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Master's Degree in Biomedical Engineering

A biomimetic and multichannel FES implementation for lower body rehabilitation protocols

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Summary

The hip and knee joints are among the most mobile joints in the body and play an important role in supporting body weight and in allowing daily activities such as walking and climbing stairs. Due to accidents or pathologies, like osteoarthritis, such joints can be damaged. In severe cases, where motor functions are limited, the patient must undergo an arthroplasty surgery, which must then be followed by a rehabilitation program for the recovery of lost muscle strength.

The aim of this project is to define two rehabilitation protocols for patients with hip and knee prostheses that can be implemented by FES. To achieve this goal, an event-driven closed-loop system for FES real-time control, designed in previous works, was used. This system allows to acquire surface ElectroMyoGraphy (sEMG) signals from a subject (therapist) through wearable acquisition boards and translate them into stimulation patterns to provide to another subject (patient) through surface electrodes in real-time. However, to optimize the acquisition sensors circuit and power consumption, instead of using all the information carried by the sEMG signal, a feature is extracted through the Average Threshold Crossing (ATC) technique. The latter is based on generating an event whenever the sEMG signal exceeds a certain threshold. By averaging the number of TC events within a time window, the ATC value is computed. Each time the therapist performs a movement, variations of ATC parameters of each muscle involved are detected and translated into modulations of current intensity (I) to send to the patient.
In this thesis, two improvements to the system software are proposed: the possibility of combining the individual muscles' activations, involved in the execution of a single movement, to generate a single stimulation pattern and the possibility of providing the patient with an alternative stimulation mode, pulse width modulation (PW). Then, the procedure followed to outline every single movement of the two protocols is described: which muscles were considered and how the relative signals were combined to obtain the activation and stimulation profiles and how the surface electrodes were positioned on the patient's muscles. Each protocol was tested on 3 subjects and for each movement, the two types of modulation (I and PW) were compared in terms of movement execution, patient discomfort and muscle fatigue. These data were obtained through forms filled out by patients at the

end of each test, in which questions were asked about the discomfort experienced during the stimulation phase, the possible onset of fatigue during the repetitions of a movement and the level of muscle fatigue perceived at the end of the session.

The results show a preference of 50-100% for PW modulation in the case of movements that involve keeping the limb in a certain position for a few seconds, as it allows to generate a greater contraction force. For movements that include small muscles, a preference of 66.7-83.3% has almost always been expressed for I modulation since it allows the patient to perform the movement without too much discomfort. Finally, the relationship between fatigue perceived during 10 movements and the charge delivery to the patient in each repetition, which is calculated by adding the charges of each pulse included in that repetition (single pulse charge delivery = IxPW of the pulse), was studied. If fatigue is perceived from a certain repetition onward, this also marks the beginning of an increase in the charge delivery to the patient.

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Chapter 1

Introduction

The purpose of this thesis is to outline a rehabilitation protocol for patients with hip prostheses and one for patients with knee prostheses both achievable through FES (Functional electrical stimulation). The hip and knee joints are among the most mobile joints in the body and play an important role in supporting body weight and in allowing daily activities such as walking and climbing stairs. Due to accidents or pathologies, like osteoarthritis, such joints can be damaged. In severe cases, where motor functions are limited, the patient must undergo an arthroplasty surgery, which must then be followed by a rehabilitation program for the recovery of lost muscle strength and a condition of complete autonomy. Such a program of muscle strengthening and movement recovery can become faster and more effective when normal physiotherapy is combined with FES. The latter, in this thesis project, has been exploited using an event-driven closed-loop system for FES real-time control, designed in previous works. This system allows to acquire surface ElectroMyoGraphy (sEMG) signals from a subject (therapist), through wearable acquisition boards, and translate them into stimulation patterns to provide to another subject (patient) through surface electrodes, which cause the contraction of the muscles on which they are positioned. Thanks to this system, the patient, helped by electrostimulation, is able to perform the 11 different movements defined for each protocol.

1.1 Skeletal Muscle

Skeletal, cardiac, and smooth muscle tissue make up the three main types of muscle tissue. Skeletal muscle is responsible for voluntary movement and belongs to the category of striated muscles as its fibers are crossed with a regular pattern of fine red and white lines, which give the muscle a distinctive striated appearance. The skeletal muscles are attached to the bones through the tendons and contribute significantly to multiple bodily functions. From a mechanical point of view, the main role of skeletal muscle is to convert chemical energy into mechanical energy to generate strength and power, determine the movements of the skeleton and maintain the posture and position of the body. Skeletal muscle also provides structural support and protection from external trauma to soft tissues, such as the abdominal wall and pelvic floor. From a metabolic point of view, the functions performed by skeletal muscle include the production of heat for the maintenance of body temperature and the regulation of entry and exit of material. The orifices of the digestive and urinary canals are surrounded by bundles of skeletal muscles, which allow voluntary control over swallowing and urination. The skeletal muscle also acts as a storage source for amino acids that can be used by different organs of the body for synthesizing organ-specific proteins. Skeletal muscle comprises about 40% of total body weight, contains 50-75% of all proteins, and accounts for 30-50% of the protein turnover of the whole body. It consists of 75% water, 20% protein, and 5% other substances, such as inorganic salts, minerals, fats and carbohydrates [1][2]. Muscle mass is determined by the balance between protein synthesis and degradation. Both processes are influenced by factors such as hormonal balance, physical activity, nutritional status, injuries or diseases.

1.1.1 Anatomy of the skeletal muscle

Skeletal muscle is formed by a set of long, cylindrical cells with fusiform ends, called muscle fibers that are grouped into fascicles. Between the different fascicles run elastic fibers, nerves, and blood vessels, which branch off to distribute to the various fibers. The entire muscle mass, that is the set of all the fascicles, is covered by a layer of dense and irregular connective tissue, called epimysium. Each fascicle is surrounded by the perimysium and, within each fascicle, the individual muscle fibers are wrapped in the endomysium, Figure 1.1. Each muscle fiber is composed of hundreds of nuclei that are located just below the sarcolemma, the cell membrane that lines the fiber. The intracellular cytoplasm is instead called sarcoplasm and contains hundreds of mitochondria and myofibrils. The activity of the former, together with the cleavage of glycogen granules (also contained in the sarcoplasm), provides the Adenosine TriPhosphate (ATP) necessary for effective muscle contractions. Myofibrils, on the other hand, are cylindrical formations as long as the entire fiber, which, having the ability to actively shorten, represent the formations responsible for muscle contraction. Around each myofibril, there is a coating, the sarcoplasmic reticulum, which is responsible for accumulating the calcium necessary for contraction. Myofibrils are in turn made up of parallel myofilaments composed mainly of two proteins, actin, and myosin. Both actin filaments (thin filaments) and myosin filaments (thick filaments) are organized into repetitive units called sarcomeres [4]. Figure 1.2 shows the structure of a

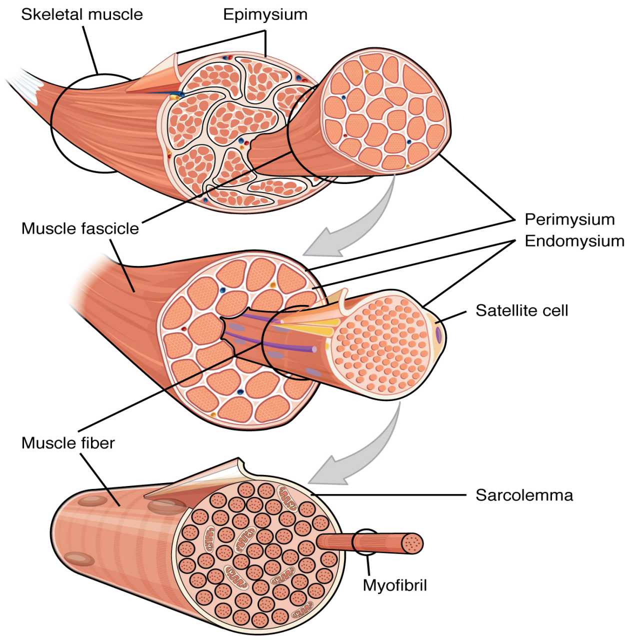

Figure 1.1: The three connective tissue layers [3].

sarcomere. The thick filaments are located in the central zone and are joined by proteins that form the M line or disc. The thin filaments are located at the ends of the sarcomere, marked by the Z lines or discs, and expand towards the M line. In the overlapping zone, thin filaments run between thick ones. Along the major axis of the myofibril, a striated pattern can be observed, due to the regular alternation of light and dark bands. The dark bands are called A bands and include the H band (thick filaments only) and the overlapping zone. Light bands are called I bands and include only thin filaments. Band A and Band I are divided in half by the H zone and Z line respectively [5]. Each thin filament has a spiral pattern and is referred to as F-actin. It consists of 300-400 molecules of G-actin, each of which contains an active site for the myosin head on the thick filament. A thin filament also contains the associated proteins tropomyosin and troponin. Tropomyosin molecules form a long chain that covers the active sites on the actin, preventing actin-myosin interactions during rest. Troponin binds to tropomyosin and helps to position it on the actin molecule. Troponin also binds to calcium ions and when this happens (shortly before contraction), troponin changes its conformation, allowing tropomyosin to move away and uncover the binding sites on actin. Once the tropomyosin is removed, the cross-bridges are formed between actin and myosin, triggering the contraction [3]. Thick filaments consist of a bundle of about 500 myosin molecules, each of which consists of a double-strand characterized by an elongated tail and a free globular head. The myosin molecules

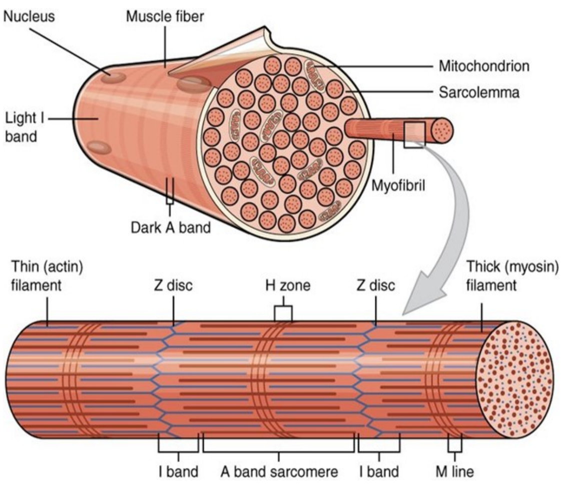

Figure 1.2: Structure of a sarcomere [3].

are oriented in order to expose the heads at the end, while the tails are grouped towards the center. At the time of contraction, it is the myosin heads that create the cross-bridges with the binding sites on actin, Figure 1.3 [4].

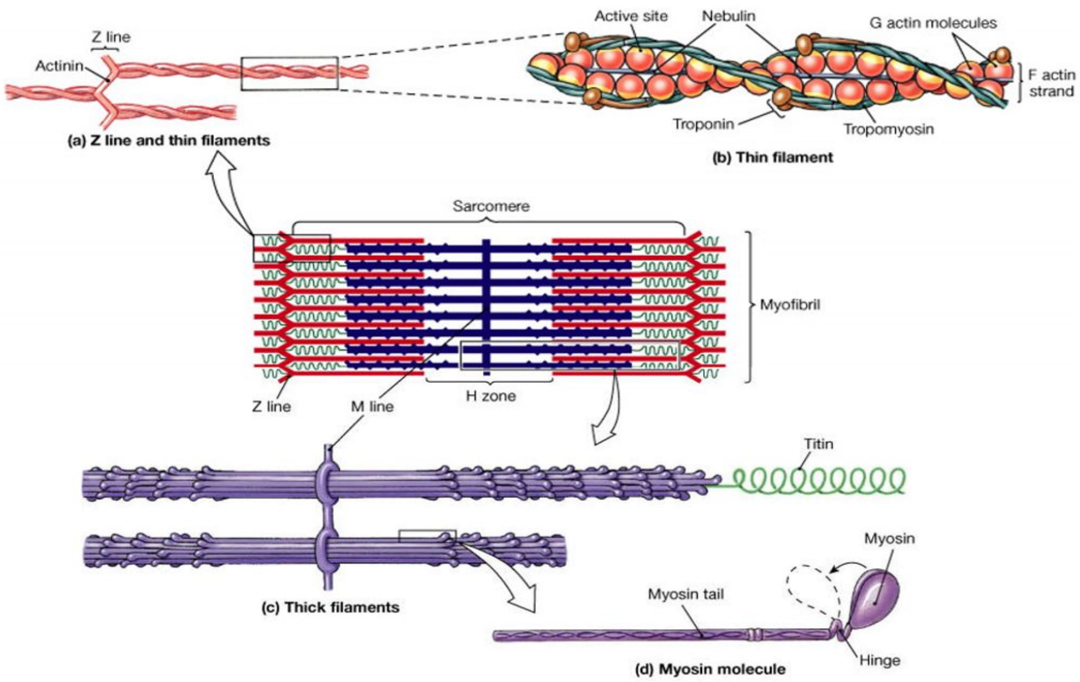

Figure 1.3: Composition of thin and thick filaments [4].

1.1.2 ATP and Muscular contraction

The contraction is the result of the interactions between thin and thick filaments of each sarcomere and is explained by the sliding filament theory. When a sarcomere contracts, the thick and thin filaments do not change in length. Thin filaments slide over thick ones, moving towards the center of the sarcomere. Therefore, during a contraction, the dimensions of the H and I bands decrease and those of the overlapping zone increase, the Z lines approach each other, and the width of the A band remains constant, Figure 1.4. The motion of muscle shortening occurs

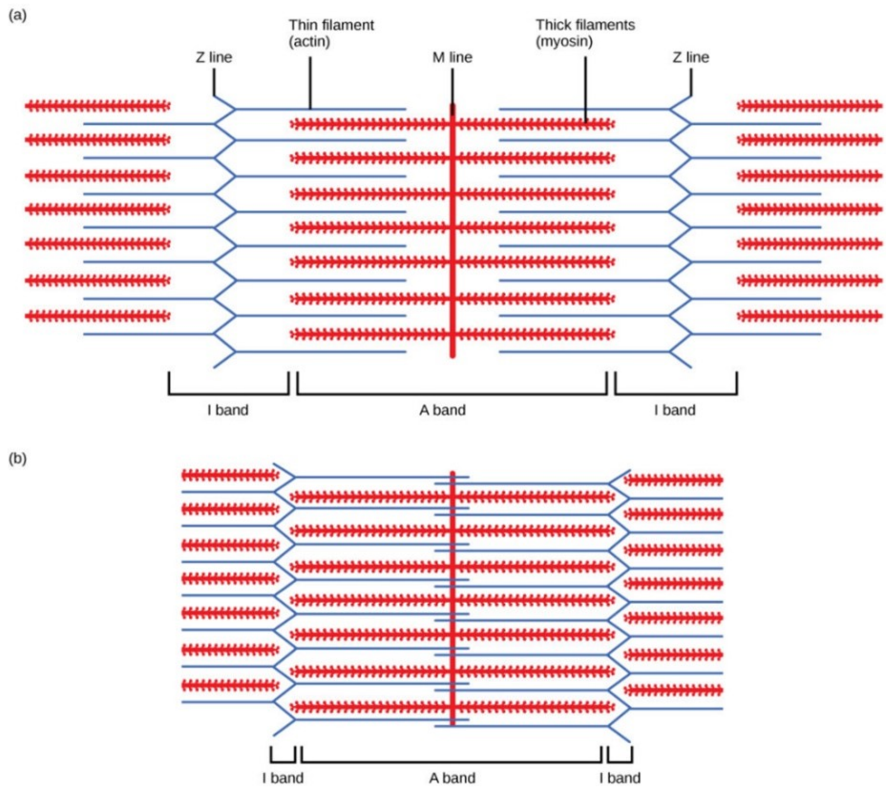

Figure 1.4: When (a) a sarcomere (b) contracts, the Z lines approach each other, the I band gets smaller and the A band stays the same width [6].

as myosin heads bind to active sites of G-actin and pull the actin inwards. This action requires energy, which is provided by ATP. Each myosin molecule is endowed with two binding sites, one, as already mentioned, for actin and one for an ATP molecule, at which enzymatic activity hydrolyzes ATP to Adenosine DiPhosphate (ADP), releasing an inorganic phosphate molecule and energy. The anchoring of ATP in the specific binding site on the myosin head leads to the detachment of the latter from the G-actin molecule, to which it was bound by the previous contraction. ATP, bound to the myosin head, is hydrolyzed to ADP and phosphate

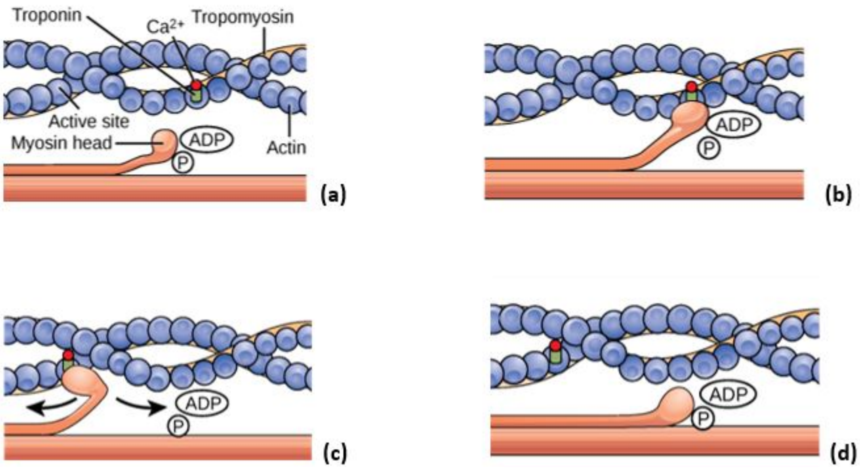

Figure 1.5: Cross-bridge muscle contraction cycle [6].

(Pi), and both products remain attached to this location, Figure 1.5a. The energy released during the hydrolysis of ATP induces a rotation of the myosin head, which is now in the high-energy conformational state ("cocked" position) and has a strong affinity for actin. At this point, if the actin-binding sites are covered, myosin remains in this high-energy configuration with ADP and Pi still attached. If, on the other hand, the binding sites are uncovered, then the cross-bridges between myosin and actin will form, Figure 1.5b. The Pi is then released, allowing the myosin to spend the immense energy in a conformational change, generating the "power stroke": the myosin heads move towards the M line, pulling the actin along with it. This movement causes a shortening of the sarcomere and the contraction of the muscle. After that, the myosin head also releases the ADP molecule and returns to its conformational state of low-energy, but the cross-bridges are still present and therefore it remains anchored to actin (rigor status, Figure 1.5c). The moment a new ATP molecule binds to the myosin head, it detaches from the actin and then a new cross-bridge cycle can start again and further muscle contraction can occur, Figure 1.5d [6].

1.1.3 Excitation – Contraction Coupling

Excitation-contraction coupling is the transduction of the appearance of an action potential in the sarcolemma in a muscle contraction. As already mentioned, the event responsible for initiating the contraction is the appearance of free calcium ions in the sarcoplasm, which are released from the sarcoplasmic reticulum when a neural signal arrives. Each skeletal muscle fiber is controlled by a motor neuron, whose cell body is located within the central nervous system, while the axon is carried to the periphery to reach the neuromuscular junction of the fiber in question. The end of the axon at the junction is called the synaptic terminal, which contains small secretory vesicles filled with acetylcholine (ACh) molecules. The latter is a neurotransmitter, a chemical used by a neuron to communicate with another cell. The synaptic terminal is separated from the motor endplates of the muscle fiber by a small space called the synaptic cleft, which contains the enzyme acetylcholinesterase (AChE), capable of degrading acetylcholine molecules. When an electrical impulse reaches the synaptic terminal, acetylcholine is released into the synaptic cleft and binds to receptors located on the motor plate, which open sodium channels to allow Na+ ions to pass through the cell. The entry of positive charges raises the potential inside the cell, leading to a depolarization. If the depolarization reaches the activation threshold of -60mV, then an action potential is generated. The initial situation is then restored through a repolarization phase, which brings the potential back to its rest value of -70mV [3]. The action potential spreads, like a wave, along the entire fiber through the invaginations of the sarcolemma, the T-tubules, to the area near the terminal cisterns, which are the calcium ion reservoirs of the sarcoplasmic reticulum. When the action potential arrives, there is an interaction between the receptors on the T-tubules' membrane and those on the surface of the terminal cisterns, the result of which is the opening of calcium channels [7]. The Ca2+ ions then flow outside the sarcoplasmic reticulum and bind to troponin, triggering contraction as described above. The generation of the action potential ceases when acetylcholinesterase hydrolyzes acetylcholine. At this point, the sarcoplasmic reticulum reabsorbs the calcium ions, and therefore the concentration of these in the sarcoplasm decays. When the concentration of calcium ions returns to normal levels, the troponin-tropomyosin complex returns to its original position, covering the binding sites and preventing the formation of further cross-bridges. No further filament sliding can occur in their absence, and thus the contraction ends. Muscle relaxation then occurs, with the fiber passively returning to its resting length [4].

1.1.4 Control of muscle contraction

A single motor neuron's set of muscle fibers constitutes a motor unit. Skeletal muscle contracts when its motor units are stimulated, and its contraction force depends on the frequency of stimulation and the number of motor units recruited. As described above, the contraction of a muscle fiber occurs when, as a result of the propagation of the action potential in the T-tubules, calcium ions are released, which then bind to the troponin, thus allowing the formation of cross-bridges. However, the amount of ions released is very small and they remain inside the cytoplasm for such a short period that not all cross-bridges are formed. As a result, the contractile strength produced by a single action potential is relatively small. If,

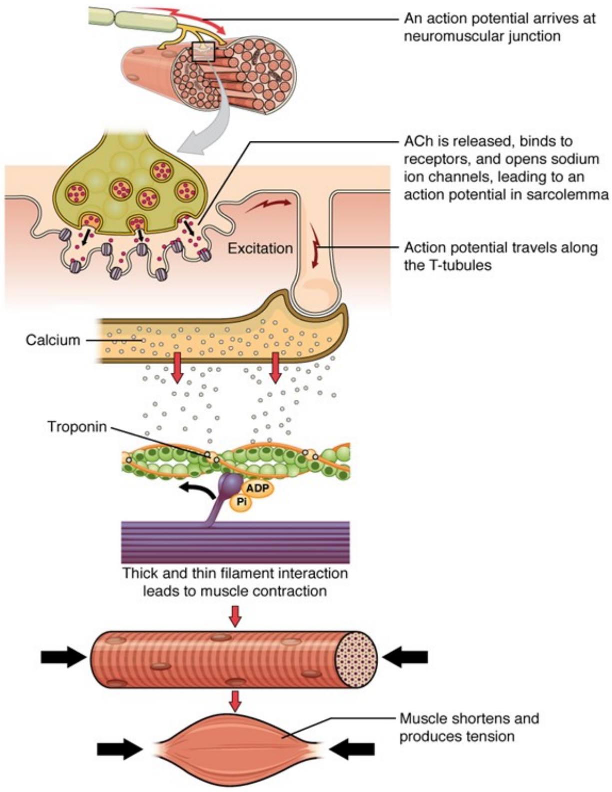

Figure 1.6: Cascade of events during excitation-contraction coupling in a skeletal muscle [3].

however, another action potential arises before the calcium ions are reabsorbed, and freed from the previous action potential, then the number of cross-bridges that are formed increases, and a higher contractile force is therefore developed. Then, by increasing the frequency of stimulation, the tension produced increases until it reaches a maximum value at which it then stabilizes (tetanus contraction) [8]. The intensity of the force exerted by the entire muscle increases with the number of motor units recruited. The order in which motor units, and also muscle fibers, are activated depends on the size of the motor neurons. Initially, motor neurons with small cell bodies are recruited, and thus are the slow fibers, and as the contraction increases, motor neurons with large cell bodies are added, with also fast fibers [9].

1.1.5 Types of skeletal muscle fibers

Muscles can perform a wide range of actions thanks to the different types of fibers that compose them [4]. Type I fibers are the fibers that contract more slowly, with a contraction time 3 times greater than the one of type IIB fibers. The mitochondria of the fibers use oxygen to produce ATP (aerobic metabolism) and can continue to generate it throughout a contraction. For this reason, type I fibers can contract for prolonged periods without showing fatigue but generating a low power of contraction. Type IIA fibers are associated with rapid contractions and, since they take advantage of both aerobic and anaerobic metabolism, they fatigue much faster than type I fibers. Type IIB fibers show the highest contraction speed (about every 0.01 s) and can develop greater forces. However, these powerful contractions require large amounts of ATP, which do not always manage to be produced by the few mitochondria present. For this reason, the main source of ATP is anaerobic glycolysis, which does not exploit oxygen to produce energy, but rather the glycogen. However, glycogen stores are limited and as a result, these fibers become fatigued very quickly.

1.1.6 Actions of Muscles

Muscles are classified according to their actions during a contraction as agonists, antagonists and synergists [10]. The agonist muscles, also called primary, are the ones responsible for the realization of a movement. For example, the biceps brachii is the agonist muscle that produces the flexion movement during elbow flexion. Antagonist muscles are those whose actions are opposite to those of agonists. When an agonist contract to produce a movement, its antagonist is stretched, helping to control the fluidity and speed of the movement. In the example of the elbow flexion, the antagonist of the biceps (agonist) is the triceps, located exactly on the opposite side of the biceps. While the biceps involves flexion of the elbow, the triceps, in addition to stabilizing the flexion, allows the extension of the elbow. The synergistic muscles participate in the action of the agonist muscles, strengthening their insertion or stabilizing their origin. Synergistic muscles become important at the starting of the movement when the agonist is stretched and therefore produces a small force. An example is shown in Figure 1.7. During the elbow flexion, the brachioradialis and brachialis act as synergistic muscles, helping the biceps pull the forearm towards the shoulder. The rotators cuff are also synergistic muscles as they stabilize the shoulder joint, allowing the biceps to exert greater force.

1.2 EMG

The Electromyography (EMG) signal is a representation of the electrical potentials generated by the depolarization of the outer membrane of muscle fibers and is detected by intramuscular or surface electrodes. In case of intramuscular recordings, the electrodes and signal sources (muscle fibers) are very close, so the filtering effect of the tissues, on the EMG signal, is very small. In contrast, for surface recordings, where electrodes are applied to the skin and not directly to the fibers,

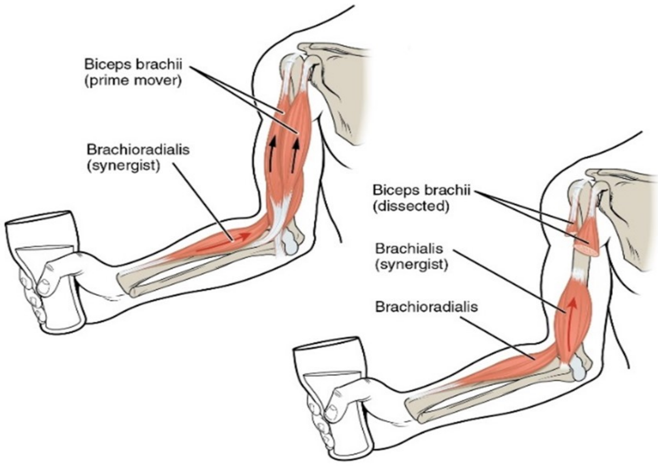

Figure 1.7: Example of agonist and synergistic muscles [10].

biological tissues have an important low-pass filter effect on the signal. From the EMG signal, it is possible to extract several features, among which one of the most relevant is the conduction velocity (CV) for its relationship with muscle fatigue.

1.2.1 EMG Signal Generation Model

When an action potential is generated, it propagates like a wave along the muscle fiber. Considering a single fiber and a pair of bipolar electrodes applied to the fiber or skin and connected to a differential amplifier, the waveform of the output signal reflects the position of the electrodes and their distance from the depolarization wave. The area of the motor endplates, where depolarization begins, is the point of the signal source. As shown in Figure 1.8, if the electrodes are positioned to the right of the depolarization source, then the wave will first meet the positive electrode and then the negative one. As the potential approaches the positive electrode, the distance between it and the depolarization front is reduced, and the amplitude of the output signal grows until it reaches a maximum that corresponds to the point where the wave is under the electrode (minimum positive electrode-depolarization wave distance). The wave then moves away from the positive electrode to approach the negative one. The amplitude of the signal then decreases until it is reset at the point where the potential is equidistant from the two electrodes. Finally, as for the positive electrode, as it approaches the negative one, the amplitude of the output signal decreases until it reaches a minimum value (minimum negative electrode-depolarization wave distance). The signal then grows until it is reset again as the potential moves away from the negative electrode. On the other hand, if the pair of electrodes is located to the left of the source, then the output signal will be characterized by a first negative fluctuation and then a positive one [11]. Extending this concept to the entire motor unit (MU), the pair of electrodes

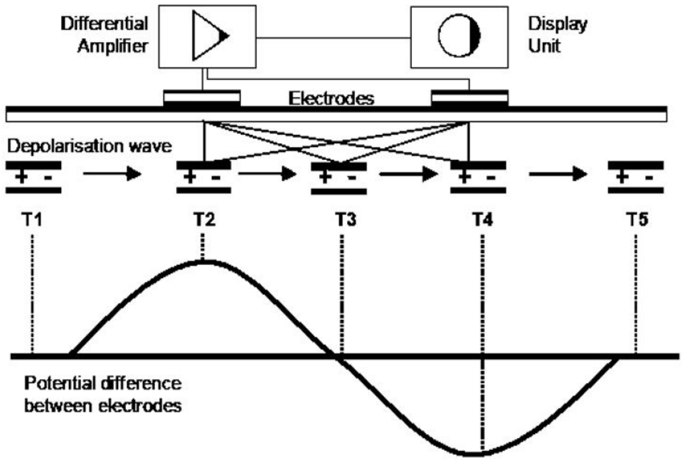

Figure 1.8: Output EMG signal model [11].

detects the potentials generated by all the fibers, which differ from each other in synchronization and amplitude. The farther the motor endplate is from the sampling points, the greater the time delay with which the potential is detected. While, as far as the amplitude of the signal is concerned, this will be greater, the more the distance between the electrodes and the single muscle fiber is low. The resulting signal is the sum of all individual contributions and is called motor unit action potential (MUAP, Figure 1.9) [12]. When the electromyographic activity

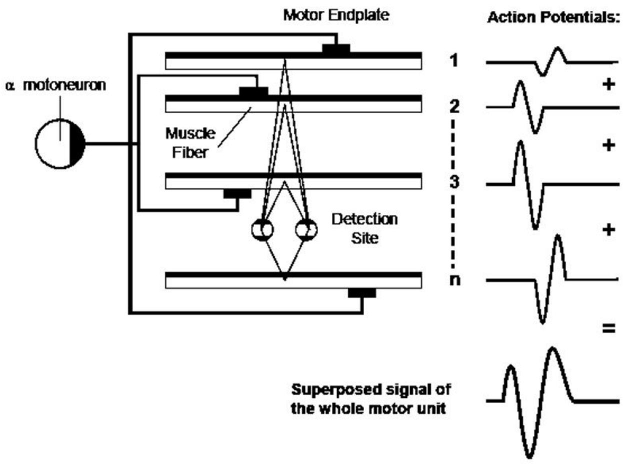

Figure 1.9: Generation of MUAP [11].

of a specific muscle is recorded, the superpose of all MUAPs of the muscle in question is detected and this signal is called *interference pattern*. The EMG signal is stochastic, therefore it is highly unlikely that the same motor task, performed with the same force, will result in the same signal patterns (non-reproducibility) [13].

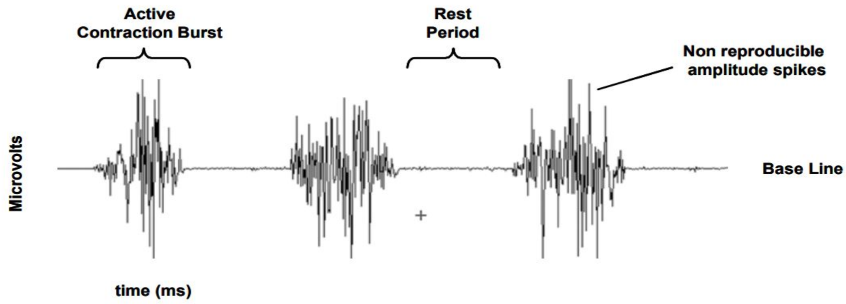

Figure 1.10: Example of raw sEMG recording [14].

1.2.2 Intramuscular and Surface Electromyography (sEMG)

The EMG signal can be detected by needle and fine-wire or surface electrodes. In the first case, the electrodes are inserted directly into the fiber. MUAPs are detected in a small volume around the tip of the needle and since the distance between the electrodes and the source is therefore minimal, the signal is less affected by noise due to biological tissues. This technique is then used to obtain localized information about both superficial and deep muscles. However, it is an invasive sampling method that can be annoying for the patient. For this reason, surface electromyography (sEMG) is very often used, in which instead the electrodes are simply applied to the skin. The main disadvantage of this technique, however, is the presence of biological tissues between the electrodes and the source (volume conductor), which increases noise on the output signal [12].The volume conductor acts as a low-pass filter on the EMG signal, and the more its thickness increases, the lower the amplitude of the signal taken. This is why the sEMG is not efficient to study the activity of deep muscles [15]. The presence of this filter reduces both the amplitude and the frequency band of the sEMG, as can be seen from Figure 1.11. The frequency band extends from 1-10 Hz up to 500 Hz, for the sEMG signal, and up to 1000 Hz for the invasively taken EMG signal.

1.2.3 Acquisition Configurations

Depending on the number of detection electrodes used, three sampling configurations can be identified [12] [16]:

Figure 1.11: Example of intramuscular and surface EMG signals recording. sEMG signal is smoothed and reduced in amplitude by the filtering effect of the subcutaneous tissues [13].

- *Monopolar*: a signal detection electrode and a reference electrode are used. This is a very noise-sensitive method as all signals close to the detection area are also detected (Figure 1.12a);
- *Bipolar or single differential (SD)*: is the most common configuration. Two acquisition electrodes are used, positioned 1-2 cm from each other, while the reference one is placed in a non-active area. The two detection electrodes are connected to a differential amplifier, which amplifies the difference between the two signals and suppresses the surrounding noise (Figure 1.12b);
- *Multipolar*: in addition to the reference electrode, more than two detection electrodes are used, placed at 1-2 cm from each other, and the signal taken passes through several differential amplifiers. This type of configuration is mainly used to reduce signals from nearby muscles (crosstalk), limit the detection volume, and increase selectivity. In particular, if three acquisition channels are used (double differential (DD) configuration), it is possible to determine the conduction velocity of the signal in the fiber (Figure 1.12b).

1.3 Functional Electrical Stimulation (FES)

Functional electrical stimulation (FES) is a technology that aims to generate a movement that mimics the normal voluntary contraction for the recovery of the functionality of the upper limbs, lower limbs and respiratory system. More specifically, it consists of applying low-energy electrical impulses to excitable tissues to perform or replace functions that have been lost due to neurological, spinal cord or musculoskeletal system damages. FES systems can be applied in different areas. They are used with paraplegic patients both for the recovery of motor activities and

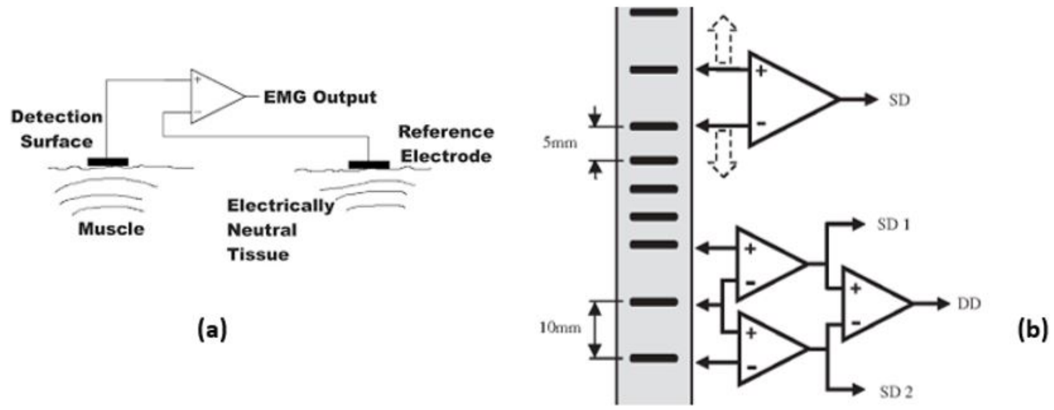

Figure 1.12: Three acquisition configurations. (a) Monopolar sampling [17]. (b) On top, bipolar or single differential sampling; on bottom, double differential sampling [18].

vital functions, such as breathing, and for the recovery of sensory functions, such as sight and hearing [19]. Another field in which FES is widely used is rehabilitation. Many patients who have suffered damage to the musculoskeletal system, such as ligament rupture or surgery, in the first phase of rehabilitation, are unable to generate sufficient muscle strength to move. This condition can lead to muscle atrophy and consequent problems in carrying out daily actions such as climbing stairs or getting up from a chair. To prevent this from happening, patients undergo FES, which helps in strengthening the affected muscles [20]. Electrical activation of muscle tissues requires at least two electrodes that produce a flow of current. These electrodes can have a monopolar configuration, in which an electrode, called active, is placed near the peripheral nerve to be stimulated and the second electrode, the return electrode, is located on a tissue that is not excitable. The second type of configuration is the bipolar one, in which both electrodes are located near the muscle to be stimulated. The electrodes used in FES are divided into [21]:

- *Transcutaneous or superficial*, applied to the skin. These are connected via cables to a stimulator, which can be worn by the patient. Being non-invasive, they are very easy to apply in a clinical setting;
- *Percutaneous*, placed directly on the muscle to stimulate. They can activate even very deep muscles and generate localized contractions. The active electrode is inserted into the muscle through a needle and is connected with cables, which come out of the skin, to an external stimulator. The return electrode is superficial;
- *Subcutaneous*, implanted in the body along with the stimulator to which they

are connected. They are used especially for long-term treatments.

1.3.1 Mechanism and pulse waveforms

Electrical impulses, transmitted by stimulating electrodes, depolarize the cell membranes of innervated neurons. If the depolarization reaches the critical threshold, an action potential is generated that propagates to the peripheral nerve endings and then, through the neuromuscular junction, is transferred to the muscle fibers, causing the contraction. FES applications for motor functions require stimulation to take place at the level of the nerve and not at the level of the muscle. This is because the critical threshold to be reached to generate an action potential in neurons is lower than that needed to produce potential in muscle fibers. So, for FES to work, the lower motor neurons, neuromuscular junctions, and muscle tissue mustn't be damaged. The FES consists of the application of rectangular electrical pulse trains described by the three fundamental parameters amplitude, pulse width (single pulse rate) and frequency. The waveform of these pulses can be monophasic or biphasic, as shown in Figure 1.13. In the first case, the waveform has only one phase, usually cathodic, which is repeated. Whereas in the second case, the wave is characterized by a first negative phase, cathodic, which generates the action potential in neurons, and a second positive phase, anodic. The second pulse balances the charge injected into the tissue from the first phase, reversing the potentially harmful electrochemical processes triggered by the first pulse [22].

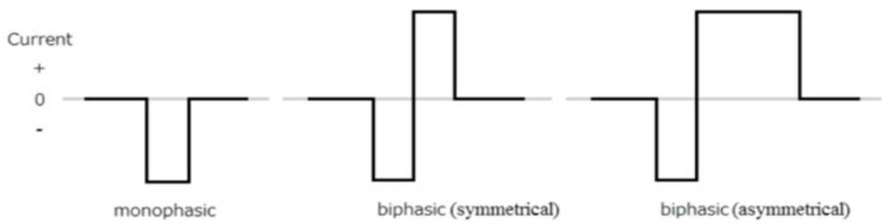

Figure 1.13: Representation of monophasic, symmetrical biphasic and asymmetric biphasic waveforms [22].

1.3.2 Muscle Fatigue

One of the main limitations of FES is the rapid fatigue of the stimulated muscles. Among the reasons that explain this phenomenon, there is the tendency of stimulation to recruit motor neurons with a large diameter, which have a lower activation threshold and which innervate the fast fibers. These fibers, type II fibers, are able to generate very strong contractions, but they are also not very resistant to fatigue. In addition, in patients with spinal cord injury (SCI), the problem of fatigue is aggravated by the transformation of slow, fatigue-resistant fibers into fast fibers due to atrophy [23].The graph in Figure 1.14 shows the percentage variation of the force of contraction over time related to the frequency of stimulation. Stimulations interspersed with a longer period, or at low frequencies, generate forces that tend to remain constant at the maximum value for fairly long periods. On the contrary, stimulations at high frequencies cause an almost immediate decay of the contracting force [24].

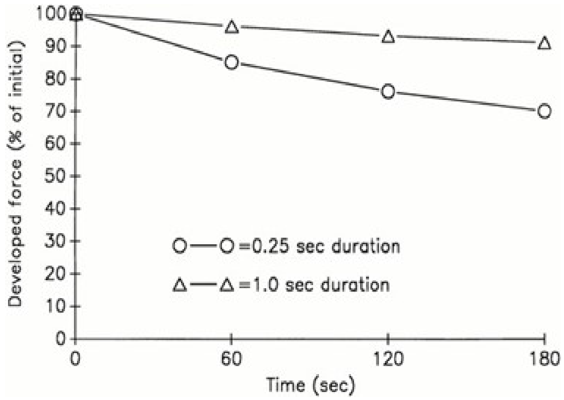

Figure 1.14: Percentage of force developed over time depending on the frequency of stimulation [24].

1.4 Anatomy of the hip joint

The hip is a ball and socket synovial joint that connects the lower limb to the pelvic girdle. The hip takes place anteriorly and laterally to the gluteal region, inferiorly to the iliac crest, and superior to the greater trochanter of the femur. The primary function of the hip joint is to provide dynamic support to the weight of the trunk and to allow the transmission of forces and loads from the axial skeleton to the lower extremities. This hip joint's ability to balance forces provides the stability needed to perform everyday activities such as standing upright, lifting weights, and raising from a chair [25]. The hip is part of the group of mobile joints, responsible of movement, called diarthrosis. Like all joints belonging to this category, it consists of three main structures [26]:

- two articular surfaces, the acetabulum, and the femoral head, both covered with cartilage;
- an articular capsule, which encloses the articular surfaces. It keeps the joint in place;
- synovial membrane, contained inside the capsule, which secretes and reabsorbs the synovial fluid. This liquid acts as a lubricant, facilitating movement between the two articular surfaces.

