ACCELEROMETRY: A METHOD FOR ANGLE ASSESSMENT OF THE LOWER EXTREMITIES WITH THE POTENTIAL OF IMPLANTATION

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ABSTRACT

A fully implantable FES system, for the restoration of locomotion with feedback control, needs an implantable angle sensor. We investigate the use of accelerometers having this potential of implantation for the assessment of the angles of the lower extremities. A theory for angle measurements based on the use of pairs of accelerometers was developed. This theory is validated with measurements during stance and walking.
We show that during stance phase accelerometers provide the lower leg angle without an integration. Furthermore they can be used to distinquish the different phases of walking.

INTRODUCTION.

Electrical stimulation of paralyzed muscles to restore functional movements (FES) can benefit from feedback control strategies [1,2]. Parameters that can be used for a feedback controller are, in the case of lower extremities, angle, angular velocity and angular acceleration of the various leg segments as well as the various muscle forces. This study concerns the assessment of the first three parameters. The systems used today for the assessment of these parameters have no, or only limited potential for implantation. Accelerometers have this potential for implantation. We therefore started to investigate their use for the assessment of the angles of the different leg segments.
The use of accelerometers for the assessment of angular velocity and angular acceleration is well established [3,4,5,6]. However the calculation of the angle from angular velocity and angular acceleration is hampered by integration drift. We will show that, under certain conditions, the angle of the leg segments can be calculated from accelerometer data without integration.

THEORY

The orientation of the lower leg is to be determined. Therefore a body-fixed frame is placed on the leg with the origin in the point of rotation (i.e. the ankle joint). The orientation of the lower leg is now given by the orientation of this body-fixed frame with respect to an inertial reference frame. (See figure 1.)

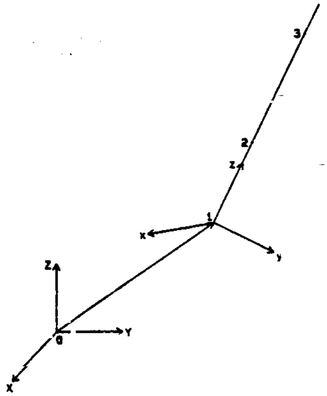

Figure 1. Points and reference systems for a simple lower leg model.
XYZ : Inertial reference system.
xyz : Body-fixed reference system.
0,1 : Origin of XYZ and xyz respectively.
2,3 : Points on the lower leg.

A seismic accelerometer with DC response is placed on the leg (point 2). The forces on the seismic mass are given by Newtons law:

$$\Sigma\vec{F} = \vec{F}_m + m\,\vec{g} = m\,\ddot{\vec{r}}_{02} \qquad (1)$$

With $\vec{F}_m$: reaction force vector on the seismic mass.
$\vec{g}$: gravitational acceleration vector.
$\vec{r}_{ij}$: vector from point i to j.
i = 0 : in inertial reference coordinates.
i = 1 : in body-fixed coordinates.
m : seismic mass.

Dividing the movement into a translation and a rotation we get

$$\ddot{\vec{r}}_{02} = \ddot{\vec{r}}_{01} + \ddot{\vec{r}}_{12} + 2\,\vec{\omega}\times\dot{\vec{r}}_{12} + \vec{\omega}\times(\vec{\omega}\times\vec{r}_{12}) + \dot{\vec{\omega}}\times\vec{r}_{12} \qquad (2)$$

Where $\vec{\omega}$ = rotation vector.

The signal from the accelerometer, called equivalent acceleration, is given by

$$\vec{a}_2 = -\vec{F}_m/m$$

$$= \vec{g} - \ddot{\vec{r}}_{01} - \ddot{\vec{r}}_{12} - 2\,\vec{\omega}\times\dot{\vec{r}}_{12} - \vec{\omega}\times(\vec{\omega}\times\vec{r}_{12}) - \dot{\vec{\omega}}\times\vec{r}_{12} \qquad (3)$$

