CZECH TECHNICAL UNIVERSITY IN PRAGUE

FACULTY OF MECHANICAL ENGINEERING

Department of Mechanics, Biomechanics and Mechatronics

System for determination of shoulder joint motion activity

MASTER'S THESIS

2016 **Bc. Adam Kratochvíl**

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ČESKÉ VYSOKÉ UČENÍ TECHNICKÉ V PRAZE

ZADÁNÍ DIPLOMOVÉ PRÁCE

I. OSOBNÍ A STUDIJNÍ ÚDAJE

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Studijní program: **Strojní inženýrství**

Studijní obor: **Biomechanika a lékařské přístroje**

II. ÚDAJE K DIPLOMOVÉ PRÁCI

Název diplomové práce:

Systém pro určování pohybové aktivity ramenního kloubu

Název diplomové práce anglicky:

System for determination of shoulder joint motion activity

Pokyny pro vypracování:

1. Rešerše literatury
2. Návrh zařízení pro měření pohybu a signálů EMG
3. Sestavení prototypu zařízení
4. Experimentální verifikace
5. Návrh vyhodnocení naměřených signálů

Seznam doporučené literatury:

1: Morlock M, Schneider E, Bluhm A, Vollmer M, Bergmann G, Müller V, Honl M. Duration and frequency of every day activities in total hip patients. J Biomech. 2001 Jul;34(7):873-81.
2: Voda, Z a kol. Průvodce světem Arduina, Nakladatelství Martin Stříž, 2015
3: Heuberer P, Kranzl A, Laky B, Anderl W, Wurnig C. Electromyographic analysis: shoulder muscle activity revisited. Arch Orthop Trauma Surg. 2015 Apr;135(4):549-63.

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Jméno a pracoviště konzultanta(ky) diplomové práce:

Datum zadání diplomové práce: **12.04.2016** Termín odevzdání diplomové práce: **12.08.2016**

Platnost zadání diplomové práce: ____________

Podpis vedoucí(ho) práce | Podpis vedoucí(ho) ústavu/katedry | Podpis děkana(ky)

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Citation of this thesis

Kratochvíl, Adam. *System for determination of shoulder joint motion activity*. Master's thesis. Czech Technical University in Prague, Faculty of Mechanical Engineering, 2016.

Acknowledgements

First and foremost I offer my sincerest gratitude to my supervisor, Prof. RNDr. Matej Daniel, Ph.D., who has supported me throughout my work on this thesis with his advices, knowledge and patience. I am deeply grateful to him for the long discussions that helped me sort out all challenges of my thesis. I am thankful to Ing. Pavel Trnka, who taught me how to solder. I would also like to thank to doc. Ing. Patrik Kutílek, Ph.D. (FBMI, CTU) and Ing. Miloslav Vilímek, Ph.D. for their help with experimental verification and to Ing. Iva Novotná (FEL, CTU) for her help with data processing.

Anotace

Cílem této diplomové práce je navrhnout a sestrojit zařízení pro určování aktivity ramenního kloubu. Takovéto zařízení bude možné používat k měření bez jakéhokoliv omezení pro pacienta. Jako základ zařízení je použita programovatelná deska Arduino Leonardo. K zaznamenávání aktivity jsou použity 2 senzory – akcelerometr a povrchové EMG. Data z těchto senzorů jsou zaznamenávána na SD kartu s frekvencí 20Hz. Činnost zařízení byla experimentálně verifikována. Byl vytvořen algoritmus pro identifikaci jednotlivých aktivit na bázi rozhodovacího stromu.

Klíčová slova: zaznamenávání pohybu, Arduino, akcelerace, povrchové EMG, ramenní kloub

Annotation

Goal of this master’s thesis is to design and build a device for determination of shoulder joint motion activity. The device can be used during daily living activities without any distractions. Programmable desk Arduino Leonardo was chosen as a basis of the device. The activity of the shoulder joint is captured by an accelerometer and a surface EMG sensor. Data from these sensors are saved on SD card with sampling frequency of 20Hz. The device was experimentally verified. An algorithm based on a decision tree was developed to identify individual activities.

Key words: motion capture, Arduino, accelerometer, surface EMG, shoulder joint

Content

1 Introduction

Functional examination of joint mobility is one of the basic examination carried out in orthopedics. The function is measured in a number of different ways, such as through the use of impairment measures, self-report measures, and physical performance measures [1]. All of these methods have unique contributions and dedicated limitations.

Clinical special tests are used to determine the tissue source of pain and are impairment-based assessments. A typical clinical examination is measuring the range of motion (ROM) of joint using goniometers [2]. It is known, that limited ROM, alone or in combination with other factors, can contribute to limited function and ultimately may have consequences for physical functioning. For example, restricted ROM at the hip may lead to a limp during gait.

However, some clinical tests do not demonstrate high levels of sensitivity and/or specificity, thus questioning the validity of use [3],[4]. A clinical examination finding of an impairment does not always correspond to a functional loss. Poor ROM and other commonly used indicators of a limp may be subthreshold for patients whose activity levels are low, thus do not compromise the individual to the point where their expectations are altered [1]. The similar problem could be observed after surgery. Some measurable surgical end-points, even when they suggest success, do not always result in satisfied patients [5].

More accurate estimation of joint movement is based on motion analysis using modern electronic motion capture (MoCap) techniques and motion parameters reflecting the complexity of movement activity. Motion analysis systems, either passive or active, require specific setup, calibration and markers placement on patient. These analyses are time consuming and are carried out mostly in biomechanics laboratories [6]. Clinical application of motion analysis is restricted to certain diagnoses like cerebral palsy in children and is not used in everyday clinical practice.

Wearable motion trackers have been introduced recently. Their accuracy is relatively low and serve to rather estimate the level of activity than to provide quantitative information on ROM and daily range of motions [7].

In addition to objective measurement of joint excurse, the joint function could be evaluated by self-reports. The self-reports with ROM may be combined in clinical score, e.g. Harris Hip Score or Mayo Hip Score [5],[8]. Self-report measures are highly subjective, should be utilized cautiously, and serve only as one component of the assessment of function.

In the orthopaedic day to day clinical life there is a strong need from the physician´s point of view to monitor progress of patient´s recovery after surgical or conservative treatment of major trauma, congenital affections, degenerative or neurogenic diseases, with respect to joint mobility control in particular. The most typical anatomical localities requiring gradual dosing of ROM are elbow, knee, shoulder and hip. The motion pattern recommended to patient in the recovery phase differs from both the type of the injury and the anatomical locality (type of natural joint motion). Therefore, there is a strong clinical need to introduce a tool helping the orthopaedic surgeon to actively control and monitor the resting, rehabilitation and progressive phases of the patient´s recovery.

2 State of Art

2.1 Anatomical analysis of shoulder joint

Glenohumeral joint (articulatio humeri) is a simple spherical loose joint (arthrodia) that means - the articulation is formed between two bones only: humerus (os humeri) and scapula (os scapulae) - spherical head of humerus (caput humeri) and glenoidal cavity of scapula (cavitas glenoidalis scapulae). The Area of the joint „ball“ is larger than the area of the joint cavity, thus such parameters allow a large range of movement between bones which provides constant contact even at large displacements from the basic position. It is the most mobile joint in the human body. The centre position of the shoulder joint is in the partial abduction (45 °) and mild ventral flexion (the position at which the joint capsule is released and contact surfaces have the largest contact area).

2.1.1 Mobility

The mobility of a shoulder joint can be described by the position of humerus with respect to three orthogonal axes - Fig. 1. The joint allows flexion, extension, abduction, adduction, rotation and circumduction (wheeling by sum of flexion, abduction, extension and adduction in the course of circumduction, the ventral side of arm flies still ahead).

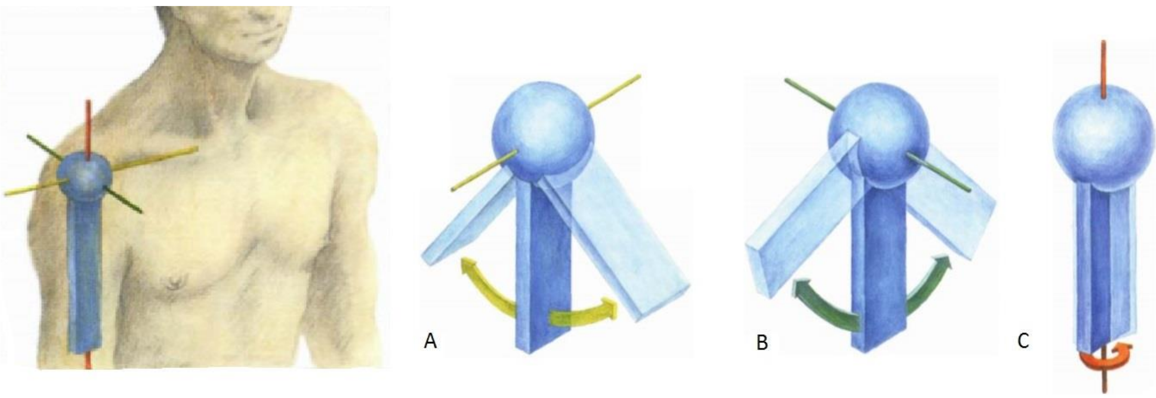

Fig. 1, A – flexion and extension, B – abduction and adduction, C – rotation [9].

The ranges of motions are presented in Tab. 1 and Fig 2.

1	Ventral flexion	0° - 90° (scapula diales up to 180°)
2	Dorsal flexion - extension	0° - 50°
3	Abduction	0° - 90° (scapula diales up to 180°)
4	Adduction and hyperadduction	0° - 75°
5	Internal rotation	0° - 90°
6	External rotation	0° - 90°

Tab. 1, the ranges of movements by shoulder joint [10].

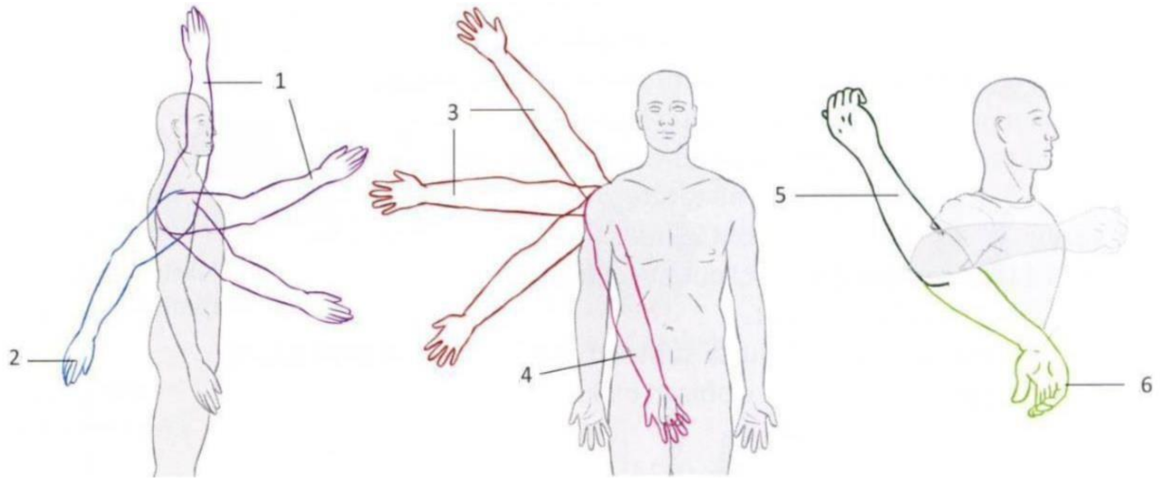

Fig. 2, movements of upper limb in shoulder joint, the movements according to Tab. 1 [10].

There is always a combination of these separated motions in real life. For example, during the most common daily activity which is walking, shoulders go forwards and backwards, that means there is a combination of ventral and dorsal flection. Arms hang loose and shoulders do opposite moves to hips in order to keep balance, so arms act like pendulums during walking [11].

As an example of a really complex motion in shoulder we could mention swimming breaststroke. Breaststroke starts in 180° ventral flection and goes down to body by adduction with external rotation and then back upwards by ventral flection with internal rotation.

Fig. 3, anatomy of breaststroke [12].

2.1.2 Ligaments and tendons

One of the most important ligament in the shoulder is glenoid labrum (labrum glenoidale - Fig. 4 [glenoid lig.]). It is a ring of fibrocartilage attached to the margin of the glenoid cavity of the scapula to increase its depth [13]. As already mentioned before, the socket is much smaller (it has only about one third of the area of the head) and shallow, so it has to be extended by this ring. Despite the presence of a glenoidal lip, the shoulder joint is most vulnerable to dislocation from all joints in human body.

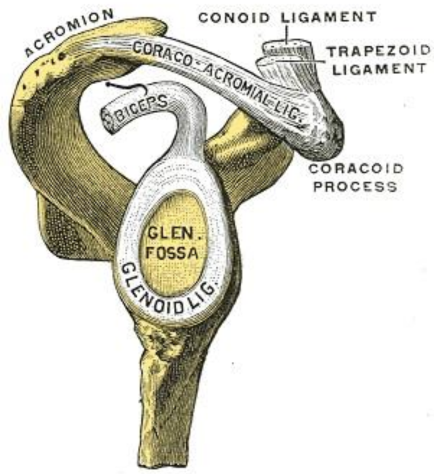

Fig. 4, labrum glenoidale – glenoid lig. [14].

Joint capsule, shown in Fig. 5, begins at the periphery of the joint socket and closes on the collum anatomicum humerus at the inner side of the joint distally further away. The synovial membrane is separated on the ventral side of the capsule into intertubercularis sulcus and along the tendon caput longum biceps brachii, thus forming the synovial packaging. Reinforcement of the capsule is provided by the closest muscles, their tendons and joint ligaments. Tendons reinforcing the capsule include the following muscles: the back - m. Supraspinatus, m.infraspinatus and m. Teres minor, the front - m. Subscapularis. File of these muscles and their tendons are clinically referred to as rotator cuff. [13]

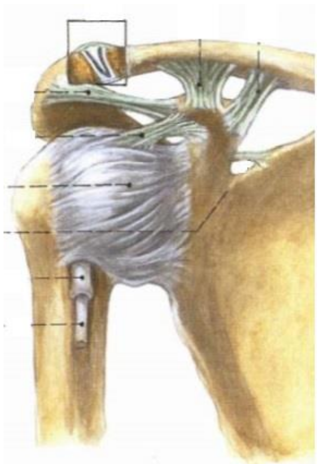

Fig. 5, joint capsule [9].

Ligaments of shoulder joint are (Fig. 6):

1 lig. coracohumerale
2 lig. glenohumeralia
3 lig. coracoacromiale

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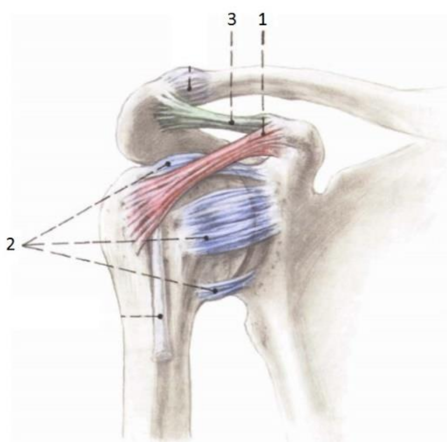


Fig. 6, ligaments of shoulder joint [9].

2.1.3 Muscles

Muscles, whose main function is to provide motion in the shoulder joint belong to the groups of muscles of shoulder and shoulder blade, chest and back. Selected muscles and their functions in relation to the shoulder joint are presented in Tab. 2.

Four of these muscles - (1) supraspinatus, (2) ifraspinatus, (3) teres minor and (4) subscapularis - are forming rotator cuff (Fig. 7) which stabilizes the shoulder joint and allows its rotation.

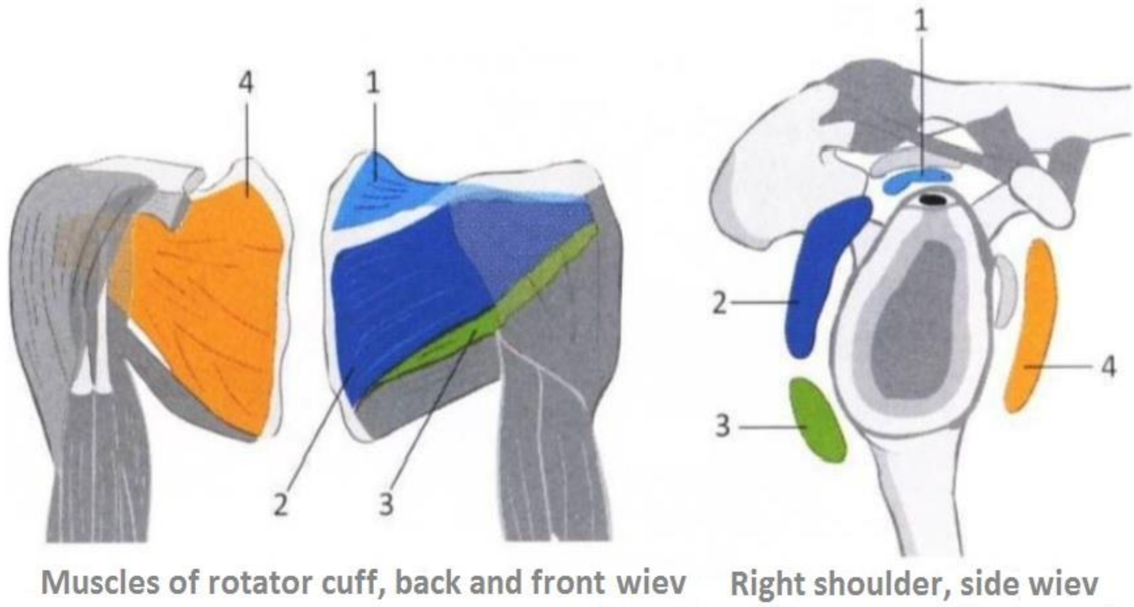


Fig. 7, rotator cuff [10].

