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A System for Detecting Stepping Cycle Phases and Spinal Cord Stimulation as a Tool for Controlling Human Locomotion

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A measurement system was developed to record angular movements at the joints of the lower limbs (the stance and swing phases) allowing the moments of detachment and contact of the foot with the support surface to be determined using sensors responding to linear and angular acceleration. We present an algorithm for triggering spinal cord stimulation in specified phases of the stepping cycle, addressing the flexor and extensor motor pools of the lower limbs. A means of triggering temporospatial spinal stimulation for the "paralyzed" limb from the "intact" contralateral limb simulating stimulation conditions for patients who have had cerebrovascular accidents was developed. It is proposed that this system can be used in therapeutic, therapeutic-preventative, and medical research institutions or at home to regulate and restore motor functions in humans.

Introduction

Restoration of locomotion in spinal trauma or stroke makes wide use of neuromuscular electrical stimulation, good results being obtained with functional electrical stimulation (FES) applied at the time points at which the corresponding muscles need to be activated (see, for example, [1, 2]). In addition, recent years have seen the effective use of epidural electrical stimulation of the spinal cord (ESSC) [3] and noninvasive transcutaneous stimulation of the spinal cord (TCSSC) [4], designed to activate locomotor neural networks controlling rhythmic lower limb movements. The use of TCSSC in the corresponding phases of the stepping cycle opens up new possibilities for controlling locomotor function and rehabilitating motor impairments [4, 5]. Thus, there is a need to develop online systems for detecting stepping cycle phases for the targeted stimulation of specific motor pools.

The literature evaluates a number of methods for recording gait phases. The most widely used are force platforms, contact switches, pressure sensors, video recordings, light-reflecting tags, accelerometers, and gyroscopes. Studies have shown that accelerometers and gyroscopes can be used to measure the kinematic parameters of the gait and evaluate the temporospatial characteristics [6, 7], joint angles [8, 9], and movement trajectories of limb segments [10].

Accelerometers — devices which measure the vector sum of the linear and free-fall (force of gravity) accelerations — are attached to the body, particularly the trunk [11, 12], and can be used to determine the moment at which the foot makes contact with the support surface, though accelerometer signals are inaccurate in terms of identifying the moment of detachment of the foot from the surface. Gyroscopes — devices measuring angular velocities — are used as detectors of stepping cycle phases in gait analysis in healthy people. They are also used in electrical stimulation systems. Comparison of measurement results obtained using gyroscopes attached to the knee to identify the phases of the stepping cycle with systems measuring pressure on contact of the foot with the support have shown that gyroscopes are a suitable and sufficiently accurate approach for detecting the gait on both level and tilted surfaces [13]. Gyroscopes attached to the hip [14] give quite clear but multiple speed change peaks at the moments at which the foot makes contact with the support surface and detaches from it. A number of studies have used gyroscopes in combination with accelerometers [15-17].

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We describe here a measuring system for online detection of stepping cycle phases using sensors positioned on the hip. The system can be supplemented by analog sensors positioned on other parts of the body, providing for determination of the kinematic parameters of movements of all lower limb segments. The system can be used to drive switching of electrical stimulators, to trigger stimulation at specified phases of the stepping cycle, and to control the electrical stimulation of impaired limbs from the movements of the intact limb in patients with hemiparesis.

Sensors for identification of stepping cycle phases based on recording of kinematic parameters — linear acceleration (accelerometers) and angular velocity (gyroscopes) — form the nucleus of the measuring system (Fig. 1). The sensor contains the following components: 1) a measuring element implemented as a three-layer accelerometer in a single LSM6DSL integrated circuit; 2) an MSP430F5510 microcontroller; 3) a radio module; 4) a power supply based on a Li-ion battery.

Algorithms

The electronic gyroscope used in the sensor provides accurate measurements of angular velocity of the subject's body part to which the sensor is attached. The algorithm is based on numerical integration of the angular velocity measured with the gyroscope with corrections from the accelerometer in the sensor element.