For static situations we get

$$a_2 \text{ (static)} = \vec{g} \tag{4}$$

Considering (4) we talk about equivalent acceleration.In view of the rigid body assumption, $\dot{\vec{r}}_{12} = \ddot{\vec{r}}_{12} = \vec{0}$,

$$\vec{a}_2 = \vec{g} - \ddot{\vec{r}}_{01} - \vec{\omega}\times(\vec{\omega}\times\vec{r}_{12}) - \dot{\vec{\omega}}\times\vec{r}_{12} \tag{5}$$

With $\vec{r}_{12} = r_{12}\,\vec{e}_{12}$ and $|\vec{e}_{12}| = 1$ equation (5) becomes

$$\vec{a}_2 = \vec{g} - \ddot{\vec{r}}_{01} - r_{12}\,\vec{\omega}\times(\vec{\omega}\times\vec{e}_{12}) - r_{12}\,\dot{\vec{\omega}}\times\vec{e}_{12} \tag{6}$$

We devide the right part of equation (6) in two parts. The first, $\vec{g} - \ddot{\vec{r}}_{01}$, is independend of the distance r_{12}, whereas the second part isn't. Usually the first part is divided out by placing accelerometers at two points (2 and 3) on one line.
With $\vec{e}_{12} = \vec{e}_{13} = \vec{e}_1$ we find :

$$\vec{\omega}\times(\vec{\omega}\times\vec{e}_1) + \dot{\vec{\omega}}\times\vec{e}_1 = \frac{\vec{a}_3 - \vec{a}_2}{r_{12} - r_{13}} \tag{7}$$

The components of the equivalent acceleration can be measured in body-fixed coordinates. So by numerical integration, equation (7) could be solved for the components of ω in body-fixed coordinates. Integrating twice we would find the rotational angles themselves. This method is hampered by integration drift, and can only be used for short intervals.
We took an alternative approach by eliminating the rotational part of equation (6). This results in :

$$\vec{g} - \ddot{\vec{r}}_{01} = \frac{r_{12}\,\vec{a}_3 - r_{13}\,\vec{a}_2}{r_{12} - r_{13}} \tag{8}$$

Suppose $\ddot{\vec{r}}_{01}$ is known in inertial reference coordinates. Then the left part of (8) is known in inertial reference coordinates whereas the right part can only be measured in body-fixed coordinates. To solve (8) we translate the left part to body-fixed coordinates by multiplying it with a rotation matrix. This leads to three equations (one for each direction) which can be solved for the three rotational angles.
The big advantage of this method is that an integration is no longer necessarry. We also see that the relation is given between the equivalent acceleration and the direction of the gravitational acceleration. This means that the rotational angles are calculated relative to the gravitational field and not relative to the

ground.
Equation (8) can be solved only when $\ddot{\vec{r}}_{01}$ is known. This means that we need a priori knowlegde about the acceleration of the body fixed frames origin. Otherwise we have to use equation (7).
In our measurements on the lower leg we assumed the movement of the leg to be two-dimensional. During stance phase we assume the acceleration of the ankle joint to be zero. (e.g. $\ddot{\vec{r}}_{01}=\vec{0}$). Under these conditions equation (8) gives :

$$- g \sin(\theta) = \frac{r_{12}\, a_{3y} - r_{13}\, a_{2y}}{r_{12} - r_{13}} \qquad (9a)$$

$$- g \cos(\theta) = \frac{r_{12}\, a_{3z} - r_{13}\, a_{2z}}{r_{12} - r_{13}} \qquad (9b)$$

With a_{iy}, a_{iz} : the (body-fixed) y and z components of the equivalent acceleration in point i.

Not shown here is that equation (8) can be generalized to any two connected segments. We can calculate, without integration, the angle between these segments using accelerometers placed on them, independent of their movements. This means that the knee angle can be calculated both during stance and swing phase.

RESULTS.

Equation (9) was first tested on a 1-degree of freedom pendulum. Data of the accelerometers was amplified and low-pass filtered (20 Hz) by a second order Butterworth filter. A potentiometer was placed on the axle for reference. The signal of this potentiometer was only low-pass filtered. The signals were sampled at 100 Hz with a 12 bit ADC on a LSI 11-23 computer. Differences between the angle calculated from accelerometer data and the angle as measured with a potentiometer have standard deviations below 0.01 rad (maximum angle 0.7 rad) and maximum differences below 0.05 rad.

