Designing an experimental setup for transcutaneous stimulation to modulate sensory feedback.

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***Abstract*— Applying a specific form of transcutaneous electro-tactile stimulation on the ipsilateral leg might cause the contralateral soleus muscle to contract, thus creating plantarflexion. If this principle is used correctly it may bring forth a type of application that could help people with balance problems in locomotion possibly caused by a type of sensorimotor disorder. For this and similar experimental research, an application is necessary that can fulfill a series of requirements.**

Keywords—muscle spindle, interneurons, short-latency crossed response, Electro-tactile stimulation, motor unit coherence

I. INTRODUCTION

A. Motivation

To prevent elderly people, amputees, people with sensory nerve issues, and people with a type of spinal injury from falling possibly due to a chronic neurological disorder affecting postural stability and gait, an electro-tactile transcutaneous stimulation might help.

A substantial portion of balance control during locomotion is in an automatic fashion. Especially stretch feedback plays a crucial role in this automatic control since it closely involves muscle contraction. This system may impair elderly people, amputees, people with sensory nerve issues, and people with a type of spinal injury, in such a way that it emerges as a problem with balance in locomotion. There might be a possibility of correcting the problems that hinder their locomotion by modulating (or augmenting) sensory feedback between the legs. Applying transcutaneous electro-tactile stimulation on the sensory afferents of the ipsilateral leg causes the contralateral leg to change its muscle contractions [3].

In the case of an unexpected perturbation to one limb, applying the right amount and type of electro-tactile transcutaneous stimulation to the ipsilateral leg will cause the contralateral legs muscles to flex or extend in such a way that the body will be stabilized and perhaps prevent the person from falling and injuring themselves.

For further research on this topic, it is important to determine the appropriate stimulation paradigms and their effect on the nervous system and motor output. This research aims to design and develop an experimental interface to investigate stimulation paradigms.

B. Design Goal

To be able to execute such an experiment or carry out similar experimental research, it is necessary to create a specific application.

The design goal is to create an experimental setup that includes software and hardware at pipeline to tune the stimulation intensity and frequency online to stimulate the targeted nerve fibers (spindle afferents) selectively, acquire high-density EMG signals from contralateral muscle and joint torque data from a dynamometer and storing these data on hard drive with appropriate sampling frequencies for each modality. This interface should include real-time monitoring of triggered averages and the ongoing EMG signal.

II. THEORETICAL BACKGROUND

To assess more specifically what requirements the application should be able to fulfil, theoretical background research needed to be employed.

It is well established that muscle spindles, cutaneous receptors and tendon organ afferents deliver information from muscle, tendon and skin on muscle displacement, velocity, and force required by the central nervous system to control these variables and to switch between phases of movement [1].

Sensory mechanism:

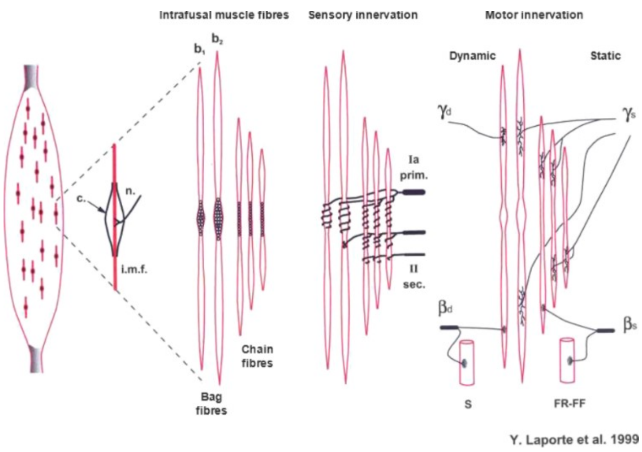


Fig. 1. "Diagrammatic representation of the mammalian muscle spindle. From left to right: parent muscle; capsule (c.) nerve supply (n.), intrafusal muscle fibre bundle (i.m.f.); typical repertory of intrafusal muscle fibres; sensory innervation showing the primary afferent (group Ia axon) innervating all intrafusal fibres and the secondary afferent (group II axon) innervating chain and bag2 fibres; motor innervation showing dynamic gamma axon restricted to the bag1 fibre and static gamma axon innervating chain and bag2 fibres. Beta innervations are

alpha motoneurons innervating both extrafusal skeletal muscle fibres and intrafusal fibres. Personal communication from Yves Laporte."[2]

Cross-spinal transmission of stretch reflex: Spindles in muscles provide feedback on muscle stretch. The identical (homonymous), antagonistic, and synergistic muscles receive this info in return. The same sensory information is also sent to the muscles in the opposing extremities, resulting in coordination between the arms and legs for balance. According to animal research, it is likely that circuitry connecting the muscles of opposite limbs exists at the spinal level.[3] Commissural interneurons(inhibitory interneurons) are a component of this circuitry. These come from the mid-lumbar region of the spinal cord, namely the dorsal horn of laminae IV, V, and VIII.

