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Stair Gait Classification from Kinematic Sensors

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1 Introduction

Gait measurement is of interest for both orthopedists and biomechanical engineers. It is useful for analysis of gait disorders and in design of orthotic and prosthetic devices. In recent years portable sensors have been studied as a complement to vision based (Morris & Paradiso, 2002). They have been used to measure both the kinematics of gait such as accelerations and angles as well as kinetics such as torques and forces. The main contribution of using portable sensors is the possibility of long time measurement of daily life situations. But the technique has also been included in active control of foot for rehabilitation. The research has mainly been focusing on the area of drop foot control see e.g. (Pappas et al., 2001: Veltink et al., 2003). The objective is to provoke foot lifting just in time for swing phase. The time to control is estimated often from gyros and force sensitive resistors. An alternative approach is to actively control foot ankle orthosis (Blaya & Herr, 2004). Recently also an active controlled foot ankle in a foot prosthesis has been studied (Svensson & Holmberg, 2006) for adapting to hill variations. But the existing systems are still limited in their capability of adapting to stair climbing. Just as orthoses, many prosthetic feet have fixed ankle position and attempting to move the body's center of mass forward may cause a sense of instability. With pressure sensors, characteristics of stair climbing and descending can be detected as the "foot down" often differs to horizontal walking. In a typical case of a sound person with two biological feet going up a stair, peak plantar pressures increase in the rear foot sensors. While at down stair there is a significant increased rate of pressure change in the frontal part. But both speed of gait and size of staircases can influence the accuracy of classification. Gyros or accelerometers have typically been attached to the waist, hip or shank. The sensors detect compensations made for different walking situations. But internal noise and temperature sensitivity of kinematical sensors tend to drift the angle estimation.

In this chapter an algorithm is presented to suit estimation of one foot angle in the sagital plane, independent on gait conditions. Only one gyro is used during swing and two accelerometers are needed for calibration during stance. Also, the sensor placement at the front of the foot avoids the need for heel strike for stance transition. Stair walking can therefore be studied. From the estimated swing trajectory three different gait conditions: up stair, horizontal and down stair are classified.

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2 Stair walking

2.1 Ascending

Movements during gait in stairs have been studied earlier (see e.g. (Andriacchi et al., 1980: McFayden et al. 1988) at hip, knee and ankle during stair walking. Normal stair ascending can be divided into three phases: weight acceptance, pull-up and forward continuation. During weight acceptance and pull-up the knee dominates with support of the hip and ankle. While, during forward continuation the ankle generates a large amount of energy. The ankle angle differs from horizontal walk mostly at the late swing phase and at the early stance. At the lift up to next staircase the edge is avoided by a small dorsiflexion and moving the knee backwards.

2.2 Descending

Also in down stair the ankle angle differs from horizontal in the swing phase when moving the limbs down (Andriacchi et al., 1980: McFayden et al. 1988). In the early stance the toes are put down before the heel. At this stage most of the energy is transferred in the knee and ankle. The hip is only dominant at the end of the downward movement of the leg into next staircase. At push-off not so large force is needed since the leg almost only has to fulfill the swing. In horizontal walk decreases the dorsiflexion when the ankle passes the lowest position during forward swing in preparation for heel down, while in stair descending this is not so crucial. Less muscle activity for vertical movements is also needed when descending.

2.3 Fixed ankle angle

Walking with fixed ankle angle, e.g. ankle foot orthoses, AFO or prostheses compensations are made with hip and knee. Studies of AFO stair walking has been done on children with neuromuscular disorders (Nahorniak et al., 1999: Thomas et al., 2002). Although there was some discrepancy found depending on test group caused by different movement strategies, differences to able-bodied could be found. An AFO prevents normal plantar flexion during weight acceptance. As solid AFO blocks dorsiflexion during forward continuation this is compensated with reduced knee flexion. Compensations at the pelvis, hip and knee mainly occurred at the late stance when lowering opposite foot to next stair. Then also the ankle flexor moment was reduced resulting in a less effective push into the next step.

There are not so many studies published about stair ambulation using prosthetic feet. But it has been observed that transtibial amputees using the *Seattle lightfoot* prosthesis have a slower velocity and asymmetrical gait pattern compared to non-amputees (Power & Boyd, 1997). This asymmetry between limbs was shown to be more significant in stair ambulation than level walking. Kinetic analysis determined significant limitations in the prosthetic ankle motion which necessitated compensatory functions at hip and knee (Schmaltz et al., 2007: Yack et al., 1999).

