Workshop on Transcutaneous Functional Electrical Stimulation

Eukene Imatz Ojanguren, Milos Kostic, Noelia Chia Bejarano and Thierry Keller

Abstract This chapter aims to give a general description of basic concepts related to transcutaneous FES. It offers examples of simple exercises to introduce the reader into the practical aspects of the application of transcutaneous FES. Different influencing aspects such as stimulation waveform, stimulation parameters, electrode type, placement, and size are analyzed. Available models related to FES that represent the electrical properties of the skin, current distribution on the skin, or either nerve excitability are presented as well, highlighting those factors that affect most transcutaneous FES applications. A practical guide on upper and lower limb is also presented, where different exercises are proposed to experience previously described theoretical aspects in practical application of FES. Finally, conclusions of the chapter and challenges observed during the exercises are described and novel techniques and technology used to overcome some of these challenges are mentioned.

1 Introduction

Functional electrical stimulation (FES) is a technique that applies electrical impulses to the peripheral nerves in order to elicit an artificially induced contraction of a muscle. The current is applied through at least a pair of electrodes that can be placed over the skin (transcutaneous or superficial stimulation), or inside the body (percutaneous

E. Imatz Ojanguren (✉) · M. Kostic · T. Keller
Rehabilitation Department - Health Division, Tecnalia Research and Innovation,
Mikeletegi Pasealekua 1, 20009 Donostia-san Sebastin, Spain
e-mail: eukene.imatz@tecnalia.com

M. Kostic
e-mail: milos.kostic@tecnalia.com

T. Keller
e-mail: thierry.keller@tecnalia.com

N. Chia Bejarano
Neuroengineering and Medical Robotics laboratory, Department of Electronics, Information and Bioengineering, Politecnico di Milano, Via Giuseppe Colombo 40, 20133 Milano, Italy
e-mail: noelia.chia@polimi.it

J.L. Pons et al. (eds.), *Emerging Therapies in Neurorehabilitation II*,
Biosystems & Biorobotics 10, DOI 10.1007/978-3-319-24901-8_11

or implantable stimulation depending on the degree of invasiveness). In any of these cases, an electric field is created between the two electrodes, producing current that propagates through the tissue and depolarizes muscle nerve fibers, generating an action potential. This action potential propagates along the nerve to the innervated muscle, where it creates a muscle contraction [23].

Neurological disorders such as brain injury or spinal cord injury can often lead to motor dysfunctions, which affect negatively to their quality of life. This inability of performing many ADL tasks can be overcome with the application of FES either by including it in rehabilitation programs or using it as an assistive device. Although its use in assistive applications is still limited, results of the application of FES in therapy have shown significant benefits [5, 18, 19].

2 Functional Electrical Stimulation (FES) Basics

2.1 Principle of Electrical Stimulation

When a muscle is physiologically contracted, the alpha motor neurons that are first activated are those innervating slow-twitch and fatigue-resistant muscle fibers. This recruitment order minimizes fatigue and guarantees that the less fatigue-resistant fibers are activated only when a higher force needs to be produced. However, when the muscle contraction is electrically induced through the skin, the recruitment order is typically inverted, because the motor neurons that are excited more easily are those with axons that have larger diameters, which usually innervate the less fatigue-resistant muscle fibers. The reality in transcutaneous FES applications is a bit more complex, since the activation of motor neurons depends not only on the size of their axons, but also on the stimulation parameters (frequency, amplitude, pulse-width) and on the distance between the motor neuron and the electrodes. Electrically-induced muscle contractions are in general more fatiguing compared to the natural activation of the muscle, because the muscle contraction is generated by continuously activating the same set of nerve and muscle fibers.

These differences between physiologically- and artificially-induced muscle contractions represent some of the challenges that have been overcome to some degree with novel devices and techniques. The choice of the appropriate electrodes and stimulation parameters is critical to produce a muscle response as natural as possible. The following sections will shed some light on this topic, giving special attention to transcutaneous stimulation.

2.2 Electrodes and Devices

The electrodes used for functional electrical stimulation can be classified into three categories, according to their invasiveness: transcutaneous, percutaneous, and implantable. Transcutaneous electrodes are the most used in practical applications, probably due to their versatility and minimum invasiveness. Percutaneous electrodes have been used in the past for better selectivity, specifically for the activation of deeper muscles that are difficult to activate from the surface. However, in chronic applications there is always a risk of infection. Nowadays percutaneous electrodes are mostly used for experimentation and neurophysiologic measurements that require high selectivity for a short period of time. Implantable electrodes are preferred as a permanent solution that minimizes the discomfort of the user since afferent skin receptors and nerves are not necessarily activated, as it is the case with transcutaneous electrodes.

Transcutaneous or surface electrodes stimulate the nerve fibers through the skin, making them the least invasive type of electrodes, but also the least selective for the activation of deeper neuronal structures. They must be carefully placed over the muscle belly or the nerve trunk, using the anatomical landmarks as guidelines. However, in most cases some readjustments are needed to obtain the desired elicited response, especially if the aim is to produce a pure movement that rotates the joint only around one axis. The muscle response depends on multiple factors, such as the properties of the muscle (muscle type, hydration, etc.), the thickness of the skin and the underlying fat tissue, and the position and exact orientation of the nerve fibers with respect to the electrodes. In fact, the high inter-subject variability constitutes one of the main drawbacks of transcutaneous electrodes, coupled with the difficulty to stimulate deep muscles, a poor cosmetic appearance, and the risk of the electrodes to come off the skin and alter the impedance during wear. Additionally, transcutaneous stimulation activates the sensory fibers and pain receptors of the skin, producing sometimes discomfort or pain to the subjects. On the other hand, transcutaneous electrodes have the clear advantage of being non-invasive and flexible to repositioning. This makes them the preferable solution in applications where the electrodes must be often relocated (e.g., when the paralysis after an injury changes over time, or when the muscle volume increases because of the training), in spasticity treatments where the antagonistic muscles must be strengthened, or in simple applications where only a few electrodes are needed to elicit a response [23]. As a consequence, main design criteria for these electrodes affect their comfort and performance; sufficient electrical surface area to prevent skin irritation, flexibility to adapt to the body surface, easy donning and doffing without being prone to come off the skin, reliable connection to the stimulator, electrical properties (stable electrical resistivity), material properties (made of hypoallergenic materials, resistant to biomedical chemicals), and convenience (reusability, low cost) [2].

Percutaneous electrodes, also called needle electrodes, are in between transcutaneous and implantable electrodes, and therefore share some of their advantages and drawbacks. They can be applied with minimal invasive procedures, since the electrodes are inserted in the body through a puncture on the skin. Only the wire tip is conductive, so, being subcutaneous, the discomfort caused to the subject is lower than with transcutaneous electrodes. However, the skin interface needs to be carefully maintained in order to reduce the risk of infection, although it has been reported only a few times in literature during application. The main risk of infection occurs through extraction and related electrode fractures that leave parts of the electrode inside the body. An infection rate of 16 % has been reported in subjects with long treatments [8]. Percutaneous electrodes are positioned with an epidermal needle next to the nerve and allow the stimulation of deep muscles. In case percutaneous electrodes move or break, the subjects do not have to undergo a surgery, since wires can be easily replaced. However, remaining artifacts are left in the body. Nevertheless, percutaneous electrodes have not been used in literature as much as transcutaneous or implantable electrodes.

Implantable electrodes are the most invasive solution, and therefore the choices of materials that are biocompatible and resistant to mechanical stress are critical in their design. They consist of two main parts: the electrode wire, often made of stainless steel, and either an electrode cuff including multiple contacts that is wrapped around the targeted nerve or a silicon embedded platinum plate (in case of epimysial electrodes) that is sewn on the epineurium [23]. The short distance between the electrode and the nerve fiber guarantees the highest sensitivity and good selectivity when properly placed. Their placement requires a surgery, which must be repeated if the electrodes break or move. Implantable electrodes are indicated for long-term applications that want to free the patient from continuous placement or repositioning of the electrodes, for systems that minimize the skin sensation (or discomfort) caused by surface electrodes, or for applications where surface electrodes cannot be applied, such as the stimulation of deep muscles [23].

Implantable electrodes have also been used in the past for high-density stimulation involving many muscles, e.g. in hand and finger movements where muscles are spatially close. Conventional surface electrodes are difficult to use for this type of applications; however, in the last few years transcutaneous electrode arrays have arisen as an alternative solution to implants. An electrode array is a matrix that contains many small electrodes that can be activated independently, and it is very useful when placed over a skin area with several motor points, allowing the simultaneous stimulation of a few muscles using only one matrix. Additionally, the high density of electrodes opens the way to new control techniques that can achieve personalized transcutaneous stimulation. In fact, several algorithms have been recently proposed to automatically calibrate the single electrode or cluster of electrodes that must be active to elicit a specific response, and this has drastically reduced the set-up times and has eliminated the need to reposition the matrix. Some other novel techniques have been proposed to change the distribution of active electrodes with time, switching between fibers and reducing muscle fatigue.