"sensory feedback elicited by tibial nerve stimulation on one side (ipsilateral) can affect the muscles activation in the opposite side (contralateral), provoking short-latency crossed responses (SLCRs)"[4], is a claim made by Gervasio, et al. According to the study of [4], which looked at whether contralateral afferent feedback was a part of the process regulating the SLCR in human gastrocnemius muscle, SLCR does have an impact on dynamic stability. This provides significant evidence for the existence of communication between opposing limbs.

Continuing on the comprehension of the communication between the opposite limbs, [4] illustrates certain neuronal pathways that are probably involved in the creation of the SLCR, as seen in Fig. 2.

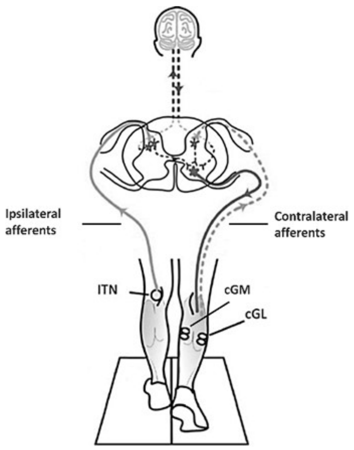


Fig. 2. "Neural pathways likely involved in the generation of the SLCR. SLCRs are elicited in the contralateral gastrocnemius medialis (cGM) and lateralis (cGL) following stimulation of the tibial nerve (iTN) in the ipsilateral leg during human walking. The current study was designed to investigate whether the activity of contralateral afferents (dashed gray lines) contributes to the SLCR. The gray and black full lines represent the ipsilateral afferent pathways and contralateral efferent pathways involved in the generation of the SLCR, respectively. Supraspinal contribution to the SLCR is not excluded (dashed black line). All other dashed lines in the spinal cord represent unknown pathways."[4]

The stimulation of the tibial nerve is depicted in [4] and Fig. 2, and [5] goes into greater detail about one potential arrangement for this stimulation.

To find the correct position for electro-tactile tibial nerve stimulation it is first necessary to understand what Hoffmann's reflex(h-reflex) and M-wave are and how they relate to the stimulation intensity.

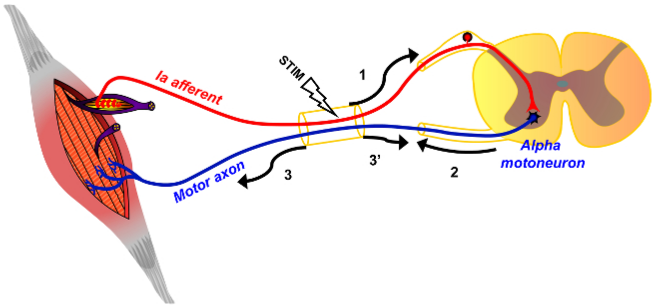


Fig. 3. "Motor and reflexive pathways activated by nerve stimulation."[11]

The explanation for Fig. 3 is as follows: "Electrical stimulation of a mixed (motor/sensory) nerve (STIM) induces a depolarization of both motor axon and Ia afferent firing. Depolarization of Ia afferents towards the spinal cord activates an alpha motoneuron, which in turn evokes an H-reflex response (pathway 1+2+3). Depending upon the stimulus intensity, motor axon depolarization evokes a direct muscular response: M-wave (pathway 3). At maximal M-wave intensity, an antidromic current is also generated (3') and collides with reflex volley (2). This collision partially or totally cancels the H-reflex response."[11].

The development of response amplitudes between the H- and M-waves differs with increasing stimulus intensity. The M wave gradually rises till reaching its maximum intensity whereas the H-reflex gradually decreases until it completely disappears from the EMG signal for a subject at rest[11]. In Fig. 4 the development of M-waves and H-reflex over the stimulation intensity can be seen.

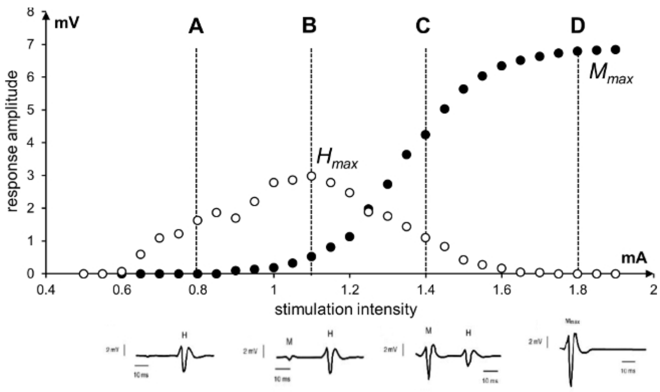


Fig. 4. "Typical recruitment curves at rest."[11]

To explain Fig. 4, [11] says the following:" Amplitudes of reflex responses (H-reflex, white round) and direct muscle responses (M-wave, black round) with increasing stimulus intensity. Bottom panels present typical traces at four progressively increased intensities (from A to B). (A)weak intensity, evoking only an H-reflex response. (B) Intensity providing the maximal H-wave amplitude (Hmax). (C) At intensity beyond Hmax, the collision between antidromic and reflex volleys induces a decrease in H response amplitude. (D) At Mmax intensity, H-reflex is totally cancelled and M-wave reaches a plateau."[11].