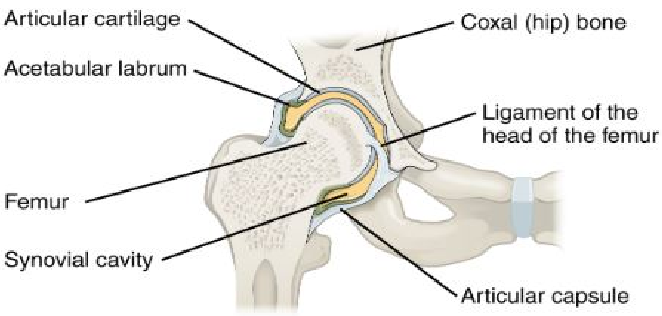

Figure 1.15: Frontal section through the right hip joint [3].

1.4.1 Articular surfaces

In a synovial joint, the articular surfaces are the two bony portions that meet to form the joint. In the hip joint, one of the two surfaces is a hemispherical protuberance, the head of the femur, which fits perfectly into the concavity of the second surface, the acetabulum. The head of the femur is covered with hyaline cartilage, except in the center, where there is a small rough depression called fovea capitis. It is the site of insertion of one end of the round ligament of the femoral head, which does not contribute to the stability of the hip, but allows the blood supply to the femoral head. The head of the femur is attached to the body of the femur thanks to the femoral neck, which varies in length depending on body size. The neck-body angle of the femur is usually $125 \pm 5°$ in normal adults, but this can vary due to pathologies show in Figure 1.16. In the conditions of coxa vara the angle is less than 120°, whereas in the coxa valga, the inclination exceeds 130°. Significant deviations of this angle from its typical range lead to having lever

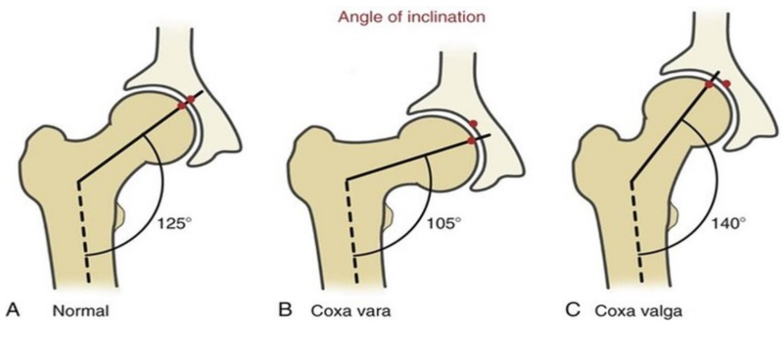

Figure 1.16: Pathologies that can lead to changes in the normal neck-body angle of femur (a). The angle decreases in the case of coxa vara (b) and increase in the case of coxa valga (c) [27].

arms, used to produce motion by the muscles, too small or too large, thus varying the ratios between the forces acting on the joint [26]. The hip joint thus becomes very unstable or is subjected to high stress and surgical correction is therefore necessary to realign the proximal femur or the acetabulum. The acetabulum is a cup-shaped depression located on the lateral inferior wall of the pelvis formed by a combination of three bones: ileum, ischium, and pubis. These innominate bones are separated from the triradiate cartilage in the skeletally immature. Between the ages of 15 and 17, the process of ossification of the cartilage begins, reaching completion usually between the ages of 20 and 25. The surface of the cavity of the acetabulum is divided into zones that play different roles and are illustrated in Figure 1.17. The horseshoe-shaped articular superior portion is called the lunate surface of the acetabulum. It is covered with hyaline cartilage and is the only part of the acetabulum that contacts the femoral head. The lower portion is the acetabular notch and guarantees the transit of the transverse acetabular ligament, gives insertion at the other end of the round ligament of the femoral head, and forms the so-called acetabular foramen, a space intended to allow the passage of the blood vessels that nourish the head of the femur. Finally, the central part of the cavity of the acetabulum is called the acetabular fossa. This area is in continuity with the acetabular notch and contains adipose tissue covered by the synovial membrane. Around the circular perimeter of the acetabulum is fixed the acetabular labrum, a fibrocartilaginous structure similar to a ring. The labrum aims to give stability to the hip, distribute the forces around the joint and promote the correct housing of the femoral head [25][26].

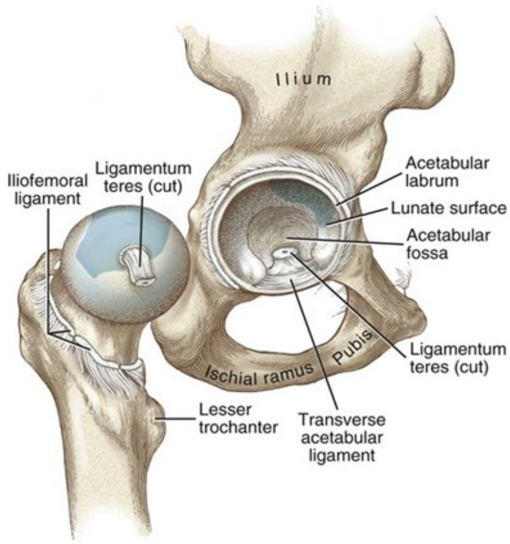

Figure 1.17: Internal components of acetabulum [27].

1.4.2 Articular capsule

The hip joint capsule is a structure of dense fibrous connective tissue that holds the femoral head and acetabulum together. In addition, it deals with the production of synovial fluid that acts as a lubricant and nourishes the joint itself. It extends from the acetabulum to almost the entire neck of the femur and has a variable thickness. The capsule is thicker anterosuperiorly where predominant stresses of weight-bearing occur and is thinner posteroinferiorly [28][29].

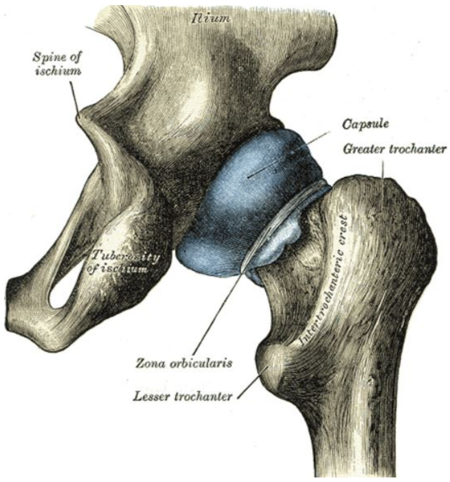

Figure 1.18: Posterior aspect of articular capsule of hip joint [28].

1.4.3 Ligaments

Three major ligaments, reported in Figure 1.19, surround the outer surface of the joint capsule, extracapsular ligaments, and have the task of strengthening the hip joint and maintaining its integrity during movements. These ligaments have a unique spiral orientation; this causes them to become tighter when the joint is extended. The iliofemoral ligament has an inverted Y-shape. It arises from the anterior inferior iliac spine and then bifurcates before inserting into the intertrochanteric line of the femur. This ligament limits extension and external rotation of the hip and assists in the maintenance of a static erect posture with minimal muscular activity. The inferior aspect of the iliofemoral ligament blends distally with the pubofemoral ligament, which has a triangular shape and covers the inferior and medial aspects of the joint capsule. This ligament tightens with hip extension and abduction. The ischiofemoral ligament originates at the level of the body of the ischium and is inserted at the base of the greater trochanter on the femur, reinforcing the capsule posteriorly. The ischiofemoral ligament also prevents hyperextension and holds the femoral head in the acetabulum [30][31].

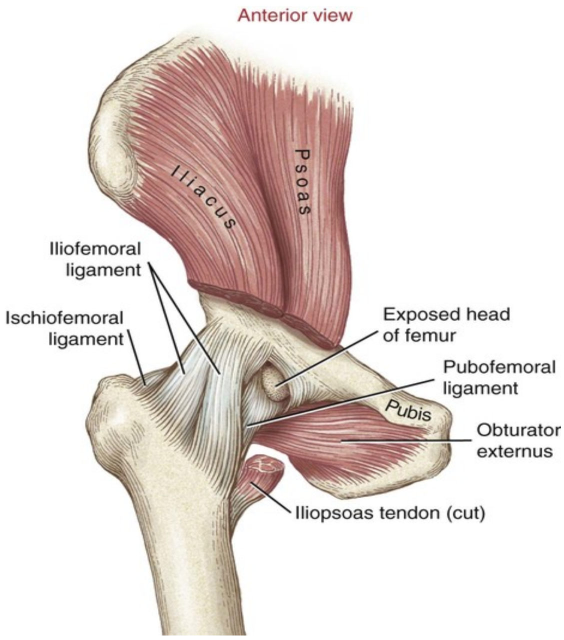

Figure 1.19: The extracapsular ligaments of the hip joint [27].

There are two further ligaments at the hip joint, that are part of the intracapsular ligaments. One is the round ligament of the femoral head or ligamentum teres which, as already mentioned, runs from the fovea capitis to the acetabular notch. A branch of the obturator artery travels through the ligamentum teres, providing the femoral head with a limited amount of its blood supply. The other one is the transverse acetabular ligament, which spans the acetabular notch, completing the "cup" of the acetabulum [27].

1.4.4 Hip muscles and movements

In addition to the primary function of weight-bear, the hip also allows the movements of the lower limb. It is a mobile joint (enarthrosis) and, being made up of almost perfectly congruent spherical articular surfaces, it permits very wide movements in all directions on all orthogonal planes. The muscles that act on the hip joint are the thigh muscles and some of the leg muscles. The thigh muscles are those that originate from the pelvis and are divided into four categories [4].

- *Gluteal muscles*: they cover the lateral surface of the ileum (gluteus maximus, medius, and minimus, tensor fasciae latae represented in Figure 1.20). The gluteus maximus muscle is the most voluminous and superficial, originating from the posterior gluteal line, iliac crest, sacrum, coccyx, associated ligaments, and lumbar fascia. The unilateral contraction of this muscle determines the extension and lateral rotation of the hip. The gluteus maximus shares an insertion with the tensor fasciae latae muscle, which originates from the iliac crest and the lateral surface of the anterosuperior iliac spine. Together, these muscles lift the iliotibial tract, a band of collagen fibers that extends along the lateral surface of the thigh and fits onto the tibia, providing important knee support. The gluteus medius and minimus muscles originate anteriorly to the gluteus maximus and are inserted at the femur's level of the greater trochanter. Both result in the abduction and medial rotation of the hip joint;

- *External rotator muscles*: they are six deep muscles positioned under the horizontal axis of the acetabulum and are inserted on the femur (piriformis, obturator externus, and internus, quadratus femoris, gemellus superior and inferior shown in Figure 1.20). All of them act by rotating the thigh laterally, and the piriformis muscle also causes abduction at the hip level;

- *Adductor muscles*: they are located below the acetabulum (adductor magnus, adductor brevis, adductor longus, pectineus e gracilis illustrated in Figure 1.21). They all originate from the pubis and are inserted, except for the gracilis (which is inserted on the level of the tibia), on the posterior surface of the femur, linea aspera. All adductors, except the adductor magnus, originate anteriorly and inferiorly to the hip joint, determining adduction, flexion, and medial rotation. The adductor magnus, depending on the stimulated portion, can cause adduction and flexion, adduction and extension, and medial and lateral rotation;

- *Iliopsoas muscle*: composed of two large muscles that are located in the medial portion of the pelvis (psoas major e iliacus in Figure 1.21). The psoas major muscle originates from the transverse processes of the last thoracic vertebrae and lumbar vertebrae and is inserted on the lesser trochanter of the femur, but, before reaching the insertion site, its tendon joins with that of the iliacus muscle, at the level of the iliac fossa. These two muscles, powerful hip flexors, pass deeply relative to the inguinal ligament and are usually referred to as the iliopsoas muscle.

  The leg muscles that are involved in hip movements are the rectus femoris and sartorius (on the front of the thigh), biceps femoris, semimembranosus e semitendinosus (hamstring muscles, on the back of the thigh). They will be explained in detail below, in the paragraph 1.5.5.

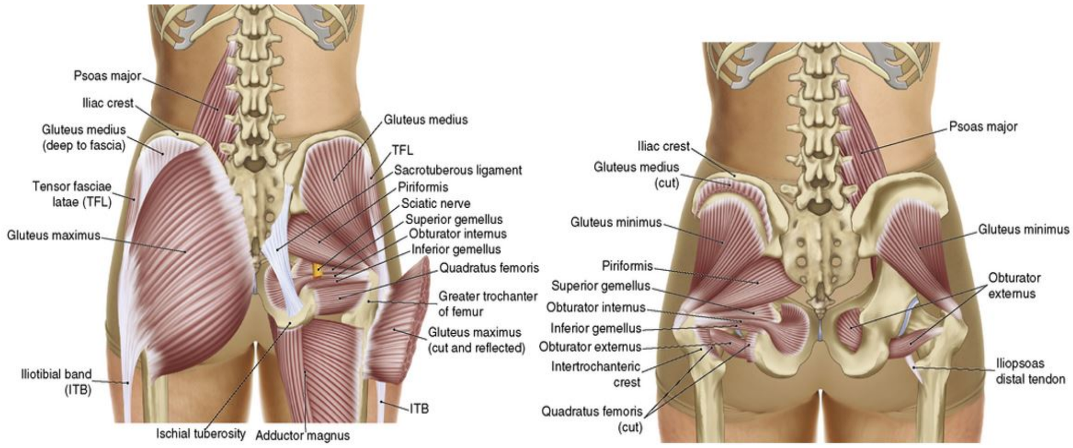

Figure 1.20: Gluteal muscles and External rotator muscles [32].

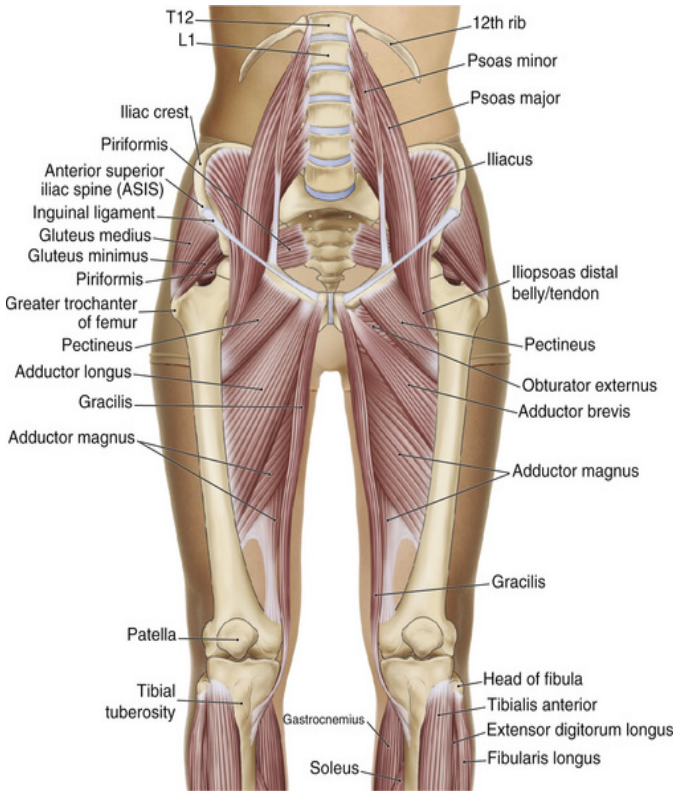

Figure 1.21: Adductor muscles and Iliopsoas muscle [32].

The hip joint is a multi-axial synovial joint that, thanks to its geometry, allows movements in three degrees of freedom: flexion, extension, adduction, abduction, external rotation, internal rotation, and circumduction.

Hip flexion occurs thanks to the action of the iliopsoas, sartorius, rectus femoris, tensor fasciae latae, adductor longus, and pectineus, which bring the front surface of the thigh closer to the trunk. Complete hip flexion occurs only when the knee is flexed and the flexion angle is about 120° during an active flexion, while it reaches 140° during passive flexion. Instead, when the knee is extended, the flexion is limited to 90° by the hamstring muscles.

The hip extension moves the lower limb posteriorly to the frontal plane. This movement also depends on whether or not the knee flexes. When the knee is extended, the extension is up to 20°, while if the knee is flexed, the movement is less wide because it is limited by the tension of the rectus femoris and by the force already expressed on the knee by the hamstrings. The massive extension reaches a maximum amplitude of 30°. The primary hip extensors include the gluteus maximus, posterior head of the adductor magnus, and hamstrings. When hip extension is performed from standing, tension on the capsular ligaments and in the flexor muscles increases. This can cause limitations of the final range of motion achieved. When hip extension is performed from standing, tension on the capsular ligaments and in the flexor muscles increases. This can cause limitations of the final range of motion achieved. The angle of hip extension can be increased by contracting together hip extensor and abdominal muscles.

Abduction and adduction are the movements that bring the lower limb closer to and away from the plane of symmetry of the body. Both occur in the frontal plane and have a free range of motion of about 45°. The angles of abduction and adduction are both greater while keeping the knee flexed. Abduction and adduction are limited by adductor and abductor muscles, respectively. In addition, the adduction is also limited by the contralateral limb. The primary hip abductor muscles include all the gluteus medius and minimus fibers and the tensor fasciae latae, while the group of adductor muscles is required for adduction.

Internal and external rotation of the hip joint occurs in the horizontal plane. It is preferable to perform these movements with the patient lying prone or sitting on the edge of a table with knees flexed at 90°. From this position, when the leg is brought in, an internal rotation angle between 30° and 40° is measured. When, on the other hand, the limb is taken out, the external rotation reaches a maximum amplitude of 60°. In addition to the external rotator muscles group, another powerful external rotator is the gluteus maximus, which generates a rotation moment that remains approximately constant between 60° and 90° of hip flexion and increases when the flexion angle is below 60°. However, the range of motion of the external rotation remains limited mainly due to the tension on the ligaments of the anterior part (iliofemoral and pubofemoral ligaments) and on the intra-rotator muscles.

The internal rotation is less free and it is mainly attended by the gluteus medius and minimus, tensor fasciae latae, and pectineus. The limiting factors, in this case, are the tension on the ligaments of the posterior part (ischiofemoral ligament) and the external rotator muscles [33].

1.5 Anatomy of the knee joint

The knee is the largest joint in the human body. It joins the lower leg and thigh bilaterally and it is mobile enough to allow the movements of the lower leg, indispensable for carrying out normal activities such as walking, running, and sitting. But it is also strong enough to be able to support the weight of the body [34]. The knee, like the hip joint, is also part of the group of diarthrosis and more specifically it is called the hinge synovial joint, which means that it allows mainly two types of movements, flexion, and extension, but also a small degree of medial and lateral rotation. Since it is a diarthrosis, the knee joint is also formed by the three typical elements, already described for the hip: the articular surfaces, which in this case are the tibiofemoral joint and the patellofemoral joint, the articular capsule, and the synovial membrane. Menisci, ligaments, and tendons are components of a fibrocartilaginous nature that are very important both for the stability and for the preservation of joint surfaces. While synovial bursae and fat pads are elements with anti-trauma functions.

1.5.1 Articular surfaces

The articular surfaces of the knee consist of the joints that compose it, the tibiofemoral joint and patellofemoral joint, shown in Figure 1.22. Both are covered with hyaline cartilage and are enclosed within the joint capsule. The tibiofemoral joint is an articulation between the lateral and medial condyles of the distal end of the femur and the tibial plateaus. The lateral and medial femoral condyles are two asymmetrical projections located at the distal end of the femur, which have a smooth convex surface, and are separated posteriorly by a deep groove known as the intercondylar fossa (or intercondylar notch). The two concave superior surfaces of the condyle located at the proximal end of the tibia and separated by the intercondylar eminence (bony protuberance) are called tibial plateaus. The tibiofemoral joint allows for knee flexion and extension and a very small degree of rotation and lateral movement. The femur patellar surface and the posterior surface of patella together form the patellofemoral joint. The patellar surface of the femur is a groove on the anterior side of the distal femur, which extends posteriorly into the intercondylar fossa. The patella is a triangular-shaped bone, with a curved proximal base and a pointed distal apex and plays an important role in the extension of the knee. It transmits the extensor force across the knee at a greater distance from the axis of rotation. In this way the moment arm increases, allowing the knee to extend further (up to 30%) without requiring additional strength to the quadriceps. The patellofemoral joint improves the quadriceps' ability to extend the knee and offers some bone protection to the anterior tibiofemoral joint [35].

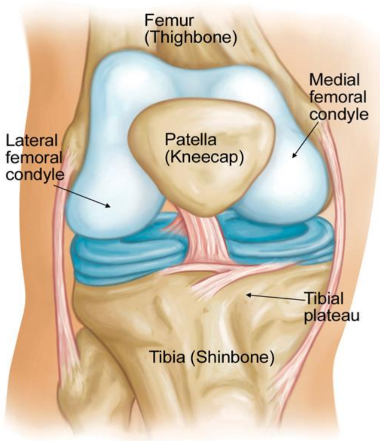

Figure 1.22: Components of articular surfaces of knee joint [36].

1.5.2 Articular capsule and synovial bursae

The articular capsule includes the tibiofemoral and patellofemoral joints and it is attached to the cartilaginous margins of the femoral and tibial condyles both anteriorly and posteriorly. The capsule is formed by an outer fibrous layer, that holds the bone components together, and an inner synovial membrane that lubricates the articular surfaces, reducing friction. Inside the joint capsule, there are also the so-called synovial bursae, synovial fluid-filled sacs that reduce the wear of the articular components within which they are located. In Figure 1.23 are shown the notable bursae of the knee joint, which include the [35]:

- *Suprapatellar bursa*, an extension of the synovial cavity of the knee, located superior to the patella between the femur and the tendon of the quadriceps muscle;
- *Prepatellar bursa* lies between the apex of the patella and the skin;
- *Infrapatellar bursa*, split into deep and superficial, and both under the patella. The deep bursa is located between the patellar ligament and the tibia. The superficial one lies between the patella ligament and the skin;

- *Semimembranosus bursa*, located in the posterior part of the knee joint, between the semimembranosus muscle and the medial head of the gastrocnemius.

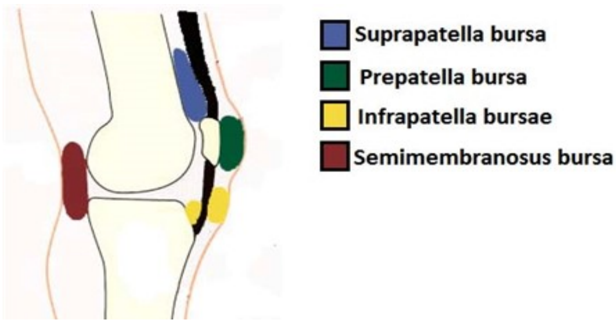

Figure 1.23: Synovial bursae of the knee joint. The tendon of the quadriceps muscle, above the patella is represented in black. The ligament of the patella or patellar tendon, under the patella, is represented in black [37].

1.5.3 Menisci

The medial and lateral menisci, illustrated in Figure 1.24, are two separate fibrocartilage crescent-shaped plates, that lie respectively on the surface of the medial and lateral tibial plateau. They function as stabilizers of the joint working together with the ligaments, and shock absorbers by increasing the surface area to further dissipate the forces and friction reducers between the articular surfaces [35].

1.5.4 Ligaments

There are four main ligaments of the knee, two intracapsular, included inside the articular capsule, and two extracapsular, outside the capsule. The common function to all knee ligaments is to provide stability to the tibiofemoral joint and prevent its dislocations. They fall into the group of intracapsular ligaments, the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL). These two ligaments connect the femur to the tibia and cross each other obliquely, hence the term “cruciate”. The ACL arises from the anterior part of the intercondylar eminence of the tibial plate, behind the attachment of the medial meniscus, and it ascends posteriorly to attach to the lateral femoral condyle [38][39]. This ligament prevents excessive knee extension and the dislocation of the femoral condyle during flexion. The PCL extends from the posterior aspect of the tibia to the medial femoral condyle and functions to prevent forward displacement of the femur on the tibia [25]. In the group of extracapsular ligaments, there are the medial collateral ligament (MCL) and the lateral collateral ligament (LCL). The MCL is a wide and flat ligament that lies on the inner side of the tibiofemoral joint. Proximally, it attaches to the medial epicondyle of the femur and, distally, to the medial condyle of the tibia. Its task is to strengthen the inner side of the knee avoiding that excessive forces (valgus stress) on the opposite side can cause a misalignment of the femur with respect to the tibia or vice versa. Furthermore, MCL contributes to conferring stability to the medial meniscus due to its connection with this structure. Thinner and rounded than the MCL, the LCL is localized on the outer part of the knee and spans from the lateral epicondyle of the femur to depression on the lateral surface of the head of the fibula. This ligament (LCL) has the same function as its medial correspondent (MCL), with the difference that the side of the knee on which it works is the external one [25][26]. Besides the ligaments already mentioned, the knee has another major ligament: the patellar ligament or patellar tendon. It is a strong, thick fibrous band that is a distal continuation of the quadriceps femoris tendon and joins the lower portion of the patella to a characteristic prominence of the tibia, located just below the condyles, the tibial tuberosity. From a functional point of view, the patellar tendon plays an important role in stabilizing the patella and supporting the femoral quadriceps during the knee extension.

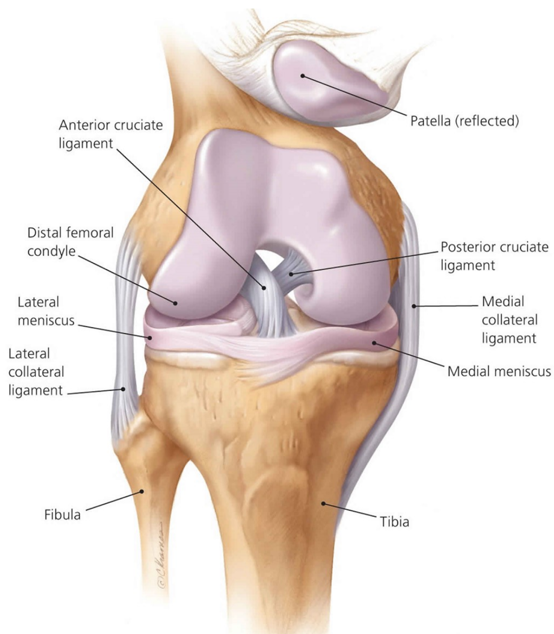

Figure 1.24: Menisci and major ligaments of the knee joint [40].

1.5.5 Knee muscles and movements

The knee joint is a hinged joint which means that almost only flexion and extension movements on the sagittal plane are allowed. Limited medial and lateral rotations (internal and external) are also possible, but only when the knee is flexed at 90° (if the knee is not flexed, the rotation occurs at the hip joint). In order to make such movements, the leg muscles listed below are recruited [4].

- *Quadriceps femoris*: it is formed by four muscles and causes the extension of the knee (Figure). It consists of three vastus muscles (vastus lateralis, vastus medialis and vastus intermedius), which originate along the body of the femur and envelop the rectus femoris muscle and all four fit onto the tibial tuberosity. The rectus femoris originates from the anterior inferior iliac spine and, for this reason, along with causing the extension of the knee, participates in the flexion of the hip. The quadriceps femoris muscles conjoin to form the patellar tendon/ligament, which crosses the knee anteriorly and inserts on the patella and tibial tuberosity;

- *Sartorius*: it is the only flexor muscle of the knee that originates on the anterior surface of the superior anterior iliac spine and is inserted on the medial margin of the tibia. In addition to flexing the knee, it is also involved in the flexion and lateral rotation of the hip joint (Figure);

- *Hamstring muscles*: this group consists of three muscles, biceps femoris, semitendinosus, and semimembranosus, which originate in the area of the ischial tuberosity and are inserted at the level of the tibia and the head of the fibula. The contraction of such muscles causes knee flexion, and they also participate in hip extension (Figure);

- *Popliteus muscle*: it originates near the lateral femoral condyle and is inserted on the posterior surface of the body of the tibia. Once the knee flexion begins, when this muscle contracts, a medial rotation of the tibia occurs and the join is blocked.

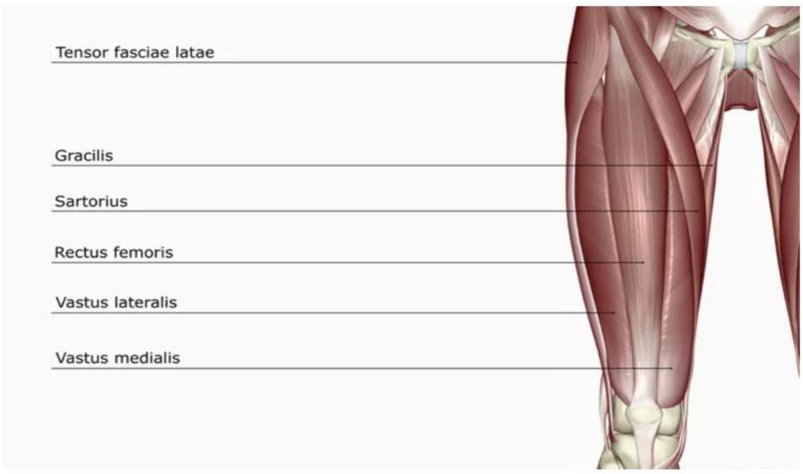

Figure 1.25: Quadriceps femoris and sartorius [41].

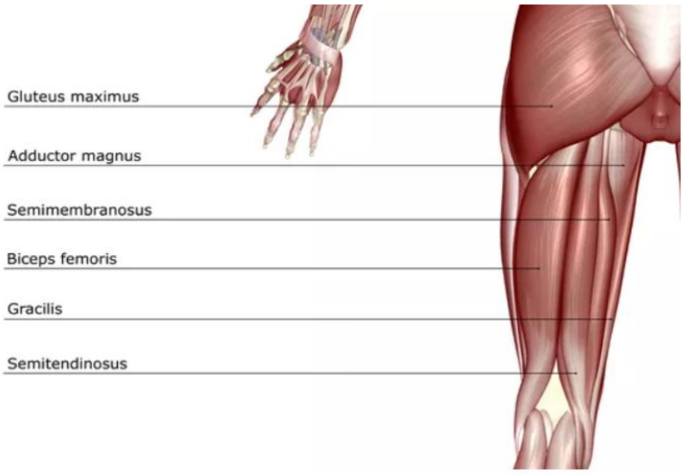

Figure 1.26: Hamstring muscles group [42].

The flexion is produced by the hamstring muscles, the sartorius, and the popliteus with the help of gracilis and gastrocnemius. Since the hamstrings are both hip extensors and knee flexors, when knee flexion is performed with extended hip, the action of these muscles is divided between the two movements and therefore the total knee flexion range reaches a maximum of 120°. Instead, when the hip is flexed, the action of the hamstrings is conveyed only on the knee flexion, which then reaches an angle of up to 140 °, with an active motion, and 160 °, with a passive motion [43].

The extension instead takes place by the action of the quadriceps muscles. In the reference position, the lower limb is already in extension, but it is possible to extend, passively, the knee by a further 5-10 ° (hyperextension). The active extension rarely and slightly exceeds the reference position and depends on the position of the hip. The prior extension of the hip facilitates the extension of the knee. The position in which you can have a maximum extension is when the knee starts from a flexion ranging from about 45 ° to 70 °. This particular extension is used frequently during daily activities, such as climbing stairs or getting up from a chair [44].

1.6 General structure of protocols

The hip and knee joints are among the most mobile joints in the body and play an important role in supporting body weight. However, they can also be subject to greater loads, considering that they are the most exploited joints to perform daily activities such as walking, climbing stairs, sitting, or getting up from a chair. Due to their continuous recruitment, the cartilage of these joints slowly deteriorates, leading, in severe cases, to arthrosis. The latter is a degenerative inflammatory pathology that affects the cartilaginous surface of the joints and manifests itself mainly in people over 50 years of age. However, it is not uncommon to find the problem even in younger subjects who have suffered trauma or surgery (e.g. meniscus removal) [45]. In the specific case of the knee joint, the degenerative process does not only affect the cartilage but also the meniscus and ligaments. Consumption of cartilage tissue leads to rubbing of the bone components of the joint, causing pain in the subject. As the pathology progresses, the joint function becomes limited, preventing the performance of the usual activities [46]. An advanced form of arthrosis often leads the affected subject to undergo an arthroplasty (total hip arthroplasty (THA) and total knee arthroplasty (TKA)), which must then be followed by an adequate rehabilitation protocol to return to a condition of complete autonomy and independence [47]. This muscle strengthening and movement recovery program, as illustrated by the different studies reported in section 2.2, is more effective and faster when normal physiotherapy is combined with FES. Hence the need to outline specific rehabilitation protocols for hip and knee joints achievable through electrostimulation.

Both hip and knee rehabilitation protocols have been outlined by referring to existing protocols used by nursing homes and hospitals and which involve the presence of the physiotherapist to help the patient retrain movements and motor functions [48][49][50][51]. In the protocols proposed in this thesis project, the help of the physiotherapist is replaced by the FES, and therefore the patient performs every single movement helped only by electrostimulation. For each protocol, 11 movements, listed below, have been defined. In particular, the movements are organized in three phases according to their difficulty and the effort required of the patient to perform them.

The first phase is characterized by all the exercises that can be carried out starting from the day after the surgery. At this stage of recovery, the muscles of the damaged joint are particularly weak, and so the execution of the movements takes place by making the patient lie on the bed. In this way, the lower limbs do not have to support the entire body weight and are not subjected to too high stress [52][53]. Normally, the physiotherapist would help the patient perform the exercises, manipulating the patient's limb and imprinting a certain force on the muscles. The FES in this case replaces the action of the physiotherapist, accompanying the patient's limb in carrying out the movement [54].

Once the patient is able to perform each exercise by himself (without FES) without perceiving pain and with a range of motion similar to that of the unaffected limb, then he can move on to the second stage. The latter provides only exercises to be performed in an upright position, in which then all the weight of the body weighs on the lower limbs, and the strengthening of the muscles intensifies [52][53]. At this stage, there is no help from the physiotherapist in guiding the patient's leg during the movement, but the use of FES is useful to help the patient perform a wider and more fluid movement.

The achievement of the third phase takes place according to the same rules defined for the transition to the second phase. It is the most advanced recovery phase, in which more complex movements are performed that require the involvement of several muscles, trained during the previous phases. It is therefore important to have first trained and adequately strengthened the various muscles individually [55][56]. At this stage, the FES replaces supports (e.g. walking frame) that the patient would normally use to complete the exercise.

The hip rehabilitation protocol includes:

- 1° phase: ankle flexion-extension; knee flexion supine; knee extension supine; hip abduction supine; hip abduction lateral; hip extension prone;
- 2° phase: hip flexion standing; hip extension standing; hip abduction standing;
- 3° phase: standing heel raises; step-up (get on a step).

The knee rehabilitation protocol includes:

- 1° phase: ankle flexion-extension; knee flexion supine; knee flexion prone; knee extension supine; hip abduction lateral; straight leg raises;
- 2° phase: knee extension seated; hip extension standing; hip abduction standing;
- 3° phase: standing heel raises; step-up.

The performance of each exercise is reported below:

- ankle flexion-extension: the patient lies supine on the bed and moves the ankle up (flexion) and down (extension) consecutively;
- knee flexion supine: the patient lies supine on the bed and bends the knee bringing it towards the chest by sliding the heel on the bed. Then it returns to the initial position;
- knee flexion prone: the patient lies supine on the bed and bends the knee bringing it towards the chest by sliding the heel on the bed. After holding this position 3-4 s, the limb is returned to the initial position;
- knee extension supine: the patient is lying supine on the bed with support (e.g: foam roller) placed below the knee and lifts the heel until it reaches the complete extension of the leg. After holding this position 3-4 s, the limb is returned to the initial position;

- hip abduction supine: the patient is lying supine on the bed and performs the hip abduction by opening the leg outwards, keeping the position for 3-4 s, and then returning to the starting position;
- hip extension prone: the patient is lying prone on the bed and raises the leg, keeping it extended, up to about 10-20 cm. After holding the position for 3-4 s, the limb is returned to the resting position;
- hip abduction lateral: the patient is lying on the side of the unaffected limb, raises the operated leg, keeping it extended, about 20-30 cm, and holds it in this position for 3-4 s. Then it returns to the initial position;
- straight leg raise: the patient is lying supine and raises the extended leg up to about 20-30 cm. After holding this position 3-4 s, the limb is returned to the initial position;
- knee extension seated: the patient is seated on the edge of the bed with the legs free to move and lifts the heel until it reaches the complete extension of the leg. After holding this position 3-4 s, the limb is returned to the initial position;
- hip flexion standing: the patient is standing and raises the leg toward the chest, keeping the knee flexed at about 90°. After holding this position for 3-4 s, the limb is returned to the resting position;
- hip extension standing: the patient is standing and raises the leg, keeping it extended, up to about 10-20 cm. After holding the position for 3-4 s, the limb is returned to the resting position;
- hip abduction standing: the patient is standing and performs the hip abduction by opening the leg outwards, keeping the position for 3-4 s, and then returning to the starting position;
- heel raise: the patient is standing and lifts his heels, shifting the weight to the toes. This position is maintained for 3-4 s, leaning on support to maintain balance if necessary, and then returns to the initial position by resting the heels on the ground. In normal physiotherapy, the first stage of rising up is performed by the patient with the help of support, on which part of the weight is discharged. The FES replaces the support, allowing the patient to move the weight on the toes on their own. The support is only used in the phase of maintaining the position in order not to lose balance;
- step-up: the patient is standing in front of a step with a maximum height of 15 cm. He puts the foot of the operated limb on the step and gets on it. Then

he returns to the initial position by descending from the step with the other leg. In normal physiotherapy, it is usually expected that the patient uses a walker or a support to help the affected leg to bring the other limb to the step. The FES replaces these support instruments, allowing the patient to perform the movement on his own. Support is used only to avoid losing the balance once the subject gets on the step.

Chapter 2

State of Art

2.1 Modulation of FES parameters

As explained in 1.3, FES consists of the application of rectangular electrical impulses. These pulses are regulated in current, rather than voltage, to prevent the contraction of the muscles from depending on a change in impedance, such as the detachment of one of the electrodes. One of the biggest limitations of FES treatments is the rapid onset of muscle fatigue during repeated contractions. The onset of muscle fatigue during electrical stimulation is strongly related to stimulation parameters such as current intensity, frequency and pulse width. The intensity of the contraction depends on the number of motor units recruited. A greater amplitude (current) or duration (pulse width) of the pulse allows to recruit more motor units and thus obtain a stronger contraction. As well as a higher frequency of stimulation leads to a faster contraction of the motor units and a greater resulting contraction force. There are several studies that, even today, try to understand how to effectively modulate the stimulation parameters to make the stimulation less uncomfortable and to reduce the onset of fatigue in the muscle fibers involved.

- [57] examines the effect of frequency modulation, combined with current intensity modulation, on quadriceps performance during repeated dynamic contractions. Seven subjects participated in two sets of experiments, in which NMES is used in knee extension exercises. In the first experiment, using the same amplitude modulation, the subjects completed four protocols:

  1. constant frequency at 20 Hz (Protocol 1);
  2. constant frequency at 40 Hz (Protocol 2);
  3. frequency decrease from 40 Hz to 20 Hz (Protocol 3);
  4. frequency increase from 20 Hz to 40 Hz (Protocol 4).

Quadriceps performance assessments were made by comparing successful run times (SRTs) between protocols. From the results obtained from this first set of experiments, it was concluded that, compared to constant frequency protocols, amplitude and frequency modulation strategies prolonged the ability of the limb to perform the desired function (knee extension). Therefore, the simultaneous modulation of amplitude and frequency allows for obtaining better performance than only amplitude modulation. In addition, analysis of the two variable frequency protocols suggests that the highest average SRT occurred in the 20-30 Hz range, indicating that this is the preferred frequency range to delay the onset of fatigue. The second experiment was used to determine a possible change in the response of muscles to low- and high-frequency stimulation. The angle of excursion of the leg subjected to constant frequency stimulation of 10 Hz and 100 Hz before and after the variable frequency protocols was compared. The results underline that there is no statistical difference in muscle response to fatigue between the two protocols.

- [58] is a study in which four different protocols are compared with each other in terms of muscle performance and time until fatigue arises. Five healthy subjects had to perform, using FES, the elbow flexion with a weight of 1 kg attached to an armband to make sure that the arm and forearm were completely extended at the end of the stimulation. The four stimulation protocols, applied in four different sessions, are described below:

  1. Standard Protocol (STD) with 20 Hz frequency, 200 µs pulse width and minimum current to evoke movement;
  2. Long Pulse Duration (LPD) with 20 Hz frequency, 500 µs pulse width and minimum current to induce the movement;
  3. High Frequency (HF) with 100 Hz frequency, 200 µs pulse width, and minimum current to perform the movement;
  4. High Amplitude (HA) used 20 Hz frequency, 200 µs pulse width, and current 25% above the minimum to evoke movement.

  Tests show that higher current amplitudes, and therefore stronger contractions, seem to affect the outcome of the tests slightly more positively and result in greater discomfort for the patient. As for the pulse width, the increase in this parameter expresses a substantial increase in the flexion angle without affecting the time until the onset of fatigue. However, the most important outcome is the discovery that the frequency of pulses is the parameter that mainly affects the onset of fatigue. Higher frequencies, such as those in the HF protocol, lead to a faster decline in the bending angle.

- In the study [59] a random modulation of stimulation parameters was tested to see if it could reduce muscle fatigue due to FES. The latter was applied, via surface electrodes, to the quadriceps and anterior tibial muscles bilaterally of seven paraplegic subjects to reproduce knee extension and ankle dorsiflexion. Four different modes of FES have been applied in random order:

  1. constant stimulation;
  2. randomized frequency, average frequency = 40 Hz;
  3. randomized current amplitude;
  4. randomized pulse width, average pulse width = 250 µs.

  The parameters were randomly modulated every 100 ms in a range of ±15% using a uniform probability distribution. No significant difference was found between the measurements of the fatigue onset time of the four stimulation modes. Therefore, the method of stochastic variation of parameters does not seem to have any effect on muscle fatigue.

- [60] compares the isometric performance and paraplegic muscle fatigue using frequency and pulse width modulation. During the tests, the eight paraplegic subjects were sitting in a semi- upright sitting position and, thanks to the FES, had to lift the leg to the maximum of its extension. The first protocol involved a stimulation with a fixed current (40 mA) and pulse width (250 µs) and a frequency variation from 10 Hz to 50 Hz with steps of 10 Hz. The second protocol instead provided a fixed current (40 mA) and frequency (30 Hz) and a pulse width variation from 200 µs to 400 µs with steps of 50 µs. The results of this study show that frequency modulation leads to different levels of fatigue: higher frequencies fatigue the muscles faster. This correspondence is not noticeable with the pulse width modulation, which, however, strongly affects the maximum muscle force produced. Finally, the study indicates the stimulation parameters considered most appropriate, for paraplegic subjects, to produce a muscle torque sufficient to perform the functional task: a frequency range between 20-35 Hz and a pulse width that can be varied up to a maximum value of 400 µs.

