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Abstract— **An embedded measurement system for foot orthosis during gait is proposed. In this paper strain gauge sensors are mounted on a foot orthosis in order to give information about strain in the sagital plane. The ankle angle of the orthosis is fixed. Strain characteristics are therefore changed when walking on slopes. It is investigated if strain information can be used for detection of inclination and estimation of inclination angle. Also walking speed influence is studied. It is shown that strain sensing only gives significant information about up hill walking. At a known walking speed ground angle can be estimated for up hill walking.**

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. But 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. [17] [11], or orthotic control [2]. FES activates the dorsiflexion muscles in the back ankle with electricity. The objective is to provoke foot lifting just in time for swing phase.

Active control of ankles is done in orthotic devices for drop foot control. Both orthotic control as orthotic rehabilitation device using pneumatic actuators [3] and with an electrical motor [2] have been proposed. The former was designed for training in hospitals or labs with compressed air facilities. The latter uses a linear motor which controls the position of a spring. This enables force or torque control across the ankle joint. The actuator minimizes the impedance in late stance to allow plantarflexion movement and introduces dorsiflexion during swing providing toe clearance. An alternative approach is the proposed ankle actuator which consists of two sets of spring-damper elements working separately: 1)at dorsiflexion to store energy and 2) at plantarflexion avoid foot drop [9]. Furthermore, portable pneumatic sources have been designed for exoskeletons [20] to assist humans carrying heavy loads. But 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 up hill 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. [7], [12],[5]). This can be seen by a shorter stride length [7] and is compensated, mostly by the knees and somewhat by the hip. Hansen et. al. [4] showed that the knee-ankle-foot system could be modeled as in Fig.1. It is stated in [8]

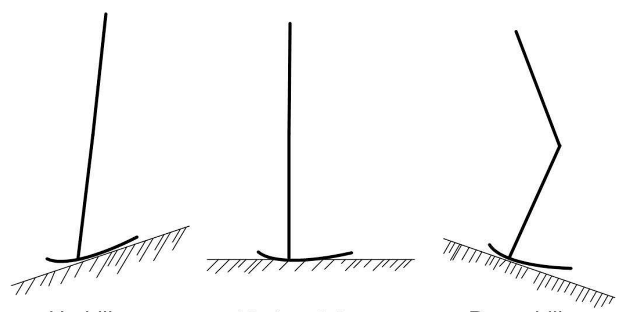

Fig. 1. Model of *Knee-ankle-foot* system [4] for different ground inclinations.

that people with reduced range of motion in one joint, their other joints compensate this. But, with elderly and prosthesis users, these changes in flexion may not be possible. Many prosthetic feet have fixed ankle position 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 *this restricted adaptation possibility causes an extra torque*. 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.

Portable gait measuring techniques are interesting both for gait analysis and active control. Automatic classification of gait phases has previously been done using various wearable sensors e.g. force resistive sensors (FSR) [15], [14], gyros [1], combinations of FSRs and gyros [11], accelerometers [19],[18] or goniometers [10]. FSRs can, although being non linear, be used as foot switches inside shoe soles for classification of stance phases. The gyro has several advantages versus accelerometers [1], [16]: 1) The gyro is not sensitive to placements as long as it is parallel to the mediolateral axis. 2) The angular velocity is less noisier since the accelerometer is a differentiated velocity. 3) The rotation can easily be estimated by integration and provides thereby spatial information. The major drawbacks though, are the sensitivity for shock due to the mechanical fastening

School of Information Science and Computer and Electrical Engineering, Halmstad University, 301 18, Sweden, Email : Wolfgang.Svensson@ide.hh.se(W.Svensson)

of the beam inside the gyro. It also requires signal processing removing the internal drift. Thus, it has a limited capability to measure ground angle.

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.2). These sensors can be valuable for active control as well as for prosthetic and orthotic design.

II. METHODOLOGY

A. Strain signal characteristics in a gait cycle

The dorsiflexion torque applied on the orthosis is assumed to increase during up hill walking but be unaffected during down hill compared to horizontal walking. This changes in torque cause a change in orthotic strain. Two different measurements were studied

- S_{mean}: The mean strain during a gait cycle which is sensitive to bias level changes in the strain signal. Here the gait cycle was from Heel rise to Heel rise.
- S_{pp}: Normalized peak to peak strain is the difference between maximum strain S_{max} and minimum strain S_{min} normalized with the time T_{maxmin} between the two peaks.

$$S_{pp} = \frac{S_{max} - S_{min}}{T_{maxmin}} \tag{1}$$

The normalization was introduced to get a more monotone relation S_{pp} to ground angle (this will be shown in the result section).