III. THE LINK BETWEEN DESIGN ASPECTS AND THE THEORY

According to [5], the metal cathode of the stimulation probe was positioned on the popliteal fossa, and the anode was a dampened 12 cm by 8 cm pad put over the patella. [4] adds to this by stating that to determine the best spot for the electrode placement, [4] delivered stimuli every 4 to 7 s and the cathode's position was changed until a stable M-wave was seen in the ipsilateral SOL muscle (iSOL) muscle, with little contamination from the stimulation artefact. Applying this principle in the design will assist to find the optimal position for the electrodes.

[6] offers details on what kind of signal can be utilized to precisely stimulate the cutaneous mechanoreceptive units and to minimally obstruct the vestibular system with regard to the sort of signal that should be modified. [6] referred to the biphasic sinusoid's use (1-ms biphasic symmetric sinusoid released at 100 Hz). The use of the biphasic sinusoid is justified by the fact that it causes less discomfort than a monophasic waveform and prevents charge accumulation because the first pulse charge is discharged by the second pulse, according to the justification given in [6]. Additionally, [8] and [9] support the application of biphasic pulse stimulation. The stimulation frequency (100 Hz) used by [6] was within the physiological discharge rate of the cutaneous mechanoreceptive units and within the range of useful frequencies for electro-tactile stimulation for sensory replacement in rehabilitation. Additionally, biphasic pulse stimulation induces a sensory adaptation in as little as 15 minutes. This suggests that biphasic stimulation is a worthwhile addition to the intended application.

According to [6], several types of stimulation are required when focusing on various areas since the type of stimulus used is specific to mechanoreceptor stimulation. [6] gives the example of galvanic vestibular stimulation and explains how this form of stimulation would be accomplished utilizing constant current stimuli provided through the mastoid process or white noise-type stimuli, which have been utilized to create postural effects. Additionally, evoked reflex responses can be triggered by stimulations with low or high bandwidths (mechanical perturbation: 0–4 Hz or electrical stimulation: 0–75 Hz), which are still distant from the stimulation frequency employed in [6] (100 Hz). Another method to elicit cutaneous reflex responses may be found in [10], which uses a Gaussian vibration stimulus composed of random frequencies between 0 and 50 Hz. Since the application's aim is electrical stimulation, it follows that frequencies between 0 and 75 Hz and at least one frequency with gaussian modulation will be employed for evaluating the application.

Using the information from the two paragraphs above, the range and types for the stimulation design can be devised and included in the design to stimulate precisely and painlessly.

The motor neuron pool discharge responses from the contralateral muscle are a good indicator of the effectiveness of the modulation signal that was supplied. In order to conduct this study, the raw EMG signal from the contralateral soleus muscle must be divided into individual motor unit action potentials. [5] provides an explanation of the EMG breakdown technique during stimulation. Including this method in the design will enable us to comprehend how muscle spindle input modulates in common oscillation.

IV. DESIGN ASPECTS

As a base for the creation of the desired application, an already existing MATLAB code with similar functionality to the desired application was used.

Based on the acquired theoretical background the following requirements have been decided, for the application to be useful in experimental research as described in I.B:

a. Creating stimuli with intensity in the approximate range of 0-25 mA. For finding stable M-wave/H-reflex and stimulation at different percentages of the acquired amplitude for max M-wave/H-reflex[4].

b. Creating stimuli with frequencies in the approximate range of 0-100 Hz more specifically 0-1Hz, 15 Hz, 25 Hz and 60 Hz. For finding stable M-wave/H-reflex, and exciting the soleus muscle spindles by stimulating the tibial nerve.

c. Creating stimuli with gaussian frequency modulation. For evoking cutaneous reflex responses [10]. For the reason that Gaussian frequency modulation can minimalize the perception of the stimulation and thus could help in creating applications that would be easier to use by people as it is less(non) intrusive.

d. Reading and displaying bipolar EMG data. By using a bipolar electrode pair and displaying the signal over time, it can be seen that after stimulation occurs it will be possible to find stable M-wave/H-reflex by looking at the EMG data. This EMG data over time also shows the delay between the two, as M-wave is closer in time to the stimulation than the H-reflex, which can then be used to discern the two [4].

e. Creating monophasic and biphasic stimulation pulses. Monophasic pulse for stable M-wave/H-reflex procedure [4] and biphasic pulse for generating less discomfort compared to a monophasic waveform and prevents charge accumulation [6].

f. Reading and displaying high-density surface electromyography (HDsEMG) data. During the experiment, the display of the muscle activity from every individual channel can be verified to see if the electrode was correctly placed. After the experiment, the raw EMG signals can be assessed by decomposing the signals into single motor unit action potentials, which can be analyzed by, for example, the mean discharge rate.

g. Reading and displaying torque data. For visual feedback for subjects. To make it possible in experiments to maintain isometric ankle plantarflexion at 20% of their maximum voluntary contraction (MVC) force it is necessary to show the subject feedback showing how much force to perform. This was done by having a dot move up and down depending on the percentage of max MVC and a constant line at 20% of their maximum MVC. If something about the EMG data seems wrong this data can be examined to look for potential abnormalities.

h. Saving all EMG and torque data. For assessing and decomposing the raw EMG signals as explained in f and g [7].

