ARTIFICIAL MUSCLES TECHNOLOGIES AND DESIGNS FOR A COUNTERMEASURE BODY SUIT

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ABSTRACT

The purpose of the on-going project DYNASUIT is to design a new intravehicular body suit for countermeasure purposes, including state of the art technologies in artificial muscles, body monitoring and biofeedback.
This paper focuses on the concepts, designs and testing of the artificial muscles subsystem. Based on a preliminary analysis and a selection of the most promising scenarios, two actuation technologies have been selected for deeper analysis in the project, namely electro-active polymers (EAP) and sigma pneumatic actuators. The paper describes the design and the expected performances of each of the actuation technologies in the suit. The physical demonstrators built during the activity in order to validate the designs and characteristics of the actuators in the context of DYNASUIT, as well as the associated results, are presented. These outputs will be used in the following of the activity to support the detailed design of the full countermeasure suit.

1. INTRODUCTION

During spaceflight, the human body is subject to physiological adaptation to the microgravity conditions. Although some effects, like vestibular disorders, only lead to temporary discomfort for some days, other reactions such as bone mineral loss or muscle atrophy impact the physical condition of astronauts in proportion to the time spent in space. This is an issue for long-duration missions such as those onboard the ISS and for future planetary exploration on the Moon and Mars. The lack of effectiveness of current countermeasure solutions and the longer exposure to microgravity for new missions require a more thorough understanding of the stakes and issues and the development of new solutions.

In this context, the ongoing ESA project DYNASUIT aims at designing a new generation of intravehicular body suit for countermeasure purposes, based on new available artificial muscles technologies, body monitoring sensors and biofeedback control strategies. This paper focuses on the concepts, designs and testing of the artificial muscles subsystem. Based on a preliminary analysis and a selection of the most promising scenarios, two actuation technologies have been selected for deeper analysis in the project, namely electro-active polymers (EAP) and sigma pneumatic actuators.

After a review of the physiological effects of weightlessness and the countermeasure scenarios, the paper describes the design and the expected performances of each of the actuation technologies in the suit. The physical demonstrators built during the activity in order to validate the muscles designs and characteristics in the context of DYNASUIT, and the associated results, are then presented.

2. PHYSIOLOGICAL EFFECTS OF WEIGHTLESSNESS ON THE BODY

The human body is capable of adaptation to environmental changes. This capability to adapt is such that the physical structure and function of many of the body’s tissues, organs and systems alter to enable life to proceed in microgravity in an efficient and economical manner. The primary systems affected are the skeletal, muscular, neurovestibular, cardiovascular, endocrine, immune and motor control systems. Adaptations occur in hours to days for some systems but can take weeks to months or even longer for others. For example the neurovestibular system will typically adapt over 3 days with symptoms ceasing by day 5 for most people, whereas the musculoskeletal system will adapt continuously for months depending on the stimulation experienced, with bone potentially still deconditioning after a year or more.

These changes, seen over hours, days, weeks and months, are a positive response to the space environment, in particular to the absence of gravity. However, they are problematic if the gravity vector is re-imposed, when coming back to Earth, during planetary excursions (increasing the risk of injury), or when Earth related achievement standards are necessary in-flight e.g. during emergencies. These are typical events that must be planned for in exploration missions. The following sections present the principal kinds of deconditioning.

2.1 Bone Atrophy

The skeletal system is weakened through the demineralisation and atrophy of bones, primarily in the bones that are normally weight bearing on earth e.g. the pelvis, femur and lower vertebrae. Although this phenomenon is not completely understood, it is relatively clear that the balance of turnover of bone is detrimentally altered in the absence of the static loading present in 1-g, and by reductions in the dynamic loading applied by impact and muscle activation.

Without countermeasures, load bearing bones can lose 1 to 2% of their density per month for extended periods leading to clinically relevant conditions in one or two years or even less. Moreover, losses still occur in spite of the crewmembers’ participation in exercise [1]. Bone atrophy is considered to be the most serious physiological consequence of body deconditioning in the microgravity environment due to the continuous nature of the deconditioning, the risks of fracture and the time required for post-mission convalescence [2].