If the relation between strain S and ground angle ϕ is linear or at least linearizable, a ground angle estimation can for each sample k be expressed as

$$\widetilde{\phi}_k = aS_k + b \tag{2}$$

The parameters a and b are identified by linear regression. Low-pass filtering to reduce the variation between steps gives the estimator

$$\hat{\phi}_k = \frac{1}{N}\sum_{i=1}^{N} \widetilde{\phi}_{k-i} \tag{3}$$

where N is the filter length.

B. Measurement Setup

The studies where done using a solid light weight carbon ankle foot orthosis. Four strain gauge sensors were glued 60mm above the orthotic sole as shown in Fig 2. 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. Furthermore was a Murata 03JA gyro attached on to the shoe, at the fifth metasarsal, only used as a reference to identify gait events: Swing and Stance. All sensors where positioned to measure walk in the sagital plane.

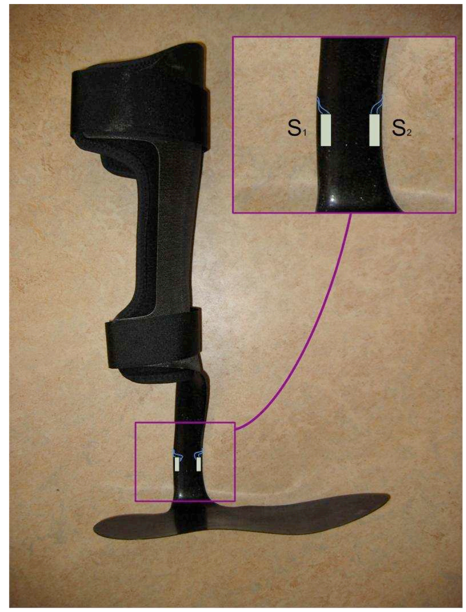

Fig. 2. Strain measurement setup. One strain gauge sensor s_1, is attached partly on the inside at front and one s_2, partly on the inside at the back. Sensors s_3 and s_4 are attached at the corresponding positions but on the outside.

C. Experiment

The experiments were conducted on one able-bodied man walking continuously at various speeds and inclinations. Horizontal walking was done at two different speeds: 1m/s and 0.5 m/s and five different inclinations: uphill at 5 and 3 degrees, horizontal (0 degree) and downhill at -3,-5 degrees. Each walking condition was measured during 90 s.

III. RESULTS

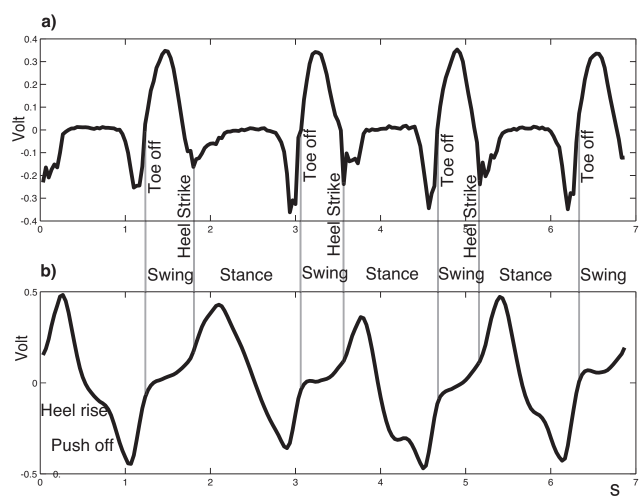

Fig. 3. Sensor data from four gait cycles: a) gyro and b)resulting signal from the strain gauges.

