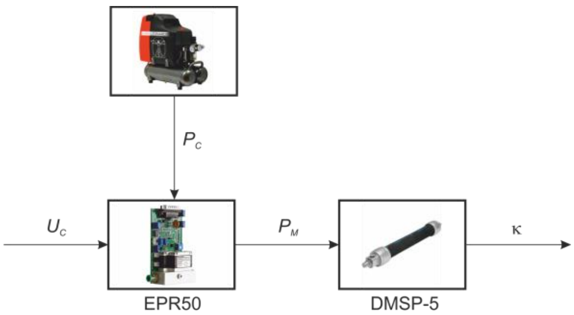

Figure 7. Control system of fluidic muscle displacement (U_C – control voltage, P_C– compressor pressure, P_m– muscle pressure, κ - muscle displacement)

To evaluate the dynamics of EPR50 pressure regulator, two steps of different height used as a pressure reference were used (300 kPa and 500 kPa). The regulator uses 0-10 control voltage range for controlling the output pressure from 0 to 700 kPa. Thus the used control voltage U_C was related to the desired muscle pressure P_{Md}in the following way:

$$U_C = \frac{10}{700} P_{Md} = 0.01429 P_{Md} \tag{3}$$

where P_{Md} – desired muscle pressure in kPa.

The responses to both steps are shown in Fig.8. These responses were filtered using moving average filter with 10 points for averaging. System Identification Toolbox was used for identifying the first-order process models using these responses.

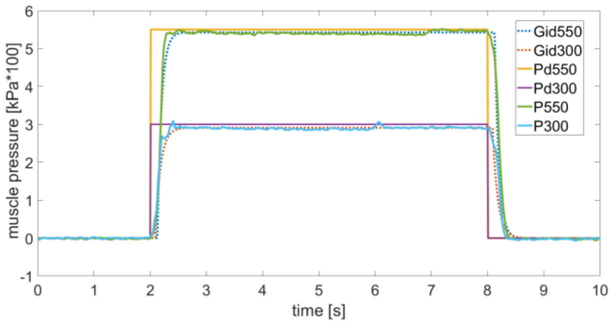

Figure 8. Dynamic response of pressure regulator to 300 kPa and 550 kPa pressure reference steps (Pd300 – 300 kPa pressure reference, Pd550 – 550 kPa pressure reference, P550 – actual response of the regulator to 550 kPa step, P300 – actual response of the regulator to 300 kPa step, Gid550 – identified transfer function for 550 kPa step, Gid300 – identified transfer function for 300 kPa step)

The following models were obtained:

$$G_{id\,300}(s) = \frac{0.98641}{0.073413s+1}e^{-0.11776s}$$
$$G_{id\,550}(s) = \frac{0.97058}{0.092507s+1}e^{-0.07916s}$$
(4)

Brief inspection of these models implies that the dynamics of regulator is fast with time constants around 0.09 and 0.07 seconds and time delay approximately 0.11 seconds and 0.08 seconds for 550 kPa and 300 kPa steps, respectively. We considered these values small enough to neglect the dynamics of pressure regulator in the preliminaryinvestigation of fluidic muscle hysteresis under dynamic conditions. The system gain (slightly under 1 in both cases) corresponds to a situation when desired muscle pressure in taken as input.

For investigation ofdynamic hysteresis in fluidic muscle, we used a triangular excitation signal with linearly decreasing amplitude shown in Fig.9. As shown in the figure, the muscle pressure was in direct proportion to control voltage signal with very small overall distortion. The amplitude of muscle pressure waveform was reduced from 500 kPa in t = 5 s to 100 kPa in t = 85 s (U_C reduced from 7.145 V to 1.429 V).

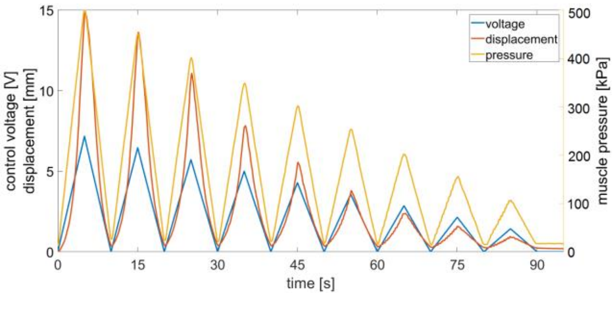

Figure 9. Dynamic responses of muscle pressure and displacement to a linearly decreasing triangular excitation waveform with frequency of 0.1 Hz