2.2 Muscle Atrophy

In the absence of adequate countermeasures, microgravity causes muscles to atrophy and alters muscle morphology, with a resulting loss of contractile mass and performance capability. This loss reduces the speed and strength of muscular contraction and thus the power capabilities. Most of the muscle losses occur in the lower limbs. For limited periods, strength reductions may be in the region of 2 to 5% per week depending on the site and function of the muscle [3]. Soleus peak power for example has been reported to be 32% lower after 6 months on ISS, even with modern exercise countermeasures in place [4].

2.3 Body Elongation

The lack of normal loading during spaceflight causes astronauts to spine elongation of up to 70mm, with significant back pain reported by approximately 50% of US crewmembers [5]. Furthermore, this elongation complicates the precise fitment of EVA spacesuits and can impact crew mission performance through poor quality sleep for those suffering discomfort.

2.4 Cardiovascular Deconditioning

The cardiovascular system deconditions due to reductions in metabolic enzyme levels, the size and quality of capillary beds and energy producing organelles. The main effects observed when returning to gravity conditions are decreases in maximal aerobic capacity, increased heart rate for any given level of exertion and orthostatic intolerance. Currently, 80% of astronauts returning from ISS experience greater than 6% loss of maximum oxygen uptake despite a rigorous countermeasure programme.

Orthostatic intolerance, the inability to maintain the standing position under +1Gz, is a consequence of cardiovascular deconditioning. Due to the risk of syncope (fainting), it is a major problem specifically during re-entry in the atmosphere. The incidence of orthostatic intolerance varies according to space mission duration and the effectiveness of countermeasures, but has been seen to be greater than 90% after some long duration missions [6].

2.5 Neurovestibular and Sensorimotor Deconditioning

The neurovestibular/sensorimotor systems are acutely affected by the loss of the gravity vector resulting in physical space sickness, disorientation and a deconditioning of the proprioception system and associated structures. Upon return to Earth, astronauts must readjust to gravity and can experience problems standing up, stabilizing their gaze, walking and turning, and retaining posture. The magnitude of sensorimotor disturbances after gravity transitions increases with the duration of microgravity exposure, which is of particular relevance to long duration spaceflight. Such disturbances may impact operational activities including approach and landing, docking, remote manipulation, extravehicular activity and egress (both normal and emergency), and thus compromise crew safety, performance and mission success. It is believed that this sensorimotor deconditioning results from in-flight adaptive changes in central nervous system processing of information from the visual, vestibular, and proprioceptive systems [7].

3. DYNASUIT SCENARIOS AND TECHNOLOGIES SELECTION

An analysis of the possible countermeasure scenarios for DYNASUIT and preliminary concepts of artificial muscles technologies have been described in [8], covering the first phase of the activity. Based on this analysis, the two most promising associations of scenarios and technologies have been selected for further detailed concepts and demonstrator experiments.

The first selected scenario is the mechanical stimulation of the foot sole with the purpose of stimulating lower leg stabilizer muscles and bones as well limiting the neuromotor deconditioning and to help maintaining the terrestrial locomotor ability. No specific medical protocol exists for this scenario. However, two strategies can be considered. The first corresponds to the simulation of a walking/running pattern, requiring a time dependent and distributed local variable mechanical pressure on the foot sole of around $10N/mm^2$ with a bandwidth of 2Hz. The second option, based on the protocol defined in [9], requires a vibration in the range of 80 to 100Hz. It is proposed to produce this varying pressure by an array of EAP stack actuators, which are in direct contact to the skin of the foot sole (Figure 1). The pressure change can be provided by the changing electrostatic pressure of the actuator, which is controlled by the applied voltage. For the DNASUIT project, a test demonstrator with a single EAP stack actuator has been built to analyse and demonstrate its functionalities in relation to the requirements of the application. The tests are described in section 4.