Other muscles which are involved in the movement of the shoulder have only a complementary function in reference to the movement and include the following muscles:

biceps brachii – ventral flexion, abduction and adduction

triceps brachii – dorsal flexion and adduction

Finally, the muscles that provide elevation or depression and rotation of the shoulder blade (which are in many cases associated with movements of the shoulder) are:

trapezius, rhomboideus major et minor, levator scapulae and pectoralis minor

Name	Origin of the muscle	Insertion of the muscle	Main function	Additional functions
Deltoideus	clavicula, acromion, spina scapulae	tuberositas deltoidea humeri	dorsal & ventral flexion, abduction & adduction	by resting tension keeps the head in the socket
Supraspinatus	fossa supraspinata	tuberculum majus humeri	abduction & adduction	external rotation
Infraspinatus	fossa infraspinata	tuberculum majus humeri	external rotation	abduction
Teres minor	margo lateralis scapulae	tuberculum majus humeri	external rotation	dorsal & ventral flexion, adduction & abduction, fixation caput humeri
Teres major	margo lateralis a angulus inferior scapulae	crista tuberculi minoris humeri	adduction & internal rotation	
Subscapularis	scapula	tuberculum minus humeri	internal rotation	dorsal flexion, adduction
Coracobrachialis	processus coracoideus scapulae	½ délky humeru	ventral flexion	adduction, internal rotation
Pectoralis major	sternum, costae 1-6, clavicula	crista tuberculi majoris humeri	adduction	ventral flexion, abduction, internal rotation
Serratus anterior	costae 1-9	margo medialis a angulus inferior scapulae	abduction	
Latissimus dorsi	Th7-12, L1-5, sacrum, costae 10-12	crista tuberculi minoris humeri	dorsal flexion, adduction, internal rotation	

Tab. 2, muscles of shoulder joint [15].

According to Tab. 2, the major muscle contributing to the movement of the shoulder is m. Deltoideus (Fig. 8), which consists of three parts – posterior, anterior and lateral. Thus its main function is to support all the four main motions - ventral and dorsal flexion as well as abduction and adduction and also secures the joint head in the socket. Therefore, in the context of my thesis I will measure surface electromyography (SEMG) of this muscle as the main representative of the shoulder muscle group.

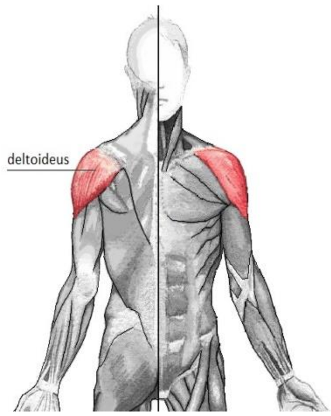


Fig. 8, m. Deltoideus [13].

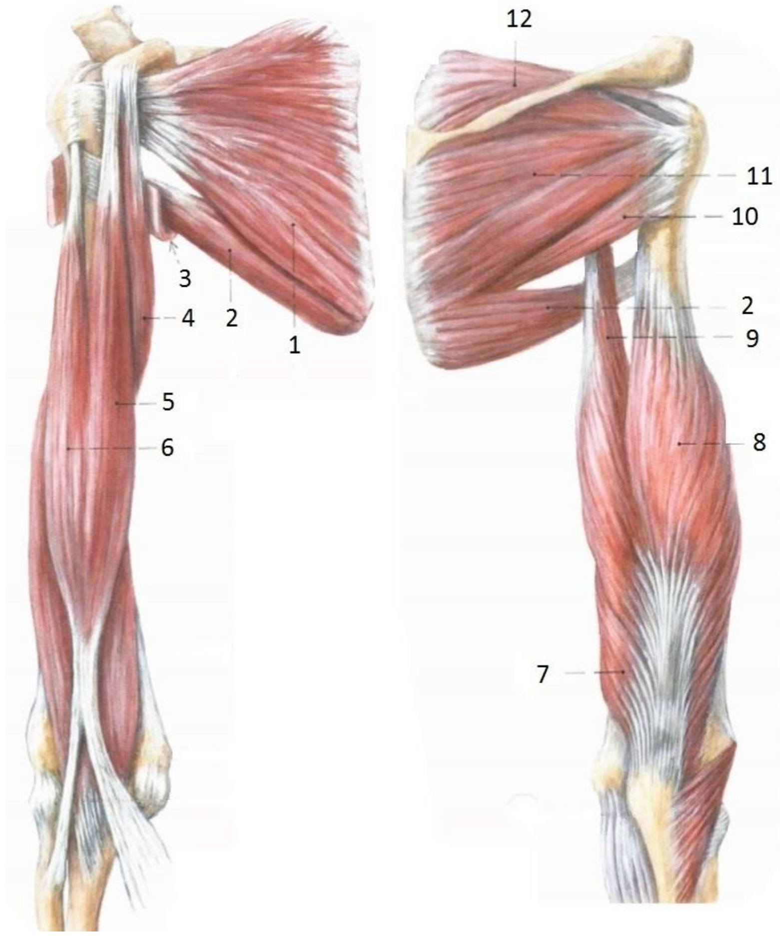


Fig. 9, muscles of arm,on the left – front wiev, on the right – back wiev [9]

In Fig. 9 is 1 – subscapularis, 2 – teres major, 3 – insertion of latissimus dorsi, 4 – coracobrachialis, 5 – biceps brachii caput breve, 6 – biceps brachii caput longum, 7 – triceps brachii caput mediale, 8 – triceps brachii caput laterale, 9 – triceps brachii caput longum, 10 – teres minor, 11 – infraspinatus, 12 – supraspinatus.

2.2 Motion capture

Motion capture (Mo-cap) is a process of recording the motions of objects or people. It is used in military, entertainment, sports, computer science, robotics and in medicine [13]. Currently, the widest use of the motion capture is in filmmaking and videogames for creating special effects. Mo-cap can be divided into two groups. In the first group there are lab systems, which are very accurate, but also really expensive according to our application and they have to be used in special environment (laboratory). They also require calibration. In the second group there are "real life" or "daily monitoring" systems such as smart bracelets and smart watches. They are much cheaper and simpler than the lab systems and they are used in daily activities. But they have to be used in combination with a smart phone and they give only basic information about subject's activities.

2.2.1 Laboratory systems

2.2.1.1 Optical systems

Optical systems use data captured from image sensors to triangulate the 3D position of a subject between two or more cameras calibrated to provide overlapping projections. Data acquisition is traditionally implemented using special markers attached to a subject. However, more recent systems are able to generate accurate data by tracking surface features identified dynamically for each particular human or object. Tracking a large number of subjects or extension of the capture area is realized by adding more cameras. Optical systems produce data with 3 degrees of freedom for each marker and information about rotation must be inferred from the relative orientation of at least three markers – e.g. markers attached to shoulder, elbow and wrist provide the angle of the elbow [16].

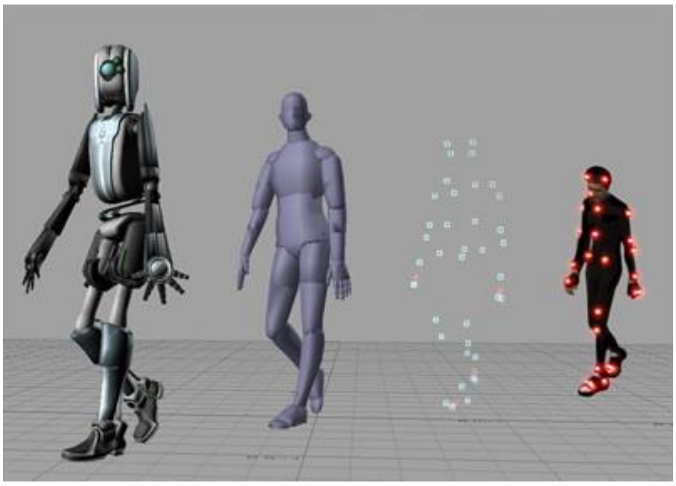


Fig. 10, processing by optical motion capture system [13].

Advantages of this method include: it provides very accurate data, the subject feels free to move due to wireless solution, more subjects can be observed simultaneously.

Disadvantages include: older systems are prone to light interference, markers can be blocked by other performers or other structures leading to potential loss of data, high cost [17].

Markerless optical systems also exist, but they need calibrated cameras and a detailed scan of the subjects. This method is still in the beginning [16].

2.2.1.2 Electromagnetic (Magnetic) systems

Magnetic systems detect position and orientation by the relative magnetic flux of three orthogonal coils on both the transmitter and each receiver. The relative intensity of the voltage and current of the transmitter allows these systems to calculate both range and orientation by meticulously mapping the tracking volume. The sensor output has 6 degrees of freedom, which provides useful results obtained with two-thirds of the amount of markers required in optical systems – one on upper arm and one on lower arm for elbow position and also angle [16]. One of the first use was in the military, where it was used to track head movements of pilots [17].

Advantages include: positions are absolute, rotations are measured absolutely and orientation in space can be determined, can be real-time, cheaper than optical, less markers than optical

Disadvantages include: response is nonlinear, especially toward edges of the capture area, the wiring from the sensors tends to preclude extreme performance movements, magnetic distortion occurs as distance increases, data can be noisy - it's not as good as optical, the markers are susceptible to magnetic and electrical interference from metal objects in the environment, e.g. wiring, which affect the magnetic field and electrical sources such as monitors, lights, cables and computers [16] [17].

2.2.1.3 Electromechanical (Mechanical) systems

Mechanical motion capture systems track body joint angles directly and are often referred to as exoskeleton motion capture systems, due to the way how the sensors are attached to the body. A performer attaches the exoskeletal structure to his body and as he moves so do the articulated mechanical parts, which measures the subject's relative motion. Mechanical motion capture systems are real-time, relatively low-cost, free-of-occlusion, and wireless (untethered) systems that have unlimited capture volume. Typically, it is a rigid structure of jointed, straight metal or plastic rods connected together with potentiometers that articulate at the joints of the performer's body [16].

The biggest advantages are: no interference from light or magnetic field and relatively low price.

Disadvantages include: the technology has no awareness of ground, equipment must be regularly calibrated, absolute positions are not known but they are calculated from the rotations [17].

Fig. 11, mechanical motion capture system (exoskeleton) [13].

2.2.1.4 Inertial systems

Inertial Motion Capture technology is based on inertial sensors such as accelerometers and gyroscopes, biomechanical models and sensor fusion algorithms. The motion data of the inertial sensors are often transmitted to a computer, where the motion is viewed and recorded. Gyroscopes are used by most inertial systems to measure rotational rates. These rotations are translated to a model in the software. Much like optical markers, the more gyros make the more natural data. No external cameras, emitters or markers are needed for relative motions, although they are required to give the absolute position of the user if desired. Inertial motion capture systems capture the full six degrees of freedom body motion of a performer in real-time [16]. For more precise information of subject location ultrasound or microwaves can be used for tracking the body of the subject and legs and arms are tracked by gyroscopes. This type of motion capture is similar to optical systems [18].

Advantages include: no need of any complicated postprocessing, ratio of portability and price, large capture areas.

Disadvantages include: lower positional accuracy and positional drift which can compound over time (can be solved by added ultrasound sensors for tracking the subject), 'floating' where the user looks like a marionette on strings, when used for filmmaking [16] [18].

2.2.1.5 Summary

All of these lab systems are good for accurate measuring. But they are really expensive and they have to be used in laboratory. That makes them not suitable for our application.

2.2.2 Daily monitoring systems

These systems are based mainly on accelerometers. There are many types of different wearable sensors. They can be divided into bracelets or watches as the two most common types. They are small and designed for daily use. But their functions are not sufficient for our application, because they are not connected to shoulder and their informative value is too low. The most common functions which these systems have are [19]:

1) Number of steps
2) Travelled distance
3) Burned calories
4) Time of sleep
5) Very active minutes

These are suitable functions for commercial use. These systems can tell when subject should go for a walk or when he should go to bed. These systems are designed to help the subject to live healthier or lose weight faster. We need a device for our application, which can tell more about what subject did during his day than what can these systems recognized.

Price of these systems starts at CZK 1,000 but these systems can be used only in connection with a smartphone (additional CZK 4,000 at least).

These systems can also use Global Positioning System (GPS), but it is not an important function for our use, because accuracy of this system is around 5 meters [20].

2.2.2.1 Fitbit flex

Fitbit flex was chosen as a representative of smart bracelets. It is a small silicon bracelet for daily use. It tracks daily activity and stores data into internal storage. It can be connected to a smartphone via Bluetooth and all tracked activities and graphs can be seen in real time. It can be used without a smartphone as well. The recorded data can be than viewed on a computer after connecting to Fitbit servers via internet [21].

It can record number of steps and walked distance, burned calories, very active time of the day and information about sleep. Battery life on one charge is 5 days of using. It is a commercial product, so it is not possible to find out how it evaluates these activities [21].

It costs around CZK 2,500, that means it is one of the more expensive bracelets.

Fig. 12, Fitbit flex bracelet and its interface [21].

2.3 Sensors

This section is intended to describe principles of sensors used mostly in motion detection and muscle activity recording.

2.3.1 Accelerometer

The function of an accelerometer is based on the property of inertia (Newton's First Law of Motion) [22]. An accelerometer measures proper acceleration, which is the acceleration it experiences relative to freefall and it is the acceleration felt by people and objects. Put another way, at any point in space-time the equivalence principle guarantees the existence of a local inertial frame and an accelerometer measures the acceleration relative to that frame. An accelerometer at rest relative to the Earth's surface will indicate 1g upwards, because any point on the Earth's surface is accelerating upwards relative to the local inertial frame [23].

Acceleration is quantified in the SI unit as metres per second squared (m/s^2) or popularly in terms of g-force (g) [23]. G-force is popular because the exact value of gravity acceleration which is not the same around the Earth has not to be known. That is caused by the rotation of the Earth. But if precise accuracy is not needed, thus average and used value for gravity acceleration is $9{,}81 m/s^2$ = 1 g-force.

An accelerometer conceptually behaves as a damped mass on a spring. When the accelerometer experiences an acceleration, the mass is displaced to the point that the spring is able to accelerate the mass at the same rate as the casing. The displacement is measured and gives the value of acceleration itself as shown in Fig. 13 [24].

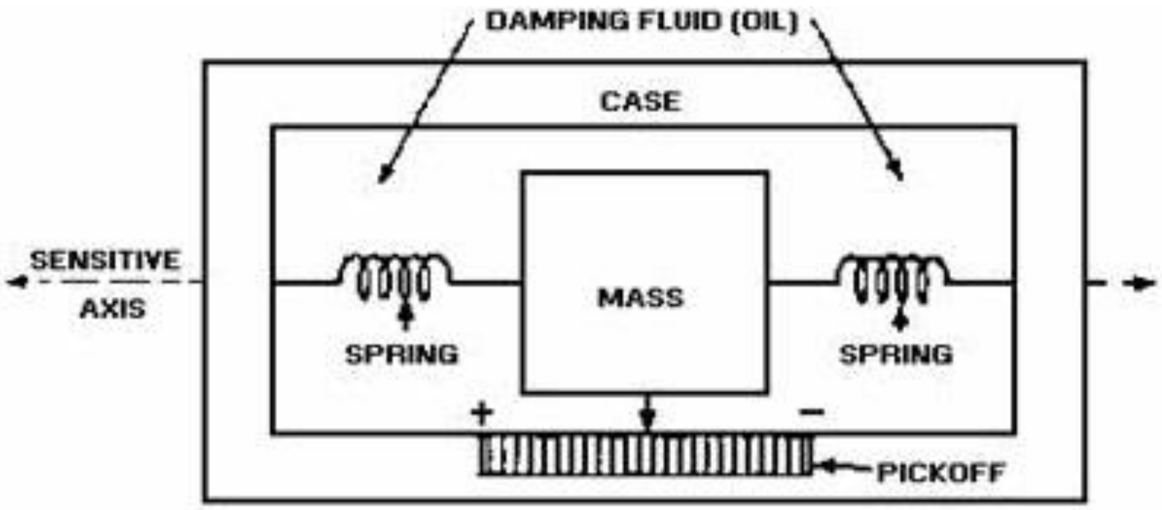


Fig. 13, principle of a basic accelerometer, this one is damped by a damping fluid [24].

In commercial devices, piezoelectric, piezoresistive and capacitive components are commonly used to convert the mechanical motion into an electrical signal.

Modern accelerometers are often really small micro electro-mechanical systems (MEMS) and they are the simplest MEMS devices possible. They consist of a cantilever beam with a proof mass (also known as seismic mass) [25].