On the other hand, more complex relationship is revealed between P_m and κ, where a non-uniform dynamic pattern of muscle displacement in response to uniform pressure pattern can be observed. Their relationship can be better evaluated using P_m-κ plane, which is shown in Fig.10. Linearly decreasing excitation signal is particularly useful for investigating the actual form of hysteresis, especially with regard to major or minor loop formation. The right ends of each of the loops correspond to desired muscle pressures at 50 kPa decrements from 500 kPa, which results from using the electronic pressure regulator. The width of given loop is directly proportional to the change of pressure in the muscle, with largest width being associated with the first loop (5.19 mm). Even larger width could be expected for the first loop in case the maximum operational pressure of DMSP-5 was used (600 kPa). In addition to a hysteretic nature of P_m-κ relationship, it is interesting to observe previously mentioned nonlinear dependence of displacement on muscle pressure in terms of their extremal values. This actually determines also the y-axis range of respective loops formed during linearly decreasing excitation. It appears that it is the consequence of specific properties of fluidic muscle as an elastic actuator.

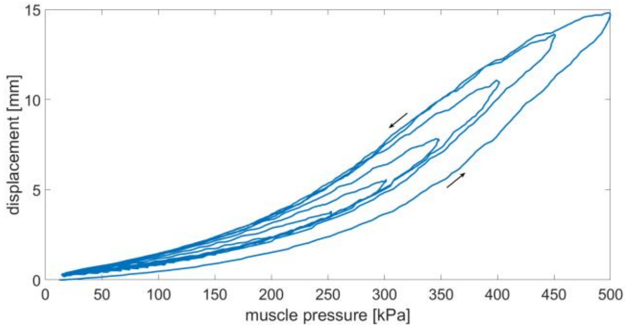

Figure 10. Hysteresis of P_m-κ relationship for a linearly decreasing triangular signal with frequency of 0.1 Hz

In the second experiment, we used the same linearly decreasing triangular waveform but this time with the frequency of 0.2 Hz (Fig.11). It can be easily observed that the nature of hysteresis curve was not significantly changed. The maximum width of hysteresis corresponding to the major loop was 5.07 mm (0.12 mm lower than for 0.1 Hz). Closer examination reveals that there are two main differences when compared to the case with 0.1 Hz (Fig.12). The actual pressure in the muscle was lower than the desired pressure due to the pressure regulator not being able to control the pressure fast enough. The distortion of the loops was caused by the oscillations of attached mass when using higher frequency reference signal.

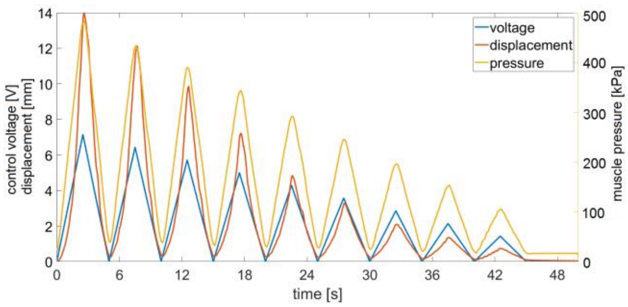

Figure 11. Dynamic responses of muscle pressure and displacement to a linearly decreasing triangular excitation waveform with frequency of 0.2 Hz

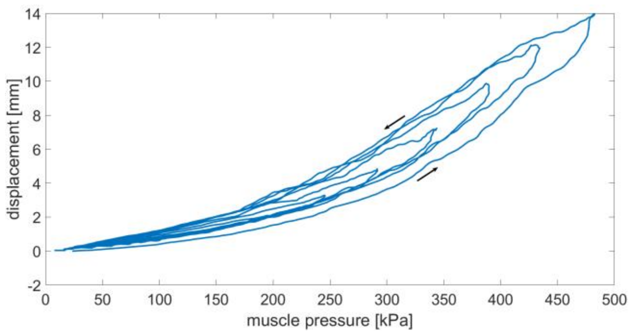

Figure 12. Hysteresis of P_m-κ relationship for a linearly decreasing triangular signal with frequency of 0.2 Hz

The effect of doubling the frequency of excitation signal can be better estimated using Fig.13 where both hystereses are overlaid. The right ends of minor loops reach lower values of muscle pressure due to the above mentioned reason of pressure control. The hysteresis curve appear to be rotated clockwise around the point of minimal pressure for a higher frequency. This is in accordance with the observed nature of its hysteresis most likely associated with the elasticity of muscle material, which manifests itself in tendency of the muscle to contract slowly when pressure is held constant. In addition to that, it can be seen that this effect apparently depends on the magnitude of muscle pressure since it is relatively insignificant for pressures in the range of 0-100 kPa and much more pronounced for pressures in the range of 300-500 kPa.