2.2.1 Influence of Stimulation Parameters

The human musculoskeletal system is highly complex, and its response to external stimuli varies between subjects, over time, muscle and fiber locations, among others. Therefore, even if some general guidelines can be established, most FES applications have an initial phase of calibration, in which several parameters are tested in order to choose the best solution. The most important parameters are the stimulation waveform, frequency, amplitude and pulse-width, and the electrode size and position. These are going to be analyzed in the following pages, providing some practical hints to the reader.

Stimulation waveform

The stimulation waveform used for FES applications is generally a train of rectangular pulses, either monophasic or biphasic (see Fig. 1). Monophasic pulses are often used for research purposes, as they do not contain a hyperpolarization pulse (anodic pulse) that balances the electrical charge applied to the muscle. Most FES applications use biphasic pulses, with a depolarization pulse (cathodic current pulse) that in most cases is rectangular, and an hyperpolarization pulse (anodic current pulse) that can have a symmetric shape, or has a longer width and a smaller amplitude (which can be useful if the amplitude is below the motor threshold) or in some cases has an exponential decreasing shape. Generally, symmetric biphasic waveforms are used to generate force in bigger muscles, whereas asymmetric waveforms are preferred for smaller muscles where higher selectivity is needed.

Stimulation frequency

A stimulation pulse produces in the muscle a twitch that fades out after 200 ms (see Fig. 2). When the stimulation signal is a train of pulses, the ratio between this muscle relaxation time and the period between pulses is crucial to understand the response of the muscle, which is highly dependent on the stimulation frequency.

When the stimulation frequency is below 10 pulses per second (10 Hz), the muscle twitches can still be differentiated from each other, and the muscle response can be characterized as a sort of tremor. This effect gradually decreases when the stimulation frequency increases, usually around the band of 10–30 Hz, until it reaches a point where it evokes a smoother muscle response known as tetanic contraction (see Fig. 2). The frequency when this happens is called the fusion frequency, and its exact value varies between subjects and fibers.

However, this does not mean that functional electrical stimulation should always use high frequencies. In fact, when the stimulation frequency keeps increasing toward the fusion frequency and above, the muscles start to fatigue very fast. The main reason is that with FES the excited fibers are all firing synchronously and not asynchronously and distributed among several hundreds to thousand fibers as it is the case in the

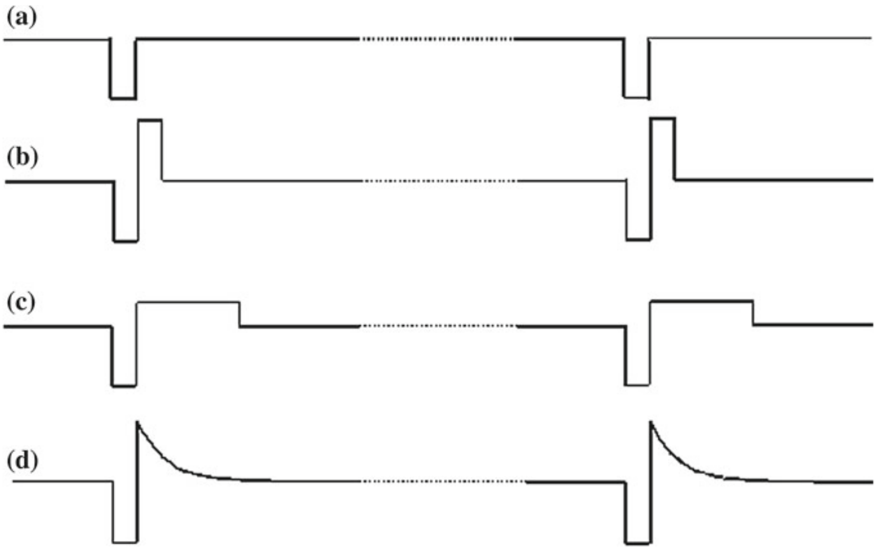

Fig. 1 **a** Commonly used pulse forms. The depolarization pulse is for all pulse forms rectangular with a pulse-width up to 300 μs and the hyperpolarization pulse is (**a**) not existent = *monophasic pulse*; **b** the same as depolarization pulse = *symmetric biphasic pulse*; **c** longer than the depolarization pulse, with a subthreshold amplitude = *asymmetric rectangular biphasic pulse*; or **d** shorter than the depolarization pulse, with a subthreshold pulse duration = *asymmetric exponentially decreasing biphasic pulse*

natural activation. In the natural case each single fiber is only activated 0.3–5 times per second allowing sufficient recovery time for recovery and energy uptake. In FES the stimulation frequencies that are normally used are in the range from 20 to 100 Hz, depending on the application.

Stimulation amplitude

Changes in the stimulation amplitude or pulse duration cause similar effects on the elicited muscle response. Higher stimulus intensities produce higher contractions through activation of more fibers, which can be measured as higher joint torques. However, there is a lower threshold under which the stimulus produces no response, and also an upper threshold of saturation where no additional fibers can be recruited. Additionally, the muscle response between the two thresholds is not linear; as it has been described before, the electrically-induced response of the muscle is created by the activation of several muscle fibers whose activation threshold depends on their diameter. Smaller myelinated nerves have closer nodes of Ranvier, which means that to elicit an action potential using FES we would need a higher voltage gradient (thus, a higher electric field and a higher current). Finally, the distance between the

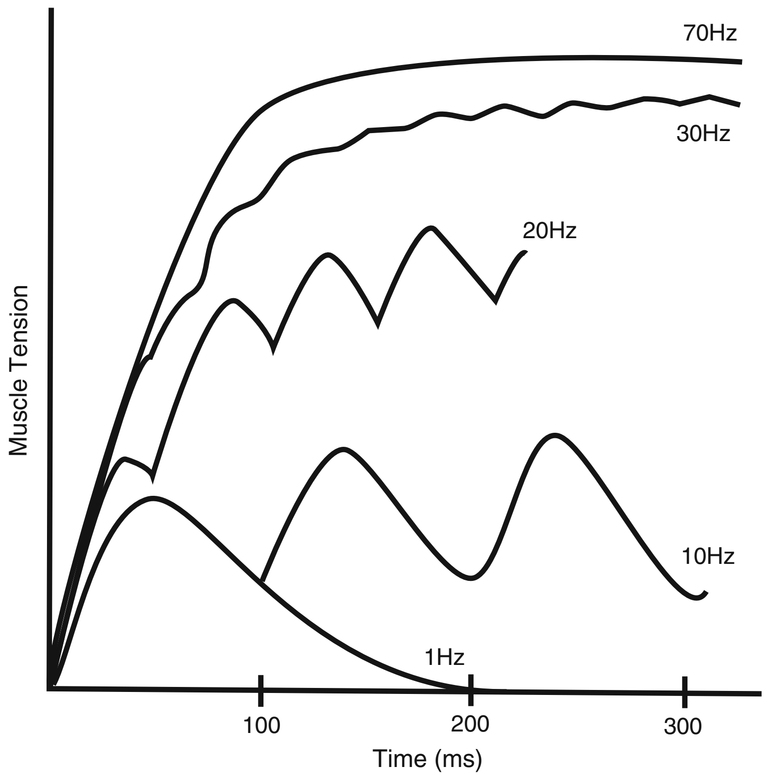

Fig. 2 Effects of stimulation frequency on muscle tension (Adapted from [3])

muscle fibers and the skin also plays a fundamental role, since the fibers closer to the electrode are reached by a stronger electric field.

It could be shown in [3] that due to afferent feedback in stroke subjects higher torques in triceps muscles could only be generated with shorter pulses, as a longer pulse duration caused antagonistic muscle responses of the biceps muscles. The curves in Fig. 3 represent the pairs of intensity current and pulse-width that elicit the same motor response in a subject, and as a rule of thumb, larger pulse-widths need to be paired with lower currents in order to obtain the same motor responses. However, there is a saturation for the pulse-width that prevents the muscle from producing a stronger contraction, even when the current amplitude is maintained and the pulse width is increased. The stimulated motor response in healthy subjects is in between the innervated motor threshold and the maximum motor response; therefore, the dotted rectangle represents the range that is normally used in FES applications. For example, for wrist extension the typical pulse widths vary from 40 to 300 μs, whereas the current amplitude is usually set up between 10 and 40 mA. However, it is very dependent on the subject pain threshold and the muscle that has to be activated. The other extreme is completely denervated muscles, e.g., in spinal cord injured patients.