V. Design of Control and Acquisition Interface

The requirements that have already been met by the base MATLAB code were a, f, g and h.

For d by changing the base code, one of the displays for HDsEMG could be used to instead show the bipolar EMG data as to show M-wave/H-reflex.

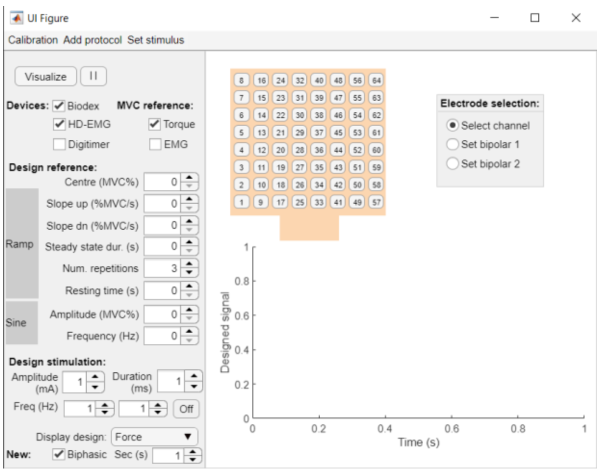
To clarify, the parameters of trigger signals were similar to the delivered stimulation. This has been measured through an oscilloscope examining the frequency and amplitude, upon which with the use of Ohm's Law was verified that the current indeed had the assigned amplitude.

When tuning the intensity and monitoring H-reflex amplitude it was possible to see an increase and decrease in its amplitude. However, depending on the subject there were some more difficulties in ascertaining the increase/decrease of the H-reflex because of noise from other areas than the targeted soleus muscle. This was most likely due to differences in muscle build or electrode positioning.

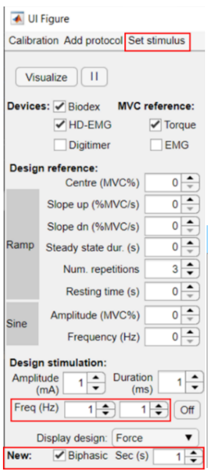
When using biphasic stimulation none of the subjects reported pain.

The resulting application following all requirements from **Fout! Verwijzingsbron niet gevonden.** is a MATLAB code that when run shows the following user interface(UI):

A:



B:



C:

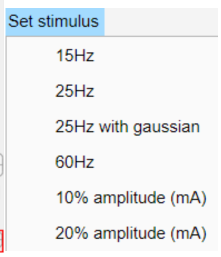


Fig. 5. MATLAB UI.

In Fig. 5 the MATLAB UI can be seen. The functions on the UI that originate from the already existing tool as seen in Fig. 5A are the following:

The “Calibration” button, this button Calibrates the Biodex and max MVC.

The “Add protocol” button adds a protocol name and creates a save file.

The “Devices” button Turns available devices on or off for data acquisition.

The “MVC reference” button Turns available MVC references on or off.

The “Design reference” button creates a function that will be displayed for the subject in the same figure as the percentage of MVC dot with the instruction to apply the right amount of force to follow the function.

The “design stimulation” button constructs the stimulation based on the inputted Amplitude(mA) and duration of the (monophasic)pulse(ms) (in the case of a Biphasic pulse the pulse takes twice as long).

In Fig. 5B elements have been included/altered to fulfil the requirements of the system.

A menu called “Set stimulus” was included for ease of use. Using this menu button the frequency and the amplitude of the stimulation can be set to the required setting for the protocol.

The “design stimulation” was changed to use frequency(“Freq (Hz)”) instead of the period.

An option was included for creating stimuli with gaussian frequency modulation Input: A (Hz), B (Hz) $|A-B|*\text{randn}+\min(A,B)$ (randn returns a random scalar drawn from the standard normal distribution) thus $|A-B|=\sigma$ and $\min(A,B)=\mu$. E.g. in the case of 25 Hz with gaussian where A=25 and B=26, σ=1 and μ=25

A checkbox was included for mono/bi-phasic stimulation(“□Biphasic”).

The option “Sec (s)” was included for changing the approximate(+ 0.2-1.0 s delay due to MATLAB code processes) stimulation train length (s) (Sec will also add a break between stimulation trains, which will be of the same length as the pulse train. E.g. if Sec = 1, a 1-second pulse train and 1 second without stimulation will alternate.)

