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Three dimensional inertial sensing of foot movements for automatic tuning of a two-channel implantable drop-foot stimulator

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Abstract

A three dimensional inertial sensing system for measuring foot movements during gait is proposed and tested. It can form the basis for an automated tuning system for a two-channel implantable drop-foot stimulator. The foot orientation and position during the swing phase of gait can be reconstructed on the basis of three-dimensional measurement of acceleration and angular velocity, using initial and final conditions during mid-stance. The foot movements during gait of one stroke person using the implanted two-channel stimulator were evaluated for several combinations of stimulation parameters for both channels. The reconstructed foot movements during gait in this person indicated that the channel stimulating the deep peroneal nerve contributes mainly to dorsiflexion and provides some reduction of inversion seen without stimulation, while the channel activating the superficial peroneal nerve mainly provides additional reduction of inversion. This agrees with anatomical knowledge about the function of the muscles activated by both branches of the peroneal nerve. The inertial sensor method is expected to be useful for the clinical evaluation of foot movements during gait supported by the two-channel drop-foot stimulator. Furthermore, it is expected to be applicable for the automated balancing of the two stimulation channels to ensure optimal support of gait.

Keywords: Inertial sensing; Peroneal nerve stimulator; Functional electrical stimulation; Stroke

1. Introduction

The impact of stroke in the life of an individual is very dramatic [1–3]. It disables a person both mentally and physically, having a detrimental effect on the motor abilities of one side of the body. It results in asymmetric body use and perception. Physically, the motor control of one side of the body may be deteriorated. Often, such a deteriorated motor function improves during the recovery period and by training, although in many cases motor control problems remain. Among others, a frequent consequence is the inability to voluntarily lift the foot at the affected side (drop foot). Since Liberson [4], FES systems have been developed to artificially activate the dorsiflexor muscles during the swing phase of gait. These systems have been reported to result in increased walking speed and reduction of physiological cost index [5]. Additional to the orthotic effect, therapeutic effects have been reported, including reduction of spasticity and additional improvement of voluntary control [6]. Stimulation of the peroneal nerve may also reduce the calf muscle stretch reflex [7].

Among the major difficulties of dropfoot stimulation are the critical positioning of stimulation electrodes on the skin to achieve a well-balanced dorsiflexion movement without excessive inversion or eversion and technical problems with cables and foot switch sensors [8]. This is one of the main reasons why several users discontinue the use of dropfoot stimulators after some time. Additional reasons for discontinuing are unpleasant skin sensations and skin irritation. A reported positive reason is that part of the users regain some voluntary control over their dorsiflexor muscles after some time [8].

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Alternative sensor modalities have been proposed for detecting the start and end of the swing phase to overcome the difficulties of using a foot-switch. These alternative modalities include inertial sensors on the shank [9,10] and signals from natural skin sensors of the foot derived using cuff electrodes around sensory nerves [11,12].

Implantable dropfoot stimulators have been developed to overcome problems with critical electrode positioning and skin sensation [13]. However, excessive eversion may still result after some time and can not be corrected because the electrode is implanted. For this reasons, a two channel implantable drop foot stimulator has been developed [14], allowing automated balancing of inversion and eversion during foot lift (Fig. 1). In the past year, this system has been implanted in four stroke patients in Twente and Salisbury [15].

Currently, the two channels of the stimulator are balanced manually. However, automatic balancing based on quantitative evaluation of foot movements by sensor measurements may be possible. For effective application of the electronic balancing feature of this two-channel implant, desired movement patterns need to be defined. Furthermore, a sensor system for objective measurement of relevant movement parameters of gait and procedures for the adjustment of the stimulation parameters of the two channels, both in time and amplitude, need to be developed [16].

It is the objective of the current paper to present the concept of automatic tuning of the two-channel implantable drop-foot stimulator and to demonstrate the feasibility of two essential prerequisites for this method. The first prerequisite is the feasibility of the 3D measurement of foot movements during gait using a three-dimensional inertial sensor system on the foot. The second prerequisite is the definition of relevant foot movement parameters which are dependent on stimulation levels.

The envisioned sensory system is not only expected to be relevant as feedback for automated tuning, but also to assess objectively the quality of gait and to track a patient's progress during rehabilitation. This does not require a fully equipped movement analysis lab with an expensive opto-kinetic measurement system and avoids the restrictions of a limited measurement space. It has the potential of being used continuously to adjust stimulation when muscles become fatigued, or when the gait pattern or the walking surface changes.