The resulting strain sensor signal increased at plantarflexion and decreased at dorsiflexion and is shown in Fig.3. At heel strike (gyro has a local peak) the strain increases from being unloaded as the foot is put down in front of the body. A non limited ankle would try to slightly plantar flex $(0-7\text{degrees})$[13]. Maximum strain is achieved and the opposite foot is lifted. The foot is then lowered (gyro goes to zero) and the the trunk moved forward decreasing the strain. In midstance, when the body is above the ankle, the orthotic ankle is unloaded and the strain is zero. The body continuous forward and the leg tries to rotate with a ankle dorsi flex as a negative strain. In late stance phase occurs an abrupt decrease of the strain. The Heel rise phase starts and followed by push-off. At the same time the body is in front of the ankle. This corresponds to in a non limited ankle reaching its maximum dorsiflexion torque [13]. At late stance, when only the toes are in contact with the ground, the load is reduced and the inherent orthotic elasticity brings the orthosis to unload. The reason why the strain not is constant zero during swing is assumed to be caused by the test person moving the foot inside the orthosis.

A. Up hill

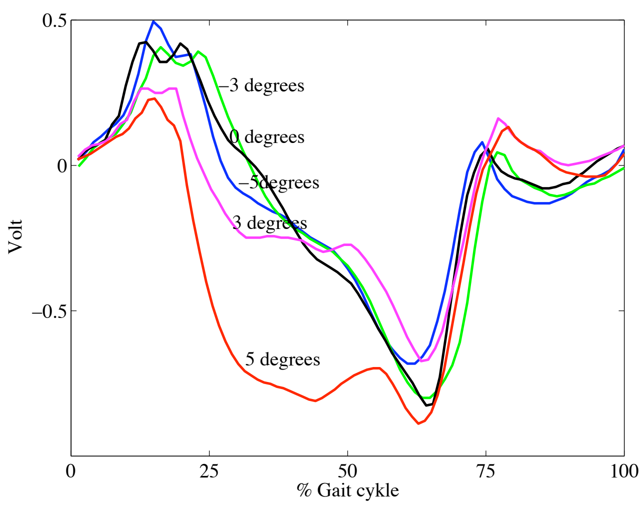

Fig. 4. Sensor data from one gait cycle, from Heel strike to Heel strike of orthotic foot for different ground inclination angles: from 5 degree down hill (-5) to 5 degree up hill (5).

During up hill walking heel strike is followed by a smaller forced plantarflexion as shown in Fig. 4. This phase is also shorter than at horizontal walking. This could support the assumption that the upper body is leaned forward [8]. If the ground angle is increased the early plantarflexion is reduced in both magnitude and time. The inclination also causes a large negative strain. As previous studies have shown is the increase of dorsiflexion a result of the compensation in the ankle foot system [4].

B. Down hill

When walking down hill no significant differences to horizontal can be observed. This supports that compensation is done at other joints [4].

C. Gait characteristics

Both gait characteristics mean and peak-to-peak show a significant change during up hill walking. In Fig. 4 it can be seen that in up hill walking (especially at 5 degrees) the strain is negative a large part of stance phase. This results in a decrease of mean strain S_{mean} as plotted in Fig.5. The normalized peak-to-peak strain S_{pp} is not influenced by the variations during swing. It was also observed that the peak positions were more distinct than heel strike.

The speed also influences the strain characteristics. It is observed that the S_{pp} difference between the two speeds is almost constant for all ground angles (see in Fig.6). This could make a speed compensation possible.

The monotone decreasing S_{pp} to ground angle relation also makes estimation possible. It is noteworthy that this peak-to-peak and ground angle relation was non monotone without the time normalization. Using eq.(3) with N=5, the estimation error was approximately one degree for uphill walking at 0.5 m/s as shown in Fig. 7. This is in the same size as kinematically based estimators although here more gait cycles were needed. The adaption time to a new slope angle was controlled by N. But at downhill there is a larger variation in the estimation although the mean estimation error does not increase. A significant separation between the walking cases for down hill was not possible.

But the gait characteristics varied between gait cycles. Further measurements, including more people, speeds and ground angles, have to be done before a complete ground angle estimator can be formulated using strain sensors.

