Basic functional electrical stimulation (FES) of extremities: An engineer's view

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VI.2: Basic Functional Electrical Stimulation(FES) of Extremites – an Engineer's View

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Abstract.

Keywords.

Introduction: Brief History of FES

Medical applications of the discharge of electric fish for the treatment of headache and arthritis were first recorded by Scribonius Largus in the second century AD. However, the first electric devices making use of static electricity were developed in the eighteenth century. These were based on the principle of using friction of insulators, such as glass, to develop high voltages due to charge separation. The ninteenth century brought a more practical stimulator with an induction coil. The so-called 'Faradic stimulator' consisted of a mechanical interrupter in series with a battery and the primary winding of a transformer. The output from the secondary winding of the transformer was a series of pulses which were similar to the stimuli of the present day electrical stimulators. The Faradic stimulator was the first device that could produce controlled and sustained muscle contractions. The output frequency could be adjusted to easily exceed the stimulation rate of 100 pulses per second and the output level was also controllable. Finally, the twentieth century brought transistors, integrated circuits, and microprocessors permitting sophisticated electronic circuits to be incorporated into very small devices.

1. FES Parameters

Functional electrical stimulation (FES) is a rehabilitation technology that uses electrical currents applied to peripheral nerves. When a stimulating current is applied to the electrodes placed on the skin overlying sensory-motor structures, an electric field is established between two electrodes and ions will create a current in the tissue (Figure 1).

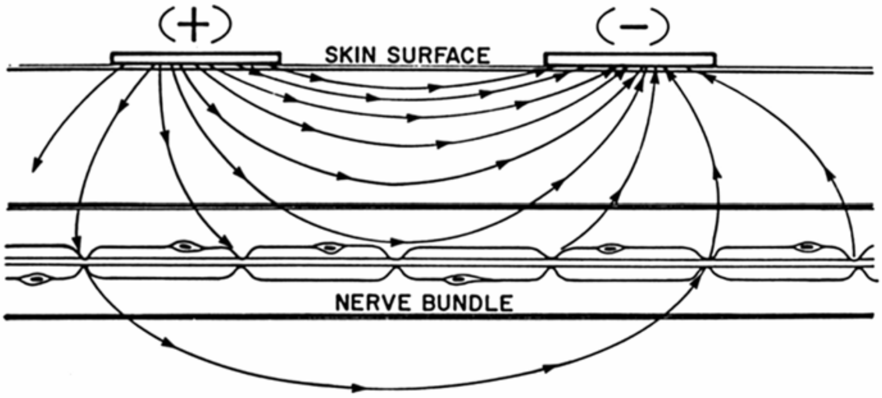

Figure 1. Electric field between a positive and negative electrode.

The ionic flow across the nerve influences transmembrane potential and can generate an action potential. The action potential propagates along the nerve causing contraction of a paralyzed muscle. In this way, FES provides restoration of movement or function, such as standing or walking by a person with a spinal cord injury.

FES is performed in a series of rectangular monophasic or biphasic (symmetrical or asymmetrical) electric pulses described by the following parameters: amplitude or intensity of pulses, frequency or pulse repetition rate, duration of single pulse, and duration of a pulse train. In most cases of surface FES applications, periodic monophasic or unidirectional pulses are used. Biphasic or bidirectional pulses prevent a slow deterioration of the electrodes, while the chemical conditions on the skin and in the muscular tissue remain unchanged.

1.1. Stimulator Circuit

With respect to the stimulator output circuit, electric pulses are either voltage- or current- controlled. Stimulators providing a *constant output voltage* can maintain the voltage desired, irrespective of resistance changes in the stimulated muscle tissue. Stimulators with current output stages make *constant current* pulses possible. An important difference between the two stimulator types becomes evident in the case of an improper contact between the electrode and the skin. In the case of a *constant current* stimulator, a smaller effective electrode surface results in greater current density, which can cause skin burns. With a *constant voltage* source, the resistance increases due to insufficient contact, which results in a decrease of current and, consequently, of the muscle response, but causing no skin damage.