Real-time determination of the stepping phase during walking can be obtained by measuring the angles of body parts with the vertical in the sagittal plane alone. Thus, sensors in the measuring system were oriented in space as follows: the X axis was to the right; when the sensor was at 0° with respect to the vertical, the Y axis faced forward and the Z axis was upward. Sensors were attached to the anterior surfaces of body parts, so that rotation of the sensor on movement in the sagittal plane was around the X axis. Thus, the angle between the vertical and the Z measuring axis of the sensor (the angle of the sensor with the vertical) is computed using the following procedure.

1) Sensors are interrogated at a frequency of 100 Hz and the following computations are run for each step:

— an intermediate estimate of the angle of the sensor with the vertical α_N^1 is made as

$$\alpha_N^1 = \alpha_{N-1} + (\omega - \omega 0_{N-1})\cdot\tau,$$

where α_{N-1} is the estimated angle between the sensor and the vertical in the preceding step, ω is the component of the angular velocity on the X axis determined by the

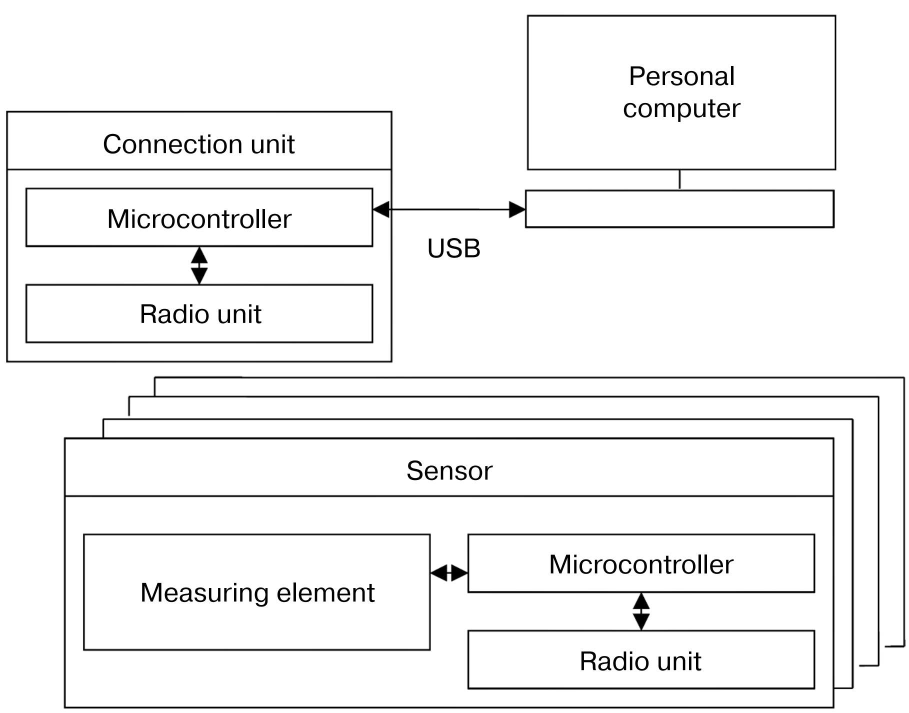

Fig. 1. Block diagram of the measuring system.

gyroscope; $\omega 0_{N-1}$ is the estimate of the gyroscope null offset in the preceding step, and τ is the measurement period (10 ms);

— the mismatch δ between the intermediate estimate of the angle of the sensor and the value determined from the accelerometer is determined:

$$\delta = \arctan(A_Y/A_Z) - \alpha^1_N,$$

where A_Y and A_Z are the accelerometer readings for the Y and Z axes respectively.

2) The angle of the sensor with the vertical and the null offset of the gyroscope reading are corrected:

$$\alpha_N = \alpha^1_N + K_\alpha \cdot \delta;$$

$$\omega 0_N = \omega 0_{N-1} - K_\omega \cdot \delta,$$

where K_α and K_ω are constant coefficients selected experimentally on the basis of improvements in the quality of transfer processes.