It has previously been reported that the dynamic-elastic-response (DER) design store energy, which is released during forward progression making it easier to run or jump and it appears not to be any significant difference between different designs during normal gait. In a study of the five most commonly used prostheses for below-knee amputation it was shown that prosthetic stair walking differs from biological (Torburn et al., 1994). But, their results showed that the dynamic response did not improve stair walking. They conclude that the reason could be that the individuals do not roll the forefoot over in the same manner as in e.g. jogging. This can be seen in the decrease in step length which restricts body weight from loading the energy storing forefoot. The shorter step length requires an increased dorsiflexion when ascending, mostly in the early stance. The limitation in the prostheses also causes a limited plantarflexion at the end of the stance phase. This is compensated by the amputees with larger anterior pelvic tilt (hip movement). Going down the DER prostheses where dorsiflexed while a biological foot is plantarflexed during heel down. But during the forward movement the flexion increased faster with a biological foot giving a larger dorsiflexion than the prosthetic. This is compensated by the amputees by less knee flexation and larger flexation in the hip.

3 Classification of stair walking

Pressure sensors have been used to characterize stair climbing and descending (Wervey et al., 1997). In comparison to level walking the stair climbing peak plantar pressures showed significant increase at the lateral foot sensors and significant decrease at all other locations. While at down stair there was a significant increased rate of pressure change in the frontal part. This work was done to understand how different particular walking activities influence the force distribution and may be useful in preventing injury to the foot.

A related area is classification of stair and level walking for monitoring daily activities especially by disabled people. Coley et al. attached a gyroscope to the shank and with wavelet transformation they were able to detect both toe-off and heel-strike (Coley et al., 2005). Furthermore, in stair descent as well as level walking there is a forward rotation of the shank about the ankle during stance while for ascent a backward rotation also is seen during foot-flat. The sign of the gyroscopic data was used to separate stair ascending from other walking conditions. The system was used on elderly hospitalized as well as healthy people in non-lab bounded surroundings. The results show that the two types of walking could be correctly classified in more than 97 percent of the cases.

With one accelerometer on each hip and ankle Kern and Schiele correctly estimated 84 percent ascending and 80 percent descending on one person. The estimator was a Bayes classifier using the mean and variance from a data window (Kern & Schiele, 2003).

Sekine et al. used a tri-axial accelerometer attached to the waist. With the wavelet transformation they were able to classify level, up and down stair walking (Sekine et al., 1998). The three types of walking patterns were categorized by comparing powers of wavelet coefficients in the vertical direction in the anterior-posterior direction. This was shown to be effective for young people (99 percent correct classification) but not for elderly people since gait changes with age. This is especially significant in lower heel down acceleration and also the shuffle which elderly do while walking. Therefore they included variance estimation and a larger difference between walking cases could be seen (Sekine et al., 2002). This was especially important looking at older people, as well as with Parkinson decease.

We propose the use of kinematical sensors attached to the shoe sole or orthosis sole. The features of using kinematical sensors are

- Foot-to-ground angle can be estimated both during stance and swing.
- The sensors do not wear out since the physical contact is limited.
- Less sensitive than pressure sensors to quality of stance phase e.g. surface roughness, balance of wearer or sensitivity in foot/skin.

By combining the signals from gyro and acceleration sensors the foot motion can be estimated and be used for classification of gait characteristics.

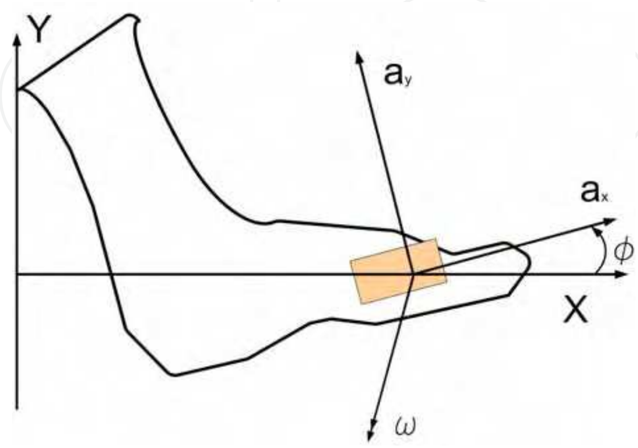

Fig. 1. Sensor setup: a gyroscope measuring ω and a two dimensional accelerometer a_x , a_y. The foot moves in the global coordinate system of X, Y.

