H-reflex conditioning in healthy individuals and an Exoskeleton for knee flexion assistance

Report

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1 Introduction

Worldwide, an overall estimated number 12 million people of affected patients are suffering from spasticity, a serious impairment in life quality for patients [25]. The chance to face spasticity as a within the first year of a stroke is 38% [34]. Compared to the total population, spasticity has an overall prevalence of 0.2% [25]. Spasticity can be caused by different diseases, like multiple sclerosis or upper motor neuron syndrome, latter shall be addressed in this research. Spasticity, or more specific hyperreflexia, is the main reason for Stiff-Knee-Gait, a disease where an overactivity of the rectus femoris prevents the patient to perform a physiological gait cycle. Reduced walking speeds due to the need of longer stance phases on the unaffected side are the daily impairments patients have to deal with. Longer distances are not feasible for those stroke survivors, about 60% of the performed work is done by the unaffected side [13]. Stiff knee gait is an abnormal moving pattern which can be characterized by a diminished and delayed peak knee flexion during the swing face of the gait cycle. It inhibits toe clearance, dorsiflexion, and therefore patients struggle to not trip or fall while they are adopting strategies like exhausting compensatory movements [27]. It is thought that the reduced knee flexion is caused by an exaggerated knee extension moment generated by inappropriate swing-phase activity of the rectus femoris muscle [28, 16], also known as hyperreflexia. Thus, reduction of rectus femoris hyperreflexia should improve the overall walking function. Today's clinical interventions still have unbearable side effects. A usual approach is to denervate the femoral nerve chemical in parts and damage the reflex arc deliberately, this shows modest results in knee flexion and does not improve walking speed, although improving the energy cost [11]. An alternative current approach is the neurotomy of the rectus femoris nerve. This seems to improve and normalize the preoperative abnormal gait through improving kinematic parameters. Although the overall muscle burst of rectus femoris is being reduced. A reduced muscle strength should be avoided, especially in patients with an reduced walking distance, this seems to be a significant drawback [6]. A novel approach of treatment should be tested for feasibility in this study. It is based on providing feedback for the patient from the muscle action potential measured to obtain a parameter for monosynaptic spinal reflex activity elicited through electrical stimulation of the peripheral nerve. This technique is known as H-reflex conditioning, which has been well characterized and studied as well in animal studies and recently got translated to human participants [30]. There is currently a lack of evidence for the feasibility of the process of conditioning in the quadriceps muscle and in post-stroke individuals. Therefore, the aim of the following study is a proof-of-concept study to examine the feasibility of the procedure, it is expected that conditioning the participants could provide a non-pharmacological, non-surgical method to enhance the life quality of individuals with neuromuscular disorders.

1.1 Theory behind the study

1.1.1 Functional chain

The functional chain how the human body performs a voluntary movement evoked through an external input has been examined for approximately 150 years now. Models have been created through the input of different studies like investigation of deficits and the corresponding lesions in the nerval system. Modern theories are built around models consisting of circuitry, analog to the electrical pendant. Volitional movements are, in traditional models, seen as a counterpart to reflexes, which are unintentional and do not need an active involvement of the higher neural structures such as hindbrain, midbrain or cerebral cortex [22]. Charles Scott Sherrington has found the mechanism behind the reflex arc-theory [24], which created the possibility to create a mathematical model to predict the output of input stimuli on the arc. Those systems have been characterized by transfer functions using a linear system approach to approximate the input / output relations of the stretch reflex in mammals [20]. In general, the reflex arc of a mammal is made up of an afferent pathway leading from a receptor to the spine and an efferent nerval pathway to the according effector, which is the muscle. Muscle control can be compared to a negative feedback control system in a technical manner – afferent inputs act as negative feedback to the supraspinal signals conducted from motor cortex via upper motor neurons followed by lower motor neurons leading to the synapse in the ventral horn of the spine. Recent anatomical models differentiate between two main different types of lower motor neurons found in the ventral horn: alpha motor neurons and gamma motor neurons, the first innervate extrafusal muscle fibers and the latter innervate intrafusal muscle fibers. This coupled system excites both muscle fiber types roughly at the same time to ensure the intrafusal fibers stay under tension while the bulk of the muscle, the extrafusal fibers, contract. This is important to maintain the stretch sensitivity of the contracted muscle during movements and prevent slacking of the sensory fibers [9]. Intrafusal muscle fibers act in the system as the specialized sensory organ detecting the amount of change in the muscle length and provide the negative feedback for the control system. Damages in the upper motor neuronal system, known as upper motor neuron lesions or pyramidal insufficiency can lead to severe impairments for the subject's gait.

1.1.2 Upper motor neuron syndrom

Upper motor neurons, which are located in the cerebral cortex, internal capsule, brain stem and the white matter of the spinal cord act as descending tracts from the brain to lower motor neurons. Therefore, damage causing lesions can prevent or alter the transmitted signals on the descending nerval pathway leading to the lower motor neurons. The main reason for occurring damage are strokes in this specific area, generated by an absence of necessary oxygen to supply the neurons, although multiple sclerosis and other pathologies can also be an underlying reason. Lesions have different effects on the muscle they are leading to – eliminating descending input from the motor cortex to inhibitory interneurons will shift the relation between input / feedback signals and hinder the underlying control system to act in a physiological manner [2]. However, following symptoms happen only due to a still intact stretch reflex arc, and intact gamma motor neurons.

Damaged lower motor neurons will prevent any involuntary or volitional innervating stimuli to be conducted to the muscle. The lack of supraspinal input to inhibitory interneurons allows gamma neurons to discharge at much faster rate, leading to exaggerated stretch reflexes. One of the results of such a pathological situation is Hyperreflexia, where myoatic (stretch) reflexes are the dominating force in the muscle and therefore will generate a spastic response for the patient [9]. In a clinical practical case are specific tests necessary to distinguish between the possible underlying reasons causing an involuntary muscle overactivity, it can also be the result of a dysfunction of segmental spinal modulation [23]. Hyperreflexia itself can be seen as a problem that mostly occurs due to underlying problems in the reflex arc. The nerval pathways reflexes use are the same as the pathways a as the input for voluntary movement carried over the corticospinal tract. Spinal interneurons are small connections within the spinal cord performing exciting or inhibiting actions. Inhibitory interneurons are the dominant variant – they coordinate muscle actions around a joint to bring together antagonists, agonists, and synergists into a myactic unit. This myactic unit fulfills regulating tasks like maintaining a proper joint stiffness. To be more specific, the Ia afferent fiber, which acts as the sensor for length changes in the muscle, is connected over a direct synaptic coupling with an Ia inhibitory interneuron which is purposing as a inhibiting element on the alpha motor neuron leading to the muscles antagonist. This allows unhindered movement of the agonist [9].

