

## Artificial sensory systems

### INTRODUCTION

Sensing is important in human motor control. It enables the control of movement tasks under variable external circumstances and supplies a person with information about the surrounding.

Certain impairments of the neuromuscular system may result in reduced sensing, which may impede motor control performance. In such cases there is a need for supplementing the reduced sensation. Furthermore, sensing is important for artificial control in assistive human motor systems (figure 9-1).

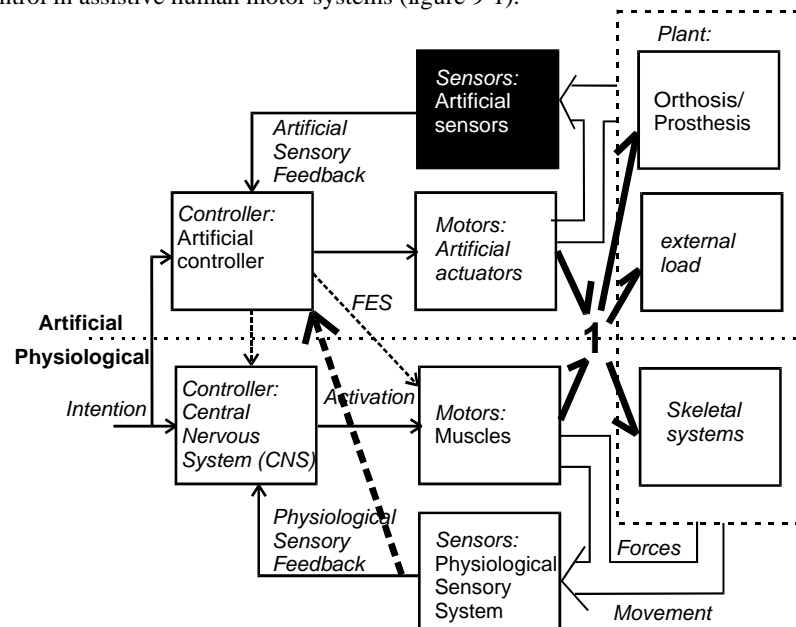


Figure 9-1 Schematic block diagram of an assistive system that supports the impaired neuromuscular systems. The sensing subsystem of assistive systems is the subject of this chapter.

### OBJECTIVES

This chapter will

- show the importance of sensing in assistive motor control systems.
- give an overview of sensing principles for signals from the CNS, force and movement
- describe how signals can be derived from the physiological sensory system and using artificial sensors
- describe how these sensory signals can be used in assistive systems
- describe several artificial sensory systems

## CONTENTS

## 9.1 Introduction

Many artificial motor support systems have no explicit control system, e.g. mechanical prostheses and orthoses. These systems are designed such that their inherent mechanical characteristics are optimized for support of a certain motor function, like standing or walking. Actually, these inherent mechanical characteristics can be physically described as a control strategy built into the mechanics of the system. This can make the system very simple and robust, but may not be very flexible if the motor tasks to be performed are variable in nature.

If the motor tasks are more variable, a more versatile interaction with the physiological control system and an explicit control system should be part of the prosthetic or orthotic system. Such a control system can only support the motor tasks of the user in an optimal way if information is available about the activities and intentions of the user and of the state of the artificial support system. This requires that sensory information is derived from the user and the support system.

There can be several reasons for deriving such sensory information (figure 9-2):

1. The sensory information may serve to detect the *intention of the user* with respect to the motor task to be performed and the way it should be performed. It can be supplied as control signals to the artificial controller (indicated with (1) in figure 9-2).
2. The information may serve *to assess the physical state of the body and characterize the interaction with the environment*. This information can be used:
  - as a feedback to the user to supply sensory information which he is deprived of because of his impairment (indicated with (2) in figure 9-2)
  - as a feedback to the artificial control system, to be used in controlling the artificial support system (indicated with (3) in figure 9-2)

Similar information streams can also be distinguished in the physiological motor control system.

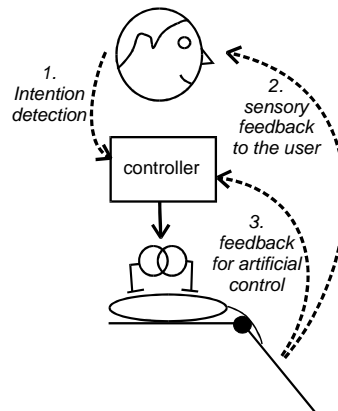


Figure 9-2 Sensory information may be derived from the body of the user in order to derive the motor intention of the user (1), to provide the user with lost sensory information (2) or in order to provide feedback to an artificial controller (3).

In this chapter, SECTION 9.2 presents the possibilities to derive sensory information from the human body (section 9.2). Signals may be derived from the physiological sensors of the body or from artificial sensors mounted on or in the body. SECTION 9.3

describes in what way this sensory information can be supplied to the user to restore lost sensory information.

SECTION 9.4 describes how the motor intention of the user can be derived. Control signals for the assistive system can be derived from these intentions. Signals may be directly derived from the Central Nervous System or by measuring movements and forces voluntarily induced by the activities of the user.

## 9.2 Deriving sensory signals

Sensory information related to motor control can be restored in several ways:

- Signals can be derived from the physiological sensors of the body if they are still present and functioning. This will be discussed in section 9.2.1.
- Alternatively, artificial sensors can be used that are mounted on external parts of the motor assistive system or even implanted (section 9.2.2).

Table 1 summarizes modalities for sensing relevant physical quantities related to human motor control. Signals may be derived from several parts of the motor control chain (figure 9-3): sensors may measure the activation of the muscles (EMG: ElectroMyoGraphy), the force or joint moments generated by the muscles or the resulting movements.

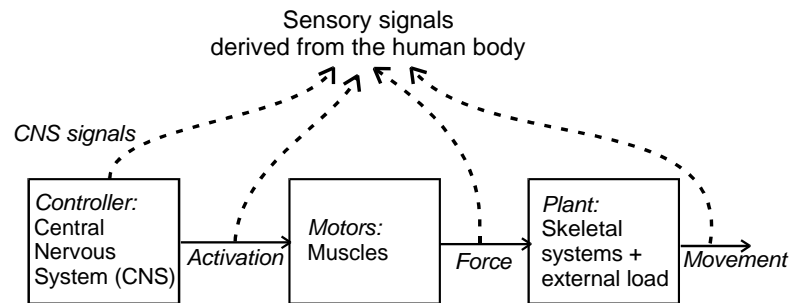


Figure 9-3 Sensory signals may be measured from several parts of the motor control chain, including muscle activation, forces and movements.

TABLE 1 MODALITIES TO DERIVE SENSORY INFORMATION FROM THE HUMAN MOTOR SYSTEM (CRAGO ET AL. 1986; VELTINK 1999).