4 Methods

The foot movement was estimated by mounting a sensor system close to the toe as shown in Fig.1. The sensor system consists of one gyroscope measuring the angular velocity, ω and accelerometers measuring a_x and a_y.
The sensors measure in a moving frame and the signals can be rotated to the fixed frame as

$$\begin{aligned} a_X &= a_x \cos\phi - a_y \sin\phi \\ a_Y &= a_x \sin\phi + a_y \cos\phi - g \end{aligned} \tag{1}$$

4.1Stance estimation

In the stationary case when the foot is at inclination ϕ the accelerometers measures are $a_x = g \sin \phi$ and $a_y = g \cos \phi$ respectively and where g is the gravitational constant. The resulting angle is then

$$\phi = \arctan \frac{a_x}{a_y} \tag{2}$$

Thus the foot-to-ground angle is defined positive for foot dorsi flexion. Since the foot is stationary during stance the angle ϕ can be estimated using the accelerometers using eq(2). The placement guarantees that the sensors are stationary during stance for all possible gait situations, even in stairs.
During swing, however, the accelerometers do not only measure the gravitational acceleration but also the acceleration of the foot. Therefore eq.(2) cannot be used for estimation of ϕ during swing. Instead, integration of the gyro signal$\omega = d\phi/dt$ can be used as an estimate.
A complete foot movement estimator should include a switching procedure between stance and swing phases.

A low pass filtered gyro signal is used to switch between the two phases:

$$\Omega_k = \frac{1}{7}\sum_{n=-3}^{3} | \omega_{k+n} | \tag{3}$$

The instance estimation motivates a symmetric low pass filter and ω is zero during stance. The detector uses $|\omega|$ since during change of rotational directional or linear movement during stair walking ω would be zero without being at stance. The transitions between the gait phases are detected by the following conditions:

- ❖ Stance: ground estimation phase. The whole foot or only the frontal part is stationary. Stance starts at sample k_{FD} when W consecutive samples of Ω are limited as

$$\Omega_k \le \gamma_s, k = k_i, \ldots, k_{i-1+W}$$
$$k_{FD} = \min_i k_i \tag{4}$$

  The threshold $\gamma_{S,}$ is chosen to be larger than the noise level at stance.
- ❖ Transitions to Swing starts with heel lift which occurs at sample k_{HL} when the condition (4) no longer is fulfilled.

Thus, the mean angle during stance is

$$\bar{\phi} = \frac{1}{M}\sum_{k=T_{start}-M}^{T_{start}-1} \arctan\frac{a_{x,k}}{a_{y,k}} \tag{5}$$

Where M=kHL-kFD and T_{start} is the start of the swing phase.

4.2 Swing estimation

At sample *k* and discrete time *h* swing integration of the gyro signal is

$$\hat{\phi}_{k+1} = \hat{\phi}_k + \omega_k h, \qquad k = T_{start}, \ldots, T_{end} \tag{6}$$

where $\hat{\phi}_{Tstart} = \bar{\phi}_{Before}$ angle at stance before swing starts from eq(5). But, to reduce the inherent bias effect, (Sabatini et al. 2005) proposed two adjustments: *i*) assuming that a step angle is almost same at the start and end the difference can be assumed to be cause by an equal noise effect. *ii*) The angle is to be adjusted during each stance using the accelerometers. A modified estimation, includes ground inclination variations between stance instances

$$\phi_k = \hat{\phi}_k \frac{T-k}{T} + \bar{\phi}_{After}\frac{k}{T}, \qquad k = T_{start}, \ldots, T_{end} \tag{7}$$

and where $\bar{\phi}_{After}$ is angle at the stance phase directly after swing and $T=T_{end}-T_{start}+1$ is the swing time.