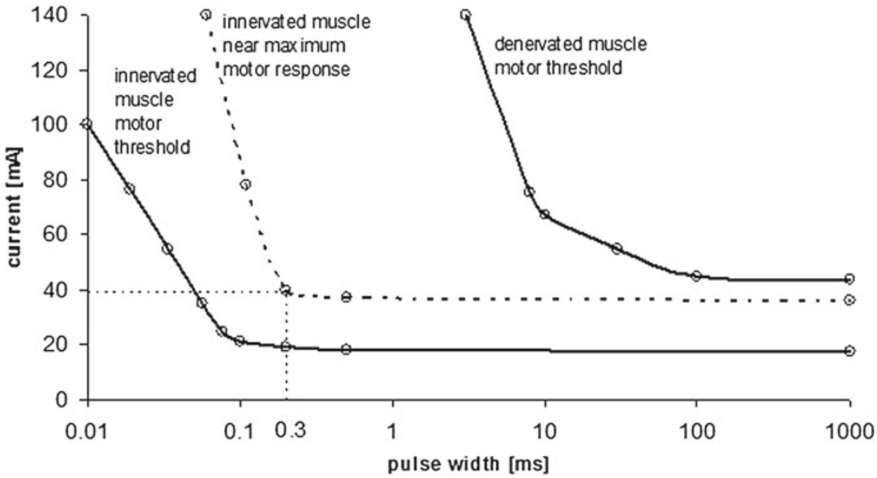

Fig. 3 Curves of equal motor response for different pulse widths and pulse amplitudes. For intact motor neurons stimulation pulses longer than 300 μs do not increase the motor response if the stimulation amplitudes are higher than 40 mA. Denervated muscles require a 100–1000 times longer stimulation pulse-widths than innervated muscles (Data from wrist extensor, surface stimulation, stimulation frequency 35 Hz). Adapted from [3]

Those muscles are no longer connected to the spinal cord (e.g., a cauda equina lesion) and get completely denervated. In such cases, it is possible to produce, after a long training period of weeks to months, muscle contractions by direct stimulation of the motor end plate instead of the nerve. However, the pulse-widths that must be used have to be 100 times larger. Here, one cannot increase the amplitude instead, as the high currents would heat the tissue to unaccepted levels.

Electrode placement

Two main aspects must be considered for the electrodes placement: the position and the distance between anode and cathode.

In order to elicit muscle responses, the active electrode (cathode) must be placed on the muscle belly, close to the innervating nerve or anywhere else where the innervating nerve is close to the surface. Placing them on the muscle belly is easier, especially in the case of the lower limbs, where the muscles are bigger and the electrodes must simply be placed along the fiber. Alternatively, the stimulation of the innervating nerve at some other location (where this nerve is close to surface) has the advantages that stimulating deeper lying muscles can be reached. Often smaller electrodes are used in this case, which sometimes can have the drawback of making the subject feel discomfort. The area that can be found to stimulate a specific muscle is called the motor point of this muscle. Some regions of the body have several motor points very close, like the arm or the hand. For such regions, the best option is to use multiple small electrodes as proposed with electrode arrays.

The second aspect that plays an important role is the distance between the anode and cathode. Considering the tissue to be homogeneous, the same electrical potential difference between the electrodes will create a stronger and more superficial electrical field if the electrodes are closer. This electrical field will activate only the superficial fibers, and with a very high intensity, which can be useful for small muscles (like the arm muscles). On the other hand, bigger muscles (like for instance the rectus femoris) require a higher distance between electrodes, where the electrical field will reach more fibers with a lower intensity. However, there is a drawback for increasing the distance between electrodes: as the electrical field spreads out over the tissue, it can activate other muscles, including the antagonists if the distance is high enough.

Electrode size

The choice of the size of the electrode (i.e., the size of their active area) depends on the dimension of the muscle or nerve that wants to be activated. As a rule of thumb, larger electrodes will reach deeper nerves, thus achieve overall stronger contractions but have a lower muscle selectivity. Small electrodes can result in very high current densities, which can lead to skin irritations or burns. Generally, small electrodes provide a better selectivity, however limit the penetration depth. The size of the electrodes that are normally used ranges between 2 and 50 cm^2; the smallest ones are used for stimulating superficial nerves, those of 6–10 cm^2 are indicated for smaller muscles, whereas electrodes bigger than 25 cm^2 or more are designed for larger muscles or dermatomes [1].

2.3 Modeling FES

The recent development of more selective FES technologies has created the need to better understand the properties of the nerve, bone, muscle, fat, and skin, and how they react when external electrical stimuli are applied. In this scope, simulations of body segment volume conductor models have been used to determine which structures are important to be considered and how stimulation parameters need to be optimized for more efficient stimulation of neural structures. This second aspect requires a two-step model, the volume conductor with all relevant physical properties and a nerve model that determines when the neuronal structure is activated. Simpler, more classical modeling of effects of electrical stimulation to tissues can be done with electrical circuit models, in case the physical structures and anatomical properties are not important.

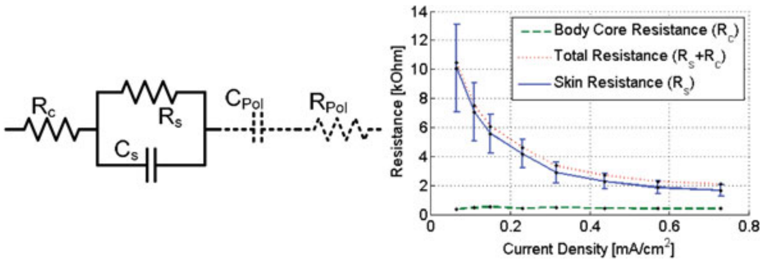

Fig. 4 Equivalent electrical model for the biological tissues (Reproduced from [11])

2.3.1 Models Used for Describing the Electrical Effects of Electrical Stimulation

Both modeling approaches characterize the full dynamics that connects the stimulation current to the nerve recruitment. In this section we first describe simplified one-dimensional electrical models and then the combination of volume conductor models with nerve activation models.

Simplified electrical models

The simplest way of modeling the skin and underlying tissues is by assuming their permittivity and resistivity is homogeneous and hypothesizing the propagation of the current impulse to occur only along one axis. This leads to the equivalent electrical model that is shown in Fig. 4, consisting on a resistor representing the fat and the muscle (Rc) in series with a RC parallel circuit representing the skin (Cs, Rs), and an additional capacitor and resistance that model the electrode polarization effects, which can be neglected for surface FES. The values of the body-core resistance, and the skin resistance and capacitance can be experimentally determined by current–voltage responses measured on the skin and with needle electrodes within the muscle tissue. Results show that the value of the body-core resistance is independent of the current density, whereas the skin resistance decreases when the stimulation amplitude increases using a specific stimulation electrode size (see Fig. 4). Hence, the skin impedance is a non-linear function of the current density [6].

Volume conductor models

A full characterization of the three-dimensional propagation of the stimulation current and the induced voltage gradients requires a more advanced model that includes more complex volumetric models, where current and voltage can have arbitrary directions. Recent studies have used a finite element (FE) model to simulate the distribution of the potential inside the biological tissues, i.e., the skin, fat, muscle and bone layers

[9]. These types of models can work with detailed geometries (reconstructed from MRI scans), are able to deal with inhomogeneities (such as sweat ducts and glands), and can simulate specific stimulation pulse waveforms, electrode positions and sizes. The equations that are used to model the propagation of the electrical field can be static or quasi-static or transient (including the inductive effects and the times of wave propagation). Additionally, each tissue layer can be defined in terms of geometry and structure, and its electrical properties can also be accurately specified (permittivity, conductivity). The inclusion of the dielectric properties of the tissues was introduced by Kuhn and colleagues [9], proving that the differences in the dielectric properties between tissue layers create capacitive effects that are not negligible for the skin. Their method obtained very good comparisons with experimental measurements on healthy volunteers.

Nerve models

The second step of modeling tissue excitability focuses on the transmembrane potentials of the nerve axons, which depend on the extracellular potentials that are calculated using the FE model of the conductor volume. The nerve model contains thousands of myelinated axons that can be activated using electrical current pulses, generating action potentials that activate the muscle fibers [12]. The main information that can be extracted from the nerve models is which factors are important in the design of the optimal transcutaneous electrical stimulation.

2.3.2 Important Factors for Nerve Activation

The analysis and simulations with these two-step models can be used to extract quantitative relations between the several factors and the stimulation outcome. The main factors that have been investigated with respect to the volume geometry were thicknesses, nerve depth, inhomogeneities, electrode position, electrode size, properties of the different layers (permittivity, resistivity), modeling equations (transient or quasistatic, including the pulse amplitude and duration), and the type of nerve model (axon diameter and diameter distribution) [9, 11, 12]. The parameters that have a stronger influence in transcutaneous electrical stimulation are the nerve depth, the resistive properties of the muscles, the axon diameters, and the axon diameter distribution. In addition, the stimulation parameters (electrode position and size, pulse-width, and amplitude) must be carefully considered when designing a FES application.