<i>Sensor</i>	Physical quantity	Transduction principles	Interaction with the body
<b>CNS signals / muscle activation</b>			
<i>EEG / MEG</i>	<i>Brain signals</i>	<i>Electric / Magnetic</i>	<i>Electrodes / coils</i>
<i>ENG / MNG</i>	<i>Signals from nerves</i>	<i>Electric / Magnetic</i>	<i>Electrodes / coils</i>
<i>EMG / MMG</i>	<i>Muscle activation</i>	<i>Electric / Magnetic</i>	<i>Electrodes / coils</i>
<b>Force</b>			
<i>Strain gauges</i>	<i>Strain of material → measure for Force / moment</i>	<i>Resistive/ piezoresistive</i>	<i>Attachment on prosthetic / orthotic elements</i>
<i>Pressure sensors</i>	<i>Pressure</i>	<i>- Force Sensitive resistors (FSR) - Force Sensitive Capacity</i>	<i>Attach between segment and environment on contact surface</i>
<i>Physiological Skin sensors</i>	<i>Skin stress (phasic rather than static components)</i>	<i>Physiological sensors</i>	<i>Derive via electrodes around sensory nerves (Hoffer et al. 1996; Sinkjaer 1999)</i>
<b>Movement</b>			
<i>Goniometer</i>	<i>Joint angle (1, 2 or 3D)</i>	<i>- Resistive - Hall effect + magnet (Johnson et al. 1999)</i>	<i>Attach to two segments connected by joint</i>
<i>Orientation sensor</i>	<i>Orientation with respect to earth coordinate system</i>	<i>- Hall effect (earth magnetic field) - inertial: gravity acceleration</i>	<i>Attach sensor to body segment</i>
<i>Rate of turn Gyroscope</i>	<i>Angular velocity (1 – 3D)</i>	<i>- inertial, coriolis / piezoresistive (Soderkvist 1994)</i>	<i>Attach to body segment (Baten et al. 1996)</i>
<i>Accelerometer</i>	<i>Acceleration (inertial + gravitational: 1 – 3D)</i>	<i>Inertial / piezoresistive, piezoelectric or capacitive [Lotters, 1995 #215]</i>	<i>Attach to body segment</i>
<i>Position sensor</i>	<i>- relative distance to objects in environment - absolute position in space</i>	<i>- radar using ultrasonic transmitters/sensors (Piezo electric) - GPS: Global Positioning System using satellite communication</i>	<i>-ultrasonic transmitters / sensors mounted on body (Shoval et al. 1998) - GPS system mounted on body</i>

Remarks:

- Skin sensors are the only physiological sensors included in this paper, because they are the only physiological sensors that have been used in human prosthetic systems at the moment (Sinkjaer 1999). The use of other physiological sensors are expected in future (Hoffer et al. 1996).
- 3D gyroscopes and accelerometers can be combined in one inertial sensory system, which gives comparable sensory information as the human vestibular system.
- In many cases, the measurement of the body position in space (e.g. ultrasonic radar, GPS) is not relevant for assistive human motor control systems, because the user is able to locate himself. This would only be required in special situations, e.g. if the user is blind (Shoval et al. 1998)

## 9.2.1 SIGNALS FROM THE PHYSIOLOGICAL SENSORY SYSTEM

In people with disorders of the Central Nervous System (CNS) (e.g. stroke or spinal cord injury), the peripheral motor system is intact in most cases. However, the information exchange between CNS and peripheral system is not present or disturbed. The muscles, being the actuators, may not receive the command to contract or receive disturbed inputs (e.g. in the case of spasticity). Comparably, the information from the peripheral sensory system (chapter 2) may not arrive by the CNS.

Sensory information may be derived from the physiological sensors of the body. This information can be used for feedback in motor control support systems or for restoring sensory feedback to the user (figure 9-2).

In order to derive sensory signals from a sensory nerve, a tripolar electrode configuration in an insulating cuff is placed around the sensory nerve (figure 9-4) (Haugland et al. 1994; Struijk 1997; Struijk et al. 1999). The tripolar configuration inside an insulating cuff ensures rejection of disturbing signals like EMG from muscles or stimulus artifacts when muscles are stimulated simultaneously. Alternatively, sensory signals can be derived with electrodes inside the nerve fiber bundles which compose a nerve, the so-called nerve fascicles (Yoshida et al. 1996). Other electrode configurations that have been proposed are a sieve electrode through which nerve fibers can grow (Kovacs et al. 1994; Stieglitz et al. 1997) and an array of electrodes which pierce the nerve (Rutten et al. 1999). It should be noted that only the cuff electrode has been used in humans (Haugland et al. 1999). The other electrode configurations have only been tried experimentally in animals.

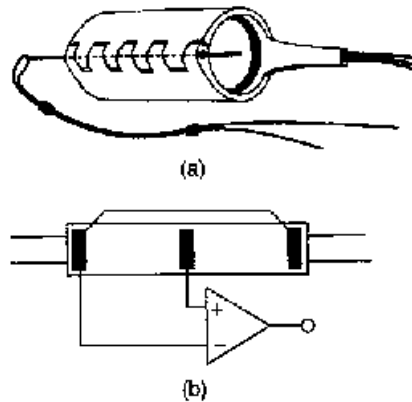


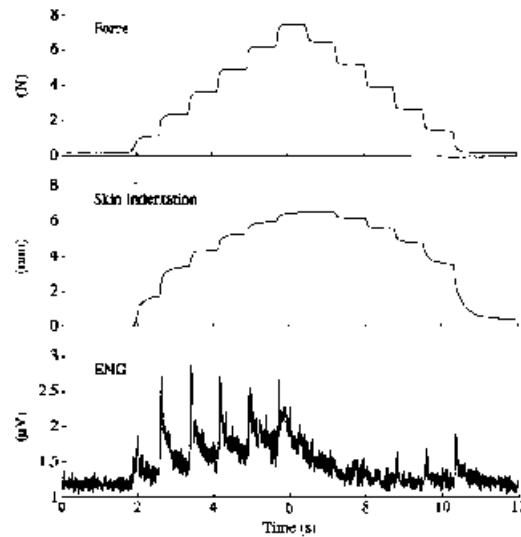
Figure 9-4 (a) Cuff electrode for measuring signals from a sensory nerve; (b) The neural signals are measured using a tripolar electrode configuration, which rejects disturbing external signals (from (Struijk et al. 1999))

Haugland et al. (Haugland et al. 1994) and other authors have shown that sensory signals generated by skin sensors can be derived using cuff electrodes around sensory nerves. These sensors appeared to have a predominantly phasic dynamical characteristic, meaning that they are mainly sensitive for changes in skin pressure. (see example 9-1). Skin sensory signals have been derived in human subjects from the skin of the foot and of the hand and have been applied as feedback in neural prostheses for upper and lower extremities fingers (Haugland et al. 1999) (see chapter 10)

More recently, Riso et al. (Riso et al. 2000) described sensory signals from mixed muscle nerves in rabbits, reporting a clear relation of these signals with joint ankle.