In Fig. 5C the “Set stimulus” menu options are shown. Based on the validation experiment the following settings were coded in for easy access:

“15 Hz”, “25 Hz”, “25 Hz with gaussian” and “60 Hz”. These options set the stimulation frequency directly.

“10% amplitude (mA) ” and “20% amplitude (mA) ”. These options change the current “Amplitude (mA)” value to 10% or 20% of its current value.

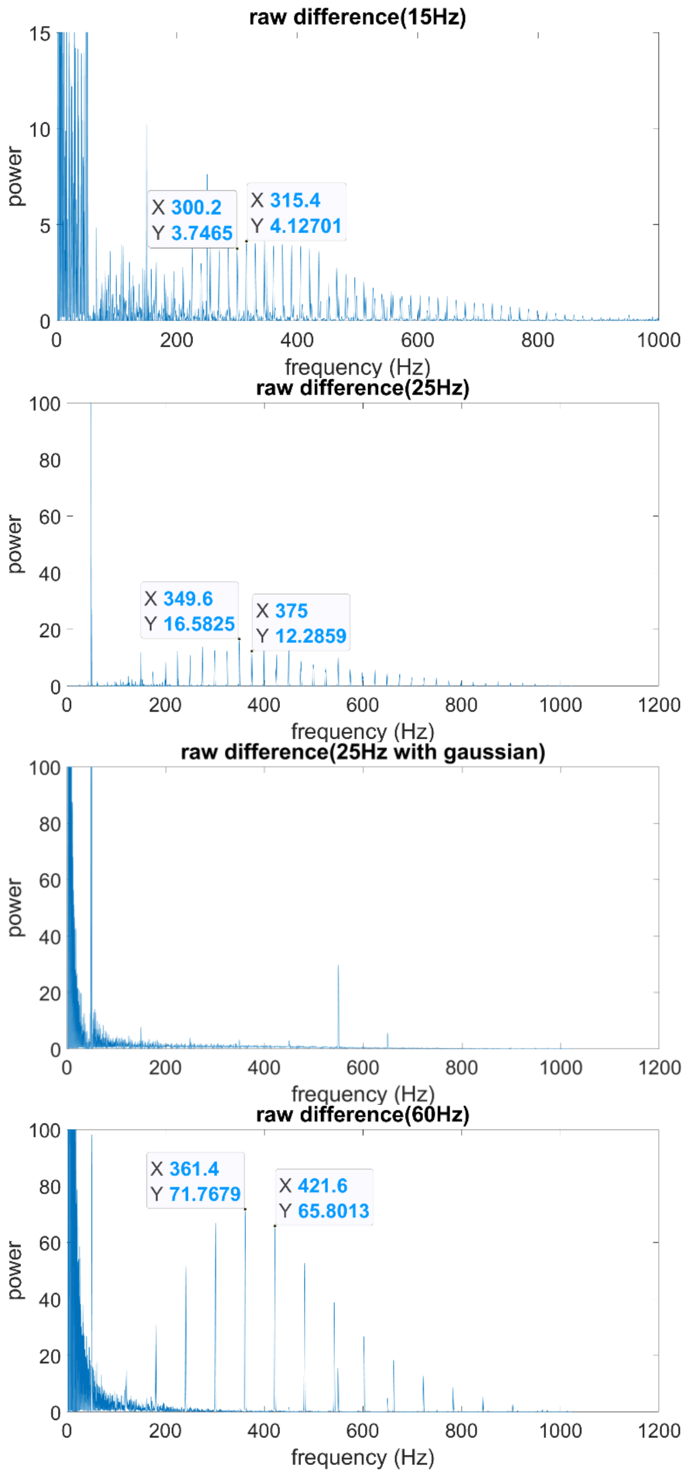


Fig. 6. Comparing HDsEMG data. “raw difference” shows the difference between the data as recorded from the experiment with and without stimulation(15 Hz, 25 Hz, 25 Hz with gaussian, 60 Hz).

From Fig. 6 it can be seen that there exist some stimulation artefacts with an interval of approximately the stimulation frequency, thus proofing that the stimulation was correctly applied to the subject as it was constructed in the model for the frequencies of 15Hz, 25Hz and 60Hz. For the stimulation “25 Hz with gaussian” there is no clear stimulation artefact as the frequencies are variable around 25Hz.

VI. Testing Functions of Components.

A. *Problem*

A set of experiments was conducted to test the functioning of the software interface, connectivity between apparatus, and their control paradigms. Furthermore, subject opinions were collected for the newly designed stimulation paradigms to clarify whether or not the subjects perceived pain during stimulation.

In this validation, several parameters were examined and the range in which these parameters were tested during the research was adjusted based on the findings of related work.

The parameters that were examined are the amplitude of the pulse and the frequency of the pulse.