- The study [61] studies the motor response to the variation of pulse parameters during ankle dorsiflexion. Seven adults with spastic diplegia participated in the study. They were placed supine with legs and feet outside the end of the table. In particular, the effects of frequency and pulse width variation, using two different constant currents (*Ia* and *Ib*), were studied. The *Ia* current was defined as the current needed to produce a 10° dorsiflexion with a pulse width of 300 µs and a frequency of 30 Hz. Whereas *Ib* was the current necessary to produce the same angle of motion but with pulse width at 500 µs and frequency at 60 Hz. For each of the current values, five values of pulse width (100, 200, 300, 400, and 500 µs) and five of frequency (10, 20, 30, 40, and 60 Hz) were combined to obtain a total of twenty-five stimulation tests to be submitted, randomly, to the subjects. The results obtained demonstrated the possibility of using pulse width modulation to regulate the motor response induced by FES. Specifically, ankle dorsiflexion can be achieved using a constant current, specific to each subject, a fixed frequency at 20 or 30 Hz and a pulse width variation between 100 and 400 µs.

2.2 Use of FES in rehabilitation

The first applications of FES were aimed at completely and permanently replacing a lost neuromuscular function. Subsequently, the use of FES was widespread also in the field of rehabilitation to retrain the patient in specific motor activities. After any damage to a specific limb or part of the body, it is crucial to follow a precise rehabilitation treatment to promote a total recovery of the limb in question. Over the years, it has been shown that simply moving the damaged extremities is not sufficient, but the active participation of the patient is necessary to obtain a movement as similar as possible to the physiological one. During rehabilitation sessions, FES is used to help the patient to perform movements. After the short-term therapeutic FES treatment, the patient is expected to be able to voluntarily perform, without the FES, the trained movements [62]. Several studies have shown that FES, combined with standard physiotherapy, leads to an increase in muscle mass and force contraction, and a faster recovery of the voluntary execution of the movement, reducing the normal rehabilitation time [63][64][65].

For example, [63] examines the long-term effect of FES cycling, applied to patients with SCI, on improving muscle performance and increasing muscle size and force generation. A total of forty-five participants were included in this study. Of these, twenty received a standard rehabilitation (control group) and twenty-five carried out an FES-assisted cycling modality (FES group). Patients of the FES group underwent sessions lasting 45-60 min with a frequency of 3 days a week for three months. The control group sessions, on the other hand, involved only passive stretching without active therapeutic physical activity. Motor, sensory, and combined motor-sensory scores (CMSS) were calculated to evaluate the effect of FES on neurological function. 80% of patients undergoing FES showed improvement in CMSS, while this occurred for only 40% of patients in the control group. In addition, an increase of 36% in the volume of thigh muscles (quadriceps and hamstrings) and a reduction of 44% in the amount of total thigh fat, were found in the FES group. Finally, greater muscle strength of the quadriceps (35%) and hamstrings (30%) in the FES group was assessed.

The study of Sabut, S. K., et al. [64] has shown instead the use of FES has a potential application as a therapeutic intervention to improve muscle strength for the correction of the drop of the foot. Ten hemiplegic subjects, able to stand alone and independently walk for 10 m, received FES therapy combined with a normal rehabilitation program 5 days a week for 12 weeks. During each session (1 hour), after performing the standard physiotherapy exercises, electrodes on the peroneal nerve and the motor point of the tibialis anterior (TA) were applied to obtain the dorsiflexion of the ankle and the eversion of the foot. A force-sensitive heel switch, placed under the heel, triggered stimulation with the oscillation phase of the gait cycle. To evaluate improvements in FES therapy, mean absolute value (MAV), root mean square (RMS), and median frequency (MF) extracts from sEMG signal of TA, pre and post-treatment, were compared. There was a significant increase in MAV and RMS at the end of FES therapy and also an improvement in MF and sEMG signal amplitude. These results indicate that FES-assisted walking, combined with standard rehabilitation, leads to muscle strengthening.
In [65] the efficacy of an FES treatment, coupled with conventional occupational therapy (COT), compared to a COT-only pathway to improve grasping is examined. Twenty-four patients with a subacute traumatic incomplete SCI were divided into the intervention group, the one that undergoes FES and COT, and the control group, which receives COT alone. Each patient received 1 hour of COT plus 1 hour of FES or 2 hours of COT for 5 days a week over 8 weeks (a total of 80 hours of rehabilitation). During each session, the subjects had to perform the movements of grasping heavy and large objects, such as water bottles, but also smaller objects, such as keys. The comparison between the two types of rehabilitation was made by calculating the scores of the Functional Independence Measure (FIM) and Spinal Cord Independence Measure (SCIM) and then performing the Toronto Rehabilitation Institute Hand Function Test (TRI-HFT). In particular, the FIM is an international standard of disability measurement in which, each of the 18 daily activities listed, is associated with a score from 1 (complete dependence on others) to 7 (complete self-sufficiency). The SCIM is a disability scale that has been specifically developed to evaluate the degree of disability in patients with traumatic and nontraumatic SCI. This scale evaluates functions in three specific areas, whose scores expand from 0 (total dependence on others) to 20 or 40 (self-sufficiency): self-care (0-20), respiration and sphincter management (range 0-40) and mobility (range 0-40). Finally, TRI-HFT evaluates gross motor function of unilateral grasp and provides a scale of scores from 0 (subject unable to reach for the object) to 7 (subject able to reach, grasp, lift, and manipulate the object appropriately). At the end of treatment, the FES group reported higher scores on all rating scales than the COT group. In particular, the average score of the FIM scale of the FES group compared to that of the COT group was 20.1 against 10. From these results, it was possible to conclude that FES treatment actually leads to a reduction in

disability and an improvement in the voluntary intake of various objects.

Not only patients with SCI or who have survived a stroke receive rehabilitation with electrostimulation. FES is also used to prevent muscle atrophy in those subjects who, due to accidents or damage, have remained motionless for a fairly long period of time. This case also includes those people who have undergone total hip arthroplasty (THA) or total knee arthroplasty (TKA) surgeries, which present muscle weakness that can lead to a decrease in the performance of daily activities such as walking and climbing stairs. Therefore, rehabilitation is crucial for muscle strengthening. Some studies [66][67][68], that prove the benefits of FES applied to these patients, are reported and decribed below.
The aim of the study [66] was to assess the effect of low-frequency electrical stimulations, associated with normal physiotherapy, on the functionality of elderly patients after a THA surgery. Twenty-nine patients were recruited and divided into two groups: the intervention group, which receives low-frequency electric muscle stimulation and conventional physiotherapy, and the control group, trained only with conventional physiotherapy. All patients of the intervention group underwent 1 hour sessions, in addition to the 2 hours of normal physiotherapy, 5 days a week for 5 weeks. Electrostimulation was applied bilaterally to the quadriceps and calf muscles. At the end of the treatment, the scores of the FIM scale, the maximum isometric resistance of the knee extensors, a 6 min walking test and a 200 m fast walking test were evaluated. Although the results from the two walking tests were similar between the two groups, a greater improvement was found in the FIM scores of the FES group compared to the control group. Furthermore, low-frequency electric stimulation led to a greater strengthening of the knee extensors than the other group, demonstrating how effective is this technology in the rehabilitation of THA patients.
Another study related to the rehabilitation of patients with THA [67], compares the results obtained from one hundred forty-three patients, who underwent an FES treatment of 15 sessions of 30 min, with those of fifty-four patients who received traditional therapy. Also in this case it has been shown that FES leads to greater muscle strengthening and also allows for reducing motor deficit.
The work of Martin, T. P., et al. [68] instead reports the analysis of the influence of FES on specific properties of the muscular fibers of the vastus lateralis on patients who are recovering from TKA. Before surgery, the vastus lateralis of these patients was characterized by a predominance of type I fibers, which were much larger than the type II fibers present in the same area. Patients who received passive physiotherapy combined with FES, at the end of treatment, reported an increase in type II fibers, while those who received only passive movement showed, in addition to an increase in type II fibers, also an increase in muscle atrophy on all types of fibers. From these data, it was concluded that FES is a valid tool to reduce the muscle atrophy that can occur with TKA.

2.3 System Description

2.3.1 Overview of the System

The system used to implement the hip and knee rehabilitation protocols, described in chapter 1, is an event-driven closed-loop system for the real-time FES control [69]. It is a system that provides modulation of the intensity of the impulse driven only by the information of the Average Threshold Crossing (ATC) applied to the sEMG signal.
ATC is a bio-inspired technique that is based on comparing the sEMG signal with a certain threshold: whenever the signal exceeds that threshold, an event is generated, producing the quasi-digital Threshold Crossing (TC) signal, as the Figure 2.1 shows. The TC signal is characterized by a digital form that carries the temporal analog information (frequency with which the muscle contracts). The ATC parameter is obtained by computing the ratio between the TC events detected in a certain time window and the length of the window itself.

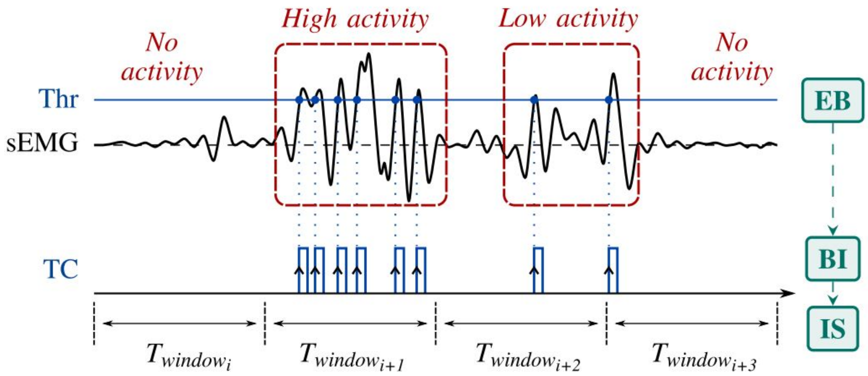

Figure 2.1: Average Threshold Crossing (ATC) technique: every time the sEMG signal crosses the threshold, a Threshold Crossing (TC) event is raised. By averaging the number of TC events in a certain time window, the ATC parameter is obtained [70].

There are several advantages related to the ATC technique. It allows to limit the extraction of all the information carried by the signal only to the events that generate a muscular action (i.e.: TC events), thus leading to a reduction in power consumption, system size and circuit complexity [71]. Moreover, this method, in which only a series of digital pulses are extracted from the starting signal, allows to better mimic the communication through neural spikes that normally takes place in a biological system. Finally, it has been shown, not only that there is a strong correlation between the ATC parameter and the muscle contraction force, but also that different muscle strength levels are represented by different ATC values [72]. The realization of the ATC technique takes place completely at the hardware level, using a voltage comparator.

The ATC-FES system, driven by TC events, works as explained below. The sEMG signal is acquired, processed and translated into an ATC value on the wearable acquisition board. ATC parameters are continuously transmitted via Bluetooth Low Energy (BLE) to a computer. At this point, a custom multi-platform software developed in Python programming language (henceforth referred simply as Python) [73], processes the received data and computes the corresponding values of current intensity (frequency and pulse width are fixed), which are then sent, via a serial communication, to the functional electrical stimulator and through surface electrodes, applied on the muscle to be trained, the muscle contraction is generated. Each time new ATC parameters are transmitted, the computer updates the data and their current intensity. The management of the connection between the acquisition boards and the computer and the control of the transmission of the stimulation parameters during the training takes place thanks to a Graphical User Interface (GUI). The system can be used in two modalities. In the therapist-patient modality, the therapist is the subject who wears the acquisition boards and from whom the sEMG signals are taken, while the patient is the second subject, who receives the corresponding stimulation parameters. In the self-administered modality, on the other hand, only one subject (patient) is involved: the data of muscle activities are acquired from the patient's healthy limb and the stimulation parameters are transmitted to the affected limb.
In the following paragraphs, the main units of the system will be analyzed in detail.

2.3.2 Acquisition and Processing of sEMG

As previously specified, the extraction of information relating only to TC events was carried out directly in hardware, but before the ATC parameter is computed, the sEMG signal is processed. Figure 2.2 shows the stages of the Analog Front End (AFE) which allow the processing of the input signal and the extraction of TC events [70].
Starting to the left of Figure 2.2, the signal is acquired through the bipolar configuration, described in subsection 1.2.3, which involves two acquisition electrodes (V_{E1} and V_{E2}), to detect the potential of muscle fibers, and a reference electrode (V_{ER}). Input and output protection are circuits that have the aim of limiting excessive voltages, with values around or higher than the supply voltage (1.8 V), which could damage the circuitry components. In addition, the output protection also has the functionality of limiting the current that is supplied to the subject in order to comply with the regulations for electrical safety (the current is kept below 10 mA so as not to cause damage or discomfort to the subject). The two impedance decoupling blocks, one for each acquisition channel, on the other hand, are voltage

followers, which provide the right impedance to the amplifier inputs to ensure that they do not remain coupled to the electrode-skin impedance.

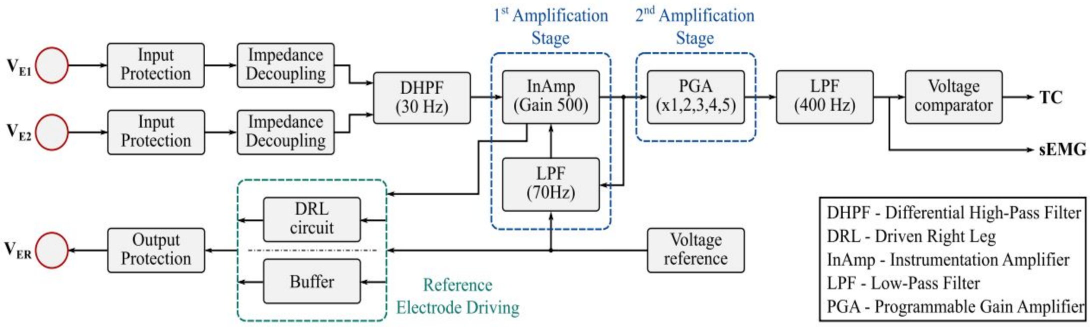

Figure 2.2: Block diagram of the Analog Front End (AFE): a high-pass filter (30 Hz) is first applied to the sEMG signal, then it is amplified in two stages (variable gain 500-2500) and finally a low-pass filter (400 Hz) is applied. Only at this point, TC events are extracted from the sEMG signal [70].

The first signal processing stage is a first-order Differential High-Pass Filter (DHPF). The sEMG signal can be affected by many noises and any chemical or physical variation at the electrode-skin interface can degrade the signal. It has been demonstrated that the input signal is often corrupted by a noise in the frequency range of 20-30 Hz.The latter is due to motion artifacts, which occur due to the relative deviation of the electrode on the skin and the electrode with its own connector. For this reason, a high-pass filter with a cutoff frequency of 30 Hz has been inserted, a value that allows the filtering of the artifacts without attenuating too much the energy content of the sEMG signal. The raw sEMG signal has a very low amplitude (values up to 10 mV), so, in order to carry out processing with digital electronics, it is necessary to amplify it. For this reason, an Instrumentation Amplifier (InAmp) has been inserted, which performs the difference of the signals at the input, amplifies it by a value equal to the gain Ad and then centers it to the value of the reference potential (V_{OUT} = Ad(V_{IN_1} $-V_{IN_2}$) + V_{REF}). The gain was set to 500 V/V because it has been considered sufficient to obtain good acquisitions from even very large muscles. Negative feedback has also been added to the InAmp reference. It is a second order Butterworth multiple-feedback low pass filter (cutoff frequency of 70 Hz to remove both low-frequency noises, introduced by the signal conditioning instrumentation itself, and residual power-line interference) that sense the amplifier output, removes unwanted noises, inverts it and returns it to the InAmp reference. Thanks to this approach, the signal obtained from this amplification stage is more stable and uniform. However, if the signal is acquired from smaller or deeper muscles, separated by a fairly thick layer of fat, a single amplification stage is not enough. Another one was added: a Programmable Gain Amplifier (PGA) able to keep the signal unchanged or amplify it by $\times 2$, $\times 3$ and

$\times 5$. Moving forward in the chain, there is a Sallen-Key Low-Pass Filter (LPF), which limits the sEMG signal bandwidth to 400 Hz. The LPF acts not only as a frequency content limiter (most relevant energy content at 50 Hz–150 Hz) but also as an anti-aliasing filter considering that, at this stage, the sEMG signal is placed in input to an Analog-to-Digital Converter (ADC). The filter, therefore, allows a minimum sampling frequency of 800 Hz. Finally, the voltage comparator was inserted to extract the TC events from the input signal. Considering that the threshold value, which defines the eventual occurrence of a TC event, can change each time depending on the environmental conditions or the muscle whose activity is recorded, a Digital-to-Analog Converter (DAC) was added to the voltage comparator [70]. The communication of the output of the AFE to the PC takes place via BLE (version 4.2), as this approach has been demonstrated to save 14% of the maximum power consumption when transmitting data from four AFE channels (four AFE outputs) [74]. To ensure the data transmission, as shown in Figure 2.3, the AFEs are coupled with a MicroController Unit (MCU) equipped with a Bluetooth antenna, which in turn communicates with the CC2540 USB dongle [75] from Texas Instrument™, which is used as an external transceiver and is serially connected to the PC. ATC data are available every 130 ms for each channel, a time that takes into account the low data rate and allows for discrimination of muscle activity [71].

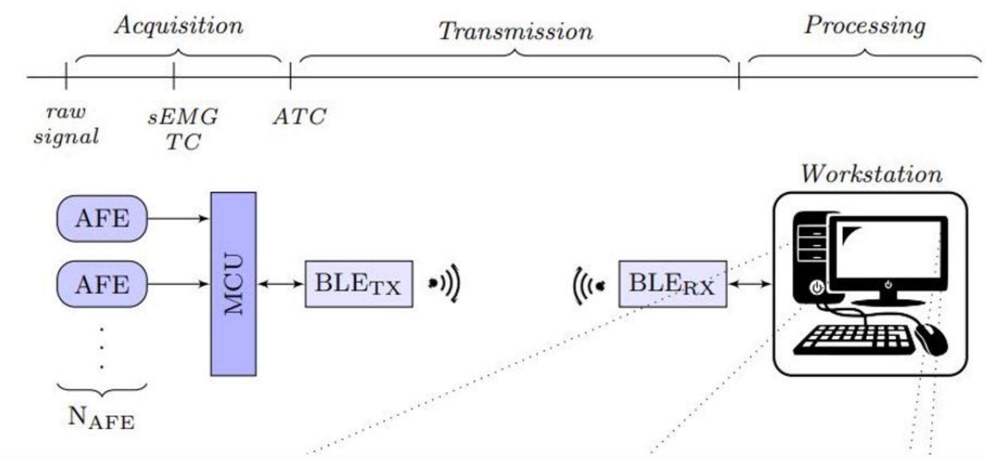

Figure 2.3: Four AFE outputs are connected to the MCU. The latter transmits ATC data via BLE using a Bluetooth antenna that communicates with the CC2540 USB dongle, also equipped with an antenna, which receives and communicates the data to the PC (Workstation) [71].

To choose the most appropriate MCU, several factors were taken into account: considering that ATC data must be sent continuously, a battery that lasts over time is needed, so good power efficiency is required to improve battery lifetime. In addition, even though the use of ATC parameters results in a very low computational weight, non-volatile and volatile memories, Flash and RAM respectively, should be large enough to store and handle large amounts of data in a short time in case of BLE connection failure. However, the MCU should have also a small size to impact as little as possible the final area of the device. The Apollo3 Blue ultra-low power microcontroller [76] from Ambiq Micro (henceforth referred simply as Apollux and used to indicate acquisition boards) has been indicated as the MCU that best fulfills the requirements listed above. In addition to the MCU, other components have also been added (illustrated in Figure 2.4): a DAC to set the threshold to be used in the ATC (explained in subsection 2.3.5), two I^2C connectors to to create, if necessary, a bridge between several boards, a Universal Asynchronous Receiver Transmitter (UART)-to-USB converter, which serves to replace BLE connectivity during diagnostics [70].

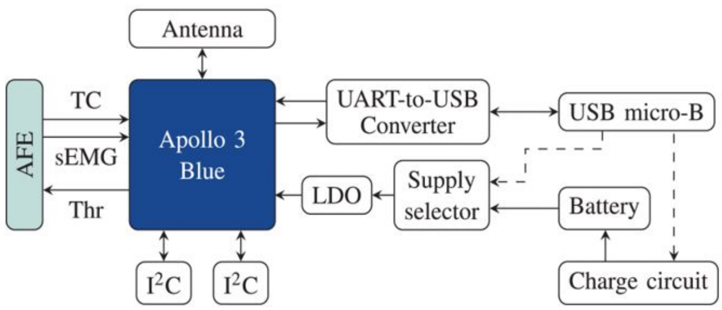

Figure 2.4: Schematic representation of the components connected to Apollo 3 Blue [70].

Figure 2.5 shows the Printed Circuit Board (PCB) arranged in the various electronic components described above: AFE, MCU and elements connected to the Artemis module (Apollo 3 Blue) [77].

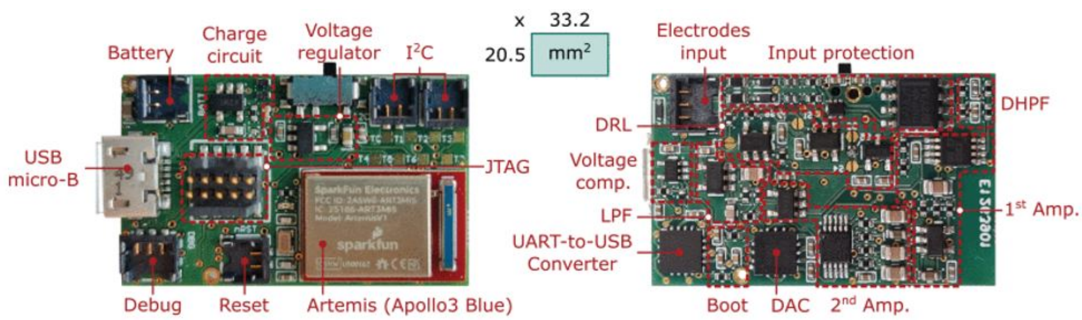

Figure 2.5: Printed Circuit Board (PCB) arrangement [70].

In Figure 2.6 are depicted the 3D-printed case, in which the PCB is contained, and the positions of the rechargeable battery and the three electrodes.

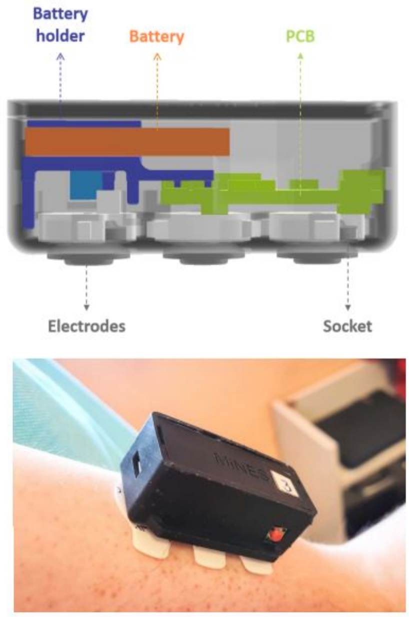

Figure 2.6: Design of the position of the battery and the electrodes inside the PCB and application of the wearable system, contained in the 3D-printed case, on the skin through sticky electrodes [70].

2.3.3 Functional Electrical Stimulator

The Functional Electrical Stimulator used is the Rehastim2 from HASOMED® [78]. It is a type IIa medical device certified according to the international standards EN 60601-1 and EN 60601-2-10 for medical devices and systems. Thanks to its rechargeable battery, it is a portable electrical stimulator that generates pulses on up to eight stimulation channels at the same time. The battery operation is particularly useful when the stimulator is used together with bio-signal acquisition devices, as it reduces the possibility of network disturbances. Looking at the Figure 2.7, it is possible to see that the stimulator is equipped with two USB ports to update the software via USB flash drive, a connection for the power supply and a USB output to interface the stimulator with an external computer, which allows the control of the stimulation parameters to provide to the Rehastim2 using the ScienceMode2 serial communication protocol. Moreover, the stimulator has a red connector for the emergency stop button, which, when pressed, interrupts the stimulation immediately, and two connectors (yellow and green) used for the two corresponding electrode cables, which connect the stimulator with the surface electrodes. Each electrode cable is divided into four stimulation channels and each channel is connected to two electrodes.

Figure 2.7: RehaStim2: connection elements [78].

The current pulses generated by the stimulator have a rectangular biphasic waveform with a balanced charge. The three basic stimulation parameters (current, pulse width and frequency) delineate the amplitude, width and period of these pulses. Looking at the Figure 2.8, the amplitude of the pulse is defined by the intensity of the current, the duration by the pulse width and between the positive and negative phases there is a fixed time delay (inter-phase interval) of 150 µs.

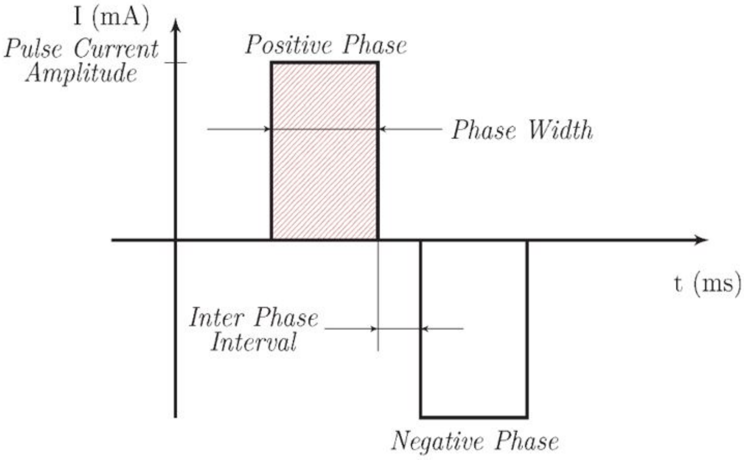

Figure 2.8: Waveform of the impulses generated by the Rehastim2: biphasic rectangular shape [79].

2.3.4 Software organization

The software of the ATC-FES system was developed [79] and updated [80] using Python, as already mentioned in subsection 2.3.1. The use of the Object Oriented Programming (OOP) paradigm to realize the architecture of the platform, has led to the creation of an application that can be updated by different people at different times (extensible), which is able to maintain the software performance regardless of the number and type of devices that can be used (scalable and modular) and to create a real-time control through multithreaded functionality. The architecture of the software consists of three layers. The bottom layer is responsible for Bluetooth communication between the acquisition boards and the computer and serial communication between the computer and the electrical stimulator, managed, respectively, by the *BLE* and *FES* objects in Figure 2.9. The middle layer, *System* in Figure 2.9, is used to connect the two objects of the lower layer: through the *BLE* object, the ATC parameters are stored, which are processed and translated into the corresponding current intensities thanks to the *System* object and finally sent to the electrical stimulator through the *FES* object. Finally, the last layer consists of the *GUI*, which allows the user to manage the whole system, as explained in more detail in the subsection below.

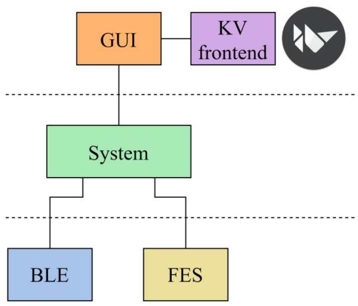

Figure 2.9: The architecture of the software is composed by three layers. Bottom layer: *BLE* and *FES* objects that manage communication between computers and external devices. Middle layer: the *System* module that processes the information collected by *BLE* and transfers it to the stimulator through *FES*. Top layer: the *GUI* class that allows any user to interact with the software [80].

2.3.5 Graphical User Interface (GUI)

The GUI is used to make the management of the connection between the acquisition boards and the PC easier and more intuitive and to supervise the stimulation phase. The interface was implemented using the Kivy Python Library [81]. When the application starts, the first screen that appears is the *Login Screen*, where the session supervisor must enter the username and password. Subsequently, the *Medical Screen* appears, in Figure 2.10, where it is possible to add the general information of the patient (name, age, gender, pathology) and, selecting the movement that will be performed in the session, the indications on the number and placement of the acquisition and stimulation electrodes are provided.

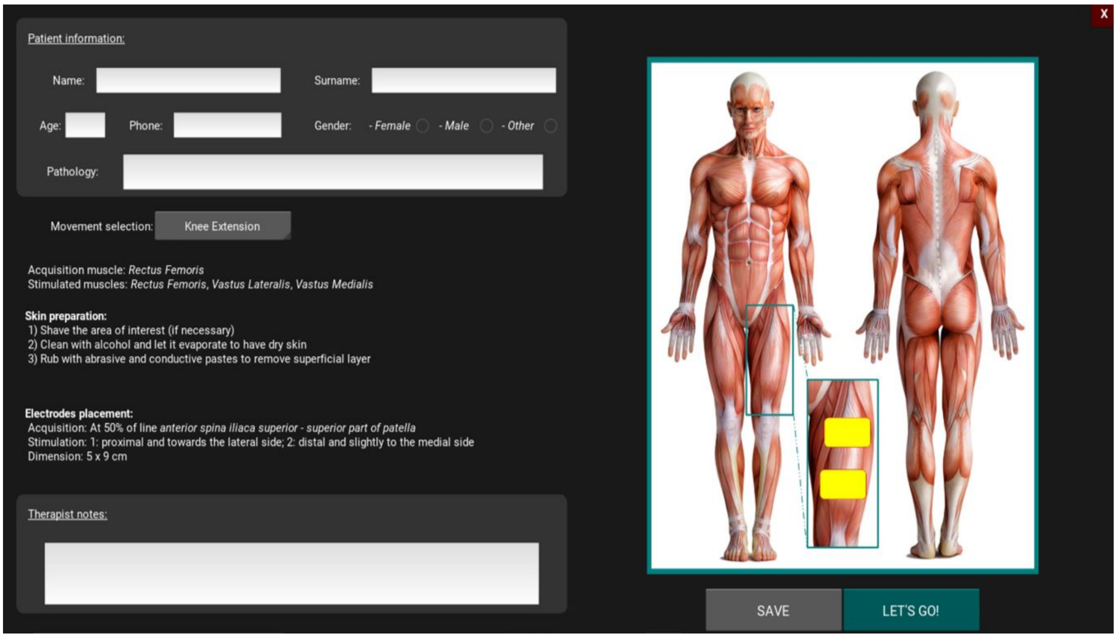

Figure 2.10: Medical Screen. For example, by selecting the "Knee Extension" movement, on the left, in addition to a summary of how to prepare the patient, it is reported that the muscles to be stimulated are Rectus Femoris, Vastus Lateralis and Vastus Medialis. Then, the exact placements of the acquisition and stimulation electrodes are indicated. On the right, there is a visual representation to give further support to the user in the positioning of the stimulation electrodes.

Pressing on *Let's Go!*, the *Stimulation Screen*, shown in Figure 2.11, is accessed. The only button enabled is *Connect*, which allows the connection of the Apollux acquisition boards with the CC2540 dongle. Pressing it, a PopUp, in which all the available Bluetooth devices are listed, appears. Once the desired devices are selected and the connection is established, the rest of the commands are enabled. At this point, the stimulation channels that will be used are selected by ticking the corresponding boxes (*Rehastim2 channels*). Each selected Apollux has to be connected to a single stimulation channel so that the data transmitted by a single acquisition board is translated into a specific stimulation pattern for a single stimulation channel, which will contract a single muscle with a pair of surface electrodes. Figure 2.12 shows an example: Apollux 2 is connected to stimulation channel 6 by selecting "Ap2" from the selector that appears at the second *Apollux channels* and Apollux 3 is connected to stimulation channel 7. It is also possible to set the PGA gain for each Apollux (x2, x3, x5).

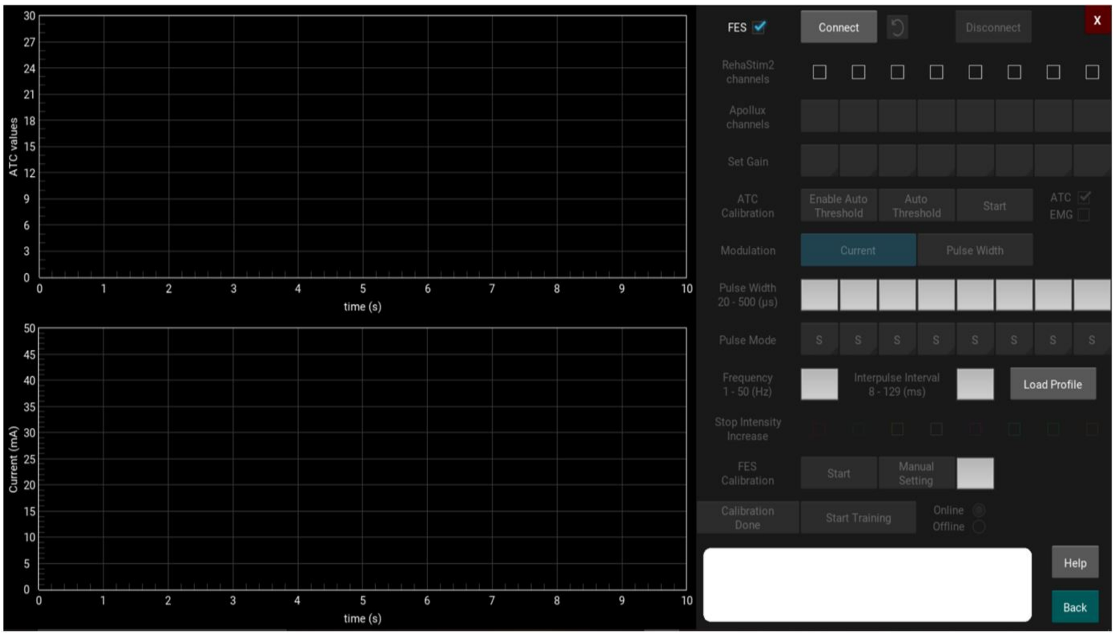

Figure 2.11: Stimulation Screen. *Connect* is the only button enabled.

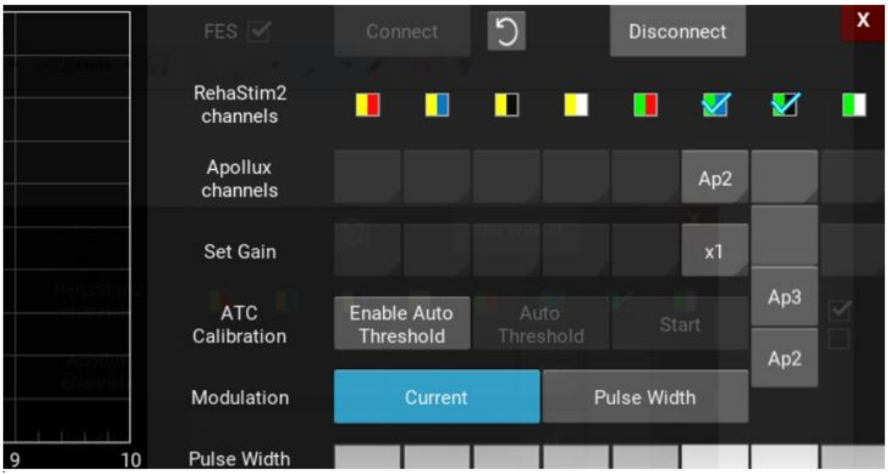

Figure 2.12: Example of association: Apollux 2 (Ap2) - stimulation channel 6 (Ch6); Apollux 3 (Ap3) - stimulation channel 7 (Ch7). At Ch6 (green and blue), Ap2 is selected from the selector and at Ch7 (green and black), Ap3 is selected.

Since ATC values and stimulation currents may be different between different subjects, even with the same muscles involved, it is necessary to assess the specific muscle activations specific for each therapist and establish what are the appropriate stimulation parameters for the subject participating in the session. Firstly, a two-step therapist calibration is performed. In the first phase, the threshold that determines the generation of a TC event is set: while the therapist remains in a resting position, the starting value of the threshold is slowly decreased until the baseline of the sEMG signal is found and at this point, the last value found is increased by 30 mV. Following this approach, the threshold is set just above the signal baseline, so that TC events are generated even when muscle activity is minimal. Instead, the second phase involves the definition of the maximum ATC produced by the therapist: the therapist repeats the movement, which will then be performed by the patient, at least four times. On each channel, the median of the maximum ATC values obtained from each repetition is computed and the maximum final ATC value is returned (Figure 2.13). After the first calibration, pulse width, frequency and interpulse interval values are manually entered and the patient calibration (*FES Calibration*) begins by pressing *Start*. As soon as the activation profile is generated by the therapist, at the end of its calibration, this is translated into a stimulation profile (Figure 2.13). Gradually, each current value of the profile, including the maximum current, is increased by 2 mA until a maximum current, which is tolerated by the patient and allows him to perform the movement, is reached. At the end of the patient calibration phase, the maximum current values for each stimulation channel are recorded and reported on the screen (Figure 2.14).

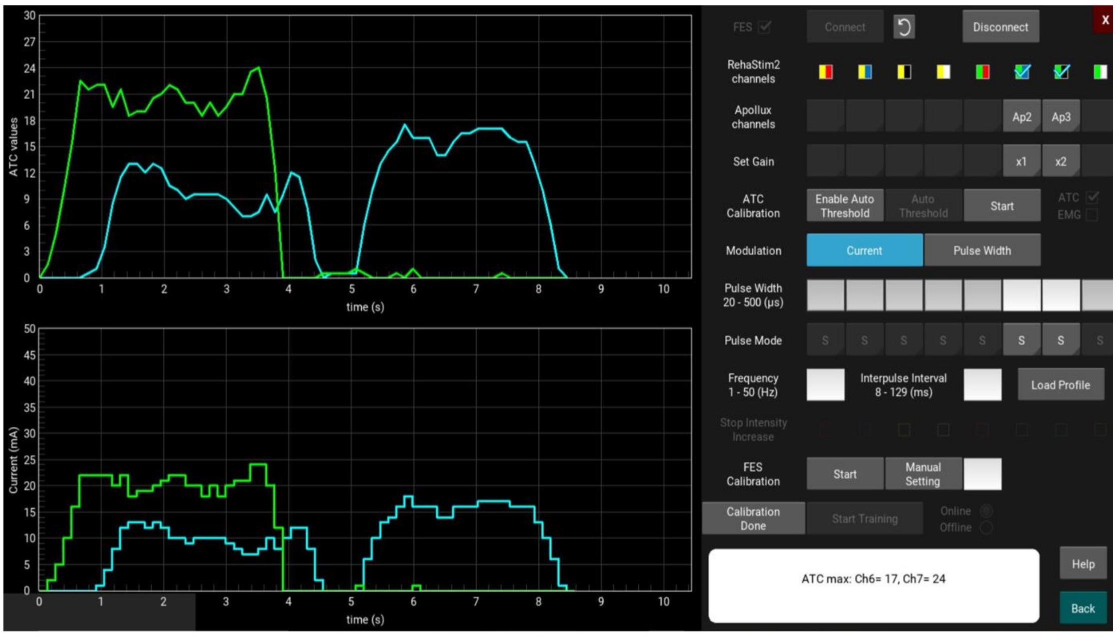

Figure 2.13: At the end of the therapist calibration, for each channel, the median profile and the maximum ATC value are returned. Top left: the median profiles returned by the therapist calibration. Bottom right: the maximum ATC values for each channel. Bottom left: therapist activation profiles translated into stimulation profiles.

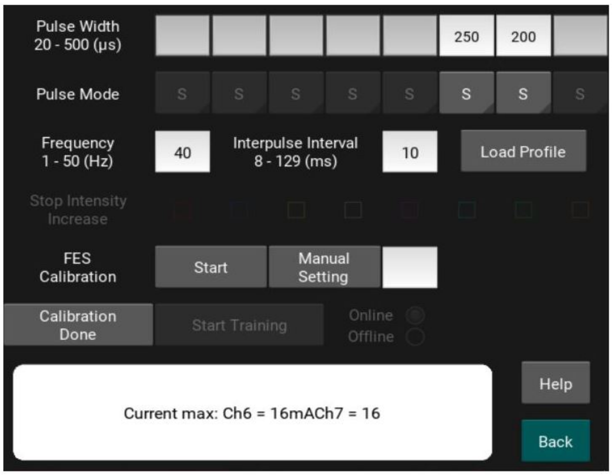

Figure 2.14: At the end of the patient calibration, the maximum current values tolerated by the subject are reported for each channel.

Once the calibration of the patient has also been completed, it is possible to move on to the training phase, explained in the section below.

2.3.6 Current Intensity Modulation

The real-time control of the FES is carried out by modulating the current intensity. In general, what happens during the training is that the therapist performs a movement, the acquisition boards record his muscle activity, ATC data are sent to the PC and are translated into corresponding values of current intensity. These current intensity values are updated each time a new ATC data is recorded. As explained in the previous section, at the end of the calibrations of the therapist and the patient, for each stimulation channel, the maximum ATC value that can be generated and the maximum current that can be tolerated (or that involves the execution of the movement) are returned, respectively. These values are used to create a Look Up Table (LUT), shown in Figure 2.15, where the maximum ATC value of each channel corresponds to the maximum current value. All other current values (from I_{max} to 0 mA) and ATC (from ATC_{max} to 0) are matched together, decreasing at each step the current of 2 mA and the ATC values of 1. However, if the maximum ATC generated by the therapist is much lower than the current value (e.g.: $ATC_{max} = 6$, $I_{max} = 40$ mA), it is not possible to scale every current value of 2 mA. Hence, the available steps (between 40 mA and 0 mA) are calculated and the decrement factor is computed accordingly. In this way, every possible ATC values, which represent the indices of the LUT columns, are related to a precise current value. During the training, each new ATC value is placed in an array, specific to each stimulation channel, where the three previous ATC values are stored. The medians between these values are computed and the results (purple cells in Figure 2.15) are used to locate the new current intensity values to be provided to the patient.