MEMS-based accelerometer with capacitors is one of the most common types. It is typically a structure equipped by two capacitors formed by a moveable plate held between the two fixed plates. Under the zero net force, the two capacitors are equal, but if the force changes, it will cause the moveable plate to shift closer to one of the fixed plates. That increases the capacitance and further away from the other fixed plate reduces the capacitance. This difference in capacitance is detected and amplified to produce a voltage proportional to the acceleration. The dimensions of the structure, since it is MEMS, are of the order of microns [23].

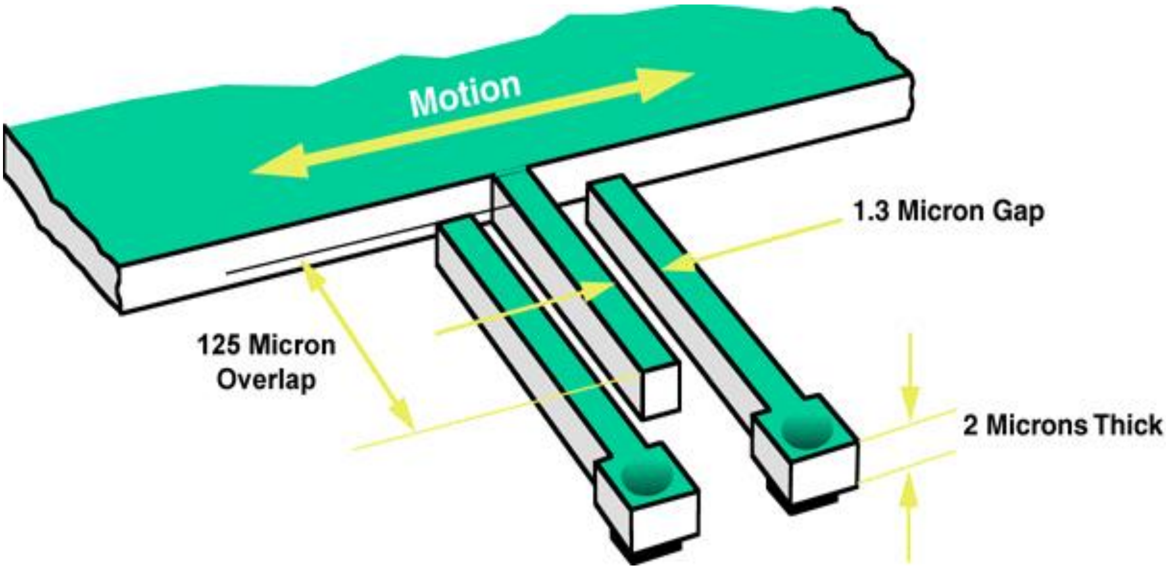


Fig. 14, principle of MEMS accelerometer based on capacitors [23].

2.3.2 Electromyography

Electromyography (EMG) is an electrodiagnostic technique used in medicine for evaluating and recording the electrical activity produced by skeletal muscles. Motor neurons transmit electrical signals that cause muscles to contract. EMG is performed by using an instrument called an electromyograph to generate a record called an electromyogram. An electromyograph detects the electrical potential between its electrodes generated by muscle cells. The electrical potential occurs when these cells are electrically or neurologically activated. The signal can be analyzed to detect medical abnormalities, activation level of muscles or recruitment order of muscles or to define the biomechanics of human or animal movement [26].

There are two kinds of EMG: intramuscular EMG and surface EMG (SEMG).

Intramuscular EMG can be performed using a different types of recording electrodes. The simplest one is a monopolar needle electrode. This can be a fine wire inserted into a muscle with a reference surface electrode. Another way is two fine wires inserted into muscle referenced to each other. Most commonly fine wire recordings are used for research or kinesiology studies.

Surface EMG assesses muscle function by recording muscle activity from the surface above the muscle on the skin. Surface electrodes are able to provide only a limited assessment of the muscle activity. SEMG can be recorded by a pair of electrodes or by a more complex array of multiple electrodes. More than one electrode is needed because EMG recordings display the potential difference (voltage difference) between two separate electrodes. Limitations of this approach include the fact that surface electrode recordings are restricted to superficial muscles, are influenced by the depth of the subcutaneous tissue at the site of the recording which can be highly variable depending of the weight of a patient, and cannot reliably discriminate between the discharges of adjacent muscles [26].

3 Aim of the thesis

Objective knowledge of joint function is crucial in therapy and rehabilitation of orthopaedic and trauma patients. There is a gap between the precise and costly laboratory systems for motion analysis requiring calibrations and extensive post-processing and simple motion trackers reporting activity level of the patient. There is currently no easily accessible device to determine joint function quantitatively not only during clinical examinations but also and mainly in everyday living. Therefore **the aim of the thesis is to design a system for objectification of functional status of the joint during clinical examination as well as daily use and for reasonable price.**

Specific aims of the thesis are:

1) Design and build a device for measuring accelerations and surface EMG. Adding a SEMG sensor is the main novelty of our system. It should be built from commercially available components. Arduino board was chosen as the basis of the device. Sensors were chosen from Analog devices Inc. and Advancer Technologies, LLC.
2) Program the device to work properly. Make the code for measuring accelerations and SEMG during daily living activities and write it on SD card for future evaluating.
3) Make an experimental verification with standardised devices. Xsens (Xsens Technologies B.V., The Netherlands) device was chosen to verify accelerometer. MA300 (Motion Lab Systems, Inc, USA) device was chosen to verify surface EMG sensor. Both of them were used due to their availability on Czech Technical University in Prague.
4) Proposal of measured data evaluating. Measure some of the basic motions and find parameters for distinction between them. Write an algorithm for evaluating the data in MatLab and verify the algorithm.

4 Methods

4.1 Construction of the device

4.1.1 Microcontroller board

Our choice of the microcontroller board is Arduino Leonardo, which is one of the cheapest boards and it has a Micro USB port. There is a possibility of connection an external shield with SD card slot and communicate with it by a SPI library (Serial Pheripheral Interface) and a SD library.

The Arduino Leonardo microcontroller board is based on the ATmega32u4 by Atmel Corporation. It has 20 digital input/output pins (of which 12 can be used as analog inputs), a 16MHz crystal oscillator, a power jack, a micro USB connection, an ICSP header and a reset button [27].

Fig. 15, Arduino Leonardo board, front and back side [27].

The board can be powered via micro USB connector or with external source such as a battery or an AC-to-DC adapter. The adapter can be connected by a 2.1mm center-positive plug into the board's power jack. Leads from a battery can be inserted in the Gnd and Vin pin headers of the power connector. The board operates on an external supply of 6V – 20V. If it is supplied with less than 7V, then the 5V pin may supply less than 5V and the board can be unstable. If more than 12V is used as an external supply, the voltage regulator may overheat and damage the board. Thus the recommended range is 7V – 12V [27].

The microcontroller ATmega32u4 has 32KB (with 4KB used for the bootloader). It also has 2.5KB of SRAM and 1KB of EEPROM (which can be read and written with the EEPROM library) [27].

The board contains a 6 channel10-bit analogue to digital converter. This means that it will map input voltages between 0 and 5 volts into output integer values between 0 and 1023. This yields a resolution between readings of 5V / 1024 units or .0049V (4.9mV) per unit [27].

The Leonardo board has different facilities for communicating with a computer, another Arduino board or other microcontrollers. The ATmega32U4 provides UART TTL (5V) serial communication, which is available on digital pins 0 (RX) and 1 (TX). The ATmega32U4 also supports serial (CDC) communication over USB and appears as a virtual com port to software on the computer. The chip also acts as a USB 2.0 device with full speed, using standard USB COM drivers. The Arduino software includes a serial monitor, which allows sending of simple textual data from and to the Arduino board. The RX and TX LEDs on the board are flashed when data is being transmitted via the USB connection to the computer [27].

The maximum length and width of the Leonardo board with the USB connector and power jack extending beyond the former dimension are 6.86cm and 5.33cm, respectively. The board is equipped with four screw holes which allow the board to be attached to a surface or case. The distance between digital pins 7 and 8 is 0.4cm, not an even multiple of the 0.254cm (0.1 inch) spacing of the other pins [27].

All of these parameters are shown in Tab. 3, which is a summary of parameters and properties of the Arduino Leonardo board.

Microcontroller	ATmega32u4
Operating Voltage	5V
Input Voltage (recommended)	7V – 12V
Input Voltage (limits)	6V – 20V
A/D converter	10bit (0 – 1023)
Digital I/O Pins	20
PWM Channels	7
Analog Input Channels	12
DC Current per I/O Pin	40mA
DC Current for 3.3V Pin	50mA
Flash Memory	32KB (ATmega32u4) of which 4KB used by bootloader
SRAM	2.5KB (ATmega32u4)
EEPROM	1KB (ATmega32u4)
Clock Speed	16MHz
Length	68.6mm
Width	53.3mm
Weight	20g

Tab. 3 Arduino Leonardo summary [27].

Specifications regarding the microcontroller ATmega32u4 and a schematic graph of the Arduino Leonardo board are added in the electronic appendix.

4.1.2 SD card shield

SD card shield by Seeed Technology (Fig. 16) was selected as a mean for saving measured data. It provides read/write on SD card via Arduino's built-in SD library. It supports SD, SDHC and Micro SD cards. It occupies only the SPI port of the Arduino board. There is still a possibility to stack on other shields that work with the unused pins. Maximum capacity of the SD card is 32GB which is sufficient for our application. After all day measuring we have around 4MB of data, so due to the storage capacity, it could store data for over 23 years of measuring [28].

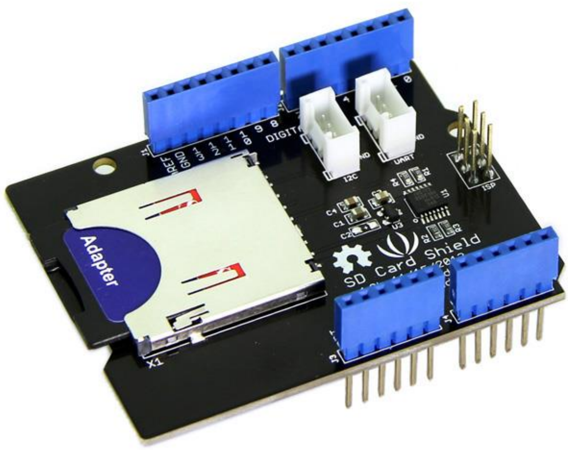


Fig. 16, SD card shield [28].

Item	Min	Typical	Max
Voltage	3.5V	5.0V	5.5V
Current	0.159mA	100mA	200mA
Supported Card Type	SD card(<=32GB); Micro SD card(<=32GB); SDHC card(<=32GB)		
Dimension	68.7mm x 53.5mm x 19mm		
Net Weight	14.8g		

Tab. 4, SD card shield summary [28].

If the SD card shield is connected to the Arduino board then the device is twice as high. But as the new dimensions are 68.7mm x 53.5mm x 27mm with weight around 35g, the size is still convenient for daily use.

A schematic graph of SD card shield is added in the electronic appendix.

4.1.3 Accelerometer

For acceleration measuring, the accelerometer ADXL335 (Analog Devices Inc., USA) was chosen. It is small 18mm x 18mm triple axis MEMS (micro electro-mechanical systems) accelerometer with low power consumption and noise.

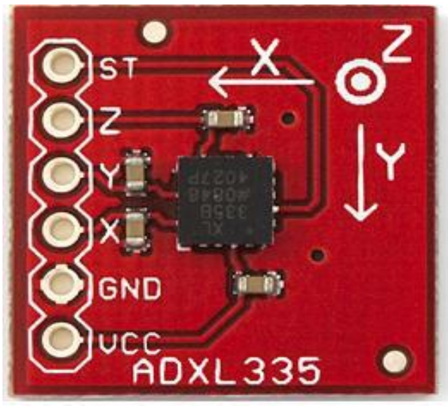


Fig. 17, accelerometer ADXL335 [29].

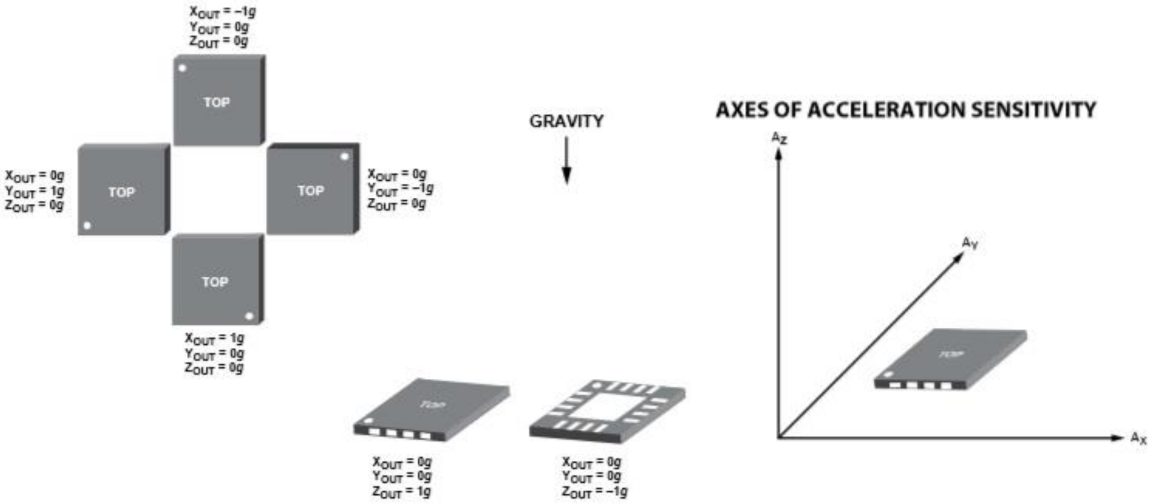


Fig. 18, axes and accelerations due to accelerometer ADXL335 [30].

The ADXL335 is a 3 axis acceleration measurement system. The ADXL335 has a minimum measurement range of ±3 g-forces (approximately ±30ms^{-2}). It has an open-loop acceleration measurement architecture which contains of a polysilicon surface-micromachined sensor and signal conditioning circuitry. The output signals are voltages that are proportional to acceleration. The accelerometer can measure both accelerations – static and dynamic. The static acceleration of gravity is used in tilt-sensing applications. The dynamic acceleration is connected with motion, shock, or vibration. The sensor is a polysilicon surface-micromachined structure built on the top of a silicon wafer. Polysilicon springs suspend the structure over the surface of the wafer and provide a resistance against any acceleration forces. Deflection of the structure is measured using a differential capacitor that consists of independent fixed plates and plates attached to the moving mass. The fixed plates are driven by 180° out-of-phase square waves. Acceleration deflects the moving mass and unbalances the differential capacitor resulting in a sensor output whose amplitude is proportional to acceleration. Phase-sensitive demodulation techniques are then used to determine the magnitude and direction of the acceleration [30].

The demodulator output is amplified and brought off-chip through a 32 kΩ resistor. The user then sets the signal bandwidth of the device by adding a capacitor. This filtering improves measurement resolution and helps prevent aliasing.

The ADXL335 uses a single structure for sensing the X, Y, and Z axes. As a result, the three axes' sense directions are highly orthogonal and have little cross-axis sensitivity. Mechanical misalignment of the sensor to the package is the chief source of the cross-axis sensitivity. Mechanical misalignment can be calibrated out at the system level.

Rather than using additional temperature compensation circuitry, innovative design techniques ensure that high performance is built in to the ADXL335. As a result, there is no quantization error or non-monotonic behaviour, and temperature hysteresis is very low (typically less than 3 mg-forces over the −25°C to +70°C temperature range) [30].

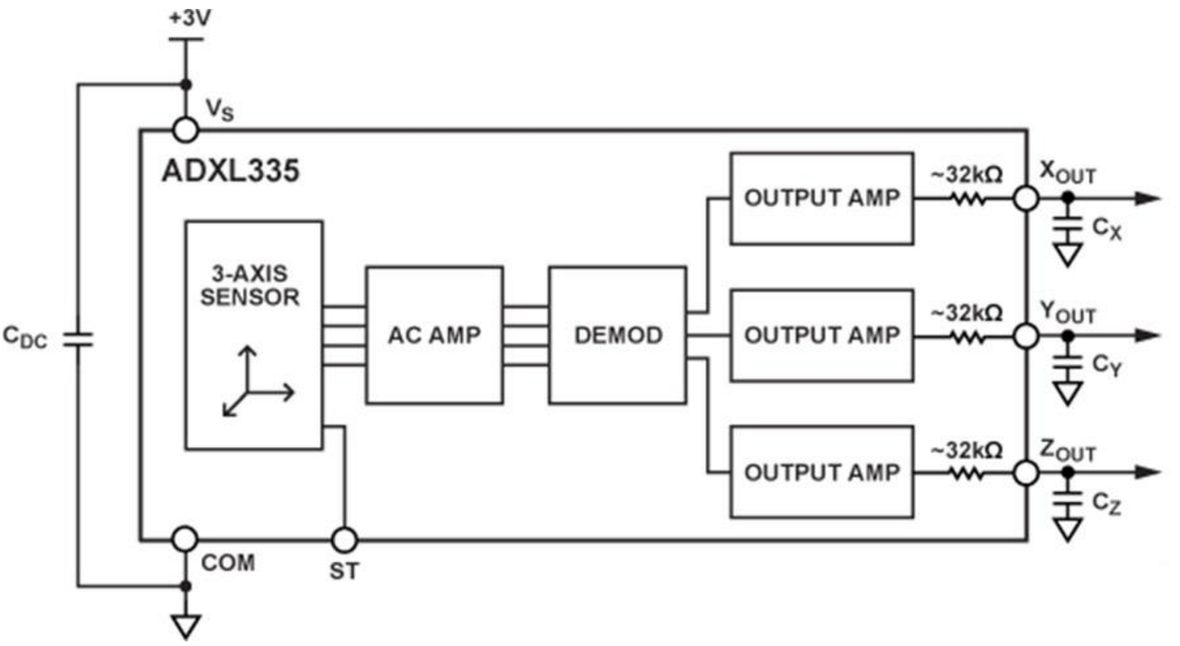


Fig. 19, ADXL335 - schematic figure [30].