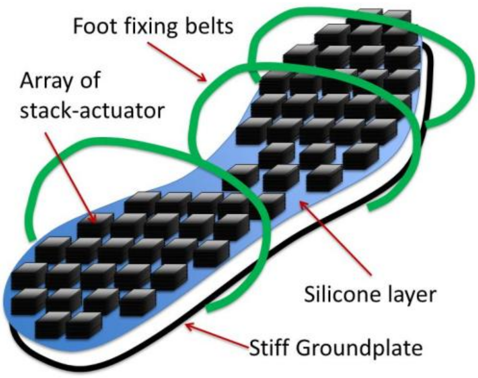

Figure 1: Arrangement and interface design of the EAP actuators for foot sole stimulation.

The second selected scenario is the production of a +Gz loading on the user in a quasi-static configuration (the contraction force being adjustable but constant in normal operations). This action is able to stimulate the body bone structure through a continuous long axis loading (from the head to the foot), graded as a replication of the +Gz profile experienced on Earth. Through this stimulation, the purpose of the suit is the reduction or prevention of atrophy of stabilizer and mobilizer muscles and bone, spine elongation and back pain. The advantages of using active components are continuous calibration, switching on/off the stimulation and easy customization of the loading profile depending on the individual (sizing, comfort) and the scenario context (e.g. countermeasure program, day schedule). The pneumatic sigma actuator is a good option for this application and is proposed for testing as demonstrator during the DYNASUIT project. This actuator can be fully integrated in the suit fabric. Moreover the required pressure for the application should be easily achieved with a hand pump, removing the need of heavy and complex annex system. The test demonstrator is described in section 5.

4. EAP STACK ACTUATORS

4.1 Principle

The contractive actuator in stack configuration currently represents the most advanced and most promising Dielectric Elastomer (DE) actuator design. This actuator consists of a large number of stacked thin silicone film pieces (ca. 80µm) which are coated with compliant electrodes (Figure 2). The electrostatic field induced actuation force and displacement is normal to the plane of the electrode and dielectric film. The obtained contraction of the actuator can be used to drive the device under a specific external service load. The design of the stack actuator is based on a series of capacitors electrically connected in parallel with alternating polarities of each layer. Therefore the actuation voltage has to be applied on two separated compliant electrodes addressing the associated conductive layers.

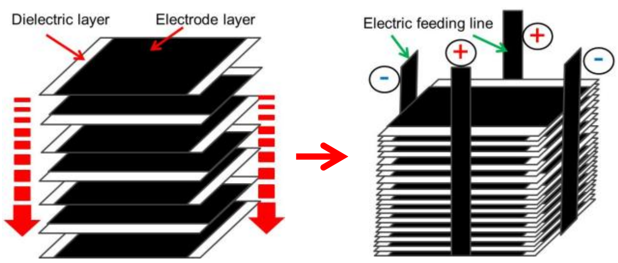

Figure 2: Assembly of the multilayer system and attachment of the feeding lines.

For the production of the actuator the "wet deposition" method has been used. Thereby the dielectric as well as the electrode material is deposited layer by layer in liquid form. After the drying (curing) process of each layer a monolithic structure consisting of many layers is obtained. Due to practical reasons this batch process is stopped after the deposition of ≈ 100 layers, which results in a series of EAP modules. These modules will be piled up and glued to form the final shape and size of the actuator. As the final procedure the feeding lines will be applied by smearing carbon powder on the outside surface of the actuator, which will be connected to the voltage wires.

4.2 Demonstrator Description

The demonstrator consists of one EAP silicone based stack actuator with the following characteristics:

- Max. edge length of the cross-section: 14mm
- Effective length of the tested actuator: 23.8mm
- Thickness of each dielectric layer: ≈ 47.6um
- Number of layers: 500
- Actuation Voltage: < 2 kV
- Switching time ≈ 5ms
- Max. required contraction: >1mm
- Min. required deformation under external load: ≈ 3%

The final real setup is shown in Figure 3. The actuator is polarized using four feeding lines with opposite lines having the same polarization. Electrical contacts consist of curved spring. A high-voltage amplifier is used to polarize the structure. A laser distance measurement positioned above the actuator is used to measure the actuator contraction when the device is activated. All the electrical signals (laser displacement, current and applied voltage) are recorded using a National Instruments interface and a Labview program. A force introduction plate, connected to a spring, can be added through a sliding system to introduce external load to the actuator. However, this has not currently been used for the tests, as the results under load can be derived from passive stress/deformation measurement.