1.2. Muscle Recruitment

A muscle recruitment curve represents the dependence of *isometric* (measurement performed at a constant muscle length) joint torque upon the FES amplitude or pulse duration (Figure 2). The joint torque is not linearly dependent upon the stimulation intensity. Two nonlinearities occur - threshold and saturation. The increase in joint torque due to an increasing amplitude of electrical stimulation occurs as a result of activating new fibers in a nerve bundle laying in an electric field between the electrodes. The main reason why all nerve fibers do not react to the same stimulation amplitude is found in the differences in the stimulation threshold and various distances from the stimulation electrodes. First, the fibers closest to the electrodes are stimulated. In addition, the fibers with a greater diameter respond earlier. Beyond a certain stimulation intensity, the force of contraction no longer increases. At such a stimulation amplitude, all nerve fibers are excited, and a further increase of the stimulus does not increase contraction. In surface FES of knee extensors for example, the values of the stimulation threshold range between 20 and 60 V, while the saturation value is between 100 and 150 V.

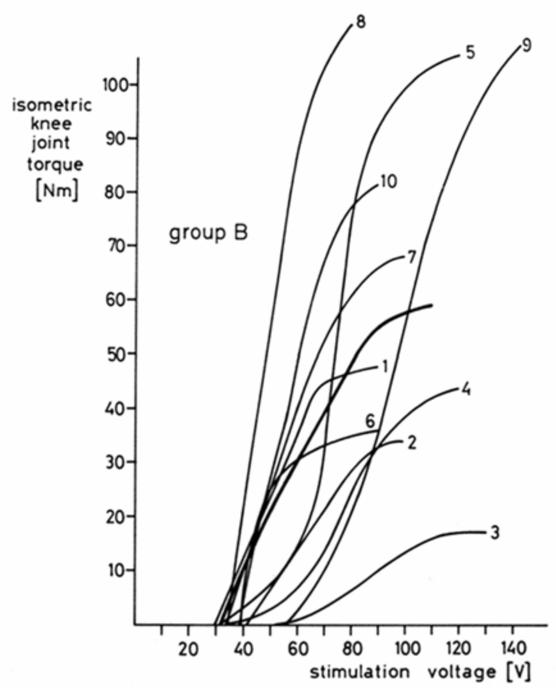

Figure 2. Muscle recruitment curves assessed in 10 paraplegic subjects after a FES restrenghtening programme.

A single stimulation pulse provokes only a short-lived muscle twitch of no more than 0.2 s. If electrical stimuli are repeated every second, a twitch occurs every second, between which the muscle relaxes. If the frequency of stimulation pulses increases up to 10 pulses per second (10Hz), between two twitches there is no time left for muscle relaxation. When measuring isometric contraction, we get twitching responses. This twitching is considerably reduced at stimulation frequencies between 15 and 20 Hz (Figure 3). At higher frequencies, the response is already smooth: this is known as *tetanic contraction*. The frequency at which the tetanic contraction occurs is called *fusion frequency*. It is not the same for all muscles and depends on properties of muscle fibers. Changes in stimulation frequencies also affect the intensity of the response. As regards the response intensity, slight losses are observed at lower stimulation frequencies. On the whole, the changing of frequency between 40 and 100 Hz causes small differences in the isometric torque as measured in the joint.

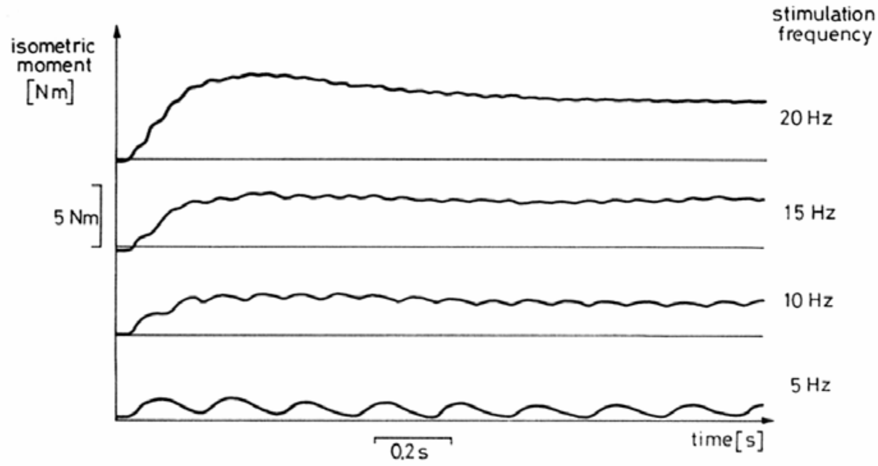

Figure 3. Influence of stimulation frequency on muscle response.