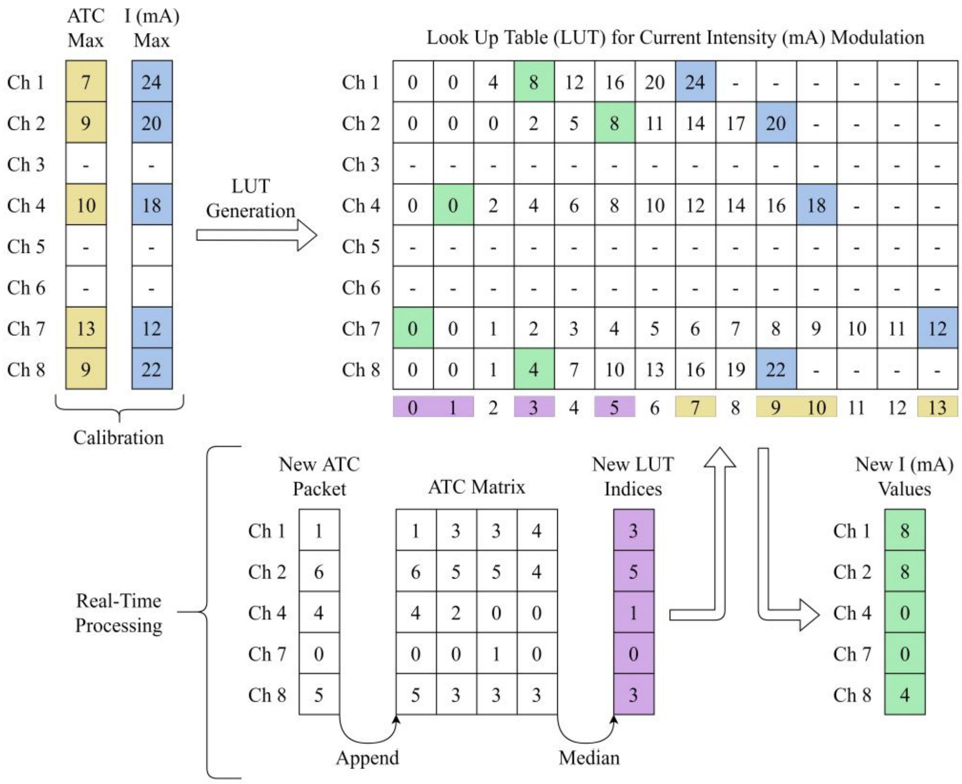

Figure 2.15: Look Up Table (LUT) built at the end of calibration. During training, each new ATC is used as an index to locate the new current parameter [80].

Following the specifications of the Rehastim2, the current can be modulated between 0-130 mA. The other two stimulation parameters are fixed and manually entered into the GUI. The pulse width can be chosen in a range of 20-500 µs with a step of 10 µs. While the frequency can be selected in a span of 10-50 Hz with steps of 5 Hz [78].

Chapter 3

Software Expansion

The purpose of this thesis is to outline a rehabilitation protocol for patients with hip prostheses and one for patients with knee prostheses both achievable through FES. In order to better define the movements of each protocol and to improve the execution of each movement through FES, changes have been introduced to the system software, which has been described in 2.3.

3.1 Versatility of ATC-FES channels connection

As explained in section 2.3.5, after connecting the Bluetooth devices to the computer and selecting the stimulation channels to be used, each acquisition board (i.e.: Apollux) must be coupled with a stimulation channel. Initially, the system allowed to associate a single Apollux with a single stimulation channel so that the ATC data, transmitted by that Bluetooth device, was translated into stimulation parameters to be sent to only that selected stimulation channel. This coupling between the acquisition board and the stimulation channel takes place at the software level using two arrays: idxBLE_from_idxFES and idxFES_from_idxBLE.

- idxBLE_from_idxFES contains the indices that identify the different Apollux devices and has a length equal to the number of total stimulation channels. Then each channel, which identifies a column of the array, is associated with an Apollux.

- idxFES_from_idxBLE includes the indices that identify the different stimulation channels and has a length equal to the number of Bluetooth devices connected to the computer. So each Apollux, which identifies a column of the array, is associated with a stimulation channel.

Before coupling, the arrays are empty (Figure 3.1a). As the device to be paired is chosen for each FES channel, the arrays are filled. Figure 3.1b shows an example of how the two arrays appear at the end of this process: the Apollux2 device is identified by index 0, Apollux3 by index 1, and the 8 available stimulation channels are indicated by the numbers 0 to 7. Apollux 2 is associated with FES channel 2 and Apollux 3 with channel 3, therefore:

- in idxBLE_from_idxFES, in the second column (column 1 - channel 2) "0" is saved, the index associated with Apollux 2, and in the third column (column 2 - channel 3) "1" is stored, Apollux 3 identifier;

- in idxFES_from_idxBLE, in the first column (column 0 - Apollux 2), "1", index of the second stimulation channel, is inserted, and in the second column (column 2 - Apollux 3) "2", number of the third stimulation channel.

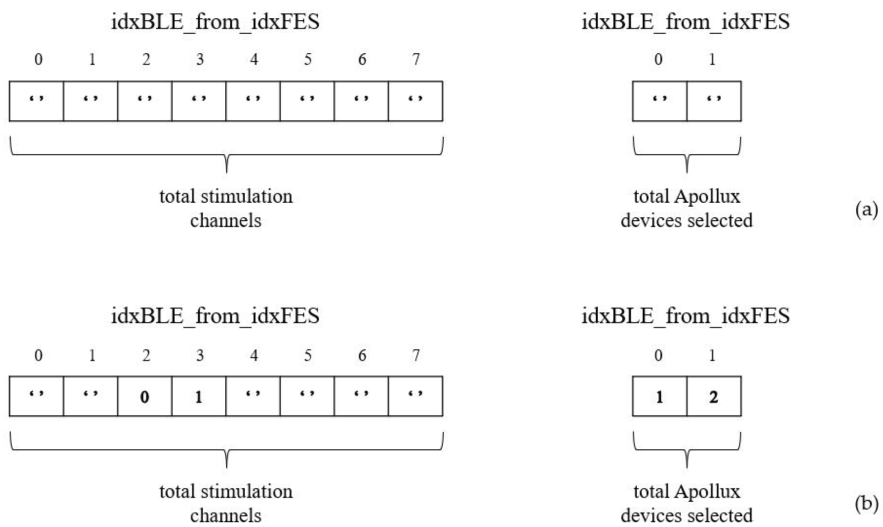

Figure 3.1: Arrays used to store and associate the indices of Bluetooth devices (Apollux) with the indices of stimulation channels. At first, both arrays are empty (a). Example of the array at the end of the coupling phase 1-1 Apollux sensors – stimulation channels (b): Apollux 2 (0) – channel 2 (1); Apollux 3 (1) – channel 3 (2). In idxBLE_from_idxFES: 0 (Apollux 2) inserted in column 1 (channel 2) and 1 (Apollux 3) inserted in column 2 (channel 3). In idxFES_from_idxBLE: 1 (channel 2) inserted in column 0 (Apollux 2) and 2 (channel 3) inserted in column 1 (Apollux 3).

In this way, however, it is not possible to connect several devices to the same FES channel or vice versa. Making the association as versatile as possible is important for two reasons:

1. Many movements, included in the protocols described in section 1.6, involve keeping the limb in a certain position for a few seconds (4-5 s), and therefore the stimulation profile must be stable enough to provide constant stimulation for a certain period of time. However, using the signal from a single muscle, it is often difficult to obtain a stabilized ATC parameter at the maximum value, leading the patient to not perform the movement correctly. Therefore it is very useful to combine signals from several muscles (connection of N Apollux sensors with 1 stimulation channel: connection N-1) to obtain a single more constant activation profile;

2. Performing a movement often involves the collaboration of different muscles (muscle synergy). This translates, in the FES application, into stimulation of several muscles at the same time to obtain a specific movement, as in the case of hip extension, which is mainly realized by the simultaneous activation of the semitendinosus and the gluteus maximus. If the activation profiles of the muscles involved are similar, it is more practical to use the same signal (one sensor) to stimulate multiple muscles (connection of 1 Apollux sensor with N stimulation channels: connection 1-N) because it is useless and expensive to use multiple acquisition boards if they all detect the signal from the same muscle fascicles. Moreover, the use of multiple boards could affect the freedom of movement of the subject and therefore, in this case, compromise the effectiveness of the system.

To achieve this type of association, the two arrays idxBLE_from_idxFES and idxFES_from_idxBLE have been transformed into the homonymous matrices. Now, idxBLE_from_idxFES is characterized, not only by a number of columns equal to the number of total stimulation channels but also by a number of rows equal to the number of Bluetooth devices connected to the computer, while in idxFES_from_idxBLE, besides the columns representing the connected Apollux, as many rows have been added as there are the stimulation channels of the system. In this way, in idxBLE_from_idxFES each Apollux index is stored in the cell determined by the row corresponding to the Apollux to be saved and by the column representing the stimulation channel to be associated with that device. The same process also occurs with idxFES_from_idxBLE, but in this case rows and columns are inverted and the indices of the stimulation channels are saved. Figure 3.2 shows an example of filling the two matrices, while Figure 3.3 shows how to select from the GUI the Apollux to associate with the various channels.

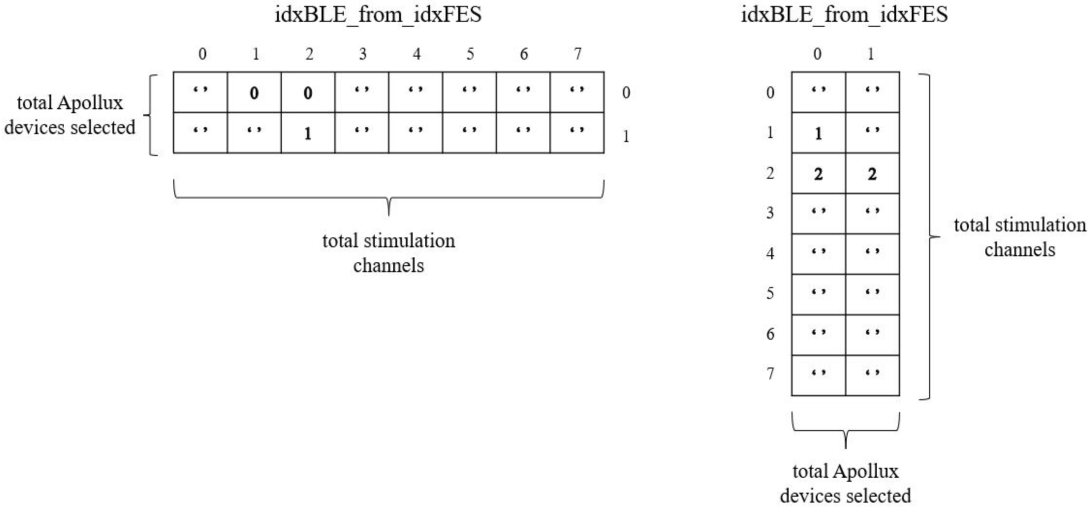

Figure 3.2: Example of associating a sensor (Ap) with two stimulation channels (1-2) and two sensors with the same stimulation channel (2-1). Ap2 - Ch2 and Ch3; Ap3 - Ch3. Then in idxBLE_from_idxFES are reported the index "0" of Ap2 at row 0 (Ap2) and columns 1 and 2 (Ch2 and Ch3) and the index "1" of Ap3 at row 1 (Ap3) and column 2 (Ch3). In idxFES_from_idxBLE the index "1" (Ch2) is written in row cell 1 (Ch2) and column 0 (Ap2) and index "2" (Ch3) in row cells 2 (Ch3) and columns 0 and 1 (Ap2 and Ap3). The association between the sensors and the channels is complete: the data of Ap2 are received by both channels 2 and 3, and channel 3 receives the combined data of Ap2 and Ap3.

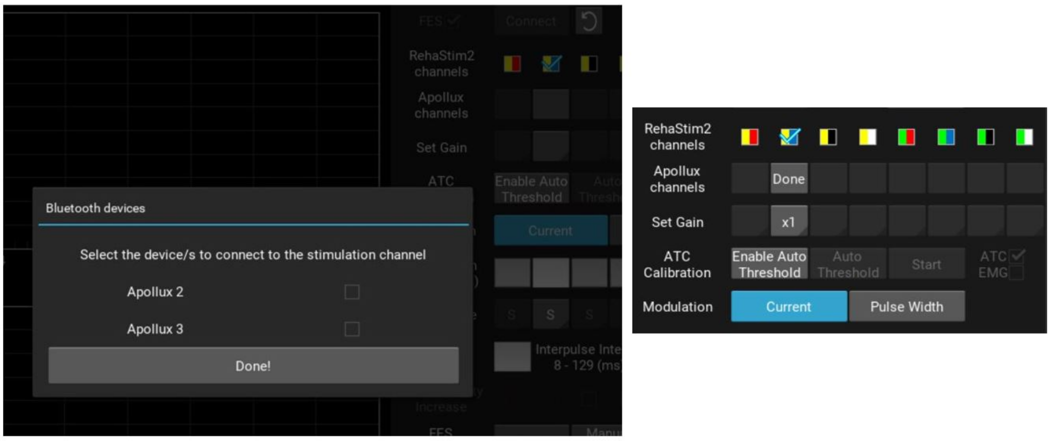

Figure 3.3: Once the stimulation channels have been selected (blue check on *RehaStim2 channels*), the corresponding button to select the Apollux is enabled (*Apollux channels*). By pressing it, a PopUp is displayed in which the devices that can be associated with that stimulation channel are listed (in the left). Once the devices are selected, by pressing *Done!*, the PopUp closes and the association with that channel is completed ("Done" then appears on the button of *Apollux channels*) (on the right).

In the case of 1-N coupling, the data provided by the sensor is processed and transmitted to the N stimulation channels, while in the case of N-1 pairing, since multiple sensors are associated with the same stimulation channel, the data from the N devices must first be combined and then sent to the channel. The mean is the operation that was chosen to combine the data and the reason will be explained later in subsection 3.1.1. The calibration of the therapist had already been illustrated in the subsection 2.3.5: after repeating the movement at least four times, the profiles of the various repetitions are aligned and the median between them is computed, obtaining a single profile (median profile) with the ATC values stored in an array. If N Apollux are associated with the same channel, a median profile is returned for each sensor and, stacking all the N profiles (the N arrays), a matrix with a number of rows equal to the total median profiles (number of Apollux associated with the channel) and a number of columns equal to the number of samples that make up the profiles is created. At this point, the element-by-element mean (rounded up) is carried out (using *atc_profile_cal* method), and a single final array (*Activation_profile*) is obtained (Figure 3.4). This resulting *Activation_profile* is the one that will be used in the patient calibration to determine the I_{max}.

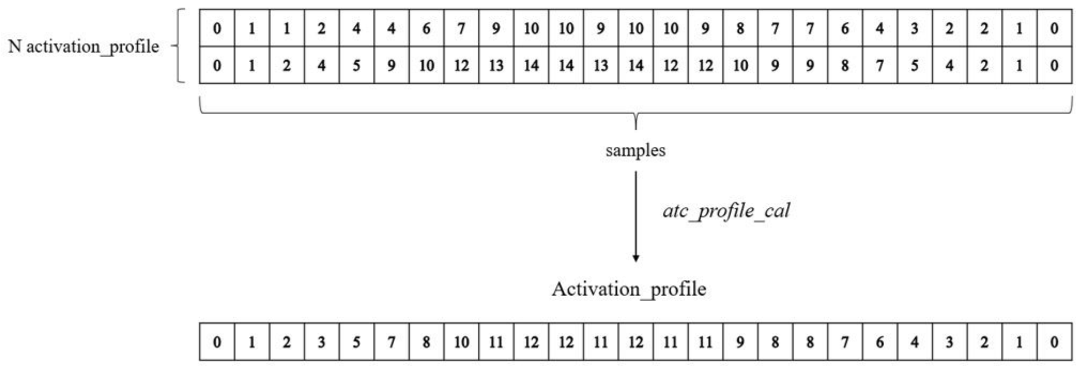

Figure 3.4: An example of what happens during the calibration phase of the therapist to compute a single activation profile: the median profiles obtained from each sensor are collected in different arrays. These are then stacked on top of each other and the element-by-element mean is carried out, using the *atc_profile_cal* method: a single activation profile is calculated for each individual stimulation channel.

Instead, during the training, there is a continuous updating of the ATC data. Each time the new ATC data are transmitted, two matrices are compiled: *mat_atc* matrix involves the new ATC values received (Nx1, with N = new ATC data sent to each channel) and *mat_fes* matrix, which has the same size (Nx1), contains the indices of the stimulation channels associated with the sensors that returned the relative ATCs stored in the *mat_atc*. If several ATC values in *mat_atc* are associated with the same stimulation channel in *mat_fes*, then the rounded-up mean between these values is performed (using *online_mean_atc* method). In the end, a single array (ATC), which contains the average ATC value for each stimulation channel, is returned (Figure 3.5).

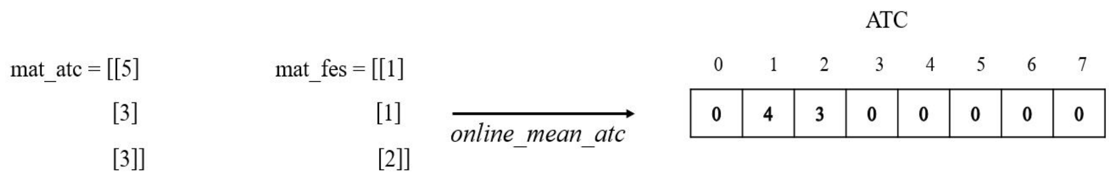

Figure 3.5: Example of what happens during training. Each new ATC data received is saved in the *mat_atc* matrix and the stimulation channel associated with the sensor that returned that ATC value is stored in *mat_fes*. The ATC values associated with the same stimulation channel (5 and 3 both sent to 1 (channel 2)) are mediated. In the end, the method that computes the average of each value, *online_mean_atc*, returns the ATC array in which: at the unused stimulation channels it has an ATC equal to 0; in channels receiving data from a single sensor, the value of ATC received (3 sent to channel 3 (column 2)) is entered; in the channels associated with multiple sensors the measured average is inserted (4 sent to channel 2 (column 1)).

3.1.1 N-1 connection: signals combination operator

If a single device is associated with multiple stimulation channels (1-N association), the data provided by the single Apollux sensor are processed and transmitted to the N stimulation channels. Instead, if multiple acquisition boards are connected to the same FES channel (N-1 association), the data, acquired by the boards, first have to be combined and then the single activation profile or ATC value can be sent to the channel. To preserve real-time FES control, it was necessary to choose a simple method of combining the signals, capable of calculating the final ATC value, for each channel, in a short time. The works in [82][83] were taken as a reference: linear operations are applied on features extracted from the sEMG signals from different channels to determine variables identifying the contraction force required for the execution of a movement (similar to the ATC parameter). In particular, in [82], during the execution of a specific motion, the root mean square (RMS) is extracted from the sEMG signal of each of the 4 acquisition channels, which is a parameter that defines muscle fatigue. Then, to calculate the total contraction intensity associated with the motion m, the RMS values of each channel are summed up. Instead, in [83] different features are extracted from the sEMG signals of the 6 acquisition channels. Among these, the mean absolute value (MAV), which is a parameter that detects muscle contraction levels, is used to compute the recorded motion speed. This speed is calculated by averaging the MAVs of all channels.

Based on these studies, the operations of mean, median, addition, and multiplication were taken into account to understand what was the best method to merge ATC data from N channels. The choice of the most appropriate combination method was made based on the most stable activation profile generated, by each of the operators, at the end of the therapist calibration phase. The activation profile is necessary to create the stimulation pattern to provide to the patient first during the calibration of the I_{max} and then during the training. As explained in subsection 2.3.6, the current calibration phase uses a fixed stimulation profile, which is only gradually increased in intensity. If this stimulation profile does not have an adequate duration of activation or if it does not remain stable for a sufficient period, the movement is not performed correctly or at all during calibration and this leads to a wrong detection of the maximum current tolerated by the patient, which could then cause them pain or acute discomfort in the training phase. The definition of a good activation profile, therefore, becomes crucial. The different operators were tested on the ATC values recorded during the execution of each movement of each protocol, described in section 1.6. Considering that the maximum number of sensors connected to the same stimulation channel was always two (two activation profiles to be combined) and that therefore the median returned the same result as the average, this was discarded immediately. Instead, multiplication and addition were tested for all movements, and, only after a comparison of the generated activation profiles, these two options were also excluded. An example, related to hip abduction, is shown in Figure 3.6, where three different activation profiles obtained by combining the activities of two muscles are represented: Tensor Fasciae Latae (TFL) and Vastus Lateralis (VL). As explained in section 3.1, at the end of the therapist calibration, a median activation profile is returned for each sensor. Figure 3.6 shows the activation profiles computed by averaging, adding, and multiplying the two median profiles. The multiplication operator was initially included in the list of combination methods to test because considering that some muscle activities can remain very low, it would have been able to generate a final activation profile with much greater maximum ATC values. However, as the example in Figure 3.6 shows, the final pattern obtained using multiplication is not only characterized by much higher ATC values than the other two but has also a shorter activation duration. As long as one of the two muscles remains inactive, the ATC data returned are set to 0 and therefore the product between these values and those coming from the other muscle remains null. Once both muscles are activated, all ATC values become greater than 0 and thus also the profile. This approach makes the use of two different muscle activities useless and causes a sudden transition from the inactivity phase to the one in which the movement begins (increase in values from 0 to 50 ATC). Therefore this option has also been excluded. Finally, the addition has also been discarded because, despite having an activation duration and a profile very similar to those of the mean, it usually leads to having a less stable profile, which is ineffective and unsuitable for the correct execution of those movements in which the maintenance of the position for a few seconds is required. Although the example shown is related to a specific exercise, this behavior occurs for all movements: the activation profiles obtained by addition or multiplication were less effective in both I_{max} calibration and training phases. For these reasons, the mean operator was chosen.

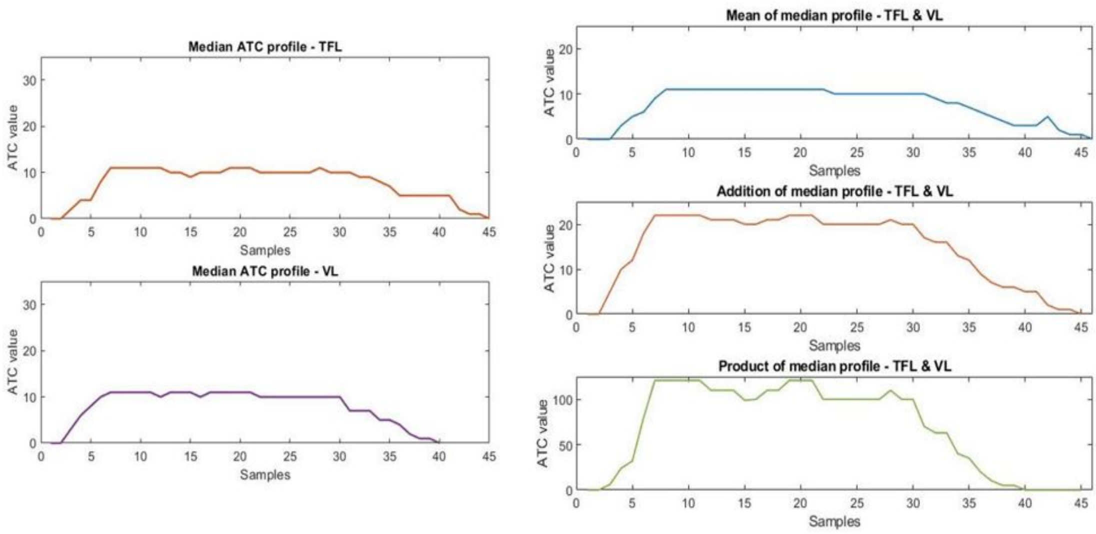

Figure 3.6: Example related to lateral hip abduction, in which Tensor Fasciae Latae (TFL) and Vastus Lateralis (VL) activities (on the left) are used to obtain an activation profile and then a stimulation pattern. The profiles were generated by averaging, adding, and multiplying the data of the two muscles (on the right).

3.2 Pulse Width Modulation

The second change concerns the addition of the option of the pulse width modulation. Several studies [59][60][61], reported in section 2.1, use this type of modulation to test its effects on movements such as knee extension or ankle dorsiflexion. The result that recurs most often is that an increase in pulse width leads to greater muscle contraction and therefore to better execution of the movement, especially if this involves very large or deep muscles. In addition, the possibility to choose between two types of modulations (current or pulse width) increases the chances of understanding what are the best combinations of stimulation parameters that allow the support of the stimulation as long as possible without straining the subject too much and that allow a performance of the movement as natural and comfortable as possible. Pulse width modulation has been designed as explained below. In order to modulate this parameter, it is necessary to establish the limits, lower and upper, within which to vary the parameter. The upper limit has been set at 500 µs, which is the maximum pulse width that can be delivered by the stimulator and a value deemed reasonable considering that the trained muscles, in both protocols, are very large (e.g.: quadriceps, biceps femoris, gluteus maximus) and therefore need high stimulation values to be contracted properly. Taking as an example the technique followed by studies such as [84][85], the lower limit is defined as the value of pulse width that causes a minimum visible contraction of the muscles (motor threshold) and is therefore specific to each subject. Using a fixed current (maximum value defined during the calibration phase of the Imax supported by the patient) and a fixed frequency, starting from 0 µs, the pulse width is gradually increased with a step of 50 µs, as shown in Figure 3.7, until a minimum muscle contraction is observed. A step of 50 µs, already used in other works [85][60], allows to discriminate well the effects of a certain pulse width from those of its consecutive value and to identify the lower limit in a short time.

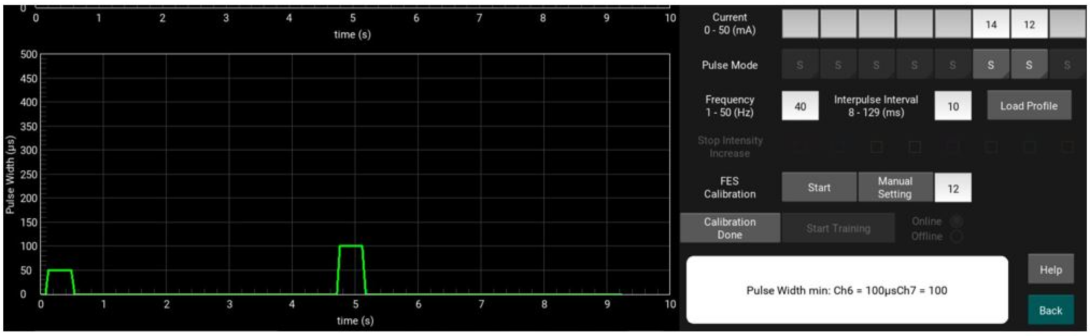

Figure 3.7: Example of minimum pulse width calibration. Using a fixed current (Ch6 = 14 mA, Ch7 = 12 mA), fixed frequency (f = 40 Hz) and interpulse interval (10 µs), starting from 0 µs, the pulse width is gradually increased with a step of 50 µs until a minimum muscle contraction is observed.

Once the upper and lower limits have been found, the Look Up Table (LUT) is determined in a similar way to the case of current modulation. The calibration phase of the therapist provides the maximum ATC of the activation profile of each channel and from the calibration of the patient, the minimum and maximum pulse width are obtained. Also in this case the maximum ATC is matched for each channel with the maximum pulse width (500 µs), what changes is how the correspondence between the other ATC values and the remaining pulse width values is assessed. In this case, the lower limit of the parameter to be modulated is a value greater than 0 (at least equal to 100 µs). Therefore in order to have the right correspondence between all the ATC values (from ATC_{max} to 0) and all the pulse width values that generate a movement (from 500 µs to PW_{min}), it is necessary to calculate the step of how much each pulse width value have to be decremented. To do this, the length of the interval within which the pulse width can vary is divided by the number of total ATCs to be coupled: step = $(PW_{max} - PW_{min})$ / ATC_{max}. In this way, each ATC value detected by the sensor during training will correspond to a certain pulse width value that certainly produces a contraction (Figure 3.8).

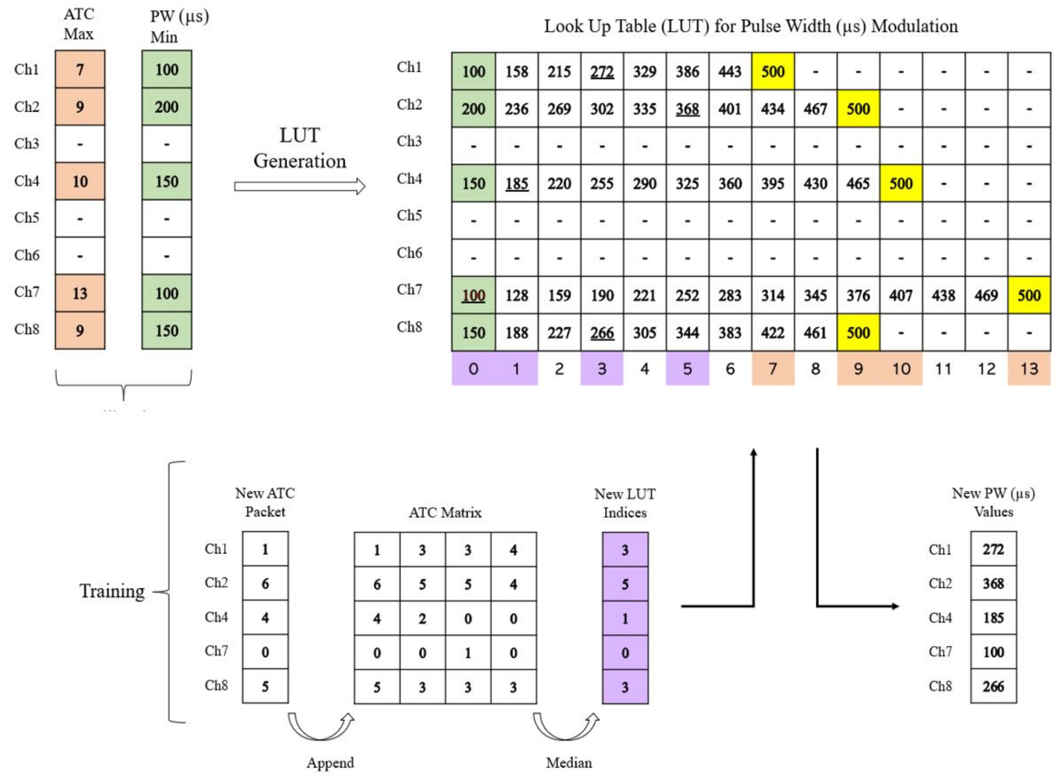

Figure 3.8: Look Up Table (LUT) built at the end of calibration. PW_{max} always remains the same for each channel (500 µs, in yellow). The step for each channel is different as it depends on PW_{min} (in green), which changes for each pair of electrodes (muscle). During training, each new ATC (in purple) is used as an index to locate the new pulse width parameter (underlined).

Chapter 4

Outline Hip and Knee Rehabilitation Protocols

As mentioned in section 1.6, both hip and knee rehabilitation protocols have been outlined by referring to existing protocols used by nursing homes and hospitals and which involve the presence of the physiotherapist to help the patient retrain movements and motor functions [48][49][50][51]. For each protocol, 11 movements have been defined. In particular, the movements are organized in three phases according to their difficulty and the effort required of the patient to perform them. The hip rehabilitation protocol includes:

- 1° phase: ankle flexion-extension; knee flexion supine; knee extension supine; hip abduction supine; hip abduction lateral; hip extension prone;
- 2° phase: hip flexion standing; hip extension standing; hip abduction standing;
- 3° phase: standing heel raises; step-up (get on a step).

The knee rehabilitation protocol includes:

- 1° phase: ankle flexion-extension; knee flexion supine; knee flexion prone; knee extension supine; hip abduction lateral; straight leg raises;
- 2° phase: knee extension seated; hip extension standing; hip abduction standing;
- 3° phase: standing heel raises; step-up.

Movements are common to both protocols because many of the muscles they work on are the same (e.g.: quadriceps femoris, biceps femoris, gastrocnemius, and tibial anterior). That is why later it will be illustrated how the individual movements were obtained. In particular, for each movement, how the activation profile was obtained, necessary to create the stimulation pattern to provide to the patient first during the calibration of the I_{max} and then during the training, will be illustrated. It is known that the current calibration phase uses a fixed stimulation profile, which is only gradually increased in intensity. If this stimulation profile does not have an adequate duration of activation or if it does not remain stable for a sufficient period, the movement is not performed correctly or at all during calibration and this leads to a wrong detection of the maximum current tolerated by the patient, which could then cause them pain or acute discomfort in the training phase. The definition of a good activation profile, therefore, becomes crucial. In addition to this, for each movement, the reasons that led to the positioning of the surface electrodes in a certain way will also be reported. This is also a fundamental aspect, considering that the success of a movement depends on the placement of the electrodes. For movements that were performed in two different initial positions, the conclusions regarding activation profile and standardized placement of the electrodes remain the same, so the procedure followed for these exercises will be described only once. It is necessary to specify that for those movements where muscle activities from two different muscles were exploited, the combination of data was performed through the operation of the mean, as explained in subsection 3.1.1

4.1 Ankle Flexion - Extension

The patient lies supine on the hospital bed and moves the ankle up (flexion) and down (extension) consecutively. It is an exercise that aims to strengthen the calf muscles and recover the mobility of the ankle, which will be particularly important when the patient resumes walking in the later stages of treatment. For this reason, it is an exercise that is performed in both protocols [52][53].

4.1.1 Activation Profile

From the information provided in an FES study applied to ankle exercises [86], it was concluded that the main muscles responsible for the dorsiflexion (flexion) and plantar flexion (extension) of the ankle are the tibialis anterior (TA) and gastrocnemius (GA), respectively. To obtain the relative activation profiles, shown in Figure 4.1, the acquisition electrodes were positioned on the two muscles following the indications of the SENIAM project [87]. During the flexion (orange profile) there is only the activation of the TA, instead, during the extension (blue profile), besides the strong activation of the GA, there is also a slight cross-talk of the TA channel. The latter, however, is associated with very low ATC values, and therefore in current modulation, this activation is translated into values of current intensity so low that they are not perceived by the patient, while in the other modulation the corresponding values of pulse width are much lower than those of the GA, so the resulting movement is, in the end, the only extension (followed by the flexion).

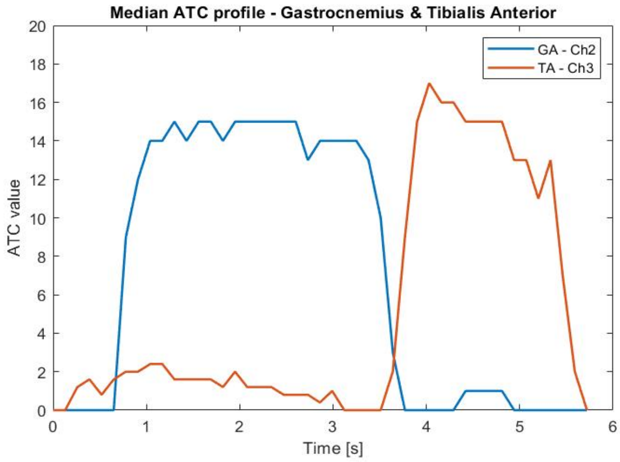

Figure 4.1: Activation profiles generated by Gastrocnemius (GA) and Tibialis Anterior (TA) during ankle extension and flexion, respectively. Slight activation of the TA even during the extension, but since it remains much lower than that of the GA, during the stimulation phase it is not perceived by the patient, making him perform only ankle extension. During the flexion, on the other hand, only the TA is activated.

4.1.2 Electrodes Placement

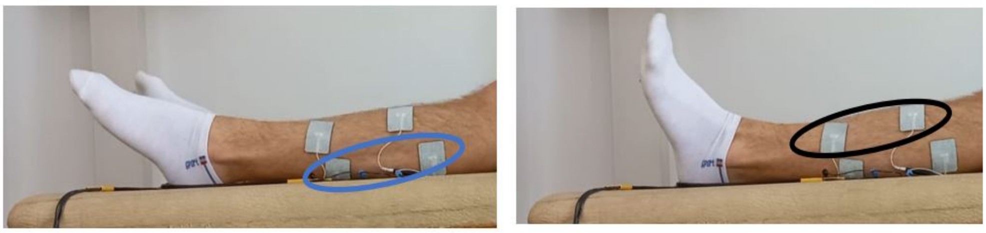

Figure 4.2: Placement of surface electrodes. The ankle extension (on the left) is made through the electrodes of channel 2 (Ch2 – in blue) placed on the GA: proximal electrode placed at 20% of the line tip of fibula - heel (GA medialis and lateralis) and distal electrode positioned at 50% of same distance (soleus). The ankle flexion (right) is realized by the electrodes of channel 3 (Ch3 – in black) placed on the TA: proximal electrode placed at 25% of the line tip of fibula - tip of medial malleolus (TA) and distal one at 60% of the same distance (extensor digitorum longus)

Since the ankle flexion-extension exercise is one of the most analyzed in studies concerning the applications of FES, it is very easy to find information regarding the relative positioning of the surface electrodes. In particular, the indications provided by [88] were followed, shown in Figure 4.2. The muscular activities, recorded from the GA and TA of the therapist, are translated into stimulation patterns that are supplied to the electrodes of the stimulation channels placed respectively on the GA (Ch2) and TA (Ch3) of the patient.

4.2 Knee Flexion - Prone

The patient lies prone on the hospital bed with his feet coming out and flexes the knee bringing the heel towards the buttock. This exercise aims to strengthen the hamstring muscles responsible for knee flexion [52].

4.2.1 Activation Profile

Considering that the muscles responsible for knee flexion are the hamstring muscles, the activation profiles generated by semitendinosus (SemT), semimembranosus (SemB), and biceps femoris (BF) were evaluated. Looking at the activation profiles in Figure 4.3, no useful activity was recorded from SemB: the values were never higher than 1 ATCs, even increasing the gain of the acquisition board. Looking at the profiles of SemT and BF, there is a clear predominance of the muscle activity of BF, which is an indication of the preferential recruitment of this muscle during the execution of the movement, as also supported by [89]. For these reasons, the SemT profile was selected as the only activation profile.

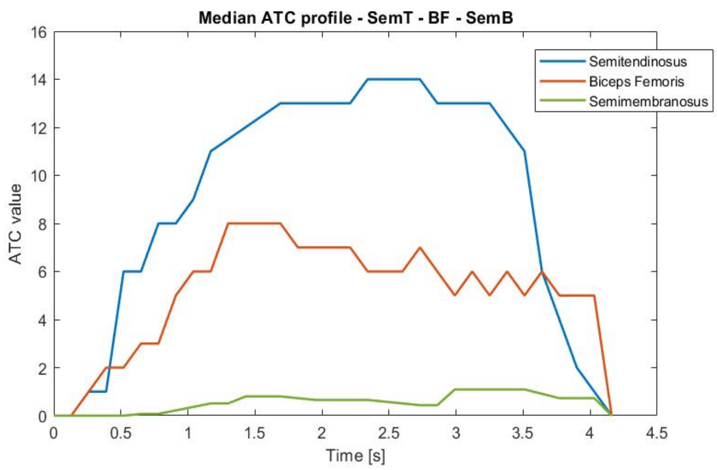

Figure 4.3: Activation profiles generated by semitendinosus (SemT), biceps femoris (BF) and semimembranosus (SemB) during prone knee flexion. SemT activates much more during the flexion than the BF.

4.2.2 Electrodes Placement

Also for this exercise, the indications provided by the [88] were followed, which suggests the use of a single stimulation channel. The two electrodes were then positioned to stimulate both the biceps femoris and semitendinosus, which are the primary muscles that allow knee flexion.

Figure 4.4: Placement of surface electrodes. The knee flexion (on the left) is realized through the electrodes of channel 2 (Ch2 – in blue) placed on the BF and SemT: proximal electrode placed at 30% of the line ischial tuberosity - medial epicondyle of the tibia (SemT and BF) and distal electrode at 70% of the same distance (SemT and BF). On the right a better view of the placement of the electrodes.

4.3 Knee Flexion - Supine

The patient lies supine on the hospital bed and bends the knee bringing it towards the chest by sliding the heel on the bed. Then it returns to the initial position. The execution of this movement occurs mainly thanks to the knee flexion, but since this also induces a hip flexion, this exercise is used by both protocols [52][53].

4.3.1 Activation Profile

The most important part of the whole movement is the flexion of the knee. Therefore, the activation profile used to carry out this first phase is the one also used in prone knee flexion (SemT), subsection 4.2.1. The return to the resting position instead takes place by extending the bent limb and therefore the activity of the various knee extensors was recorded, but the only muscle that reported a slight activation was the rectus femoris (Figure 4.5).

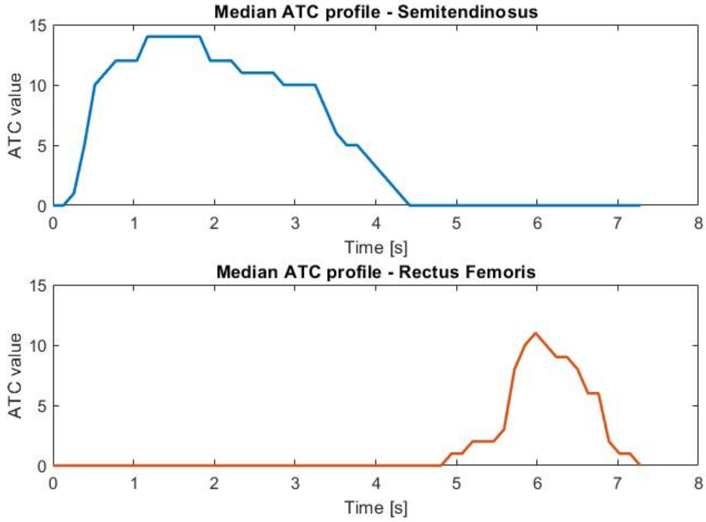

Figure 4.5: Activation profiles generated by semitendinosus (SemT) during flexion and rectus femoris (RF) during the return to the starting position. The activation of the SemT is much longer than the RF since the flexion movement requires a longer activation time to be performed correctly: if the SemT is activated only for a couple of seconds, the knee flexion stops at the beginning or middle of its total range of motion. For RF, on the other hand, a contraction force of a few seconds is enough to return the leg to its initial position, probably because the quadriceps femoris muscles are much more responsive to stimulation.

4.3.2 Electrodes Placement

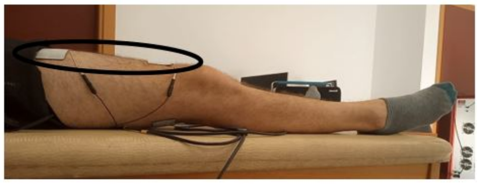

Figure 4.6: Placement of surface electrodes. The knee flexion (on the left) is realized through the electrodes of channel 2 (Ch2 – in blue) placed on the BF and SemT, like in the knee flexion prone. The return to the resting position (on the right) is obtained by the electrodes of channel 3 (Ch3 – in black) placed on the RF and VM: proximal electrode at 40% of the line anterior spine iliac superior - proximal border of patella (RF) and distal one at 80% of the same distance (RF - VM).