2. Automatic tuning of the two channel drop-foot stimulator

In order to tune the two-channel dropfoot stimulator automatically, we propose to define relevant movement parameters which would serve as desired reference values to be achieved through automatic adaptation of stimulation parameters [17]. This concept of automatic tuning is schematically presented in Fig. 2. The vector $\vec{m}_n$ represents the movement parameters measured in walking cycle n, $\vec{s}_n$ represents the stimulation parameters applied in the same cycle. The stimulation parameters are determined from the desired movement parameters using an estimated relation. This relation is recursively identified from measured foot movement parameters and the applied stimulation levels for each step.

Relevant movement parameters may be defined for several instances in the swing phase. This includes mid-swing, when foot-orientation and -lift determine foot clearance, and just before foot-strike at the end of the swing phase. Heel-first landing without excessive inver-

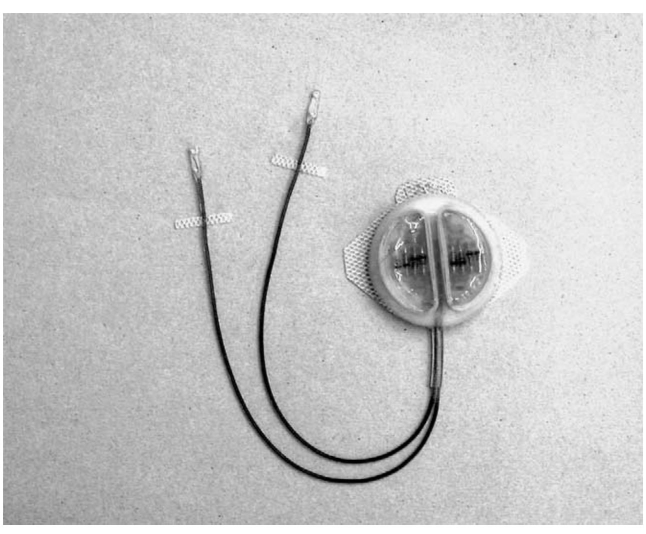

Fig. 1. Two-channel drop-foot stimulator for independent stimulation of the deep and superficial branches of the peroneal nerve [14,15].

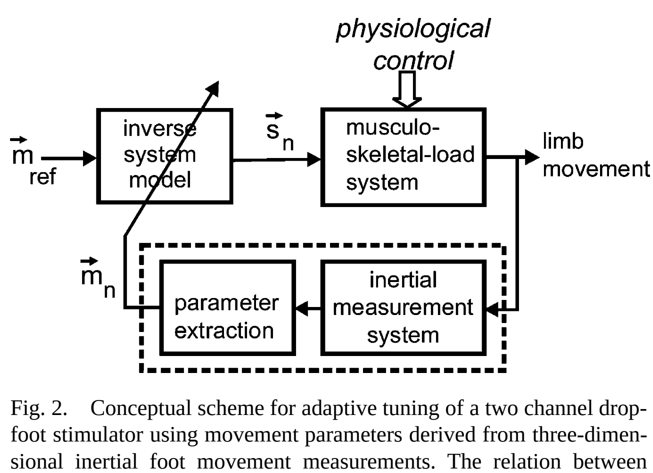

Fig. 2. Conceptual scheme for adaptive tuning of a two channel drop-foot stimulator using movement parameters derived from three-dimensional inertial foot movement measurements. The relation between stimulation and movement parameters are recursively identified. This relation is used to calculate stimulation parameters on the basis of desired movement parameters. The vector $\vec{m}_n$ represents the movement parameters measured in walking cycle n, $\vec{s}_n$ represents the stimulation parameters applied in the same cycle.

sion or eversion contributes to stable gait and optimal loading transition from swing to stance. Heel-first landing also avoids excessive stretch reflexes in the calf muscle which may result from fast dorsiflexion when the forefoot hits the ground first [18].

The current paper considers a subset of relevant foot movement parameters, being the sagittal and coronal orientations of the foot just prior to foot-strike. With respect to stimulation parameters, only the stimulation levels of both channels were considered.

3. Methods

3.1. Assessment of 3D foot movement during gait using inertial sensors on the foot

Accelerometers measure the inertial and gravitational forces acting on a mass. The signal of a three-dimensional accelerometer $\vec{s}_{\mathrm{a}}$ can be expressed as follows:

$$\vec{s}_{\mathrm{a}} = K_{\mathrm{a,s}}(\vec{a}-\vec{g}) + \vec{k}_{\mathrm{a,o}} \quad (1)$$

$\vec{a}$ representing the inertial acceleration and $\vec{g}$ the gravitational acceleration in sensor coordinates, $K_{\mathrm{a,s}}$ is the sensitivity matrix and $\vec{k}_{\mathrm{a,o}}$ the offset vector.