1.3. Muscle Fatigue

A low stimulation frequency results in less pronounced fatigue of the neuromuscular system. An electrically stimulated muscle fatigues more quickly than in the case of voluntary contraction. The main reason is the reversed recruitment order. It is not too serious an oversimplification to think of human muscle as a combination of slow fibers, capable of sustaining low levels of contractile activity without fatigue for prolonged periods, and fast fibers, capable of developing large forces, but fatiguing so rapidly that they can be used only in intermittent activities. Some muscles are predominantly made up of slow fibers, some predominantly of fast, and some of a given mixture of the two. All muscle fibers innervated by the same motoneuron have been found to be of the same type. Motoneurons innervating predominatly slow muscles have axons of small diameter, and those supplying fast muscles have axons of a larger diameter. In a voluntary contraction of a normally innervated muscle, the slow fibers are recruited first, and as increased muscle force is required, the fast fibers are recruited. Slow fibers are, therefore, activated frequently, while fast fibers are employed only infrequently, during a burst of intense activity. When applying electrical stimulation, fibers with a greater diameter respond earlier. This are motoneurons innervating fast muscles. The normal order of recruitment order is, therefore, reversed resulting in an increased fatiguing of electrically stimulated muscle. In addition, by electrical stimulation, the same nerve fibers are stimulated all the time, whereas with a healthy muscle the work is divided among different motor units of the same muscle. Also, due to relatively high stimulation frequency, the transmitter in the neuromuscular junctions is being exhausted, so the muscle stimulated soon shows signs of fatigue.

1.4. Duration of Stimulus

In a fashion similar to amplitude, pulse duration also exerts a direct effect upon the intensity of contraction. Here again, this is determined by threshold value and response saturation. When applying surface stimulation electrodes, the accompanying unpleasant sensation (when preserved) or even skin damage is due mainly to an excessively long duration of a stimulus, therefore short durations are used (0.1 to 0.5 ms), while the force of a paralyzed extremity is controlled by increasing the stimulus amplitude. Changing the pulse duration has little or no effect on stimulated muscle fatigue.

Functional movement of a paralyzed extremity cannot be obtained by a single electric stimulus, but requires a series of stimuli of a certain duration, following one another at an appropriate frequency. Such a series of stimuli is called a *stimulation pulse train*. In FES training of atrophied muscles, a stimulation pulse train is followed by a pause, and then by another stimulation train. The relationship of train duration and pause is often called the *duty cycle* and exerts an influence upon the fatigue of a stimulated muscle.

2. FES Electrodes

2.1. Anode & Cathode

A surface stimulation electrode is a terminal through which electrical current passes into the underlying tissue. At the electrode-tissue interface, a conversion occurs between the current of electrons driven through the wires coupled to the stimulator and the current of ions in the tissue. An electrode is usually made of metal. However, it may be made of a nonmetal, commonly carbon. The electrode through which current passes from the metallic or nonmetallic conductor *to* the tissue is called the *anode* and that through which current passes *from* the tissue to the conductor is the *cathode*. In electrical circuits, the current flows from the terminal at higher electrical potential to the terminal at lower electrical potential. In this way the *anode* is the *positive electrode* and the *cathode* the *negative electrode*.

2.2. Unipolar & Bipolar

Besides distinguishing between positive and negative electrodes, we also speak about *unipolar* and *bipolar* electrical stimulation techniques. With *unipolar* stimulation, one electrode is often considerably smaller than the other, whereas the electrodes used in *bipolar* stimulation both have the same size. In *unipolar* stimulation the smaller electrode is *negative* and is also called an *active electrode* due to the fact that in its vicinity there occurs depolarization of the membrane of nerve fibers. In motor nerve stimulation, the active electrode is positioned as closely to the *motor point* of the muscle as possible. The *motor point* is a site on the skin, where the amplitude of the *stimulus* required to fully activate the muscle is at a *minimum*, and where all of the motor nerve fibers are closest to the stimulating electrode. In multichannel electrical stimulation systems, it is possible to have a single anode and several independent cathodes or to have anodes and cathodes that are galvanically separated.

Let us examine four properties of surface stimulation electrodes and electrodes positioning, which influence the effectiveness of electrical stimulation: electrode size, polarity of electrodes, resistance, and distance between the electrodes.

