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Design of an electrohydraulic hand orthosis for people with Duchenne muscular dystrophy using commercially available components

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Abstract—People with Duchenne muscular dystrophy are currently in need of assistive robotics to improve their hand function and have a better quality of life. However, none of the available active hand orthoses is able to address to their specific needs. In this study, the use of hydraulic technology is proposed in the design of an active hand orthosis. Commercially available components were used to identify where customization is necessary for a new electrohydraulic hand orthosis. The presented prototype was able to move four finger modules with a single actuator. The finger modules were separable and had a total mass of only 150 g, whereas the valve manifold added another 250 g. Results revealed that the prototype was able to function well with full flexion/extension cycles up to 2 Hz, but with hysteretic losses between 37–81% of the total input energy. Specialized valves and slave cylinders are required to increase efficiency at higher speeds and to obtain more robust sealing performance.

I. INTRODUCTION

Duchenne muscular dystrophy (DMD) is an X chromosome-linked progressive neuromuscular disease, resulting in disability and shortened life expectancy [1]. It is the most common and severe form of muscular dystrophy [2]. Recent advances in technology and medicine rapidly increased the life expectancy of people with DMD [3]. However, their hand function is rather limited [4]. Therefore, in order to have a better quality of life, they find themselves in need of assistive robotics [5].

The importance of the hand function is highlighted by the great amount of orthotic devices developed in the last decades [6]. In contrast, at the moment, the only existing devices to support the hand function of people with DMD are static hand splints [7]. Their aim is to postpone the development of contractures and maintain a satisfactory range of motion [7]. Evidence suggests that people with DMD can greatly benefit from the use of orthotics [8], even

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passive [7]. Nevertheless, those are not sufficient to assist in activities of daily living (ADLs).

As with any orthosis for daily assistance, it should be small, light-weight, comfortable [9] and provide enough force to support during ADLs. Creating a small and light-weight orthosis has proven to be challenging considering the small amount of design space that is available on the human hand. For people with DMD in particular, the interaction forces should not be too high due to an increased sensitive skin, and the hand orthosis needs to be donned onto the fingers one by one to more easily accommodate contracted fingers. Due to these factors, none of the existing devices meets the specific needs of people with DMD, hence a different approach is required in designing a dynamic hand orthosis.

The goal of this paper is to present the design of a hand orthosis that is actuated by an electric motor and uses miniature hydraulics as a method to transfer mechanical work. In particular, a prototype of the design is made as a platform to examine the use of commercially available components in terms of pressure and speed limitations. This helps to identify the bottlenecks of such a system and research focus for future iterations.

II. MINIATURE HYDRAULICS

In this study, the use of miniature hydraulics was explored for an active hand orthosis for people with DMD. The concept of using hydraulics to provide force and motion is certainly not new. Examples can be found at large scale in heavy machinery in construction (e.g., excavators, cranes). Also at micro-scale, hydraulics is being used in lab-on-a-chip systems. Between those scales, where hydraulic components are sized in the range of several millimeters, it becomes more feasible for orthotic applications. As long as this miniaturization is coupled with an increase in system pressure, hydraulic systems can be more compact and light-weight than an electromagnetic equivalent [10]. There are not many other applications that operate on this scale and therefore components are hard to find. Nonetheless, it is useful to find the limitations on what is currently commercially available and which components are hindering further miniaturization of hydraulics.

A hydraulic system can be largely characterized by the type of actuator that is used. Following the classification in [11], fluidic actuators can be either elastic or inelastic. Within the elastic class, McKibben-type actuators are easiest to obtain, whereas piston-cylinders with contact seals are

the most widely available inelastic actuators. McKibben-type actuators can provide the highest forces relative to their cross-sectional area (i.e., actuator stress [12]). However, as is the case with all elastic actuators, available force from the actuator decreases as stroke is increased [13] and system pressure is limited by the material [14]. This is not the case with piston-cylinders, which can provide the maximum rated force over the full range of stroke [15]. Moreover, the available stroke relative to initial length (i.e., actuator strain [12]) is higher for piston-cylinders. Based on these observations, piston-cylinders were chosen as the most suitable type of actuator.