This measurement algorithm gives errors even for ideal sensor readings, as the accelerometer signal depends not only on the sensor orientation (not only the angle between the sensor and vertical), but also on the linear accelerations experienced by the sensor. However, as the mean acceleration over the time period of the step is zero, the accelerations were comparable with or smaller than free-fall acceleration and the values of the coefficients K_α and K_ω were small, so errors introduced by linear accelerations were minor.

The **sensor/personal computer connection unit** was linked with the sensors via a radio channel and with the computer via USB. It included a connector for the stimulator and carried out the following functions: interrogation of the sensors at a specified frequency (generally 15 Hz); transmission of data to the computer; analysis of sensor signals; generation of synchronization signals to trigger stimulation. The measuring system (Fig. 1) included sensors, connecting unit, and personal computer. The software consisted of three elements: 1) the sensor microcontroller program; 2) the connecting unit microcontroller program; 3) the computer control program.

TABLE 1. Errors in Detection of the Moment of Detachment of the Foot from the Surface and Contact with the Surface Using the Sensor System Described Here (% of the duration of the stepping cycle in four subjects)

Subject No.	Version 1		Version 2	
	Stance phase	Swing phase	Stance phase	Swing phase
1	3.76 ± 0.44	−2.02 ± 0.73	2.86 ± 0.74	−5.88 ± 0.94
2	−3.29 ± 1.10	0.16 ± 0.46	−5.18 ± 1.40	9.29 ± 1.36
3	1.86 ± 0.83	−1.00 ± 1.16	1.55 ± 1.54	1.84 ± 1.53
4	5.44 ± 0.48	−0.22 ± 0.70	2.51 ± 1.49	−5.07 ± 0.78

Online Algorithms for Detecting the Stepping Cycle

Version 1. The moment of placing the foot on the support (the end of the swing phase and the beginning of the stance phase) was taken as the time point at which hip flexion started (Fig. 2). This moment was determined from the change in sign of the derivative of the signal at the hip joint from plus to minus. The moment of detachment of the foot from the support (the end of the stance phase and the beginning of the swing phase) was taken as the moment at which hip flexion started. This time point was determined by the change in sign of the derivative from minus to plus.

Version 2. This was developed to allow electrical stimulation in the appropriate phase of the stepping cycle to be delivered in patients with hemiparesis, when the preserved leg demonstrated a relatively normal movement pattern and the afflicted leg could move nonuniformly, with jerks, etc., making it impossible to use the signal from the sensor attached to it to trigger stimulation. This algorithm provided the opportunity to use the signal recorded from the preserved leg to be used for stimulation of the afflicted leg. This is carried out by transferring the patterns of time points at which stepping cycle phases were detected from the preserved leg to the afflicted leg with an offset of one half of the duration of the stepping cycle (by $T/2$).

Offline analysis of stepping cycle phase detection was performed by comparing the signals recorded using the measuring system developed here with the time points of detachment of the foot from and contact of the foot with the support determined from traces of the coordinate markers on the subject's heel (recorded using a Qualysis video analysis system). The mean error of detection of stepping cycle phases directly from the signal from the leg performing the step (version 1) ranged in different subjects from −3% to +5% of the duration of the stepping cycle (Table 1). When detecting stepping cycle phases using the sensor applied to the essentially healthy leg making the step (version 2), subjects' errors were in the range −6% to +10%. Comparison of the spreads of mean error values showed that the standard deviation in version 1 in all subjects was significantly smaller than in ver-