Micro-machined rate gyroscopes measure angular velocity $\vec{\omega}$. In 3-D, the signal vector $\vec{s}_{\mathrm{g}}$ can be expressed as follows:

$$\vec{s}_{\mathrm{g}} = K_{\mathrm{g,s}}\vec{\omega} + \vec{k}_{\mathrm{g,o}} \quad (2)$$

$K_{\mathrm{g,s}}$ is the sensitivity matrix and $\vec{k}_{\mathrm{g,o}}$ the offset vector.

3D foot position and orientation can be estimated using the signals of an inertial sensor system existing of a 3D accelerometer and a 3D rate-gyroscope. These 3D sensors can consist of three uniaxial sensors. It requires integration of the angular velocity $\vec{\omega}$, measured with the gyroscopes (2), to orientation [19] and double integration of acceleration $\vec{a}$ (1) in inertial reference frame coordinates. This requires that the orientation of the sensor system at each time instance is taken into account and that the acceleration of gravity $\vec{g}$ is subtracted (1). It should be noted that it is not self-evident, even remarkable, that this can be done sufficiently accurately. Moreover, small errors in the estimated offsets, $\vec{k}_{\mathrm{a,o}}$ and $\vec{k}_{\mathrm{g,o}}$, and gains, $K_{\mathrm{a,s}}$ and $K_{\mathrm{g,s}}$, may result in integration drift, yielding further errors in estimated positions and orientations. However, in the special case of human gait, certain initial and final conditions can be assumed (e.g., foot flat on the ground, vertical position at the beginning and end are equal) and the integration time is limited (in the order of one second).

The algorithm for deriving 3D foot position and orientation during gait was based on these principles and consisted of several components (Fig. 3). Initially, the orientation of the sensor module on the foot is determined from a static stance measurement, measuring the gravity acceleration vector in sensor coordinates using the 3D accelerometer. The beginning and end of each stride are determined on the basis of relevant signal characteristics. At the end of each stride, the angular velocity measurements are integrated [19], and drift is compensated applying the initial and end-conditions for a stride. The resulting foot orientation as a function of time is subsequently used to express the 3D accelerometer signals in inertial coordinates using a coordinate transformation. The gravity component of the accelerometer signal can then be subtracted, resulting in the acceleration as a function of time. This acceleration is integrated, again imposing the initial and end-conditions for a stride to avoid drift.

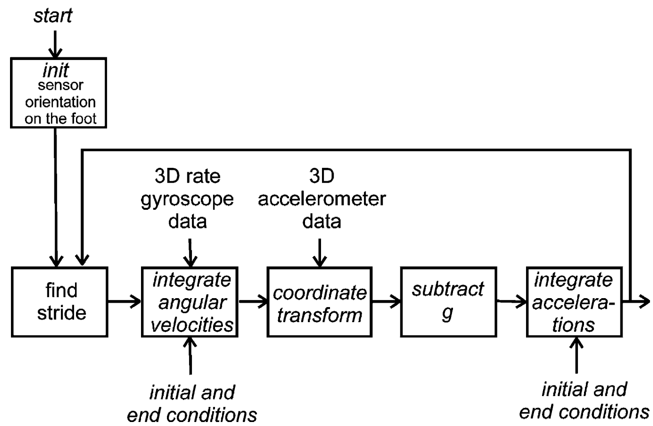

Fig. 3. Schematic representation of the algorithm for deriving 3D foot position and orientation during gait on the basis of 3D inertial sensor signals. After static initialization of the orientation of the sensor on the foot, the measured angular velocities are integrated taking into account the initial and end conditions of a step. The accelerometer signals are expressed in inertial coordinates using the resulting orientations at any time instance during the step. After subtraction of the gravity acceleration, the accelerations are integrated twice to yield positions as a function of time, taking into account relevant initial and end conditions of a step.

In an earlier application of this method speed and traversed distance were measured during gait and running in healthy persons. It was found that the errors in the estimated total distance traveled were below 3%. As a reference, the actual traveled distance was measured.

It should be noted that the foot orientations are considered relative to the inertial coordinate system of the environment and not relative to the lower leg. This is justified by the fact that the movement of the foot relative to the ground is important when evaluating foot clearance and foot-strike.