## EXAMPLE 9-1

**SKIN CONTACT FORCE INFORMATION IN SENSORY NERVE SIGNALS RECORDED BY IMPLANTED CUFF ELECTRODES (HAUGLAND ET AL. 1994).** Haugland et al. investigated the relation between force applied on a skin area and the electroneurographic signal (ENG) derived from the tibial nerve of a cat, which is a sensory nerve. An example of the ENG response to a staircase skin force profile is shown in figure 9-5. The ENG signal was band pass filtered (1 KHz – 10 KHz), rectified and bin integrated (taking average over subsequent small intervals). Figure 9-5 clearly illustrates that the skin sensors are very sensitive for changes in skin pressure, but also sense low frequency components. Also, pressure increase elicits larger responses than pressure decrease.



*Figure 9-5 Example response of ENG to a staircase pressure profile applied to the skin (from (Haugland et al. 1994)). Please note the phasic character of the neural responses, which is different for increasing than for decreasing skin pressure.*

Haugland et al. derived a mathematical model of the relation between the applied contact force and the ENG (figure 9-6). This model gave a good prediction of ENG depending on applied skin pressure. However, the model contains several nonlinear components which can not be inverted (e.g. rectification and saturation). For this reason, skin pressure can not be estimated from the ENG signals. Additional assumptions need to be made for such an estimate.

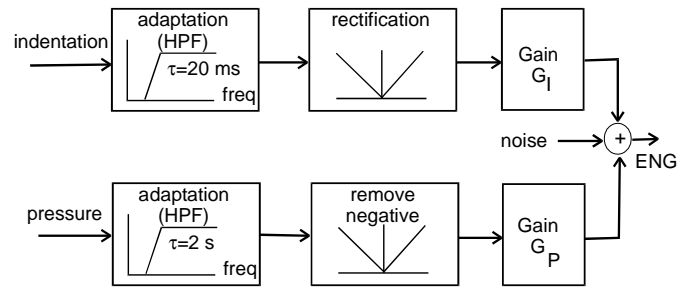


Figure 9-6 Model relating skin indentation and skin pressure to ENG. Note that this model can not be inverted (Haugland et al. 1994).

#### PROBLEM 9-1

CONSIDER THE MODEL RELATING SKIN INDENTATION AND SKIN PRESSURE TO ENG (FIGURE 9-6)

- Reconstruct the predicted ENG signal for the skin pressure and skin indentation signals depicted in figure 9-5. How well is the measured ENG signal predicted?
- Skin indentation and skin pressure are related by the mechanical characteristics of the skin. Give a sketch of this relation on the basis of the signals depicted in figure 9-5.
- Why is it not possible to estimate skin pressure and skin indentation from measured ENG on the basis of the model depicted in figure 9-6?

#### 9.2.2 ARTIFICIAL SENSORY SYSTEMS

Artificial sensors that measure movement or force provide an alternative for deriving signals from physiological sensors. Movement support systems should be useable in any place. Therefore, sensors that provide feedback in these systems should preferably not depend on provisions in the environment, but be mounted on or implanted in the body. In order to observe the state of the human multi-segmental body system and provide sensory feedback, movements of body segments and interaction forces with the environment need to be measured. The following examples describe several sensors for measuring body movements.

#### EXAMPLE 9-2

##### IMPLANTABLE JOINT ANGLE SENSOR BASED ON THE HALL EFFECT

A Hall sensor measures a component of a magnetic field: an electrical current is induced in a planar strip of semiconductor material. If the magnetic field has a component perpendicular to the direction of the electrical current and the plane of the semiconductor material a Lorentz force acts on the charge carriers in the material, inducing an electrical field perpendicular to the current, which can be measured as a potential difference.

If a magnet is positioned in the bone at one side of a joint and the Hall sensor in the other bone, a signal can be measured which is related to joint angle, and, thus, the joint angle can be measured (figure 9-7).

Other joint angle sensors (e.g. potentiometric sensors) are also available, but have only been developed for external use.

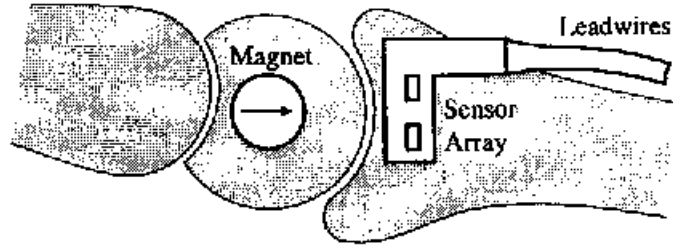


Figure 9-7 Hall effect based joint angle sensor (from (Johnson et al. 1999)). The magnet is placed in the bone at one side of the joint, while the Hall sensor is placed in the bone at the other side.

### EXAMPLE 9-3

#### THREE DIMENSIONAL ACCELEROMETER [Lotters, 1998 #223]

An 3D accelerometer measures the inertial forces acting on a mass. When dividing the resulting force by the mass on which it acts, the sum of acceleration  $\vec{a}$  (2<sup>nd</sup> derivative of position) and gravitational acceleration  $\vec{g}$  (figure 9-8) results:

$$\vec{s}_a = \vec{a} - \vec{g} \quad (9-1)$$

The gravitational component provides information about the inclination with respect to the vertical.

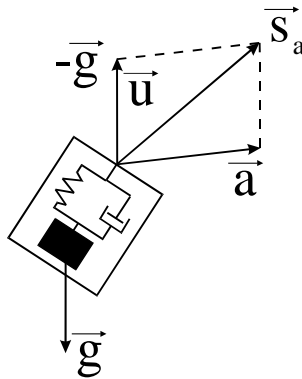


Figure 9-8 An accelerometer measures the sum of the inertial acceleration  $\vec{a}$  and the gravitational acceleration  $\vec{g}$  acting on on a mass.

The 3 components of acceleration can be measured using three uniaxial accelerometers. Alternatively, they can be measured with an inherent 3D accelerometer consisting of a mass in a box, suspended by springs (figure 9-9) [Lotters, 1998 #223]



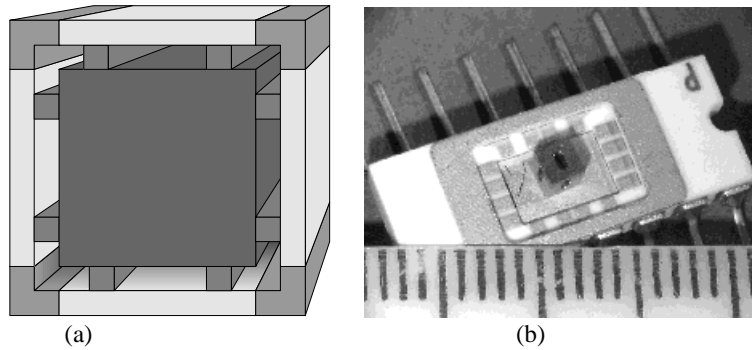


Figure 9-9 3-axial accelerometer (from [Lotters, 1998 #223]) consisting of a mass in a box, suspended by springs. The distances between mass and box are measured capacitively at all sides, yielding the inertial forces acting on the mass. (a) schematic drawing, (b) photograph of triaxial accelerometer (dimensions  $2*2*2 \text{ mm}^3$ ).