(i) Electrode size: Electrical stimulation is applied to a nerve fiber, since muscle fibers have a considerably higher stimulation threshold. Thus we can say that larger electrodes are used to stimulate the nerve endings spreading all over the underlying tissue, whereas smaller electrodes are applied to influence the nerve when the latter come closer to the skin. Using larger electrodes, stronger contraction is obtained along with a reduced current density and a less pronounced unpleasant sensation on the skin. However, large electrodes permit no selective choice of a desired movement of the stimulated paralyzed extremity. The active areas of electrodes range between 2 cm^2 and 50 cm^2. Electrodes of 2 cm^2 to 4 cm^2 are used to stimulate the nerves near the surface, those of about 8 cm^2 for the stimulation of smaller muscles, while electrodes of 25 cm^2 or more are used in case of larger muscles.

(ii) Polarity:A positive and a negative electrode are placed along the muscle to be stimulated. Considering their polarity, the electrodes are positioned so as to provoke an optimal movement from the functional point of view. Stronger movement is usually obtained by placing the positive electrode distally.

(iii) Resistance:It is desirable that the resistance should be as low as possible in order to avoid energy losses before the stimulation has reached the neuromuscular tissue. The impedance between the electrode and the skin is frequency dependent. The DC (or low frequency) impedance tends to be several orders of magnitude larger than the impedance at higher frequencies. Nominal values of 2 kΩ are encountered. Contact conduction is increased by moistening the electrodes with water or special conductive electrode gels. Adipose tissue offers high resistance to electrical currents and so higher stimulation amplitudes must be used, causing pain in the skin as a side effect. Bones, too, are very bad conductors of electric current; electrical stimulation cannot reach muscles which are behind them.

(iv) Distance between electrodes:The greatest current density appears at the skin-electrode contact and tends to decrease with distance from the electrodes as the flow spreads out over a larger area. Closely spaced, small electrodes generally make the effective area of stimulation rather superficial due to the lower impedance of the current path through proximal tissue. Deeper tissues will be stimulated by using a greater distance between the electrodes. Increasing the electrode separation leads in general to an increase of the maximal achievable force. If the skin between the electrodes is moist, this causes the current between the electrodes to flow to the skin which results in a burning sensation and a slight or no muscle contraction at all.

2.3. Electrode design criteria

The design criteria for surface stimulation electrodes are: physical comfort to the skin, electrical surface area greater than four square centimeters to prevent skin irritation, use of hypo-allergenic materials, flexibilty to follow body surface, ease of attachement and ability to remain in position for a duration of at least one active day, reusable, low cost, reliable means of connection to stimulator, resistant to medical solvents and electrode gels, low and stable electrical resistance.

The simplest surface electrodes consists of a metal plate or metal wire mesh coated with fabric or sponge. Common materials used are stainless steel, silver-silver chloride, platinum, or gold. For safety purposes, the upper part of the electrode is covered with a non-conductive material. The electrode is applied after having been moistened with water. Such electrodes are usually fixed on the extremity by means of velcro or elastic bands. With the development of longer term surface electrodes, these electrodes are often used for site evaluation either in pain treatment or stimulation provoking muscle contraction. Small, button-shaped electrodes of similar design are highly suitable for the stimulation of a single nerve. Here, the metal plate is coated with several layers of gauze so that the electrode might retain the moisture required for as long as possible.

Surface electrodes made of silicone impregnated with carbon are applied to the skin surface with conductive gels and held in place with adhesive patches. A too thinly or unevenly spread gel increases current density at certain points, thereby bringing about a danger of burns. Electrodes may be left on the skin for several days a time. Another important property of electrodes made of conductive rubber is their flexibility making them adaptable to any part of the body. These electrodes can be shaped by cutting, so as to adapt them as much as possible to a proper stimulation site.

Conductive adhesive gel electrodes provide self-adhesion when applied (Figure 4). Karaya gum was the first of these adhesive gels. Later a variety of electrically conductive polymers were developed, enabling good contact with the irregularities of the skin surface and a more uniform current density at the electrode-skin interface. These electrodes can be used for extended periods with minimal skin irritation. An important breakthrough in electrode design was made by Axelgaard. This is a flexible electrode which adheres well to the patient's skin, is easily removed therefrom, and is able to move with the patient's skin ensuring proper placement of the electrode. The electrode is in the form of a knit conductive fabric. Conductive fibers include a blend of stainless steel and polyester. The fabric can be stretched up to about 20%. A conductive adhesive fills interstitial areas of the knit fabric and adheres the electrode to the patient's skin. A non-conductive sheet on the other side of the knit fabric prevents undesired electrical contacts.