Hydraulic components can be connected with each other using flexible hoses, effectively providing a flexible transmission of mechanical energy. This makes it possible to place heavier equipment (e.g., pump, energy storage) away from the hand and, for example, fix it to the wheelchair. In contrary to a Bowden cable system, a hydraulic transmission can provide a more transparent force efficiency that is independent of any bends in the hose [16]. The only losses in efficiency are related to wall friction within the hoses and small added friction coefficients due to smooth bends [17]. These were assumed negligible compared to O-ring friction in the actuators.

Only a few examples can be found where a hydraulic system was used in a hand orthosis [13], [14], but these systems use elastic actuators that limit the available force for larger joint angles and maximum system pressure. As a related application, more examples can also be found in hand prosthetics [18], [19], [20]. Similar to orthotics, reported arguments are generally aimed towards minimization of added volume and mass—one of them resulting in the most light-weight hand prosthesis [20]. All examples, however, rely on customized components and do not provide an insight on their accessibility compared to, for example, electromagnetic systems.

III. DESIGN CRITERIA

The design criteria for the presented hand orthosis were similar to a mechanism that was designed in a previous study by Bos et al. [21]. Specifically, the number of degrees of freedom, size of the design domain and maximum resultant force on the skin were kept the same. In this design, the thumb was not included but assumed to be fixed in opposition. Other criteria were adjusted and added in order to accommodate this particular target group. A summary of the criteria is given in Table I. In this section they are elaborated in more detail.

The main objective of a hand orthosis that provides assistance during ADLs is to provide sufficient grasping force and range of motion. In Kargov et al. [22], the highest joint moment in the human hand was measured at 0.1 Nm in order to grasp a bottle of approximately 0.5 kg. In order to be able to include a larger variety of graspable objects and levels of skin hydration (which affect attainable friction forces with the skin [23]), a value of twice this magnitude was chosen, namely 0.2 Nm per joint. To accommodate movement, an

TABLE I
LIST OF DESIGN CRITERIA FOR THE HAND ORTHOSIS.

Criterion	Description [metric]	Implementation
Grasping force	Joint moment [Nm]	0.2 Nm per joint
Range of motion	Flexion angle range [°]	20–80° per joint
Degrees of freedom	Functional degrees of freedom [#]	1
Design domain	Cross-section [mm×mm]	26×17 mm
Comfort	Resultant force on skin [N]	<5 N
	Shear force [N]	0 N
Mass	Mass on hand [kg]	<200 g
Speed	Cycle frequency [Hz]	2 Hz
Wearability	Easy donning/doffing	Separable finger modules

average range of finger flexion angles of 20–80° were chosen which relate to a functional range of motion [24].

Only forces normal to the fingers' skin surface contribute to an increase in grasping force. Shear forces, on the other hand, increase the resultant force and the related risk of skin tissue damage, while they do not contribute to grasping force. This means that shear forces should be avoided at all costs.

The orthosis should not interfere with the hand too much. This is reflected in the criteria for range of motion, but should also include a limit on the perceived mass. Similar systems mention a mass limit around 500 g [25], [14] that is placed on the hand, where heavier components can be placed on more proximal locations or even attached to the wheelchair. For people with DMD, however, it is believed that this value should be even lower because any additional weight will cause more strain on already weakened muscles or on the potential arm supports. Hence, a mass limit of 200 g was imposed on this design. Other factors that may increase the perceived mass are stiff elements that connect the hand module with the heavier components. We aimed to keep this added stiffness minimal.

The speed at which the hand orthosis is able to move, should not be a limiting factor to achieve natural human movement and to be able to respond to fast changes in intention. Based on speeds that occur in the hand while performing ADLs, cycle frequencies between 0.5-1.6 Hz have been reported in the design of hand orthoses and prosthesis [26], [14]. Additionally, a recent study with one person with DMD, implies that people with DMD can process signals and respond successfully with finger movements no faster than 2 Hz [27]. Hence, a full cycle frequency of 2 Hz was deemed sufficient and chosen as criterion.

One of the complications regarding people with DMD, is the occurrence of contractures [28]. This can make it difficult and even painful for these individuals to don glove-like systems, which requires all fingers to be stretched simultaneously. Therefore, in order to improve wearability of the hand orthosis, easy donning/doffing should be facilitated by incorporating separate finger modules, allowing each finger to be fitted one-by-one.

IV. DESIGN DESCRIPTION

An illustration of a single finger module of the hand orthosis is shown in Fig. 1.