EXAMPLE 9-4

IN-USE CALIBRATION OF 3D ACCELEROMETER [LOTTERS, 1998 #201]

A 3D accelerometer measures inertial acceleration as well as gravitational acceleration (equation 9-1). The gravitational component provides a means of calibrating the accelerometer for  $-g$ ,  $0 \text{ m/s}^2$  and  $+g$ .

In practical use of accelerometers for human movement analysis quasistatic periods with no or little inertial acceleration occur regularly. The measured acceleration is only or mainly the gravitational acceleration. These quasistatic periods can be detected and used in the calibration of the sensor during use.

PROBLEM 9-2

IN-USE CALIBRATION OF 3D ACCELEROMETER

Explain how a 3D accelerometer can be calibrated during use if quasistatic intervals occur regularly for several orientations of the sensor [Lotters, 1998 #201].

EXAMPLE 9-5

DETECTION OF POSTURES AND MOVEMENTS USING ACCELEROMETERS (VELTINK ET AL. 1996A)

If human mobility is supported using an assistive device, it can be important to detect what postures and movements the user performs. This information can be used to detect intention and supply feedback for the control of the assistive system (see section 9.1).

Detection of posture or movement and, in case of posture, distinction of the posture can be done using a few uni-axial accelerometers on segments of the body (e.g. trunk and thigh) (Veltink et al. 1996a). By detecting whether the signals vary with time it can be assessed whether the person moves or is in a posture (figure 9-10). If a posture is attained, the posture can be distinguished by combining the signals of the accelerometers on several body segments, which now act as an inclinometer (under static conditions, the accelerometer signal only consists of the gravity acceleration) (figure 9-11).

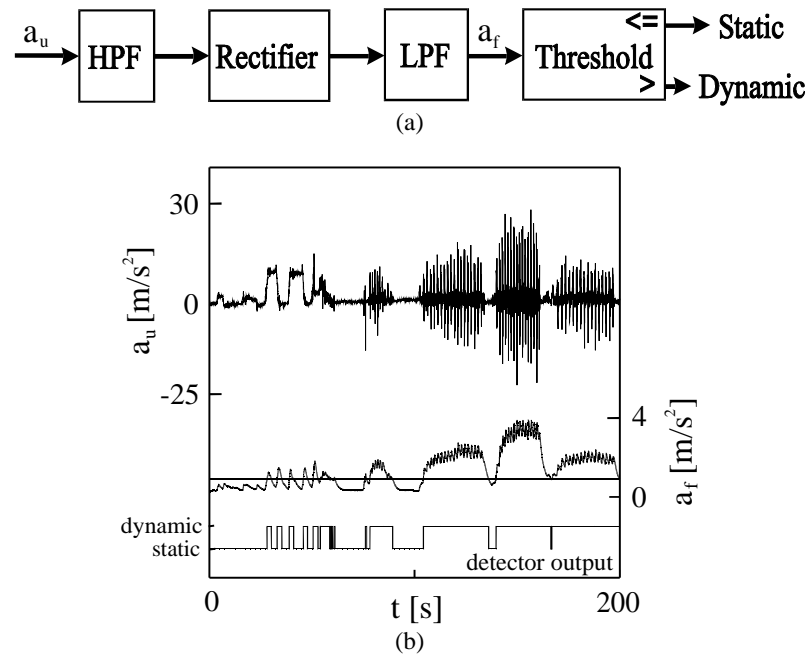


Figure 9-10 (a) detector for postures ('static') and movement ('dynamic'), which can be applied to a signal measured with a uni-axial accelerometer on a body segment; HPF: high pass filter, LPF: low pass filter; (b) example of detection of posture and movement periods from an accelerometer signal on the upper leg (Veltink et al. 1996a). The upper signal is the sensor signal, the signal in the middle is the output of the movement detector displayed in figure (a). After providing a detection threshold, the output of the posture-movement detector is obtained.

### PROBLEM 9-3

#### POSTURE-MOVEMENT DETECTOR

Explain the purpose of the high pass filter (HPF), rectifier, low pass filter and threshold operations in the posture-movement detector of figure 9-10.

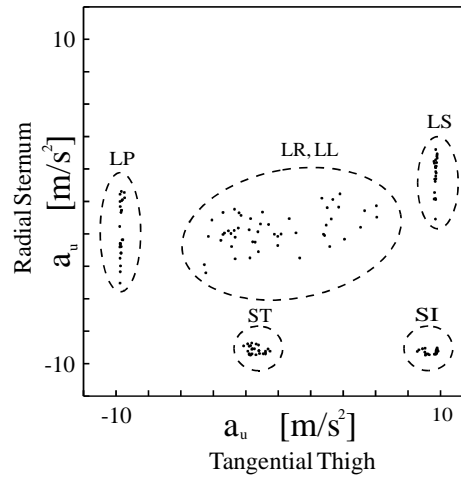


Figure 9-11 Under quasi-static circumstances (no or little body movement) the body posture can be assessed by combining the inclinometer reading of accelerometers on several segments of the body (Veltink et al. 1996a). In this example a tangentially placed accelerometer on the thigh and a radially placed accelerometer on the trunk are combined (ST: standing; SI: sitting; LS: lying supine; SP: lying prone; LR: lying on the right side; LL: lying on the left side).

## PROBLEM 9-4

## DISTINGUISHING POSTURES

What accelerometer should be added to be able to distinguish lying on the left and right side, which could not be distinguished in the example presented in figure 9-11 (LR, LL)?

## EXAMPLE 9-6

## OPERATION PRINCIPLE OF RATE GYROSCOPE

If a mass  $m$  is rotated with angular velocity  $\vec{\omega}$  while it has a velocity  $\vec{v}$ , a Coriolis force  $\vec{F}_C$  is acting on the mass (figure 9-12):

$$\vec{F}_C = 2m\vec{\omega} \times \vec{v} \quad (9-4)$$

This phenomenon is used to measure angular velocity with rate gyroscopes: the mass is vibrated in one direction. A vibration in a perpendicular direction will also occur if the mass is rotated around an axis that has a component perpendicular to the direction of vibration.