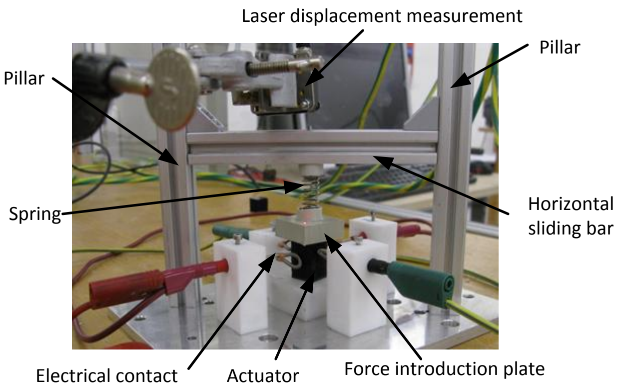

Figure 3: EAP stack actuator demonstrator setup.

4.3 Experiments and Results

Several experiments have been conducted in order to characterise the EAP stack actuator.

The first experiment was the determination of the passive (not activated) modulus of the actuator on a standard tensile-pressure test machine. It varies between 0.6MPa and 1MPa, depending on the strain domain. It is stiffer than the same structure made of pure silicone, due to the stiffening effect of the electrodes, which consist of small but very stiff particles. The stiffness can be adjusted by changing the deposition rate of the electrode material but this has also an influence on the electrical conductivity. This data can be used to estimate the force applied by the actuator by combining it with the amount of contraction, observed in the other experiments.

The second experiment was the measurement of the system response for slow variation of the actuation voltage (< 0.1 Hz, between 0 and 2kV), with no external load. A maximum contraction of 3% has been reached at 40V/um field, ≈ 2kV actuation voltage. This corresponds to the maximum actuation voltage that can be safely applied to the actuator before risks of damages (due to electrical breakdown in the insulator). The obtained deformation would be insufficient considering that we require 3% of deformation under load, corresponding more or less to an additional 3% of deformation with no load. Conclusively this means that the maximum electro-static field of 40V/um should be increased up to approximately 50V/um. This topic still represents the major research and development field at EMPA and comprehensive activities are under way to improve the most promising technology. This level can be achieved by working in clean room environment and using highly precise fabrication tools. Moreover materials with extremely high purity have to be used in order to avoid any contamination and single imperfections.

The third experiment aimed at determining the actuator deformation under higher bandwidth and cyclic operations with square and sinusoidal voltage solicitation (Figure 4). As expected and mentioned above the amplitude of the contraction is decreasing by increasing activation frequency. As a matter of fact the current limitation of the power supply in combination with electrode resistance does not have the capability to completely charge and discharge the actuator within two actuation periods. Moreover the materials viscosity property is limiting the speed of deformation and thus is reducing the maximum deformation capability. Especially in the low frequency domain the difference of single polarity activation and alternating polarity activation is essential. The pulse solicitation provides higher deformation than the sinusoidal approach (Figure 4). This demonstrates the importance of completely discharging the actuator before the next charging phase. This can be achieved in a best way by alternatingly changing the polarity of the capacitors. This experiment demonstrated the capability of the present actuator design for higher bandwidth actuation, although, as above, the resulting amplitude reduction at high actuation frequency has to be assessed for the specific application.