From eq.(1) a_X , a_Y can be estimated and since the velocity is zero at stance, can by integration, v_X and v_Y be estimated as

$$\hat{v}_{X,k+1} = \hat{v}_{X,k} + a_{X,k}h$$
$$\hat{v}_{Y,k+1} = \hat{v}_{Y,k} + a_{Y,k}h \tag{8}$$

which also is compensated for bias effect as

$$v_{X,k} = \frac{T-k}{T}\hat{v}_{X,k} \tag{9}$$
$$v_{Y,k} = \frac{T-k}{T}\hat{v}_{Y,k}$$

Finally is the swing position relative the stance in the X and Y direction found with an integration of the velocity

$$y_{k+1} = y_k + v_{Y,k}h \tag{10}$$
$$x_{k+1} = x_k + v_{X,k}h$$

4.4 Classification of a step

Non gait swings are discriminated by requiring the forward movement to be large than the foot length. To further reduce the influence of estimation error the resulting inclinations was used for classification. Thus, the resulting classification variable, when x_T>0.2m is

$$\mathrm{Q_T} = \frac{\mathrm{y_T}}{\mathrm{x_T}} \tag{11}$$

Classification of stair up stair up, horizontal and stair down is then made by thresholding

Up:	$\gamma_U < Q_T$
Horizontal:	$\gamma_D < Q_T < \gamma_U$
Down:	$Q_T < \gamma_D$

4.5 Classification during swing

It would be interesting to investigate if it is possible to classify a step while it is taken. If so, it would open the possibility for online adjustment of controllable orthoses and prostheses. At the end of forward swing the largest part of vertical motion also is completed. Using the forward acceleration sensor this *swing end* can be detected. First, a time instant just before the forward acceleration is passing a lower threshold γ_L

$$\mathrm{k_L} = \min_k \arg\{a_x(k) < -\gamma_L\} \tag{12}$$

Then the time τ in swing, corresponds to the first proceeding acceleration maximum is calculated as

$$\tau = \min_k \arg\{a_x(k) > a_x(k \pm 1)\} \tag{13}$$

The foot angle can then be estimated by eq.(5-7) by replacing T with τ and assigning $\overline{\phi}_{After} = \overline{\phi}_{Before}$. In the same way the velocities $v_{X,\tau}$ and $v_{Y,\tau}$ and the positions x_τ and y_τ are estimated from eq.(8-9) and eq.(10) respectively. Thus, the resulting classification variable, when x_τ>0.2m is

$$Q_\tau = \frac{y_\tau}{x_\tau} \tag{14}$$

Classification of is also here made by thresholding.

4.6 Measurement system

The sensor system consists of one gyroscope *Murata ENC-03J* measuring the angular velocity, ω and a two dimensional accelerometer *ADXL 311* measuring a_x and a_y. Signals are sampled at 50 Hz with 10 bit AD converters with a low cost, off the shelf, 40 MHz PIC18F microprocessor. The signals are smoothed by a first order analog low-pass filter with cut-off frequency of 50 Hz. Furthermore, to analyze the system performance on-line, a *Bluetooth* unit enables logging of data to a PC.

4.7 Experiments

The evaluation was performed by letting four healthy men walk in a corridor and up and down stairs in the lab building at self selected speeds. In average each person walked 40 steps in the lab corridor. When walking in the stairs only one foot was put on each staircase. Each staircase was 17cm high and 31cm deep. The stair consisted of 10 staircases where only one foot was placed per staircase. Halting was not considered in this study. Each person walked six stairs up and six down. In between the stairs was a small platform resulting in one or two horizontal steps. Measurements were done with and without wearing a foot-ankle orthosis. Walking down stairs with an orthosis this foot was partly put in front of the staircase thereby avoiding the limitation which the ankle stiffness causes.

5 Results

The test showed several occurrences where the using $|\omega|$ avoids erroneous stance starts where the gyro signal is zero during several samples. At high speed walking though, it was hard to detect stance phases. Decreasing the window size Wmakes it easier to detect stance but resulting in poor ground angle estimations.