The electrodes of the channel that receives data from the sensor placed on the SemT (knee flexion) have been positioned following the same rules used for the prone knee flexion. Whereas the electrodes of the second stimulation channel used to extend the knee were both placed on the RF, thus only partially following the indications reported by [88], which instead provides for the placement of the proximal electrode on the vastus lateralis (VL). However, considering that VL stimulation can cause pain or acute discomfort in some cases, RF was chosen.

4.4 Knee Extension - Supine and Seated

In the supine knee extension, the patient is lying supine on the bed with support (e.g: foam roller) placed below the knee, while in the seated extension, the patient is seated on the edge of the bed with the legs free to move. From these positions, the subject lifts the heel until it reaches the complete extension of the leg. After holding this position for a few seconds (3-4 s), the limb is returned to the initial position [49]. In the case of the seated extension, the initial position involves flexion of the knee of 90°, while in the extension from supine this angle is much smaller (30°-40°) [51]. This implies that, when the patient performs the movement while sitting, the range of motion is much wider, and much more muscle strength is required than in the supine position. Knee extension exercise is aimed at strengthening the muscles of the femoral quadriceps, which are the main extensors of the knee. For this reason, considering that the goal of the knee rehabilitation protocol is to work on the muscles that affect the joint, the exercise is performed in both versions: the supine one trains the knee extenders without stressing them too much and the sitting one intensifies the strengthening of the muscles [52]. As for the intervention of hip prostheses, it is essential to act not only on the specific muscles of the joint (gluteus, tensor fasciae latae, iliopsoas) but also on the other muscles of the leg to then subsequently guarantee an easier and faster recovery of the functions that involve the entire lower limb, such as walking or climbing stairs. For this reason, this protocol also includes the exercise of extension of the knee from supine to train the quadriceps femoris [53].

4.4.1 Activation Profile

The main challenge in the realization of this activation profile was to obtain a signal that was stable enough to allow the maintenance of the knee extension for a few seconds. The individual profiles generated by the muscles of the femoral quadriceps were analyzed, taken at the two points where the stimulation electrodes are positioned (on the patient): 40% and 80% of the distance between the superior anterior iliac spine and the proximal border of the patella (defined according to [88]). Among the signals, at 40% the vastus medialis (VM) does not appear because at this distance the recorded profile was too small and therefore unusable. For both sampling points, as shown in Figures 4.7 (on the left), the most stable profiles are those of the vastus lateralis (VL at 40% and 80%), which, however, taken individually, do not have a sufficiently constant activation to allow the leg to remain in extension for 3–4 s. By averaging between the two profiles (Figures 4.7, on the right), a profile that for at least 4 s manages to remain at the maximum ATC value is achieved.

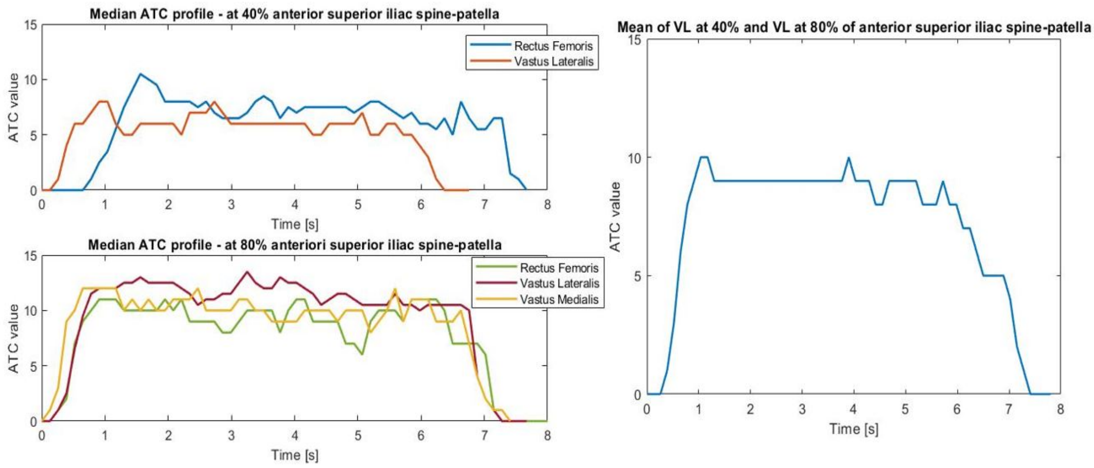

Figure 4.7: Activation profiles returned by the individual muscles of the quadriceps femoris (RF, VM and VL) at 40% and 80% of the superior anterior iliac spine distance – proximal border of the patella (dp-asis, on the left). On the right, the mean between profiles of VL at 40% and at 80% of dp-asis.

4.4.2 Electrodes Placement

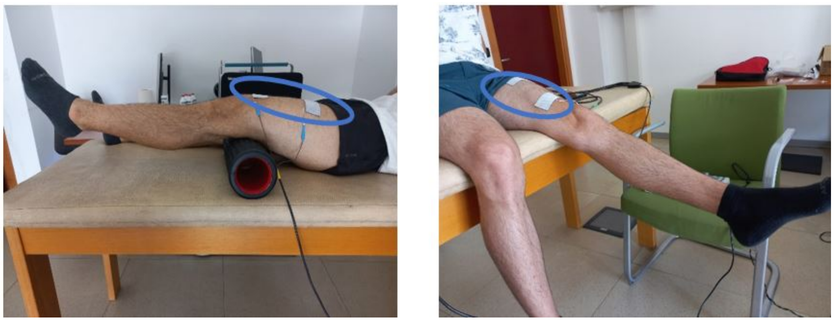

Figure 4.8: Electrodes placement for the knee extension (supine position on the left and seated position on the right): proximal electrode at 40% of the line anterior superior iliac spine - proximal border of patella (RF) and distal at 80% of the same distance (RF-VM).

The positioning of the electrodes was performed following the indications of [88], but placing the proximal electrode on the RF and not on the VL (as in the supine knee flexion) for reasons related to the discomfort perceived by the patient during stimulation. Although the activation profile is taken from the VL, it can be used to contract RF and VM since the timing and forms of activation of the three muscles are the same.

4.5 Straight Leg Raise

The patient is lying supine and raises the extended leg up to about 20-30 cm. It holds the position for 3-4 s and then returns to the resting position. Such exercise allows the strengthening of the quadriceps muscles and is therefore recommended for knee rehabilitation. Although the movement is also achieved through a hip flexion, it is not suggested for hip protocol as it subjects the newly implanted prosthetic implant to too high stresses [51].

4.5.1 Activation Profile

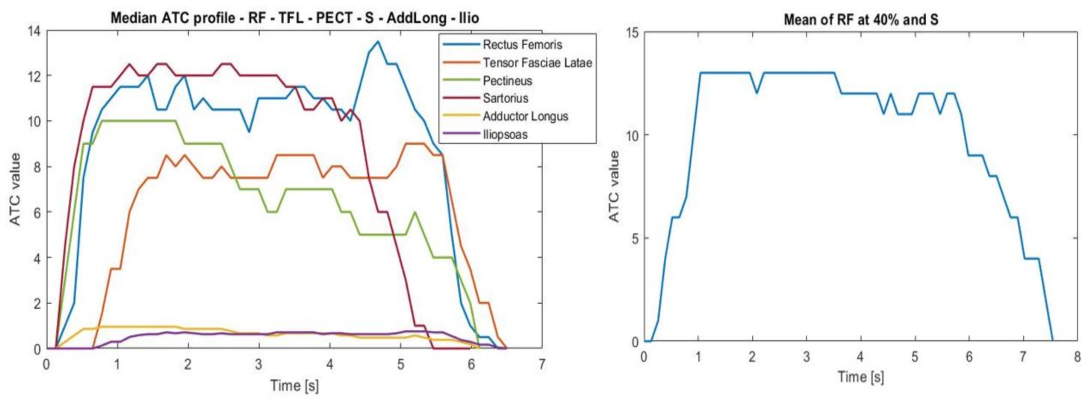

Figure 4.9: Activation profiles returned by the primary flexor muscles (RF, S, TFL and PECT, on the left). PECT has a strong activation that then decreases during the execution of the exercise. RF is the first muscle to activate, along with S, and the one with the longest duration. TFL is the last one to be activated and has a less intense activity, but the signal generated remains particularly stable. On the right, the mean is computed between the RF and S, which are the muscles with the highest activity and similar activation/deactivation times.

Since this movement consists of an extended leg hip flexion, the activation profiles were recorded by the primary flexor muscles, which according to [27], are the iliopsoas, RF, sartorius (S), tensor fasciae latae (TFL), adductor longus and pectineus (PECT). To place the acquisition electrodes on the various muscles, several indications were followed. The distances reported in SENIAM [87] and study [90] were used for TFL and, PECT and adductor longus, respectively. While for iliopsoas, S, and RF the work in [91] was taken as a reference. In Figures

4.9 (on the left) the signals acquired by RF, S, TFL and PECT appear, while for iliopsoas and adductor longus no activity has been detected. The best profiles in terms of activation intensity and signal stability are those of RF, S and TFL, which activates slightly after the first two. However, to make the activation profile as stable as possible, to allow the patient to keep the leg up for 3-4 s, only the RF and S signals were mediated, as they are the muscles that most of all are activated (on the right).

4.5.2 Electrodes Placement

Two stimulation channels and so four stimulation electrodes in total were used. Based on the analysis of muscle activity described above, two stimulation electrodes were placed on the TFL-S and S (channel 2 -Ch2). These two electrodes allow the raise of the leg (flexion of the leg). While the electrodes of the second channel have been placed on the RF and have the task of keeping the leg extended throughout the movement and helping in lifting the leg.

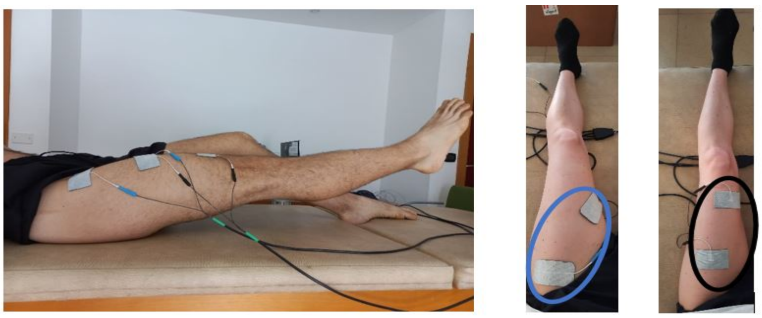

Figure 4.10: Electrodes placement used for straight leg raise. In stimulation channel 2 (Ch2, on the left in blue): proximal electrode placed at 1/6 of the line anterior superior iliac spine - lateral femoral condyle (TFL and S) and distal one on S, at 100 mm from the point of origin of S (70mm from the anterior superior iliac spine). For stimulation channel 3 (Ch3, on the right in black), the electrodes were positioned as for knee extension.

4.6 Hip Abduction - Supine and Standing

The patient is lying supine on the bed or is standing and from these initial positions, he performs the hip abduction by opening the leg outwards, keeping the position for a few seconds, and then returning to the starting position [49]. Since the aim of both exercises is to strengthen the hip abductors, they are therefore included in the hip rehabilitation program [53]. However, hip abduction from standing, which involves standing on only one limb, is an exercise that is also used to retrain the ability of the patient to maintain balance. For this reason, it is a movement also included in the knee rehabilitation protocol [51].

4.6.1 Activation Profile

The activities are acquired from the muscles that [27] describes as the primary and secondary hip abductors in the frontal plane. Among these, however, only the most superficial muscles were considered, from which it is possible to collect a good activity with surface electrodes. In particular, gluteus medius (GMed) and TFL were analyzed among the primary muscles and S and RF among the secondary ones. The profiles obtained are reported in Figure 4.11 (on the left), in which is possible to see that the activation of GMed, as also reported by the study [92], remains minimized during hip abduction in the frontal plane.

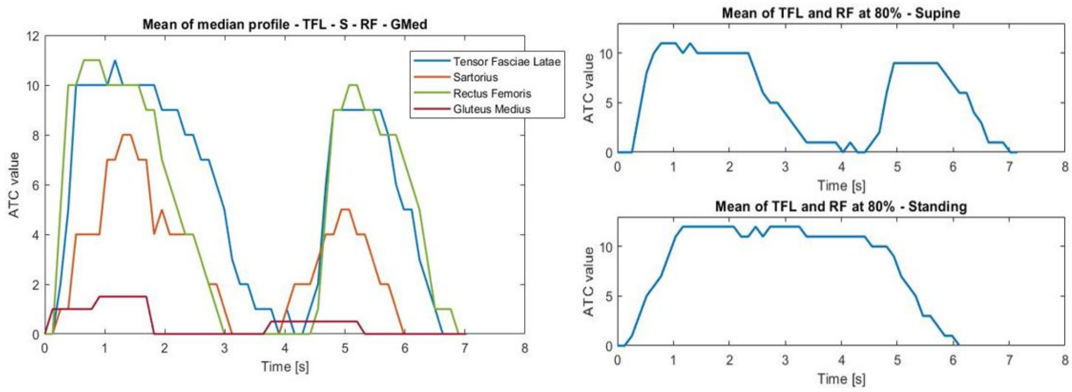

Figure 4.11: Activation profiles returned by the main abductor muscles (RF, S, TFL and GMed, on the left). GMed has almost no muscle activity. S activates for a too short time period to achieve abduction. TFL presents a strong and long activation, but it takes too much time to deactivate, so the patient perceives a single stimulation (instead of two). RF is faster to deactivate and activate and therefore two distinct phases are always obtained, but sometimes it does not remain too stable. On the right are reported the two activation profiles receive for the supine and standing positions.

The acquisition electrodes were positioned on TFL, S and RF following the same indications described in Straight Leg Raise. The profiles of the three muscles are characterized by two activation phases: the first one is necessary to bring the leg outwards and keep it a few seconds in this position and the second curve is used instead to bring the leg back to the initial position. The two curves must be separated by a phase of non-activity or very low activity to return the leg to the initial position. Only when the leg is already open to the outside (position reached with the first curve), the stimulation pattern is able to bring the leg back to the axis of the body. These considerations were made only for the supine version of the movement as standing only needs an activation curve to perform the abduction, while the return to the resting position occurs thanks to gravity. Looking at the signals, the greatest activations are obtained from TFL and RF. However, neither signal, taken individually, is good as an activation profile. The TFL presents an activation duration in the first curve sufficient to achieve the hip abduction and hold the position, but it takes too long to deactivate or reach very low ATC values (pause between one curve and another last less than 1 s). This translates, at the level of stimulation, into the perception of a single wave of stimulation instead of two. On the other hand, the RF is able to deactivate and reactivate very quickly (pause of more than 1 s), but it is often not too stable. By averaging between the two signals, an activation and stable profile is obtained, which, in the case of supine abduction, is characterized by two distinct activation curves (on the right).

4.6.2 Electrodes Placement

The electrodes were placed on the muscles which, as described above, are most activated and therefore only one stimulation channel was used: one electrode has been placed on the TFL (in the same place used for the acquisition electrode), but slightly shifted inwards to contracts also the RF and the other one has been positioned on the RF. The electrode on the TFL is the one that mainly allows the abduction and the electrode on the RF is used to keep the leg extended throughout the movement and to help in the abduction.

Figure 4.12: Electrodes placement for the supine and standing hip abduction, which involves only one stimulation channel: proximal electrode at 1/6 of the line anterior superior iliac spine - lateral femoral condyle (TFL) and distal at 80% of the line anterior superior iliac spine - proximal border of patella (RF).

4.7 Hip Abduction - Lateral

The patient is lying on the side of the unaffected limb, raises the operated, keeping it extended, leg about 20-30 cm, and holds it in this position for 3-4 s. Then returns to the initial position [49]. Since lateral hip abduction allows a strengthening of hip abductors (TFL and S), gluteus muscles, and also quadriceps femoris muscles, it is a recommended exercise for both patients with THA and TKA [49][52].

4.7.1 Activation Profile

Even though the movement performed by the joint is the same, lateral hip abduction has a different muscular activity from those observed in the supine and standing positions. According to the article [93], in the lateral position, the muscles that show the highest activations are TFL, gluteus medius and maximus (GMed and GMax). While the study [94] reports that, for angles of hip abduction from 0° to 30°, a strong activation also occurs at VL. The acquisition electrodes were placed on each of these muscles following the directions of the SENIAM [87], except for RF which was acquired as in the case of the supine hip abduction. Looking at the Figure 4.13 (on the left), TFL and VL prove to be the best both in terms of activation intensity and signal stability. While RF was not considered in this version of the movement since it is much more unstable and therefore less effective. Both the individual TFL and VL profiles are stable enough to ensure a stimulation pattern that allows the patient to lift the leg and hold it in place for a few seconds during the I_{max} calibration phase. However, this does not always occur during the training phase, in which between two consecutive ATC values there may be a variation such as to induce a shaky stimulation, which prevents the correct execution of the movement. To be sure that the signal was consistent even during the training, the average between TFL and VL (on the right) was performed.

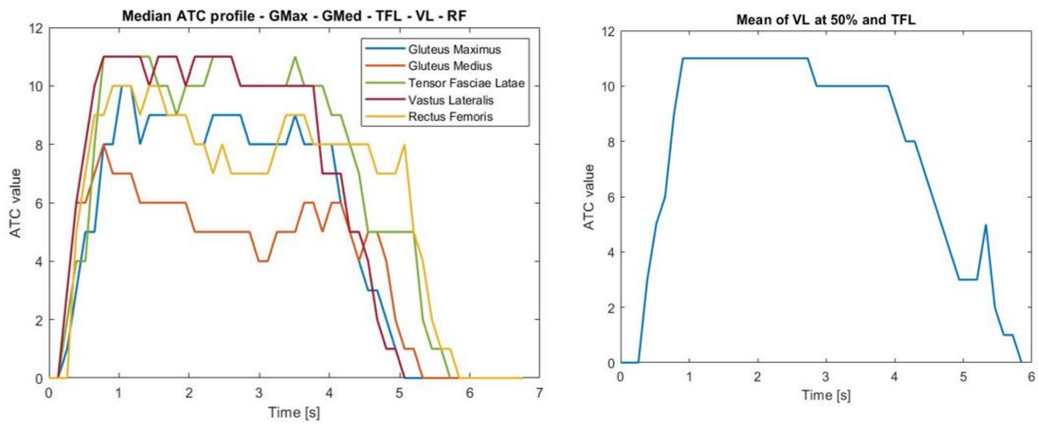

Figure 4.13: Activation profiles returned by TFL, Gmax, Gmed, RF and VL (on the left). Gluteus muscles and RF signals are too unstable. On the right, activation profile calculated through the average performed between TFL and VL.

4.7.2 Electrodes Placement

In the case of supine abduction, hip abduction occurs parallel to the plane of the bed and therefore the force required to perform the movement remains the same along the entire length of the limb.

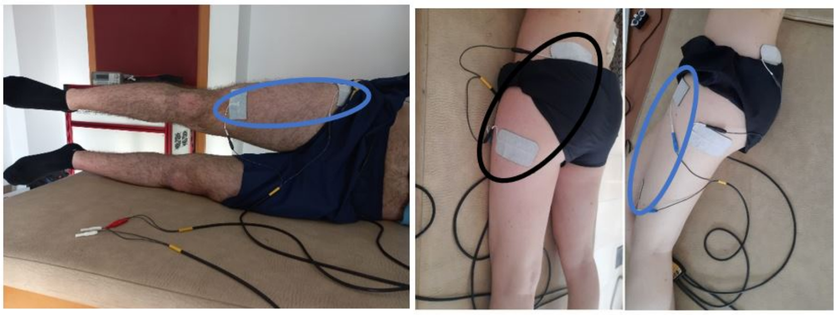

Figure 4.14: Electrodes placement for the lateral hip abduction, which involves two stimulation channels. For stimulation channel 2 (Ch2, in blue), the electrodes are placed as for the supine and standing hip abduction. For stimulation channel 3 (Ch3, in black): proximal electrode positioned at 50% of the line iliac crest - trochanter (GMed) and distal at below the greater trochanter along the line of TFL.

In the lateral case, on the other hand, the leg is moved perpendicular to the bed, and as the leg is lifted an increasing force of contraction is required. So just the only two electrodes used for supine abduction are no longer sufficient to achieve the desired movement, but another stimulation channel must be used. Following the indications of [88], two additional electrodes were placed at the gluteus medius and the TFL.

4.8 Hip Extension - Prone and Standing

The patient is lying prone on the bed or is standing and raises the leg, keeping it extended, up to about 10-20 cm. After holding the position for 3-4 s, the limb is returned to the resting position. The exercise performed from the prone position is included in the hip rehabilitation protocol as it has the task of strengthening the hip extensors (gluteus muscles and hamstring muscles) [53]. Instead, the extension from standing is a type of exercise that is not only useful to recover the balance of the patients but is also a preparatory exercise for the retrain of proper walking. For these reasons, it is included in both rehabilitation programs [51].

4.8.1 Activation Profile

The muscles that participate most in this movement, as reported by studies like [95] [96], are the SemT and the gluteal muscles. In particular, as evidenced by [97], the hip extension takes place thanks to progressive muscular activations: even before the extension of the limb begins, RF and SemT activations occur (RF remains inferior to SemT) followed by GMax activation as soon as the leg is raised. This time delay that elapses between muscle activations, is much more evident in the case of prone extension, while when the exercise is performed from standing, the muscles are activated almost at the same time. The considerations just made are summarized in Figure 4.15 (on the left). To avoid losing the information on the activation delay of GMax, for the prone extension two different activation profiles are used (on the right): one obtained from the average between SemT and Gmax (in blue), used to contract the RF, and the other obtained instead considering only the GMax (in orange) and used to contract the gluteus muscles. While in the case of the standing hip extension, it is possible to use only one activation profile for all the muscles since there is no need to contract the gluteus muscles later.

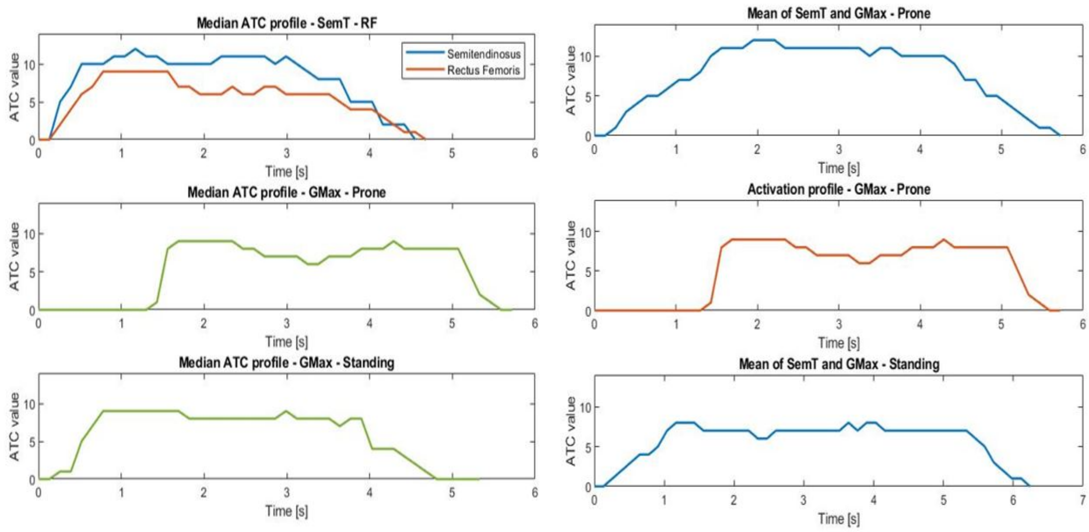

Figure 4.15: Activation profiles obtained for the muscles involved in the hip extension (RF, SemT and Gluteal muscles, in the left). On the right the activation profiles used for the calibration and training phase in the two different versions of the movements. For prone extension, the mean between SemT and GMax is performed to obtain a more stable signal when both muscles are activated and is used to contract the RF. The activation of the GMax alone is used to provide a later contraction of the gluteus. In standing hip extension only the profile of the mean is used to generate a simultaneous contraction on all the muscles of the patients involved (RF and gluteus).

4.8.2 Electrodes Placement

The positioning of the electrodes was chosen following the information provided by [88], which indicates two electrodes placed on the gluteus muscles for the realization of the extension.

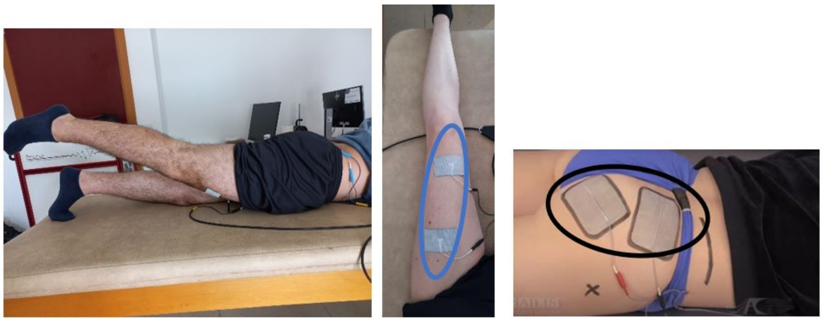

Figure 4.16: Electrodes placement for the prone and standing hip extension, which involves two stimulation channels. For stimulation channel 2 (Ch2, in blue), the electrodes are placed as for the knee extension. For stimulation channel 3 (Ch3, in black): proximal at 50% of the line iliac crest - trochanter (GMed) and distal at 50% of the line sacral vertebrae - greater trochanter (GMax).

However, the contraction of these muscles alone did not allow the patient to lift the leg, so another pair of electrodes was added on the RF. When the RF contracts it detaches the patient's leg from the bed, making it easier to perform hip extension. Moreover, the contraction of the RF keeps the knee extended throughout the movement. The pair of electrodes was not placed on the hamstring muscles because, although they are involved in hip extension, their contraction only led to undesired knee flexion.

4.9 Hip Flexion - Standing

The patient is standing and raises the leg toward the chest, keeping the knee flexed at about 90°. After holding this position for 3-4 s, he returns to the starting position. As this exercise is intended to strengthen hip flexors is recommended only for patients with THA [49].

4.9.1 Activation Profile

The hip flexion and the straight leg raise are the same exercise, but in the first one the knee is flexed and in the second one, the knee is extended. So since both movements involve the same muscles, the same signals were used for the delineation of the activation profile. From the analysis of the signals of the hip flexors, also in this case, the best activations were found for RF and S (Figure 4.17, on the left), and by mediating them the activation profile was obtained (on the right).

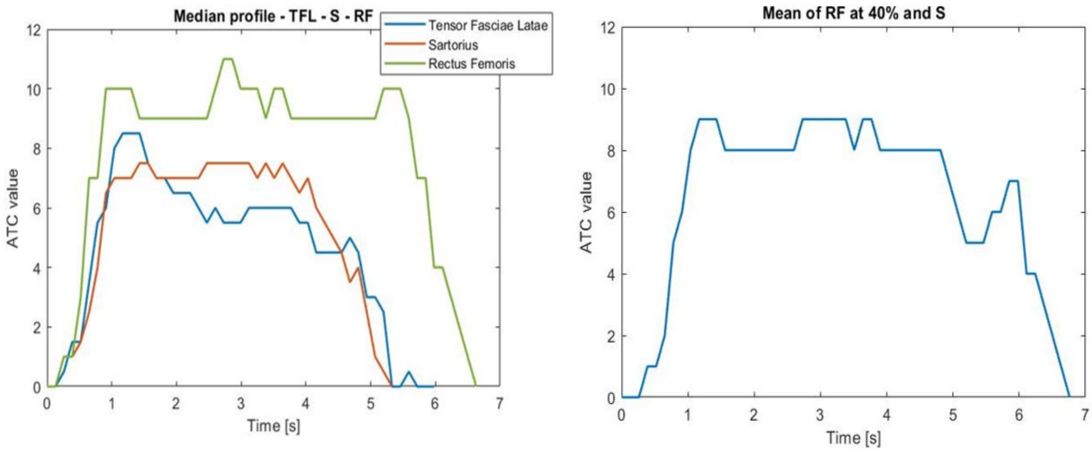

Figure 4.17: Activation profiles returned by the primary flexor muscles (RF, S and TFL, on the left). On the right, the mean computed between the RF and S.

4.9.2 Electrodes Placement

Figure 4.18: Electrodes placement for the standing hip flexion, which involves two stimulation channels. For stimulation channel 2 (Ch2, in blue), the electrodes are placed as for the knee flexion. For stimulation channel 3 (Ch3, in black): proximal electrode at 1/6 of the line anterior superior iliac spine - lateral femoral condyle (TFL and S) and distal electrode on the S at 100 mm from the point of origin of (70mm from the anterior superior iliac spine).

As mentioned before, since the hip flexion is practically the same exercise as the straight leg raise, the positioning of the electrodes was also established following the same rules. So to achieve hip flexion, two electrodes were placed on TFL and S. Moreover, considering that the knee remains flexed, the electrodes of the second stimulation channel were placed on the SemT and BF, as for the knee flexion. Although the activation profile is generated by two muscles completely different from the hamstrings (S and RF), it was considered reasonable to use it to create the stimulation pattern to be provided to the SemT since the activation of the latter during knee flexion is similar to the one returned by the average of S and RF.

4.10 Heel Raise

The patient is standing and lifts his heels, shifting the weight to the toes. This position is maintained for 3-4 s, leaning on support to maintain balance if necessary, and then returns to the initial position by resting the heels on the ground. It is an exercise that is usually performed both by patients with THA, to strengthen mainly the gluteus muscles, and by patients with TKA, to strengthen the calf muscles. In addition, it is an important exercise for the recovery of balance [55][56].

4.10.1 Activation Profile

This type of movement is carried out through the action of the muscles of the entire lower limb. The lifting of the heels is generated mainly by the action of the GA, which induces the plantarflexion of the ankle and leads the patient to shift the weight on the toes. The action of the GA, however, remains concentrated in the calf area, so if the patient were electrostimulated only in this area, he would tend to raise his heels with his knees bent. In order for the patient to be able to remain on the toes while keeping the entire leg extended, the simultaneous activation of the muscles of the quadriceps femoris is necessary. Based on what has just been said, the signals coming from GA and quadriceps muscles were analyzed and obtained by positioning the acquisition electrodes according to the indications of the SENIAM [87]. In Figure 4.19 (on the left), the acquired profiles are represented.

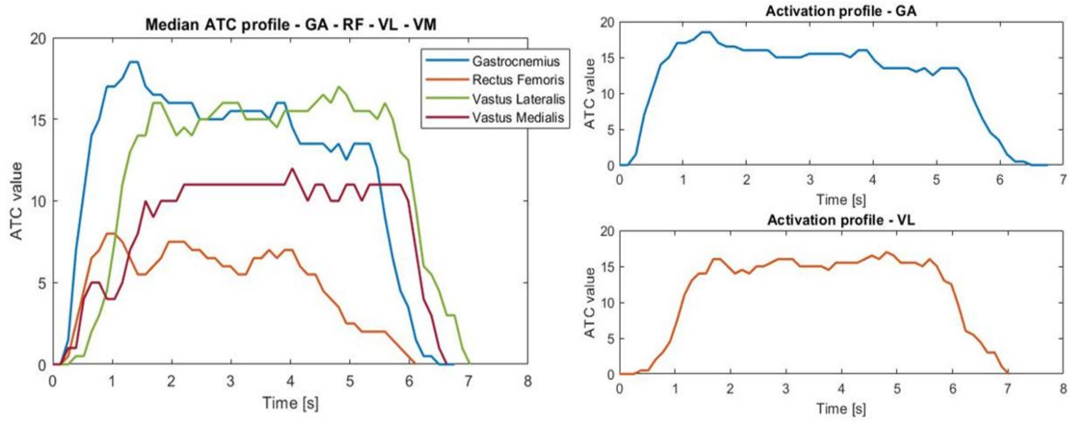

Figure 4.19: Activation profiles returned by gastrocnemius (GA) and quadriceps femoris muscles (RF, VM and VL, on the left). On the right, the two activation profiles chosen.

Looking at the profiles generated by the muscles of the femoral quadriceps, the VL is the muscle that shows the greatest activity. Since both VL and GA signals manage to remain stable at the maximum ATC value even during the training phase, they were chosen as the two activation profiles (on the right) to be provided each to a stimulation channel. In addition, since these muscles are activated almost at the same time, they generate two stimulation patterns that simultaneously contract the calf and thigh muscles, allowing the patient to perform the movement fluidly.

4.10.2 Electrodes Placement

Figure 4.20: Electrodes placement for the heel raise, which involves four stimulation channels. For stimulation channels 2 and 3 (Ch2 and Ch3, in blue): proximal electrode at 20% of the line tip of fibula - tip of medial malleolus (GA medial and lateral) and distal at 50% of the same distance (soleus). For stimulation channels 4 and 5 (Ch4 and Ch5, in black): proximal electrode at 40% of the line anterior superior iliac spine - proximal border of patella (RF-VM) and distal at 80% of the same distance (RF-VL).

By placing the electrodes only on the affected limb, the movement would be induced only on one leg, forcing the patient to follow the stimulation by voluntarily contracting the unaffected leg. The execution of the exercise in this way is particularly difficult for the subject, who feels a different contraction between the two legs and does not understand how to perform the movement correctly. To make the execution of the movement easier and more intuitive, it was decided to place the electrodes also on the undamaged leg. The stimulation channels are therefore a total of four: two positioned on the calves of each leg (as mentioned in [88]) and the other two on the femoral quadriceps (as for knee extension).

4.11 Step-up

The subject is standing in front of a step with a maximum height of 15 cm. He puts the foot of the operated limb on the step and gets on it. Then he returns to the initial position by descending from the step with the other leg. It is an exercise that is performed at an advanced stage of the rehabilitation program and helps the patient to retrain the action of climbing the stairs without any external support. For this reason, it is included in both the hip and knee rehabilitation protocol [55][56].

4.11.1 Activation Profile

In order to carry out the step-up movement in the simplest way, the exercise has been divided into two phases. The first one consists of placing the foot on the step, which is realized by a hip flexion movement. The second phase consists of the extension of the knee (pull-up phase) to bring the other limb to the step. However, it is necessary that the height of the step remains limited (no higher than 15 cm), otherwise, the FES applied to the quadriceps muscles alone is not able to lift all the weight of the body. So to perform the two movements consecutively (hip flexion + knee extension), two activation profiles are required: one that allows only the flexion of the hip and one for the extension of the knee. The muscles, already seen in standing hip flexion, straight leg raise and knee extension, were then analyzed to see if they show different activations for this specific exercise. Figure 4.21 shows the signals acquired by the muscles TFL, S, RF (at the same distances specified for straight leg raise) and VL (in the same two points seen for knee extension) and the two phases of movement separated by the dotted line in black.

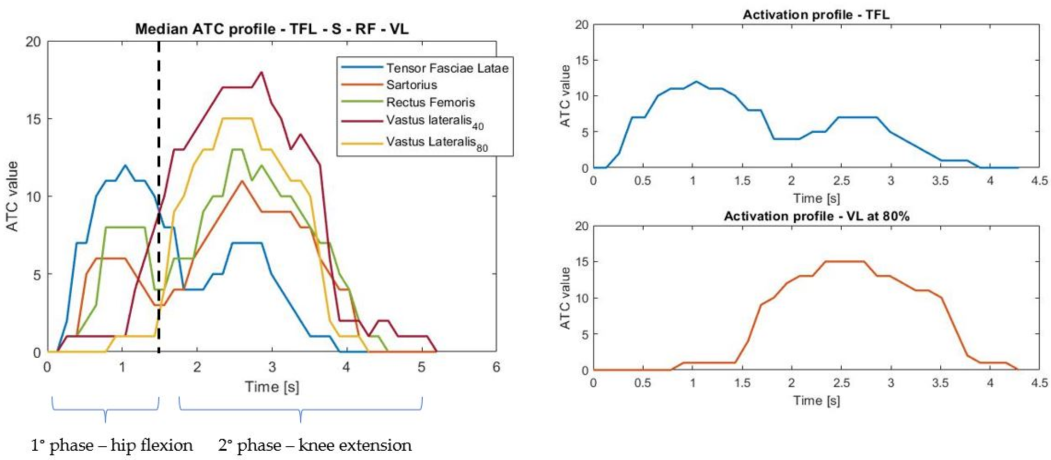

Figure 4.21: Activation profiles generated by hip flexors and knee extensors (TFL, S, RF and VL, on the left). The black dotted line divides the foot on step stage from the pull-up stage. On the right the two activation profiles chosen.

During the first phase TFL, S and RF are activated. The problem with S and RF, which were chosen to perform hip flexion in both straight leg raise and standing hip flexion, is that they exhibit their main activity during the second part. So although they would allow obtaining the flexion in the first part, when the subject has to climb on the step, instead of receiving the contraction only on the knee extensors, he is also electrostimulated on the hip flexors and the pull-up phase does not occur correctly. TFL, on the other hand, is the best choice because it has the greatest activation during the first phase and the lowest one in the second part and therefore does not prevent the extension of the knee. As for the choice of the second activation profile, the VL at 80% of the line anterior superior iliac spine - proximal border of the patella has the advantage of remaining completely inactive throughout the first part and it activates only as soon as the pull-up phase begins. In addition, immediately after the VL at 40% of the same distance, it is the muscle that has the greatest activation.

4.11.2 Electrodes Placement

Since the movement is composed of hip flexion and knee extension, the electrodes have been positioned as indicated in the sections of the relevant exercises. Then three stimulation channels were used in total: two for hip flexion and one for knee extension.

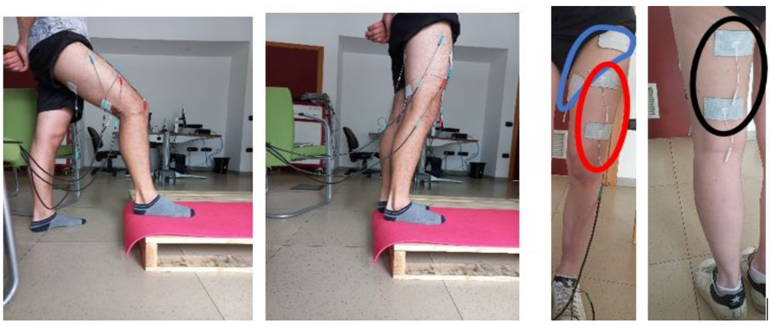

Figure 4.22: Electrodes placement for the step-up, which involves three stimulation channels. Stimulation channels 2 and 3 were used for hip flexion. For Ch2 (in blue): proximal electrode at 1/6 of the line anterior superior iliac spine - lateral femoral condyle (TFL and S) and distal one on S at 100 mm from the point of origin of Sartorius (70mm from the anterior superior iliac spine). For Ch3 (in black): proximal at electrode at 30% of the line ischial tuberosity -medial epicondyle of the tibia (SemT and BF) and distal one at 70% of the same distance (SemT and BF). For stimulation channels 4 (Ch4 , in red) electrodes are positioned as for knee extension.

Chapter 5

Experimental Tests

Experimental tests were conducted to compare any differences between intensity current (I) and pulse width (PW) modulation. The movements of each protocol were performed using both modes of stimulation and comparisons were made in terms of movement execution, discomfort perceived by the patient and muscle fatigue. A total of 6 male subjects aged between 23 and 27 years were recruited, divided equally between the two rehabilitation protocols. Considering that each protocol is composed of 11 movements and that each movement is repeated twice (once for each modulation), each subject underwent 22 stimulation sessions performed on 11 different days: the sessions that involved the execution of the same movement were done on the same day to make the comparison between the two modulations easier for the patient. In each session, the ATC-FES system was used in the therapist-patient mode, in which the first subject (therapist) generates the acquisition profiles and the consequent stimulation pattern is provided to the second subject (patient). The tests were conducted following the rules of the Bioethics Committee of the University of Turin [98].

5.1 Subjects Preparation

At the beginning of each test, acquisition and stimulation electrodes were applied to the therapist and patient respectively, following the instructions given for each movement in Chapter 4. To ensure that the electrode adheres well to the skin, both the therapist and the patient must undergo a specific skin preparation [87]: the area of application of the electrodes first is shaved and then cleaned with alcohol. The acquisition electrodes, shown in Figure 5.1a, applied to the therapist are the Kendall™ H124SG model electrodes produced by Covidien [99]. They are single-use latex-free foam electrodes with a pre-gelled adhesive on the back with non-irritating gel and an Ag/AgCl chloride coating. While the electrodes used for the patient are surface electrodes of two different sizes depending on the movement performed. The smallest electrodes, in Figure 5.1b, are the PG470W model electrodes by Fiab [100], with size 3.5x4.5 cm and were applied only for movements involving gastrocnemius and tibialis anterior (ankle flexion-extension and heel raise movements). For all the other exercises, larger electrodes were chosen, in particular those by HASOMED® [101] of size 5x9 cm, in Figure 5.1c. Both types of stimulation electrodes are reusable, so during the applications, the gel they are equipped with tends to wear out and then it becomes necessary to add more conductive gel in order to ensure good contact between electrodes and the skin.

(a) (b) (c)

Figure 5.1: Acquisition electrodes (a) and stimulation electrodes of two different sizes: the smaller ones (b) applied on smaller muscles such as the gastrocnemius and the tibialis anterior and the larger ones (c) used on all other muscles (quadriceps femoris, tensor fasciae latae, hamstring muscles).

5.2 System Calibration

Once all the electrodes had been placed on both the therapist and the patient, the calibration of the ATC-FES system has to be. In the first phase of calibration, the therapist has to stay still in the resting position in order to define the threshold for the application of the ATC technique. Once the threshold has been defined, in order to obtain the activation profiles and the maximum ATC values that the therapist can generate, s/he repeats at least four times the movement to evaluate. At the end of this stage, the calibration phase of the patient begins, which, if the modulation used is that in current, consists only of the calibration of the maximum current tolerated by the patient, if instead the modulation in pulse width has been selected, then the minimum pulse width necessary to generate a visible contraction of the muscles (motor threshold) must also be determined. In both cases, before evaluating the maximum current, it is necessary to manually enter the values of the remaining stimulation parameters, which will then remain fixed. Frequency and interval interpulse are set to predetermined values [102], while the pulse width is set at 500 µs because:

- in the case of PW modulation, as explained in section 3.2, this value represents the maximum limit of pulse width variation and is therefore used to determine the maximum current tolerated by the patient in the worst stimulation conditions (PW reaches the maximum);
- in the case of I modulation, considering that the muscles to be contracted are very large (e.g.: quadriceps, biceps femoris, gluteus maximus), it was found that lower pulse width values often did not allow to obtain a complete or correct execution of the movement.