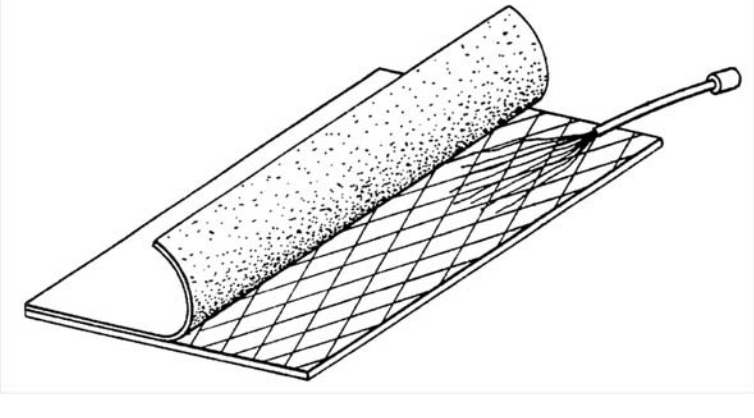

Figure 4. A conductive adhesive gel electrode with a portion of nonconductive sheet peeled back showing the knit fabric.

2.4. Problems with electrodes

When improperly handled, electrodes can damage the skin in the contact area. Burns typically occur underneath the anode, but not the cathode, when using identical surface electrodes. Another problem resides in a precise electrodes positioning along a muscle. Sometimes a displacement of a few millimeters completely changes the muscle response. This happens when a selected nerve (e.g. peroneal nerve) ought to be stimulated by surface electrodes. Surface electrodes may excite pain receptors in the skin, although patients sensibility may be reduced to such an extent that the sensation of pain is not critical. Another problem is undesired motion of the skin with respect to the neuromuscular tissue. Even though an electrode seemingly occupies the same place all the time, its distance from the nerve is not constant. This is one of the reasons why the movements caused by electrical stimulation cannot be easily repeated. Another limitation is that small muscles usually cannot be selectively activated and deep muscles cannot be stimulated without first exciting the superficial muscles. Relatively high voltages, sometimes in excess of 100V, between electrode pairs cause hazards for the patients and the personnel that treat them. Finally, the applicability of the surface stimulation electrodes depends on fixation problems. Stretchable garments with electrodes already mounted in appropriate locations have been developed by several manufacturers to simplify the application of electrodes to the skin surface. In the case of lower limb stimulation, fixation problems can be overcome by specially designed trousers carrying stimulation electrodes and cables. Such stimulation equipment is comfortable and easy to handle. In the non-invasive, upper limb neuroprosthesis, the surface stimulation electrodes were built into an elegant, self-aligning, and flexible splint. The splint provides additional fixation of the wrist joint and allows the entire electrode array to be positioned within a few seconds.

It is not difficult to realize that most of the inconveniences of surface stimulation electrodes can be overcome by the use of implanted electrodes. Nevertheless, because of their simple non-invasive application, surface electrodes will remain of use in therapeutic treatments.

3. Electrical stimulators

Electrical stimulators comprise an input circuit, pulse generator, output stage, and power supply (Figure 5). The input into the electrical stimulator is represented by a control signal automatically switching electrical stimulation pulses on or off or may be under the patient's voluntary control. The output electric pulse current is led, via electrodes, to the selected stimulation site. The type of stimulation pulse is determined by a pulse generator. The output pulses provided by the circuit are considerably lower than the surface stimulation pulses required for a functional movement of an extremity but are of an appropriate frequency and duration. The output stage ensures energy for the electrical stimulation of paralyzed muscle in either constant current or the constant voltage outputs. In case of current stimulation, the internal resistance of the end stage is considerably higher than the tissue resistance between the electrodes. The current source of stimulation pulses provides a constant current irrespective of the resistance of the skin and the tissue between the electrodes. In the case of constant voltage output stages, skin resistance is lower than that between the electrodes, and the stimulator provides a constant voltage independent of the skin and tissue resistance. A power supply provides the energy necessary for the operation of particular electronic circuits (low voltage) and the electrical stimulation itself (high voltage). Electrical stimulators are usually battery powered with stimulation pulses with an amplitude of more than 100 V. A high voltage for the output stage is obtained from low battery voltage by means of a voltage convertor.