Whole manual with all specifications and schematic graph is added in the electronic appendix.

4.1.4 Surface EMG sensor

Muscle sensor v3 (Advancer Technologies, LLC, USA) was chosen for measuring muscle activity. One of its biggest advantages is the output signal. It doesn't provide a raw EMG signal, which can be compared to a sinus wave. The output is amplified, rectified, and smoothed signal as shown in Fig. 21. The sensor outputs 0-Vs Volts depending on the amount of activity in the selected muscle, where Vs signifies the voltage of the power source.

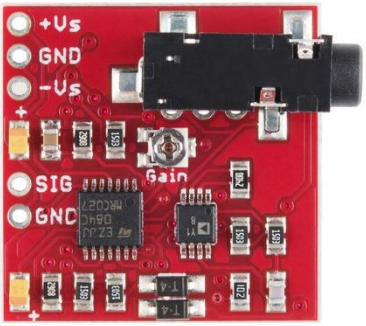


Fig. 20, SEMG sensor [31].

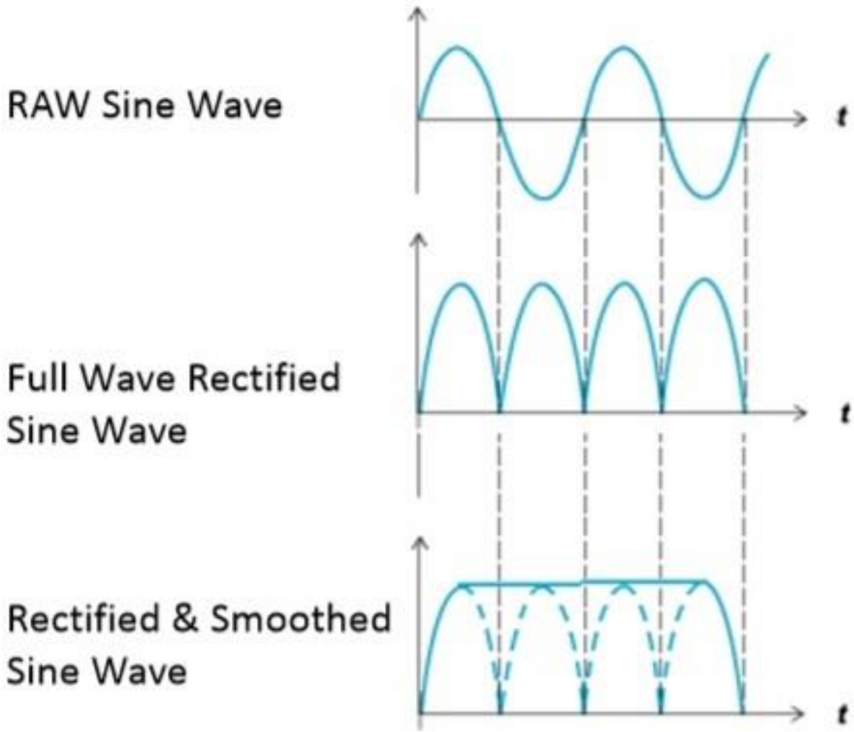


Fig. 21, EMG postprocessing [32].

This sensor uses bipolar differential SEMG which is actually the most common form of SEMG. Bipolar SEMG uses 3 electrodes instead of 2 as monopolar SEMG does. Bipolar SEMG gives much cleaner signal. It is caused by differential amplification in bipolar recordings that are able to selectively amplify the difference in the signal (from the muscle action potential) while supressing the common signal, i.e. the background noise. That is also the reason why the double-differential electrodes are even more effective in improving the signal-to-noise ratio.

Additional advantage of this sensor is its relatively small size 25mm x 25mm. On the other hand, the sensor must be externally powered by two 9V batteries, which is a disadvantage making the device bigger and heavier.

The whole SEMG measuring sensor consists of 3 electrodes with wires, sensor itself (Fig. 20) and two 9V batteries.

Electrical specifications are shown in Tab. 5.

Parameter	**Min**	**Typical**	**Max**
Power Supply Voltage (Vs)	±3V	±5V	±30V
Gain Setting, Gain = 207*(X /1kΩ)	0.01Ω (0.002x)	50kΩ (10,350x)	100kΩ (20,700x)
Output Signal Voltage (Rectified & Smoothed)	0V	-	+Vs
Differential Input Voltage	0mV	(2-5)mV	+Vs/Gain

Tab. 5, Electrical specifications of muscle sensor v3 [32].

The whole manual for Muscle sensor v3 is added in the electronic appendix.

Fig. 22, Top Trace electrode [33]

SEMG Top Trace electrodes (Fig. 22) are used for measuring. It is a foam pregelatinized Ag/AgCl electrode for connecting by stud. It is round with a diameter of 50mm. It is suitable for all applications that require a strong grip. The electrode should not remain on the skin for longer than 2 days according to producer’s recommendation [33].

4.1.5 Power source

There are two options how to power the Arduino board. It can be fed by a power jack or via a micro USB, thus one option is for example a 9V battery with a homemade connector to power jack. The second option is a power bank which is mostly used as an external battery for mobile phones. Both of these options have advantages and disadvantages. Power bank is more expensive than normal battery (70 cents for a 9V battery and around 20 euros for a power bank). But power bank can be used repeatedly and it shows the battery level and this is not possible with normal battery, thus it can die during the measuring. For that reason power bank was chosen as our power source for the Arduino board.

Fig. 23, Power bank CONNECT IT POWER BANK CI-247 [34].

Decisive criteria for choosing the right power bank were size, capacity, price, separated in and out ports and the battery level scale on the power bank.

After considering all of these criteria the CONNECT IT POWER BANK CI-247 was chosen. This power bank has sufficient properties with low costs.

CONNECT IT POWER BANK CI-247 summary is in Tab. 6.

Capacity	5200mAh
Input voltage	DC 5V
Output voltage	DC 5V
Input current	1A
Output current	2.1A
Weight	120g
Length x width x height	74mm x 53mm x 25mm

Tab. 6, CONNECT IT POWER BANK CI-247 summary [34].

4.1.6 Case

For using the device on a daily basis a suitable case has to be used. It is Vanguard BIIN II 7H (Fig. 24) with external dimensions of 150mm x 68mm x 85mm and internal dimensions of 115mm × 40mm × 70mm. The dimensions of the Arduino board with a SD card shield are 68.7mm x 53.5mm x 27mm and the dimensions of the battery are 74mm x 53mm x 25mm, thus it fits perfectly. Materials of the case are 210D Polyester outside and 150D Polyester inside [35].

The case can be attached to belt or shoulder strap can be used.

Fig. 24, Vanguard BIIN II 7H case [35].

4.1.7 Wiring

The board should be on a belt of the subject, that means in the height of waist and the sensors have to be on shoulder. The subject has to move with measured shoulder without any limitations, then the wires has to be at least 1m long. One meter long wire for Arduino is not a commercially available part. The longest wires that can be bought are about 15cm long. Thus this component of the device have to be custom-made.

Basic single-pin wires are used with Arduino boards. On one end there is a connector with a crimp connecter housing for a connection with the Arduino board and the second end is soldered to the sensors. Soldering of the wires to the sensors is necessary. Without it an uninterrupted connection along whole measuring cannot be guaranteed and there could be missing or distorted data.

For soldering the solder – S-Sn60Pb40E is used, which is a 1mm tin wire with an active rosin in the middle.

The wires are united together by spaghetti to be as a one thicker cable. All the wires were made long enough for subject to move freely in shoulder joint with the sensors on.

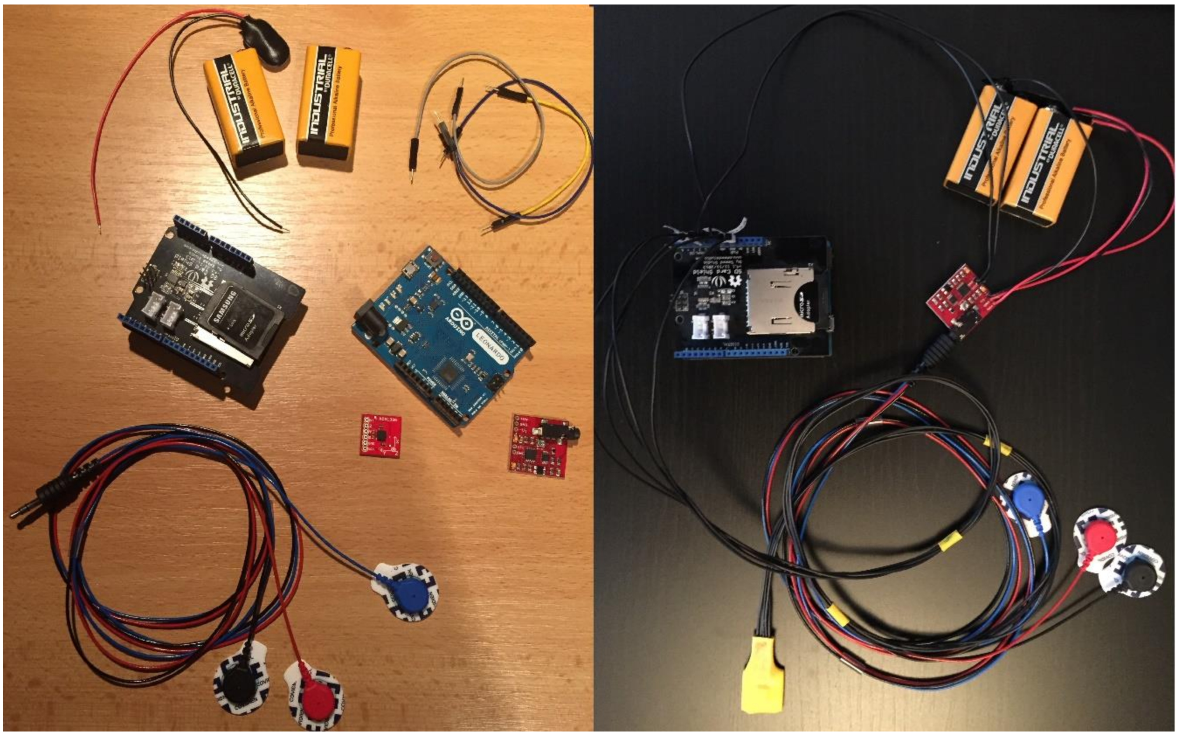


Fig. 25, components before and after soldering and adjustments.

4.1.8 Price of the device

Part	Price [CZK]
Arduino Leonardo	590
SD card shield	388
Accelerometer ADXL335	597
Muscle sensor v3	1,134
Power bank CI-247	449
Case Vanguard	449
Wires	20
Sum	3,627

Tab. 7, summary of prices of parts of our device.

As shown in Tab. 7, the final price of our device is CZK 3,672. It is a really low price compared to Xsens (Xsens Technologies B.V., The Netherlands) with which a verification of our accelerometer has been done. Just one sensor of Xsens motion capture device costs around CZK 20,000.Compared to a smart bracelet which costs around CZK 1,000 it is still a low price, because our device has a surface EMG sensor and not just an accelerometer. Moreover, in order to use the smart bracelet, a smartphone is needed, which even leads to a higher price compared to our device.

Additional costs are the electrodes, but they cost only CZK 2.50 per piece.

4.1.9 Experimental setup

For evaluating, methods like mean and standard deviation are used and the acceleration is evaluated as the sum of the tree vectors x, y, z. That makes the system robust against a slight disorder in sensor placement.

Accelerometer is placed in to analog pins in the Arduino boards. Axis x is in pin 0, axis y is in pin 1 and axis z is in pin 2. Ground is placed in the ground pin in the Arduino board and VCC is placed into 5V pin in the Arduino board. SEMG signal is placed in analog pin 3 in the Arduino board and ground is placed into ground as shown in Fig. 26.

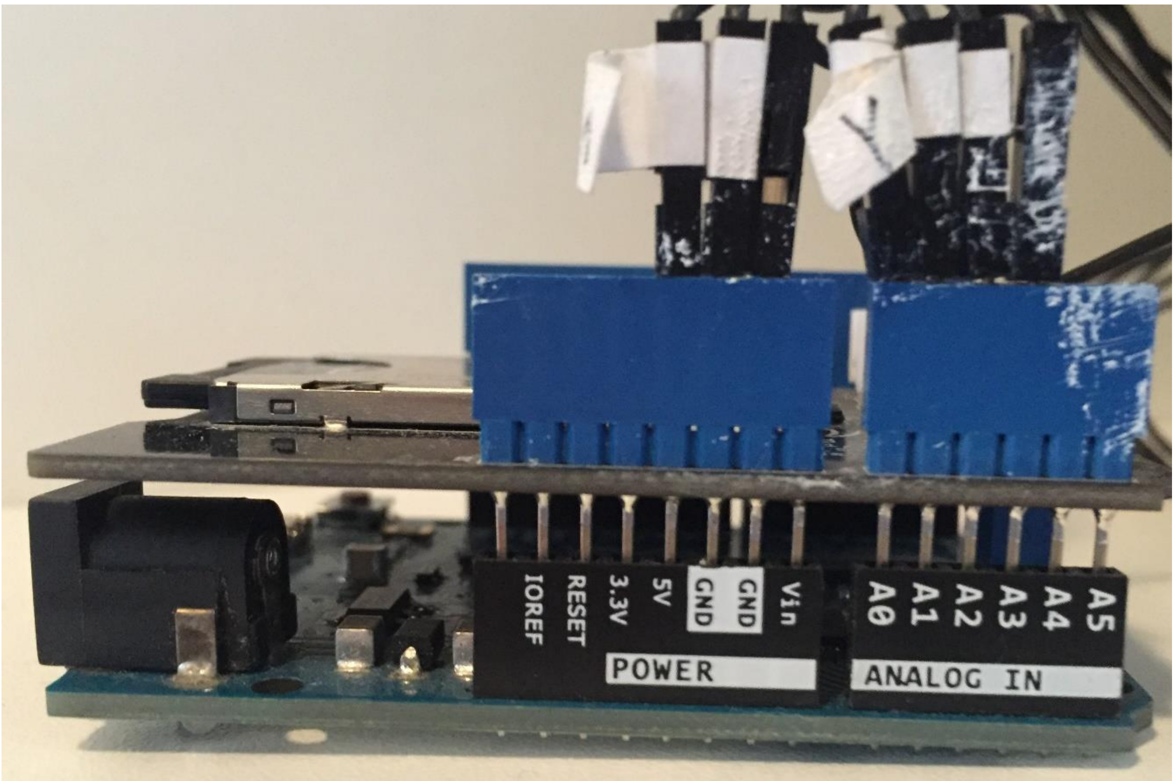


Fig. 26, senzors connected to the Arduino board.

The power source is placed in to the micro USB port on the board. Two 9V batteries are connected by clips to the SEMG sensor.

In Fig. 27, there is shown how to place the SEMG electrodes and the accelerometer. The black cable is placed on the electrode, which is placed on the bony part of the body near to the deltoid muscle. The red cable is placed on the electrode, which is in the middle of the muscle body. The blue cable is placed on the electrode, which is at the end of the muscle. The accelerometer is placed between the electrodes with the red and blue cable. Axis x of the accelerometer is in extension of the hand in basic position as shown in Fig. 27. Axis y heads backwards and axis z is directed away from the body. The accelerometer is anchored by a tape.

Fig. 27, placements of sensors.

The electrodes are placed with respect to the source [36] and the frontal plane of the body. The deltoid muscle can be divided into three parts – anterior, middle and posterior. We place the electrodes on the middle part of the deltoid muscle (Fig. 29) due to two reasons.

The first reason to choose the middle part of deltoid muscle is that it is the only part of the muscle which participates on flexion, abduction and also on extension in shoulder as shown in Fig. 28.

The second reason why to choose electrodes placement on the middle part of deltoid muscle is that with SEMG it is not possible to measure only one specific muscle. Advantage of the deltoid muscle is that it is a large muscle and it is directly on the surface with no other muscle crossing it over. So if we put electrodes in the middle of this big muscle then we have a probability high enough that we are measuring deltoid muscle EMG only.