A. Hydraulic system

1) *Slave actuators*: To the authors' knowledge, there is no commercially available hydraulic piston-cylinder that has a diameter of $< \text{Ø}8$ mm and states to be able to work with pressures of >2 MPa. Nonetheless, a line of small pneumatic cylinders from Festo were used. They are rated at 0.7 MPa for pneumatic use, but their ability to work with water and much higher pressures were tested. In particular, to maximize available actuator strain, $\text{Ø}4$ mm cylinders with 20 mm stroke were used (EG-4-20-PK-2, Festo).

2) *Medium*: Because a hand orthosis requires a close interaction with a human operator, the medium should not cause damage to the skin or stains on fabric in case of any leakage. Therefore, water was chosen as medium. It is a neutral fluid and has been used in other miniature hydraulic systems as well [13], [20].

3) *Transmission*: The flexibility of the hydraulic transmission can have a large influence on the risk of leakages and perceived mass of the hand orthosis. Low-pressure hoses are generally more flexible than high-pressure hoses and available in the smallest diameters. Using them at higher pressures, however, can cause them to expand, which decreases the bulk stiffness and increases hysteresis of the hydraulic system. Maximizing flexibility was considered to be more important than overall system stiffness and hysteresis. For this reason, a standard polyurethane hose that was compatible with the fittings on the slave actuators was used (PUN-H-3x0.5, Festo).

4) *Circuit*: To minimize the added stiffness from the hydraulic transmission between the heavier components and the hand orthosis, it was facilitated with only one hose. This way, only one master cylinder could be used and all supported degrees of freedom of the hand orthosis were underactuated. For each finger, this means that slave cylinders will curl around any object's shape. Across the fingers, the hand can adapt to three-dimensional shapes (e.g., spheres) as well [29], [20]. The overall schematic of the hydraulic system is shown in Fig. 2, which is very similar to the one shown in [20]. The only added feature is a set of electrically operated valves that allow to control which fingers to move and which to block—an approach which can also be seen in Kargov et al. [18].

5) *Valves*: In order to support multiple fingers and be able to switch between different grasp types, a hydraulic hand orthosis should also contain valves that are able to facilitate this functionality. As displayed in Fig. 2, each valve can block or permit the flow of pressurized fluid so that specific groups of slave actuators can be selected to move. In the presented design, 2/2-way on/off normally-closed valves were chosen. The smallest available valves with highest rated pressure from Festo were explored. Because it used a smaller nozzle diameter of 0.65 mm, a 3/2-way miniature valve with blocked exhaust provided the 2/2-way function at a rated pressure of 0.8 MPa (MHA1-M1H-3/2G, Festo).

B. Hand orthosis mechanism

1) *Modular approach*: In order to accommodate the specific needs of people with DMD, the hand orthosis should allow each finger to be fitted individually. The weaker hand muscles combined with possible contractures make it very difficult to align all fingers properly at the same time. This restricts the use of any glove-like orthosis. As an alternative, we chose to design separate finger modules that can be donned one-by-one.

2) *Compliant mechanism*: To minimize the shear forces, a compliant mechanism was developed with the purpose of absorbing shear forces. More specifically, the parts that interface with the metacarpals and phalanges of each finger were connected by flexure elements. These flexures should be strong enough in tensile direction to absorb the shear forces, but should also accommodate the joint motion with minimal added stiffness and allow small joint misalignments. This part was made by 3D-printing polyamide (PA 2200) using selective laser sintering (SLS). Due to the strength of the material, the flexures were implemented with a zig-zag-like shape in order to reduce overall bending stiffness. Additionally, this also allowed for slight elongations of the flexures, making the interface compliant to disturbances.

3) *Mechanical transmission*: The total range of motion for each finger joint that can be achieved by direct actuation with a linear actuator is limited to a certain amount. In Fig. 3, it can be seen that the desired moment arm (r) and range of motion (θ) affect the (x, y)-placement of the proximal connection (A) of the actuator. Moving the distal connection (B) as proximal as possible is desirable, because this maximizes the attainable range of motion and avoids the actuator rod to hit the skin. However, this placement is limited by the slave cylinder's initial length (L_0) and possible interference with the second cylinder. Especially due to these factors, placement was also a practical consideration, where a moment arm of $r = 20$ mm was used for each slave cylinder and the dorsal distance from the joint center was limited to $y = 30$ mm. In combination with the chosen slave cylinders, this allowed for a theoretical flexion angle range of $10\text{--}80^\circ$.

