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Ground angle estimator from an ankle foot orthosis based on strain sensing and Fourier series

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Abstract— **A ground angle estimation technique for use on ankle-foot-orthosis AFO, during gait is proposed. Strain gauge sensors were mounted on a foot orthosis in order to give information about strain in the sagittal plane. The ankle angle of the orthosis was fixed. Strain characteristics were therefore changed when walking on slopes. It was investigated if strain information could be used for detection of inclination and estimation of inclination angle. With a Fourier series representation of the strain during a gait cycle, ground angle at different walking speeds and inclinations could be estimated with similar accuracy as previous studies using kinematically based estimators. This indicates that embedded strain sensors can be used for online control of future orthoses with inclination adaptation.**

I. INTRODUCTION

ANKLE and foot muscles disorders affect the human gait and are commonly treated with orthoses to partially compensate functional loss. Typical assisting functions are: Provide ankle stability during stance, simulate push off effect during late stance, keep the toes off ground during swing, assist poor functional muscles and decrease pain by limiting motion. Orthoses have been passive and purely mechanical. However, decreasing size and cost of electronics have made it possible for active solutions of assistance.

These approaches are functional electrical stimulation(FES), see e.g. [16] [9], or orthotic control [2]. FES activates the dorsiflexion[1] muscles in the back ankle with electricity. The objective is to provoke foot lifting just in time for swing phase.

However, existing systems are still limited in their capability of adapting to new inclining circumstances i.e. hills or heel hight variation.

Studies have shown that when able-bodied walked uphill, a tilting of the trunk can be observed and this was assumed to help in creating a forward momentum [7],[8],[6]. These compensations due to uphill walking are primarily adjusted in ankle foot system [4]. While in downhill, compared to level walking, the joints have to absorb more energy caused by the combined forward and downward movement (see e.g. [5],[7],[10]). This can be seen by a shorter stride length [7] and is compensated, mostly by the knees and somewhat by the hip. It is stated in [8] that people with reduced range of motion in one joint, their other joints compensate this. However, with elderly, orthotic and prosthesis users, these changes in flexion may not be possible. Many of the assisting devices have fixed ankle positions and attempting to move the body's center of mass forward may cause a sense of *instability* when walking on inclinations. Our hypothesis is that, *using an AFO with a fixed ankle angle, adjusted to be comfortable at horizontal gait an extra (uncomfortable) torque will be introduced when walking in inclinations*. This would therefore be measurable and useful for estimating the ground angle. In the future this could be used in ankle control for adapting to the ground inclination.

In previous studies, for slope estimation, kinematical sensors have been used i.e. gyroscopes and accelerometers. Aminian *et al.* used feedforward neural networks to estimate both incline (98% correct) and speed of walking [1]. Their input data was 10 summary statistics measurements (median, variance etc.) used from accelerometers at the trunk. Rehbinder and Hu [11] observed that at low accelerations in gait, an accelerometer can be used as gravity sensor. The methods applicability on human gait has been shown by e.g. [12] and [13]. By positioning the sensor on the foot/shoe sole ground angle was estimated during stance when the foot was still. The problem with estimation ground angle using kinematical sensors is that the sensors must be motionless. Thus only the stance phase can be used. On uneven ground the sensors will always move during stance.

Torque is typically estimated from measuring ground reaction forces and ankle angle. Instead of using previously mentioned sensors, the use of strain gauge sensors as an indirect measurement is investigated. Another feature of strain sensors is the small size that allows embedded constructions (see Fig.1). Furusho *et al.* developed a controllable brake for ankle orthoses with force and bending sensors [3]. They showed that by positioning strain sensors on the orthosis shank, gait phases could be detected for ankle control in normal walking situations. We have in a previous report shown that strain sensing changes for different inclinations [15]. In that study was maximum, minimum and mean during a cycle of strain compared for different inclinations. It was shown that strain sensing gave significant information about uphill walking. At a known walking speed ground angle can be estimated for uphill walking. In this paper, a Fourier series representation is used for strain gait cycle. With this representation, a slope angle estimator can be constructed with the same accuracy as for kinematical sensors. It does not depend on stance phase and is thereby less sensitive to ground variations. However, the main improvement is that walking speed information is not needed.