1.1.3 Spinal stretch reflex

Examinations for the integrity of the reflex arc and the spinal stretch reflex itself can be done with few equipment. Through tapping a tendon muscle fibers are being stretched same as the dynamic component of the muscle spindle what activates the Ia afferent sensory fibers. The signal of the high rate of muscle length changing is coupled over the direct connection between Ia afferent sensory fibers with alpha motor neurons, who themselves fire the signal back to the muscle trying to reset it in the original position and prevent further, too fast, stretching which could damage the integrity [9]. In addition, the spindle afferent has often monosynaptic connections with the synergistic muscles, as well as being connected to Ia inhibitory interneurons acting on the opposing antagonistic muscles to keep them relaxed and provide an unhindered movement for the reflex [9]. Physicians investigate the knee jerk reflex using a tendon or reflex hammer to check the integrity of the nerval pathway leading from muscle spindles and the Ia afferent branch back to the alpha motor neurons. This provides a simple investigation for a patient's motor neuron lesions, which would be shown in either abnormal high reflex responses, Hyperreflexia (Upper motor neuron syndrome) or abnormal low down to even a lack of reflex responses (spinal or nerval root damages). Nevertheless, an exam like this revealing the spinal stretch reflex is very difficult to standardize. It brings the problem of the hammer impact position, the exerted force by the physician and the subjective grading. A more technical approach to investigate the reflex arc integrity leads to the Hoffmann- reflex.

1.1.4 Hoffmann- reflex

The Hoffmann- reflex is the electrical pendant of the body's mechanical nerval response observed in a lot of mammals, the spinal stretch reflex. It is electrically induced, analogous to the reflex hammers input for the spinal stretch reflex. Applying an electrical stimulus on a nerve creates several responses in a human body, which can be either seen directly through a muscle reflex twitching or, using an Electro-myography device (EMG), through a change in the electrical potential on the skin surface around the nerve. It is an estimation of alpha motor neuron excitability in a specific nerve while intrinsic inhibition [37] and excitability [4] through interneurons remain constant. The main difference for the examination result comparing a Spinal-Stretch-Reflex (SSR) and the H-Reflex is the bypassing of the muscle spindles [21] using an electrical stimulus and therefore creating an valuable assessment tool to evaluate the neurologic function for different time dependent responses. Providing an electrical stimulus to a peripheral nerve stimulates the Ia afferent sensory fibers, which lead back to the synaptic joint and provide a stimulus to the motor neurons of the corresponding muscle [4], as well as stimulating the efferent arcs alpha motor neurons, who are transmitting the potential in distal

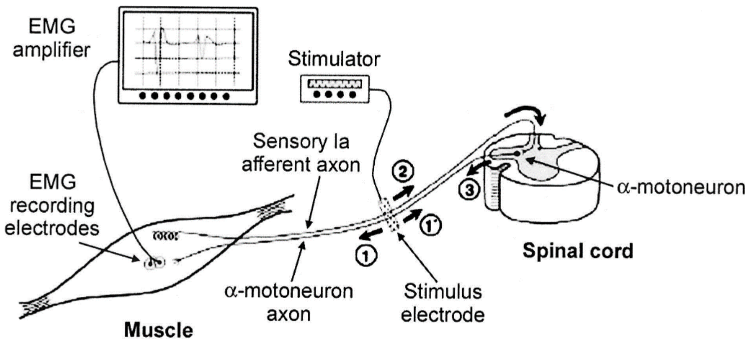

Figure 1.1: Electric potential measurement schematic for spinal pathway

direction eliciting a first muscle response, precisely seen in the soleus muscle potential 3-6 ms after the stimulus, called the M-wave [14]. The H-wave is likely to appear 28-35 ms following the stimulus. Those timings depend on the accounting limb length, increasing the pathway is increasing the time gaps between stimulus, M-wave and H-wave [14]. To stay correct, there is a third response observed in muscles action potential, the F- wave. It seems to appear at the same time as the H-reflex. Taking a look on the transmission path, the H- wave is a response on the elicited action potential traveling orthodromically along the Ia afferent sensory pathway entering the spinal cord, where it crosses the synaptical gap onto alpha motor neurons where it is transmitted back to the muscles where it evokes the muscle response. The F- response, appearing at high stimulus inputs, itself is not an reflex. It is generated through an antidromically transmission path of the action potential in the motor neurons back to the spine, where too low amplitudes will fail to depolarize the cell body. If the amplitude of the stimulus has been high enough, the input elicits a backfiring of motor neurons in orthodromic direction from the spine to the according muscle. The F-response is hard to standardize, it will generate a varying population of motor units who are responding to the stimulus [29]. An Figure of the description above is shown in 1.1, here can be observed the Ia sensory axon leading the signal back to the spine where it is being conducted over a monosynaptic gap to the alpha motor neurone where it can be seen on the EMG graph.

Figure 1.2 shows how a typical EMG response after a stimulus, which is shown as the high peak at the beginning of the graph, does look like

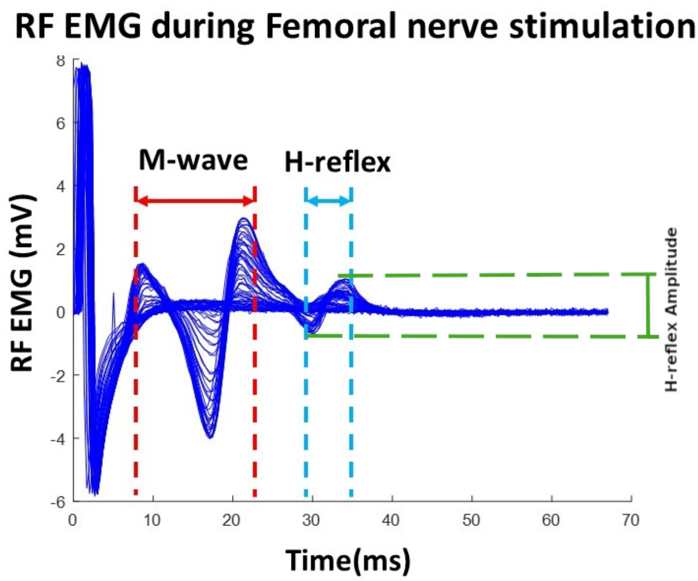

Figure 1.2: H- reflex as seen on the EMG for the rectus femoris

while using a proper stimulus amplitude. The H-reflex and the M-wave can be distinguished very well. To standardize the measurements, the ration Mmax amplitude divided by Hmax amplitude is obtained. This allows for comparison of the results over different sessions.