2.3.3 Selectivity and Comfort of Electrodes

The combination of the volume conductor and the nerve models is leading to a full understanding of the dynamics that connect the transcutaneous stimulation parameters with the response of the nerves and fibers. This has opened a research line that

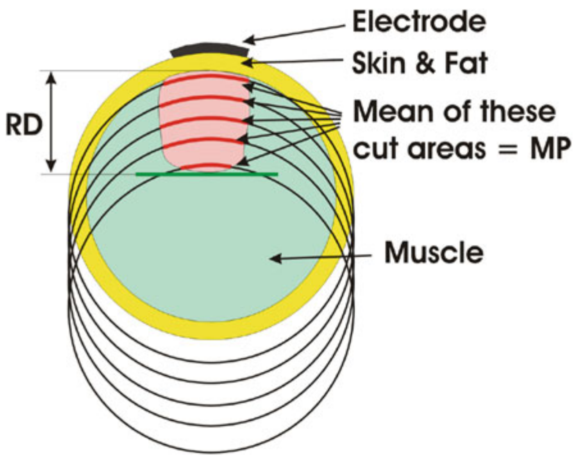

Fig. 5 Definition of the Recruitment Depth (*RD*) and Mean Profile (*MP*) (Reproduced from [10])

aims at optimizing the FES applications, which is translated into stimulating muscles with the highest selectivity, while causing the lowest discomfort to the subject. However, reaching both goals simultaneously is not simple. The two main factors that influence selectivity and comfort are the electrode size and the electrode materials. The electrode is usually built by stacking two layers of conductive hydrogel, a conductive substrate (made of fabric or carbon film) and a substrate. This construction provides a distributed current density over the electrode surface, prevents the skin from burning and guarantees a low electrode-substrate-skin impedance in order to excite the minimum number of afferent nerves [7]. Therefore, when designing a FES application, it is much easier to reach the selectivity and comfort specifications by changing the electrode size. The size of the electrodes is usually chosen in function of the stimulation site, depending on the size of the muscle that wants to be contracted and the proximity of the adjacent muscles that want to be avoided. Smaller electrodes produce more selective activations; however, it might be difficult to activate deep muscles as they generate high current densities that could cause discomfort to the subject [13]. The optimal electrode size can be assessed by combining transcutaneous stimulation FE models (that will establish the selectivity) and experiments in subjects (who will characterize the perceived comfort). The electrical field that is created by the electrodes will penetrate the skin with a certain recruitment depth and width, shaped as a lobe in a cross-sectional view of the muscle (See Fig. 5).

Therefore, the selectivity can be defined as proportionally dependent on the Recruitment Depth (RD) and inversely related to the area that is affected by the lobe, i.e., the Mean Profile (MP) [10, 13]:

$$\text{Selectivity} = \text{RD}/\text{MP}$$

On the other hand, the estimated comfort can be assessed as a combination of the pain threshold (ThePain) and the current density of the electrode (CD). None of these

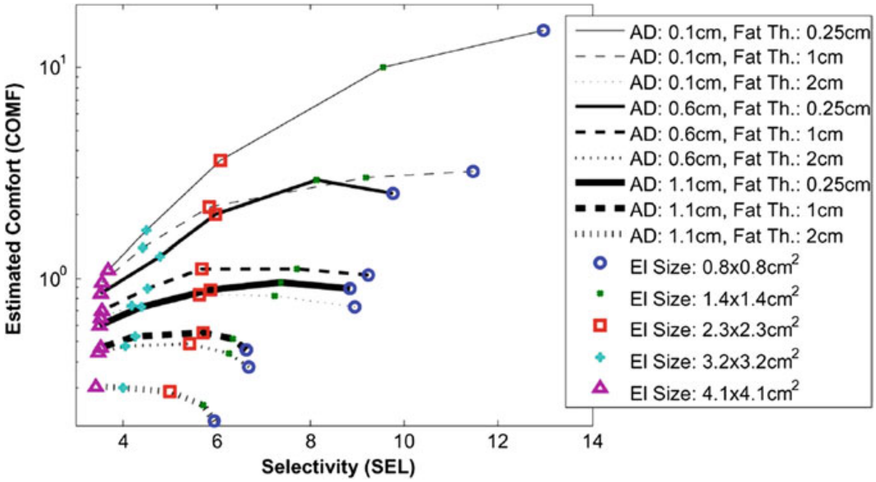

Fig. 6 Comfort versus selectivity, for a various set of Activation Depths (*AD*), Fat Thicknesses (*Fat Th*), and Electrode Sizes (*El Size*) (Reproduced from [11])

parameters can be measured directly; therefore, it has been suggested in literature to measure the perceived pain threshold as the average on ten healthy subjects, and to calculate the current density by dividing the amplitude of the stimulation by the area of the electrode [13]:

$$\text{Comfort} = \text{ThePain}/\text{CD}$$

A recent study in literature [13] simulated the selectivity and comfort for a set of electrode sizes, activation depths, and fat thicknesses (Fig. 6). Results show that both selectivity and estimated comfort are strongly influenced by the three factors, and they all play a role in the muscle response. This study also indicates that smaller electrodes produce more selective muscle activations and tend to cause less discomfort in the subject when the current density is kept constant. However, the estimated comfort reaches a maximum that does not always correspond to the smaller electrode. In fact, for high activation depths and thick fat layers, smaller electrodes can be less comfortable than larger electrodes. The reason for all this is twofold: electrical stimulation is more tolerable when it is felt on a smaller portion of the skin (less receptors are activated), and the current with smaller electrodes (at a specific current density) penetrates less deep into the tissues than with a larger electrode.

3 Upper Limbs—Practical Guide

Upper limbs are involved in a countless amount of activities of daily living (ADL) that include reaching, grasping, and manipulating objects.

This section aims to be a practical introduction to transcutaneous FES on upper limbs, where practical exercises are suggested in order to experimentally prove the basic concepts described in the previous section and check at first hand the main challenges related to transcutaneous FES application in upper limbs.

General notes In order to use transcutaneous functional electrical stimulation in efficient and safe manner it is necessary to obey certain elementary rules.

As explained in the introduction, FES is achieved by applying electrical current to the muscle or nerve fibers through electrodes placed on the skin surface. It is important that the surface of the electrode is conductive, adhesive, and has good contact with skin along the entire surface. Most commercial electrodes used today have a layer of conductive gel which ensures these properties. In any case, before placing the electrodes one should check if the gel is moist and adhesive, as dried up electrode can cause discomfort or pain during stimulation.

When starting the stimulation it is advised to start with the lowest recommended amplitude and gradually increase it until the required response is achieved.

3.1 Upper Arm

Upper arm muscles act mainly at shoulder and elbow joints; therefore, the main functional task carried out by these muscles is the reaching task.

3.1.1 Anatomy

Anterior and posterior views of arm muscles are shown in Fig. 7. However, for the following exercises we will briefly describe the prime movers and those muscles located distally starting at the shoulder.

Deltoid is one of the most important muscles involved in shoulder motion. When its lateral or all fibers are active, deltoid causes abduction of the arm, but it is also responsible for flexion/extension of the arm when its anterior/posterior fibers are activated isolatedly. This muscle is innervated by the axillary nerve. Biceps and brachialis are prime movers of elbow flexion, and are innervated by the musculocutaneous nerve. Finally, triceps muscle is the prime mover of elbow extension and is innervated by the radial nerve. In Fig. 8 we can see an scheme of the peripheral nervous system relevant to the arm.