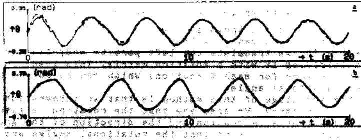

Figure 2. The angle of the lower leg as a function of time.
—— Calculated from accelerometer data.
· Calculated from potentiometer data.

For the measurements on the lower leg we placed the accelerometers on a bar which was then attached to the leg with VELCRO straps. A potentiometer was placed on the ankle joint and connected with the bar. Placing the foot firmly on the ground, the shank was rotated in the sagittal plane. Two typical results, after applying a least square correction (to compensate for offset and gain errors), are given in figure 2. The differences are quantisized in figure 3.

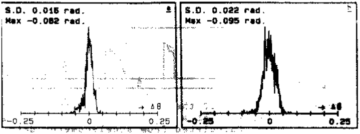

Figure 3. Difference histograms between the lower leg angle as calculated from accelerometer data and as calculated from potentiometer data.

Standard deviations are below 0.04 rad and maximum errors are below 0.1 rad.
As mentioned before equation (8) can only be used during stance phase. So we have to be able to distinquish between stance and swing phase. This can also be done with the accelerometers. From (8) we see that during stance when $\ddot{r}_{01} = \vec{0}$ the modulus of the right

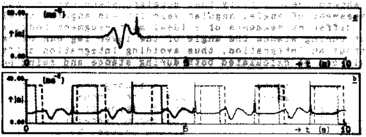

Figure 4. Equivalent acceleration in the ankle during walking as a function of time.
a. 1 step.
b. Walking, combined with footswitches.
(High is on the ground, low is of the ground.)
.........: Footswitch under the heel.
- - - - : Footswitch under the toes.

side of (8) should be equal to g ($g = 9.81\ m/s^2$). In figure 4 we see this modulus for 1 step and during walking. In figure 4b the equivalent acceleration signal is combined with the signal from footswitches under the heel and the head of the first metatarsal bone. We can see that the equivalent acceleration of the ankle closely correlates with the phases of walking. (Heel-off, toe-off, heel-contact, toe-contact.) So accelerometers can not only be used to calculate the angle of the lower leg but they can also show whether we are in swing or stance phase. For those phases during walking where the whole foot is on the ground we calculated the

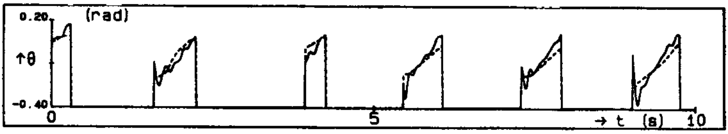

Figure 5. The angle of the lower leg during the stance phase of walking as a function of time. (See figure 4b.)
——— Calculated from accelerometer data.
- - - Calculated from potentiometer data.

angle of the lower leg using the accelerometer data and compared it with the angle according to the potentiometer (Figure 5). Standard deviations are comparable to those obtained during the measurements with the foot on the ground.

DISCUSSION.

Most sensors available for the assessment of the feedback parameters (angle, angular velocity and angular acceleration of the lower extremities) have severe limitations for their use with FES. Therefore we started research into the use of accelerometers. We showed theoretically that accelerometers can be used for the assessment of angle, angular velocity and angular acceleration of the different segments of a ideal multi-segment body. We found that during stance the angle of the lower leg can be calculated without an integration, thus avoiding integration drift. The knee angle can be calculated both during stance and swing phase without integration.

First measurements show that this theory can be used favourably for the assessment of the angle of the (non-ideal) lower leg. After our initial measurements (rotation of the shank during stance) [7] we now extended the use of accelerometers to the detection of the different walking phases. We also used the accelerometers for the angle assessment of the lower leg during the stance phase of normal walking.

Experiments during the swing phase must show whether accelerometers can be used to calculate the angle of the lower leg during all phases of normal walking.

For the stance phase preliminary data [8] indicates that the accuracy obtained with accelerometers is high enough for the development of a feedback controller.

Because of their small size, mass and low energy consumption it should be possible to develop a fully autonomous and implantable system for the assessment of the feedback parameters using accelerometers.

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