Once the values have been entered from the GUI, the current calibration process begins, which involves the gradual increase in intensity until a current tolerated by the patient, which allows the performance of the movement, is reached. If during the training phase, the current value thus found should cause pain or be too weak, then it can be changed manually. As mentioned above, if the modulation chosen is the current one, the calibration phase ends here, instead, if the modulation is the one in pulse width, then the minimum pulse width capable of causing a visible contraction in the patient's muscles is defined.

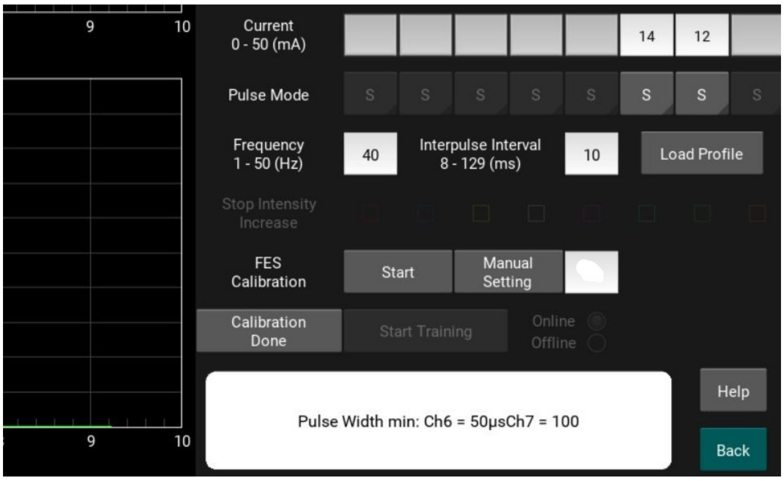

Figure 5.2: Minimum pulse width calibration. The frequency and interpulse interval values used for the previous calibration phase of the maximum current are left. While in the boxes of the selected stimulation channels the maximum current values found at the previous current calibration step are entered.

The same frequency and interval interpulse values, used in the current calibration stage, are maintained and the maximum current values of each stimulation channel, found in the previous step, are entered, as shown in Figure 5.2. At this point, the pulse width is progressively increased with steps of 50 µs until a visible contraction of the muscles is observed. At the end of this phase, the system is calibrated and the stimulation session can begin.

5.3 Session Organization

Before starting each session, the type of modulation to use (I or PW) is randomly chosen. Each session consists of a series of steps described below. After the calibration phase of the ATC-FES system, the patient undergoes a first warm-up phase, in which the frequency is set to 35 Hz and the movement is repeated every 10 s 5 times. The purpose of the warm-up is to warm up the patient's muscles and accustom him to the type of stimulation, considering that the two modulations imply different perceptions of stimulation. Being, therefore, an adaptation phase, the frequency of the warm-up was kept slightly lower than that used in the training to avoid providing too many impulses to the patient's muscles not yet warmed up and causing excessive fatigue. Frequencies below 35 Hz were initially considered for the warm-up phase, but they elicit a tapping effect where individual pulses can be distinguished and so they were excluded. 35 Hz instead is a value that allows fewer pulses and therefore less charge delivery (defined as IxPW generated with each pulse) to the patient compared to the training phase, but that does not produce annoying sensations. Then follows a 10 min of resting period, which is considered a reasonable time for an almost total recovery from accumulated muscle fatigue according to several articles [103][104][105]. After that, the subject enters the training phase, in which the frequency is increased to 40 Hz, according to the indications of HASOMED [102], and the movement is repeated 10 times with pauses of 10 s between one repetition and another. At the end of this stimulation session, a 20 min break is taken, as suggested by [106], to leave the subject all the time necessary to recover the muscles, and then the entire session (warm-up + training) is repeated using the other type of modulation.
At the end of each stimulation session, the subjects were asked to fill out a form containing questions related to comfort/discomfort and any muscle fatigue perceived both during the warm-up and training phases. This form, which will be explained in more detail in chapter 6, has proved to be a very useful tool to better highlight any differences perceived by patients, at the level of stimulation, between the two modulations.

Chapter 6

Results and Discussion

Since the two protocols have many movements in common, which were then performed by the subjects of both groups, the data obtained from the experiments are reported divided by movements and for each movement, the results are presented as explained below.

First, a brief analysis of the execution of the movement and the perception of stimulation of the subject is conducted in the two cases of modulations of pulse width (PW) and current intensity (I). Next, a study of the charge delivery to the patient for each of the 10 repeated movements during the training and the possible onset of muscle fatigue was performed. The charge delivery for an individual pulse is measured as the product between the current and the pulse width of the pulse itself (IxPW). So considering that during the execution of a movement lasting 5 s about 200 pulses are sent to the patient (training stimulation frequency of 40 Hz) and that the charge for each pulse is different from the one of the other impulses because it depends on the current and pulse width values, the total charge delivery for each movement is obtained by adding all the charges associated with every single pulse. In this way, the trend of the total energy value supplied to the patient during the 10 repetitions of the training is obtained. The onset of muscle fatigue perceived during the training was instead obtained through the answers returned by the patients in the form at the end of each session. The form submitted to the subjects was the same for each movement, test, and modulation, so to identify each session a specific code was used for each subject, movement and type of modulation. If the patient felt fatigued, he was asked to indicate the repetition from which this condition arose and which muscles were involved.
Subsequently, a more general picture of the energy supplied to the patient during the training is presented, reporting, for each subject, the distributions between the 10 charge values explained above. In this way, the median value of the charge delivery to the subject in the two cases of modulation is highlighted, which is compared with the index of muscle fatigue perceived by the subject for that particular type

of modulation. The index of muscle fatigue is another parameter obtained from the form, in which patients were asked to judge the state of the stimulated muscles at the end of the session. Subjects had to respond using a scale of 1 to 5, where 1 indicates muscles that are not at all tired and 5 excessively tired muscles.
Finally, the preference of the subjects between the two modulations is reported based on the discomfort perceived. These data were also obtained through the form, where the subjects were asked to indicate on a scale from 1 (pain) to 10 (comfort) the degree of comfort/discomfort experienced.

Since the warm-up phase aims to accustom the patient to the type of stimulation and warming up the muscle, only the data obtained from the training have been considered to make comparisons between the two modulations.

6.1 Ankle Flexion - Extension

The subject moves the ankle down (extension) and up (flexion) while lying supine. Since this movement is part of both rehabilitation protocols, it was performed by all six subjects. The movement was performed correctly with both types of modulation, but the range of motion always remained quite limited due to the resting position: the ankle is blocked by the bed and therefore the movements are less wide than they would be if the patient were sitting with his legs outside the bed without any impediment between the foot and the floor.

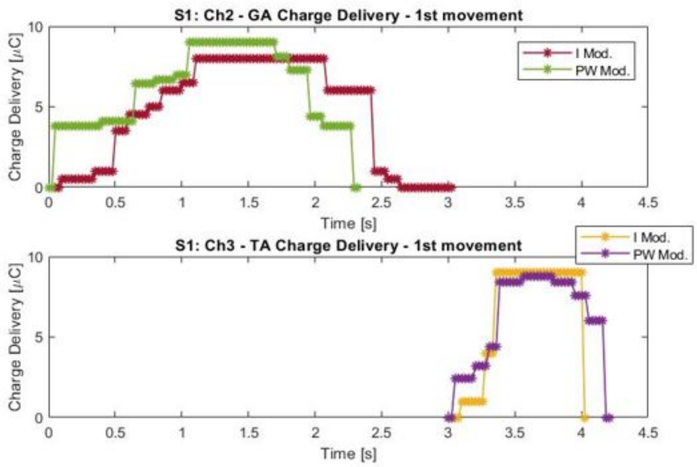

Figure 6.1: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = Gastrocnemius (GA); Ch3 = Tibilias Anterior (TA). First the ankle extension is performed (Ch2 active first) and then the flexion (Ch3 is the second to activate).

All subjects agreed to define PW modulation as more intense and strong from the first seconds in which the stimulation begins. Figures 6.1 gives an example of what happens at each repetition and represent the charge values (IxPW of the pulse as

previously explained) provided to the patient at each pulse of the movement. The graph shows the charge trends, during the first movement, for both stimulation channels used: stimulation channel 2 (Ch2) on the gastrocnemius (GA) and channel 3 (Ch3) on the tibialis anterior (TA). For both channels, as soon as the movement begins, the charge delivery to the patient is much higher in the case of modulation in PW. This translates into a stronger and more intense stimulation from the first moments, while the modulation in I, as described by the subjects, is lighter and therefore less annoying, especially in the gastrocnemius area.
I modulation did not cause any kind of fatigue for any subject. While in the case of PW modulation, only 3 subjects (S1, S2, S3) perceived fatigue limited to the gastrocnemius area during the last 3 movements. Looking at Figure 6.2, which shows the trend of the total charge delivery for each of the 10 movements (measured as explained above), there is an actual increase in the charge on Ch2 (gastrocnemius) for subjects S1, S2, and S3 in the last 2-3 repetitions (dotted black line represents when muscle fatigue occurs).

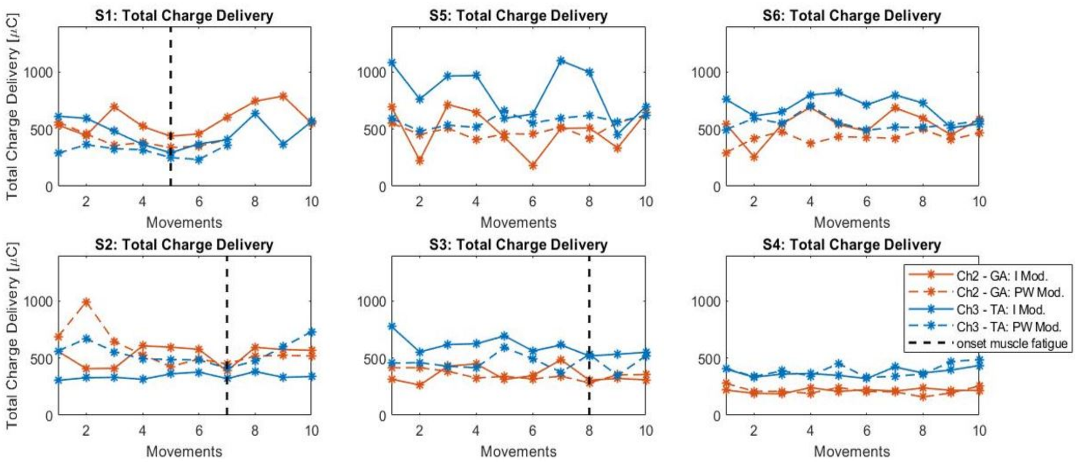

Figure 6.2: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

Figure 6.3 shows, for each subject and stimulation channel, the distributions of the 10 values of charge delivery represented in Figure 6.2. Looking at the distributions of the 3 subjects who felt muscle fatigue on the gastrocnemius during PW modulation (S1, S2 and S3), the median value (numerical values reported in Table 6.1) is greater in the case of the I modulation for subjects S1 and S2, while S3 has a slightly higher median value for PW modulation. In addition, the maximum value reached, in the I mode, is much greater for the S1 and S3 subjects than for the PW mode.

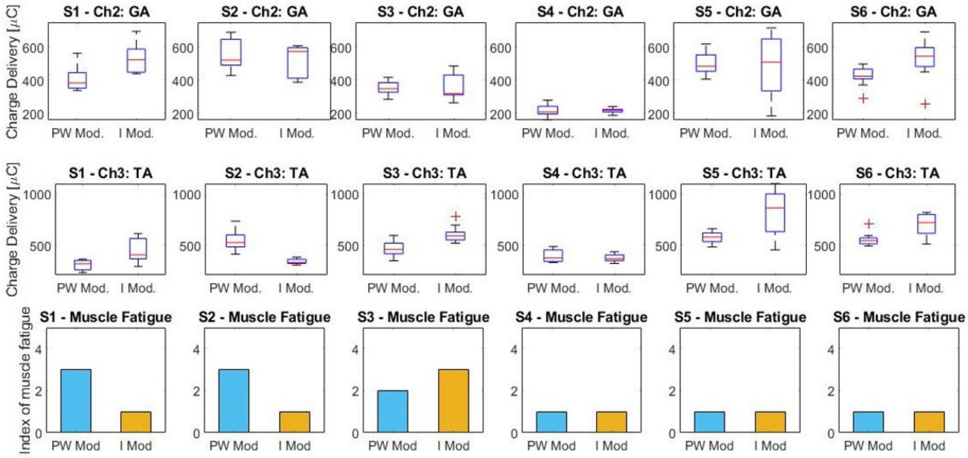

Figure 6.3: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (Gastrocnemius). Second row: distribution of charge delivery to channel 3 (Tibialis Anterior). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Median value of charge delivery distribution [μC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	383.6	524.0	484.4	509.0	424.1	545.5
Ch3	318.8	407.0	579.3	864.5	544.7	721.5
	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	521.9	573.0	349.0	320.3	208.4	218.0
Ch3	525.3	333.0	459.2	590.0	375.5	368.5

Table 6.1: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2= Gastrocnemius; Ch3= Tibialis Anterior.

Finally, the last line of Figure 6.3, shows the indices of muscle fatigue at the end of the training. Subjects S1 and S2 have perceived greater fatigue for the PW modulation, even though the median charge delivery is lower than that of the I modulation. The subject S3 presents exactly opposite results, a superior charge in PW modulation, but more fatigue perceived in the other mode. While subjects S4, S5, and S6 did not experience any fatigue. Therefore, it is not possible to associate the modulation considered more tiring with a higher average charge value delivery to the patient. Figure 6.4 show the comfort indices for the two modulations. The modulation in PW was evaluated by 4 subjects as less comfortable and, of these, 2 subjects (S1 and S2) expressed an acute discomfort in the upper area of the gastrocnemius (near the position of the proximal electrode), due to which, the subject S1 had to interrupt the session (execution of 7 movements and not 10).

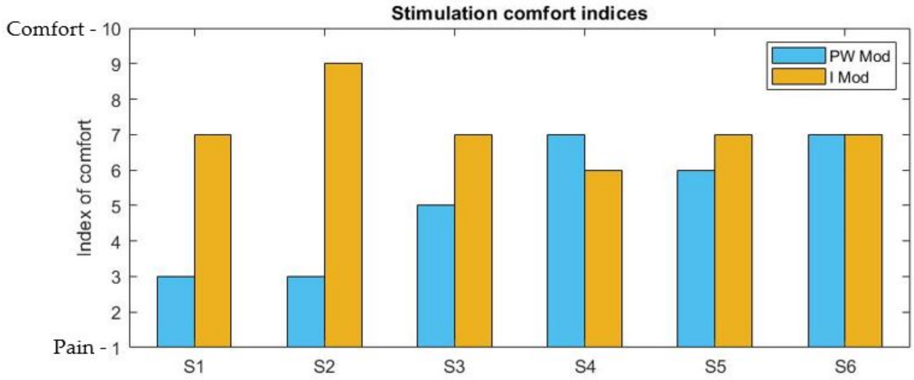

Figure 6.4: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.2 Knee Flexion Supine

The subject is lying supine and flexes the knee bringing it towards the chest while dragging the heel on the bed. Then he returns to the resting position.

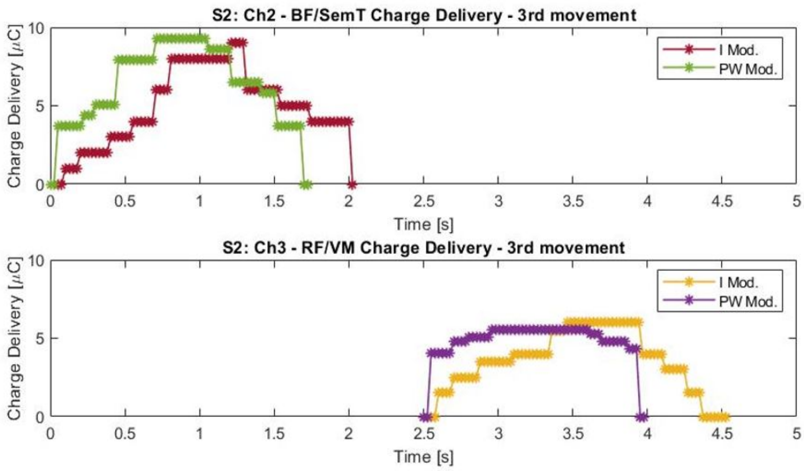

Figure 6.5: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = biceps femoris and semitendinosus (BF/SemT); Ch3 = rectus femoris and vastus medialis (RF/VM). First the knee flexion is performed (Ch2 active first) and then the extension to return to the resting position (Ch3 is the second to activate).

Since this movement is included in both rehabilitation protocols, it was performed by all six subjects. The movement was performed correctly with both types of modulation. However, the modulation in PW allowed achieving the complete flexion of the knee without any help from the voluntary contraction of the subject, while in the I modulation the angle of flexion of the knee remained slightly lower and only thanks to the muscular strength impressed by the subject a complete flexion was achieved. Despite this, all subjects agreed to define PW modulation as more intense and strong from the first seconds in which stimulation begins. The Figure 6.5, as explained for the previous movement in section 6.1, shows the course of the charge, during the third movement, for both stimulation channels used: stimulation channel 2 (Ch2) on the biceps femoris and semitendinosus (BF/SemT) and channel 3 (Ch3) on the rectus femoris and vastus medialis (RF/VM). For both channels, as soon as the movement begins, the delivery of the charge to the patient is much higher in the case of modulation in PW. This leads to a stronger and more intense stimulation from the first moments, while the modulation in I, as described by the subjects, is lighter and therefore less annoying, especially in the RF area. Moreover, since stimulation with PW modulation is so intense, it can be sustained for just over 1 s while the I modulation allows, both in the bending phase and in returning to the resting condition, a longer stimulation without creating too much discomfort to the patient. However, these considerations, for this specific exercise, are not particularly relevant considering that, both the knee flexion phase and the extension one, are easily performed in about 1 s, which is a sustainable stimulation time with both types of modulation. I modulation did not cause any kind of fatigue for any subject. While in the case of PW modulation, only 2 subjects (S1, and S6) began to perceive fatigue, limited to the RF area, starting from the 7th movement onwards. Looking at Figure 6.6, it is possible to see that the charge remains constant at a high value, on both stimulation channels, for the subject S1 and that instead there is an increase in the parameter, on both channels, for the subject S6.

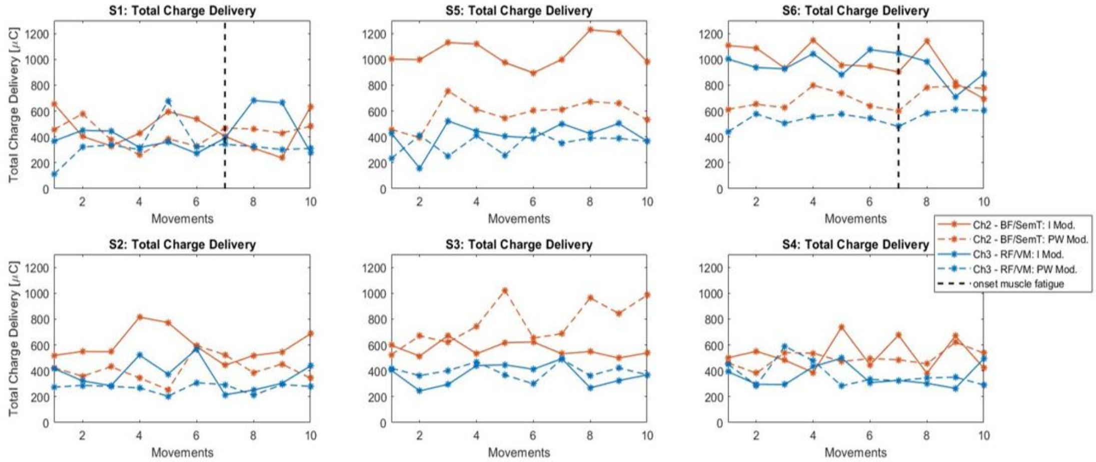

Figure 6.6: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

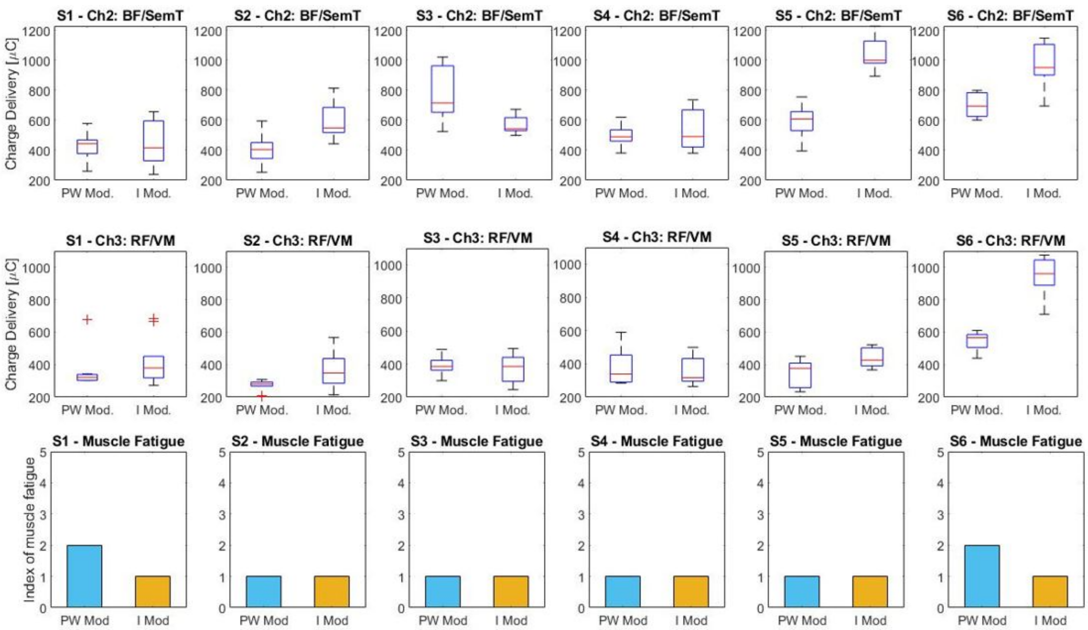

Figure 6.7: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (BF/SemT). Second row: distribution of charge delivery to channel 3 (RF/VM). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Even in this case, however, it cannot be said that there is a real relationship between a higher median charge value and a greater degree of fatigue. Looking at Figure 6.7 and Table 6.2, the two subjects who experienced muscle fatigue on RF during training in PW modulation (S1 and S6), show median values of distribution on channel 3 (RF) higher in the case of I Modulation. However, looking at the fatigue indices, both subjects overall declared higher muscle fatigue in the PW case. Any outliers, such as those of the distribution of channel 3 of S1, are due to the fact that the value of energy supplied to the patient at each repetition depends on the muscle activation of the therapist: less activation means less charge and vice versa. Therefore, considering that the therapist tends to activate his muscles with lesser or greater strength to obtain respectively less discomfort or greater contraction force to perform better the movement, some repetitions may be characterized by muscle activations, and therefore charge values, outside the normal range of variation of the parameter.

Median value of charge delivery distribution [µC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	443.9	416.8	609.0	1001	695.2	951.8
Ch3	322.2	379.8	377.40	428	566.4	960.5
	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	404.9	549.0	715.5	543.8	490.0	491.5
Ch3	280.0	348.3	385.9	386	339.2	317.5

Table 6.2: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2= BF/SemT; Ch3= RF/VM.

Figure 6.8 show patients' preferences on the type of modulation to use for supine knee flexion. 4 out of 6 subjects considered PW modulation as less comfortable because, despite allowing a complete knee flexion, it is a too strong stimulation to execute this type of movement, in which the limb does not have to be completely raised (all the movement of flexion of the leg and return to the initial position takes place always sliding the heel on the bed) and so even a small contraction is sufficient to perform it.

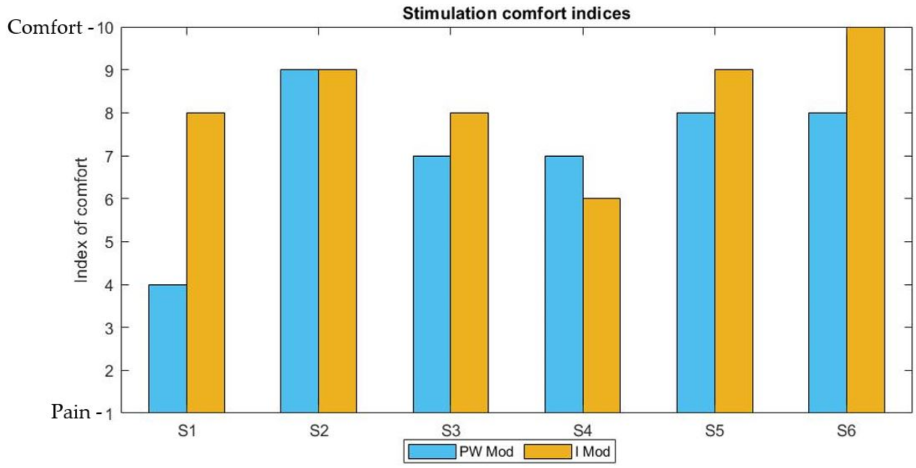

Figure 6.8: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.3 Knee Flexion Prone

The subject is lying prone and flexes the knee bringing the heel towards the buttock, maintains this position for a few seconds, and then returns the leg to the initial position. Since this movement is included only in knee rehabilitation protocols, it was performed only by 3 subjects (S2, S3, and S4). Both the execution of the flexion and the maintenance of the position are better performed with PW modulation.

Figure 6.9: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = biceps femoris and semitendinosus (BF/SemT). Knee flexion is performed, this position is maintained for a while and then the leg is returned to the resting position.

The current modulation is sometimes too weak and causes the knee flexion movement to be barely hinted at. PW modulation, on the other hand, produces a greater force of contraction, allows not only to reach greater flexion angles but also to maintain the position of knee flexed without the help of the patient's voluntary contraction. This behavior can be observed from the charge provided at each impulse in Figure 6.9, which shows the example relating to the second movement of the subject S3. As always, the charge provided in the case of PW mode is immediately higher and therefore translates into a more effective stimulation, in terms of movement performance, from the first moments. On the contrary, the I modulation remains initially lower and therefore does not allow the execution of the movement immediately.

No fatigue was felt during the 10 repetitions of the training, as can be seen from the graphs of the charge delivery at each movement in Figure 6.10.

Figure 6.10: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Subjects of knee rehabilitation protocol S2, S3 and S4. The dotted black line represents the movement in which muscle fatigue arises.

Only at the end of the tests, the subject S4 felt slight fatigue in the semitendinosus area, and in particular, the perceived fatigue in the case of PW was greater than that of I mode. Looking at the charge distributions and their median values (Figure 6.11 and Table 6.3), the modulation considered more tiring is not associated with the higher median charge value.

Median value of charge delivery distribution [µC]

	S2		S3		S4	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	545.6	768.0	341.0	582.0	480.3	526.5

Table 6.3: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2= BF/SemT.

The subject S3 instead reported at the end of both sessions a fatigue index equal to 3, specifying however that this was not caused by stimulation, but by a physical activity carried out during the day, so his results of muscle fatigue were not considered. Moreover, in the S3 charge distribution, there are outliers for both modulations always caused by the fact that the charge values for each movement depend on the muscular activation of the therapist, which changes depending on whether the patient is perceiving too much discomfort or if more contraction force is needed to perform correctly the movement.

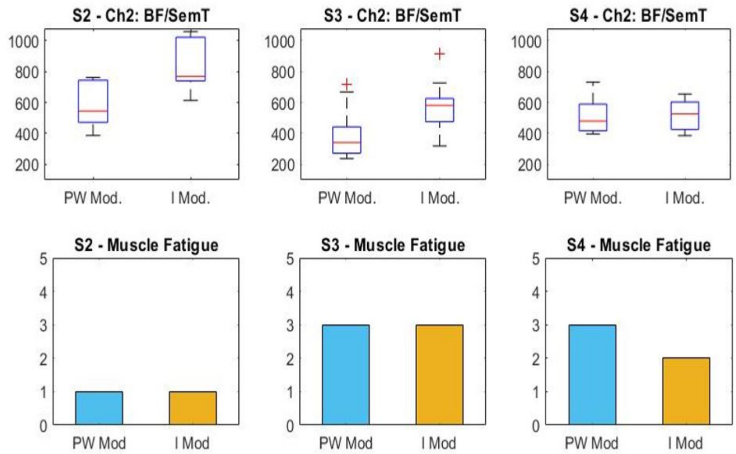

Figure 6.11: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (BF/SemT). Second row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

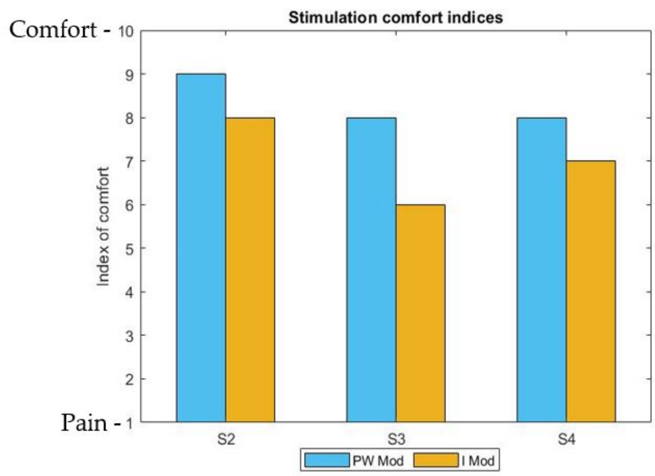

Figure 6.12: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

As for the discomfort experienced during the stimulation phase, all the subjects expressed a preference for modulation in PW because this stimulation was more stable and constant and better guided the limb in the execution of the movement. While I modulation, not only was perceived as more discontinuous but was often too weak and therefore forced the patient to compensate largely with their muscle activity to perform the complete movement, resulting in a higher level of discomfort.

6.4 Knee Extension Supine

The subject, lying on his back and with a foam roller below the knee, raises the heel to complete the extension of the knee. It holds the position for a few seconds and returns to the starting position. This movement was performed by all six subjects. During the knee extension phase, both modulations allow performing the movement. The differences are only related to the maintenance phase, which has a maximum duration of 3 seconds, after this period a cramp can occur. I modulation has been described as less intense and less stable and, therefore, ineffective in keeping the leg in extension. To stay in this position for a few seconds, the subjects had to use their muscle contraction. In contrast, PW modulation stimulation allowed the subjects to maintain knee extension for a few seconds without them having to compensate with a voluntary contraction. This may be due to the fact that in PW modulation the current is always constant at its maximum value, from when the stimulation begins, as reported in terms of charge delivery per pulse in Figure 6.13.

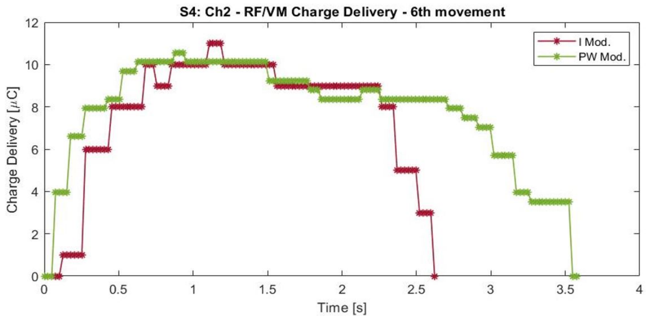

Figure 6.13: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = rectus femoris and vastus medialis (RF/VM). Knee extension is performed, this position is maintained for a while and then the leg is returned to the resting position.

This allows not only to reach the complete extension from pulse width values just above the lower limit but also to support the final position even if the pulse width value decreases slightly. In I modulation, on the other hand, the current starts from 0 and rises until it reaches its maximum value, and usually, the complete extension is reached only for current values slightly lower than this maximum (2-3 mA below maximum). That is why it is more difficult to keep the knee in extension only with stimulation in this case. Small variations in current from its maximum value tend

to return the leg to the resting position.
Regarding muscle fatigue, 3 subjects (S3, S4, S5) claimed to have felt, during PW modulation, fatigue in the area of the rectus femoris and vastus medialis, where the distal electrode is positioned, starting from the seventh movement. As shown in the Figure 6.14, there is an increase in the charge delivery for subjects S3 and S4, while for S5, from the seventh repetition, the charge remains constant at a high value.

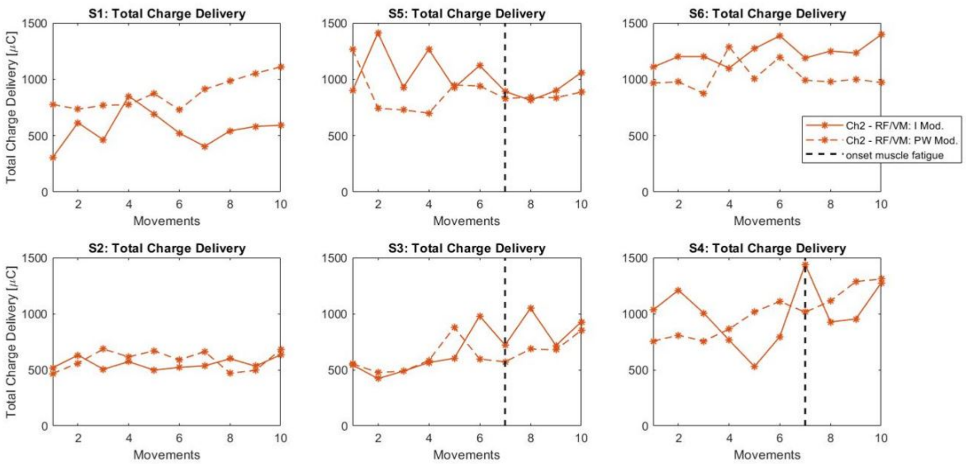

Figure 6.14: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

Looking at the charge distributions, Figure 6.15 and Table 6.4, for subjects who have perceived greater fatigue in PW mode (S3, S4, S5), only for the S4 subject there is a correspondence between an increase in the charge and the muscle fatigue index.

Median value of charge delivery distribution [μC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	825.9	562.5	838.3	927.0	985.0	1218.0
	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	602.3	534.3	589.9	659.8	1014.9	979.0

Table 6.4: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2= RF/VM.

For both S3 and S5, which expressed an equal index of fatigue for the two cases, the median charge delivery for the modulation in I is slightly higher. Also in this case, therefore, it is not possible to associate the modulation judged more tiring, with a greater overall charge. Finally, in the S1, S5, and S6 charge distributions, there are outliers for both modulations caused, as explained before, by the fact that the charge values for each movement depend on the muscular activation of the therapist, which changes depending on whether the patient is perceiving too much discomfort or if more contraction force is needed to perform correctly the movement.

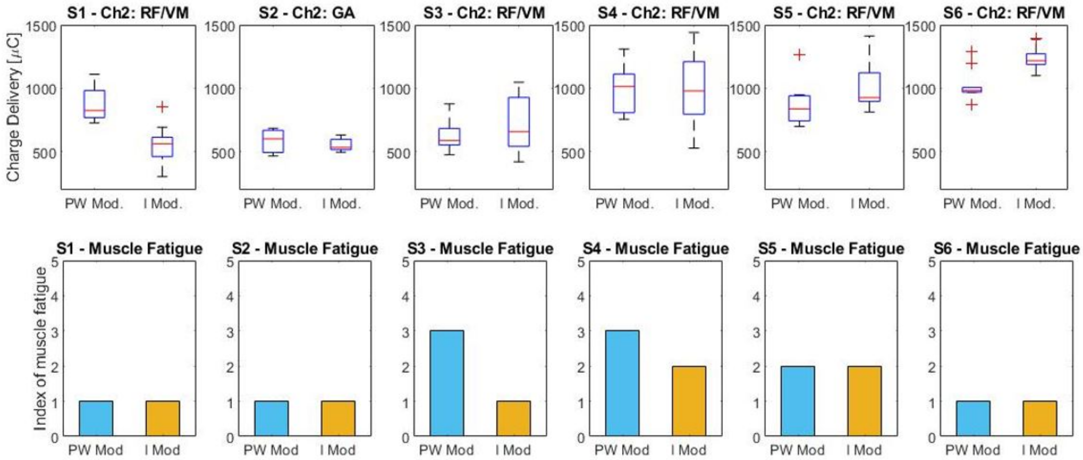

Figure 6.15: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (RF/VM). Second row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

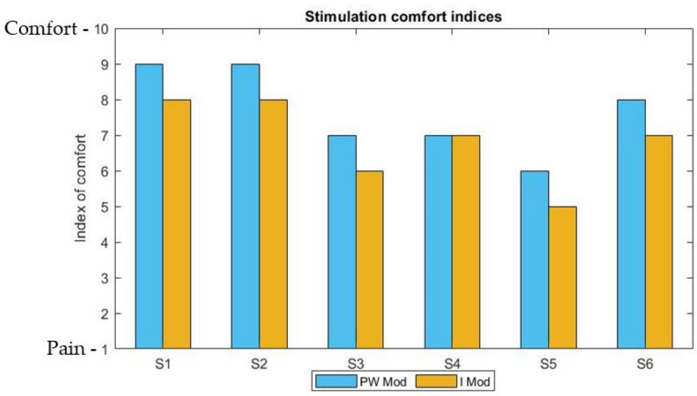

Figure 6.16: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

Despite the slight increase in fatigue, 5 subjects expressed a preference for the PW modulation, as shown in Figure 6.16. In this case, the stimulation is continuous and more stable, especially in the maintenance phase, thus translating into a more comfortable stimulation, that helps the patient more to understand what is the movement that he has to perform. While the discontinuity of current modulation turns out to be less effective and more uncomfortable.

6.5 Hip Abduction Supine

The subject is lying on his back, opens his leg outwards, and then returns the limb to the initial position. So in this case there is a first stimulation to perform the abduction of the leg and a second stimulation to bring it back to the resting position. Since this movement is included only in hip rehabilitation protocols, it was performed only by 3 subjects (S1, S5, and S6). The I modulation has been evaluated as too weak and shaky, so the resulting movement remains very limited (opening angle of about 10°-20°) and the return phase is often not performed therefore the patient must return the limb to the initial position with his muscle contraction. On the contrary, the modulation in PW allows to obtain a very decisive and wide movement (angles of 30°-40°) and to return each time leg to the initial position without the compensation of the patient's muscle strength. This behavior reflects on the amount of charge delivery to the patient at each pulse during the single repetition of the movement. In Figure 6.17, the graph of the second movement performed by the subject S6 is shown: since the modulation in PW is more intense, the abduction movement is performed within 2 s (first stimulation), while the I modulation is weaker, especially in the first moments, so it is necessary to prolong the stimulation time, about 3 s, to achieve the same movement, with a range of motion that however remains more limited.

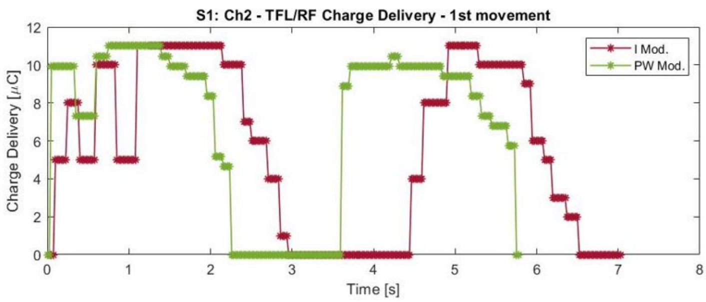

Figure 6.17: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = tensor fasciae latae and rectus femoris (TFL/RF). The abduction of the hip (leg open to the outside) is first performed and then the limb is returned to the initial position thanks to the second stimulation.

Regarding muscle fatigue, only one subject S5 perceived muscle fatigue during the PW modulation starting from the 5th movement. In fact, in the relative graph of the trend of the charge delivery at each repetition, it is evident that the parameter remains about constant at a high value for the entire second half of the training.

Figure 6.18: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Subjects of hip rehabilitation protocol S1, S5 and S6. The dotted black line represents the movement in which muscle fatigue arises.

Looking at the charge distributions, Figure 6.19 and Table 6.5, and the overall fatigue indices obtained at the end of each test, for the subject S5, it is possible to observe that as the median value of the charge is greater for the PW case the same thing applies to the fatigue index. While, the exact opposite occurs for the subject S6, whose charge distribution is shifted to higher values for the current mode and whose muscle fatigue is found only in PW mode.

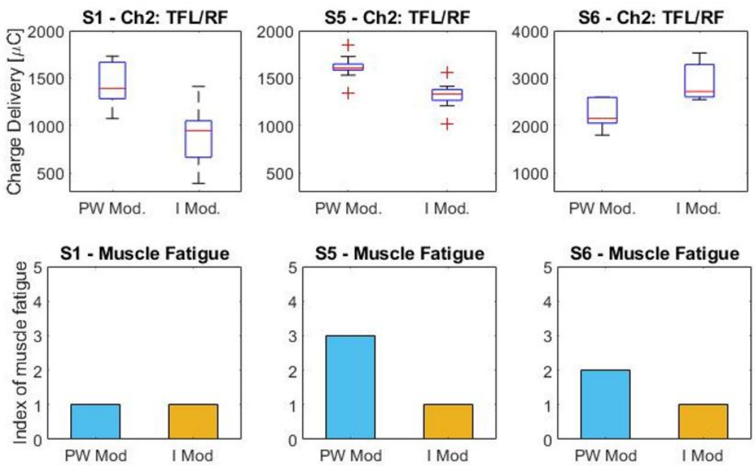

Figure 6.19: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (TFL/RF). Second row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Median value of charge delivery distribution [µC]						
	S1		S5		S6	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	1392.0	947.3	1606.0	1332.0	2154.7	2720.0

Table 6.5: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects.Ch2= TFL/RF.

Despite the increased fatigue experienced with modulation in PW, all subjects preferred this type of stimulation for the execution of the supine hip abduction. Even though the presence of fluctuations of the ATC value in the activation profile, the stimulation is perceived as continuous, stable, and intense thanks to the fact that the current is kept fixed at its maximum value and therefore small variations in the pulse width parameter are not perceived by the patient. Consequently, PW modulation leads the limb in the execution of the whole exercise without creating any type of discomfort. Whereas with I modulation, variations in the activation profile result in discontinuous and less effective stimulation, which becomes annoying for the patient.