1.1.5 Recruitment curve

There are several things to consider while working a long-term study measuring and comparing the H-reflex. The H-reflex itself reveals on the EMG at low levels of stimulation. While increasing the stimulus amplitude, the threshold to depolarize the alpha motor neurons is achieved at some point and there can be seen a M- response as well. Further, increasing the stimulus input the H- reflex can be observed at higher amplitudes, at some point while still maxing the amplitude it starts to decrease again, till it finally vanishes and M-wave achieves its maximum and remains stable [14]. The H- wave is getting decreased with higher amplitudes because of the occurrence of the F- wave, which is depolarizing the motor neurons in antidromic direction. As the neurons need some time to build up an chemical potential difference again, they can not be depolarized in this time and therefore they can not be fired for the H-reflex response itself, lowering its amplitude till it vanishes. In this studies there have been trials to obtain the highest possible amplitude of the H-reflex at this specific day with this specific electrode positions. This has been done performing a recruitment curve. Therefore, the stimulus has to be increased gradually in intensity from zero to a point where increasing the amplitude does not increase the amplitude of the M-wave anymore [14] till there can be a graph like Figure 1.3 obtained where the stimulus amplitude eliciting Hmax is clearly visible. In Figure Illustration 32: Recruitment cure is shown how an increasing stimulus does increase the M- response till a maximal value, but enhancing the stimulus amplitude over a certain value lowers the H-reflex amplitude. The perfect stimulus input for obtaining the H- reflex is shown here at $x = 3$. The absolute values of the reflex and M-wave depends on the physical condition of the subject, a specific day, and the electrode placement. This is the reason for necessary daily evaluation of the recruitment curve. Measurements for the study have been examined on the quadriceps femoris muscle, the frontal knee extender. The according nerve for this muscle is the femoral nerve, therefore it has been aimed to get as close as possible to this nerve with the electrodes. The femoral nerve runs down from the pelvis to the anterior thigh and provides the nerval signals to flex the hip joint and the extenders to quadriceps, which includes rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius muscles. It also consists of branches providing sensory functions for the thigh.

1.1.6 Spinal plasticity

The point of the study is to examine if the technique of H-reflex operant conditioning, which takes effect of the spinal plasticity effect, is feasible in the rectus femoris. Spinal plasticity is known as short to long term alternations of the spine's neural pathways, adapting to some external recurring stimulus. A lot of literature available is modeling the spine as hard-wired neural pathway, not able to change over time, and as a transmitter of information to and from the brain. Also it is widely known that spinal

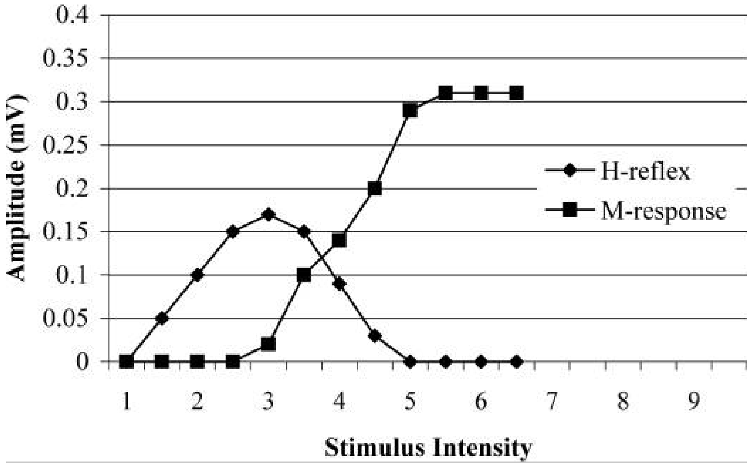

Figure 1.3: Example of stimulus amplitude dependency of stimulus intensity curve [14]

cord acts as a repository of different reflex functions reacting on a lot of external stimuli automatic, bypassing the brain. Newer models show that the spine instead is consisting of flexible and dynamically changing pathways showing characteristics of learning and memory in the intact mammal. Long excitability alterations are likely to an change of interneuron cell receptor sensitivity, this can last up to a few hours or a day which is an interesting thing to consider as while-training effect during one study session. There is another stage of excitability changes, known by the term spinal fixation. Here a longer, more intense stimulus can increase the excitability of the stimulated reflex pathways for several days and leaves a memory trace in the spinal cord. This fixation is achieved by stimulation or lesions of different brain regions. It is also possible to produce such a response by stimulation of a peripheral nerve, or the underlying reason can be an inflammation of a peripheral area such as a knee joint. Fixation seems comparable to long-term potentiation, a long- term learning and adapting strategy of the memory in mammals. A non associative excitability alternation can be permanent, therefore an extended stimulus input onto reflex arc will be applied. This may destroy inhibitory interneurons, who are regulating the excitability balance [5, 35]. A newer case study by Thompson et el. seem to be quite successful if using a proper conditioning protocol including feedback for participants could alter the H-reflex as well [31], although there is still a lack of long term studies investigating the stability of adapted changes over several years.

2 Operant conditioning of rectus femoris

2.1 Test setup for study

2.1.1 Devices

A standard differential EMG preamplifier has been used to obtain the H-reflex, namely the BORTEC AMT-8 EMG Amplifier. It features a Bandpass-filter between 10 – 1000 Hz and 8 input channels. Assuming this device has several unpublished filter techniques included, for example to measure the potential without having to deal with a lot of electrical disturbances, a standard EMG device has filters included. If they are actually implemented as analog, physical variant or included in after the analog-digital converter in digital form depends on the device. As an example a notch filter designed with cut off frequencies between 49-51 Hz would be suitable to filter power line humming (in countries with a 50 Hz power line) in the EMG graph, nevertheless it has to be considered every applied filter will probably lower or hide parts of the signal actually being useful. To digitize the amplified and prefiltered EMG signals the PCIe-6259 card manufactured by National Instruments has been used, it had been put in a dedicated observation PC running Windows 10. This provided the interface which generated the data stream for the running Software. This card serves as an USB multifunctional I/O device including 32 analog inputs, 4 analog outputs, and 48 digital input/outputs. The software to trigger the stimulus and save the data read at the right timing after the stimulus was EPOCS. It has been developed for stimulation and data acquisition purposes. Also a custom MATLAB script was used to test the setup. For the stimulation device there has been used a constant current pulse stimulator suitable to use with surface electrodes, the Digitimer DS8R. It has proven useful the stimulator can be triggered using a TTL input (external PC with custom software). For the pulse width 1000 µs had been chosen, the shape was a rectangle input and the amplitude was obtained through the recruitment curve, although it is possible to variate the device's output between 0 – 1000 mA constant current.

2.1.2 Participants and tasks

As the study is done on individuals, there have been recruited a total of 3 healthy persons. There have not been any special restrictions to choose the subjects besides showing no gait impairments. For the study the surface EMG of rectus femoris was measured, while the femoral nerve has been stimulated on the skin surface. For providing the feedback of the H-reflex for the subject, there has been established a monitor interface with a visual feedback of the rectus femoris H- reflex and the current muscle potential obtained by the EMG and provided by the EPOCS software. The optimal stimulus amplitude has been obtained using the recruitment curve. The task of the subjects was to maintain a stable voluntary muscle potential, therefore they were given a visual feedback using the EPOCS software. This is due to the possibility of altering the H-reflex while flexing the rectus femoris, if this was the case the software did not provide a stimulus input and postponed the next stimulus till the participant was able to provide the needed EMG activity again.