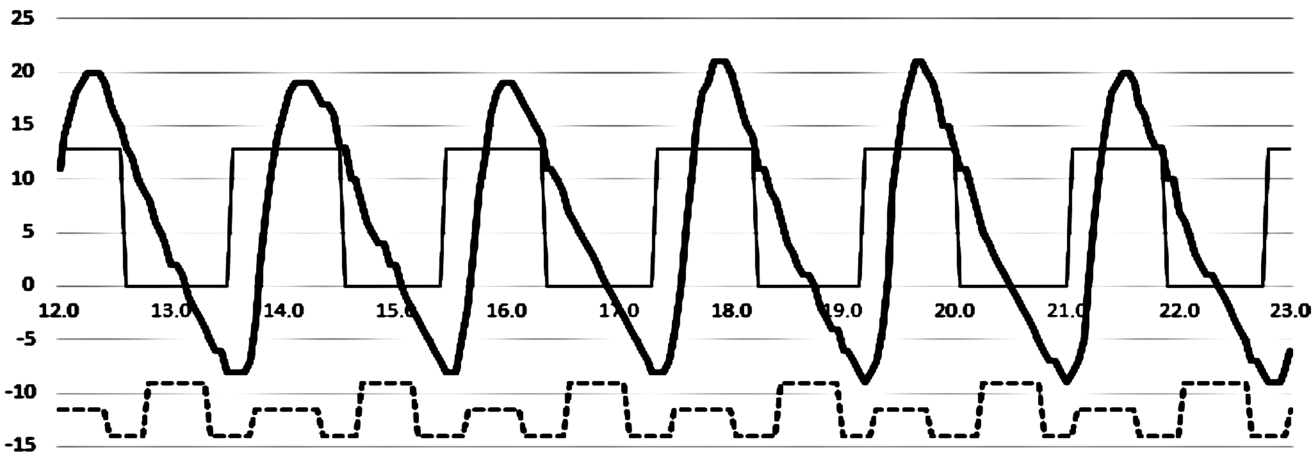

Fig. 2. An example of the operation of the online stepping cycle phase detection algorithm during walking on a treadmill: the thick continuous line shows the angle at the hip relative to the vertical recorded using the measuring system described here (the upper parts of the plot show flexion of the hip, the lower parts show extension); the thin continuous line shows the signal from the contact sensor on the right heel (the upper parts of the plot show detachment of the heel from the surface); the dotted line shows operation of electrical stimulation (upper part of plot); the shorter rectangles show activation of stimulation of the motor pools of the right-sided flexors and the taller rectangles show stimulation of the left side. For further explanation see text.