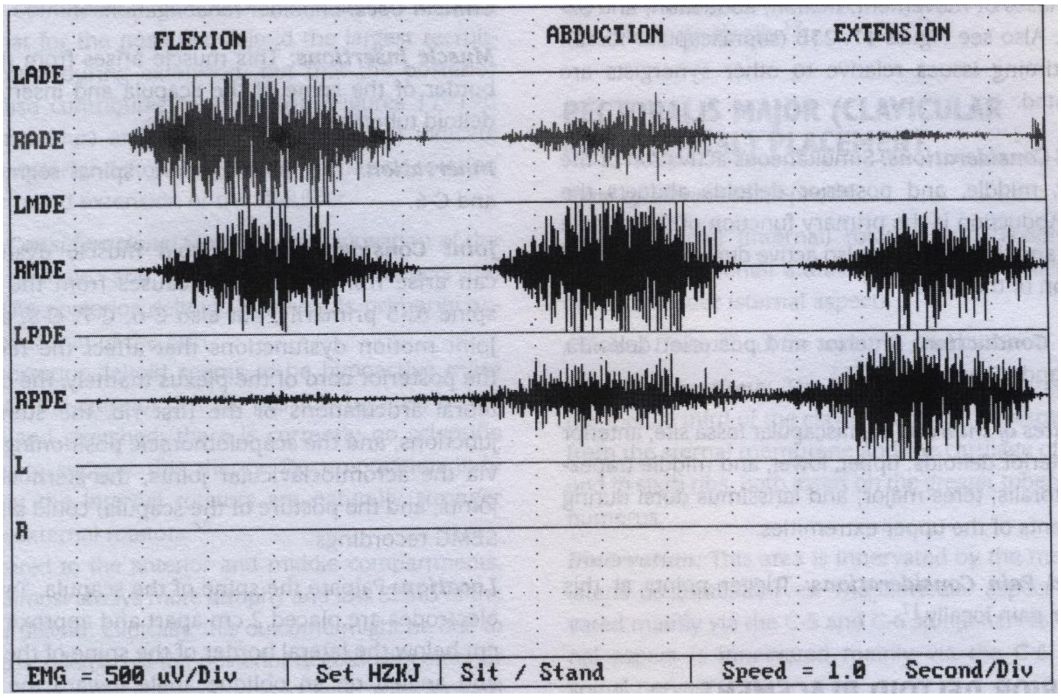


Fig. 28, EMG of the deltoid muscle, the first line refers to the anterior part, second one to the middle part and third one to the posteroir part of the deltoid muscle [36].

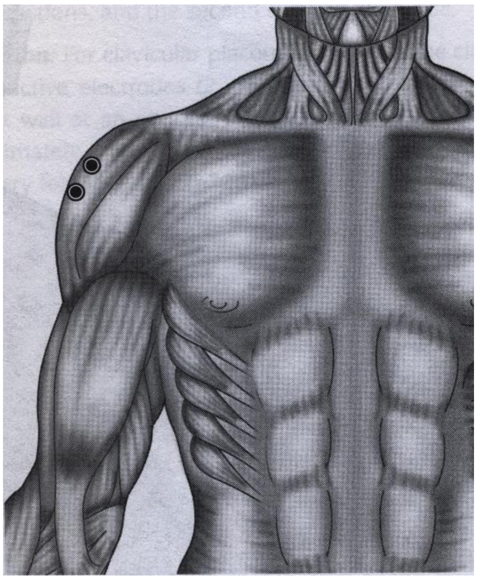


Fig. 29, middle part of deltoid electrodes placement [36].

In Fig. 30, a subject wearing the device is presented. Accelerometer and electrodes from the SEMG sensor are hidden under sleeve of a T-shirt, the only visible part is the case on the belt.

Fig. 30, wearing the device.

The accelerometer can be sow in to rubber for easy placing and removing without any assistance.

During the daily measuring, the sensors are attached by a big transparent waterproof plaster. This plaster is almost non removable, thus all sensors will stay at their positions.

4.2 Programming

Arduino has its own open-source software (IDE - Integrated Development Environment) for writing a code and an easy upload to the board. The code is similar to the C language and uses libraries, which make it easier to use.

4.2.1 Main sources helping with writing the code

To learn how to write the code and make programs these sources have been used:

[37] is Arduino's original web page, where every library that was already added to the Arduino code can be found. There are also many examples from the easiest one to the really complex codes.

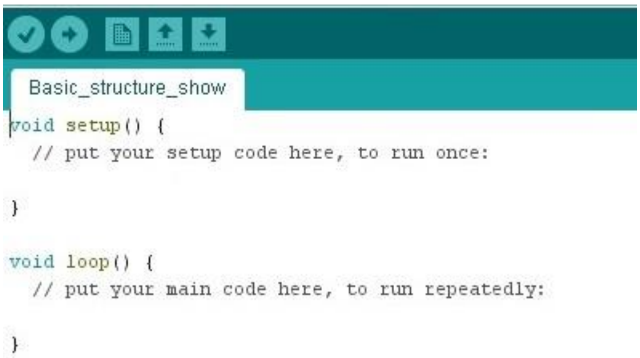
[38] is a forum on the official Arduino's page. User-written codes can be found on the webpage and also various Arduino-related topics are being discussed there. The forum is widely used, so the new questions are answered promptly. The forum is divided into sections such as – Using Arduino (Installation, Project Guidance, Programming questions, Sensors, etc.), Topics (Science and Measurement, Robotics, Device hacking, etc.), Development, Products and also International, where support for non-English speakers can be found as well (mainly in German and French).

[39] is a basic manual written in Czech language, which is also helpful for beginners. Basic structure of the code with commentary is to be found there, including tips on how to do basic things and operations such as variables, constants, data types, arithmetic operations, functions, etc.

[40] is also a manual but updated and much more comprehensive compared to [39]. It has 280 pages and it includes everything from different products of Arduino to the programming.

[41] is web http://arduino.cz/, which is a Czech analogy to web http://arduino.cc. There are forums, tutorials, articles about Arduino, etc. Source [40] is also from that web as a collection of most important articles and tutorials in one electronic book.

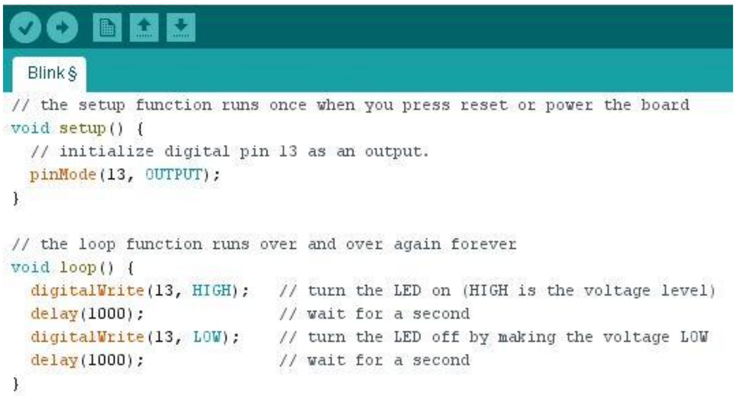
4.2.2 Basic structure of the program



```
Basic_structure_show
void setup() {
  // put your setup code here, to run once:

}

void loop() {
  // put your main code here, to run repeatedly:

}
```

Fig. 31, basic structure of Arduino program.

The basic structure is very simple and consists of two main parts as depicted in Fig. 31 - void setup and void loop. The setup code is put in to the void setup which runs once in the beginning of the program as serial communication and its frequency, launching of libraries, setting digital pins, etc. The program itself is written in the void loop, which will run repeatedly. A basic program – Blink is shown in Fig. 32. It turns on and off an LED with a period of one second. In order to use libraries, it is necessary to define them before void setup e.g. #include<SD.h>, which consists of the code to cooperate with the SD card.



```
Blink §
// the setup function runs once when you press reset or power the board
void setup() {
  // initialize digital pin 13 as an output.
  pinMode(13, OUTPUT);
}

// the loop function runs over and over again forever
void loop() {
  digitalWrite(13, HIGH);   // turn the LED on (HIGH is the voltage level)
  delay(1000);              // wait for a second
  digitalWrite(13, LOW);    // turn the LED off by making the voltage LOW
  delay(1000);              // wait for a second
}
```

Fig. 32, Blink program.

4.2.3 Development of the programme

As we worked on the improving of the device during the time, also our program was being improved. There were two previous versions of the program that were used before the final version.

<u>The first version</u> – This version had only a basic code for displaying four main values (accelerations in axes x, y, z and SEMG) in serial monitor in the Arduino IDE without any sampling frequency or saving on a SD card.

<u>The second version</u> – It was programmed to make a string out of our four values plus it added time to this string using the SimpleTimer library and it saved the string of data into a CSV (comma separated values) file on a SD card. This file was created as the programme was launched. There were problems with this programme as it used only the delay command at the end of the void loop, which provided break in milliseconds before the next loop of the program. That led to a sampling frequency slow down. To explain what cause the sampling frequency slow down, we have to say that the program was written to open the file on the SD card and find the end of the file. Then wrote the measured values at the end of the file and closes the file. As the file got bigger and longer by consisting of more and more values it took longer to find the end of the file and that slowed down the sampling frequency by the time.

Before the final code was written two criterions were set. At first we needed to keep the sampling frequency during the measuring the same, because we wanted to use FFT (Fast Fourier Transform) for evaluating the measured data by a frequency analysis. To do that it is necessary to have constant sampling frequency. The second criterion was to make a new file every hour of measuring, because we didn’t want to have all the data from all day measuring in one file in case that the file is corrupted and all day of measuring would be lost.

4.2.4 Final programme

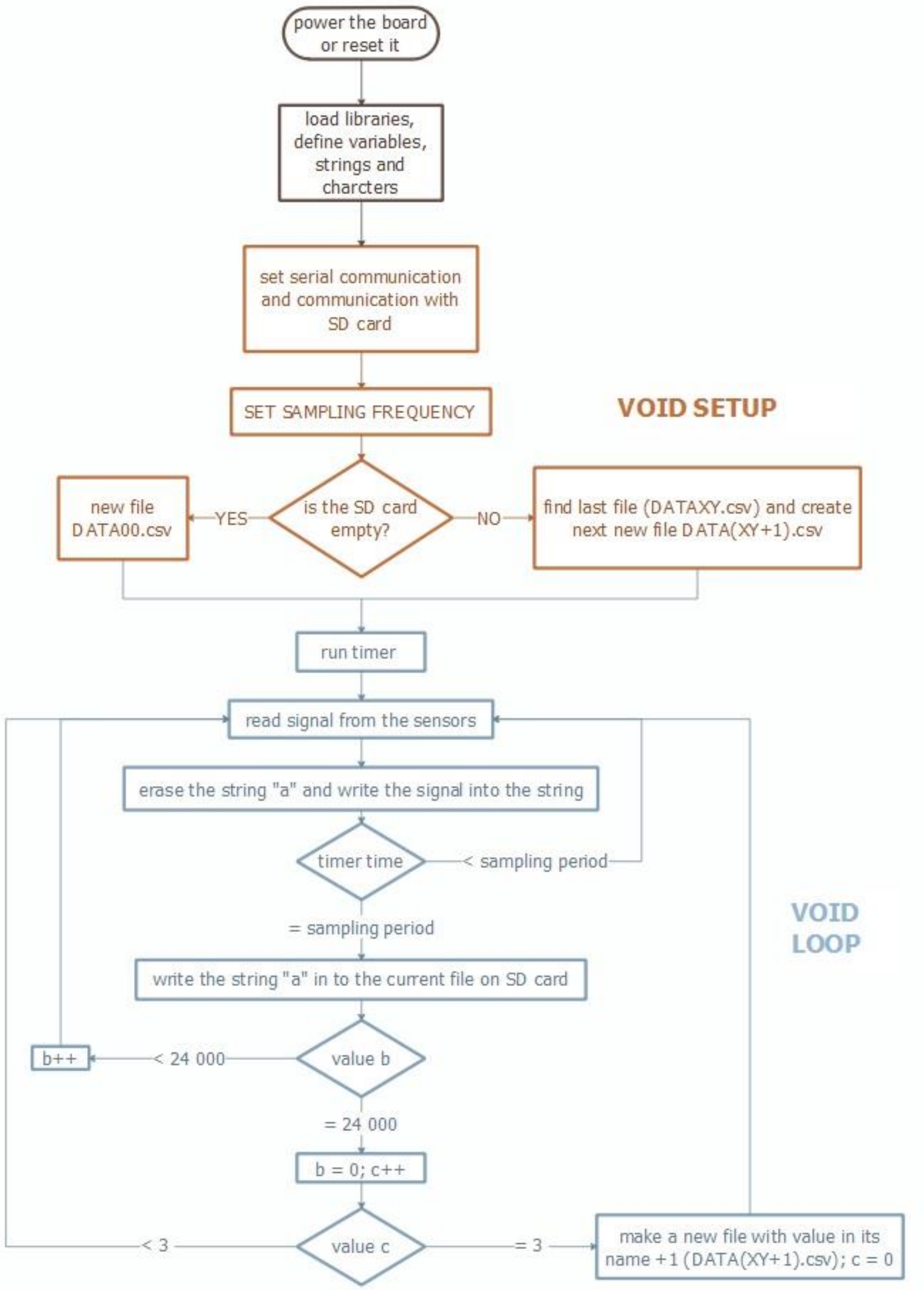


Fig. 33, diagram of algorithm of the final programme.

The programme uses three libraries to work properly. The SD library, which allows for reading from and writing to a SD card. The SPI (Serial Peripheral Interface) library, which allows to communicate with SPI devices (such as shift registers, sensors and SD cards) with the Arduino as the master device and the Simpletimer library to keep constant sampling frequency.

```
1. #include <SD.h>
2. #include <SPI.h>
3. #include <SimpleTimer.h>
```

The Simpletimer library provides time which runs from the beginning of the program. Then the sampling frequency can be set. The sampling frequency is used to call a function every e.g. 50 milliseconds while the program is running as shown in row 4 of the code.

```
4. timer.setInterval(50, RepeatTask);
```

The sampling frequency is set at 20 Hz. This may seem not enough since we measure surface EMG, but according to our way of evaluating of the SEMG it is enough. Please refer to Chapter 4.4 for more information regarding the evaluation. The sampling frequency is set according to the Nyquist-Shannon sampling theorem (1). It states that there is always a loss of information during a sampling of the analog signal (continuous-time signal), but to minimize this loss the sampling frequency has to be at least twice the highest frequency of the harmonic part of the analog signal [42].

$$f_s > 2\, f_{max} \tag{1}$$

where f_s is the sampling frequency and f_{max} is the maximal harmonic frequency

We are interested about movements with harmonic frequencies around 1Hz or 2Hz, thus sampling frequency of 20Hz is sufficient.

The variables, characters, constant pins, serial communication and SD communication have to be set.

```
5. int c; int b; byte i; String a;
6. const int chipSelect = 4;
7. Serial.begin(9600);
8.  pinMode(10, OUTPUT);
9.  SD.begin(chipSelect);
```

In the beginning, the programme searches through the files on the SD card. If the SD card is empty, the program creates the DATA00.csv file. If there are already some files, the program finds the last one and creates a file with a subsequent number. It also allows more day measuring without any support.

```
10. char fileName[] = "DATA00.CSV";
11. for (i = 1; i <= 99; i++) {
12.    if (SD.exists(fileName)){
13.      fileName[4] = i/10 + '0';
14.      fileName[5] = i%10 + '0'; } }
```

The program itself starts the timer in the void loop (row 10). Then it erases string *a* (row 11). After that it saves the four values from sensors to the string *a* and makes commas between the values (rows 12 – 15)

```
15.  timer.run();
16.  a = "";
17.  for (int analogPin = 0; analogPin < 4; analogPin++) {
18.       int sensor = analogRead(analogPin);
19.       a += String(sensor);
20.        if (analogPin < 3) {a += ",";}}
```

The RepeatTask function is the writing function that is called every 50ms. It opens the current file and writes the string *a* and closes the file. Every time this happens, +1 is added into the variable *b*.

```
21. void RepeatTask()
22.  {b++;
23.  File dataFile = SD.open(fileName, FILE_WRITE);
24.  dataFile.println(a);
25.  dataFile.close();}
```

When *b* reaches the value of 24,000 the programme adds +1 into the variable *c* and sets b = 0 (rows 21 – 23). And when *c* reaches 3 – that means 72,000 cycles of the writing function which means one hour of measuring, then the programme goes through the names of the files on the SD card and creates the subsequent one. Then it sets *c* = 0 (rows 24 -29) and it starts from the beginning again.

```
26. if(b>=24000){
27.  c++;
28.  b=0;}
29. if(c>=3){
30.   for (i = 1; i <= 99; i++){
31.    if (SD.exists(fileName)){
32.      fileName[4] = i/10 + '0';
33.      fileName[5] = i%10 + '0';}}
34. c=0;}
```

It needs to be done this way with the variables as the number of 72,000 needs to be reached, but the maximum value of a variable int (integer) is 32,767 as it is a 16-bit value.

The final version of the programme is in the electronic appendix.

4.3 Verification of sensors

4.3.1 Verification of accelerometer

The aim of the verification is to prove that the custom developed device consisting of the ADXL 335 accelerometer (Analog Devices Inc., USA) works properly. Verification has been performed using the Xsens sensor (Xsens Technologies B.V., The Netherlands) as a reference. Xsens makes inertial motion tracking systems. It costs hundreds of thousands of Czech crowns and it measures accelerations with precision of $10^{-6}ms^{-2}$. Compared to our system, it is much more expensive, but also bigger as shown in Fig. 34 and it is also a laboratory system.

Fig. 34, size comparison of Xsens and our device based on Arduino board.