Only the two proximal joints of the finger (i.e., metacarpophalangeal and proximal interphalangeal joint) were supported in flexion/extension. Using results from robotic grasping, two phalanges are sufficient to reach a stable grasping position [30]. The most distal joint (distal interphalangeal joint) was only protected against overextension by extending the orthosis interface (see Fig. 1).

V. DESIGN CHARACTERIZATION

A. Prototype

A photo of the resulting prototype in several poses is shown in Fig. 4. For purposes of characterizing the hand orthosis, it was fitted onto mock-up fingers. They were manufactured by 3D-printing polyamide (PA 2200) and showed an anthropomorphic shape. Each joint contained an integrated leaf spring of the same material which added a small but non-linear dummy stiffness, whose resting position

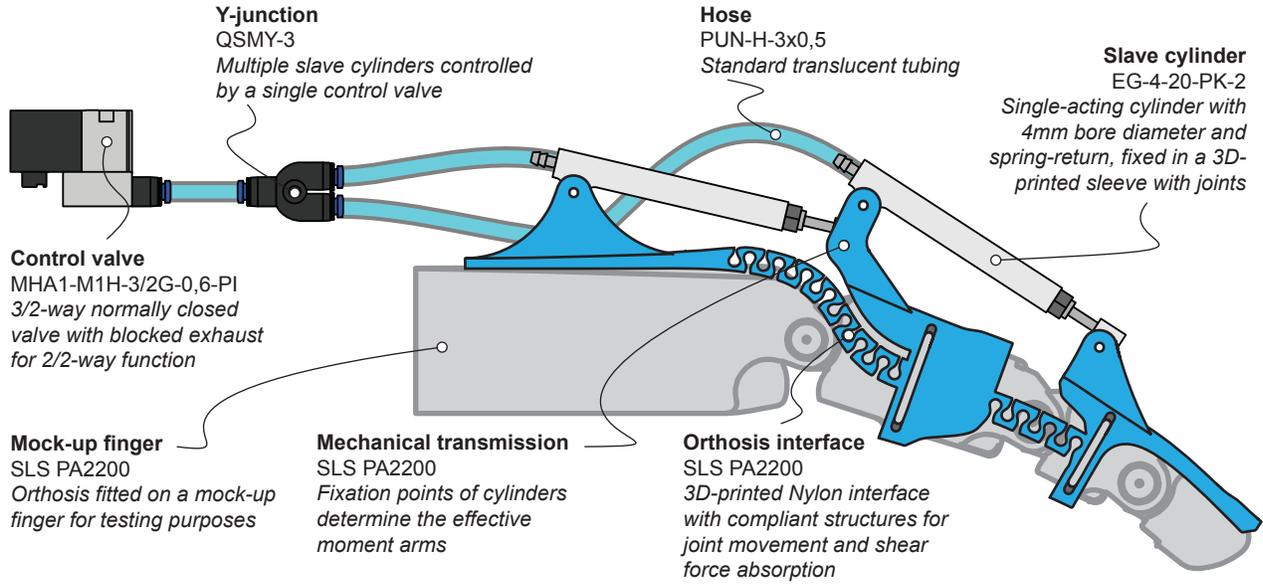


Fig. 1. Illustration of a single finger module of the hand orthosis.

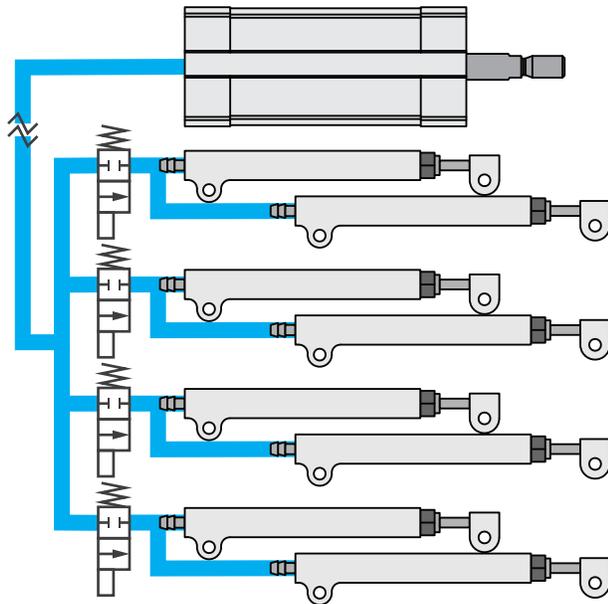


Fig. 2. Schematic of the hydraulic system, where one master cylinder underactuates several slave cylinders. Each finger is equipped with two slave cylinders, whose movement can be blocked or allowed by the 2/2-way valves.

was in a slightly flexed position of 20° . The added stiffness was characterized with a linear stiffness of approximately 0.01 Nm/rad around its resting position.