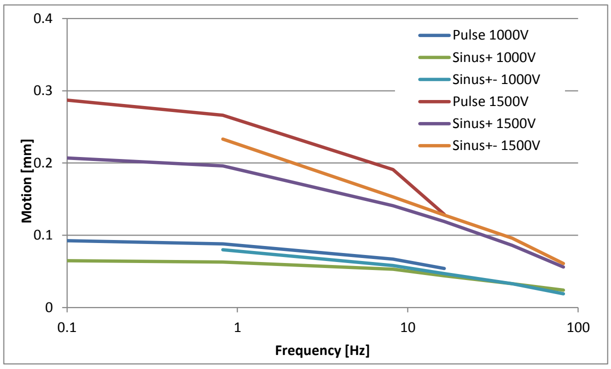

Figure 4: Peak deformation (motion) in dependence of the actuation frequency

Finally, the power consumption of the demonstrated stack actuator has been determined as a function of the actuation frequency (Figure 5). Obviously and as expected, the consumed electric power is increasing with higher actuation frequency. Considering the foreseen number of actuators embedded on the sole, and with a required limit of 30W for the actuation subsystem, 10Hz is a first approximation of the upper frequency limit. However this value is true when all actuators are in operation simultaneously. The total power consumption can be reduced when many actuators are in operation in phase shifted mode, so that the stored electric energy in one actuator can be used to charge another actuator. In this case a smart and highly sophisticated controller is strictly necessary to transfer electric energy from one actuator to another actuator and thus to avoid electric energy loss at the actuator discharging process (producing heat), which in this case is strongly recommended.

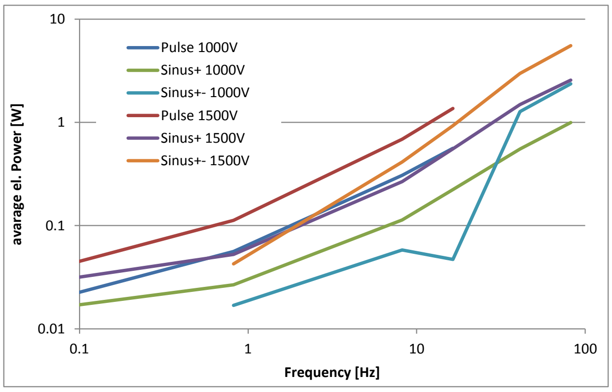

Figure 5: Power consumption of one stack actuator as a function of the actuation frequency.

5. SIGMA PNEUMATIC ACTUATORS

5.1 Principle

The construction of the sigma actuator is shown in Figure 6. An airtight bladder, made of a thin foil is placed in a pocket, inserted in the suit between two fabric layers. Upon inflation, the thickness of the bladder increases pressing on the ambient fabric which results in a contractive force. The generated force is given by:

$$F = p \cdot s \cdot l \cdot \frac{\cos\varphi}{\varphi} \qquad (1)$$

with F the contraction force, p the air pressure, s the width and l the length of the actuator and φ an angle related to the contraction of the air bladder. The maximum usable contraction force happens at the beginning of the contraction and then decreases. The contraction is typically limited to 20% to keep reasonable high forces.

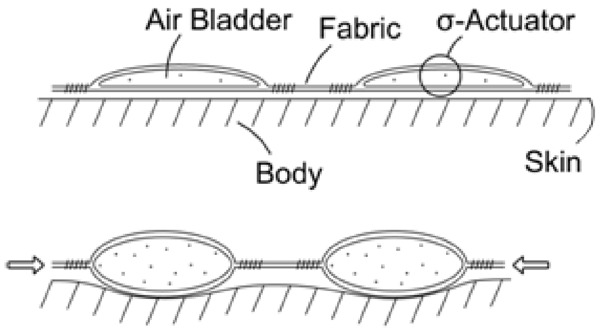

Figure 6: Cross sectional view of two sigma actuators in the relaxed (upper) and activated stated (lower).