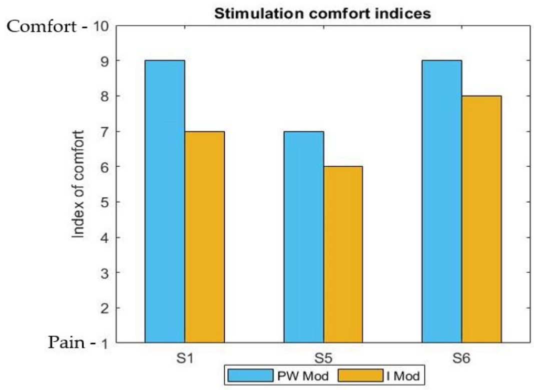

Figure 6.20: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.6 Hip Extension Prone

The subject is lying prone, lifts the leg, and holds the position for a few seconds. Then the limb is returned to its initial position. Since this movement is included only in hip rehabilitation protocols, it was performed only by 3 subjects (S1, S5, and S6). Both achieving and maintaining hip extension were best performed with PW modulation. The latter was perceived as more intense and strong stimulation, able to bring the leg to a height of about 20-30 cm and to keep it in this position without the help of the subject's voluntary contraction. I modulation was felt, instead, as weaker and lighter, especially in the gluteus area, and did not always result in sufficient stimulation to lift the leg. In cases where the movement was executed, the range of motion was more limited, with a height reached by the leg not exceeding 10 cm, and in the maintenance phase, the patient was forced to add his own muscular contribution to the stimulation. The Figure 6.21 shows the usual graph of charge delivery values at each pulse related to, in this case, the first repetition of the movement of the subject S6. Hip extension occurs initially thanks to stimulation on RF and VM (channel 3) and then stimulation on the gluteus medius and maximus (GMed/GMax, channel 2) is added. Since the modulation in PW is more intense, the extension of the limb is carried out within about 0.5 s and for the remaining 2 s the position is maintained. I modulation is weaker, especially in the first moments, so a longer stimulation time is required to achieve the same movement, but with a more limited range of motion. The fact that current modulation is less effective in keeping the leg in extension, as already explained for the supine knee extension, could be due to the fact that in PW modulation the current is always constant at its maximum value from when the stimulation begins and this allows not only to reach the complete extension from pulse width values just above the lower limit but also to support the final position even if the pulse width value decreases slightly. In I modulation, on the other hand, the current increases slowly and the movement is completed only for current values close to the maximum (2-3 mA below the maximum). So small variations in current result in less effective stimulation.

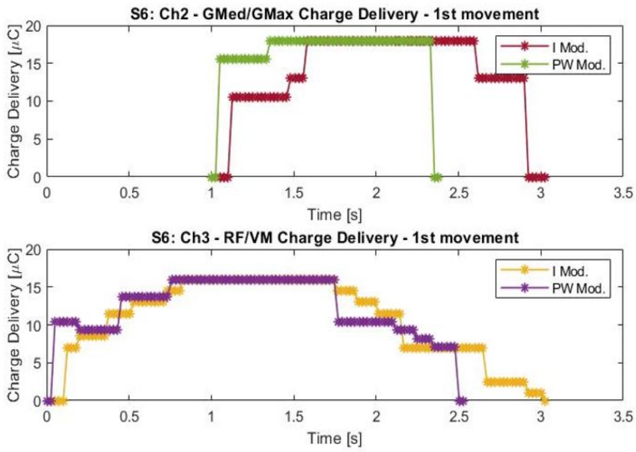

Figure 6.21: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = gluteus medius and maximus (GMed/GMax); Ch3 = rectus femoris and vastus medialis (RF/VM). The extension is started by stimulation on the RF and then the stimulation of channel 2 of the gluteus medius is added.

As regards muscle fatigue perceived during the training, only the S5 subject reported strong fatigue centered in the RF area (near the distal electrode) starting from the 6th movement onwards in both modulations. In particular, the perceived fatigue was greater in the PW case, and in fact, as can be seen in Figure 6.22, during the last 4 charge values related to channel 3 (RF, blue line), there is an increase in the parameter, while in I modulation, the energy simply remains constant at high values.

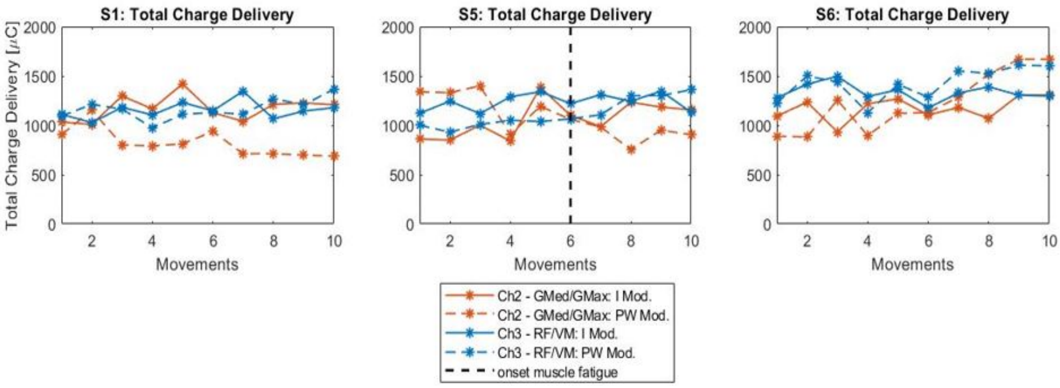

Figure 6.22: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Subjects of hip rehabilitation protocol S1, S5 and S6. The dotted black line represents the movement in which muscle fatigue arises.

Looking at the charge distributions, Figure 6.23 and Table 6.6, and the overall fatigue indices for the subject S5, it is possible to observe that even in this case, the charge distributions associated with higher values are the ones relating to I modulation, while the modulation considered as more tiring is the one in PW.

Median value of charge delivery distribution [μC]

	S1		S5		S6	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	797.1	1187.0	1021.5	1051.0	1192.6	1199.0
Ch3	1148.3	1146.0	1058.5	1244.3	1470.0	1317.0

Table 6.6: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects.Ch2= GMed/GMax; Ch3 = RF/VM.

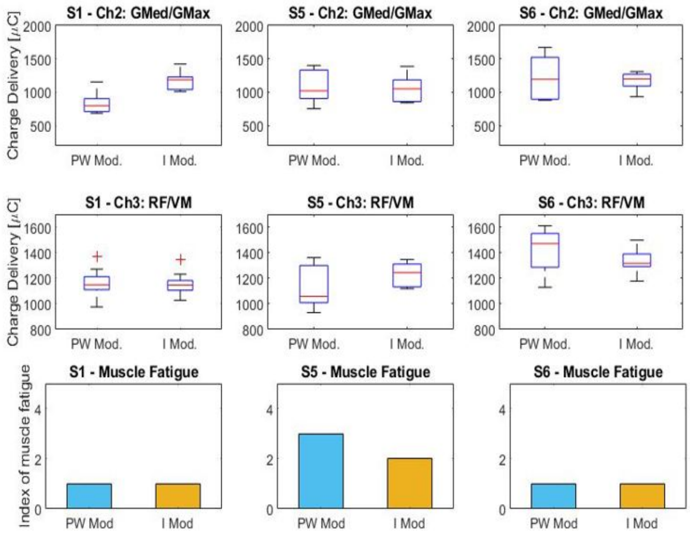

Figure 6.23: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (GMed/GMax). Second row: distribution of charge delivery to channel 3 (RF/VM). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

All 3 subjects preferred modulation in PW. The latter is perceived as a continuous and stable stimulation that, although it is intense and can generate slight fatigue (i.e.: S5), does not generate excessive discomfort in the patient. On the contrary, with I modulation, variations in the activation profile result in a discontinuous and shaky stimulation that does not allow the patient to be guided by the stimulation and therefore generates discomfort.

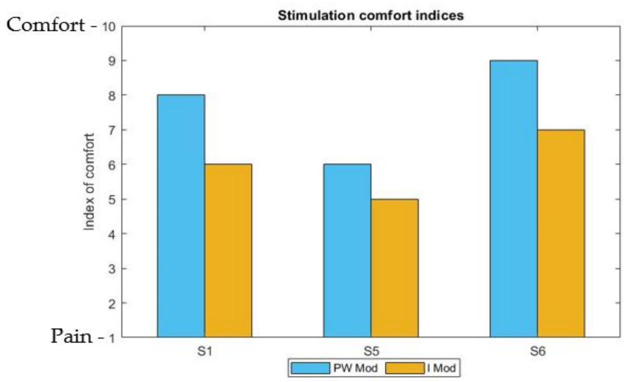

Figure 6.24: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.7 Hip Abduction Lateral

The subject, lying on the side of the healthy limb, raises the operated leg, keeping it extended, and holds it in this position. Then return to the initial position. This movement was performed by all six subjects. The movement performed was very limited for all subjects and all the cases: the maximum height reached by the leg never exceeded 15 cm. As regards the execution of the abduction movement, this was performed with both modulations. The differences are only related to the maintenance phase. I modulation has been described as less intense and less stable and, therefore, ineffective in keeping the leg in abduction without the help of the voluntary contraction of the subject. In contrast, PW modulation was intense enough to keep the leg up for 3-4 s without the subject's intervention. This difference between the two types of stimulation is always linked to the fact that in the PW mode, the current remains constant at its maximum value and this allows the execution of the movement starting from pulse width values even slightly above the lower limit (PW min). Whereas in I modulation, the current starts from 0 and rises until it reaches its maximum value and usually, to raise the leg by at least 10 cm, a current that is slightly lower than the maximum (2-3mA lower) is required. For this reason, even small variations in current from its maximum value prevent the limb from being kept in the final position reached. Figure 6.25 shows the concept just explained through the values of the charge delivery to the subjects for each pulse: PW modulation provides more energy to the patient right away, while modulation takes longer to reach the same charge levels.

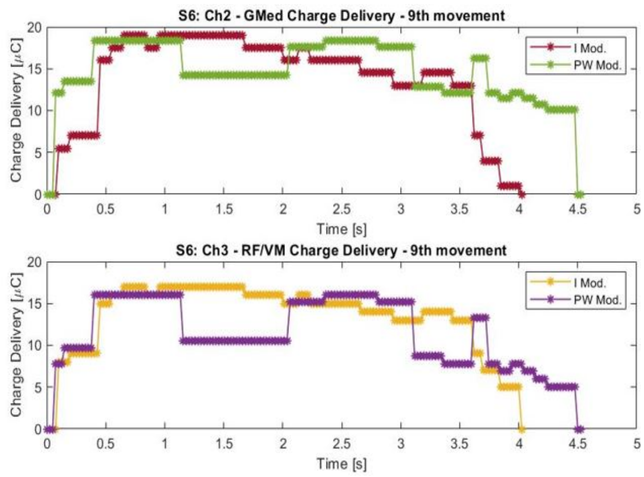

Figure 6.25: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = gluteus medius (GMed); Ch3 = rectus femoris and vastus medialis (RF/VM). The abduction is performed thanks to the contraction provided by both stimulation channels simultaneously.

Subjects S1 and S5 are the only ones who have experienced fatigue during the training and this condition has arisen starting from the 5th and 7th repetition respectively. For both subjects, this fatigue remained limited to the area of the RF muscle and was felt only with PW modulation (channel 3, dotted blue line). The Figure 6.26 shows that for the S1 there has been an increase in the charge parameter provided to the subject from the moment the fatigue has appeared, while for the S5 the parameter remains constant at high values from the 7th movement onwards.

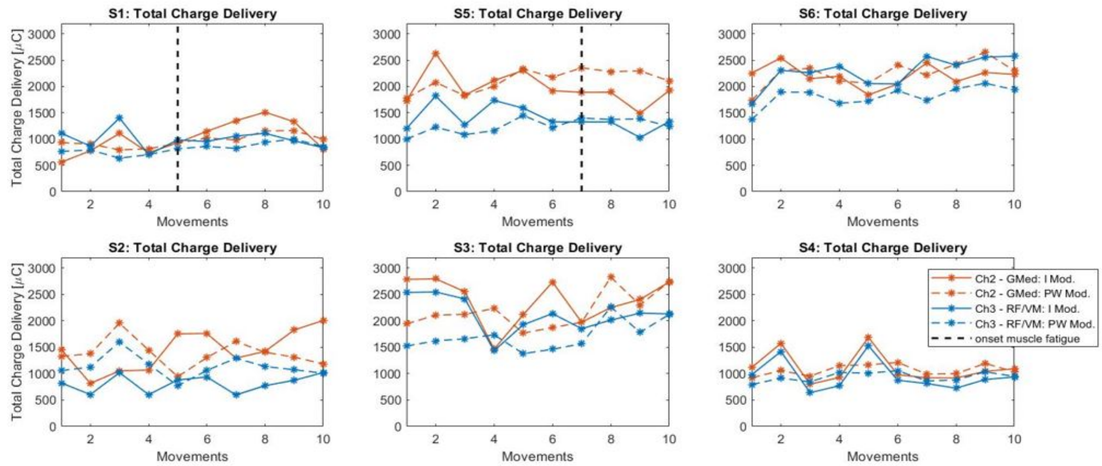

Figure 6.26: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

Both the subjects S1 and S5 have therefore defined the session with PW as more tiring (muscle fatigue indices), however, the median value of the charge distribution is greater in the I modulation case, as shown in Figure 6.27 and Table 6.7. The subject S4 instead felt mild muscle fatigue, always limited to the RF area, at the end of both stimulations (same index of muscle fatigue). The outliers present in the distributions of S2, S3, S4, S5, and S6 are due to those repetitions in which the activation of the therapist was too weak or too strong. If the patient feels discomfort, the therapist tries to decrease the amount of charge that is sent to the patient by contracting his/hers muscles less, while if the stimulation is not strong enough to perform the movement, then the muscle activation of the therapist is increased. Not all charge variations, however, fall within the same range and therefore sometimes result in outliers of the charge delivery.

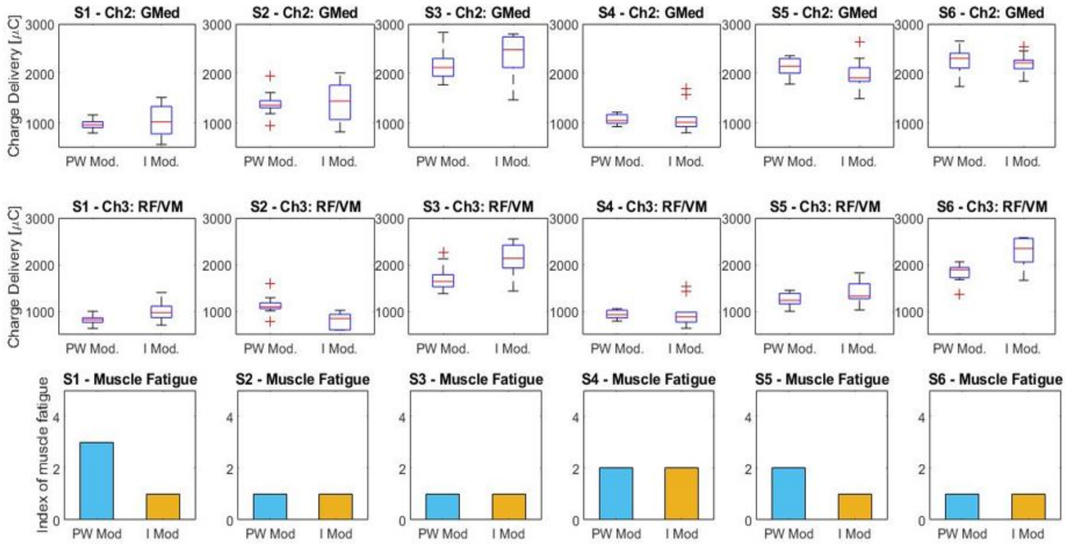

Figure 6.27: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (GMed). Second row: distribution of charge delivery to channel 3 (RF/VM). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Median value of charge delivery distribution [µC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	958.1	1018.5	2140.9	1905.3	2304.2	2212.0
Ch3	818.9	974.3	1235.0	1326.0	1891.6	2346.0
	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	1352.6	1439.0	2116.4	2478.3	1047.4	1012.5
Ch3	1099.8	850.0	1640.8	2135.0	932.8	883.0

Table 6.7: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2 = GMed; Ch3 = RF/VM.

Despite the increased fatigue felt in the case of PW modulation, this was the preferred stimulation of 5 out of 6 patients. The subject S4 did not feel any difference between the two conditions. Even though the presence of fluctuations of the ATC value in the activation profile, the stimulation with PW modulation is perceived as continuous, stable, and intense thanks to the fact that the current is kept fixed at its maximum value and therefore small variations in the pulse width parameter are not perceived by the patient. Consequently, PW modulation leads the limb in the execution of the whole exercise without creating any type of discomfort. Whereas with I modulation, variations in the activation profile result in a discontinuous stimulation that causes discomfort to the patient, who does not understand the movement he has to perform.

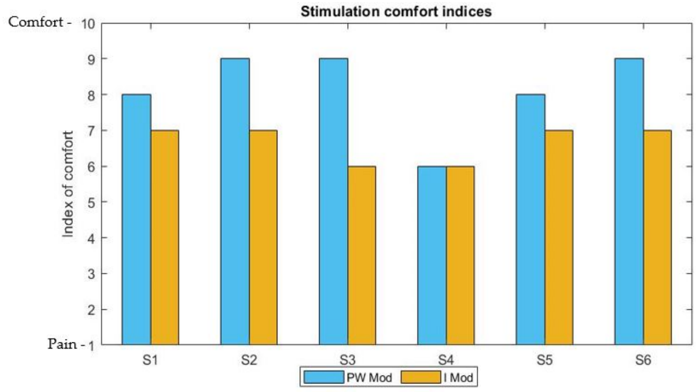

Figure 6.28: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.8 Straight Leg Raise

The patient is lying supine and raises the extended leg. He holds the position and then returns the leg to the initial position. Since this movement is included only in knee rehabilitation protocols, it was performed only by 3 subjects (S2, S3 and S4). The movement performed with the PW modulation allowed to raise the leg more than the I modulation. In addition, I modulation has been described as less intense and less stable and, therefore, ineffective in keeping the leg raised and extended without the help of the voluntary contraction of the subject. In contrast, PW modulation was intense enough to keep the leg up for 3 s, a longer stimulation time sometimes led to cramps, without the subject's intervention. This difference between the two types of stimulation is always linked to the fact that in the PW mode, the current remains constant at its maximum value and this allows the execution of the movement starting from pulse width values even slightly above the lower limit (PW min). Whereas in I modulation, the current starts from 0 and rises until it reaches its maximum value and usually, to raise the leg by at least 10 cm, a current that is slightly lower than the maximum (2- 3mA lower) is required. For this reason, even small variations in current from its maximum value prevent the limb from being kept in the final position reached. Figure 6.29 shows the concept just explained through the values of the charge delivery to the subjects for each pulse relative to the 2° movement performed by subject S2: PW modulation provides more energy to the patient right away, while modulation takes longer to reach the same charge levels.

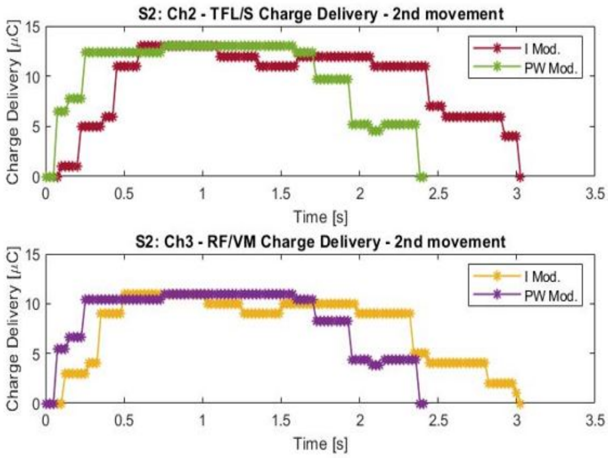

Figure 6.29: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = tensor fascia latae and sartorius (TFL/S); Ch3 = rectus femoris and vastus medialis (RF/VM). The raise of the leg is performed thanks to the contraction provided by both stimulation channels simultaneously.

All 3 subjects experienced increased fatigue during PW modulation training. This fatigue has always been limited only to the RF area (near the distal electrode), line dotted in blue in Figure 6.30, and has occurred starting from the 7th repetitions for the S2 and S4 subjects, where in fact there is an increase in charge delivery during the last 3 movements, and from the 5th repetition for S3, in which instead the parameter remains about constant throughout the second half of the training.

Figure 6.30: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Subjects of knee rehabilitation protocol S2, S3 and S4. The dotted black line represents the movement in which muscle fatigue arises.

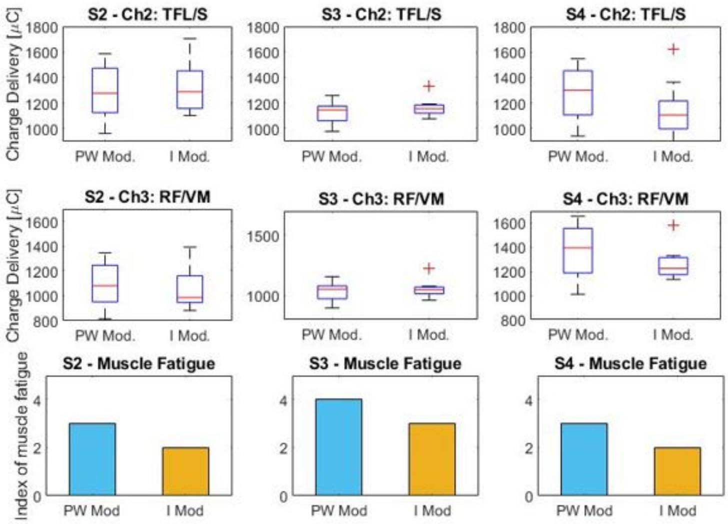

Figure 6.31: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (TFL/S). Second row: distribution of charge delivery to channel 3 (RF/VM). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Median value of charge delivery distribution [μC]

	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	1277.3	1289.5	1147.1	1156.0	1301.8	1106.8
Ch3	1080.8	987.0	1051.9	1046.5	1394.8	1224.0

Table 6.8: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2 = TFL/S; Ch3 = RF/VM.

As reported in the last line of the Figure 6.31, all 3 subjects indicated PW modulation as the most strenuous stimulation. In particular, S2 and S4 assigned the same fatigue indices, while the subject S3 assigned a score greater than one point to both fatigue indices. It should be noted that the greater fatigue perceived by S3 could be influenced by the fact that he has carried out sports (swimming) before undergoing the stimulation session. Looking instead at the first two rows of Figure 6.31 and the Table 6.8, it is clear that the median values of the charge delivery to the patient, in particular relating to the channel most involved in fatigue (channel 3, RF), are greater in the case of PW modulation. So in this case the modulation considered most tiring is also associated with a greater median value of total energy supplied to the patient. The outliers in S3 and S4 distributions, as explained above for other movements like knee flexion supine and hip abduction lateral, are due to those repetitions in which the activation of the therapist was too weak or too strong. Although PW modulation generates greater fatigue, it was rated by all three subjects as the most comfortable stimulation to perform this type of exercise. The I modulation, discontinuous and too sensitive to small variations in current, does not allow to keep the leg raised and often contracts the muscles in a spasmodic way without making the patient understand the movement to be performed. This, in addition to preventing the correct execution of the exercise, is a source of discomfort for the patient.

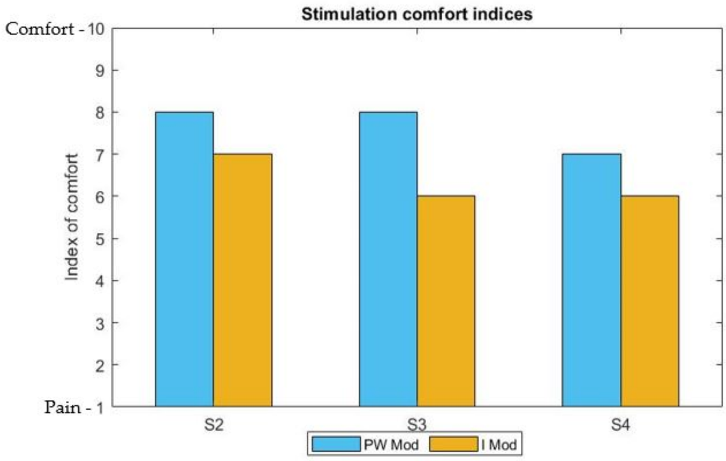

Figure 6.32: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.9 Knee Extension Seated

The patient is seated on the edge of the bed with the legs free to move. He lifts the heel until it reaches the complete extension of the leg and holds this position. Then the limb is returned to the initial position. Since this movement is included only in knee rehabilitation protocols, it was performed only by 3 subjects (S2, S3 and S4). The movement is good both with the PW and I modulations, however only the first one allowed a complete extension of the leg, and the second instead brought the leg to an extension of about 160°. In addition, in the case of PW modulation the extension of the leg occurs much faster than in the case of I mode: as shown in the Figure 6.33, PW modulation carries out the extension of the leg in just over 0.5 s (when the growth trend ends it means that the extension has been reached), while I modulation takes at least 1 s. The main difference between the two types of stimulation, however, is related, also in this case, to the holding phase of the extension. The PW modulation, which is felt as stronger and more intense, despite the small oscillations around the maximum value of pulse width, translates into a continuous and stable stimulation, while the one in current, since is more sensitive to variations of the maximum current value, as already explained above, translates into a more discontinuous muscle contraction and therefore does not always allow to keep the leg in extension.

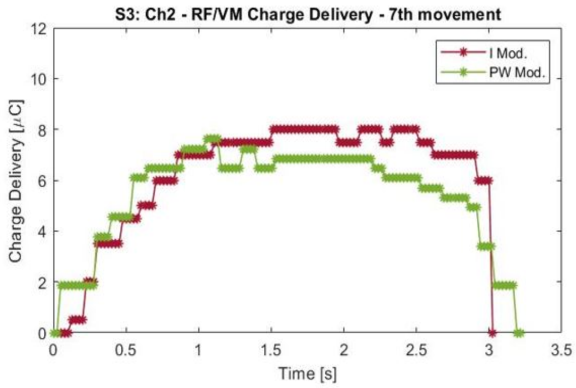

Figure 6.33: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = rectus femoris and vastus medialis (RF/VM).

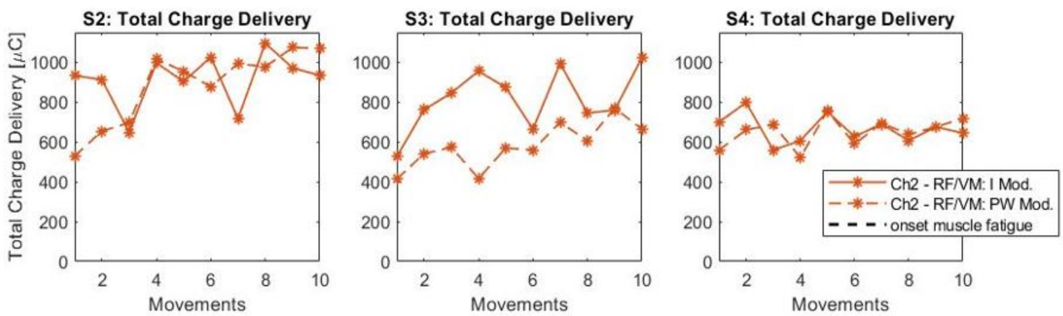

Figure 6.34: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Subjects of knee rehabilitation protocol S2, S3 and S4. The dotted black line represents the movement in which muscle fatigue arises.

No fatigue was felt either during the training or at the end of each session. In fact, the index of muscle fatigue assigned by each subject for each modulation was equal to 1, as shown in Figure 6.35 third row. The outliers that occur in the distributions of the subject S2, Figure 6.35 second row, as explained above for other movements like knee flexion supine and hip abduction lateral, are due to those repetitions in which the activation of the therapist was too weak, as the 3rd and 7th repetitions in I modulation, Figure 6.34.

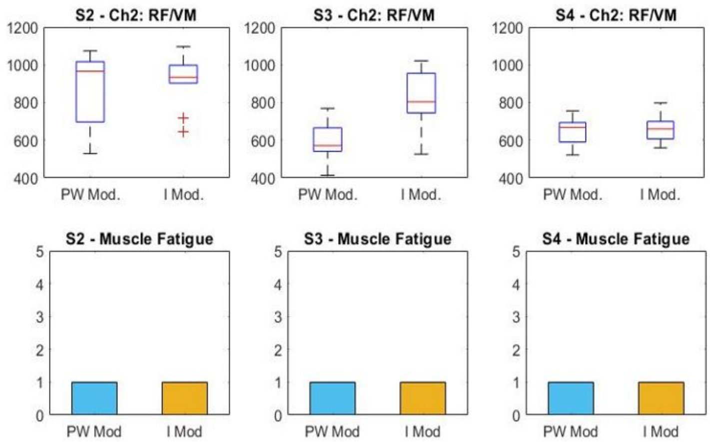

Figure 6.35: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (RF/VM). Second row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Median value of charge delivery distribution [µC]

	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	965.8	933.5	414.0	803.8	668.2	660.0

Table 6.9: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2 = RF/VM.

Neither of the two modulations generated particular discomfort, therefore the preferences of the subjects were assigned only based on the fact that the I modulation sometimes presented oscillations of the current values that resulted in discontinuous and sudden contractions of the muscles, leading to slight discomfort. S4 stated that he did not perceive any difference during the two stimulation sessions.

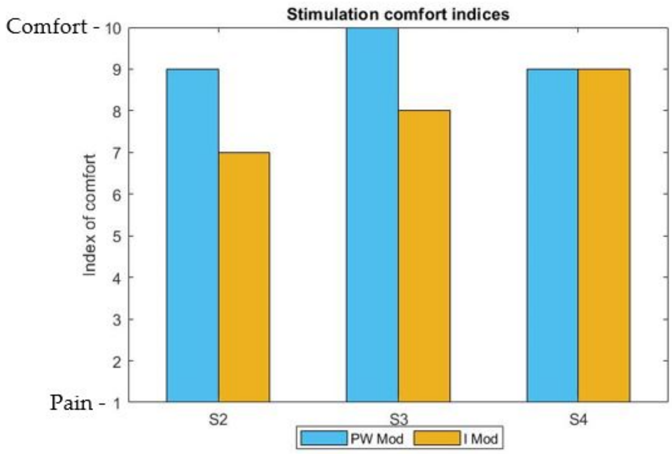

Figure 6.36: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.10 Hip Flexion Standing

The patient is standing and raises the leg toward the chest, keeping the knee flexed at about 90°. After holding this position, he returns to the starting position. Since this movement is included only in hip rehabilitation protocols, it was performed only by 3 subjects (S1, S5 and S6).

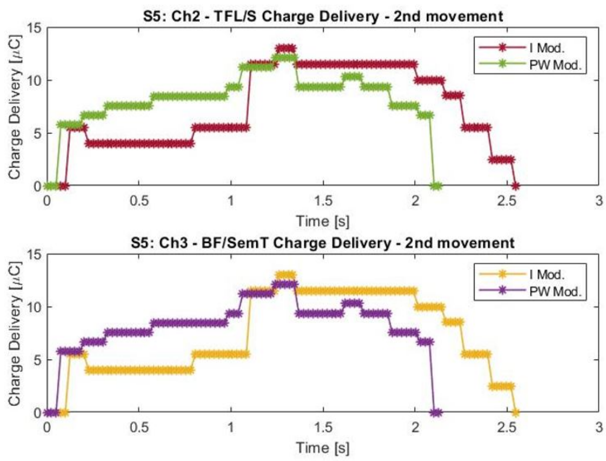

Figure 6.37: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = tensor fasciae latae and sartorius (TFL/S); Ch3 = biceps femoris and semitendinosus (BF/SemT).

Hip flexion takes place in both cases and in particular, PW modulation allows the leg to be brought to a slightly higher height. However, the subjects pointed out that in order for the stimulation to be able to raise the leg, it is useful to slightly shift the body weight on the healthy limb, to prevent the foot from getting stuck on the floor. In the Figure 6.37, it is possible to see that in PW modulation, already starting from pulse width values just above the lower limit, the leg reaches the final position (growth trend of the charge profile ends at about 0.5 s, then the maintenance phase begins), while with the I modulation more than 1 s is required to reach the maximum current and charge values and so the final height of the limb. The most significant difference as regards movement performance is that while the PW modulation generates a stimulation that grows gradually, allowing a more fluid movement, the I modulation often grows impulsively and therefore translates into a more jerky movement. During the position maintenance phase, the PW modulation is more effective because, despite the fluctuations in the pulse width value around its maximum, the leg is held in place. This, however, as already explained for previous movements such as knee extension supine or hip abduction lateral, is not as well realized for I modulation. Therefore, in presence of variations, the leg is kept in flexion thanks above all to the contraction of the patient.

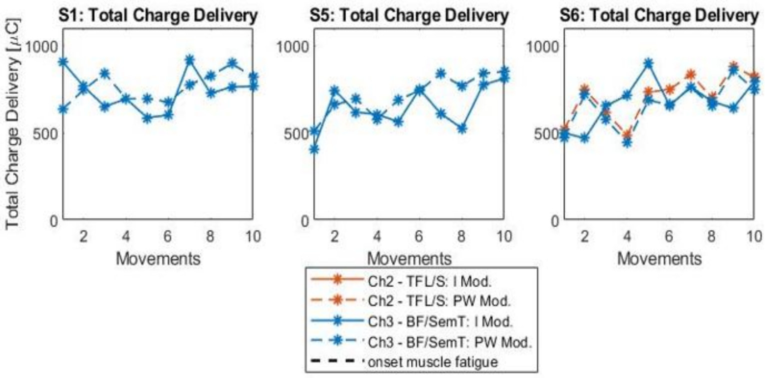

Figure 6.38: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Subjects of hip rehabilitation protocol S1, S5 and S6. The dotted black line represents the movement in which muscle fatigue arises.

No fatigue was felt either during the training or at the end of each session. In fact, the index of muscle fatigue assigned by each subject for each modulation was equal to 1, as shown in Figure 6.39 (third row). For subjects S1 and S5 (both modulations) and S6 (I modulation only), the stimulation parameters used for the two stimulation channels are the same, so the charge values provided at each repetition, Figure 6.38, and consequently also the respective distributions, Figure 6.39 first and second row, are the same.

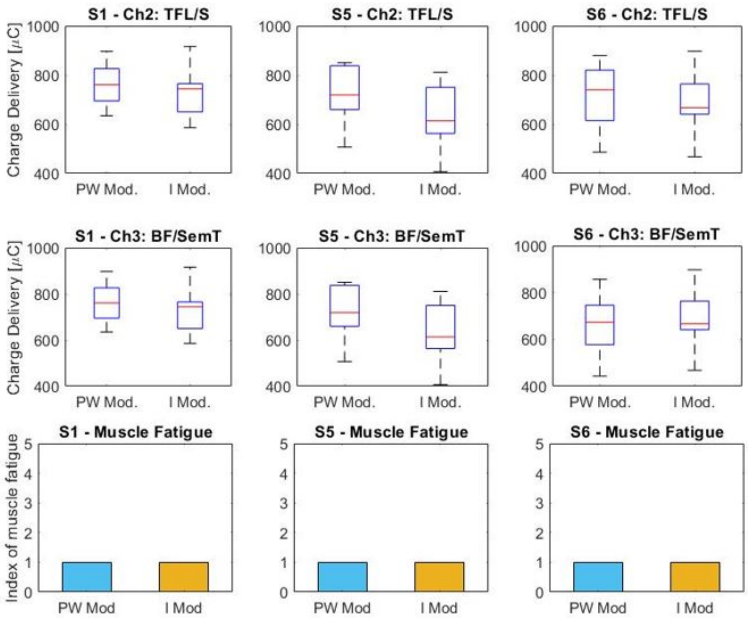

Figure 6.39: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (TFL/S). Second row: distribution of charge delivery to channel 3 (BF/SemT). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Median value of charge delivery distribution [μC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	760.6	744.0	718.7	613.8	740.0	667.0
Ch3	760.6	744.0	718.7	613.8	740.0	667.0

Table 6.10: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. Ch2 = TFL/S; Ch3 = BF/SemT.

The only subject who experienced significant discomfort was subject S5. In particular, the sudden and intermittent stimulation realized by the I modulation was a source of discomfort, especially in the sartorius area. This condition also occurred with PW modulation, but in this case, the perceived discomfort was less due to the continuous and gradual delivery of the stimulation throughout the movement. Subjects S1 and S6 did not perceive particular discomfort in either of the two modulations but agreed with S5 in defining PW modulation as a more comfortable stimulation able to accompany the limb during the execution of the movement with gradual and more constant intensity.

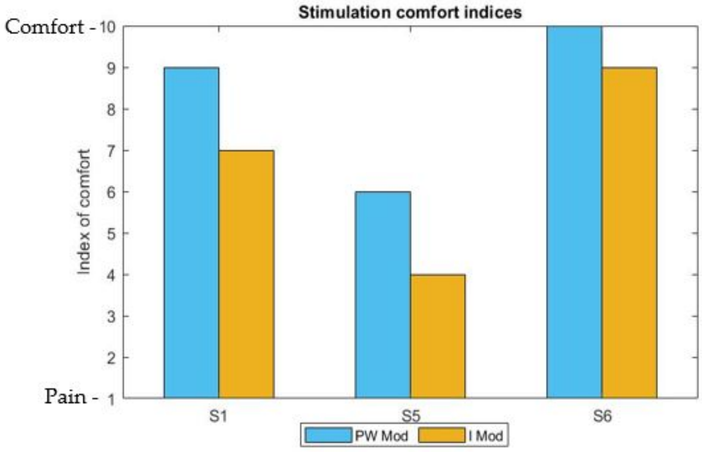

Figure 6.40: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.11 Hip Extension Standing

The subject is standing and lifts the leg back, keeping the knee extended. The position is maintained for a few seconds and then the limb is returned to the starting position. This movement was performed by all six subjects.

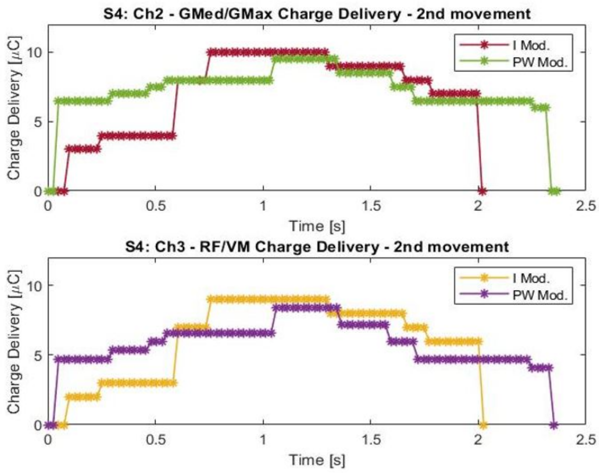

Figure 6.41: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = gluteus medius and maximus (GMed/GMax); Ch3 = rectus femoris and vastus medialis (RF/VM).

The execution of the movement takes place with both modes, but with PW modulation a greater angle of extension is reached. 5 out of 6 subjects agreed that PW modulation is more intense, grows gradually and stimulation perceived is continuous, allowing the movement to be performed smoothly and to maintain the extended position for a few seconds without the patient helping with his own contraction. I modulation, on the other hand, is often weak and uncertain and therefore ineffective in keeping the leg raised without the help of the patient's muscle contraction. In addition, as can be seen from the example given in Figure 6.41, I modulation is often characterized by stimulation that grows briefly at the beginning and then has a sudden increase to the upper limit. This leads the subject to perform the movement in a nonuniform and unnaturally way. Only one subject did not perceive any difference in the execution of the movement and in the perception of the stimulation. No perceived fatigue during training. Only at the end of the PW session, S1, S2 and S3 reported experiencing slight fatigue in the RF area. The subject S4, on the other hand, perceived slight fatigue, always in the same area, at the end of both stimulation sessions. Looking at the Figure 6.43 and the Table 6.11, S2 and S3 present, for the stimulation channel related to the most fatigued muscle (RF, channel 3), greater median values of the charge distributions for modulation in PW, while for S1 the opposite situation occurs. It is therefore not possible to say that the modulation that has produced greater fatigue is related to higher values of charge delivery. The outliers in S1, S2, S4 and S6 distributions, as explained above for other movements like knee flexion supine and hip abduction lateral, are due to those repetitions in which the activation of the therapist was too weak or too strong.

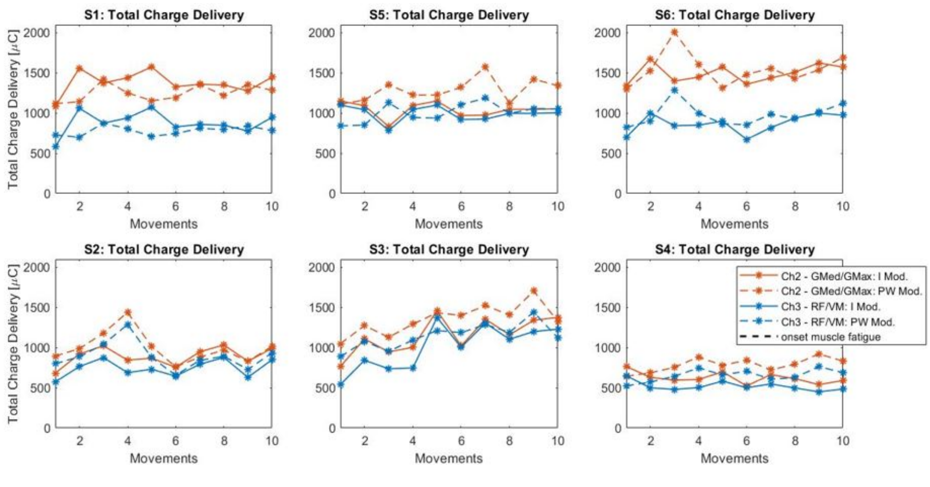

Figure 6.42: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

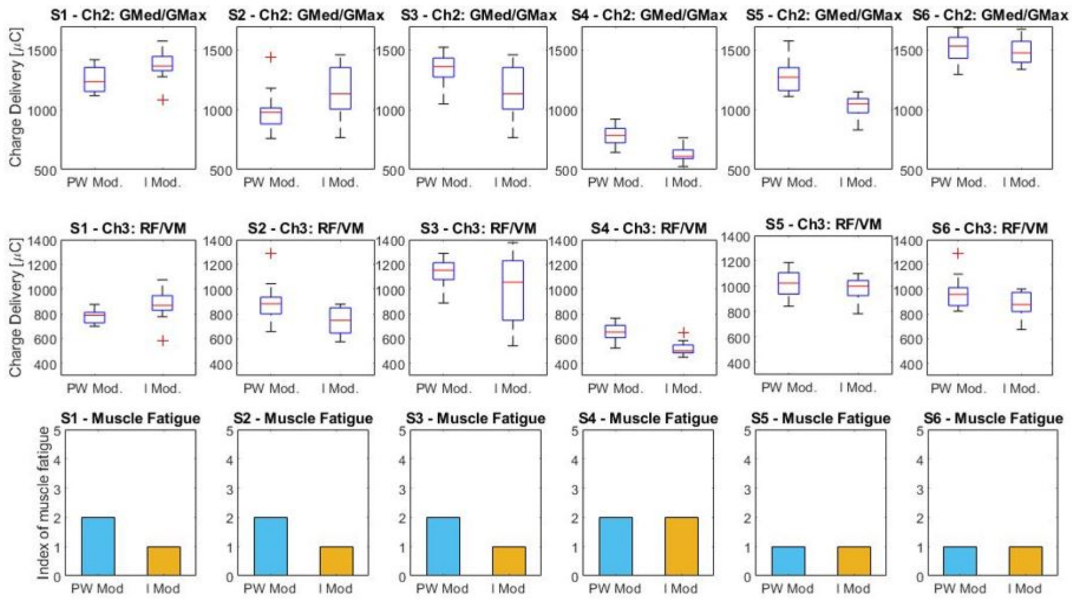

Figure 6.43: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (GMed/GMax). Second row: distribution of charge delivery to channel 3 (RF/VM). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

Median value of charge delivery distribution [µC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	1234.8	1367.0	1272.9	1050.5	1533.5	1477.5
Ch3	788.7	867.0	1023.5	1000.5	957.5	875.0
	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	978.8	1134.3	1362.5	1134.3	786.0	609.5
Ch3	882.3	748.0	1152.8	1054.8	652.5	501.0

Table 6.11: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2 = GMed/GMax; Ch3 = RF/VM.