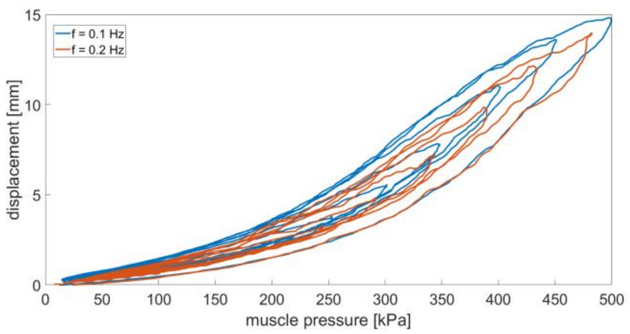

Figure 13. Direct comparison of P_m-κ hysteresis for two frequencies of excitation signal

To test the evolution of hysteresis when excited with arbitrary signal we used a triangular signal with randomly selected peak values of control voltage (i.e. desired muscle pressure). Its waveform as well as the responses of muscle pressure and muscle displacement can be seen in Fig.14. This signal was obtained by modifying the distribution of peak values of 0.2 Hz signal over y-axis. Hysteresis curve formed in response to this signal is shown in Fig.15 showing a complex character with multiple crossings.

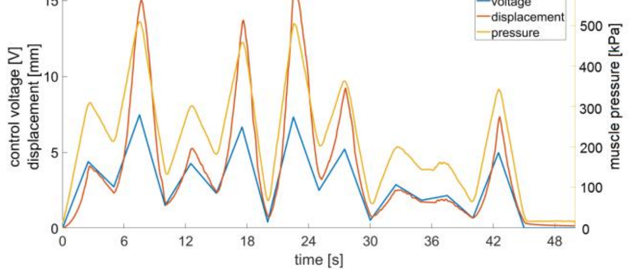

Figure 14. Dynamic responses of muscle pressure and displacement to a random triangular excitation waveform

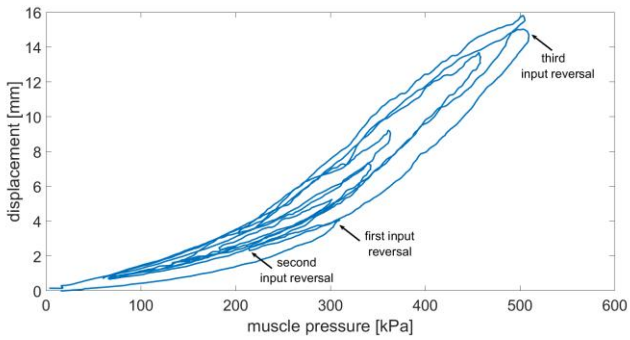

Figure 15. Hysteresis of P_m-κ relationship for a random triangular excitation waveform

The figure also shows the first three input reversals occurring at 2.5, 5 and 7.5 seconds which indicate the process of evolution of hysteresis curve. This curve helps to evaluate the non-local character of hysteresis of fluidic muscles since it is clear that there are points which are not uniquely associated with only one state of the system and future evolution of the hysteresis from that point is determined also by the previous input history.

In the last test of our preliminary investigation of DMSP-5 hysteresis we carried out a quasi-static measurement using a stair signal with 50 second period. The excitation signal from a stair generator approximated the waveform of triangular signal with linearly decreasing amplitude similar to that used in previous tests. The period of 50 seconds was selected based on the previous experiments to achieve quasi-steady state with only minimal changes of displacement. The resolution of excitation signal amplitude was selected to be 0.25 V which corresponded to 217 levels of amplitude to achieve reduction from 500 kPa to 100 kPa with 50 kPa decrements. This number of levels was also equal to the number of samples we selected from the whole signal after experiment – we selected each 4900-th sample from the resulting 1080000 samples. The results of this resampling for both the excitation signal and the responses can be observed in Fig.16.