2.1.3 Study design

The conditioning protocol itself was chosen to be based on a well-established existing one. The first 6 sessions are baseline sessions, followed by 24 conditioning sessions. In total, the timespan for the conditioning should have been 10 weeks, so 3 sessions per week. The participants were instructed to maintain a natural standing posture and a stable background EMG, while the H-reflex was elicited. M-wave size was kept constant, and no feedback of the H-reflex was given to the subject. Each baseline session consisted of

225 H- reflex control measurements. The training sessions started off with 20 trials to build a within- session baseline, followed by 3 runs of 75 training trials, in total 245 trials for a subject. For the training trials, the participants were given an additional feedback of their current H-reflex amplitude in form of a amplitude bar. Their task was to lower the height of the rectus femoris reflex bar while maintaining the background EMG. There was no explicit strategy given, as former studies have shown successful approach strategies are quite individual. A successful trial, talking about a H-reflex below the within session threshold, was followed by an immediate feedback. More precisely, the color of the H-reflex amplitude bar changed color and the shown success rate increased [30]. After completion of the training trials, 4 follow up sessions in the next 3 months should be done. To compare the obtained data, the first 6 baseline session results should have been averaged and compared to the average of the last six training sessions. The differences should have been compared using a linear mixed effects model.

2.1.4 Electrode placements

The participants have been tested for the peak of the muscle bulk while flexing it to ensure the electrode is placed correctly. In total, there are 6 surface electrodes placed for each measurement session. The stimulus electrode is placed on the femoral nerve, just at the point where it starts to split into different branches. The anode for stimulation is located in a straight, horizontal line back through the body on the gluteus maximus. There is placed one electrode on the vastus lateralis and the vastus medialis each, those are for measuring the effect of a H-reflex conditioning on surrounding muscles to rectus femoris, but this data is collected only for further examination. One electrode is obtaining the muscle potential at rectus femoris, this one is used to record the background EMG and the H-reflex after a stimulus. Because the used EMG device has an active driven feedback to ensure very few disturbances, there is another electrode placed on the knee patella to couple negative electrical disturbances on the skin surface.

2.1.5 H-reflex normalization

The slightly different electrode placements are a reason for the different absolute H-reflex amplitudes each session. Also because the amplitude itself varies between different participants there is a need for normalization. Those variations can also result from different skin resistances or subcutaneous fat. For example the H-reflex can be normalized with the use of the M-wave. The lowest stimulus amplitude eliciting a maximal M-wave has to be found. Adjustment of the stimulation intensity has to be performed to produce a H-reflex with an equal amplitude to some percentage of Mmax, and this stimulation intensity is being used to run the trials. However the stimulus percentage is an arbitrary choice but has to be chosen at the up-sloping part of the recruitment curve. This method of normalizing also allows for an evaluation of the same portion of alpha motor neurons every trial if wished, for example a maximal M- wave is conducted using all available motor neurons while 10% of the amplitude are using about 10% of the depolarizable motor neurons [14]. The normalization method used for this study is to standardize the Hmax/Mmax ratio. The Hmax amplitude is an indirect estimation of the number of motor neurons being recruited and the Mmax value represents all motor neurons available, so the ratio Hmax / Mmax can be interpreted as a specific part of the whole motor neuron pool. Important for data collection is to assure a recording of raw H-reflex and M-wave amplitudes to detect a changing in the M-wave and therefore the ratio. For multiple sessions using slightly different electrode placements the latter one is preferred, because those movements make it hard to assume the same portion of the motor neurons is being stimulated.

2.1.6 Expectations for the outcomes

It is expected to achieve an overall decrease in average rectus femoris H-reflex magnitudes in healthy individuals without showing any side effects. If this expectations are met, a new study including post-stroke participants can be done. In theory, it should drive their overall gait pattern back towards a more physiological one. There are no significant side effects expected, since the whole procedure takes effect of the body intern, physiological possibilities of axons to adapt, more precise the spinal plasticity in a human body. Treatment of patients itself is quite excessive, although it could be included or even combined in a physiotherapeutic training and be examined outside of a clinical environment. A drawback to consider is the time dependence, following this procedure will take about 10 weeks to show first results, although a trend can be estimated during the procedure. Also, it is likely patients will need follow up sessions where subjects are shown their reflexes again to give the brain the possibility to adapt the reflexes again. In the lab there have been trials showing that the effect will vanish over time without any follow up sessions. Also, cost is a significant factor, because the needed time for the treatment is immense. Building up a good environment, for example in a physiotherapeutic's office, could enable mentally fit patients a treatment on their own, only the electrode placements have to be done by someone with medical background knowledge. Another possible outcome in the long run is a treatment for multiple patients done by one medical staff member. Table 2.1 arranges the outcomes for baseline sessions for 3 participants which could be observed before the restrictions regarding Covid-19 got more strict and banned the study. Each row represents a participant, the results shown in a bar graph on the left and the same values presented in a linear graph on the right column.