5.2 Demonstrator Description

The +Gz loading demonstrator consists in an active wearable bodysuit providing a compression force between the shoulders and the crotch (Figure 7). For this demonstrator, six sigma actuators are integrated in the fabric on the front and on the back, in a series configuration. All actuators can be independent of each other pressurized. Theoretically, two sigma actuators in series would be enough to produce the required force and contraction. The setting with 6 actuators allows optimizing both the number and location of the actuators. The current demonstrator addresses only the upper body stimulation, although a full body suit (acting also on the legs) would be necessary for a full +Gz compensation.
In order to generate high loads, it is very important that the fabric has a high elastic modulus. If the fabric is too elastic, inflation of the actuator mainly leads to a stretching of the fabric around the bladder and no tension forces are generated. The silver fabric is polyamide from the company Schoeller textiles (product

nr. 1455). It was chosen due to the expected low strain, which, however, was not observed during the experiments. The bladder material is a thin PU-foil. A small electrical pump SPV6T is used to inflate the actuators.

Figure 7: Test of the DYNASUIT +Gz suit demonstrator with a human subject.

The +Gz demonstrator suit has been evaluated in three different configurations:

- Setup 1: mounted on a test bench to measure the intrinsic characteristics of the suit.
- Setup 2: mounted on the EMPA manikin "Nick" that is equipped with load sensors to measure the total vertical force between the upper and lower body (here shoulder and hip), and small pressure probes at dedicated locations to measure the pressure on the skin
- Setup 3: donned by a human subject for qualitative evaluation of usability and comfort aspects.

5.3 Experiments and Results

The first experiment, on setup 1, consisted to measure the displacement of different weights attached at the bottom of the suit, under activation. This experiment illustrated the non-linear behaviour of the sigma actuator; the main part of the lifting being obtained at the lower pressure before saturating (when the pocket takes a circular shape). • The suit is quite well stretched under loading, indicating that probably a stiffer fabric should be used. A better choice would be a thicker fabric, or none woven fabric. The optimization of the mechanical properties of the fabric is, however, a rather complex task.

The second experiment, on setup 2, aimed at measuring the vertical force produced by the suit, as a function of the air pressure for a full activation of the suit (2x6 sigma actuators). The air pressure in the pockets was increased by the electrical pump to the maximal value of about 700mbar (maximal pressure delivered by the pump). The vertical force experienced between the shoulder and the crotch as a function of the actuator inside air pressure is illustrated in Figure 8. The force increases strongly for low pressure values and levels off at higher pressure values at a level around 45kg (corresponding to the compression requirement of the suit). This is a typical behaviour of sigma actuators.

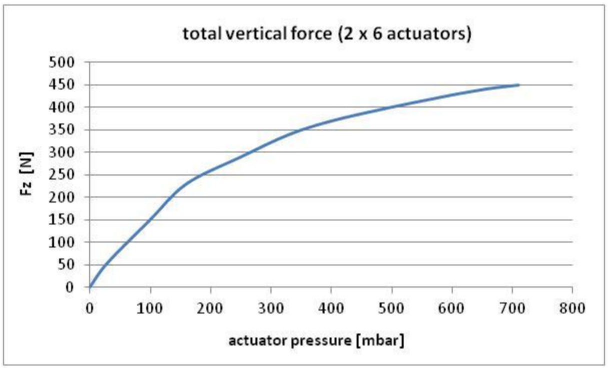

Figure 8: Vertical force Fz between the shoulder and the crotch as a function of the sigma actuator pressure.

The skin pressure at the shoulder as a function of the air pressure is shown in Figure 9. The maximal skin pressure is about 55 mmHg (73 mbar) and might be locally even higher (for instance, at the region closed to the neck, the strain of the fabric was very high, indicating local high skin pressure values). This is higher than the assumed maximal tolerable value of 40 mmHg. However, the skin pressure can be further reduced by a suit which fits perfectly to the person wearing. Another approach could be the introduction of more fixation point, e.g. at the middle of the upper body, in order to reduce the skin pressure and improve the comfort.