A MoCap system Xbus kit was used. It is a lightweight device which uses datalogger (Xbus Master) and units (called MTx units) for the segment orientation measurement. The MTx unit is an accurate 3DoF tracker (dimensions 38mm × 53mm × 21mm, weight 30g), providing a drift-free 3-D orientation as well as kinematic data: 3-D acceleration and 3-D turn rate (rate gyro). Therefore, an Xbus kit can also be seen as a representative of gyroscope systems to measure orientation. The MTx unit was calibrated before the verification measurement. The MTx unit was placed on the subject's shoulder [43].

The MTx unit is bigger as it uses triplet of each sensors (accelerometer, gyros) and that also allows the high accuracy of measurement.

MTx unit specifications are in Tab 8.

Interfaces	RS-232, RS-485, RS 422 (max 921k6 bps) and USB (ext. converter)
Operating Voltage	4,5V – 30V
Power Consumption	350mW
Operating temperature range	-40 ºC to +85ºC
Gyro Bias Stability	20deg/h
Timing accuracy	10ppm
Dimensions	38mm x 53mm x 21mm
Static accuracy (roll/pitch)	< 0.5deg
Static accuracy (heading)	< 1deg
Dynamic accuracy	2deg RMS
Angular resolution	0.05deg
Weight	30g

Tab. 8, MTx unit specifications [44].

Our accelerometer was attached to Xsens sensor and Xsens sensor was attached to a shoulder of the testing subject. Then the subject repeated three motions of abduction and adduction. Coordinate systems of both sensors were leveled. The sampling frequency of both systems were set at 50Hz.

4.3.2 Verification of SEMG sensor

The aim of the verification procedure is to prove that the custom developed device consisting of a Muscle sensor v3 (Advancer Technologies, LLC, USA) works properly. Verification has been performed using MA300 (Motion Lab Systems, Inc, USA) as reference.

The MA-300 EMG system consists of two units a backpack and desktop unit and a single thin coaxial connecting cable. The subject carries the backpack, attached to a belt, with EMG pre-amplifiers and 8 foot switches. EMG, foot switch and other signals are digitized and processed within the back-back and transmitted as digital information to the desk top unit over the cable. This single core cable is eighteen meters long and weighs less than 160 grams. It does not encumber the subject in any way. The standard MA-300 system does not use radio or infrared telemetry and can be used in almost any environment with any of the restrictions of comparable telemetry systems. By digitizing all signals at the subject, the MA-300 guarantees a clean signal without any degradation from the transmission of analog signals [45].

MA300 characteristics are presented in Tab 9.

Number of EMG channels	16
EMG signal output level	±5V Full Scale
Standard EMG Bandwidth	20Hz to 2,000Hz at -3dB.
Built in Low Pass Filter	-3dB at 350, 500, 750, 1000, 1250, 1500, 1750 and 2000Hz. Set by user accessible switch when power is applied to the backpack
Number of foot switches	8 switches supported, independent of EMG channels.
Signal connection	18m using standard RG-174 coaxial cable (4.8mm diameter, total weight 160g). Cable length can be up to 36.5m
Electrical Isolation	1500V DC Applied part
EMG pre-amplifier input noise	Less than 2µV RMS nominal, C.M.R.R. >100dB at 40Hz.
AC input rating	100V – 240V, 50VA, 50/60Hz

Tab. 9, MA300 characteristics [45].

The Surface EMG pre-amplifiers supplied by the MA-300 are single, miniature, modular, surface-mount pre-amplifiers with a built-in instrumentation amplifier using a dual differential input configuration, full static, muscle stimulator protection and includes a Radio Frequency Interference filter. It uses medical grade stainless steel dry button skin contacts to sense the EMG signal at the skin surface and also provide a ground reference. A single shielded high flex miniature cable connects to a unique flexible, six pin Harwin connector [45].

EMG Pre-amplifier Characteristics are presented in Tab. 10.

Input Impedance	>100,000MΩ
Input Configuration	Dual Differential
Input Protection	± 40V DC
Equivalent Input Noise	less than 2FV RMS nominal.
C.M.R.R.	>100dB min at 40Hz
Bandwidth (-3 dB)	10Hz to 2,000Hz
Standard Gain	20 (± 2%)
Body size	38mm x 19mm x 9mm
Weight	20g
Connector	Harwin, 6 pin female

Tab. 10, EMG Pre-amplifier characteristiscs, Source [45]

Fig. 35, MA300 Desktop unit and EMG Pre-amplifies [45].

EMG Pre-amplifie from MA300 was placed on the biceps brachii (Fig. 36) of the testing subject. Electrodes from our SEMG sensor were placed on the same muscle of the testing subject right next to the electrodes from MA300. The sampling frequency of the MA300 was set at 1000Hz and the sampling frequency of Arduino was 50Hz, thus this should be another source of deviation of both signals. The testing subject did multiple flexions and extensions in elbow to activate the measured muscle and also maximal isometric contraction was performed. All these motions were did in 60s.

The sequence of movements was as following:

1) maximal isometric contraction for circa 10s
2) 4 flexions and extensions of forearm
3) 4 flexions and extensions of forearm with increasing effort
4) 3 fast flexion and extensions of forearm

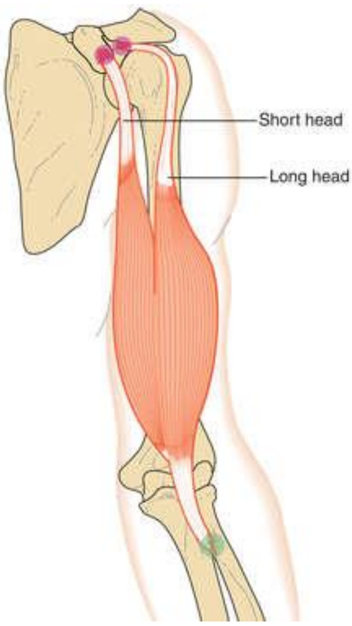


Fig. 36, Biceps brachii on left arm of human body [46].

4.4 Data evaluation

Our goal is to find a method how to separate groups of activities which have something in common to define executed activities during daily living. The files with measured data contain one hour of measuring each. We use a one minute window for evaluating the data. That means we separated the record into one minute records (Fig. 37) and find a parameter for each minute of the record, which defines the executed activity in that minute.

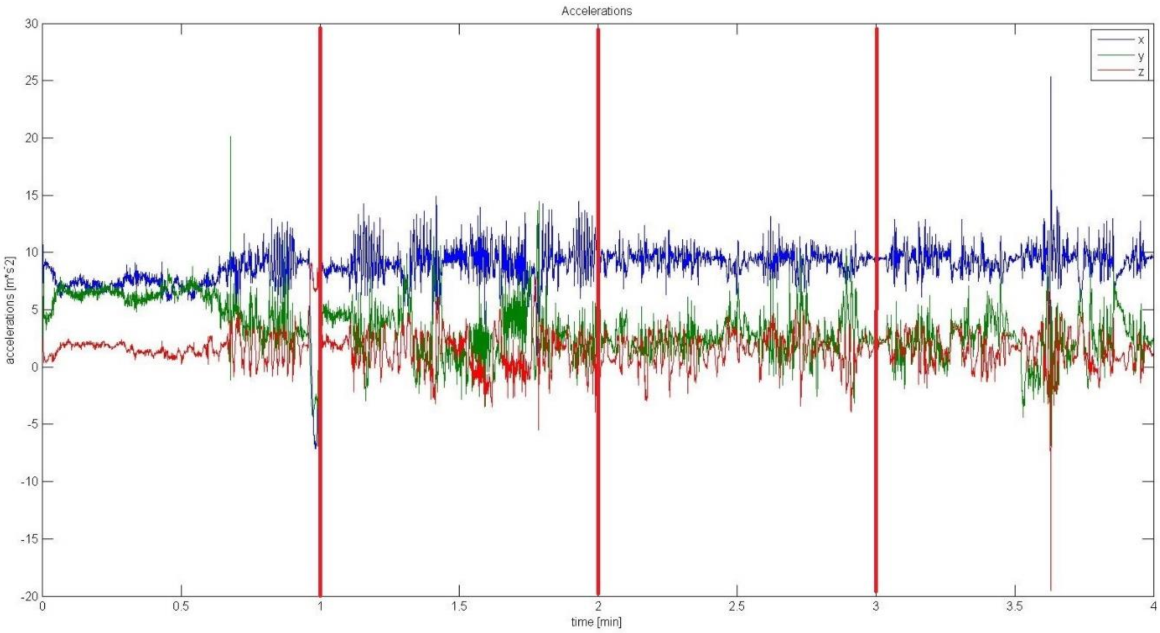


Fig. 37, creating one minute windows of the record.

4.4.1 Acceleration evaluation

More mathematical methods are used to find the parameters for acceleration. These are:

1) Mean (Arithmetical average)
2) Standard deviation
3) Fast Fourier Transform for the frequency analysis of the signal

4.4.1.1 Mean and standard deviation

For a random variable vector *A* made up of *N* scalar observations, the mean is defined as (2).

$$\mu = \frac{1}{N}\sum_{i=1}^{N} A_i \tag{2}$$

For a random variable vector *A* made up of *N* scalar observations, the standard deviation is defined as (3). Where μ is the mean from (2)

$$S = \sqrt{\frac{1}{N-1}\sum_{i=1}^{N} |A_i - \mu|^2} \tag{3}$$

4.4.1.2 Fast Fourier Transform (FFT)

The Fourier transform (4) decomposes a function of time into the frequencies which it consists. It is similar to how a musical chord can be expressed as the amplitude or loudness of its constituent notes. The Fourier transform of a function of time itself is a complex-valued function of frequency, whose absolute value represents the amount of that frequency present in the original function, and whose complex argument is the phase offset of the basic sinusoid in that frequency. The Fourier transform is called the frequency domain representation of the original signal [47].

$$\hat{f}(\xi) = \int_{-\infty}^{\infty} f(x) e^{-2\pi i x \xi} dx \tag{4}$$

For any real number ξ. The independent variable *x* represents time [s] and the transform variable ξ represents frequency [Hz].

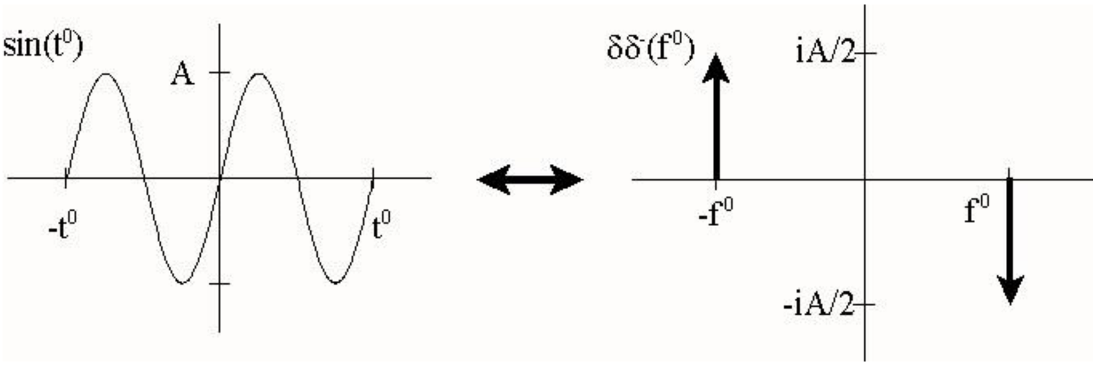


Fig. 38, Fourier transform [47]

FFT is an algorithm which computes the discrete Fourier transform (DFT) (5) of a sequence. An FFT rapidly computes such transformations by factorizing the DFT matrix into a product of sparse (mostly zero) factors. As a result, it manages to reduce the complexity of computing the DFT from O (n^2), which arises if one simply applies the definition of DFT to O (n log n), where n is the data size [48].

$$X_k \stackrel{\text{def}}{=} \sum_{n=0}^{N-1} x_n e^{-2\pi ikn/N}, k \in Z \tag{5}$$

Due to periodicity, the customary domain of the actually computed k is [0, N − 1]. That is always the case when the DFT is implemented via the Fast Fourier transform algorithm.

Using Euler's formula (6), the DFT formula can be converted to the trigonometric form (7) [47]. From this trigonometric form follows, that every signal can be divided into sinuses and cosines.

$$e^{ix} = \cos x + i \sin x \tag{6}$$

$$X_k = \sum_{n=0}^{N-1} x_n \left(\cos\left(-2\pi k \frac{n}{N}\right) + i \sin\left(-2\pi k \frac{n}{N}\right)\right), k \in Z \tag{7}$$

Where N is number of time samples, n is current sample which is considered (0,…,N-1), x_n is the value of the signal at time n, k is the current frequency which is considered (0Hz up to (N-1)Hz) and X_k is the amount of frequency k in the signal.

Frequency analysis of the signal recorded during running (Fig. 39) is in Fig. 40.

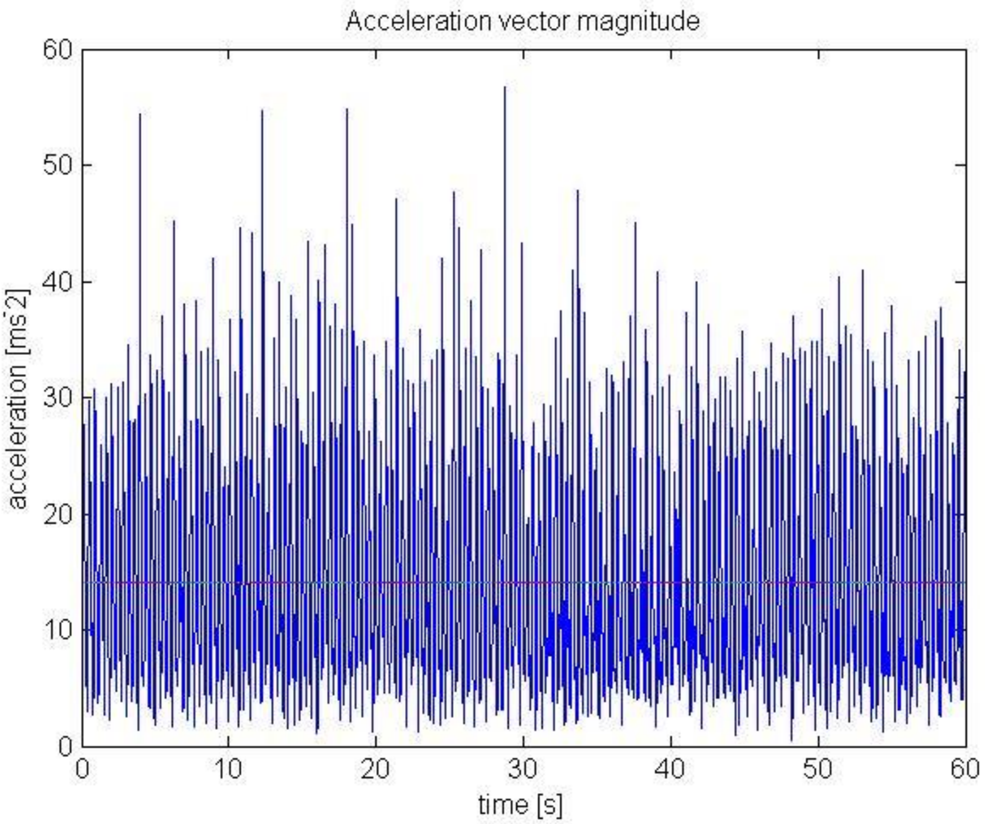


Fig. 39, sample of signal recorded during runing, 1 min of recording

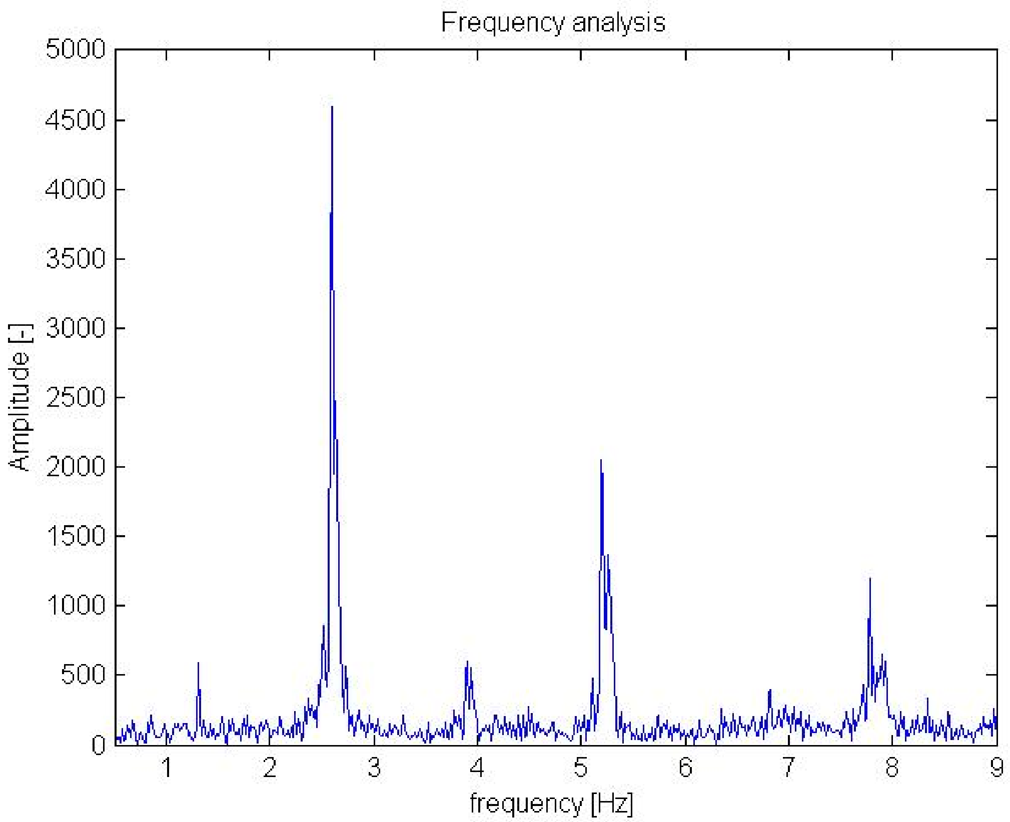


Fig. 40, frequency analysis of signal in Fig. 39.