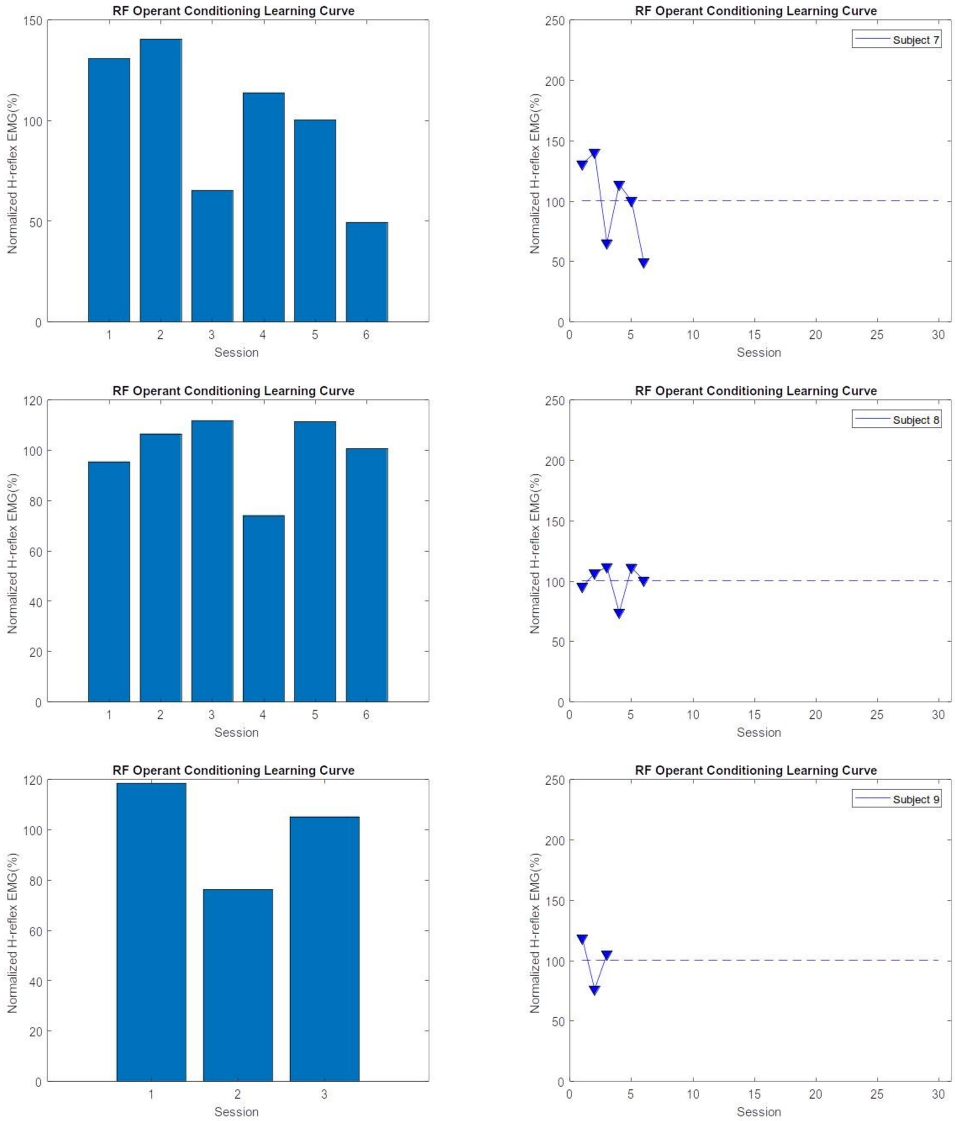

Figure 2.1: Measurement results for the baseline sessions

2.1.7 Situation with Covid-19, outcomes

Due to the situation with Covid-19 in Texas the trials came to a quicker end as originally planned, concrete there had been preparations, theoretical research and some data could be collected including 6 baseline measurements. Unfortunately after this session, access to the lab and the equipment got restricted, and the other lab employees are still in home office at the time this report is composed. Because the research background topic has proofed very interesting, of global importance, and may it could show a solution to enhance a lot of patients life quality without a lot of side effects there has been research for a different approach to help those patients using an exoskeleton, explained in the next section. Outcomes of the obtained 6 measurement sessions are illustrated and included on the following pages. The bar height for the figures arranged in Table 2.1 represents the mean of 225 trials of each session, 3 runs with 75 trials, and each sessions H-reflex has been normalized by M-max amplitude for session to session comparison. There cannot be done any trend estimation, as there has not been any feedback for the participants during the baseline sessions.

3 Exoskeleton for knee flexion assistance

A further project which was, same as the operant conditioning, aimed to improve the life of patients facing the daily struggles with stiff knee gait. Therefore literature research for an external, active drivable device to assist stroke patients with bending their knee has been done. The focus has been laying on general exoskeletal devices, not limited to the lower limb. There has been the assumption device designs for example built, as example, for the arm could provide important design inputs for external actuators.

3.1 General information

Powered exoskeletons allow individuals to move more freely, farther distances and in a more physiologic gait pattern. The usage of robotics to assist limb movements has been proven to affect psychological well-being by improving quality of life through enhancing independence, decreasing anxiety and depressive symptoms [32]. Wearable robots developed as an result of a transition from seeing robots only useful to replace humans in tedious and repetitive actions to a scenario where robots are actively interacting with humans. This is being expanded from a sole exchange of information like in teleoperation tasks, to service robotics where they stay in close interaction with humans including physical and cognitive modalities [19]. The term wearable robot refers to a person oriented robot, worn by human operators to supplement a limb function or even to replace it completely. In first case the robot is working in parallel with the human body, the latter case in series supplementing the limb in its function. To link the needed information between a robot and its user, a robot–human interface is needed. This can be either a physical or a cognitive interface, first shall be focused on in this study due to its simpler implementation and no expected drawbacks for an assistive device. In terms of implementation this will be a explicit developed interface to support the flow of power between the human and the robot actors. It is based on a set of actuators fixed to a either rigid or flexible frame which is worn and therefore connected to the human skeletal system. Because of the close and rigid connection between human and robot there have be a lot of considerations met in terms of safety [19]. An exoskeleton is a special species of wearable robot, where the design aims to map the exoskeletons kinematic chain onto the human limb anatomy. There is a specific term to describe how well the exoskeleton can mimic the human limb joints, namely the kinematic compliance, which is a key aspect to consider while developing this sort of robot. However, the point of this literature state of the art research is to estimate current developments in lower limb exoskeletons. Those can be classified mainly due to their particular application as assistive device for human impaired movement or human power augmentation [19]. Today exoskeletons are considered as one of the most challenging areas of robotic research, it is a interdisciplinary field requiring knowledge of materials, electronics, mechanics, sensors, controls, intelligence, communication technologies, power sources and actuation [3]. Due to the inherently coupling between exoskeleton and body limbs there have been developments for safer actuation systems providing a link between "hard" actuation systems like hydraulic or electronic devices. For example pneumatic actuator systems mimicking the human muscle profile have proven as a more sufficient actuation technology for providing force inputs [3]. There are many different designs and approaches to transmit external forces actuated by an body- external system to the human body, one of the most obvious classification is the differentiation between the general material solidity and the effects on the body, how in particular the force is contributed to the patient.

3.1.1 Classification of exoskeletons

Exoskeletons built without a rigid frame, instead using a soft, more compliant and elastic material like neoprene, are considered soft exoskeletons.

The aim and use case depends very much on the pathology, as well as the mobility of the patient. Soft exoskeletons are the preferable choice in terms of portability to a low mass and inertia, a comfort sit and more forgiving if the size is not perfect adapted for a specific patient. Another significant advantage compared to rigid devices are the unrestricted joint movements possible by the wearer, which should make adapting to different scenarios or wearers less complex. In other terms, this minimizes the unintentional interference between the wearable robot and the wearer, which gives the possibility to preserve more natural gait mechanics. To build such exoskeletons, there have been developed different materials and shaping. In general, there are purely passive devices such as the commercial available Ski-Mojo, a passive device worn on the lower limb. This device uses its material properties to dampen the shock occurring during skiing, and is able to reduce the muscle action potentials measured by an EMG [17]. The next step would be a quasi-passive device, which is storing energy in an internal structure and is designed to have a control system which is releasing the stored force on demand of the user [10]. The more advanced ones do have a active actuation system, either stored "offboard", transmitting the mechanical force for example through a Bowden cable system [15] or pressurized air using a pneumatic system, or the force is generated directly on the suit. Rigid exoskeletons instead are using a rigid frame which is attached to the user. A obvious drawback of this system is the lack of adaptability to different users, they have be built modular or a new device for each size. To provide a compliant actuation characteristic, most rigid exoskeletons make use of compliant actuators. However, a massive advantage compared to its soft counterpart is the high possible torque output and the broad actuation bandwidth achievable for the system. Also force measurement is implementable in the exoskeleton hinges to provide a force controlled system instead of merely relying on EMG obtained values. There have been trials to build a rigid exoskeleton without any fixed hinges, instead using a pulley system. This is a possibility to get rid of the rigid connections, interfering involuntary with the wearer [8]. A significant drawback is the required force field control system because of the force feedback coupling between the single parts.