3.2. Experimental setup

The experimental setup is schematically depicted in Fig. 4a. Fig. 4b provides a picture of the setup on the

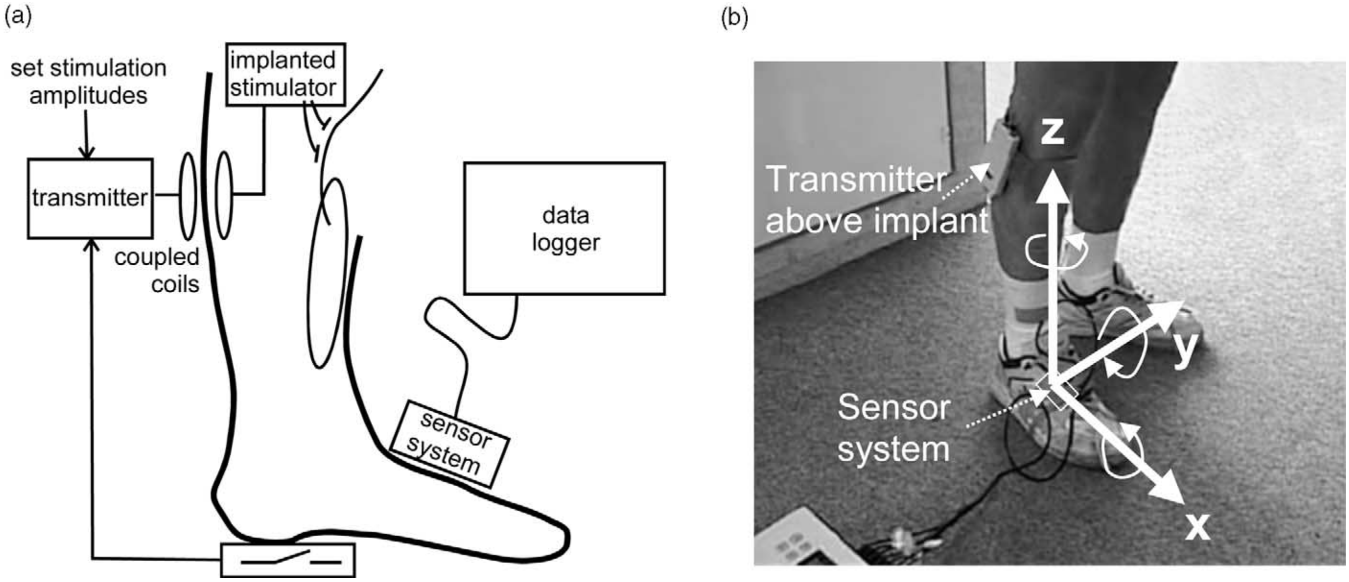

Fig. 4. Schematic representation (a) and picture (b) of the experimental setup. The implanted two-channel dropfoot stimulator is powered and controlled by an external transmitter through an inductive coupling. The stimulation timing is controlled using a footswitch and the balance between both channels is adjusted manually on the transmitter. The inertial sensor was taped to the shoe, but can also be strapped under the shoelaces. The sensor signals were sampled using a portable datalogger. The inertial coordinate system is indicated in Fig. b. The *x*-axis coincides with the direction of progression during gait, the *z*-axis is directed vertically and the *y*-axis is defined perpendicular to *x*- and *z*-axes. Positive *x*, *y*, *z* orientations are indicated.

leg of a subject. The implanted two-channel stimulator activated two branches of the peroneal nerve. Channel 1 activated the deep branch and channel 2 the superficial branch. The deep branch mainly results in dorsiflexion, the superficial branch results in eversion of the foot. The stimulation levels and timing of stimulation pulses were controlled and power was supplied by an external transmitter, which was inductively coupled to the implant [14]. The stimulation levels could be adjusted manually using potentiometers on the transmitter. The timing of the stimulation was controlled by a heel switch, assessing heel-off and heel-down instances [4].

The inertial sensor system was attached to the upper aspect of the foot (instep) (Fig. 4). It consists of two two-axial accelerometers (Analog Devices ADXL 210), perpendicularly oriented such that a three-axial accelerometer is obtained and three perpendicularly oriented rate gyroscopes (Murata ENC-03J). Fig. 4b defines the inertial coordinate system *xyz* that was used to express foot orientation and position relative to the ground. The *x*-axis was defined in the direction of progression, the *z*-axis vertically and the y-axis perpendicular to the *x*- and *z*-axes. Foot orientations are presented as Euler angles with respect to the reference orientation during midstance.

The sensor signals were sampled at a rate of 300 samples per second using a portable datalogger. Before sampling, the signals were filtered using an analog presampling low-pass filter with a cut-off frequency of 50 Hz.

3.3. Experimental procedure

Experiments were performed on one male stroke person, age 38 years, who was involved in a clinical pilot study for evaluation of the two-channel implantable drop-foot stimulator [15]. The experiments fitted in the experimental procedures of this pilot study, which was accepted by the local Medical Ethical Committee. The subject had given informed consent before his participation in the study.

The dependency of the movement parameters upon the stimulation levels of both channels was evaluated by adjusting the stimulation amplitudes of both channels to values equivalent to 0.50 and 100% recruitment levels, 0% being stimulation threshold. The equivalent stimulation levels were determined in a prior measurement of recruitment curves for both channels, assessing 3D isometric ankle moments as a function of stimulation amplitude for both channels, while the subject was seated [15].