Each metacarpophalangeal joint expressed an active range of motion of $30\text{--}65^\circ$ flexion, for each proximal interphalangeal joint this was $10\text{--}45^\circ$. The amount of finger elements involved in the (grasping) motion could be selected with on/off valves and, for example, allowed it to switch from a power grasp to pointing with the index finger (see Fig. 4).

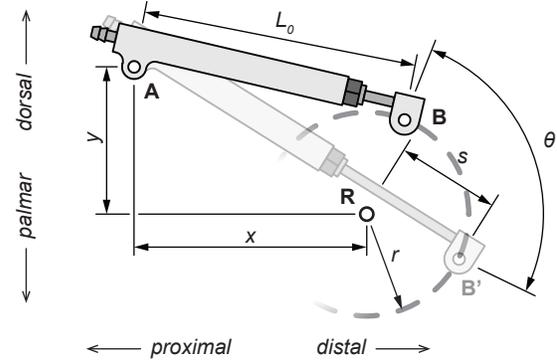


Fig. 3. Illustration showing the relation between the slave actuator's stroke (s), initial length (L_0) and placement of the proximal connection (x, y) with the moment arm (r) and resulting joint range of motion (θ) around the finger joint (R).

The maximum occupied cross-section on top of each finger was $33 \times 19 \text{ mm}$ (height \times width). Each finger module could be donned/doffed separately, where the mass on the hand for all four modules was measured at 150 g . The valve manifold added a mass of 250 g . The grasping force and interaction forces were not measured in this study.

B. Dynamic behavior

To evaluate the dynamic behavior of the system, a large electric motor (AKM22C-BNCNC-00, Kollmorgen) with spindle drive was used to move the master cylinder. The master cylinder was subjected to full stroke cycles at different frequencies and the resulting system pressure was recorded with a pressure sensor (3500-B-0040-A-01-B-0-00RS, Gems Sensors & Controls). The range of frequencies were chosen to be between $0.5\text{--}5 \text{ Hz}$ with steps of 0.5 Hz , well below and above the required full cycle frequency of 2 Hz . All

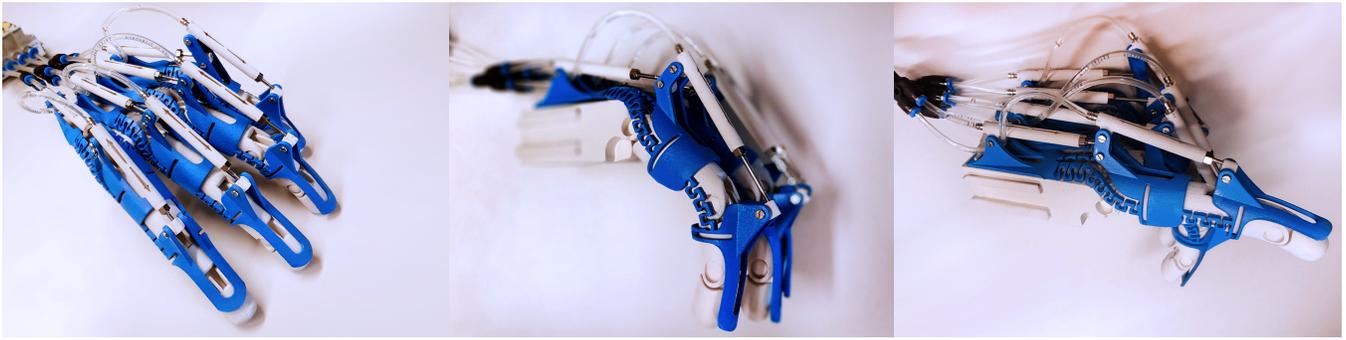


Fig. 4. Photos of the prototype fitted on a mock-up left hand with, from left to right, fully extended position, fully flexed position and pointing with index finger.