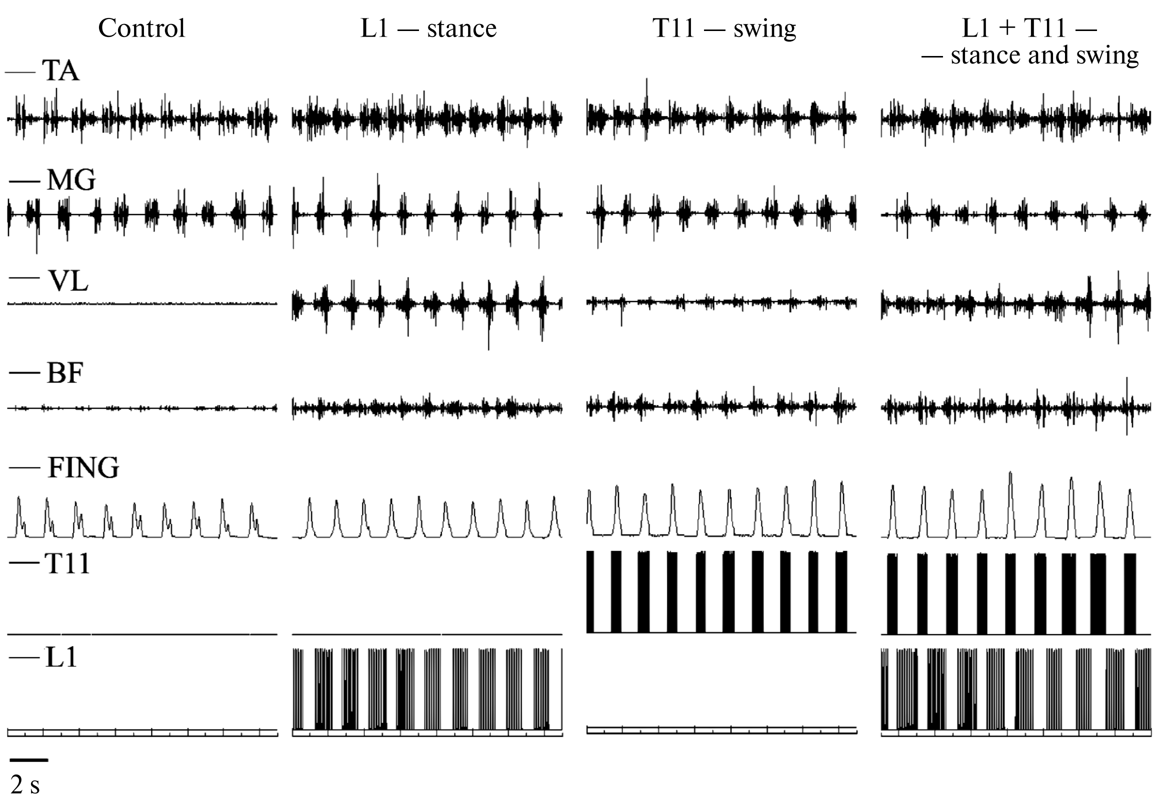

Fig. 3. Walking on the treadmill with rhythmic stimulation of the spinal cord in different phases of the stepping cycle: above — EMG of the right leg muscles; TA — tibialis anterior; MG — medialis gastrocnemius; VL — vastus lateralis; BF — biceps femoris; Fing — vertical component of the position of the marker attached to the right great toe; T11 and L1 — markers for electrical stimulation of the corresponding segments of the spinal cord.

sion 2 ($p = 0.05$, Fisher's test). This provides evidence that detection accuracy using sensors on the contralateral leg (version 2) somewhat degrades the accuracy of the stimulation start time. Nonetheless, detection errors were not large, and the stimulation algorithm using sensors located on the preserved side could be used in patients with hemiparesis.

An example of spinal cord stimulation using detection of stepping cycle phases is shown in Fig. 3.

Spinal cord stimulation used a noninvasive spinal cord electrical stimulator [18]. The stimulating electrodes were positioned between vertebrae T11/12 (stimulation of flexor motoneuron pool) and L1/2 (stimulation of extensor motoneuron pool) at a distance of 1 cm to the right of the midline of the spine. Stimulation of the flexor pool was at a frequency of 30 Hz and stimulation of the extensor pool was at 15 Hz.

With the subject walking on a treadmill without stimulation (control conditions), rhythmic EMG activity was seen mainly in the distal leg muscles (TA and MG) (Fig. 3). Stimulation of the extensor pools in the stance phase increased EMG activity in the thigh muscles, with predominant activation of the extensor muscles (VL). Stimulation of the flexor pools in the swing phase led to increased activation of the flexor muscles (TA and BF) and increased lifting of the limb. Stimulation of the flexor and extensor pools respectively in the swing phase and the stance phase increased lifting of the leg (the Fing amplitude) and increased EMG activity in the thigh and leg muscles.

Thus, the results obtained here provide evidence that addressed stimulation of the flexor and extensor motoneuron pools can effectively control locomotor activity in humans.

Conclusions

The algorithm for detection of stepping cycle phases in terms of changes in the angle at the hip joint provides for detection of the stance and swing phases with a mean error of no more than 5% of the duration of the stepping cycle. Detection of stepping cycle phases using sensors attached to the contralateral side gave a mean error of no more than 10%. This method provides not only for detection of the stance and swing phases, but also analysis of the kinematic characteristics of gait; it also provides addressed electrical stimulation of the flexor and extensor motor pools at specified phases of the stepping cycle.

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