The subject was asked to walk up and down a corridor with several combinations of the stimulation levels of both channels, evaluating the movement parameters. The dependency of the movement parameters upon the stimulation levels was subsequently evaluated for all steady state walking cycles.

As a reference, the gait of a 25 year male healthy person, walking slowly compared to his comfortable speed, was measured using the same sensory system.

The same movement analysis procedures were used on this person.

4. Results

4.1. Foot positions and orientations and derived movement parameters

Examples of sensor signals and derived foot orientations and positions with maximum and no stimulation during three steps are given in Figs. 5 and 6. The stroke subject walked at a velocity of 0.9 m/s without stimulation and 1.1 m/s with maximum stimulation. As a reference, the same is shown for a healthy person, walking at 1.2 m/s. The *x*-component of position is not shown because it is relatively large (forward progression) and not of primary interest for foot clearance. Fig. 5 depicts the measured accelerometer and gyroscope signals, Fig. 6 the derived orientation and yz-position signals. Foot-strike is clearly distinguishable by the fast change of gyroscope signal around the *x*-axis, especially in the trial without stimulation (Fig. 5d), indicating initial lateral foot-strike and subsequent fast diminishing inversion when the foot hits the ground. Please note that in the trials of the stroke subject, quite representatively, the gyroscope offsets had drifted although the sensors had been calibrated at the beginning of the experiment, resulting in the angular velocities in each direction not to be zero on average (Fig. 5d and e). The method of reconstructing foot orientations and positions presented above cancels the effect of this offset drift by applying suitable initial and final conditions for each step.

Clear differences between the orientations in the sagittal and coronal planes are evident when comparing both stimulation conditions (Fig. 6a and b). Excessive inversion during the swing phase when no stimulation is applied results in a fast change of foot orientation vector when the foot hits the ground at the end of the swing phase, as illustrated by the fast change of the x-component of the Euler angle representation of foot orientation. This corresponds to the earlier mentioned fast change of the *x*-component of the gyroscope signal, and indicates that the foot orientation is not optimally prepared for the load acceptance when the foot hits the ground, hampering gait. Also, toe lift is smaller at foot-strike when no stimulation is applied, in contrary to the case when both channels are stimulated at full recruitment level.

The trial on the stroke person with maximal stimulation on both channels (Fig. 5b and e) shows comparable orientations and *yz* movements as the healthy person (Fig. 6e and f). It should be noted however, that the gait pattern of healthy subjects may vary considerably between persons too. The healthy person, of whom the movements are shown, walks with a remarkable medial movement of the foot (positive *y* direction) at the beginning of the swing phase (Fig. 6f), which is not seen as extensively in the stroke person. Also, the reconstructed vertical position (*z*-position) becomes negative during the stance phase, especially in the healthy person (Fig.