B. *Apparatus and Instrumentation*

To stimulate the tibial nerve an anode (9 cm × 5 cm) and cathode (3.5 cm × 2 cm) were placed over the patella and popliteal fossa, respectively. At each combination of stimulation intensity and frequency, high-density surface electromyography (HDsEMG) signals of contralateral (non-stimulated side) Soleus muscle was recorded using a semi-disposable 64-channel electrode (TMSi).

Electrical stimulation was delivered to the ipsilateral tibial nerve using the Digitimer DS7*, an electrically isolated constant current stimulator which is CE certified. Stimulations were delivered as a square waveform or Gaussian noise with frequency modulations. The biphasic current was used as the alternating direction of current flow largely avoiding the chemical build-up at the electrode/skin interface and preventing pain due to stimulation [6][8][9]. Multichannel EMG signals of the soleus muscle were measured on the unstimulated side during isometric contraction. The HDsEMG signal was acquired using a 2x64 channel TMSi Refa EMG acquisition system. Simultaneously, the oscillations in plantar flexion force(torque) were measured using a load cell (Biodex system) to estimate the influence of force output on the unstimulated joint. The HDsEMG and force data were stored on a hard disk. For outputting the appropriate signal to make the stimulator create the stimuli and read the data from the EMG acquisition system and the Biodex and send the data to the MATLAB application the NI USB-6003 DAQ USB Device was used.

C. *General experimental setup*

The participants were seated in a comfortable position on a chair. While the knee of the contralateral (unstimulated) leg was positioned at an angle of approximately 120° and the knee of the lateral (stimulated) leg was positioned at a comfortable angle of approximately 180°. The right (contralateral to the stimulated leg) foot was positioned on a footplate and the left (ipsilateral) foot was supported by a cushion.

Fig. 7. Positioning of legs during experimental setup.

To stimulate the tibial nerve an anode and cathode were placed over the patella and popliteal fossa, respectively. After this, the skin at the electrode location (lateral Soleus muscle and lateral ankle) was cleaned by shaving, and dirt was removed with scrub followed by alcohol to obtain low impedance and a bipolar electrode pair was placed on the cleaned skin as well as a reference electrode that was placed on the ankle. The most suitable stimulation position was found by finely sliding the cathode electrode on the skin and checking the visual feedback from the MATLAB code on a clear M-wave/H-reflex.

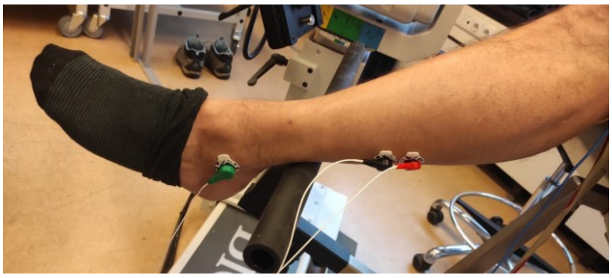


Fig. 8. Positioning of bipolar electrode pair and reference electrode during experimentation.

Once the most suitable stimulation position was found, the bipolar electrode pair was removed to not affect the HDsEMG, the skin at the electrode location (contralateral Soleus muscle and contralateral ankle) was cleaned by shaving, and dirt was removed with a scrub followed by alcohol to obtain low impedance and a semi-disposable 64 channel electrode was placed on the cleaned skin as well as a reference electrode that was placed on the ankle.

Fig. 9. Positioning of a semi-disposable 64-channel electrode and reference electrode during experimentation.

The subjects were then asked to perform isometric ankle plantar flexion contraction at maximum voluntary contraction (MVC) for 5 seconds then relax for 5 seconds and repeat two more times. For the MVC, the subjects were instructed to press down on a footplate so the soles of the feet remained flat on the footplate to ensure that the force produced, was a result of the rotation at the ankle joint.

During the experiment, the subjects were instructed to maintain isometric ankle plantarflexion at 20% of their maximum voluntary contraction (MVC) force. The subjects were shown visual feedback to assist them in maintaining a contraction force at fixed MVC.

D. *Experiment*

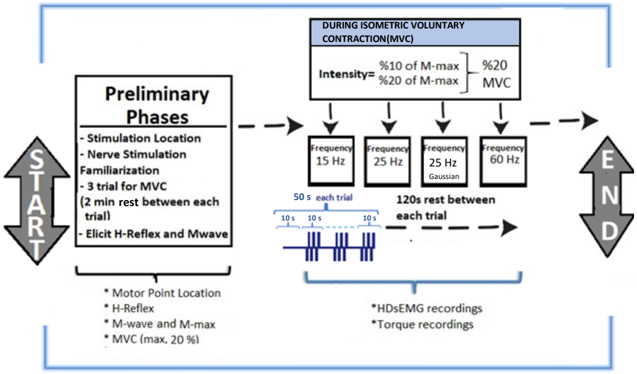


Fig. 10. Experimental procedure to test each possible set of stimulation paradigms, stimulation control and signal acquisition interface.

