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

Wolfgang Svensson and Ulf Holmberg

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 dorsiflexion1 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 height 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. Rehinder 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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1Here we use the term dorsiflexion when describing ankle angles between toe and calf which are less than 90 degrees.
Let $t_{\text{start}}(0)$ be any initial sampling instant. Find the start of gait cycle $k$, defined at $t_{\text{start}}(k)$ by using auxiliary sampling instant $t_{\text{aux}}(k)$, according to

\[
\begin{align*}
    t_{\text{aux}}(k) &= \arg \min_{t>t_{\text{start}}(k+1)} S(t) < \gamma_1 \\
    t_{\text{start}}(k) &= \arg \min_{t>t_{\text{aux}}(k)} S(t) > \gamma_2
\end{align*}
\]

The strain during one cycle $k$ can be denoted with a $N$ dimensional vector $s_k$

\[
s_k = \begin{pmatrix}
S(t_{\text{start}}(k)) \\
\vdots \\
S(t_{\text{start}}(k+1) - 1)
\end{pmatrix} \in \mathbb{R}^N 
\]

The index $k$ will be omitted for simplicity except when the gait number is specific for the expression. By choice of thresholds $t_{\text{start}}$ 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

\[
a = \begin{pmatrix}
a_1 \\
\vdots \\
a_n
\end{pmatrix},
b = \begin{pmatrix}
b_1 \\
\vdots \\
b_n
\end{pmatrix}
\]

the Fourier coefficients can written in matrix form

\[
a_0 = \frac{1}{N} \sum_{i=0}^{N-1} s(i) = \text{mean}[s] \\
\begin{pmatrix}
a \\ b
\end{pmatrix} = \frac{2}{N} S^T s
\]

where the matrices $C \in \mathbb{R}^{N \times n}$ and $S \in \mathbb{R}^{N \times n}$ have the elements $C_{ij} = \cos(ij\Omega)$ and $S_{ij} = \sin(ij\Omega)$. Furthermore, letting $I = [1 \ldots 1]^T \in \mathbb{R}^{N \times 1}$ the basis functions and representation are

\[
E = \begin{bmatrix} 1 & C & S \end{bmatrix}, \quad x = \begin{pmatrix} a_0 \\ a \\ b \end{pmatrix}
\]

The strain can thus be approximated as

\[
\hat{s} = Ex \in \mathbb{R}^{N \times 1}
\]

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

\[
x = \arg \min_x \| s - \hat{s} \|^2 \in \mathbb{R}^{2n+1}
\]

That is $\dim(x) = 2n + 1 < \dim(s) = N$ where $N$ varies with speed while $n$ is constant.
D. Inclination estimation

Model the ground angle \( \phi_k \) as a linear combination of the Fourier coefficients

\[
\hat{\phi}_k = x^T_k \theta \tag{7}
\]

and minimize the modeling error

\[
e_k = \phi_k - \hat{\phi}_k \]

by least squares method

\[
\theta = \arg \min_{\theta} \sum e^2
\]

The ground angle can then be estimated by

\[
X \theta + e
\]

the parameters in \( \theta \) are identified by linear regression:

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

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:

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

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: 1 m/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.

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 downhill (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
measured strain $p$ during one gait cycle and its Fourier series approximation using 4 harmonics.

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

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ground angles at two different speeds. The estimators are identified individually for each speed. Left estimation tuned for slow speed (0.5 m/s), Right estimation tuned for fast speed (1.0 m/s).

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**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 (0.5 m/s), Right estimation tuned for fast speed (1.0 m/s).

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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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**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.

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