3.1.2 Actuators

Actuators can be differentiated in compliant actuators, or non-compliant actuators. Compliant actuation systems try to mimic human muscle properties with including some mechanic elastic element in series, like tethered systems using spring systems, more flexible bowdenBowden cable systems or even pneumatic artificial muscle systems where the compressor is linked over this flexible muscle to the exoskeleton. Another approach for actuator design is to use a direct coupled actuator, like a electric motor directly linked over a stiff structure to the human limb. This benefits the weight of the device, a drawback is the possible distal mass (inertia) and the need of a way more advanced control system, also safety is something to concern. Another drawback of including an elastic element in series is the reduced bandwidth of the actuation system. There have been studies using a "quasi -direct drive" method to couple the actuation system to the body, which are promising to have a high torque output and a suitable bandwidth while still providing a compliant character [36]. An also mentionable approach to design actuators is to use a variable stiffness systems, where the series elastic element can be adapted to each use case. Bandwidth of the force actuation is getting variable then.

3.1.3 Sensors and control systems

Sensing and predicting the motion a wearer is willing to do is an advanced topic. One common approach, predestined for exosuits, is to measure EMG muscle activities of the wearer. If the wearer tries to use a muscle voluntarily, there is a higher amplitude of muscle activity obtained, although also involuntary muscle activities will higher the amplitude and will generate a force input from the exoskeleton. In general robotics, there are different ways to control a robot. One way is to measure excerpted force, this will lead to another approach which is including a force measurement device on a rigid part attached to the body to sense the wearers intentions, or even to include such a torque sensor on the hinge of the rigid exoskeleton. A third common place to measure the wearers intention would be a force feedback under the wearers foot, like in a treadmill or in the shoe. This allows to differentiate between gait phases and to provide a right timed input. For the controlling of the actuation system itself, usually in the series elastic system is included a strain sensor to measure the torque exerted by the motor. However, another approach to control the system and provide a proper assistance would be a system to measure the position of the wearer in a defined room, this is called spatial controlling, but there have not be found a lot of literature investigating this method. Design goals Based on further knowledge of the current state of the art, a compliant powered actuation was considered as needed for human assistance purposes. Therefore following requirements have been defined as aims [3].

1. Safety: Exoskeleton is inherently coupled with the wearers body; Therefore fault proof safety is a crucial thing unacceptable to be not met
2. Comfort of wearing: System shall be comfortable to wear, the participant should like to don the suit to enable long term assistance
3. Low mass: Structure shall be portable, enhance the comfort and avoid energy waste
4. Interface: Interface sensing the motion predictions obtained to calculate movements of the exoskeleton shall be precise, therefore a accurate force feedback and motion sensing is necessary
5. Range of motion: Device shall enable the user to do at least a regular squat motion and bend the knee (assisted) about 110 degrees
6. Compensation for gravity: Any forces due to the mass of the device shall be compensated
7. Complexity: It shall be easy to maintain and to adapt for different user sizes, also this can be considered as a factor of cost
8. Donning and Doffing: Donning and Doffing shall be as easy as possible to provide a solution users can wear by themselves

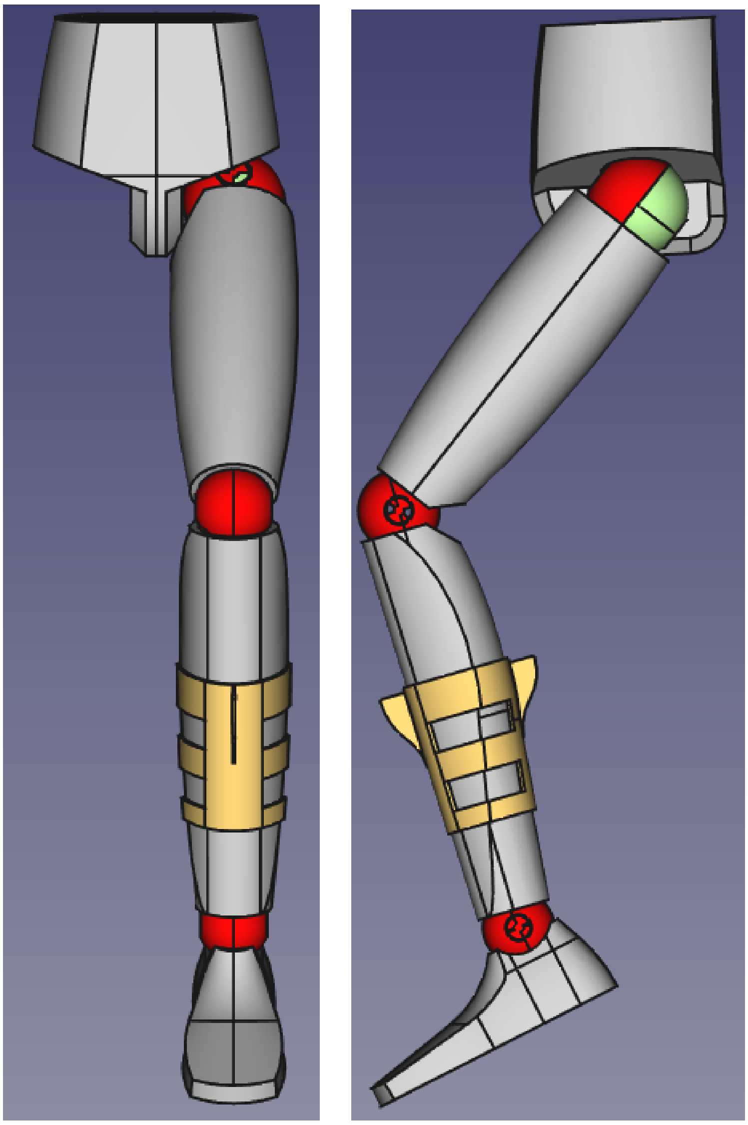

Figure 3.1: Exoskeleton design

3.1.4 Simulation

During Covid-19 quarantine an approach to simulate the kinematic and static forces resulting on a human leg has been investigated. Therefore a model, a mannequin, in human size has been drawn in a 3D- CAD application and a possible brace to attach to the human. Unfortunately there has not been any possibility to get the needed simulation software from the academic workspace on the personal computer, so this can be further investigated after the Covid-19 restrictions have ended. This model is depicted in Figure 3.1.