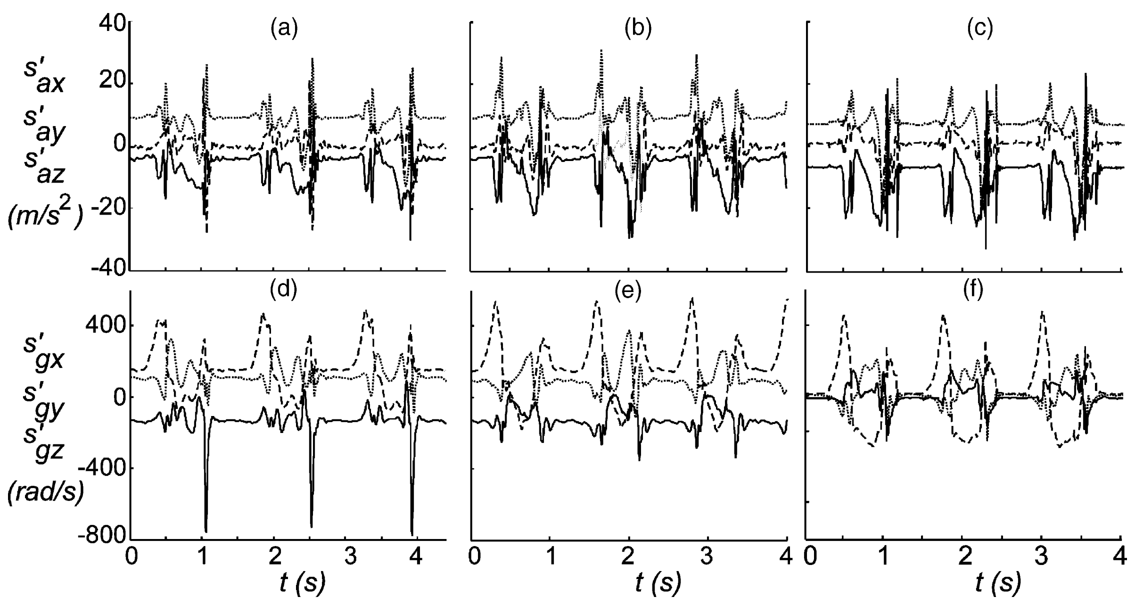

Fig. 5. Inertial sensor signals measured on the stroke person without stimulation (a, d) and with maximal stimulation on both channels (b, e). The walking velocity was 0.9 m/s without stimulation and 1.1 m/s with maximal stimulation. As a reference, the sensor signals measured on a healthy person, walking at 1.2 m/s, are shown (c, f). The 3D accelerometer signals are presented in (a), (b) and (c), 3D gyroscope signals in (d), (e) and (f). The initial calibration at the start of the experiment was applied to these signals (indicated by the accents). Please note the offset change due to drift in the gyroscope signals in the measurement of the stroke subject. The sensor signals are represented in sensor coordinates. The signals are marked according to the corresponding axis during mid stance: The signals in the direction of the *x*-axis (direction of progression) are indicated by solid lines, *z*-axis signals (vertical) by dotted lines and *y*-axis signals (perpendicular to *x* and *z*) by dashed lines.

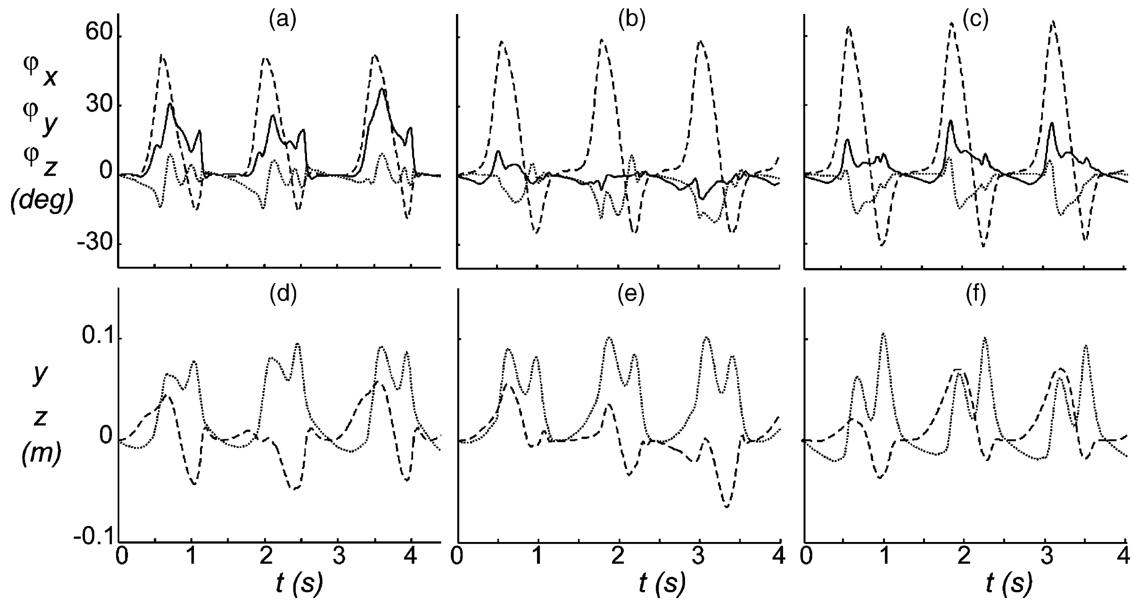

Fig. 6. Reconstructed foot orientations and *yz*-positions for the troke subject without stimulation (a, d) and with maximal stimulation at both channels (b, e). As a reference, the reconstructed foot orientations and positions on the healthy person are shown (c, f). Reconstructed orientations are presented as Euler angles with respect to the reference orientation during mid-stance in (a), (b) and (c). The *yz*-positions are presented in (d), (e) and (f). The reconstructed orientations and positions are expressed in inertial coordinates. The signals in the direction of the *x*-axis (direction of progression) are indicated by solid lines, *z*-axis signals (vertical) by dotted lines and *y*-axis signals (perpendicular to *x* and *z*) by dashed lines.

6f). This may be due to the fact that the beginning of each step was chosen rather early in the stance phase. During push-off, the height of the sensor, which is on the instep of the foot, may become lower than during early stance.

Fig. 7 indicates how the movement parameters considered in this study are derived from the orientation and position signals. They concern sagittal and coronal orientations of the foot just before foot-strike.