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[1]Here we use the term *dorsiflexion* when describing ankle angles between toe and calf which are less than 90 degrees.

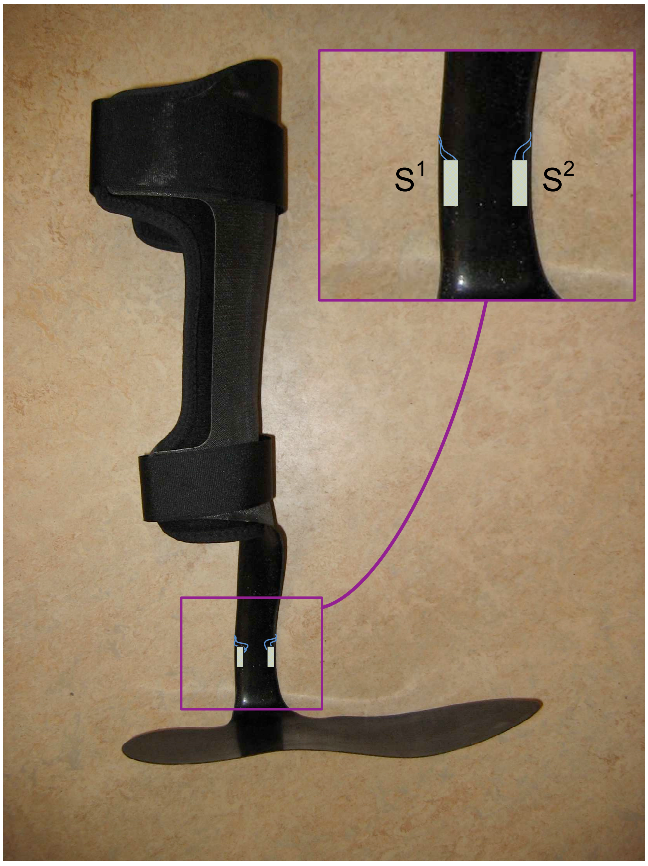

Fig. 1. Strain measurement set up. One strain gauge sensor, S^1, is attached partly on the inside at back and one, S^2, partly on the inside at the front. Sensors S^3 and S^4 are attached at the corresponding positions but on the outside.

II. METHODOLOGY

A. Setup

The studies were done using a solid, lightweight carbon ankle foot orthosis. Four strain gauge sensors were glued 60mm above the orthotic sole as shown in Fig 1. The sensor signals were combined into one via a (full Wheatstone) bridge. The combined amplified sensor signal was sampled with an off-the-shelf *PIC* processor with a 10-bit AD-converter at 50Hz sampling frequency. The signal was logged into a PC with *Sysquake* software using *Bluetooth* modules. The sensors where positioned to measure gait in the sagittal plane[2].

B. Gait cycle detection

Both gait cycle and events can be estimated from the strain sensor. The cycle is here defined as the period between two consecutive events. With strain signal S defined as negative in dorsiflexion and introducing the thresholds $\gamma_1 < \gamma_2$ a gait cycle start is found by:

[2]Sagittal plane is a plane that is the median vertical longitudinal plane that divides a bilaterally symmetrical human into right and left halves or planes parallel to that.

Let $t_{start}(0)$ be any initial sampling instant. Find the start of gaitcycle k, defined at $t_{start}(k)$ by using auxiliary sampling instant $t_{aux}(k)$, according to

$$\begin{aligned} t_{aux}(k) &= \arg \min_{t > t_{start}(k-1)} S(t) < \gamma_1 \\ t_{start}(k) &= \arg \min_{t > t_{aux}(k)} S(t) > \gamma_2 \end{aligned}$$

The strain during one cycle k can be denoted with a N dimensional vector $\mathbf{s}_k$

$$\mathbf{s}_k = \begin{pmatrix} S(t_{start}(k)) \\ \vdots \\ S(t_{start}(k+1)-1) \end{pmatrix} \in \mathcal{R}^N \quad (1)$$

The index k will be omitted for simplicity except when the gait number is specific for the expression. By choice of thresholds t_{start} comes close in time to *Toe off*.