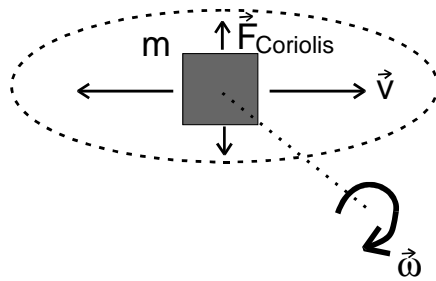


Figure 9-12 On a mass  $m$  with velocity  $\vec{v}$  a Coriolis force  $\vec{F}_C$  is acting when it is turned with an angular velocity  $\vec{\omega}$  having a component perpendicular to  $\vec{v}$ . When the mass is vibrated in one direction a vibration in the perpendicular direction will also occur if the mass is rotated.

## PROBLEM 9-5

## 3D INERTIAL SENSOR

- Explain that the triaxial accelerometer depicted in figure 9-9 can also be used as a rate gyroscope if the mass is vibrated.
- In how many directions should the mass be vibrated in order to be able to measure all three components of angular velocity?

## EXAMPLE 9-7

## OBTAINING ORIENTATION FROM ANGULAR VELOCITY

A 3D rate gyroscope provides 3D angular velocity information  $\vec{\omega}$ . In many applications it is important to estimate orientation  $\vec{\theta}$  of a body segment. This can be obtained by integrating the angular velocity. However, this can not be done by straightforward integration but by solving the following differential equation (Bortz 1971):

$$\dot{\vec{\theta}} = \vec{\omega} + \frac{1}{2}\vec{\theta} \times \vec{\omega} + \left(1 - \frac{\theta \sin \theta}{2(1 - \cos \theta)}\right) \vec{\theta} \times (\vec{\theta} \times \vec{\omega}) \quad (9-5)$$

## PROBLEM 9-6

## 3D ORIENTATION ESTIMATION FROM 3D ANGULAR VELOCITY MEASUREMENTS

- Why can 3D orientation not be obtained by straight forward integration of each component of 3D angular velocity as measured by the rate gyroscope?
- From 3D angular velocity measurements 3D orientation change can be measured. What initial information is required in order to estimate absolute orientation?

## PROBLEM 9-7

## 3D ACCELEROMETER / RATE GYROSCOPE COMPARED TO THE HUMAN VESTIBULAR SYSTEM

Show that a 3D inertial sensor system, consisting of 3D accelerometer and rate gyroscope gives comparable information as the human vestibular system (see chapter 2).

## PROBLEM 9-8

## ORIENTATION INFORMATION PROVIDED BY 3D ACCELEROMETER AND 3D GYROSCOPE

- Explain that both 3D accelerometer and 3D rate gyroscope provide orientation information.
- The 3D accelerometer does not provide any information about one component of orientation. Which component?

Both 3D accelerometer (example 9-3) and 3D rate gyroscopes provide orientation information. These can be fused using a Kalman filter and a model of both systems (Luinge et al. 1999).

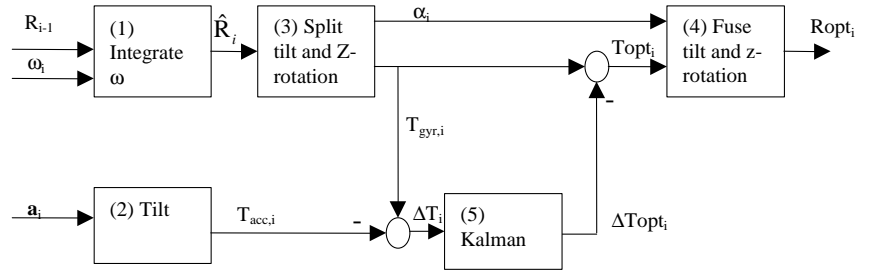


Figure 9-13 Optimal estimation of orientation ( $R_{opt}$ ) by fusing the information of a 3-axial accelerometer ( $\mathbf{a}$ ) and a 3-axial rate gyroscope ( $\omega$ ).  $T$  is tilt,  $\alpha$  is orientation around vertical (Luinge et al. 1999).

PROBLEM 9-9

OPTIMAL ESTIMATION OF ORIENTATION (FIGURE 9-13)

- What orientation information can be derived from the 3D accelerometer?
- Under what condition is this information reliable?
- What orientation information can be derived from the 3D rate gyroscope?
- What problem occurs when estimating orientation for a long period of time by integrating angular velocity?
- Describe how the information from the 3D accelerometer can be optimally “fused” with the information from the 3D rate gyroscope for estimating orientation.

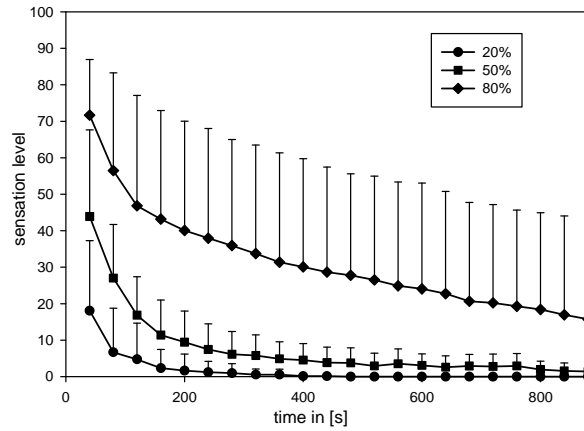
9.3 Sensory feedback to the user

As depicted in figure 9-2 sensory information may be derived for feedback in artificial motor control or to supplement the reduced sensory information of the user of the motor assist system. This restoration of sensory information to the user is essential for giving him the opportunity to control motor tasks.

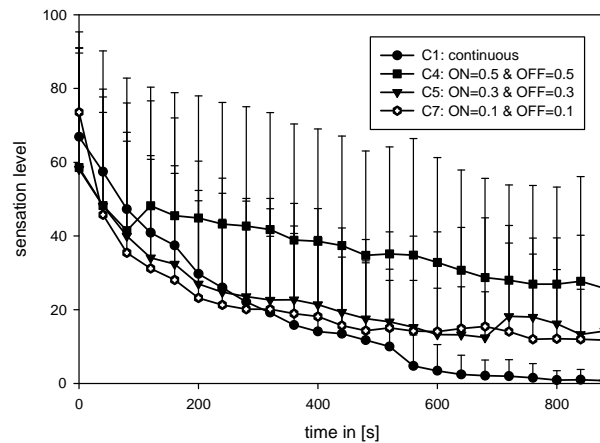
Several alternatives exist for giving sensory feedback to the user of a prosthetic or orthotic system:

- The position and force sensations may be transferred from a body part or prosthesis part without sensation to a healthy body part, mapping force on force and position on position (Extended Physiological Feedback: EPP). This method was first proposed by Simpson (Simpson 1974; Doubler et al. 1984) and has the advantage that the transferred sensation is very natural.
- Using other sensory channels to restore lost sensory information (movements / forces). This can be done by visual or auditory feedback (Smith 1990), or by tactile feedback (both mechanically or by electrical stimulation of the skin) (Szeto et al. 1990; Kaczmarek et al. 1995). When stimulating the skin sensors, account should be taken of the dynamic characteristics of these sensors. The sensation adapts within minutes for continuously supplied signals and it takes several 100 ms before a steady level of sensation is obtained when a stimulus is applied (Figure 9-14). This should be taken into account when optimizing an electrocutaneous interface for feedback.
- A third alternative is to supply the sensory information directly to the nerves that carry sensory nerve fibers by stimulation of these nerve fibers. Preferably, the

provided information should be adequately mapped on the stimulated sensory nerve fibers, in accordance with the information previously transmitted by these nerve fibers (Riso 1999). Direct sensory information transfer to the sensory nerves has not been done in humans yet.



a.



b.