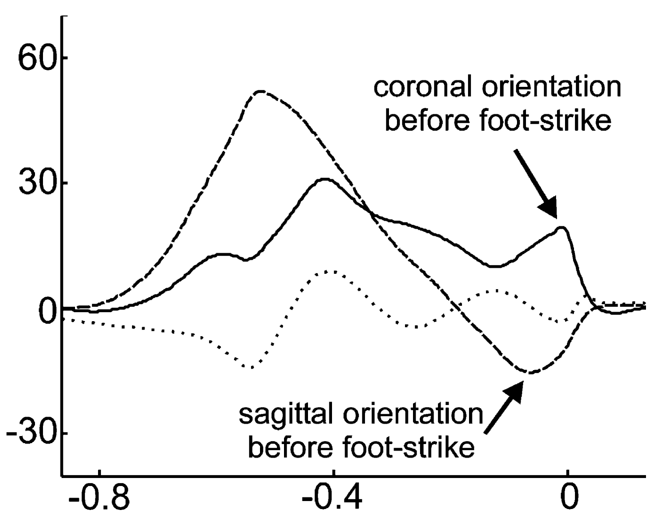

Fig. 7. Relevant parameters of foot movement can be defined and evaluated from the orientation and position data reconstructed from the 3D inertial sensor signals. The movement parameters considered in this paper, sagittal and coronal orientations of the foot just before foot-strike, are indicated in the reconstructed orientations for one sample stride with no stimulation. Signals in *x*-direction are indicated with solid lines, *y*- axis dashed and *z*-axis dotted. Time zero indicates the moment the foot is flat on the ground.

4.2. Influence of the stimulation levels on the movement parameters

When considering the influence of the stimulation levels of both channels on the movement parameters (Fig. 8) it is evident that channel 1 (deep branch of peroneal nerve) has a major influence on foot movement in the sagittal plane just before foot-strike (Fig. 8b), while this movement parameter is not dependent on the stimulation level of channel 2 (superficial branch of peroneal nerve). In contrast, both channels influence coronal foot orientation just before foot-strike (Fig. 8a). Note that these orientations are expressed relative to the floor and that, according to the applied sign convention, a negative sagittal plane orientation corresponds to a foot orientation with toes lifted with respect to the heel (Fig. 8b).

5. Discussion

5.1. Accuracy of derived foot orientations and positions

The accuracy of the derived foot orientations and positions depends on the validity of the assumed initial conditions and the performance of the inertial sensor system. The absolute accuracy remains to be compared with a reference 3D movement analysis system. However, for