As can be seen from Fig. 10 the experimental procedure consists of eight different trials. For every trial the stimulation occurred in a ten-second pulse train followed by ten seconds without stimulation, this continued till three ten-second pulse trains had been executed and were then followed by 120 seconds of rest before starting the next trial.

The first four trials were performed at an intensity of 10% of the amplitude that was used when a maximum M-wave was found and for every individual trial the frequencies 15 Hz, 25 Hz, 25 Hz with gaussian frequency modulation and 60 Hz were used respectively.

For the next four trials, an intensity of 20% of the amplitude that was used when a maximum M-wave was found was used and for every individual trial the frequencies 15 Hz, 25 Hz, 25 Hz with gaussian frequency modulation and 60 Hz were used respectively.

E. *Statistical analysis*

Outside of the created application can the data recorded from the experiments be analyzed to give some more meaningful results for experimental research. Firstly filtering the raw HDsEMG signal as can be seen in Fig. 11.

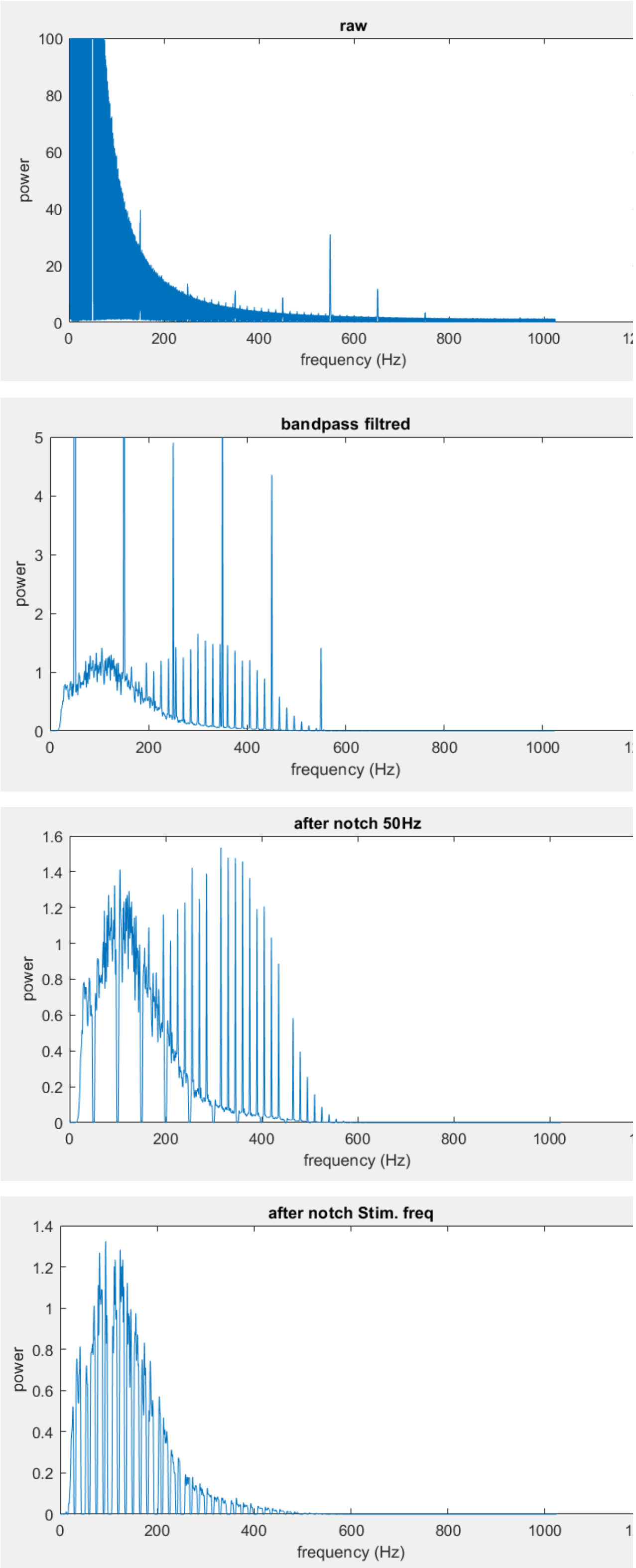


Fig. 11. Filtering HDsEMG data. “Raw” shows the data as recorded from the experiment. “Bandpass filtered” shows the data after it has been bandpass filtered. “after notch 50 Hz” shows the signal from “Bandpass filtered” after it has been notch filtered at 50 Hz to remove noise from the city line power. “after notch Stim. freq” shows the signal from “after notch 50 Hz” after it has been notch filtered at the frequency of the stimulation to remove stimulation artefacts. The data is filtered in these three steps to make the individual motor units stand out as much as possible, as otherwise, the decomposition algorithm will pick up the higher amplitude noise instead of the lower amplitude motor units.

The filtered signal can then be decomposed into single motor unit action potentials, which can be analyzed by, for example, the mean discharge rate.