*Figure 9-14 (a) Sensation adaptation for continuous electrocutaneous stimulation at several stimulation levels (20, 50 and 80% of range between sensation and pain threshold). The sensation is reduced within minutes after it is applied. The speed of adaptation depends on stimulation level. (b) Intermittent stimulation at 50% duty cycle and varying cycle time. Intermittent stimulation reduces adaptation, but stimulation bursts should be long enough (several 100 ms) to reach a steady sensation level. For this reason the results are different for the three cycle times, although the number of stimuli per time unit are the same. These are experimental results obtained by ir. D. Buma*

## PROBLEM 9-10

## SUSTAINED ELECTRO CUTANEOUS STIMULATION

- a. What is the relation between the dynamics of skin sensory responses depicted in figure 9-5 and the sensation characteristics of continuous and intermittent electrocutaneous stimulation shown in figure 9-14? Are they in agreement?
- b. Can you explain why short stimulus bursts (0.1 s) could result in a lower sensation level than longer stimulus bursts (e.g. 0.5 s).

#### 9.4 Deriving the motor control intention of the user control signals from the Central Nervous System

The artificial motor control systems should support the motor intention of the user in an optimal way. Therefore, control signals may have to be derived from the body of the user, giving information about the intention of the user (*SECTION 9.1, figure 9-2*). Signals may be directly derived from the Central Nervous System (*SECTION 9.4.1*) or by measuring movements and forces voluntarily induced by the activities of the user (*SECTION 9.4.2*).

##### 9.4.1 DERIVING CONTROL SIGNALS FROM THE CENTRAL NERVOUS SYSTEM

Except for sensing movements and forces, also control signals may be derived from several levels of the Central Nervous System (CNS) by measuring EMG (ElectroMyography), ENG (ElectroNeuroGraphy) or EEG (ElectroEncephalography). These signals can be used to control electrical stimulation of paralyzed muscles (Graupe et al. 1998; Thorsen et al. 1999) or artificial actuators in prostheses (Parker et al. 1986).

##### 9.4.1.1 *Electro and Magnetic EncephaloGraphy (EEG and MEG)*

It has been shown that movement intentions can be detected from multi-electrode EEG measurement (27 electrodes at distances of 2.5 cm) above the right and left sensorimotor cortex areas (Pfurtscheller 1999). Upon a visual cue on a screen, healthy subjects were asked to imagine a one-sided hand movement. By applying suitable temporal and spatial filters, the movement intention could be detected. It should be noted that, until now, the demonstrated information capacity of this channel is small (in order of 1 bit/s).

##### 9.4.1.2 *Electro- and Magnetic MyoGraphy (EMG and MMG)*

Muscles are actuated by electro-chemical signals which are transported by the  $\alpha$ -motor neurons. These signals induce currents in the tissue of the body, which, in turn, induce an electrical potential field in the tissue. As a result of this field, potential difference signals can be measured from electrodes in the tissue or on the skin. These signals related to the activation of muscles are called ElectroMyography (EMG).

As the muscle consists of many asynchronously activated motor units, each of which consist in turn of groups of muscle fibers, the EMG consists of a weighted sum across both space and time of a large number of individual events (Hogan et al. 1980a; Hogan et al. 1980b; Graupe 1989). Therefore, EMG can be described as a random process having a Gaussian distribution, a zero mean and a variance  $\sigma^2$ , being a memoryless function of activation level (figure 9-12). Therefore, the power spectral density of the

EMG signal  $S_M$  can be expressed as a zero-Mean, Gaussian white noise process passed through a linear constant-coefficient filter  $H$ , which is subsequently multiplied by a static function  $\sigma$  giving the overall activation state of the muscle:

$$S_M(f) = Q \cdot |H(f)|^2 \cdot \sigma(A)^2 \quad (9-7)$$

where  $Q$  is a constant (Hogan et al. 1980a).

Under isometric circumstances, the activation level  $A$  may be a function of muscle force and therefore, the variance of the EMG signal may also be a nonlinear function muscle force.

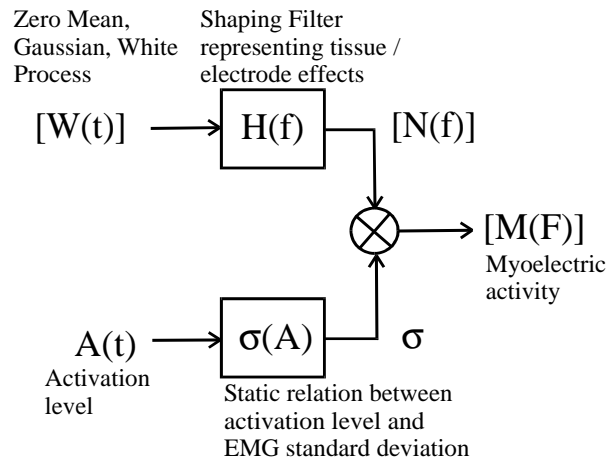


Figure 9-12 Functional model of EMG, being a band limited, zero-mean Gauss Markov process  $[N(t)]$ , amplitude modulated by a static function of overall activation level of the muscle (Hogan et al. 1980a).

On the basis of this random process model of EMG, Hogan and Mann (Hogan et al. 1980a) derived the maximum likelihood estimate of activation level (figure 9-13). It consists of a prewhitening filter, followed by a variance estimator. Subsequently, the square root operation delivers the optimal estimate of the standard deviation of the EMG signal. The prewhitening filter removes the correlation between successive samples of the myoelectric activity introduced by the shaping filter  $H(f)$ . Hogan and Mann propose that the required prewhitening may also be implemented by reducing the spacing between the EMG electrodes, thus increasing the bandwidth of the EMG, modifying the spatial filtering involved in measuring EMG. It should be noted however, that reducing the spacing between the electrodes may result in a smaller part of the muscle actually contributing to the measured signal.



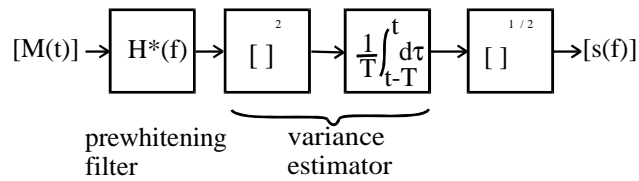


Figure 9-13 Schematic representation of the maximum likelihood estimator of the standard deviation of the EMG signal (Hogan et al. 1980a).