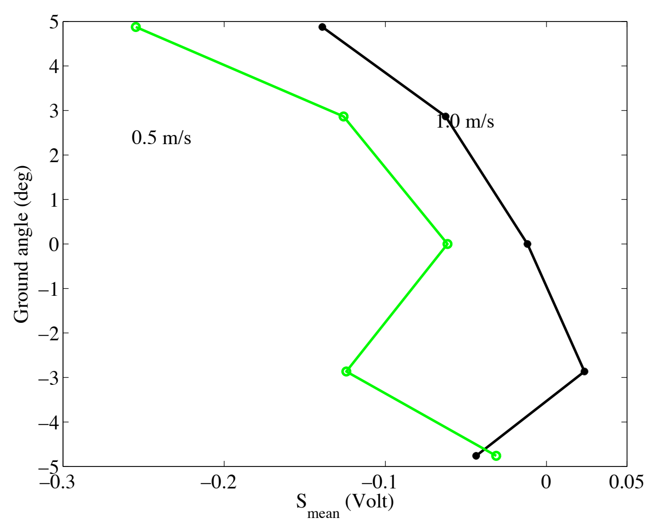

Fig. 5. The average of all gait cycles of the strain mean S_{mean} at different speeds and slope angles.

IV. CONCLUSIONS

An embedded measurement system for foot orthosis during gait is proposed. Portable cheap sensors are suitable in active control rehabilitation equipment such as prostheses and orthoses. In this paper strain gauge sensors are mounted on a foot orthosis in order to give information about strain in

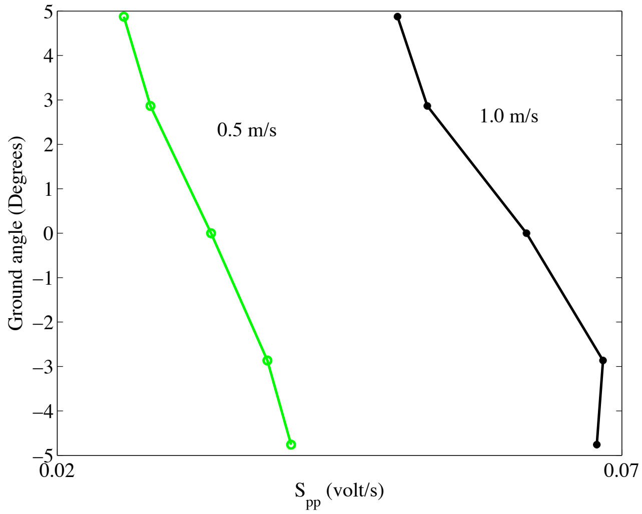

Fig. 6. The average of all gait cycles of the normalized difference S_{pp}, maximum to minimum during a gait cycle, at different speeds and slope angles.

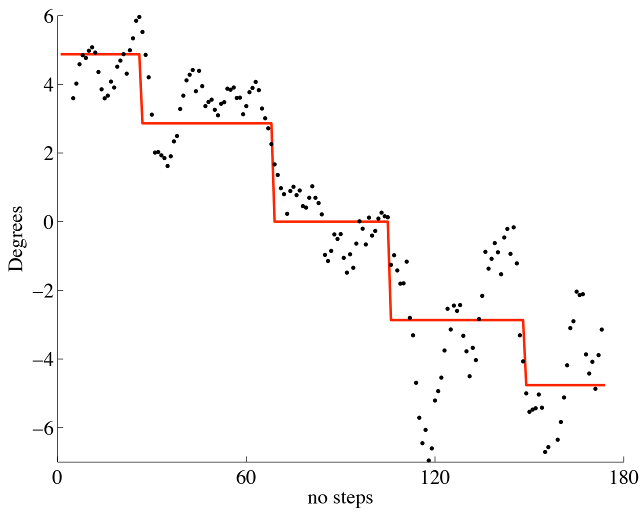

Fig. 7. Ground angle estimation $\hat{\phi}$ (dotted) and true ground angle (solid) at 0.5 m/s walking speed.

the sagital plane. With a fixed ankle angle strain is changed when walking in hills.

Two gait characteristics are studied: mean strain during a gait cycle and the difference between maximum and minimum strain when the foot is on the ground. Both show a significant change during up hill walking compared to level walking while no significant difference is shown walking down hill. One reason was the larger variation between gait cycles in down hill. This study shows similarities with previously reported studies on torque variations of human gait in inclinations. From the tests, it was also observed the speed influence on strain. The orthosis design has to be further studied and if the approach is transferable to orthotic users with muscle disorders. At a known walking speed, ground angle can be estimated for up hill walking. Measurement variations between gait cycles were reduced by a low-pass filter which improved the estimation. This sensing technique may open possibilities for active control foot prostheses and orthoses.

V. ACKNOWLEDGMENTS

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