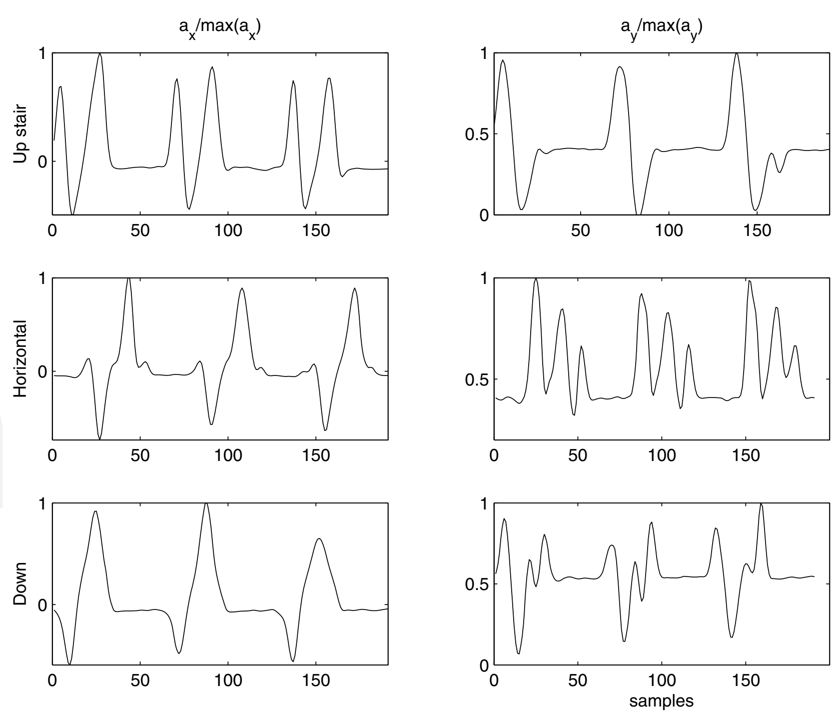

Fig. 2. Sensor signals $a_{x\ and}\, a_y$ at different gait conditions.

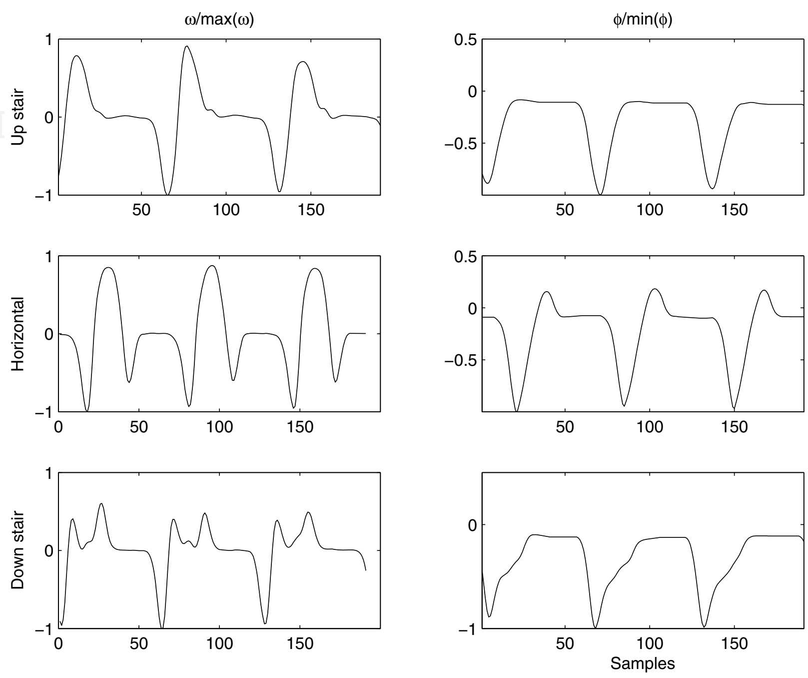

Fig. 3. Gyro signals and estimated foot-to-ground angle at different gait conditions.

From the angle estimation, as shown in Fig. 3 (Horizontal) it can be observed that the angle reaches its minimum when the heel lift phase is completed. This is followed by a calf pendulum movement forward in the swing phase which ends when the angle is maximal. The heel touches ground and the foot blade is brought down. The foot lift is similar for stair walking. During ascending is no maximum observed since the foot mostly is brought down with the toes first. Often only the frontal part is in contact with the staircase.

At descending the foot is not largely flexed since the foot forward motion brings clears the toe from ground. Here it can also be seen that the foot is adjusted so that the toe is brought down first. The time for first foot contact to ground is approximately the same for ascending and descending. The lack of heel down is seen as reduced variations at foot down.

Walking with orthoses no larger differences appear in the sensor data for the horizontal case. At ascending is the heel or at least the whole foot down. The reason why there are no more differences support the previous studies that knee and hip compensate for ankle stiffness. But at descending the foot is clearly brought down with the heel first. At stance end the foot rolls over the staircase edge and the frontal part moves down at foot lift. Only a small lift is needed to clear from the stair and start the downward motion. This rolling is partly caused by the ankle stiffness and makes the stance phase period a bit shorter.