3.1.2 Exercises

The aim of the following exercises is to experiment basic application of FES on the upper arm and see the effects of location of electrodes and parameters regarding motor outcome and discomfort. As the aim is to generate a contraction of a big

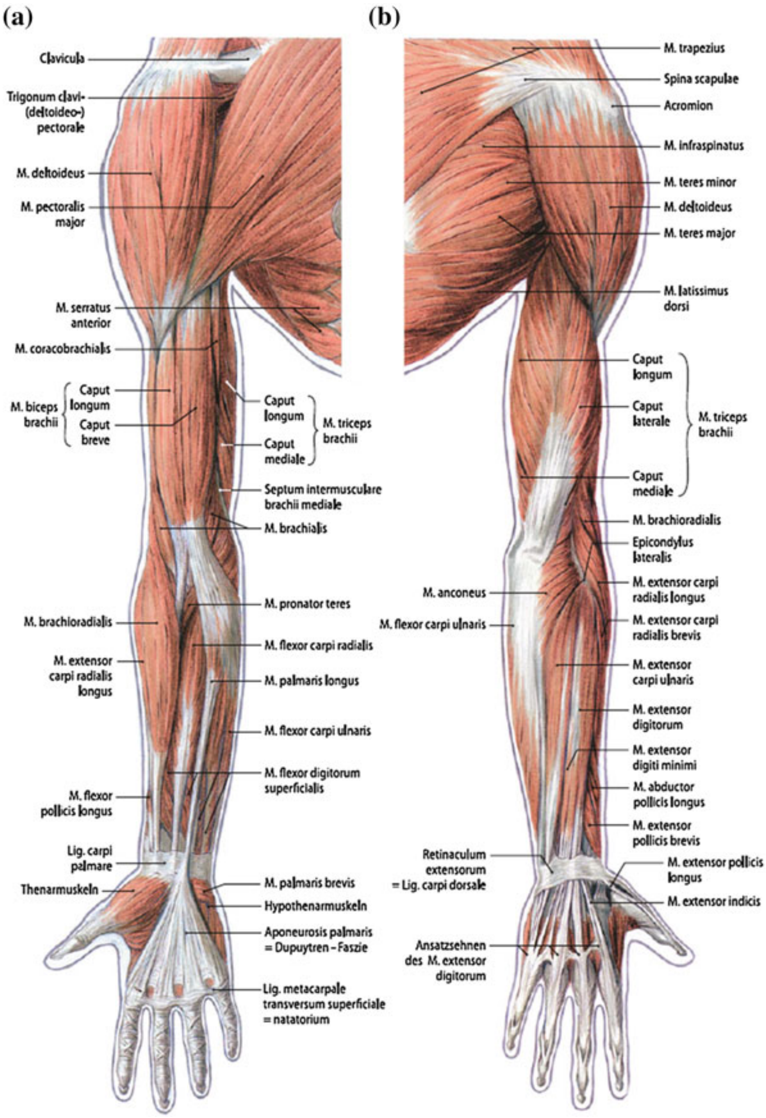

Fig. 7 **a** Anterior view of arm muscles. **b** Posterior view of arm muscles [17]

muscle, in these exercises electrodes are placed over the top and bottom sides of the targeted muscle belly (along the muscle) with enough distance between them to ensure that the current goes deep enough to reach the motor nerve fibers and generate an action potential [3].

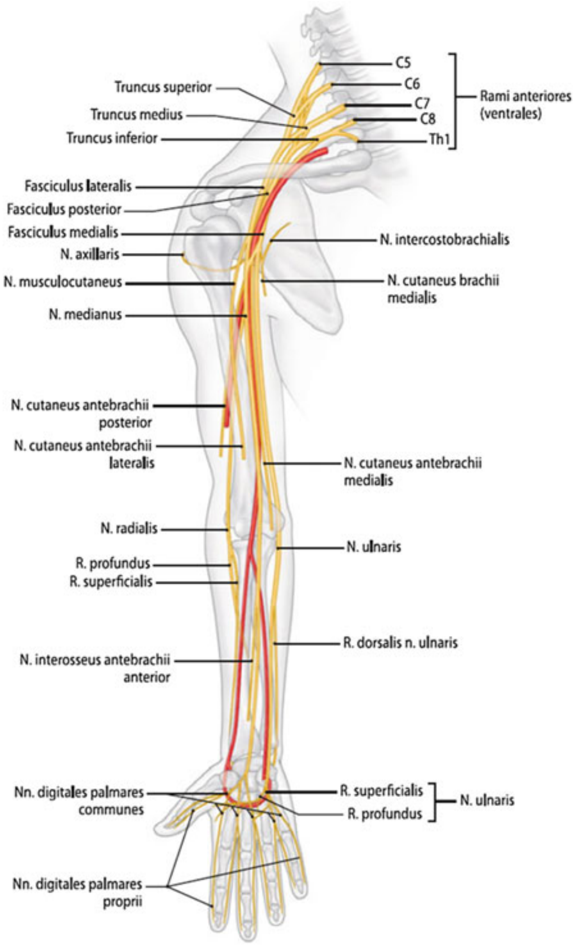

Fig. 8 Scheme of upper arm nerves [17]

If the exercises are carried out in group, the high inter-subject variability in terms of sensation (amplitude tolerance) and motor response to FES application will be observed.

Arm movements at the shoulder are primarily performed by the deltoid muscle, as previously described. In this exercise we will see that flexion, extension, and abduction of the arm at the shoulder can be achieved by acting in the different fibers that form the deltoid. Figure 9 shows some examples of electrode placement for different movements of the arm. Parameters used during the the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 25 Hz
- Pulse-width: 200 µs
- Amplitude: upon tolerance

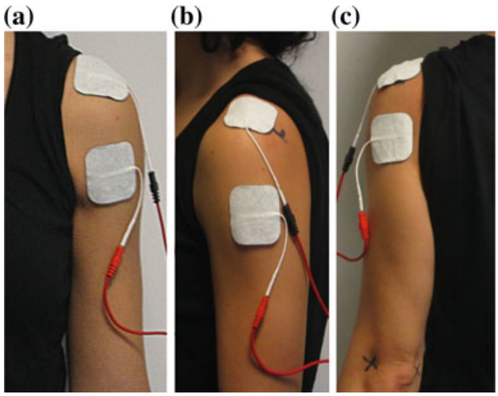

Fig. 9 **a** Example of electrode position for shoulder flexion. **b** Example of electrode position for shoulder abduction. **c** Example of electrode position for shoulder extension

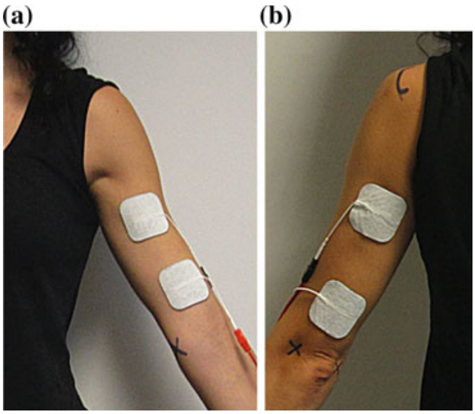

Fig. 10 **a** Example of electrode position for elbow flexion. **b** Example of electrode position for elbow extension

After performing stimulation of the deltoid muscle with different electrode placements, we should be able to see the different arm movements generated when applying FES to motor nerves that innervate different muscle fibers.

Elbow flexion is mainly operated by the muscles located in the anterior part of the upper arm. The objective of this exercise is to get an elbow flexion and observe the effects of changes in pulse-width on outcome and discomfort. An example of electrode placement for elbow flexion is shown in Fig. 10. Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 25 Hz
- Pulse-width: 100–300 μs
- Amplitude: upon tolerance

After applying FES to the anterior part of the arm with different pulse-width values, we should be able to see that muscle contractions are stronger when higher pulse-width values are used and, on the contrary, higher amplitudes are needed to

achieve a motor response when lower pulse-width values are used. This fact proves the concept shown in Fig. 3 and explained in Sect. 2. Regarding discomfort, in general the longer the pulse-width is, the more uncomfortable the stimulation becomes.
Elbow extension is performed by the triceps muscle, located on the posterior part of the upper arm. The objective of this exercise is to get an elbow extension and observe the effects of changes in frequency on outcome and discomfort. An example of electrode placement for elbow extension is shown in Fig. 10. Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 5–50 Hz
- Pulse-width: 200 μs
- Amplitude: upon tolerance

After applying FES to the posterior part of the arm with different frequency values, we should be able to see that the outcome to low frequencies are short muscle twitches or fasciculations, whereas we get tetanic muscle contractions as we go higher in frequency values. This fact proves the concept shown in Fig. 2 and explained in Sect. 2. Regarding superficial discomfort, in general, high frequencies (35–50 Hz) are more comfortable than middle frequencies (20–35 Hz), although they carry a big disadvantage in terms of fatigue that will be discussed more in detail in Sect. 4.

3.2 Forearm and Hand

Forearm and hand muscles are mainly involved in wrist, finger, and thumb movements. Therefore, the main functional tasks carried out by these muscles are the grasp and manipulation of objects.

Although a healthy human is able to grasp precisely different types of objects in a great variety of manners without a special effort, it is an extremely complex task that is the result of a combination of perfectly synchronized muscle contractions. The objective of these exercises is to explore the effects of the application of transcutaneous FES onto the forearm while trying to achieve a functional grasp by combining different wrist, finger, and thumb movements.