4.4.2 SEMG evaluation

Sampling frequencies of the EMG signal are usually in range 100Hz – 1000Hz. The sampling frequency of the device is 20Hz, thus thresholding of the signal is the only reasonable evaluation. It is reasonable and also perfectly suitable for our application.

The SEMG sensor is not able to set the base line at 0, thus the first operation is to subtract the minimum value of the signal from each value of the signal (Fig. 41). The second thing is to define the thresholds and determine the muscle activity as low, medium and high.

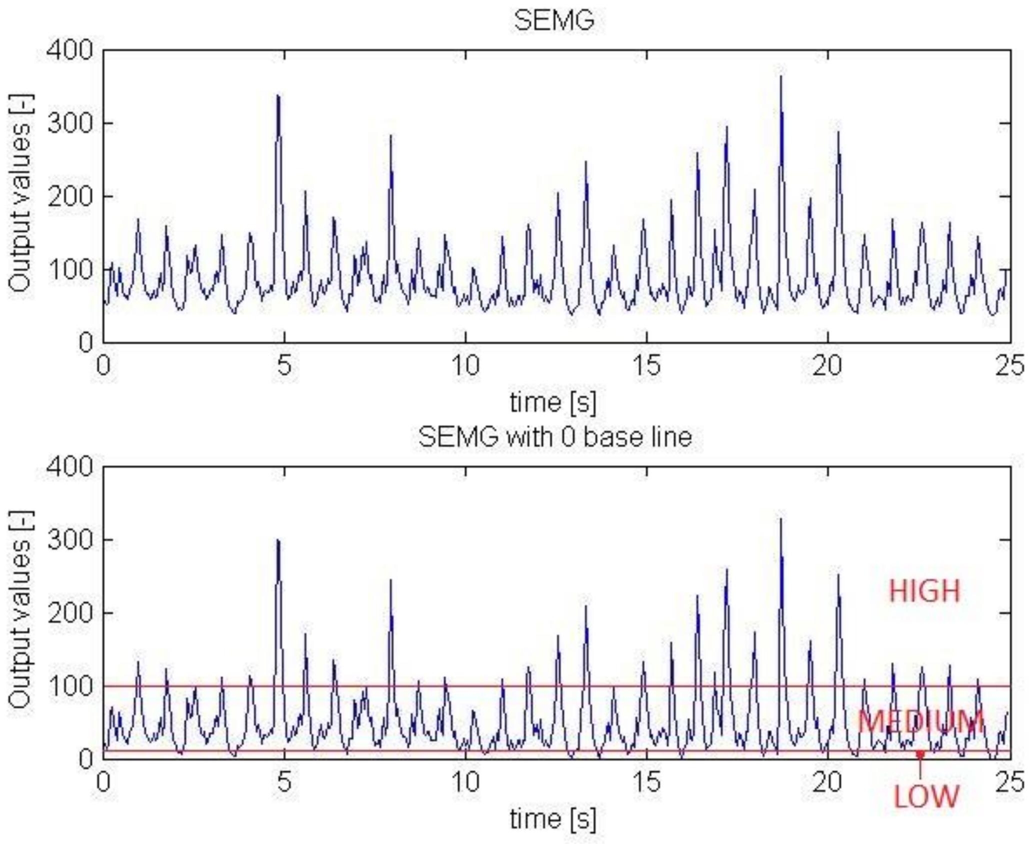


Fig. 41, SEMG evaluating (thresholds are only tentative).

4.4.3 Tested activities and subjects

To test the suitability of the device, basic activities have been chosen. The subjects were asked to perform all activities with self-chosen comfortable speed. These activities should be recognized by the decision tree from the measured data. The basic activities are:

1) Walk with or without a load
2) Run
3) Sit
4) Lie
5) Static manual activities (e.g. writing, working on computer)
6) Dynamic manual activities (e.g. washing dishes, cleaning)

The tested subjects are presented in Tab. 11.

Subject	Gender	Age	Tracked shoulder
1	Male	25	Right
2	Male	25	Right
3	Male	26	Right
4	Female	26	Right
5	Female	27	Left
6	Female	26	Right

Tab. 11, tested subjects.

The decision tree algorithm will be developed after parameters are found for each of the chosen activity for each of the subjects (1-5). The algorithm will be then tested on subject 6.

The parameters are:

1) mean of the acceleration vector magnitude
2) standard deviation of the acceleration vector magnitude
3) mean of the SEMG signal
4) mean of the angle of the acceleration vector magnitude in x-y plane
5) Frequency with maximal amplitude
6) Amplitude of the frequency (5)

The subjects in the tested group are almost in the same age, thus the decision tree with parameters based on that group should not work on subjects of different age.

5 Results

5.1 Sensor verification

5.1.1 Accelerometer verification

Acceleration from both sensors in axes x, y, z are shown in Fig. 42, 43, 44. Small differences in axes y and z were caused by slight disorder of the axes rotation and by the conversion of values from the A/D converter to accelerations in ms^{-2}. The disorder can be seen in the beginning of the measuring as the base lines are in different levels.

This meassuring verified that our small, cheap sensor ADXL335 gives trustworthy accelerations. It is shown in the Figs that our sensor gives similar accelerations as costly Xsens system and it is fully useable for our application.

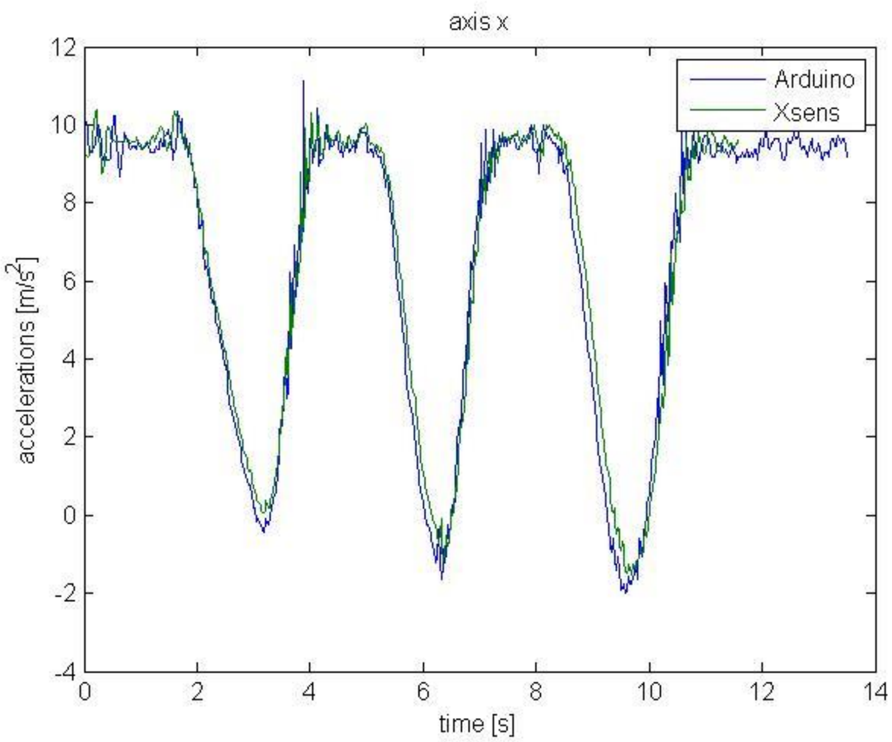


Fig. 42, accelerations in axis x of both sensors during abduction and adduction

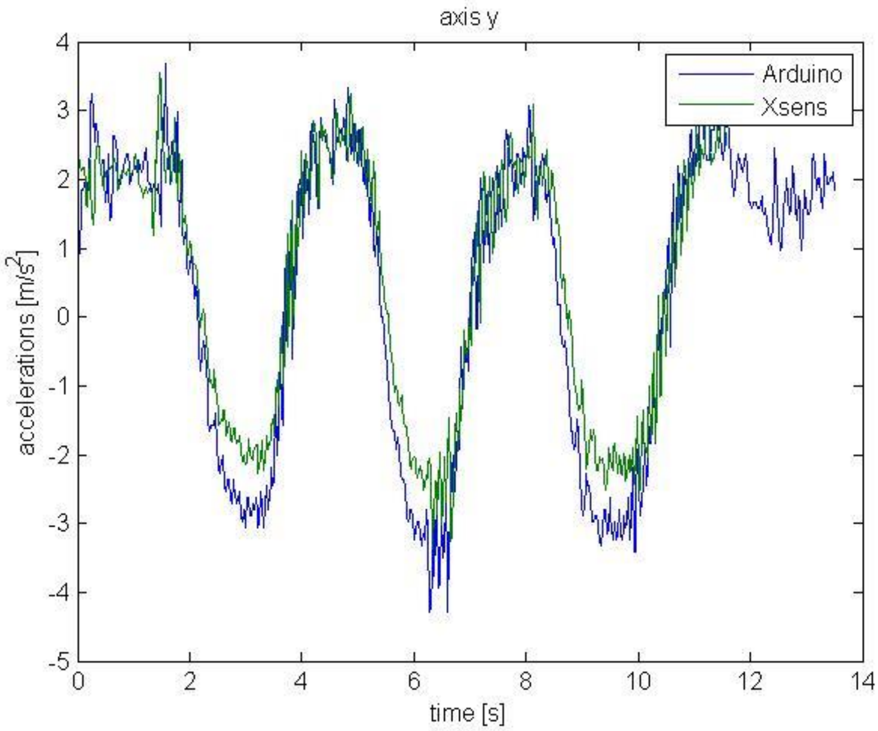


Fig. 43, accelerations in axis y of both sensors during abduction and adduction.

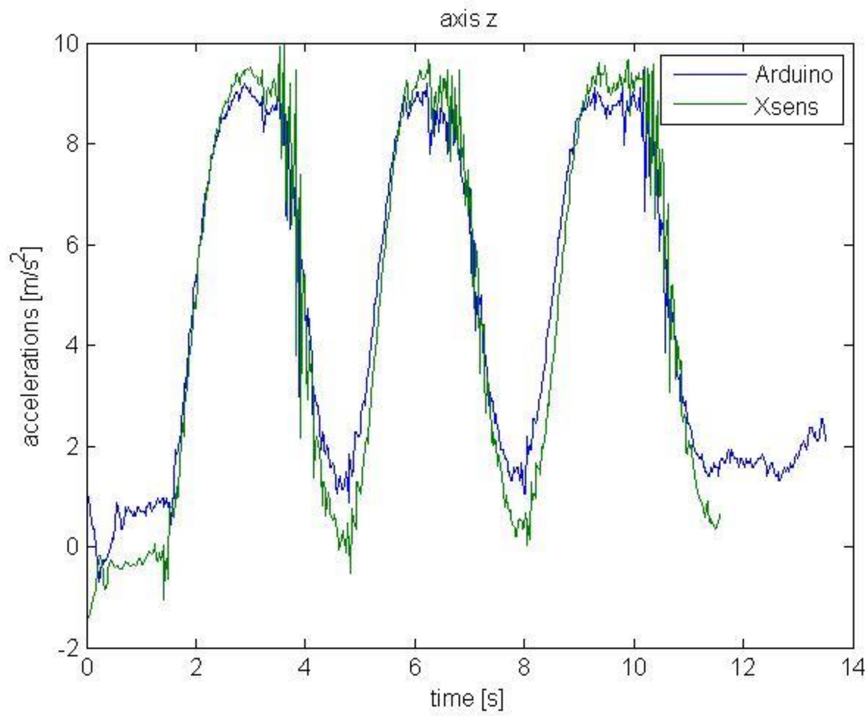


Fig. 44, accelerations in axis z of both sensors during abduction and adduction.

5.1.2 Surface EMG sensor verification

In Fig. 45 there are presented output data from the MA300. It is a raw EMG signal. In Fig. 46 there are depicted output data from the Muscle sensor v3 connected on Arduino. This signal is already rectified and smoothed by the sensor.

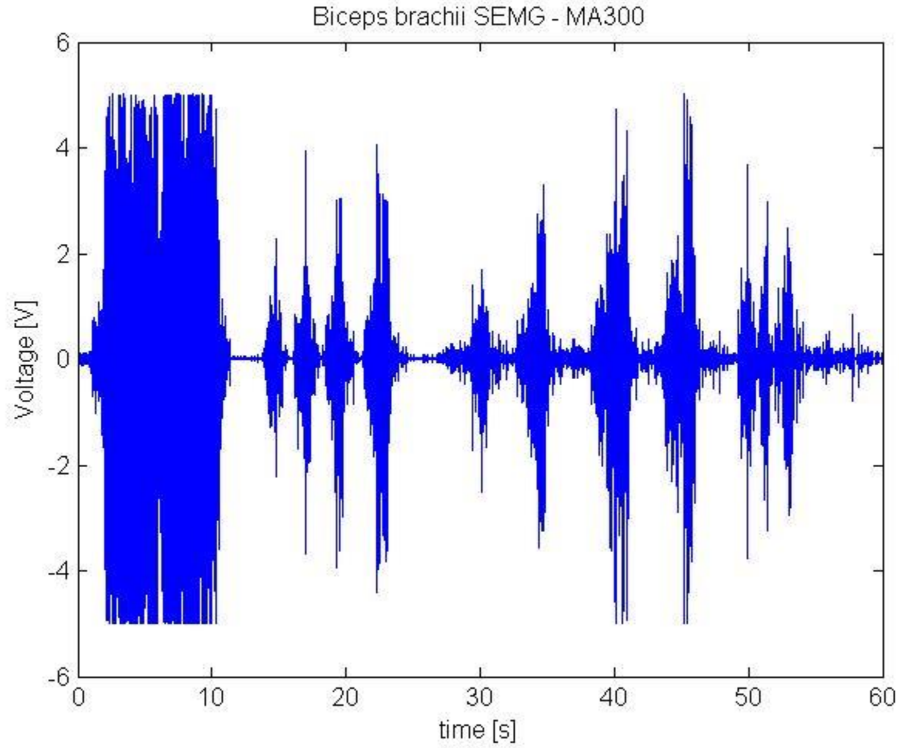


Fig. 45, SEMG of Biceps brachii measured by MA300.

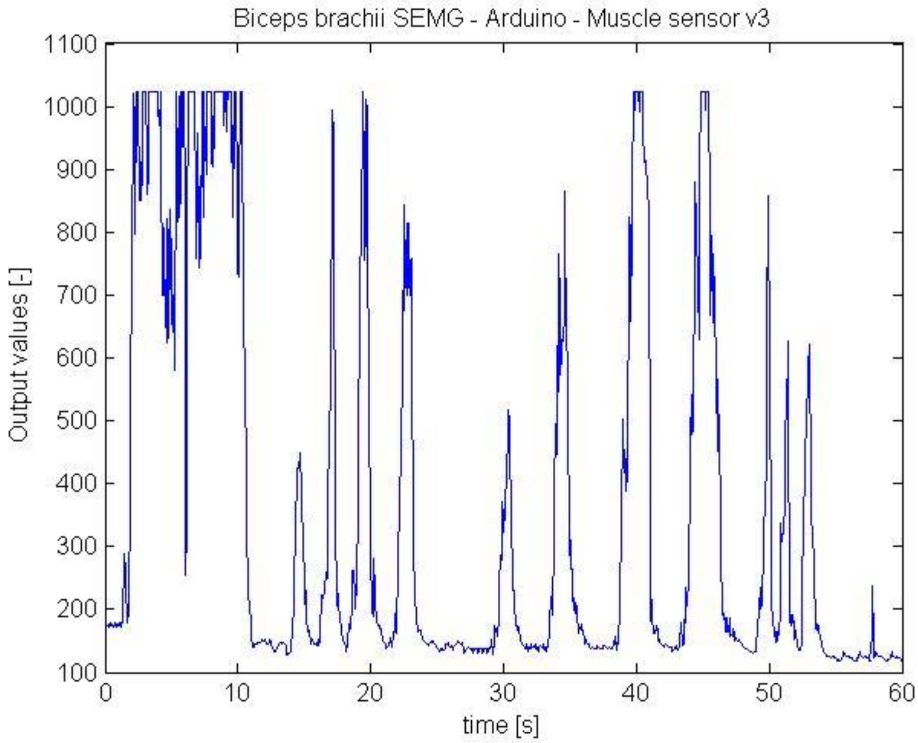


Fig. 46, SEMG of Biceps brachii measured by Muscle sensor v3.

Arduino gives the maximum value during the maximal isometric contraction as is presented in Fig 46. This maximum value of 1,023 occurs also during flection with maximum effort. But these are not activities performed during an ordinary day.