F. Experimental Results

Some example results can be seen in Fig. 12 that show that the created application can be used for similar experimental research.

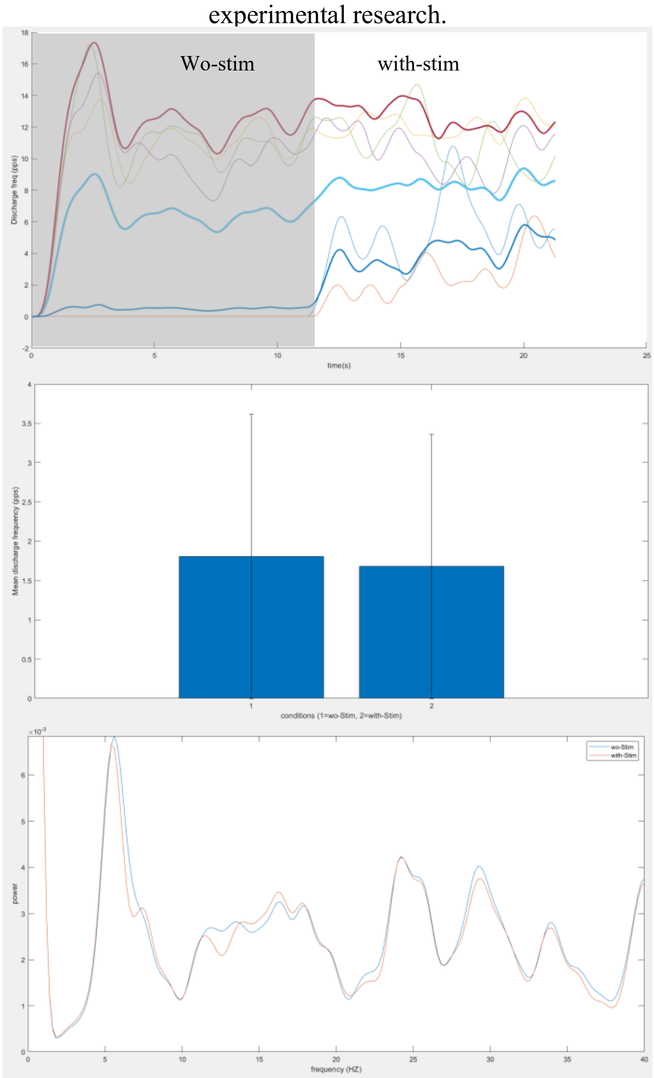


Fig. 12. Experimental results. Discharge freq (pps) over time (s), Mean discharge frequency (pps) for conditions (1=wo-Stim, 2=with-Stim) and power over frequency (Hz) (power spectrum of the compound spike trains of motor units) respectfully.

Furthermore, from the collected subject opinions it can be concluded that the stimulations for the before-mentioned paradigms did not result in any perceived pain during stimulation.

VII. Discussion

In the creation of the application some challenges needed to be overcome for the application to meet the set requirements. To make it possible for the application to create stimulation pulses in the range from 0-100 Hz it was necessary to prepare a full pulse train of the desired length instead of constructing the signal pulse by pulse, as there was some background computing occurring after every construction which caused a delay of approximately 0.3 s. In turn, the stimulation frequency did not reach higher than approximately 3 Hz.

To create the option of having stimuli frequencies with gaussian frequency modulation, the code needed to be altered to work with frequency instead of the period as the calculations that would have been needed to get the same result as for using frequency as input were much more complex.

While experimenting some other challenges also came to light. One such challenge was that the quality of the available electrodes used for stimulating was non-ideal, making it harder to find the correct position of the tibial nerve. This can easily be prevented in future research by ordering a better electrode in advance.

Another major challenge resulted from having to use the Biodex near the stimulator, as the electromagnetic noise from the large motor(s) and the relatively unshielded stimulator connection to the NI DAQ USB Device caused the stimulator to not function properly as well as caused noise on the measurements from the EMG acquisition system. To resolve this problem the cables connecting the stimulator and the NI DAQ USB Device needed to be better protected by only using a short coax cable that had to be split on one end to fit in the NI DAQ USB Device port. This can however be improved upon by shielding the NI DAQ USB Device or stimulator better.

In future work on this application, it would also be advisable to examine more the Creation of the stimuli(intensity, frequency, etc.) and add a description of exactly how much the stimulation diverges from the expected/desired as this might influence the results.

For reading and displaying and saving data, it might be beneficial to verify if no data is lost or wrongly recorded due to inefficient code, hardware or software limitations.

This application could be expanded upon to be able to test in real locomotion conditions instead of this isolated setup, as there might be other factors worth taking into account when applying similar experimental research to real-life applications.

VIII. Conclusion

In conclusion, it can be said that the created application, although not perfect in all aspects would still more than suffice for the type of experimental research similar to the experiment performed in “Testing ”.

Acknowledgement

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