3.2.1 Anatomy

Most forearm muscles act at wrist, finger, or thumb joints. The forearm muscles that generate finger and thumb movements are known as the extrinsic muscles of the hand, which arise in the forearm and insert into the digits via long tendons crossing the wrist. The intrinsic muscles of the hand, which arise and insert within the hand, are involved in finer finger and thumb movements where thumb is controlled by the thenar muscles. However, for practical reasons, in most applications transcutaneous FES is limited to stimulation of the extrinsic muscles of the hand. Figure 7 shows anterior and posterior superficial views of arm muscles.

Most muscles located on the anterior part of the forearm generate flexion movements at the wrist and fingers and are mainly innervated by median or ulnar nerve. Conversely, muscles located on the posterior part of the forearm generate extension movements at the wrist and fingers and are innervated by the radial nerve and its branches. Although there are forearm muscles that act on the thumb, thenar muscles are the prime movers of thumb abduction and opposition, which are innervated by median or ulnar nerves. In Fig. 8 we can see an scheme of the peripheral nervous system relevant to the arm.

3.2.2 Exercises

The objective of the following exercises is to explore the effects of the application of transcutaneous FES onto the forearm and try to achieve selective movements by finding different motor points.

It is important to point out that the neuromuscular system that controls hand movements is extremely complex, with numerous muscles located at different depths from the skin surface and innervated by branches of the same nerve. Therefore, small electrodes are suggested for finding independent motor points. We call motor points to those skin surface points where, when transcutaneous FES is applied, a determined and, if possible, isolated muscle contraction is obtained. When muscles are deep or are located under other muscles, it can be very difficult or impossible to activate them without activating neighboring motor nerve fibers as well.

For compensating this selectivity issue, a system with multi-field electrodes was used for these exercises [21, 26]. Two regular electrode matrices of 16 small fields each were used, one of them was placed covering the extensor muscles and the other one was placed covering the flexor muscles. One big common anode was used for each electrode matrix, which was located over the tendinous area of the wrist in its respective side. Electrode configuration is shown in Fig. 11. This approach of small cathode matrix and big common anode is used in order to generate an action potential in a precise location of the forearm without generating any at the wrist.

It can be that certain movements are not achieved in the following exercises in some subjects and it is due to inter-subject physiological variations. Relative position of muscles, nerves, and skin changes slightly among individuals, which can make it difficult to find some motor points in some subjects. If the exercises are carried out in a group, the high intersubject variability in terms of sensation (amplitude tolerance) and motor response to FES application will be observed once more.

Parameters used in the following exercises should be:

- Waveform: biphasic symmetrical
- Frequency: 25 Hz
- Pulse-width: 200 μs
- Amplitude: upon tolerance

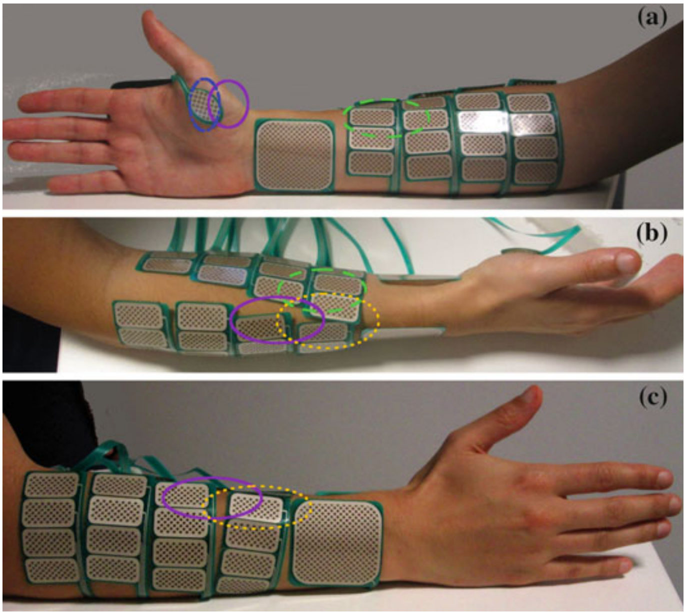

Fig. 11 Approximated areas where motor points are expected to be found. Each color aims at a different thumb movement. *Orange dotted*—extension, *purple solid*—abduction, *blue dash-dotted*—opposition, *green dashed*—flexion. **a** Anterior view of thumb motor points. **b** Radial view of thumb motor points. **c** Posterior view of thumb motor points

Thumb movements are controlled by muscles located both on the forearm and on the palm. In this exercise, we will try to find motor points that generate thumb extension, flexion, abduction, adduction, and opposition. Figure 11 shows areas where the corresponding motor points are usually found.

After applying FES on forearm and palm with different electrode placements, we should be able to see the different thumb movements generated when applying FES to the different motor points.

Wrist and finger flexors are located in the anterior part of the forearm. In this exercise, we will try to find motor points that generate combined wrist and finger flexion as well as isolated wrist flexion and finger flexion. Figure 12 shows areas where the corresponding motor points are usually found.

After applying FES on the anterior forearm with different electrode placements, we should be able to see the different wrist and finger movements generated when applying FES to the different motor points.

Wrist and finger extension is controlled by the muscles located in the posterior part of the forearm. In this exercise, we will try to find motor points that generate combined wrist and finger extension as well as isolated wrist extension, index extension, and finger extension. Figure 13 shows areas were the corresponding motor points are usually found.

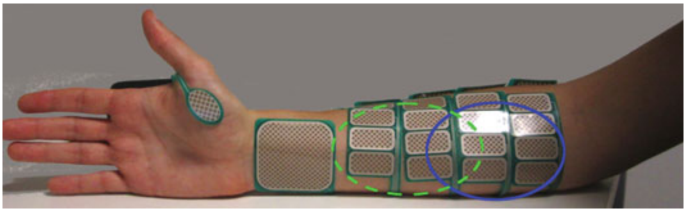

Fig. 12 Approximated areas where motor points are expected to be found. Each color aims at a different flexion. *Blue solid*—wrist flexion, *green dashed*—finger flexion

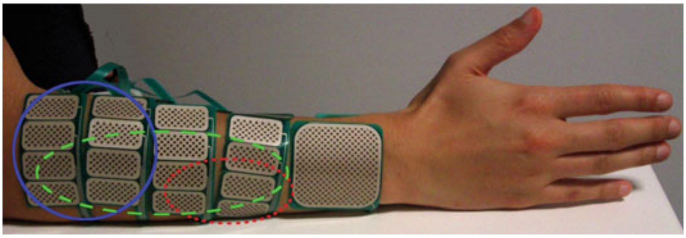

Fig. 13 Approximated areas where motor points are expected to be found. Each color aims at a different extension. *Blue solid*—wrist extension, *green dashed*—finger extension, *red dotted*—index extension

After applying FES on the posterior forearm with different electrode placements, we should be able to see the different wrist and finger movements generated when applying FES to the different motor points.

Grasp sequences Many grasp classifications have been suggested in literature, but most taxonomies distinguish at least three main grasping postures, which are palmar or power grasp, pitch or precision grasp, and lateral grasp [14]. In these exercises we play around to find sequences of activations in order to achieve different grasps. The objective is to achieve a palmar grasp (bottle), a lateral grasp (credit card), and a precision grasp (very small object). Quick design of simple sequences is suggested to be designed by a trial-and-error method, where motor points are manually activated and deactivated. An example of a simple sequence for the palmar grasp is presented as follows:

Palmar grasp sequence *Activate wrist extensors⟶Activate finger extensors⟶ Activate thumb abductor⟶Deactivate finger extensors⟶Activate finger flexors⟶Deactivate thumb abductor*

4 Lower Limbs—Practical Guide

In this section, we will describe practical guidelines for applying functional electrical stimulation for activation of various muscles in lower limbs in order to actuate different joints. Through these exercises we will get familiar with the challenges present in this field, such as muscle fatigue and electrode placement. These challenges, as well as techniques for overcoming them will be discussed.

General notes In order to use transcutaneous functional electrical stimulation in efficient and safe manner it is necessary to obey certain elementary rules.

As explained in the introduction, FES is achieved by applying electrical current to the muscle or nerve fibers through electrodes placed on the skin surface. It is important that the surface of the electrode is conductive, adhesive, and has good contact with skin along the entire surface. Most commercial electrodes used today have a layer of conductive gel which ensures these properties. In any case, before placing the electrodes one should check if the gel is moist and adhesive, as dried up electrode can cause discomfort or pain during stimulation.

When starting the stimulation it is advised to start with the lowest recommended amplitude and gradually increase it until the required response is achieved.

4.1 Upper Leg

Upper leg muscles are the actuators of knee joint, and they participate in the actuation of the hip. In this section we will focus on the functionality of the knee.

Knee flexion is actuated by hamstring muscles; semimembranosus, biceps femoris and semitendinosus, all innervated by the sciatic nerve. In order to activate these muscles, large electrodes are placed on the posterior side of the thigh, as shown in Fig. 14.

Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 50 Hz
- Pulse-width: 300 μs
- Amplitude: 20–45 mA (upon tolerance)

Fig. 14 Approximated areas where motor points for knee flexion are expected to be found

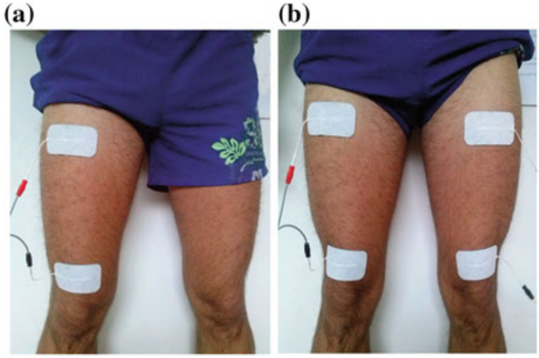

Fig. 15 Approximate electrode placement for **a** knee extension and **b** fatigue exercise

The subject should lie on the belly with his legs laid freely. When the stimulation starts, and the activation threshold is reached, the knee of the stimulated leg should start to flex, lifting the foot and the lower leg.

Knee extension is actuated by the quadriceps muscles; rectus femoris, vastus medialis, vastus intermedius, and vastus lateralis. These are innervated by the femoral nerve. In order to activate these muscles, large electrodes are placed on the anterior side of the thigh, as shown in Fig. 15a.

Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 50 Hz
- Pulse-width: 300 μs
- Amplitude: 20–45 mA (upon tolerance)

The subject should sit on a high chair with his legs hanging freely. When the stimulation starts, and the activation threshold is reached, the knee of the stimulated leg should start to extend, lifting the foot and the lower leg.

Influence of stimulation frequency on muscle fatigue can be easily demonstrated in a setup similar to the one in the previous exercise, but with both legs stimulated at the same time. Electrodes should be placed on each leg, as described in the knee extension exercise and shown in Fig. 15b. Two stimulators should be used, each connected to the electrode pair on one leg. Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 25 Hz left leg, 80 Hz right leg
- Pulse-width: 300 μs
- Amplitude: 20–45 mA (upon tolerance). It should be set individually to produce the same response as in the previous exercise.

When the parameters are set, both stimulators should be activated at the same time. Initially both legs should rise, the same way as in the previous exercise.

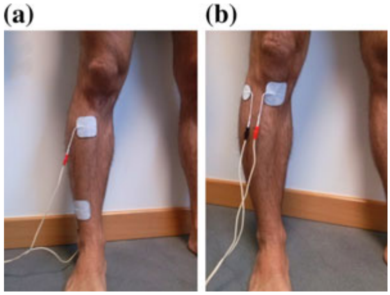

Fig. 16 Approximate electrode placement for **a** muscle belly stimulation and **b** peroneal nerve stimulation

After a short time, the fatigue in the right leg will become noticeable. First it will start to oscillate and then it will drop, while the other leg will remain lifted.

After a few minutes of rest, the stimulators should be switched to perform a control measurement. The effects should be the same, just mirrored.

This effect represents one of the main challenges in this field, and is discussed in more detail in further text.

4.2 Lower Leg

Lower leg muscles are the actuators of the ankle joint.

Ankle dorsal flexion is actuated by the tibialis anterior muscle, which is innervated by the deep peroneal nerve. Two methods for ankle dorsiflexion are described: direct stimulation of the muscle and stimulation of the peroneal nerve.

Stimulation at the belly of the muscle In order to activate this muscle, medium-size electrodes are placed laterally on the anterior part of the leg. Anode is placed distally, closest to the ankle, while the cathode is placed on the belly of the muscle, as shown in Fig. 16a. The belly of the muscle is located by palpation during several voluntary contractions. Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 50 Hz
- Pulse-width: 300 µs
- Amplitude: 15–35 mA (upon tolerance)

Stimulation of the peroneal nerve The biggest challenge in achieving dorsal flexion by stimulation of the peroneal nerve is cathode placement. The cathode should be a small size electrode placed near the head of the fibula, where peroneal nerve is superficially located. Anode should be a medium-size electrode, placed over or under the knee, as shown in Fig. 16b. It might be necessary to perform several adjustments of cathode position, in order to achieve good flexion. Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 50 Hz

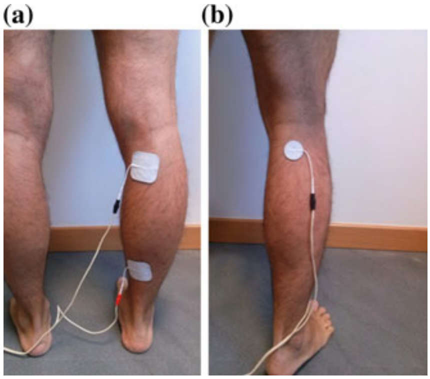

Fig. 17 Approximate electrode placement for **a** muscle belly stimulation and **b** peroneal nerve stimulation

- Pulse-width: 300 μs
- Amplitude: 7–15 mA (upon tolerance)

Even though this type of stimulation is more complicated from the electrode placement perspective, it is preferable because the muscle is excited in a more physiological manner, which postpones fatigue and requires less external energy (lower pulse amplitudes).

Ankle plantar flexion is mostly actuated by the gastrocnemius and soleus muscles, which are both innervated by the tibial nerve. Two methods for ankle plantar flexion are described: stimulation at the belly of the muscles and stimulation of the tibial nerve.

Stimulation at the belly of the muscle In order to activate this muscle, medium-size electrodes are placed medially on the posterior part of the leg. Anode is placed distally, closest to the Achilles tendon, while the cathode is placed on the belly of the gastrocnemius muscle, as shown in Fig. 17a. The belly of the muscle is located by palpation during several voluntary contractions. Parameters used during exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 50 Hz
- Pulse-width: 300 μs
- Amplitude: 15–35 mA (upon tolerance)

Stimulation of the tibial nerve The cathode should be a small size electrode placed on the posterior part of the leg, near the kneepit (popliteal fossa), where tibial nerve is superficially located, as shown in Fig. 17b. Anode should be a medium-size electrode, placed over or under the knee (as shown in Fig. 16b). It might be necessary to perform several adjustments of cathode position, in order to achieve good plantarflexion. Parameters used during the exercise should be:

- Waveform: biphasic symmetrical
- Frequency: 50 Hz

- Pulse-width: 300 μs
- Amplitude: 7–15 mA (upon tolerance)

Even though this type of stimulation is more complicated from the electrode placement perspective, it is preferable because the muscle is excited in a more physiological manner, which postpones fatigue and requires less external energy (lower pulse amplitudes).

5 Conclusions

Transcutaneous FES applications have been developed over more than 50 years with some success, however, limited to a few applications. In the clinical and sports environment, the technique is in wide use for non-selective muscle training, muscle endurance and relaxation and the treatment of pain complementary to TENS. One of the few successful applications of FES is the drop-foot stimulator, a device that helps to improve foot clearance during swing phase. In recent years, a new approach to address the main disadvantage of FES, the need to place the electrodes precisely by a skilled and experienced user has brought new attention to the field. A new multichannel stimulation electrode technology using arrays of electrode pads within a single physical electrode required new investigations of electrode properties, current pathways in human tissues, and the resulting activations of neural tissues. Models with physiological properties have been developed and used to simulate the current pathways and related nerve activations in inhomogeneous and realistic tissue configurations.

5.1 Modeling

Modeling of transcutaneous electrical stimulation that combines volume conductor and nerve models has proven to be a very useful tool to provide a full understanding on the activation dynamics of the musculoskeletal system, and a valuable information source to be considered when designing transcutaneous electrical stimulation applications. The simulation results obtained with such models could estimate the dominant parameters in the transcutaneous stimulation; these are the muscle resistivity, the electrode–axon distance, and the axon diameter. Additional factors mainly concerning the stimulation waveform (frequency, pulse-width, amplitude) nevertheless are also crucial and should be carefully selected depending on the application. The size of the electrodes has a strong influence on the selectivity and the comfort. The stimulation of superficial nerves has proven to be more comfortable when using small electrodes of about $1 \times 1\,\text{cm}^2$, however, larger electrodes are the better option as the activation depth increases. Finally, the relation between selectivity and electrode size is purely monotonic; the smaller the electrodes, the better the performance at targeting single and small superficial muscles.