C. Strain signal characteristics in a gait cycle

The dorsiflexion torque applied on the orthosis was assumed to increase during uphill walking but be unaffected during down hill compared to horizontal walking. These changes in torque caused a change in orthotic strain.

Another way to characterizing gait cycles is to use Fourier series by assuming that the strain is a periodic signal $S(t+N) = S(t)$ of length N. If $\Omega = 2\ \pi/N$ is the normalized frequency and introducing the column vectors

$$a = \begin{pmatrix} a_1 \\ \vdots \\ a_n \end{pmatrix}, b = \begin{pmatrix} b_1 \\ \vdots \\ b_n \end{pmatrix} \quad (2)$$

the Fourier coefficients can written in matrixform

$$\begin{aligned} a_0 &= \tfrac{1}{N}\textstyle\sum_{i=0}^{N-1} \mathbf{s}(i) \quad = mean[s] \quad (3) \\ a &= \frac{2}{N}\mathcal{C}^T s \\ b &= \frac{2}{N}\mathcal{S}^T s \end{aligned}$$

where the matrixes $\mathcal{C} \in \mathcal{R}^{N\times n}$ and $\mathcal{S} \in \mathcal{R}^{N\times n}$ have the elements $\mathcal{C}_{ij} = \cos(ij\Omega)$ and $\mathcal{S}_{ij} = \sin(ij\Omega)$. Furthermore, letting $\mathbf{1} = [1\dots1]^T \in \mathcal{R}^{N\times 1}$ the basis functions and representation are

$$E = [\ \mathbf{1}\ \ \mathcal{C}\ \ \mathcal{S}\ ],\ x = \begin{pmatrix} a_0 \\ a \\ b \end{pmatrix} \quad (4)$$

The strain can thus be approximated as

$$\hat{\mathbf{s}} = Ex \in \mathcal{R}^{N\times 1} \quad (5)$$

Its also noteworthy that the approximation introduces a lower dimension. Although $\mathbf{s}, \hat{\mathbf{s}} \in \mathcal{R}^N$ a cycle can be represented by

$$x = arg \min_x \|\mathbf{s} - \hat{\mathbf{s}}\|^2 \in \mathcal{R}^{2n+1} \quad (6)$$

That is $dim(x) = 2n+1 < dim(\mathbf{s}) = N$ where N varies with speed while n is constant.

D. Inclination estimation

Model the ground angle ϕ_k as a linear combination of the Fourier coefficients

$$\hat{\phi}_k = x_k^T \theta \quad (7)$$

and minimize the modeling error

$$e_k = \phi_k - \hat{\phi}_k$$

by least squares method

$$\theta = arg \min_{\theta} \sum e_t^2$$

The ground angle can then be estimated by (7). With M steps data

$$\begin{aligned} \phi &= [\ \phi_1 \ \ \ldots \ \ \phi_M\ ]^T \\ X &= [\ x_1 \ \ \ldots \ \ x_M\ ]^T \\ \rightarrow \phi &= X\theta + e \end{aligned} \quad (8)$$

the parameters in θ are identified by linear regression:

$$\theta = arg \min_{\theta} \|\phi - \hat{\phi}\|^2 = (X^T X)^{-1} X^T \phi$$

Notice, that by including data corresponding to different walking speeds the estimator will be insensitive to speed variations. To quantify the estimation quality, the root mean square error RMSE:

$$RMSE = \sqrt{\frac{\sum_{k=0}^{M-1} (\phi_k - \hat{\phi}_k)^2}{M}} \quad (9)$$

is used for M steps data.