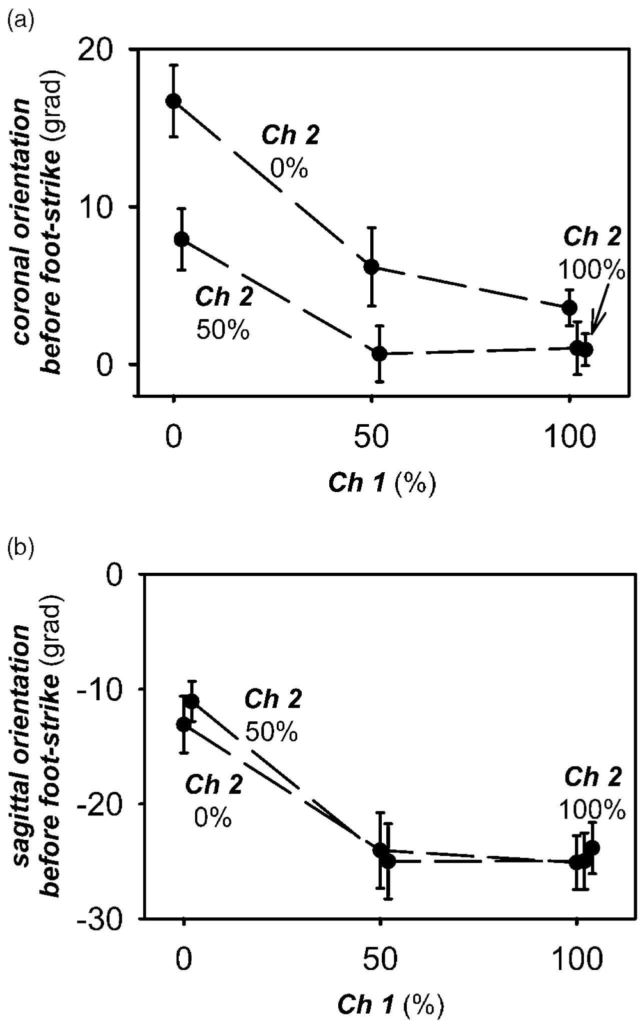

Fig. 8. Influence of stimulation levels on the movement parameters defined in Fig. 5 for one stroke person walking with the implanted dual-channel drop-foot stimulator. Mean orientations in the coronal (a) and sagittal (b) planes just before foot-strike and corresponding standard deviations are shown. The recruitment level (0, 50, 100%) of channel 1 (Ch 1) is presented along the horizontal axis, the recruitment level (0, 50, 100%) of channel 2 (Ch 2) is represented as a parameter of the curves. Please note that 100% recruitment of channel 2 was only applied in combination with 100% recruitment of channel 1, In the case of 0% recruitment threshold stimulation was applied. Note that these orientations are expressed relative to the floor and that, according to the applied sign convention, a negative sagittal plane orientation corresponds to a foot orientation with toes lifted with respect to the heel.

this specific application it is important to note that the inertial sensor system and the algorithm for calculating position and orientation is sufficiently sensitive to measure the influence of stimulation levels on the foot movements of drop-foot patients. Earlier experiments learned that the error in the estimation of walking distance is below 3%. This would result in a maximal error of 3 cm in x-direction for a stride of approximately 1 m. The errors in y and z-direction presented in this study are expected to be considerably lower, since the initial and end conditions in these directions are more stringent than in x-direction: the z displacement over a stride is assumed to be zero and the walking direction is defined such that the net y displacement is zero. The relative orientation errors are expected to be considerably smaller than the errors in position, since the calculation of positions follows from the orientation calculations after several operations which are expected to contribute markedly to the overall position errors. They include the coordinate transformation on the basis of calculated orientations, the subtraction of gravity acceleration and the double integration of acceleration to position, constrained by initial and end conditions.

The reduction of integration drift by application of initial and end conditions for a stride assumes walking over a horizontal flat surface. Increased errors can be expected in the case of stair walking or walking on irregular surfaces. With regards to the validity of the initial conditions it is important to correctly identify the moment in time in which the patient is in single-stance on the sensor-equipped foot. This phase is indicated above as ‘mid-stance’. During single-stance the full weight of the body rests on one foot and it can be expected that there will be no, or little, motion of the foot with respect to the ground and the foot will be flat on the ground. Hence the initial conditions are satisfied optimally and any moment during single-stance can serve as the start of integration (t=0). During stance, the assumption of zero velocity may not be exactly met at the position of the sensor system, because the foot continues to move even if the sole of the foot is stable. The deviation of this assumption is expected to be largest in walking direction (x-direction).

5.2. Interpretation of dependency of foot movement parameters upon stimulation levels

Channel 1 results in dorsiflexion and reduction of inversion. Channel 2 results in additional reduction of inversion. This agrees with anatomical knowledge that channel 1 (deep branch of the peroneal nerve) mainly activates tibialis anterior, which has a major dorsiflexion contribution. It also agrees with the fact that channel 2 (superficial branch of the peroneal nerve) activates, among others, the peroneal muscles, which mainly contribute to eversion around the ankle. It should be noted that gravity acts as antagonist to these muscles during the swing phase, resulting in plantar flexion and inversion.

When tuning the stimulator, dorsiflexion should first be adjusted with channel 1. Subsequently, remaining inversion can be corrected with channel 2. This procedure can be used for the automatic tuning procedure proposed in Fig. 2.

The movement restrictions and effect of stimulation levels on foot movements may differ between subjects, although the general effect of stimulation of deep and superficial branches of the peroneal nerve on dorsiflexion and eversion will remain.

It should be noted, that the stimulation of muscles acting around the ankle does not only influence the orientation of the foot, but may also influence the whole gait pattern of the user, because he or she will adapt to the imposed changes. For example, it may affect the amount of lifting of the foot and the circumduction of the affected leg, which are both under the voluntary control of the user.

Furthermore, foot orientation is not only influenced by the muscles innervated by the branches of the peroneal nerve, but also by the calf muscles, which are normally nearly inactive during the swing phase. However, in the case of hyperreflexia, spastic contractions of the calf muscle may influence foot movement. This is especially evident during the stance phase if the forefoot first lands on the ground, generating excessive stretch reflexes which may even result in clonus [18]. During the swing phase these reflexive components may also be present, for example if the calf muscle is stretched above the stretch-reflex threshold velocity when stimulating the dorsiflexors at the beginning of the swing phase. This is one of the reasons why a balanced and well-tuned stimulation of the dorsiflexors during the swing phase is required.

In conclusion, the current study indicates that foot movements can be reconstructed from 3D inertial sensor signals such as to distinguish the change of foot movements during swing when the stimulation parameters of the two channels are adapted. This sensitivity is a prerequisite for an automated balancing system for both stimulation channels of the dropfoot stimulator. The feasibility of such an automated balancing system is still to be demonstrated.

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