5.2 Precision and Selectivity

As described before, one of the main challenges that affects mainly to upper limb applications is the selectivity issue. Simply analyzing an anatomy book we can realize how complex the upper limb system is, composed by around 50 muscles innervated by different branches of different motor nerves corresponding to the peripheral nervous system [17]. This redundant system allows us to reach, grasp, or manipulate objects in a wide variety of ways for the same task. Although unconsciously humans optimize the muscle activation patterns in order to perform the task successfully minimizing the needed energy, controlling this complex system with transcutaneous FES is challenging and we could point out three main reasons. First of them is the difficulty of achieving precise control of joints. Taking the reaching task, as an example, biomechanical models show that the human arm has 7 degrees of freedom (DOF) used to reach objects on the 3D space [24]. Although there could be ways of controlling 7 DOF for reaching precisely certain points in space by means of transcutaneous FES, it is extremely difficult to control each of these joints precisely enough to expand this success to any point in space. This is due to the reverse recruiting order of muscles [4, 25], where we activate big strong muscle fibers before smaller muscles responsible for precise movements, causing difficulties when fine-tuning control of reaching is needed. The second reason why controlling upper limbs for functional tasks is challenging is the issue of selectivity. Hand movements are a result of complex contraction patterns of muscles located both at the forearm and hand. As transcutaneous FES is applied over the skin surface, current is spread over different tissues before reaching the targeted motor nerve fiber, which becomes a problem when we pretend to apply it in a structure of small muscles, located at different depths, one on top of the other, and innervated by branches of the same nerve, like forearm muscle structure. As explained in Sect. 2, several factors affect the current distribution when FES is applied on the body surface and it is difficult to find isolated motor points or to generate action potentials in certain motor nerve fibers without acting in the neighboring nerve fibers as well [12]. Finally, another big challenge related to the application of transcutaneous FES on upper limbs is the high inter-subject and intra-subject variability in terms of motor response to FES, which makes it unlikely to find a unique solution for every user. At present, selectivity issues related to surface FES are being slightly overcome with the use of multifield electrodes, which bring diverse benefits like reduction of number of cables, improving selectivity or possibility of automatically determining customized stimulation patterns for each user, among others. [16, 20, 21].

5.3 Fatigue

As mentioned before, functional electrical stimulation can quickly lead to metabolic muscle fatigue. In normal conditions, muscle fatigue appears after prolonged intensive muscle activity, and is essentially a state in which the ability of muscle to generate force deteriorates. This can be caused either by inability of the nerve to produce excitation signals (neural) or by the inability of muscle fibers to contract (metabolic).

Electrical stimulation leads to metabolic muscle fatigue. Particularly when trying to achieve functions which require high force output from large muscle groups, such as standing and walking. The reason for this is the unnatural regulation order of muscle fibers.

Physiologically, the muscle fiber regulation follows the Henneman's size principle. This means that small, slow-twitch muscle fibers are activated first. These fibers are fatigue resistant, but generate low force. When additional force is required, large fast-twitch, muscle fibers are recruited. These generate high force, but they are less fatigue resistant.

On the other hand, when muscles are activated by electrical stimulation, first fibers to be recruited are those with biggest voltage gradient, i.e., the largest fibers. Therefore, the recruitment order is opposite from the physiological, which leads to faster fatigue.

One way to postpone muscle fatigue is to stimulate the nerve which innervates the muscle we want to activate. There the recruitment is performed by the nerve, i.e., it is performed in the physiological order. Even though this is the optimal method, it is not always achievable with transcutaneous stimulations. It can be achieved only if the nerve is close to the surface, and even then electrode placement is a nontrivial problem.

When nerve stimulation is not possible, alternative solution is distributed low-frequency stimulation, performed with multipad electrodes [15, 22]. Here, instead of using one large electrode to activate large muscle group, e.g., quadriceps, many smaller electrodes are used asynchronously so that while some of the fibers are contracting, the others have time to recover.

References

1. Bajd, T.: Surface Electrostimulation Electrodes, p. 1–5. Wiley, New york (2006)
2. Bajd, T., Munih, M.: Basic functional electrical stimulation (FES) of extremities: an engineers view. Technol. Health Care **18**(45), 361–369 (2010)
3. Baker, L.L., McNeal, D.R., Benton, L.A., Bowman, B.R., Waters, R.L.: Neuromuscular electrical stimulation: a practical guide. Rehabilitation Engineering Program, Los Amigos Research and Education Institute, Rancho Los Amigos Medical Center, USA (1993)
4. Bigland-Ritchie, B., Jones, D., Woods, J.: Excitation frequency and muscle fatigue: electrical responses during human voluntary and stimulated contractions. Exp. Neurol. **64**(2), 414–427 (1979)
5. Bogataj, U., Gros, N., Kljajic, M., Acimovic, R., Malezic, M.: The rehabilitation of gait in patients with hemiplegia: a comparison between conventional therapy and multichannel functional electrical stimulation therapy. Phys. Ther. **75**(6), 490–502 (1995)
6. Keller, T., Kuhn, A.: Electrodes for transcutaneous (surface) electrical stimulation. J. Autom. Control. **18**(2), 35–45 (2008)
7. Keller, T.: Surface functional electrical stimulation (FES) neuroprostheses for grasping, vol. 14481. ETH Zurich, Switzerland (2001)

8. Knutson, J., Naples, G., Peckham, P.H., Keith, M.: Electrode fracture rates and occurrences of infection and granuloma associated with percutaneous intramuscular electrodes in upper-limb functional electrical stimulation applications. J. Rehabil. Res. Dev. **39**(6), 671–684 (2002)
9. Kuhn, A., Keller, T.: A 3D transient model for transcutaneous functional electrical stimulation. International Functional Electrical Stimulation Society Conference, Montreal, Canada, vol, 10, p. 385 (2005)
10. Kuhn, A., Keller, T.: A selectivity measure for transcutaneous electrodes. In: International Functional Electrical Stimulation Society Conference, Philadelphia, PA (2007)
11. Kuhn, A.: Modeling transcutaneous electrical stimulation, vol. 17948. ETH Zurich, Switzerland (2008)
12. Kuhn, A., Keller, T., Lawrence, M., Morari, M.: A model for transcutaneous current stimulation: simulations and experiments. Med. Biol. Eng. Comput. **47**(3), 279–289 (2009)
13. Kuhn, A., Keller, T., Lawrence, M., Morari, M.: The influence of electrode size on selectivity and comfort in transcutaneous electrical stimulation of the forearm. IEEE Trans. Neural Syst. Rehabil. Eng. **18**(3), 255–262 (2010)
14. MacKenzie, C.L., Iberall, T.: The grasping hand. North-Holland, Amsterdam (1994)
15. Malesevic, N., Popovic-Maneski, L., Schwirtlich, L., Popovic, D.B.: Distributed lowfrequency functional electrical stimulation delays muscle fatigue compared to conventional stimulation. Muscle Nerve **42**(4), 556–562 (2010)
16. Malesevic, N., Popovic, L., Bijelic, G., Kvascev, G.: Muscle twitch responses for shaping the multi-pad electrode for functional electrical stimulation. J. Autom. Control **20**(1), 53–57 (2010)
17. Zilles, K., Tillmann, B.N.: Anatomie. Springer-Lehrbuch (2010)
18. Miller, R.C., Popovic, M.R., Thrasher, T.A., Verrier, M.: Functional electrical stimulation therapy improves grasping in chronic cervical spinal cord injury: two case studies. J. Autom. Control **18**(2), 53–61 (2008)
19. Popovic, M.B., Popovic, D.B., Sinkjr, T., Stefanovic, A., Schwirtlich, L.: Clinical evaluation of functional electrical therapy in acute hemiplegic subjects. J. Rehabi. Res. Dev. **40**(5), 443–454 (2003)
20. Popovic, D.B., Popovic, M.B.: Automatic determination of the optimal shape of a surface electrode: selective stimulation. J. Neurosci. Methods **178**(1), 174–181 (2009)
21. Popovic-Bijelic, A., Bijelic, G., Jorgovanovic, N., Bojanic, D., Popovic, M.B., Popovic, D.B.: Multi-field surface electrode for selective electrical stimulation. Artif. Organs **29**(6), 448–452 (2005)
22. Popovic-Maneski, L., Malesevic, N., Savic, A., Keller, T., Popovic, D.B.: Surfacedistributed low-frequency asynchronous stimulation delays fatigue of stimulated muscles. Muscle Nerve **48**(6), 930–937 (2013)
23. Rattay, F.: Electrical Nerve Stimulation: Theory, Experiments and Applications, p. 264. Springer, Wien (1990)
24. Rose, D.J., Christina, R.W.: A multilevel approach to the study of motor control and learning. Allyn and Bacon, Boston (1997)
25. Sheffler, L., Chae, J.: Neuromuscular electrical stimulation in neurorehabilitation. Muscle Nerve **35**(5), 562–590 (2007)
26. Velik, R., Malesevic, N., Popovic, L., Hoffmann, U., Keller, T.: INTFES: A multi-pad electrode system for selective transcutaneous electrical muscle stimulation. In: 16th Annual Conference of the International Functional Electrical Stimulation Society, Sao Paulo (2011)