The estimation of a_Y shows in Fig 4. a large increase when ascending to lift the foot up to next staircase. In the same way is a large descending causing a large decrease. During horizontal walk the foot accelerates upward during lift and then down during the initial forward swing. At the last part of the swing it again accelerates upwards preparing for heel down. Comparing the horizontal accelerations a_X reveal no large differences between different gait conditions and are therefore omitted.

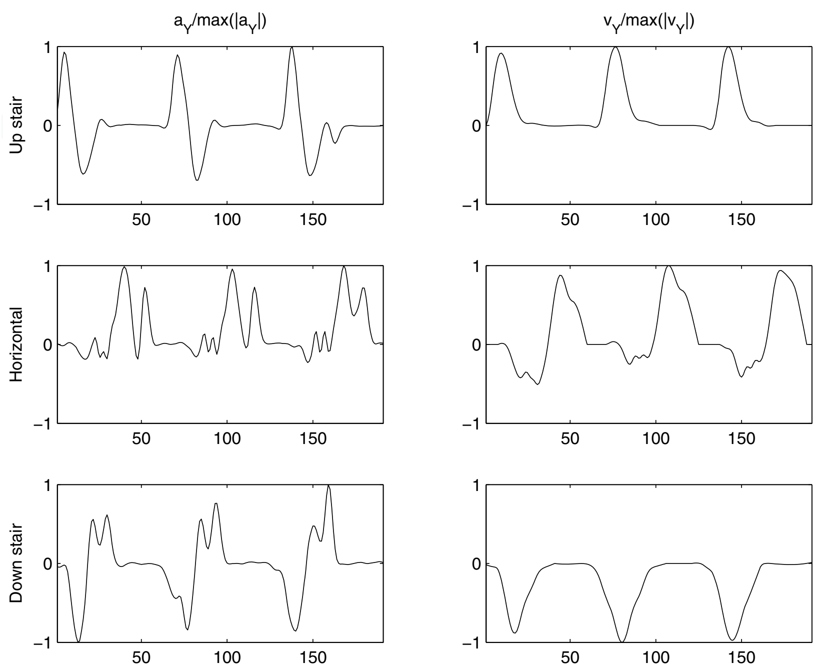

Fig. 4. Estimated vertical acceleration a_Y and velocity, v_Y at different gait conditions.

For the same reason v_X is not shown. But in the vertical velocity, as shown in Fig. 4, are the differences in gait conditions even more apparent. The positioning of the sensor causes a velocity variation which at a first glance may as if the foot was moving downwards. But this is only the sensor (due to a rounded front foot) and since the final position is of interest this is assumed to not cause any misclassifications.

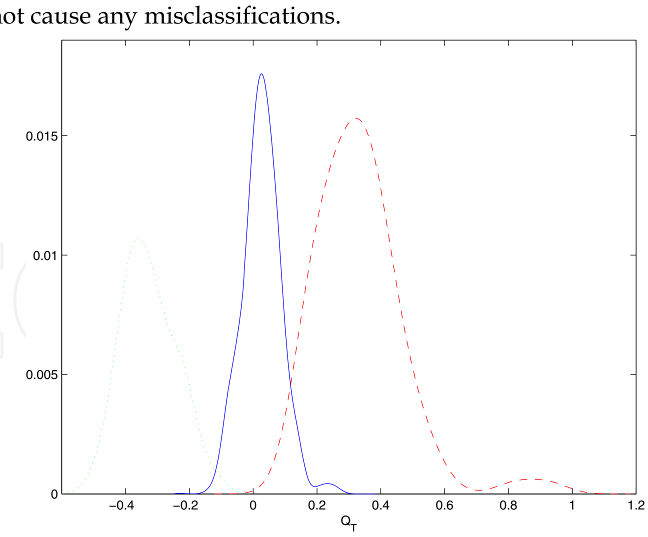

Fig. 5. Histogram of the classification quotient QT plotted using the complete step walking up-stair (red --), horizontal (blue solid) and down stair (green dotted).

In Fig. 5 is the histogram of the classification quotient Q_T plotted. The averages differ distinctly. The deviation at horizontal walk is not large. It was noted that some of the deviations were caused by rotations and sliding of the feet when walking in curves, especially at the platforms in the stairs. Possible boundaries for a Bayesian classifier and their probabilities are

Ascending : $P(Q_T > 0.12 \mid \text{Ascending}) = 96\%$
Horizontal: $P(-0.12 < Q_T < 0.12 \mid \text{Horizontal}) = 92\%$
Descending: $P(Q_T < -0.12 \mid \text{Descending}) = 93\%$

$P(Q>\gamma \mid C)$ denotes the probability of $Q>\gamma$ given class C.