E. Experiments

The experiments were conducted on one able-bodied man walking continuously at various speeds and inclinations. Walking was done at two different speeds: 1m/s and 0.5 m/s and five different inclinations: uphill at 4.7 and 2.45 degrees, horizontal (0 degree) and downhill at -2.6 and -4.8 degrees. Each walking condition was measured during 90 s. The data set was split as $60/40\%$ used for estimation and validation respectively. Typical sensor data from one gait cycle for different ground inclination are shown in Fig. 2.

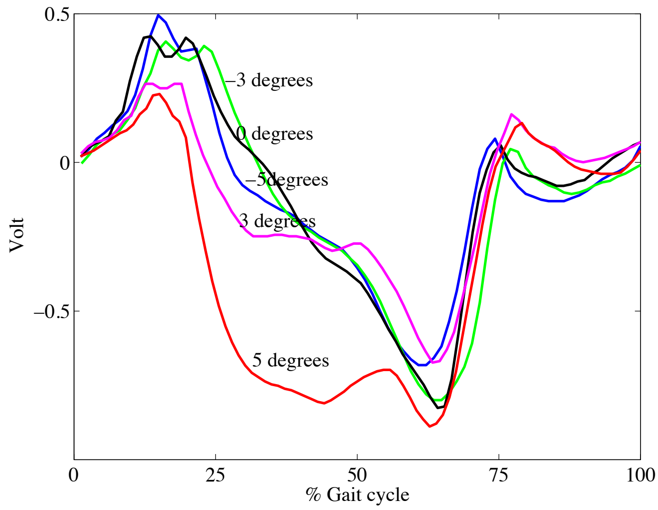

Fig. 2. Sensor data from one gait cycle, from Heel strike to Heel strike of orthotic foot for different ground inclination angles: from 5 degree down hill (-5) to 5 degree uphill (5).

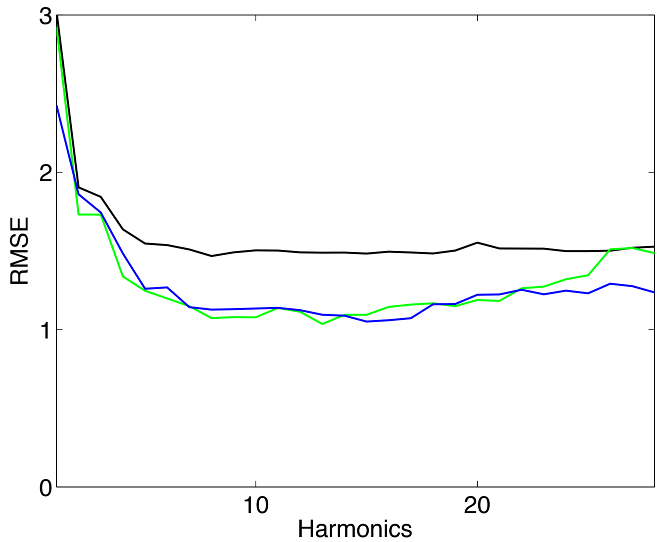

Fig. 3. $RMSE$ at different Fourier series harmonics of the gait cycle with a model for slow walking speed(blue), fast (green) and both speeds simultaneously (black).

III. Results

With the Fourier coefficient representation the ground angle estimation varied with the dimension of input vector x. The $RMSE$ for different harmonics n of the gait cycle are shown in Fig. 3. For small harmonics the $RMSE$ was larger than at higher harmonics. Thus most information was found in the lower frequencies and an increasing dimension of input data did not significantly improve the estimation. This would also explain why the mean values, used in the previous study [15], was not sufficient as estimator measurement. The size of x was chosen to $n = 8$ (= 17 coefficients) which was the $RMSE$ minimum. It could be seen, as in Fig. 4, that this reduced representation captures the most of the strain signal behavior.

Constructing an estimator for each speed $RMSE$ was approximately 1 degree both at uphill and down hill (see Figures 3 and 5). Making a speed independent estimator the error increased to almost 1.5 degree. This higher $RMSE$ was mainly caused by a decreased possibility of estimating down hill walking at lower speeds (see Fig. 6). Still this error was in the same size as kinematically based estimators [14].