The same experiment has been conducted with half of the actuators activated. Only about half of the vertical force is then obtained. With 2x6 actuators a pressure of 130 mbar is needed to apply a vertical force of 20 kg, while the full pressure of 700 mbar is needed for the same vertical force for 2x3 actuators. This is a difference of more than a factor 5. The lower pressure means less energy for inflation and more comfort for the person wearing the suit. Thus, a setting with more actuators is to be preferred. The skin pressure is about 30 mmHg for both settings at 20 kg vertical force (acceptable value). The stretching of the fabric plays a crucial role in the suit. In fact, the elasticity limits the maximal applicable vertical force of the sigma actuators. The material and number of actuators need to be carefully balanced.

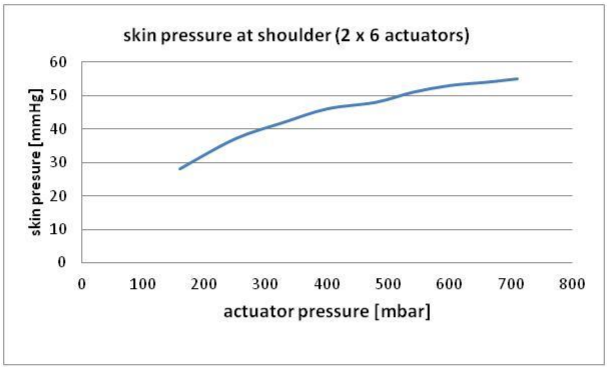

Figure 9: Skin pressure at the shoulder as a function of the sigma actuator pressure.

For the last experiment, the suit was tested on human subjects (male and female). It was activated with a medium pressure value of 500 mbar. The resulting compression force was well perceptible between the shoulders and the crotch. The following observations, related to the comfort of the suit, have been made:

- The pressurized actuators gave somehow a stiff feeling in the circumference. This could be improved by dividing the actuators in two or three chambers in the circumference.
- Bending forward and backward and sideward was not restricted by the suit (in the activated or deactivated state).
- Wearing the suit for longer times results in sweat underneath due to the plastic bladder material. For a future design this could be resolved with a breathable bladder.
- The female subject reported some pressure on the breasts (at 450 mbar). This could be released by a slight forward bending. The suit also pressed some seems of the bra into the skin which was felt unpleasant. For female subjects the suit needs to be carefully designed at the breast region. This might be done by positioning the actuators more towards the hip.
- At high pressure the suit was uncomfortable at the crotch especially for male subjects. A better load introduction into the hip is needed.

The load introduction at the crotch of this demonstrator is not acceptable for humans. The solution was chosen due to its simplicity and with the main tests situation on a manikin in mind. For human subjects the load needs to be distributed over the buttocks and upper legs in order to be comfortable.

6. CONCLUSIONS

This paper provided information about the design, building and testing of two artificial muscles demonstrators with the purpose to demonstrate their feasibility for integration in a countermeasure suit. The two technologies investigated are the EAP stack actuators aiming at providing vibration stimulation on the foot sole or on the body, and sigma pneumatic actuators able to produce axial loading on the body.

The feasibility of using EAP technology to provide vibration stimulation has been highlighted, although the achievable deformation obtained with the current actuator (3% in free deformation) is below the required deformation for the foot sole applications. New techniques of manufacturing for silicone based actuators could improve the achievable deformation until 10% in a short/medium time frame (+/- 1 year). These actuators would then be compatible with the requirements of this application. This demonstrator has only one actuator. Further tests shall be performed with a full integration of several actuators with feeding lines and energy transfer between the stacks, as well as tests with human for validation.

A 2x6 sigma actuator suit has been built and tested in three different configurations, including tests on human subjects. The feasibility to provide axial loading has been clearly demonstrated. However, the load introduction to the body, the stiffness of the material, the shape and numbers of sigma actuators have to be optimized to improve the comfort and the usability of the suit. The current version of the suit addresses only the upper part of the body. The concept shall have to be extended to the lower part (legs), where the deconditioning is very important.

The purpose of the next phase of the project is to provide one detailed design of complete countermeasure suit with the goal to assess the physiological impact and technical limitations for a full suit design. The two technologies address complementary scenarios and are independent in terms of integration. For these reasons, both will be investigated for integration in the detail suit design.

7. ACKNOWLEDGEMT

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