3.2 Development of a possible actuation system

To provide a sufficient force coupling to the human limb, there is an popular actuation system tested using pressurized air, named pneumatical artificial muscle or PAM. A widely used system is the braided McKibben muscle, named after the first time implementer. In opposition to a system with air cylinders which are including heavy parts like pistons and barrels as guidance this system is making use of a combination of elastic materials to reduce the weight and enhance wearing comfort for the patient. The system itself is built upon an elastic body material which is acting as an elastic bladder expanding when filled with pressurized air. Thus to the expanding in the vertical direction, the elastic material translates the volume expansion into a linear contraction movement, therefore no moving rigid components as seals or lubricants are needed. One important advantage of the implementation of a PAM is the high compliancy and therefore a muscle like behavior if the bladder is being extended in a linear direction, for example by a antagonist muscle system. The compliancy can be examined if the elastic base material is being stretched in a linear movement, the length extension does not cause an enlargement of force introduced on the limb by this muscle, it stays the same and is not related to any external extension forces. For such a system different fluids can be used, although air adds even more compliancy due to the possible compression of the medium. To provide an additional security feature, the inner membrane of the muscle is encased by a woven fiber shell. The diameter is chosen to prevent a toofar expansion of the inner bladder through providing this external housing restricting the volume expansion. The loose-woven nature of the outer shell provides a flexible solution to restrict contraction while still adding little weight to the actuator, nevertheless it should be considered using the PAM up the limit of contraction adds additional wear to the inner membrane, therefore it has been thought better to use an air muscle capable keeping about 25% reserve to the maximum contraction possible. Some drawbacks restrict the use of PAMs in industrial environments, such as the lack of precise position control possibilities, difficulties occurring with high speeds and repeatability plus the additional wear of the system which is causing high maintenance. Also the responsiveness is quite different compared to a pneumatic cylinder system, which use cases are long actuation distances with small forces while air muscles usually provide high forces over small contractions [1]. The theoretical working principle showing the inflation process of an air muscle is shown in Figure 3.2, there can be seen the length shortage with the according force F while the volume of the muscle is being increased using pressurized air. A commonly accepted formula is describing the relation between the diameters of the used membrane D_0, the Pressure P and angle of the braiding resulting in the force is, with regard to Figure 3.2,

$$F = (\pi D_0^2 P/4 \cdot sin^2\Theta_0)[3(1-\Delta L/L_0)^2 cos^2\Theta_0 - 1] \tag{3.1}$$

according to [33, 7, 12]. This formula is a rough estimation and does not include parameters like the wall thickness S and it is considered as a perfect cylinder. Due to the nonlinear material behavior, air compressibility and the complicated structure the building of more sophisticated models has been proven difficult. It can be said lower initial braid angles and higher muscle diameters yield to higher output forces, although the length of the muscle and the contraction status, the membrane material (for example rubber, or silicone based rubber) can lead to very different results [1].

Controlling such an actuator is difficult, a standard approach like a linear PID control can cause problems like phase-delay and unpredictable oscillation including overshoot, which is critical to user safety. There are approaches to use a combination of a sliding mode controller with a PID controller to get better results, called PSCM, Proxy-based sliding mode control [26]. Another variant is to implement a sliding mode function in a

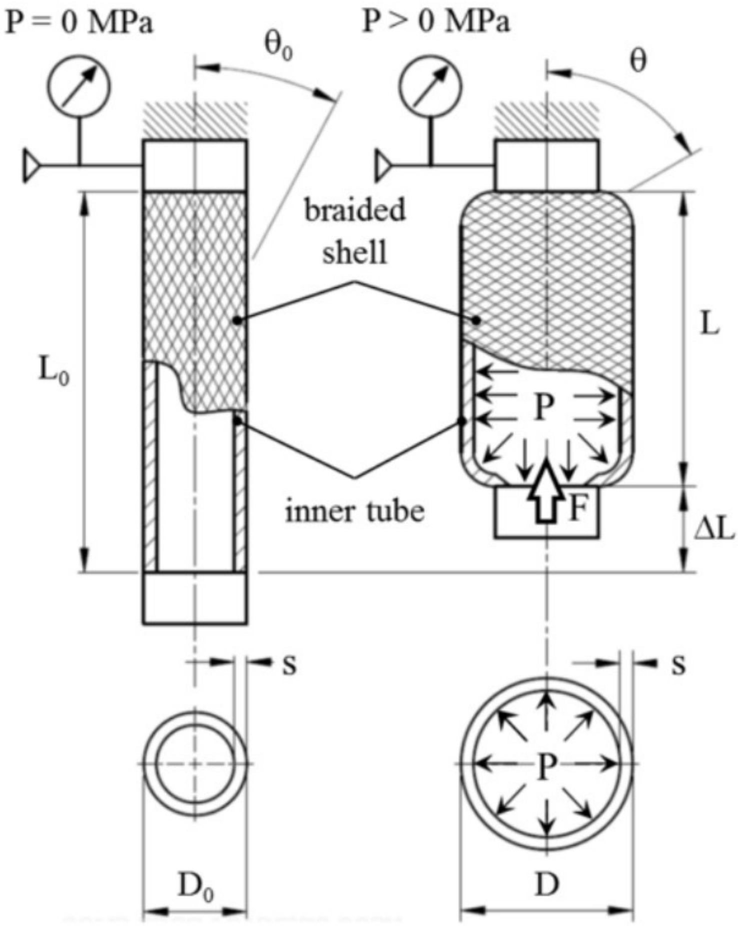

Figure 3.2: Length shortage or contraction pressurizing a PAM
[1]

model-based approach to get better results [26]. However, the feasibility to build such a system has been tested using standard materials, cheap and widely available to improve maintenance costs.

3.2.1 Implementation of a pneumatic actuation system

The aims of this system were to provide a cheap system build of widely available parts and to measure the performance of it, and the feasibility to implement it as a simple assistive force input for patients. As inner air membrane a 6mm x 9mm diameter Latex tube has been assembled into a expandable braided cable sleeve usually used for cable protection with a diameter of 12mm. As an input there has been used a pneumatic 4 x 6 mm diameter polyurethane hose which is mounted in the expandable air membrane. To close the front part of the tube and create air tightness, standard hose clamps and shrinking tubes have been used, there is also the attachment placed to connect it to rigid points. To evaluate this system, a standard 12V miniature diaphragm pump by Parker, the D732-23-01 membrane pump, is connected to the polyurethane hose. Pressure is sensed using a diaphragm pressure sensor by Honeywell, the 100PGAA5, the sensor is connected in parallel to the pneumatic muscle. To control the system a Atmel Atmega328p Evaluation board (Arduino Nano) is used, this is driving a custom built power stage build disconnected from the µC using a optocoupler for safety and a MOSFET switching device. This on the other side is driving a 24V PWM modulated signal to a solenoid valve, Festo CPE14-M1BH-3GL-1/8. Those are used in a 3 way 2 position variant.

Figure 3.3 shows the schematic of the test setup and gives an overview how the PAMs where implemented. Control of the device was implemented in the Arduino software, and user inputs are read using external button interface and a potentiometer to regulate the duty cycle on the fly of the PWM applied to the transistor stage. The PWM frequency output from the Arduino was set to 250 Hz which has been proven to provide the most stable solution for solenoid valves in fast switching environments [18]. A schematic of the software on the µC is shown in Figure 3.4. Basically the µC is checking if a button is pressed, and outputs a PWM signal of 250 Hz with