PROBLEM 9-11

EMG SIGNAL GENERATION

The filter H represents the filtering influence of the tissue volume conductor and the geometric relation between the source of the EMG signal, being the action potentials which run along the muscle fibers initiating the contraction of these fibers, and the position of the electrodes where the signals is registered (figure 9-14).

Questions:

- Explain how the action potential, being a spatially distributed current source at the muscle fiber membrane which travels along the muscle fiber, can give a time dependent potential difference between the EMG electrodes.
- Explain how the tissue volume conductor can influence the frequency content of the EMG signal measured at the electrodes
- Explain how the distance between the EMG electrodes in the case of bipolar measurement can influence the frequency content of the EMG signal initiate the muscle fiber contractions running along the muscle fibers

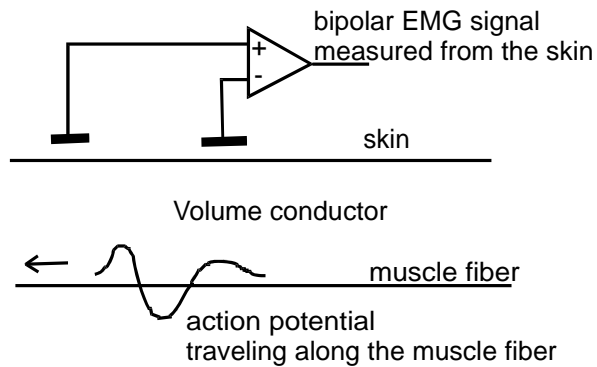


Figure 9-14 Schematic representation of the EMG measuring process, in which an spatially distributed source signal traveling along the muscle fiber generates a time varying potential difference between the EMG electrodes.

EMG has been used as a control source in several applications:

- EMG controlled hand and elbow prostheses (Parker et al. 1986; Abul-Haj et al. 1990a; Abul-Haj et al. 1990b)
- Graupe (Graupe et al. 1988; Graupe 1989) used above lesion EMG of paraplegic patients to control FES supported mobility. In fact, he detected the intention of the user to perform a certain mobility task (e.g. standing up, making a step) by constantly identifying the parameters of autoregressive dynamic model of EMG. Each intention was associated with a certain subspace of the space spanned by these parameters.
- Thorsen (Thorsen 1999; Thorsen et al. 1999) developed a muscle activation booster for partly paralyzed muscles. The muscles were stimulated at an intensity which depended on the remaining voluntary EMG as measured in between the stimulation pulses.

#### 9.4.2 DERIVING MOTOR CONTROL INTENTION FROM BODY MOVEMENTS AND FORCES

Users of motor support systems have mostly a remnant voluntary control of their motor function. When support systems are required, this voluntary control is insufficient to perform required motor tasks without help. However, the voluntary activity may still contribute to the task to a large degree. Except for contributing to the execution of motor tasks, the remaining voluntary activity may be useful for detecting the motor intentions of the user. These may be detectable from the activation of muscles, forces applied on the environment and resulting body movements. Graupe (Graupe et al. 1988; Graupe 1989) showed that EMG of voluntary controlled above lesion muscles can be used to detect motor task intention in mobility of paraplegics. Andrews et al. (Andrews et al. 1989; Andrews 1995) and Veltink et al. (Veltink et al. 1995; Veltink et al. 1996b) showed that mobility intention of paraplegics can be derived by a combination of force and movement sensors suitably placed on the body. This intention detection schemes are mostly combined with a finite state machine description of relevant motor tasks. State changes are detected on the basis of the measured sensory signals. Kirkwood et al. (Kirkwood et al. 1989) demonstrated that an optimal set of sensors for detection of defined state changes can be determined by machine learning methods. Detection of state transfers can also be performed using Neural Network Methods (Heller et al. 1993; Popovic et al. 1993; Kostov et al. 1995).

#### PROBLEM 9-12

WHICH SIGNAL CAN BE CONSIDERED TO REPRESENT THE INTENTION OF A PERSON BEST:

- a. a signal measured from the nervous system
- b. interaction forces of body segments with the environment
- c. signals from sensors which measure the movements of body segments

Explain your choice.

## REFERENCES

- Abul-Haj CJ and Hogan N (1990a): Functional assessment of control systems for cybernetic elbow prostheses - Part I: description of the technique. *IEEE Transactions on Biomedical Engineering* 37: 1025-1036.
- Abul-Haj CJ and Hogan N (1990b): Functional assessment of control systems for cybernetic elbow prostheses - Part II: application of the technique. *IEEE Transactions on Biomedical Engineering* 37: 1025-1036.
- Andrews BJ (1995). *On the use of sensors for FES control*. 5th Vienna International Workshop on Functional Electrostimulation, Vienna, pp. 263-266.
- Andrews BJ, Barnett RW, Phillips GF and Kirkwood CA (1989): Rule-based control of a hybrid FES orthosis for assisting paraplegic locomotion. *Automedica* 11: 175-199.
- Baten CTM, Oosterhoff P, Kingma I, Veltink PH and Hermens HJ (1996). *Inertial sensing in ambulatory load estimation*. 18th annual International Conference of the IEEE-EMBS, Amsterdam, IEEE, pp. 2 pp.
- Bortz JE (1971): A new mathematical formulation for strapdown inertial navigation. *IEEE transactions on aerospace and electronic systems* 7: 61-66.
- Crago PE, Chizeck HJ, Neurman MR and Hambrecht FT (1986): Sensors for use with functional neuromuscular stimulation. *IEEE Transactions on Biomedical Engineering* 33: 256-268.
- Doubler JA and Childress DS (1984): An analysis of extended physiological proprioception as a prosthesis-control technique. *Journal of Rehabilitation Research* 21: 5-18.
- Graupe D (1989): EMG pattern analysis for patient-responsive control of FES in paraplegics for walker-supported walking. *IEEE Trans Biomed Eng* 36: 711-9.
- Graupe D and Kohn KH (1998): Functional neuromuscular stimulator for short-distance ambulation by certain thoracic-level spinal-cord-injured paraplegics. *Surg Neurol* 50: 202-7.
- Graupe D, Kohn KH and Basseas S (1988): Above- and below-lesion EMG pattern mapping for controlling electrical stimulation of paraplegics to facilitate unbraced walker-assisted walking. *J Biomed Eng* 10: 305-11.
- Haugland M and Sinkjaer T (1999): Interfacing the body's own sensing receptors into neural prosthesis devices. *Technol Health Care* 7: 393-9.
- Haugland MK, Hoffer JA and Sinkjaer T (1994): Skin contact force information in sensory nerve signals recorded by implanted cuff electrodes. *IEEE Transactions on Rehabilitation Engineering* 2: 18-40.
- Heller BW, Veltink PH, Rijkhoff NJ, Rutten WL and Andrews BJ (1993): Reconstructing muscle activation during normal walking: a comparison of symbolic and connectionist machine learning techniques. *Biol Cybern* 69: 327-35.
- Hoffer JA, Stein RB, Haugland MK, Sinkjaer T, Durfee WK, Schwartz AB, Loeb GE and Kantor C (1996): Neural signals for command control and feedback in functional neuromuscular stimulation: a review. *J Rehabil Res Dev* 33: 145-57.
- Hogan N and Mann RW (1980a): Myoelectric signal processing: optimal estimation applied to electromyography - Part I: derivation of the optimal myoprocessor. *IEEE Transactions on Biomedical Engineering* 27: 382-395.
- Hogan N and Mann RW (1980b): Myoelectric signal processing: optimal estimation applied to electromyography - Part II: experimental demonstration of optimal myoprocessor performance. *IEEE Transactions on Biomedical Engineering* 27: 396-410.
- Johnson MW, Peckham PH, Bhadra N, Kilgore KL, Gazdik MM, Keith MW and Strojnik P (1999): Implantable transducer for two-degree of freedom joint angle sensing. *IEEE Trans Rehabil Eng* 7: 349-59.