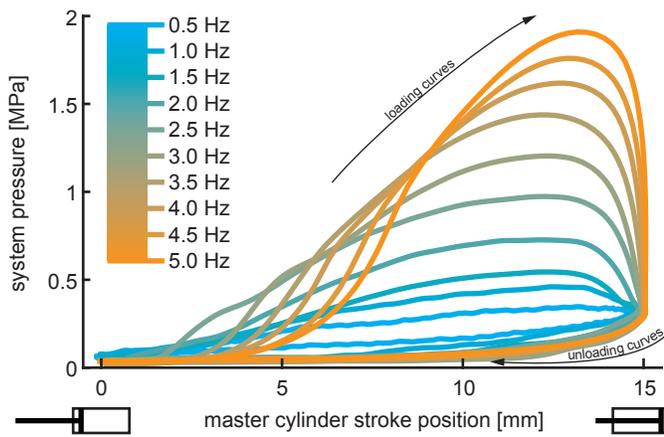


Fig. 5. Loading and unloading curves of the system pressure versus master cylinder stroke. Different full cycle frequencies were imposed at the master cylinder and affect the measured system pressure.

valves were opened such that all slave cylinders and mock-up fingers moved along, imposing the maximum load for free movement (i.e., no grasping).

Fig. 5 shows the measured system pressures as a function of the master cylinder stroke. Each curve represents the average of 10 cycles during steady-state operation. Due to the return springs in the single-acting cylinders and stiffness in the joints of the mock-up fingers, the curves show a distinguished loading (i.e., fingers flexing) and unloading (i.e., fingers moving back to resting position) curve. The peak value of the loading curve clearly increases with increasing frequency, whereas the unloading curve shows a lower limit due to the passive elements. After approximately 2.5 Hz the shape of the load curves slightly changes, where pressures below atmospheric pressure were measured and an overall delayed increase towards the peak pressure can be observed.

Fig. 6 shows energy values that correspond to the tested frequencies. The energy expenditure of the system was estimated by integrating the measured pressure over displaced volume ($E = \int p dV$), where the loading curves determined the input energy and the unloading curves the output energy. The difference between these values was determined as the hysteresis for one full cycle. In these results it can

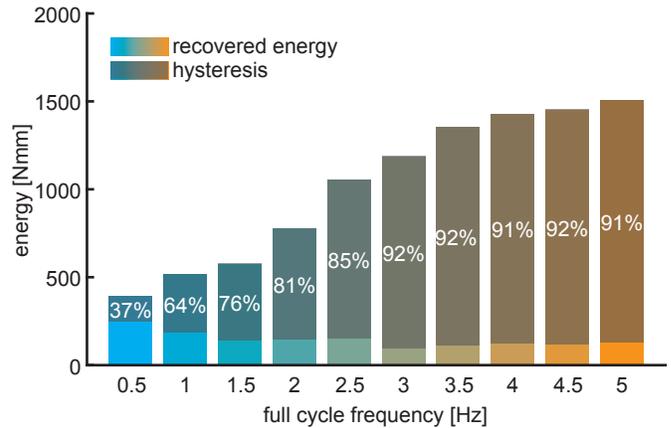


Fig. 6. Calculated energy values for each tested full cycle frequency. The total height indicates the required input energy to fully load the system (reaching maximum finger flexion). After unloading, a percentage (indicated in white) is lost due to hysteresis and the rest is recovered from the passive elements.

be seen that an increase in speed comes at the cost of increased hysteresis and total input energy. The system is most sensitive to changes in speed around 2 Hz, which is also reflected by the altered shapes of the pressure curves in Fig. 5 and implies a change in dynamic behavior. At 2 Hz, both relative hysteresis and total input energy have doubled in magnitude compared to the results at 0.5 Hz.

C. Components

The components of the hydraulic system that were used in the design showed high resilience to increased system pressures. A static measurement revealed that the valves were no longer able to block fluid flow at pressures above 1.5 MPa. The slave cylinders showed no signs of leakage at pressures up to 2 MPa. With closed valves, a decrease in bulk stiffness was observed for increased system pressure, which imply radial expansion of the hose material. At full cycle frequencies higher than 2.5 Hz, a substantial increase in air bubble formation was observed in the hydraulic hoses and coincide with the observed changes in dynamic behavior. Additionally, after one month, one of the valves showed signs

of corrosion inside and needed to be cleaned in order to function properly.