IV. Conclusions

An embedded measurement system for foot orthosis during gait is proposed. Strain gauge sensors were mounted on a foot orthosis in order to give information about strain in the sagittal plane. With a fixed orthotic ankle angle, strain is changed when walking in hills.

With a Fourier series representation of the strain during a gait cycle, the ground angle at different walking speeds could be estimated with similar accuracy as previous studies using

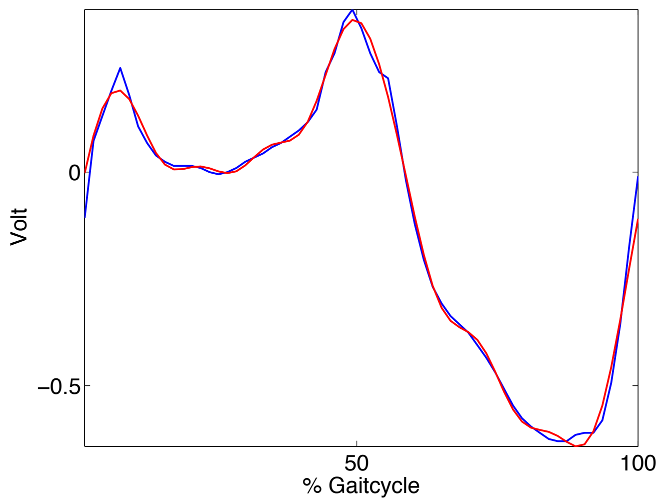

Fig. 4. Measured strain (blue) during one gait cycle and its Fourier series approximation(red) using 8 harmonics.

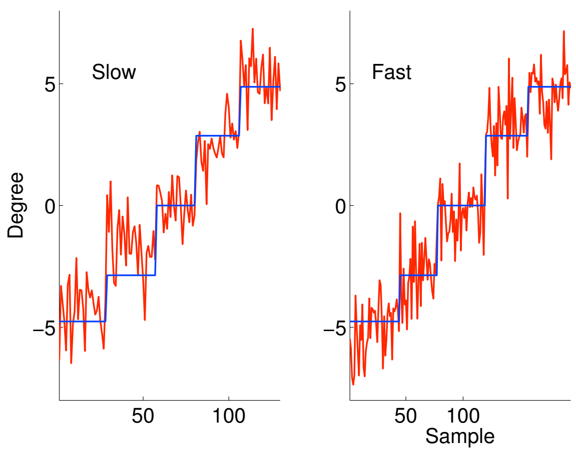

Fig. 5. Ground angle estimation $\hat{\phi}$ (red), using Fourier series, at different ground angles (blue) at two different speeds. The estimators are identified individually for each speed. Left estimation tuned for slow speed (0.5m/s). Right estimation tuned for fast speed (1.0 m/s).

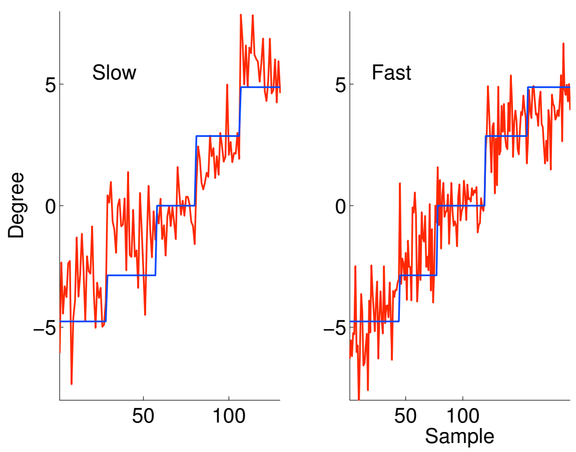

Fig. 6. Ground angle estimation $\hat{\phi}$ (red), using Fourier series, at two speeds on different ground angles (blue). One single estimator is used, tuned for both slow and fast gait.

kinematically based estimators. Although no significantly variations in the strain size could be observed, in comparing downwards to level walking, ground angle estimation was possible. Further improvements are possible by averaging estimations in time. This sensing technique may open possibilities for active control of foot prostheses and orthoses.

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