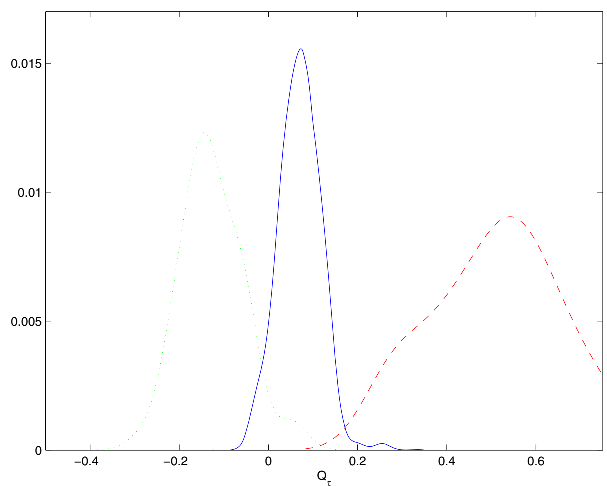

Fig.6. Histogram of the classification quotient Qt when estimating during swing step walking up-stair (red --), horizontal (blue solid) and down stair (green dotted).

Also when an estimation of motion during swing is used the different gait conditions can be separated.

Possible classification boundaries and probabilities are

Ascending: $P(Q_\tau > 0.17 \mid \text{Ascending}) = 99\%$
Horizontal: $P(-0.02 < Q_\tau < 0.17 \mid \text{Horizontal}) = 94\%$
Descending: $P(Q_\tau < -0.02 \mid \text{Descending}) = 91\%$

The estimation instance τ of using an early detection is 45 % descending, 43% horizontal and 65% at ascending of the total swing time T. Available control time is approximately 30% of T if the adjustment is to be done before the foot hits ground.

Tests were also done without using bias compensation (7) and (9). These results showed, at descending the deviation from average is too large to be used for classification. This motivates the use of a bias compensating weight. It is interesting to notice that the classification improves for ascending and horizontal gait when classifying during swing is used rather than waiting for the complete step. The bias compensator in these cases seems appropriate.

6 Discussion

This chapter describes a compact portable measuring system which both can be used for analysing able-bodied as well as foot orthosis control. Mounting the sensors on the frontal part of the foot enables detection of stance phase from the gyro signal in stair walking. By sensor fusion a model-free gait classifier be can used. Also an on-line motion estimator makes classifying during swing possible.

7 Future research

Test on able bodied people have shown promising results. Sekine et al. recognized that people with malfunctioning lower extremities are harder to classify. Further studies on patients should be done to show if orthotic walk is distinct. Moreover, a complete system has to include classification of gait or no gait conditions. Finally has the usability for control to be developed.

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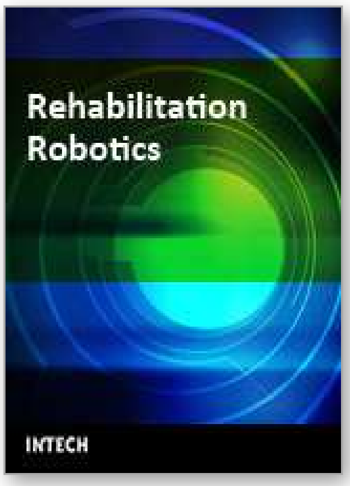

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The coupling of several areas of the medical field with recent advances in robotic systems has seen a paradigm shift in our approach to selected sectors of medical care, especially over the last decade. Rehabilitation medicine is one such area. The development of advanced robotic systems has ushered with it an exponential number of trials and experiments aimed at optimising restoration of quality of life to those who are physically debilitated. Despite these developments, there remains a paucity in the presentation of these advances in the form of a comprehensive tool. This book was written to present the most recent advances in rehabilitation robotics known to date from the perspective of some of the leading experts in the field and presents an interesting array of developments put into 33 comprehensive chapters. The chapters are presented in a way that the reader will get a seamless impression of the current concepts of optimal modes of both experimental and ap- plicable roles of robotic devices.