VI. DISCUSSION

The valves were the main limiting factor in terms of speed, pressure and mass. They contained the smallest orifices in the system with 0.65 mm in which fluid speeds can rapidly increase, making turbulent flow possible after 1 Hz in this design. This caused fluid friction to increase in magnitude and can be seen by the fast increase in hysteresis for increasing cycle frequencies. The Venturi effect caused local pressures to drop below the medium's vapor pressure and cavitation bubbles were clearly visible. Additionally, the valves had the lowest maximum pressure and the valve manifold added a mass which was almost twice as much as all the finger modules. Combined with the observed corrosion, different valves are required with increased water compatibility, larger nozzle diameter, higher pressure limit and lower mass.

The slave cylinders were able to accommodate water at higher pressures and did not add much volume and weight to the orthosis. Their initial length, however, required the fixation points to be stacked on top of each other. This caused the mechanism to have larger protrusions and have a lower margin on the range of motion to make up for relative movement of the compliant interface. Moreover, it did not allow for much variations in moment arms, which can provide more human-like force distributions [22] and increased grasping stability [30]. The U-cup sealing elements in the slave cylinders were not able to seal against vacuum pressures, which occurred when the master cylinder was retracting faster than the return springs could follow. Custom slave cylinders can improve the design when they have a higher actuator strain, stronger return springs and more robust sealing elements.

A substantial number of air bubbles were visible in the system at higher cycle frequencies and affected the dynamic behavior of the hand orthosis. This is due to the combination of pressure drops below vapor pressure inside the valves and air being sucked in at the slave cylinders' sealing elements. The compression and expansion of the air bubbles slightly increased the amount of recovered energy and the added volume caused the pressure at rest to increase. More importantly, the added compliancy delayed the system pressure's response to the master cylinder's position input. At the highest frequencies, it attenuated the motion of the slave cylinders—explaining the lower increase rate in required input energy. Vacuum treatment of the water or using a different medium with lower vapor pressure can decrease these effects. Alternatively, the speed of the master cylinder needs to be limited according to the system's capabilities.

Even at the lowest frequency, the total efficiency is quite low compared to a previous hydraulic master-slave system [16]. Naturally, the mechanism and mock-up hand add more losses to the system, but expansion in the hoses' material also add hysteresis to the system. Hoses with higher radial stiffness can alleviate this effect. In its intended application, this will become more important as system pressures will be

higher when objects need to be grasped and more realistic joint stiffnesses are present—including the possibility of increased stiffness due to muscle contractions. Human testing is required to validate this increase in pressure, but was considered out of scope of this study.

The presented design lacks an integrated energy storage and pump that controls the master cylinder. For a fully functional and controllable electrohydraulic hand orthosis, these components are indispensable as well and need to be optimized for minimal volume occupation. It is also possible to replace the master cylinder with a different fixed displacement pump (e.g., gear pump), but this would also require the addition of a reservoir with return line or additional valve.

To further decrease weight and increase portability, less finger modules can be used. For example, depending on the wearer's situation, supporting only the index and middle finger can already suffice in improving the ability to grasp some daily objects [31]. However, supporting as many fingers as possible also encourages more finger movement and can possibly retard the development of contractures. It is unclear whether such an increased portability can outweigh this potential rehabilitative effect and requires further research. Either way, with the modular approach of the presented design it is relatively easy to attach or detach finger modules.

VII. CONCLUSION

This study has shown that it is possible to create an electrohydraulic hand orthosis using commercially available hardware components, with the only exception being the 3D-printed structure which was custom-printed by a 3D-printing company. The prototype underactuated all flexion/extension movements with a single functional degree of freedom and consisted of separable finger modules. It was able to operate within the 2 Hz limit, but came at the cost of a large increase in energy losses. The mass was exceeded due to a heavy valve manifold and the desired range of motion and design domain were not reached due to low actuator strains in the slave cylinders. Evaluating the feasibility of the design and the individual components revealed possible bottlenecks and further improvements, indicating that the used methodology may still be a feasible solution for people with DMD. Specifically, specialized valves can decrease hysteresis, increase attainable system pressures and decrease mass; custom slave cylinders can reduce actuator strain, reduce occupied volume and improve sealing performance; and, stiffer hoses can reduce bulk stiffness and hysteresis, which can increase the system's overall efficiency.

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