The development of electronics made it possible for Wladimir Liberson to develop a stimulator to preventing foot-drop in hemiplegic patients. It was triggered by a heel switch in the shoe of the affected leg. Stimulation electrodes were positioned above the peroneal nerve in the popliteal fossa, behind the knee joint. Each time the patient raised the heel, the heel switch triggered the stimulator, causing the nerve to cause the extensor group of muscles to contract and dorsiflex the ankle and so lift the foot. Advanced versions of these dropped foot stimulators were developed in Ljubljana, Slovenia and Salisbury, UK. Both stimulators were applied to a large group of stroke patients. In the WalkAide peroneal nerve stimulator, a tilt sensor triggers ankle dorsiflexion.

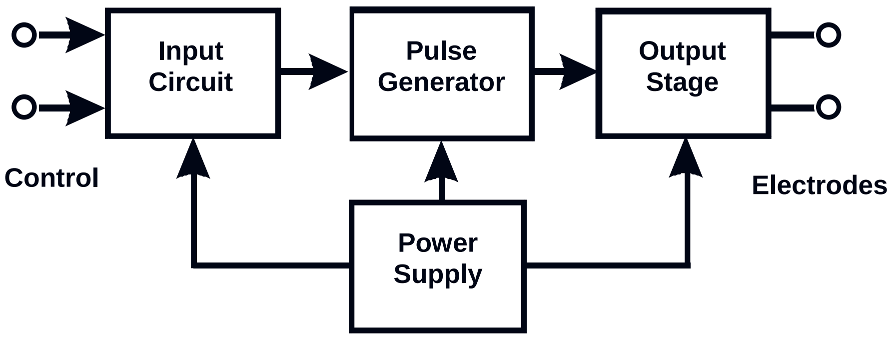

Figure 5. Block scheme of an electrical stimulator.

A minimum of four channels of FES were used for synthesis of a simple reciprocal gait pattern in completely paralyzed paraplegic subjects. During reciprocal walking, the stimulator must be controlled through three different phases of walking: right swing phase, double stance phase, and left swing phase. This is achieved by two hand-push buttons built into the handle of the walker or crutches. When niether of the push buttons is pressed, both knee extensors are stimulated providing support to the body. On pressing the push button in the right hand, the peroneal nerve is stimulated in the right leg eliciting the flexion reflex presented by simultaneous hip and knee flexion and ankle dorsiflexion. The same is true for the left leg. The Ljubljana FES walking system consists of two small two-channel stimulators attached to each leg. Only three electrodes are applied to a single leg in order to produce knee extension and flexion responses. As both activities never occur simultaneously, the distal electrode placed over the knee extensors represents the common electrode for both stimulation channels. Using the same principles as the Ljubljana FES system, and adding two channels of stimulation to both hip extensors, the FDA-approved Parastep surface stimulation system was developed. A new multipurpose programmable transcutaneous electric stimulator, Compex Motion, was developed to allow users to design various custom-made neuroprostheses, neurological assessment devices, muscle exercise systems, and experimental setups for physiological studies. The Compex Motion stimulator can generate any arbitrary stimulation sequence, which can be controlled in real-time using any external sensor.

Despite the fact that correct application of a stimulator implies no danger either for the patient or therapist, we wish to emphasise some important safety points. When buying a new electrical stimulator, be sure that it carries the CE mark. Every stimulator has its own characteristics which must be indicated in the instructions of use, an obligatory accompaniment of any commercially-available stimulator. Special attention should be paid to stimulation parameters: pulse duration, frequency, and maximal current or voltage. Also of utmost importance is the information on the proper use of surface electrodes. The use of electrical stimulators might be dangereous in case of patients with an implanted pacemaker. Simultaneous use of electrical stimulators and high frequency surgical devices is prohibited. Electrical stimulators might not work properly in close proximity to microwave devices. Transthoracic positioning of FES electrodes may cause fibrillation of the heart. The stimulator should not be switched on in case of short-circuit of the electrodes. It is true, that electrical stimulators are made in such a way that a short-circuit of several minutes does not damage them. However, one should not take unfair advantage of this property.

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