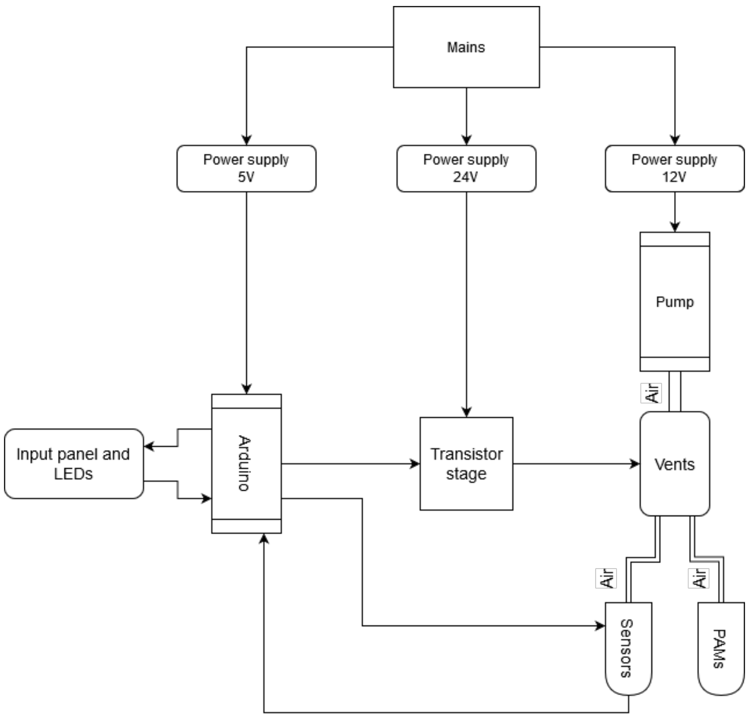

Figure 3.3: Schematic of the test setup to evaluate a pneumatic actuation system

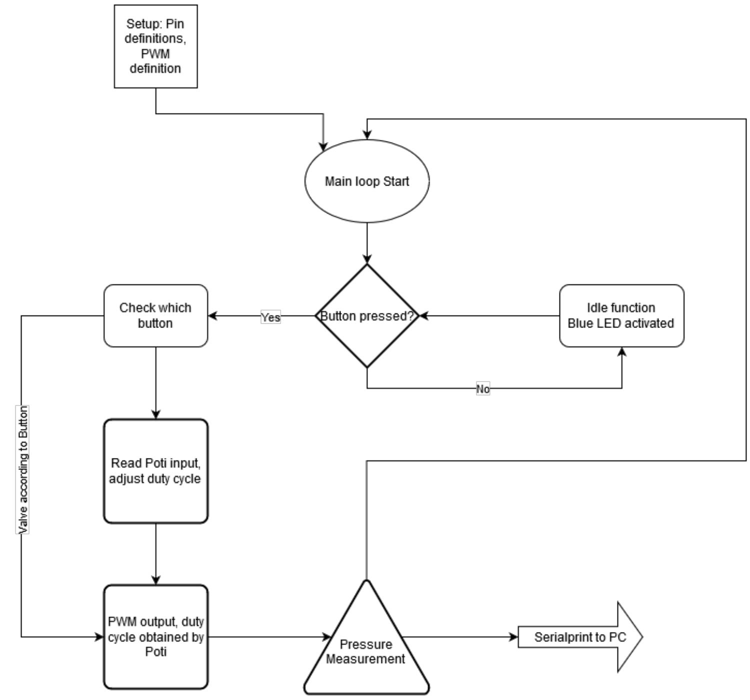

Figure 3.4: Software flow diagramm for PAM evaluation

the duty cycle obtained by a potentiometer at the according valve output pin.

The actuator itself was built with an initial length of $125mm$, when the pressurized air was applied it shortened to $105mm$ leaving a ΔL of $20mm$ or in relative percentage 16% of the initial length. It also has been found there is the need of a proper air tank system to be able to provide a steeper input response, the obtained pressure graph is shown in Figure 3.5. This graph shows the responsiveness of the actuation system when directly connected to an air compression system, it is shown the relative pressure surpassing the ambient pressure over time in seconds. To get a more advanced responsiveness with a stiffer curve as it will be needed in an exoskeleton there has to be included an proper air tank to increase fluid flow to the PAM. The regulation of the air flow has been stable on a wide range of the duty cycle, leaning towards the lower end the valve has shown unstable controlling what can be expected given the solenoid nature while

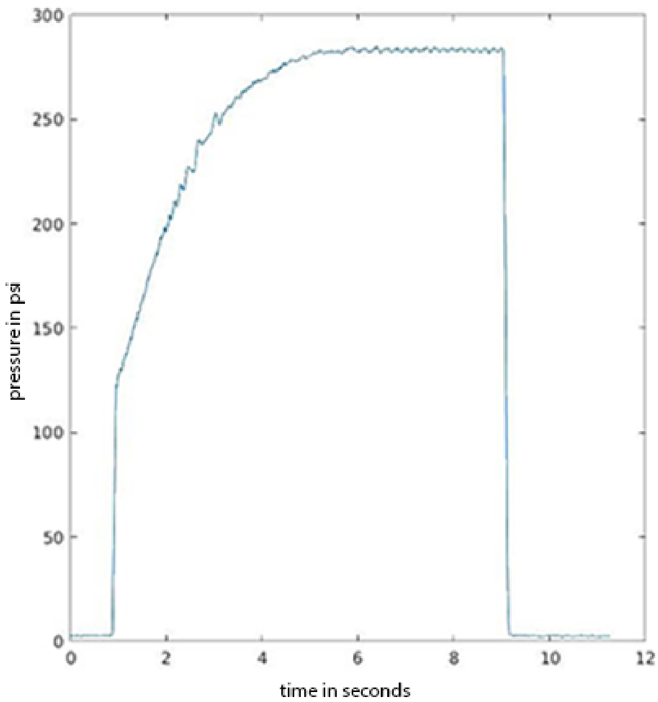

Figure 3.5: Pressure response in the PAM, measured with parallel capacitive sensor

driving the valve in full position mode.

4 Conclusion

There are several ways to improve overall life quality for patients suffering Stiff Knee Gait. The results in H-reflex adaption through proper training proticols could not be investigated, but there has been a proof of feasibility of an implementation of the training setup into a rehabilitation training setting. It is possible for experienced medical personnel to set up a patient for treatment in a few Minutes preparation time, considering the total amount of conservative training this should be easily implementable in a clinical rehabilitation setting. It is expected follow up sessions are needed every now and then to prevent the H-reflex excitability wandering back to pretraining setting. As alternative there has been research for mechanical solutions to provide a assisting device to enhance a patient's overall knee movements and gait possibilities. It has been shown there are many different possibilities to provide a actuation force with an external source to the lower limb, also have been stated different approaches to provide this. For customer friendly devices, respective donning and doffing of the device, flexible solutions have proven suitable. They also seem to be able to be produced with justifiable costs and provide a decent improvement for the wearer. Therefore a design example for a patient- actuation system link has been shown, with a flexible standard structure only few adaptions have to be made while transiting to a different patient. In terms of actuation there has one approach been tested, the PAM. Pneumatic muscles are lightweight solutions able to provide an impressive force output compared to stiff devices. Because of the necessary air supply system they seem to be suitable for repetitive movements over a longer period of time while staying in the same location, like gait training on a treadmill. Regarding their excellent compliance due to the inherent lack of stiff devices and air as a energy transmission system, pneumatic actuators have natural secure behavior and do not need as many advanced safety features as actuators based on electrical motors directly coupled over a gear system to the human limb.

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