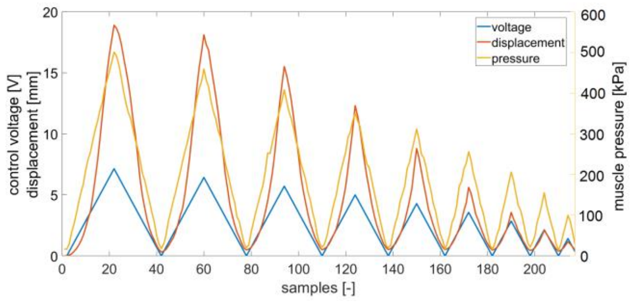

Figure 16. Quasi-static responses of muscle pressure and displacement to a stair signal approximating the triangular excitation signal with linearly decreasing amplitude

The reason for choosing 4900-th sample was that it was close to the end of stair signal period so that the muscle had enough time to attain the steady-state. Of the note is non-smooth character of muscle pressure response that was caused by the closed-loop control of this variable where the actual value could be changed momentarily as a result of controller counteracting the effect of disturbances. This is clearly visible in t = 85 s, where a tooth in pressure response was formed due to the above mentioned reasons.

In accordance with previous depictions, the hysteresis of P_m-κ relationship is shown in Fig.17. Despite the non-smooth nature of this hysteresis (due to the reasons mentioned before), the formation of a major and minor loops is comparable to previous dynamic tests. The maximum width of the hysteresis determined for the major loop was 5.26 mm, which is 0.05 mm higher than for 0.1 Hz and 0.19 mm higher than for 0.2 Hz. These results are summarized in Tab.1.

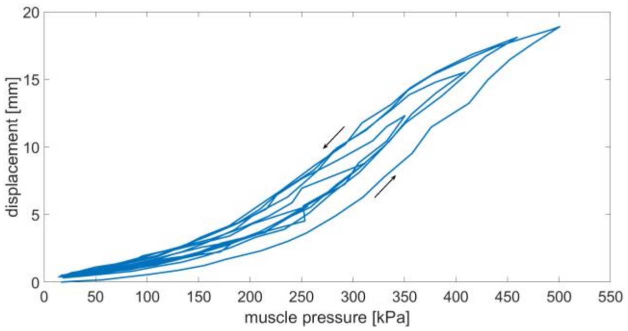

Figure 17. Quasi-static hysteresis of P_m-κ relationship for a stair signal approximating the triangular excitation signal with linearly decreasing amplitude

The results shown in Tab.1 supports the observation that the increasing frequency of excitation signal reduces (although only slightly) the width of hysteresis. On the other hand, it is easily that this hysteresis is not rate-dependent in the sense of being a simple dynamic lag, which disappears when frequency goes to zero [Bertotti 1998]. Quite the contrary, the width of hysteresis was increased for quasi-static measurement which could be associated with the observed effect of slowly increasing contraction of the muscle when pressure was kept constant.

Frequency, f [Hz]	Maximum hysteresis width for major loop [mm]
0	5.26
0.1	5.19
0.2	5.07

Table 1. Maximum width of hysteresis for the major loop for three ferquencies (0 Hz is used to indicate the quasi-static measurement)

5 CONCLUSIONS

We presented a preliminary investigation of hysteresis ofP_m-κ relationshipDMSP-5 fluidic muscle. Special attention was paid to determine the nature of this hysteresis, especially with regard to its non-locality as well as the confirmation of its rate-independence. It was observed that the hysteresis is mainly caused by the apparent elasticity of rubber tube material, which resulted in increasing contraction of the muscle for constant pressure with the effect being more pronounced for higher values of pressures. In accordance with this, the measured maximum widths for given frequencies were 5.26 mm, 5.19 mm and 5.07 mm. By using a random triangular excitation signal, we could observe the complex character of hysteresis curve with a number of crossings which attest to non-locality of this hysteresis where a given point in P_m-κ is associated with possibly infinitely many further paths determined by the previous input history.

The investigation of hysteresis of pressure-displacement relationship in PAMs is important for high-performance position control since in that case the hysteresis should be compensated by using its inverse model. There are, of course, many hysteresis models each with its pros and cons which might be important with regard to the intended application. The results supports the need for more complex models which can be used for hystereses with non-local memory (Preisach, Bouc-Wen, Prandtl-Ishlinskii, expanded input space model and others). In future work it will be needed to select one of these models to construct an inverse model of hysteresis usable for its compensation within the structure of model-based (possibly) nonlinear position control of fluidic muscle.

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