- Kaczmarek KA and Bach-Y-Rita P (1995). Tactile displays. In: W. Barfield and T. A. Furness III, eds., *Virtual Environments and Advanced Interface Design*. New York, Oxford, Oxford University Press. pp. 349-414.
- Kirkwood CA, Andrews BJ and Mowforth P (1989): Automatic detection of gait events: a case study using inductive learning techniques. *J. Biomed. Eng.* 11: 511-516.
- Kostov A, Andrews BJ, Popovic DB, Stein RB and Armstrong WW (1995): Machine learning in control of functional electrical stimulation systems for locomotion. *IEEE Trans Biomed Eng* 42: 541-51.
- Kovacs GT, Storment CW, Halks-Miller M, Belczynski CR, Jr., Della Santina CC, Lewis ER and Maluf NI (1994): Silicon-substrate microelectrode arrays for parallel recording of neural activity in peripheral and cranial nerves. *IEEE Trans Biomed Eng* 41: 567-77.
- Luinge HJ, Veltink PH and Baten CT (1999): Estimating orientation with gyroscopes and accelerometers. *Technol Health Care* 7: 455-9.
- Parker PA and Scott RN (1986): Myoelectric control of prostheses. *CRC Critical Reviews in Biomedical Engineering* 13: 283-310.
- Pfurtscheller G (1999): Bio-inspired systems - EEG-based brain-computer interface. *lecture notes in computer science* : 250-254.
- Popovic DB, Stein RB, Jovanovic KL, Dai R, Kostov A and Armstrong WW (1993): Sensory nerve recording for closed-loop control to restore motor functions. *IEEE Trans Biomed Eng* 40: 1024-31.
- Riso R, Mosallaie FK, Jensen W and Sinkjaer T (2000): Nerve cuff recordings of muscle afferent activity from tibial and peroneal nerves in rabbit during passive ankle motion. *IEEE Transactions on Rehabilitation Engineering* : accepted.
- Riso RR (1999): Strategies for providing upper extremity amputees with tactile and hand position feedback - moving closer to the bionic arm. *Technology and Health Care* 7: 401-409.
- Rutten WL, Smit JP, Frieswijk TA, Bielen JA, Brouwer AL, Buitenweg JR and Heida C (1999): Neuro-electronic interfacing with multielectrode arrays. *IEEE Eng Med Biol Mag* 18: 47-55.
- Shoval S, Borenstein J and Koren Y (1998): The Navbelt - A computerized travel aid for the blind based on mobile robotics technology. *IEEE Transactions on Biomedical Engineering* 45: 1376-1386.
- Simpson DC (1974). The choice of control system for the multimovement prosthesis: extended physiological proprioception. In: P. Herberts, eds., *The control of upper-limb prostheses and orthoses*. Springfield, C.C. Thomas. pp. 146-150.
- Sinkjaer T (1999). *Interfacing the bodies own sensing receptors into neural prosthesis devices*. International Biomechatronics Workshop, Enschede, pp. 193-199.
- Smith RV (1990). Audiobiofeedback for tasks. In: R. V. Smith and J. H. Leslie, eds., *Rehabilitation Engineering*. Boca Raton, CRC Press. pp. 29-78.
- Soderkvist J (1994): Micromachined gyroscopes. *Sensors and Actuators A* 43: 65-71.
- Stieglitz T, Beutel H and Meyer I-U (1997): A flexible, light-weight multichannel sieve electrode with integrated cables for interfacing regenerating peripheral nerves. *Sensors and Actuators* 60: 240.
- Struijk JJ (1997): The extracellular potential of a myelinated nerve fiber in an unbounded medium and in nerve cuff models. *Biophys J* 72: 2457-69.
- Struijk JJ, Thomsen M, Larsen JO and Sinkjaer T (1999): Cuff electrodes for long-term recording of natural sensory information. *IEEE Eng Med Biol Mag* 18: 91-8.
- Szeto AYJ and Riso RR (1990). Sensory feedback using electrical stimulation of the tactile sense. In: R. V. Smith and J. H. Leslie, eds., *Rehabilitation Engineering*. Boca Raton, CRC Press. pp. 29-78.
- Thorsen R (1999): An artefact suppressing fast-recovery myoelectric amplifier. *IEEE Trans Biomed Eng* 46: 764-6.

- Thorsen R, Ferrarin M, Spadone R and Frigo C (1999): Functional control of the hand in tetraplegics based on residual synergistic EMG activity. *Artif Organs* 23: 470-3.
- Veltink PH (1999): Sensory feedback in artificial control of human mobility. *Technology and Health Care* 7: 383-391.
- Veltink PH, Bussmann HB, de Vries W, Martens WL and Van Lummel RC (1996a): Detection of static and dynamic activities using uniaxial accelerometers. *IEEE Trans Rehabil Eng* 4: 375-85.
- Veltink PH, De Vries W, Baardman G, Hermens HJ, Sweeney P and Van Riel W (1995). *Design of an intention detection system for FES assisted mobility in paraplegics*. 5th Vienna International Workshop on Functional Electrostimulation, Vienna, pp. 43-46.
- Veltink PH, De Vries W, Hermens HJ, Baardman G, IJzerman M, Heinze S, Nene AV, Zilvold G and Boom HBK (1996b). A comprehensive FES control system for mobility restoration in paraplegics. In: A. Pedotti, M. Ferrarin, J. Quintern and R. Riener, eds., *Neuroprosthetics - from basic research to clinical applications*. Berlin, Springer - Verlag. pp. 163-169.
- Yoshida K and Horch K (1996): Closed-loop control of ankle position using muscle afferent feedback with functional neuromuscular stimulation. *IEEE Transactions on Biomedical Engineering* 43: 167-176.