4 subjects expressed a preference for modulation in PW because even if the stimulation is more intense and slightly more tiring, it is delivered continuously throughout the movement allowing a smooth and natural performance of the exercise. I modulation was evaluated as more uncomfortable due to the fact that

the stimulation sometimes grows suddenly generating discomfort, especially in the RF area. For S4 there was no difference in the perception of the stimulation and S3 has rated I modulation as a more comfortable stimulation since it was less intense and therefore caused less discomfort, especially on the quadriceps.

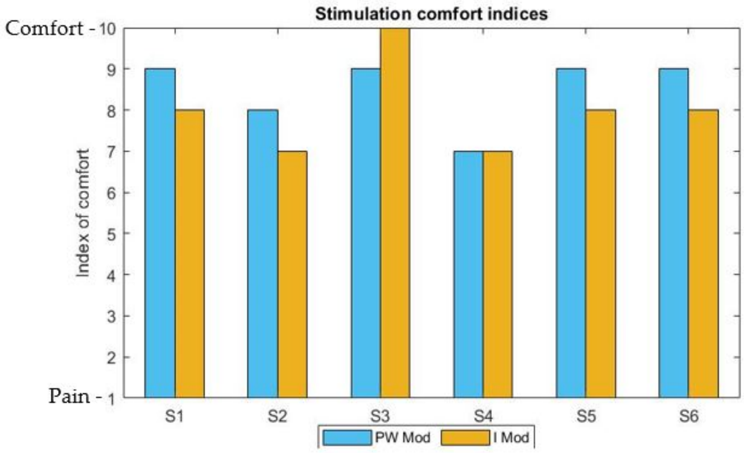

Figure 6.44: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.12 Hip Abduction Standing

The subject is standing and performs the hip abduction by opening the leg outwards. He holds the position and then returns to the starting position. This movement was performed by all six subjects.

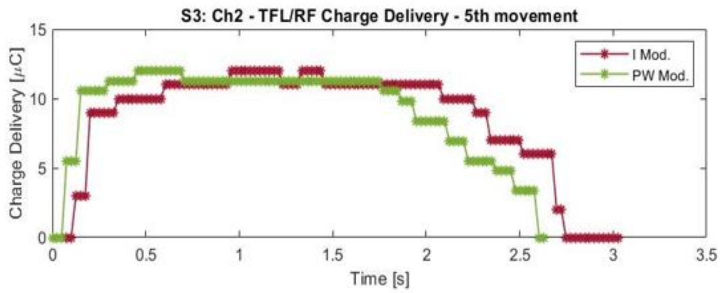

Figure 6.45: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = tensor fasciae latae (TFL/RF).

The movement is well performed with both modulations. All subjects have defined PW modulation as more intense and characterized by a continuous stimulation, while current modulation is milder and takes longer to raise the leg, therefore a higher stimulation time is required to perform the movement. As shown in the example in Figure 6.45, the maximum charge value is reached after 0.5 s with PW modulation, while 1 s of stimulation is required to achieve the same value with I modulation.

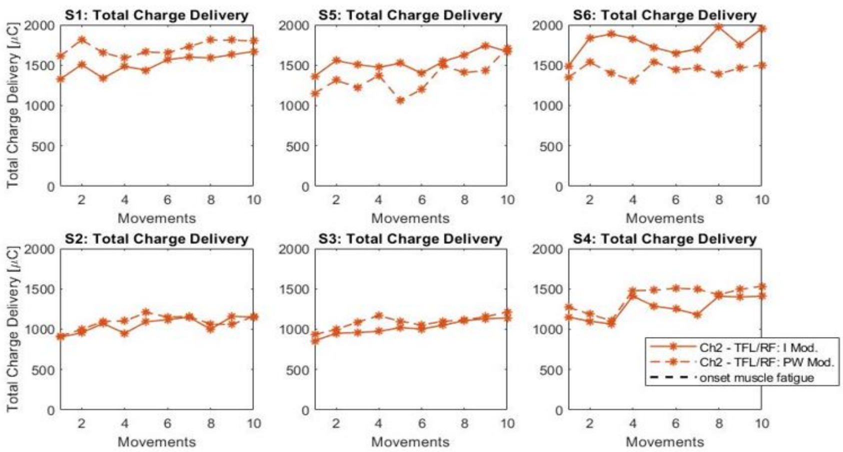

Figure 6.46: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

No fatigue was perceived either during the training phase or at the end of the two stimulations. In fact, the index of muscle fatigue assigned by each subject for each modulation was equal to 1, as shown in Figure 6.47 (third row).

Median value of charge delivery distribution [μC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	1696.7	1539.0	1343.4	1534.5	1452.8	1790.0
	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	1100.4	1082.0	1100.0	1011.5	1481.2	1267.3

Table 6.12: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2 = TFL/RF.

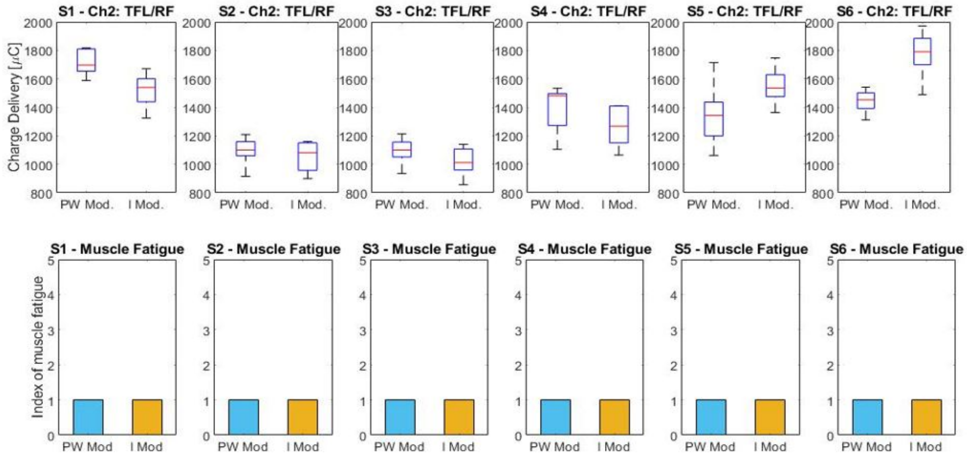

Figure 6.47: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (TFL/RF). Second row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

In general, no subject has experienced particular discomfort and in fact, the indices of comfort are almost all close to the maximum. Two subjects, S1 and S3, did not express any difference in the discomfort experienced during the stimulations. S2 preferred for I modulation because it was less intense and allowed to obtain a good movement as well. Finally, the subjects S4, S5 and S6 chose PW modulation because the stimulation, as long as it was intense, always remained continuous and accompanied the limb throughout the duration of the movement without creating discomfort. I modulation, on the other hand, although it also allowed a good movement, sometimes it was slightly more unstable and therefore a little uncomfortable.

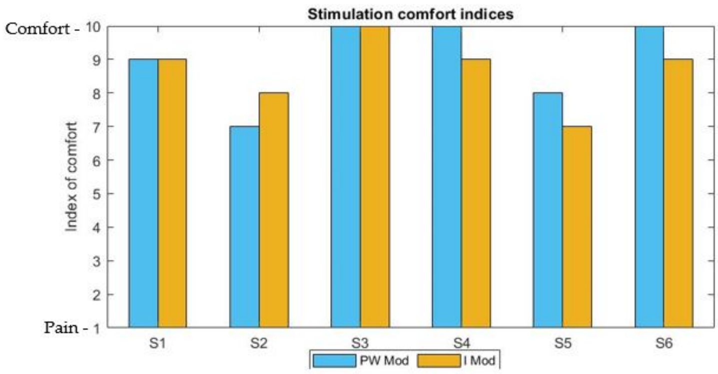

Figure 6.48: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

6.13 Heel Raise

The subject is standing and lifts his heels. He holds this position and, if necessary, uses support to maintain the balance, then returns to the initial position. This movement was performed by all six subjects. Both modes allow the execution of the movement, but the subjects suggested that to help the stimulation lift the heels, it is necessary to start from an initial position in which the weight is slightly shifted to the tips of the feet. Everyone agreed that the stimulation in PW was much more intense than that with I modulation, as shown by the charge delivery trends of both stimulation channels reported in Figure 6.49: as soon as the stimulation begins, the PW modulation immediately reaches higher charge values than those achieved by I modulation. The main difference in the execution of the movement was found during the holding of the position on the tips of the feet: muscle activations, especially those of the GA, sometimes present variations of the ATC parameter from its maximum value which, in the case of I modulation, result in a slightly unsteady stimulation that can make the subject lose the balance, while in the case of PW modulation these fluctuations are not perceived and the stimulation remains stable for the duration of the exercise.

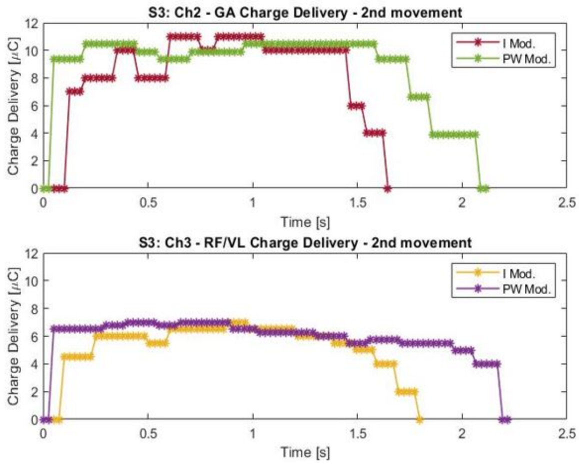

Figure 6.49: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = gastrocnemius (GA); Ch3 = rectus femoris and vastus lateralis (RF/VL). 4 channels were used in total, two in one leg and two in the other. Channels Ch2 and Ch5 were placed on GA of each leg and Ch3 and Ch6 on RF/VL. The stimulation profile of Ch5 was equal to that of Ch2 and that of Ch6 was equal to Ch3.

Only subject S3 stated that he began to experience fatigue in the RF/VL area starting from the 6th movement of the PW modulation training. As represented in the S3 graph in Figure 6.50 (channel 3, dotted blue line), there is indeed an increase in the value of the charge delivery from the 6th to the 7th movement, and from this repetition onwards, the parameter remains constant at this value.

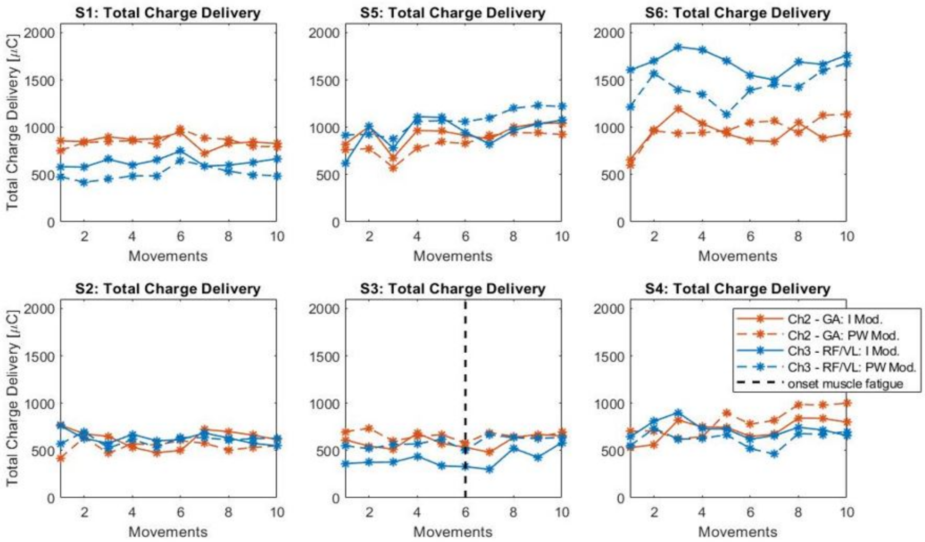

Figure 6.50: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

The subjects S1, S3, S6 were the only ones to have experienced certain fatigue in the quadriceps area at the end of the session with PW modulation and each of them reported a muscle fatigue index equal to 2, Figure 6.51 third row. Figure 6.51 second row and Table 6.13 shows the median values of charge delivery on RF/VL (channel 3, muscles most involved in fatigue), which are greater in the case of PW modulation only for the subject S3.

Median value of charge delivery distribution [μC]

	S1		**S5**		**S6**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	844.3	852.5	837.2	963.0	962.2	931.5
Ch3	488.2	614.8	1065.5	989.0	1412.7	1695.0
	S2		**S3**		**S4**	
	PW Mod	**I Mod**	**PW Mod**	**I Mod**	**PW Mod**	**I Mod**
Ch2	553.3	594.0	658.9	594.0	798.2	747.5
Ch3	620.8	621.3	594.9	376.5	647.3	722.5

Table 6.13: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2 = GA; Ch3 = RF/VL.

S1 and S6 have instead higher charge values in the case of I modulation. It is therefore not possible to say that the most tiring modulation (PW mod) is associated with greater charge distributions. The outliers in S1, S4, S6 distributions, as explained above for other movements like knee flexion supine and hip abduction lateral, are due to those repetitions in which the activation of the therapist was too weak or too strong.

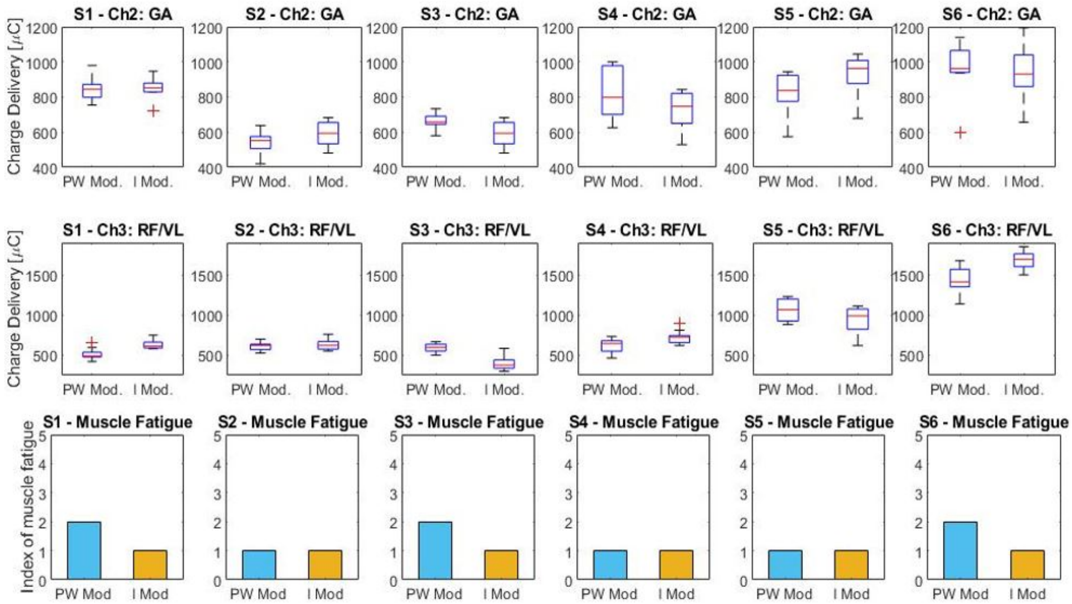

Figure 6.51: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (GA). Second row: distribution of charge delivery to channel 3 (RF/VL). Third row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

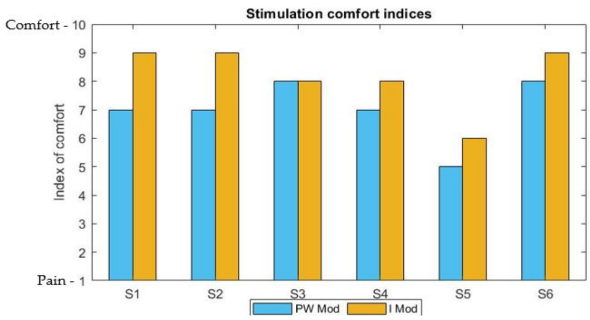

Figure 6.52: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

Finally, 5 out of 6 subjects indicated modulation in pulse width as quite uncomfortable in the GA area, although it is more effective in maintaining the tiptoe position. The current modulation, on the other hand, was less intense and therefore caused less discomfort, still allowing the correct execution of the movement. Only subject S3, despite having defined PW modulation as the most strenuous stimulation, did not feel any difference in terms of discomfort between the two types of stimulation.

6.14 Step-up

The subject puts the foot of the operated limb on the step and gets on it. Then he returns to the initial position by descending from the step with the other leg. This movement was performed by all six subjects.

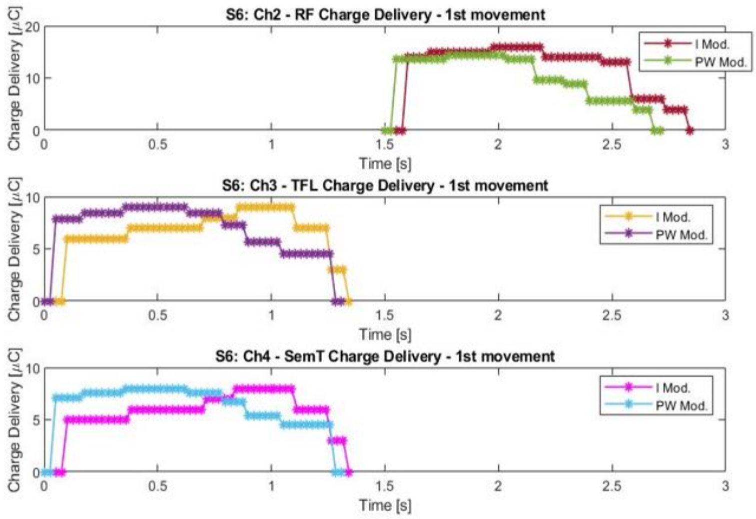

Figure 6.53: Charge delivery to the patient at each pulse (f = 40Hz). Ch2 = rectus femoris (RF); Ch3 = tensor fasciae latae (TFL); Ch4 = semitendinosus (SemT). First the hip flexion is performed (stimulation on Ch3 and Ch4) and the patient puts his foot on the step. Then the pull-up phase takes place (stimulation on Ch2) to get the patient on the step.

The step-up movement, as already explained in section 4.11, is divided into two phases: the first part consists of hip flexion, with which the subject places the foot of the operated limb on the step; the second phase is instead made up of a knee extension, with which the subject extends the leg and climbs on the step. The first part of the movement (hip flexion) was performed well with both modulations. The difference was found during the second phase (pull-up): the stimulation provided with the modulation in pulse width was intense enough to allow the subject to get on the step. In contrast, current modulation often did not provide strong enough stimulation to perform the movement, despite the fact that the maximum current values were increased by 4 mA compared to those used in PW modulation. Using I modulation, the subject began the pull-up phase, but then the contraction force was not enough to end the movement and at this point, the patient intervened with his voluntary contraction to get on the step. Since I modulation is weaker, it was necessary to provide stimulation for a longer time to achieve movement. As shown in Figure 6.53, in the graph relating to the stimulation channel of the rectus femoris (Ch2), which is the muscle involved in the pull-up phase, modulation in PW allows the performance of the knee extension in about 0.5 s (when the constant trend ends and the decrease of the charge begins, the subject has extended almost the entire knee), while the I modulation needs at least 1 s, almost twice the time, to perform the same movement. Subjects S3, S5, S6 experienced fatigue in the RF area during PW modulation training. In particular, for the subject S3, who felt the onset of fatigue starting from the 7th repetition, an increase in the charge delivery on the stimulation channel of the RF (channel 2, orange line dotted in Figure 6.54) can be observed. For subjects S5 and S6, for whom fatigue has arisen starting from the 5th and 6th repetition respectively, the parameter tends to remain constant at a high value.

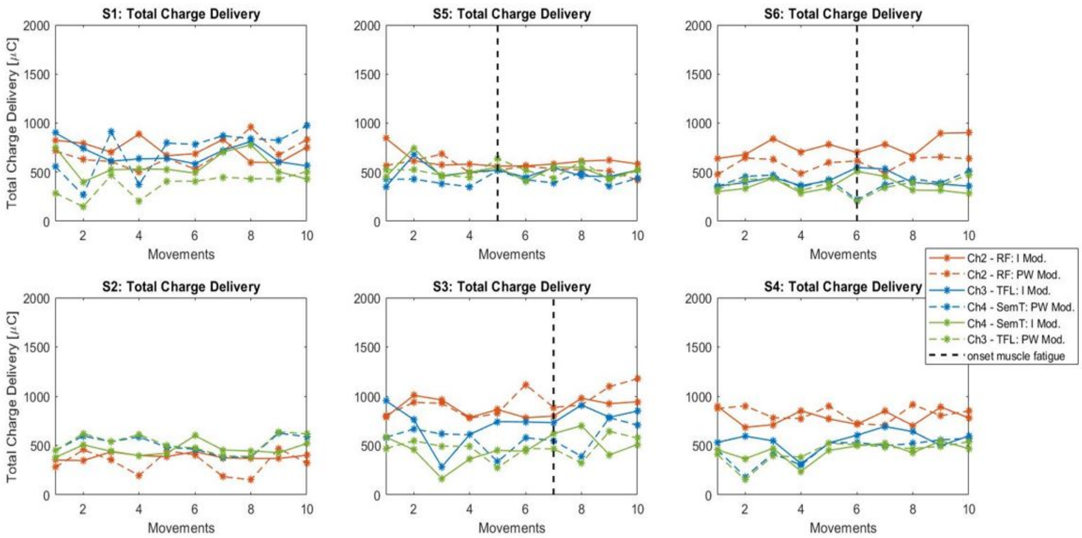

Figure 6.54: Trend of the total charge delivery during the 10 repetitions for each subject and stimulation channel and both types of modulation. Top row: subjects of hip rehabilitation protocol (S1, S5, S6). Bottom row: subjects of knee rehabilitation protocol (S2, S3, S4). The dotted black line represents the movement in which muscle fatigue arises.

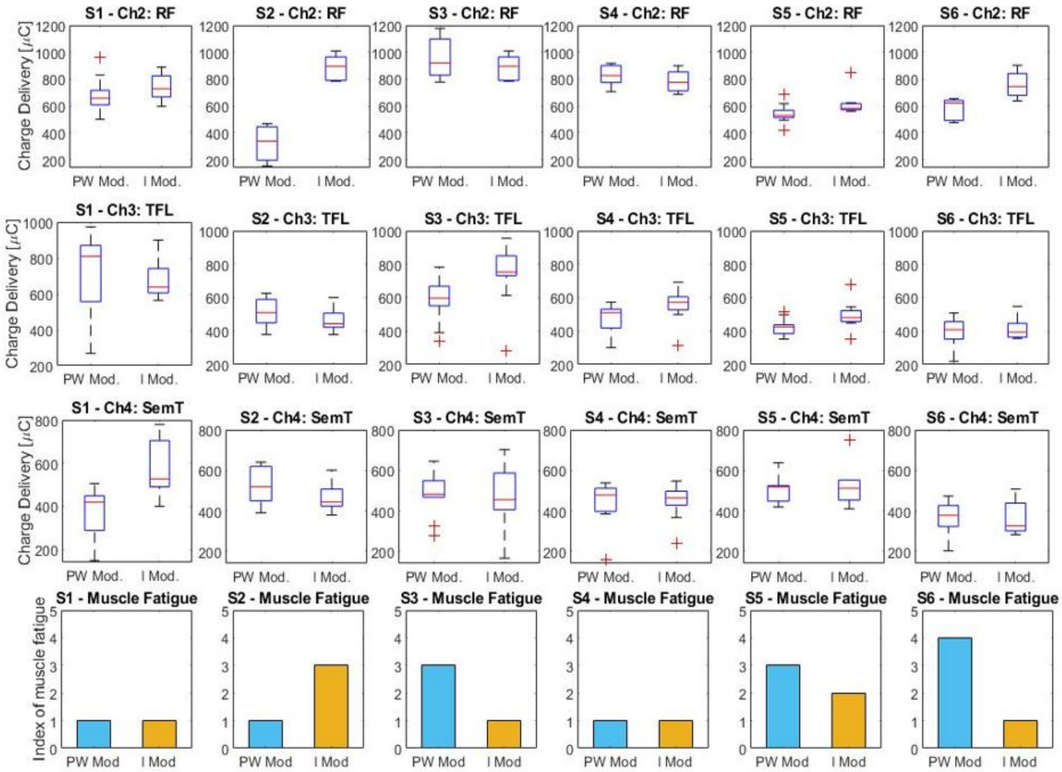

Figure 6.55: Boxplot of the 10 charge delivery values collected in a training in each stimulation modality and representation of index of muscle fatigue perceived. First row: distribution of charge delivery to channel 2 (RF). Second row: distribution of charge delivery to channel 3 (TFL). Third row: distribution of charge delivery to channel 4 (SemT). Fourth row: Index of muscle fatigue perceived (1 – muscles not tired; 5 – muscles extremely tired).

As reported in Figure 6.55 third row, subjects S3, S5, S6 declared that they experienced increased fatigue at the end of the PW modulation session, while S2 felt increased muscle fatigue after the I modulation training. However, looking at the median charge values of the distributions related to channel 2 (RF), in Figure 6.55 first row and Table 6.14, it cannot be said that the modulation considered more tiring is the one that provides a greater charge to the patient: this matching is visible only for S2 and S3, while for S5 and S6 the major median values are associated with the I modulation, for which an index of muscle fatigue of 1 has been expressed (no fatigue). The outliers in S1, S3, S4, S5 distributions, as explained above for other movements like knee flexion supine and hip abduction lateral, are due to those repetitions in which the activation of the therapist was too weak or too strong.

Median value of charge delivery distribution [µC]						
	S1		S5		S6	
	PW Mod	I Mod	PW Mod	I Mod	PW Mod	I Mod
Ch2	658.3	728.5	528.3	583.0	621.2	744.5
Ch3	809.7	638.0	425.5	480.5	408.3	394.0
Ch4	418.7	526.5	517.5	510.8	377.5	325.5
	S2		S3		S4	
	PW Mod	I Mod	PW Mod	I Mod	PW Mod	I Mod
Ch2	338.6	895.0	919.8	895.0	827.2	775.5
Ch3	510.2	444.0	596.4	753.0	510.2	572.0
Ch4	518.6	444.0	481.6	455.0	478.0	463.5

Table 6.14: Median value of charge delivery distribution among 10 movements performed during training for both stimulation's modulation and all subjects. S1, S5, S6 hip rehabilitation protocol's subjects. S2, S3, S4 knee rehabilitation protocol's subjects. Ch2 = RF; Ch3 = TFL; Ch4 = SemT.

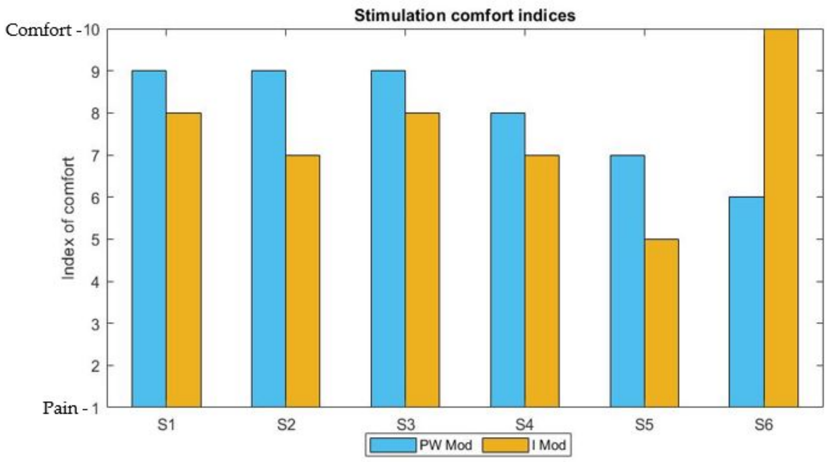

Figure 6.56: Stimulation comfort indices. Each subject expressed the degree of comfort/discomfort perceived during the training of both modulations. 1 means pain, while 10 comfort.

Finally, the subjects assigned the scores to the discomfort perceived during the training. These data are related only to the pull-up phase, considering that the first part was natural and comfortable for all subjects. Despite the greater fatigue related to PW modulation, 5 out of 6 subjects preferred this stimulation because it was strong enough to progressively extend the knee until the patient got on the step. During the I modulation instead the stimulation allowed to begin the extension of the knee, but then, about halfway up the climb, the movement was interrupted because the stimulation was not strong enough to complete the climb.

The muscles then continued to contract without producing a progressive extension of the knee, thus leading to a discomfort concentrated especially on the quadriceps area. Subject S6 is the only one who has not encountered this problem because its high current values allowed, even in the case of I modulation, to completely extend the knee and then climb on the step. Despite this, he perceived I modulation as more comfortable than the PW modulation, which was too intense.

6.15 Overall results

6.15.1 Hip rehabilitation protocol

As explained in chapter 1, the hip protocol consists of movements divided into 3 phases depending on the difficulty and effort required of the patient to perform them.

The results obtained from the tests related to those exercises which belong to the first phase of the protocol and include the stimulation of small muscles such as the gastrocnemius (GA) or dynamic movements of the operated limb, suggest the use of current modulation to obtain a better performance of the movement. In the exercise of ankle flexion-extension the muscles involved, TA and GA, are small muscles that do not require a high contraction force to perform a movement that is very limited and does not involve a weight lifting. Considering that for TA and GA pulse widths around 200-350 µs are sufficient to obtain a foot extension and flexion [102], the modulation in pulse width, which provides for a variation of the pulse width up to a maximum of 500 µs, is too intense and can lead to the generation of cramps, especially on the GA. Moreover, since the current remains fixed at the maximum value tolerated by the patient from the moment the stimulation begins, the contraction force generated is very high even for pulse width values just above the lower limit. This results in a too elevated contraction that is therefore useless for the execution of this exercise. On the contrary, in current modulation, the current intensity varies from 0 to a maximum value, and with it also the contraction force, which initially remains very low and then gradually increases. This causes the stimulation to be more comfortable and to obtain an adequate contraction force for this type of exercise. In the exercise instead of knee flexion supine, the limb is never lifted from the bed and is only dragged up and down. For this reason, even this type of exercise does not require a high contraction force, but a stimulation that allows the limb to move smoothly and quickly. The contraction produced by the modulation in pulse width generates a contraction force so high that it brings the knee up to the chest. So even in this case, the force of contraction is exaggerated to perform the exercise, which involves only a simple flexion of the knee without detaching the limb from the bed. The current modulation, on the other hand, remains lighter, allowing a natural and fluid performance of the movement

without causing discomfort.
The remaining 4 exercises of the first phase of the protocol (knee extension supine, hip abduction supine, hip abduction lateral and hip extension prone) involve lifting of the operated limb from the bed and the maintenance of this position for a few seconds. So for these exercises, the generation of a sufficiently high contracting force becomes of primary importance. The results obtained from the experimental tests have confirmed that the modulation in pulse width allows a better performance of the movement. The latter, as already explained above, is characterized by a fixed current at the maximum value, which allows producing an intense contraction from the first moments of the stimulation, lifting the limb from the bed even for pulse width values slightly higher than the lower limit. The final position of the leg is thus reached for pulse width values even quite lower than the upper limit (300-400 µs, upper limit = 500 µs) and therefore, during the holding of the position, if the stimulation pattern is characterized by fluctuations around the upper limit, they are not perceived at the level of muscle contraction, which remains continuous and constant ensuring the maintenance of the desired position for a few seconds. In current modulation, on the other hand, intensity of at least half of the maximum current value are required to begin the execution of the movement, and the final position of the limb is reached only for current values equal to the maximum (I_{max}) or slightly lower than it (1-2 mA). For this reason, any variations in the stimulation signal from the I_{max} cause a non-continuous and intermittent contraction, which prevents the patient from keeping the leg in the specific position. Modulation in PW is, therefore, preferred by subjects, despite causing greater fatigue.

In the second phase of the rehabilitation program are planned the 3 standing exercises (hip flexion standing, hip extension standing and hip abduction standing), which always include a phase of maintenance of the final position reached by the leg and therefore a greater force of contraction. The results obtained for these exercises were the same as for the limb lifting exercises belonging to the first phase: the modulation in PW, although it is more tiring, allows to obtain a better movement performance, especially in the holding position phase.

Finally, the third phase involves the heel raise and step-up movements for which two different results have been obtained. For heel raise, mixed results were returned from the subjects. The modulation in pulse width allows better maintenance of the position on tiptoe, but generates a great discomfort in the GA area, as explained above for the ankle flexion- extension. Current modulation is more comfortable, but may be slightly more unstable if the stimulation pattern presents current intensity variations from the maximum value. However, 5 out of 6 subjects gave more importance to the discomfort experienced and chose current modulation, which still allows obtaining a good movement.
The step-up exercise consists of two movements, hip flexion and knee extension. The first movement, necessary for the patient to put his foot on the step, does not

involve maintenance of position or sensitive stimulation muscles, so no difference has been detected between the two modulations in this first part. The second on the contrary, necessary for the patient to get on the step, requires a great force of contraction on the quadriceps. During the I modulation the stimulation allowed to begin the extension of the knee, but then, about halfway up the climb, the movement was interrupted because the stimulation was not strong enough to complete the climb. The muscles then continued to contract without producing a progressive extension of the knee, thus leading to a discomfort concentrated especially on the quadriceps area. The modulation in PW instead, despite being more tiring, was strong enough to progressively extend the knee until the patient got on the step and for this reason, it was preferred by the subjects.

The results showed also that, for those subjects who perceived fatigue during a particular training, this fatigue, from the moment it arises, is accompanied by an increase in the charge delivery parameter at each repetition or its stabilization at a high value. This means that as the training continues if more and more energy is supplied or if it continues to be delivered at a high value, this causes progressive fatigue in the patient that lasts until the end of the session. However, it has not been proven that the modulation considered more tiring is also the one associated with higher median values of charge distributions. This could be due to the fact that during a training, the charge values provided at each repetition vary depending on the muscle activation of the therapist. If the patient feels discomfort, the therapist tries to decrease the amount of charge that is sent to the patient by contracting his/hers muscles less, while if the stimulation is not strong enough to perform the movement, then the muscle activation of the therapist is increased. So even a few low charge values could decrease the entire distribution and cause lower charge median values to be associated with modulations with higher indices of muscle fatigue and vice versa.

6.15.2 Knee rehabilitation protocol

As explained in chapter 1, the knee protocol consists of movements divided into 3 phases depending on the difficulty and effort required of the patient to perform them.

Considering that almost all the movements belonging to the first phase of the protocol are the same as in the hip protocol, for all the exercises in common, the same results reported in the previous paragraph apply. The only exercises that are not present in this rehabilitation protocol are hip abduction supine and hip extension prone, which are replaced by knee flexion prone and straight leg raise. Both require a high force of contraction to perform the movement and to maintain, for a few seconds, the final position reached by the leg. So as explained before, the best modulation is the one in pulse width, which causes greater fatigue, but allows a constant and continuous stimulation, which is very effective during especially the maintenance phase.

In the second phase of treatment, the only exercise of the hip protocol that does not appear in this one is hip flexion standing, which is replaced with knee extension seated. This type of exercise also requires a high contraction force to achieve knee extension and to be able to hold it for a few seconds. So also in this case the subjects preferred pulse width modulation for the correct execution of the movement.

Finally, the third phase of this rehabilitation program includes the same movements also defined for the hip protocol: heel raise and step-up. The considerations for both exercises are reported in the previous paragraph.

The results relating to the charge delivered and fatigue, described above, also apply to the movements of this protocol. Providing a constant or continuously increasing value of energy to the patient can lead to muscle fatigue that lasts throughout the session. However, it cannot be said that a higher median value of charge delivery is associated with the modulation considered more tiring.

Chapter 7

Conclusions

In this thesis, two lower limb rehabilitation protocols that can be implemented through FES are presented: hip protocol and knee protocol.

To outline the two protocols, an event-driven approach applying the Average Threshold Crossing (ATC) technique to the surface ElectroMyoGraphic (sEMG) signal was exploited. The ATC technique is based on generating an event whenever the sEMG signal exceeds a certain threshold. By averaging the number of TC events within a time window, the ATC value is computed. Therefore, using the ATC-FES system in the therapist-patient mode, each time the therapist performs a movement, ATC values are extracted from his/her muscle activations through wearable acquisition boards. The ATC parameters are then sent via Bluetooth to the computer, which processes them and translates them into stimulation patterns to provide to the patient. In particular, these stimulation profiles are based on current intensity modulations which, thanks to the surface electrodes, contract the patient's muscles in order to realize the movement that the therapist is performing.

In this work, first of all, two improvements have been proposed to the software of the ATC-FES system. The first concerns how to obtain a stable activation profile and its use in the contraction of one or more muscles. In particular, by connecting multiple acquisition boards to a single stimulation channel (therefore only one muscle is stimulated), the activation signals of all the muscles involved in the execution of a single movement are combined to obtain a single more stable activation profile to provide to the only stimulation channel involved. If, on the other hand, a single acquisition board is coupled to multiple stimulation channels (therefore more muscles are contracted), the same activation profile is used to generate a single stimulation pattern to be sent to all the FES channels. This avoids the use of multiple boards to record signals with similar activations.
The second addition to the software concerns the possibility of providing the patient

with an alternative stimulation mode to that based on current intensity: pulse width modulation (PW).
Subsequently, the movements of each protocol (11 for each protocol) were defined following a similar procedure for all. First, an analysis of the different muscular activities involved in a specific movement was performed. Then the best muscle signals were combined to achieve the most stable and effective activation profile. Finally, a study of the most standard possible placement of surface electrodes on the patient's muscles was done.

Experimental tests were conducted to validate the two protocols and to compare any differences between intensity current (I) and pulse width (PW) modulation. A total of 6 male subjects were recruited, divided equally between the two rehabilitation protocols. The movements of each protocol were performed using both modes of stimulation and comparisons were made in terms of movement execution, discomfort perceived by the patient and muscle fatigue. These data were obtained through forms filled out by patients at the end of each test, in which questions were asked about the discomfort experienced during the stimulation phase, the possible onset of fatigue during the repetitions of a movement and the level of muscle fatigue perceived at the end of the session.
Since almost all movements are in common between the two protocols, the results obtained apply to both treatments. Tests have shown that:

- for movements, such as ankle flexion-extension and heel raise, involving small muscles (e.g.: gastrocnemius, tibialis anterior) a preference within the range 66.7-83.3% was expressed for current modulation, which allows the execution of the movement without generating discomfort. Instead, considering that for TA and GA pulse widths around 200-350 µs are sufficient to obtain a foot extension and flexion and that modulation in pulse width provides for a variation of the pulse width up to a maximum of 500 µs, this type of stimulation is too intense and can lead to the generation of cramps, especially on the GA;

- for movements that do not require a high contraction force, but a stimulation that allows the limb to move smoothly and quickly, such as knee flexion supine, where the limb is never lifted from the bed and is only dragged up and down, the contraction produced by the modulation in pulse width generates a contraction force exaggerated to perform the exercise. On the other hand, current modulation, for which a preference of 66.7% has been expressed, remains lighter, allowing a natural and fluid performance of the movement without causing discomfort;

- for movements involving large muscles (e.g.: quadriceps femoris, hamstring muscles, gluteal muscles) and that involve lifting of the operated limb from the bed and the maintenance of the position for a few seconds or the lifting of a significant load (e.g.: step-up movement), the generation of a sufficiently high contracting force becomes of primary importance. The modulation in pulse width, in these cases, is the one that allows a better performance of the movement. Since it is characterized by a fixed current at the maximum value, allows producing an intense contraction from the first moments of the stimulation. In addition, any fluctuations in the stimulation pattern around the maximum value are not perceived at the level of muscle contraction, which therefore remains continuous and constant for the duration of the stimulation. On the contrary, current modulation requires intensity of at least half of the maximum current value to begin the movement, and the final position of the limb is reached only for current values equal to the maximum (I_{max}) or slightly lower than it (1-2 mA). For this reason, any variations in the stimulation signal from the I_{max} cause a non-continuous and intermittent contraction, which prevents the patient from generating an effective contraction force.

The results showed also that, for those subjects who perceived fatigue during a particular training, this fatigue, from the moment it arises, is accompanied by an increase in the charge delivery parameter (calculated as the product of amplitude and pulse width of each pulse, IxPW) at each repetition or its stabilization at a high value. This means that as the training continues if more and more energy is supplied or if it continues to be delivered at a high value, this causes progressive fatigue in the patient that lasts until the end of the session. However, it has not been proven that the modulation considered more tiring is also the one associated with higher median values of charge distributions. This could be due to the fact that during training, the charge values provided at each repetition vary depending on the muscle activation of the therapist. If the patient feels discomfort, the therapist tries to decrease the amount of charge that is sent to the patient by contracting his/hers muscles less, while if the stimulation is not strong enough to perform the movement, then the muscle activation of the therapist is increased. So even a few low charge values could decrease the entire distribution and cause lower charge median values to be associated with modulations with higher indices of muscle fatigue and vice versa.

Although the results obtained from the tests are promising, the two modulations should be tested on patients with real motor limitations to understand if the two stimulations actually contribute to the recovery of movement and muscle strengthening and if one of the two modulations is more effective than the other.

Finally, the current system could be further improved by also inserting the frequency modulation option, which could be the real key to decreasing muscle fatigue as suggested by many studies [57][59][60][61].

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