In Fig. 47 there are output data from both sensors together. It shows that our cheap SEMG sensor really measures the surface EMG signal. It also proves that the value of our EMG is trustworthy for thresholding, which is used for evaluation. The Arduino output signal was converted to volts for the purpose of comparison with the MA300 output signal. Also the signal from MA300 was rectified for comparison with the signal from Arduino.

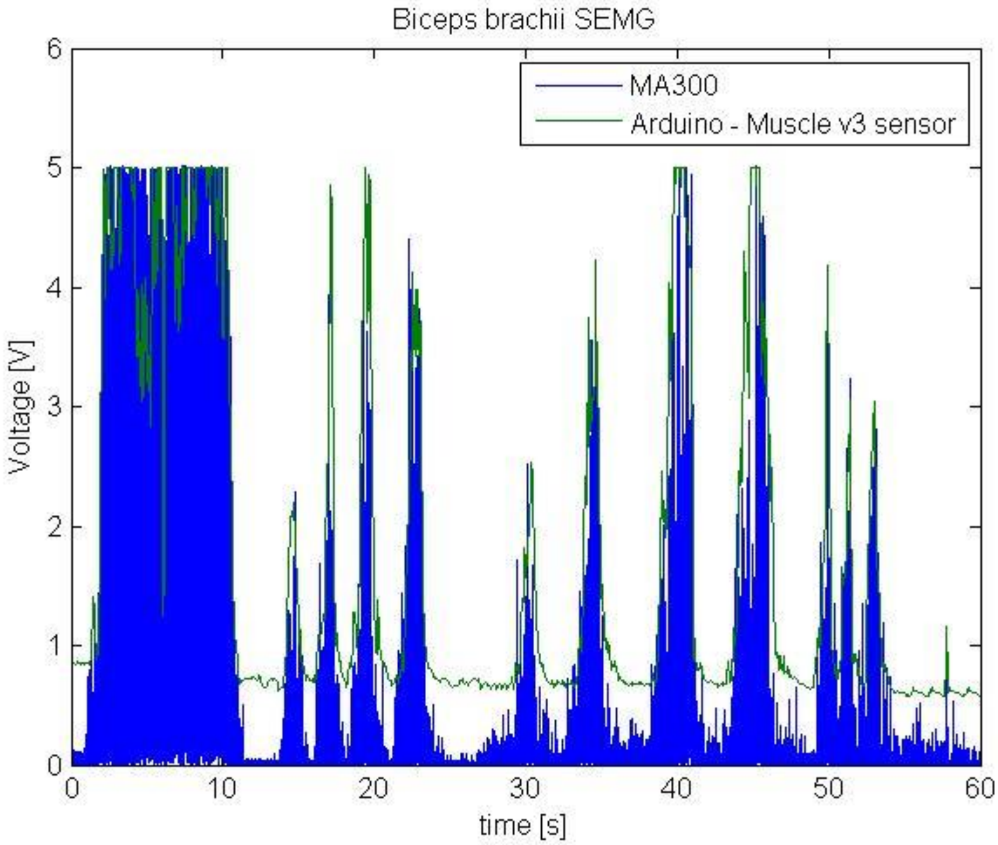


Fig. 47, comparison between both SEMG sensors.

The Arduino sensor is not able to set the base line to 0 as seen in Fig. 47. That is caused by the placement of the reference electrode on a bony part of the body which is not close enough to the measured muscle.

5.2 Data evaluation

5.2.1 Parameters and the decision tree

Subjects 1-5 from Tab. 11 performed activities from chapter 4.4.3 1) – 6). The final parameters defining the decision tree are based on parameters measured on these subjects. Selected parameters and graphs are in Appendix 1. The decision tree shown in Fig. 48 was developed after parameter evaluation.

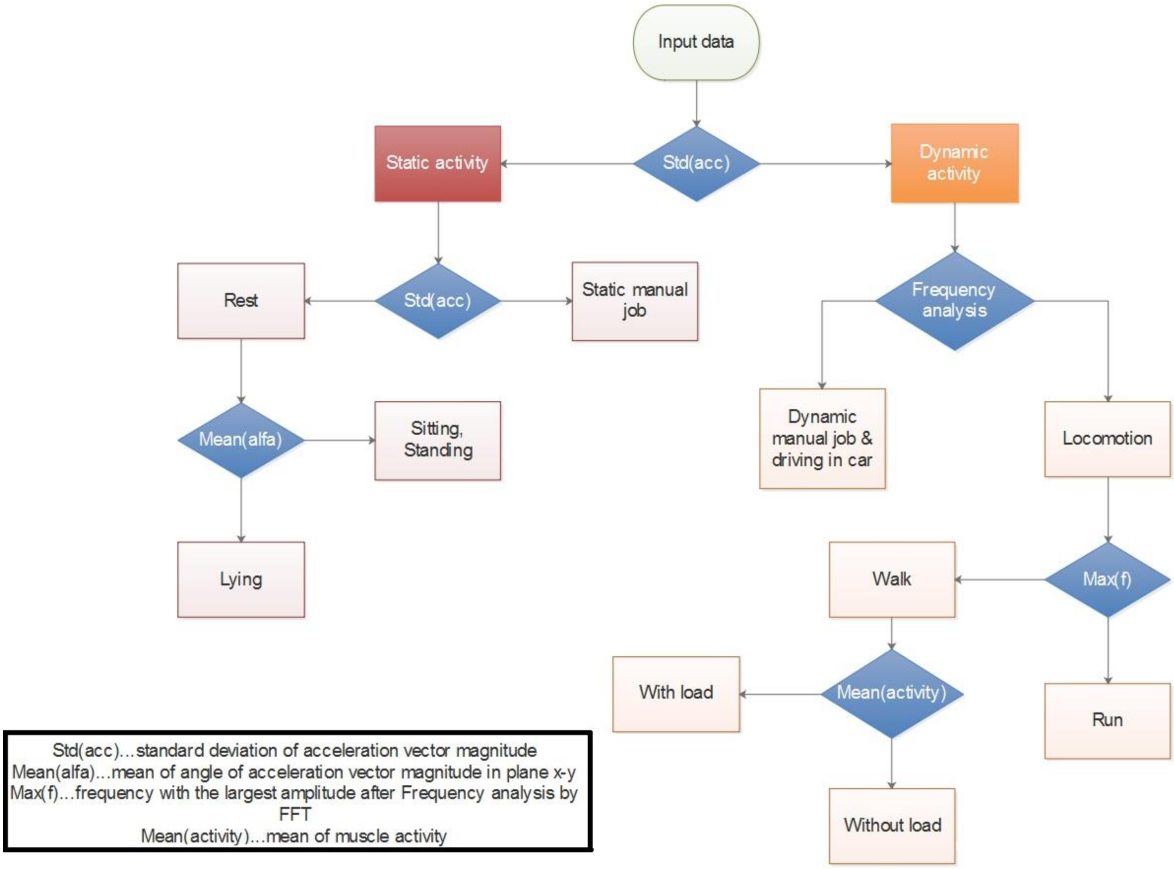


Fig. 48, the decision tree.

The decision tree algorithm is written in MatLab (The MathWorks, Inc., USA) and is provided in the electronic appendix. The algorithm starts with uploading of measured data. These are uploaded automatically directly from the device after determining the number of files. This part of program is shown in rows 1 – 7.

```
1.  clc, clear all;
2.  numfiles = 10;
3.  mydata = cell(1, numfiles);
4.  for k = 1:numfiles
5.    myfilename = sprintf('DATA%02d.xlsx',(k-1));
6.    mydata{k} = importdata(myfilename);
7.  end
```

When all measured data is uploaded into a cell in MatLab, the programme goes through each file (cell) and saves acceleration and SEMG into variables X, Y, Z and A. Then it creates one minute windows of the measured data and calculates parameters for each window.

```
8.  for k = 1:numfiles
9.      X = (mydata{1,k}(:,1)-504)/114*9.81;
10.     Y = (mydata{1,k}(:,2)- 512)/110*9.81;
11.     Z = (mydata{1,k}(:,3)- 489)/111*9.81;
12.     A = mydata{1,k}(:,4) – min(mydata[1,k}(:,4));
13.     N = (1:1200:length(X));
14.     for i = 1:(length(N)-1)
15.         B = (N(i):(N(i+1)-1));
16.         x = X(B);
17.         y = Y(B);
18.         z = Z(B);
19.         a = A(B);
20.     acc = sqrt(x.^2+y.^2+z.^2);
21.     alfa = atand(y./x);
22.     Sacc = abs(fft(acc));
23.     lenHalfFftacc = round(length(Sacc)/2 + 1);
24.     Sacc = Sacc(1:lenHalfFftacc);
25.     facc = 20/2*linspace(0, 1, lenHalfFftacc);
26.     [row,col] = find(Sacc == max((facc>0.5).*Sacc'));
27.     faccmax(i,k) = facc(row);
28.     Saccmax(i,k) = Sacc(row);
29.     meanacc(i,k) = mean(acc);
30.     stdacc(i,k) = std(acc);
31.     meana(i,k) = mean(a);
32.     meanalfa(i,k) = mean(alfa);
33.     end
34. end
35. save parameters.mat faccmax Saccmax meanacc stdacc meana
    meanalfa k
```

Acc is calculated in the row 20, which is an acceleration vector magnitude. In the row 21, alfa is calculated, which is the angle of acc in the x-y plane. In the rows 22 – 25, the Fast Fourier transform is performed for the frequency analysis. Frequency with maximum amplitude is found in the row 26. We are looking for frequencies over 0.5Hz only, because we are interested in motions around (1 – 3)Hz. Afterwards, all the parameters are calculated and saved (rows 27 – 32).

Performed activities are then evaluated from the parameters. At first the activities are divided into static and dynamic activities according to the decision tree in Fig. 48. If the standard deviation of acc (std(acc)) is bigger than $0.5ms^{-2}$, then it is a dynamic activity. If the std(acc) is smaller or equal to $0.5ms^{-2}$, then it is a static activity.

Static activities:

When the std(acc) is smaller or equal to $0.13ms^{-2}$, the subject is in a rest (standstill). The standstill is divided into a sitting/standing or a lying position. That is recognized by the mean of the angle alfa. When the angle is in the range of $(-45 – 25)^0$ then the subject is sitting or standing. If the angle is in the range of $(-90 – -45)^0$ or $(25 – 90)^0$ then it is lying. When the std(acc) is bigger than $0.13ms^{-2}$, then the subject is performing a static manual job. Static manual jobs are e.g. writing either on computer or by hand.

Dynamic activities:

The main signal evaluating the dynamic activities is the Frequency analysis by FFT. When there is a maximum amplitude over 750 and its frequency in the range of 1.6Hz – 2Hz, then the subject is walking. If the maximum amplitude is over 2 500 and its frequency is in the range of 2Hz – 3Hz, then the subject is running. In Fig. 49, frequency analysis of running and walking is presented as an example of graphs of these activities.

If the mean of the muscle activity during walking is bigger than 40, then it means that the subject is carrying some load in his hand.

If there is no dominant frequency, it is a dynamic manual job like e.g. cleaning dishes. Or the subject can be in a car or a public transport.

The muscle activity is evaluated by thresholding and classified as small, medium and high. The thresholds are set at the output value of 40 between the low and the medium and at the value of 200 between the medium and the high as shown in Fig. 50.

All the static activities and their parameters are shown it Tab. 12. All the dynamic activities and their parameters are shown in Tab. 13.

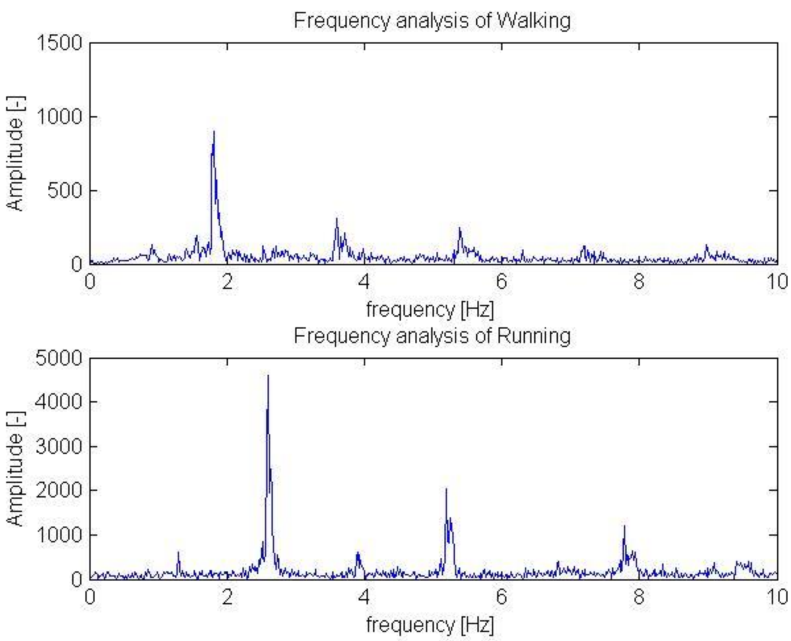


Fig. 49, frequency analysis of walking and running

Fig. 50, muscle activity axis.

Static activities = std(acc) ≤ 0.5ms^{-2}		
Activity	Parameter	Value or Range
Static manual jobs	Std(acc)	> 0.13ms^{-2}
Rest	Std(acc)	≤ 0.13 ms^{-2}
Sitting/Standing (Rest)	Mean(alfa)	$(-45 – 25)^0$
Lying (Rest)	Mean(alfa)	$(-90 – -35)^0$ & $(25 – 90)^0$

Tab. 12, static activities and their parameters.

Dynamic activities = std(acc) > 0.5ms^{-2}		
Activity (Superior activity)	Parameter	Value or Range
Dynamic manual job	Frequency analysis	without dominant frequency
Locomotion	Frequency analysis	with dominant freq.
Walk (Locomotion)	f(max) & S(max)	(1.6 – 2⟩Hz & > 750
Run (Locomotion)	f(max) & S(max)	(2 – 3)Hz & > 2500
Walk with load (Walk)	mean(a)	> 40
Walk without load (Walk)	mean(a)	< 40

Tab. 13, dynamic activities and their paramaters.

5.2.2 Experimental verification of the parameters

The decision tree with all of the parameters have been verified on the subject 6. She did the same activities (1) – 6) presented in chapter 4.4.3) as other subjects (1 – 5) did. She was asked to perform all activities with self-chosen comfortable speed. The results of the decision tree are presented in Tab. 14.

Activity	Time [min]	Recognized activity	Time [min]	Match
Walking without load	4	Walking	4	100%
Walking with load	3	Walking with load	3	100%
Running	2	Running	2	100%
Static manual job	6	Static manual job	6	100%
Dynamic manual job	7	Dynamic manual job	7	100%
Rest – Sitting	5	Sitting (Static manual job)	1 (4)	20%
Rest – Lying	3	Lying	3	100%
Driving in Car	47	Driving in Car (Static manual job)	26 (21)	55%

Tab. 14, recognized activites by the decision tree algorithm.

5.2.3 Evaluation of a measurement

Subject 1 was equipped by the device on an ordinary workday morning. He went to the university in Dejvice by public transport and then to Karlovo náměstí. He had consultation on both places. Then he went home by public transport and did some work there. The measurement lasted 241min.

The output is in the form of graphs. Four graphs were chosen as an example. All executed activities with their respective duration are shown in Tab. 15. The muscle activity is shown in Tab. 16.

In the first chart (Fig. 51), a comparison between a dynamic and a static activity performed during the measuring is shown. 131 minutes of dynamic activity and 110 minutes of static activity have been performed.

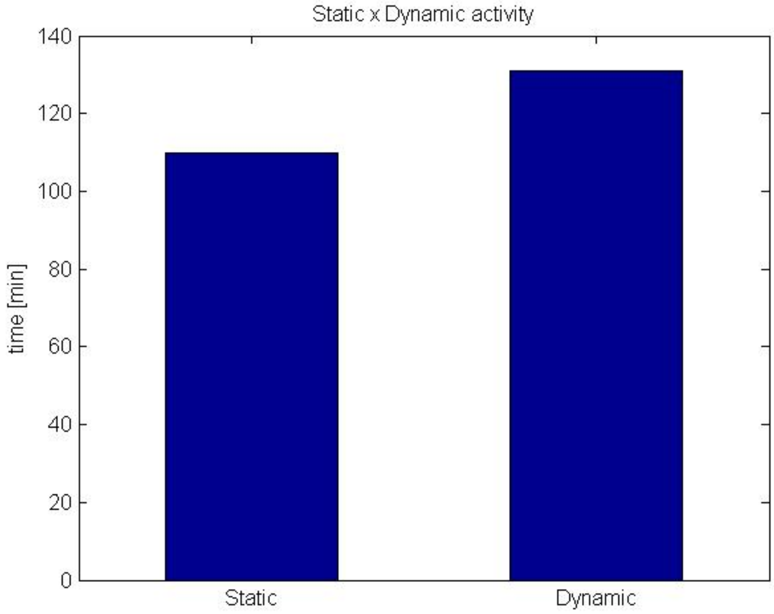


Fig. 51, static and dynamic activity.

In the second chart (Fig. 52), a comparison between walk with and without a load is depicted. He walked 41 minutes. From these forty-one minutes was thirty-nine minutes without a load and two minutes with a load.

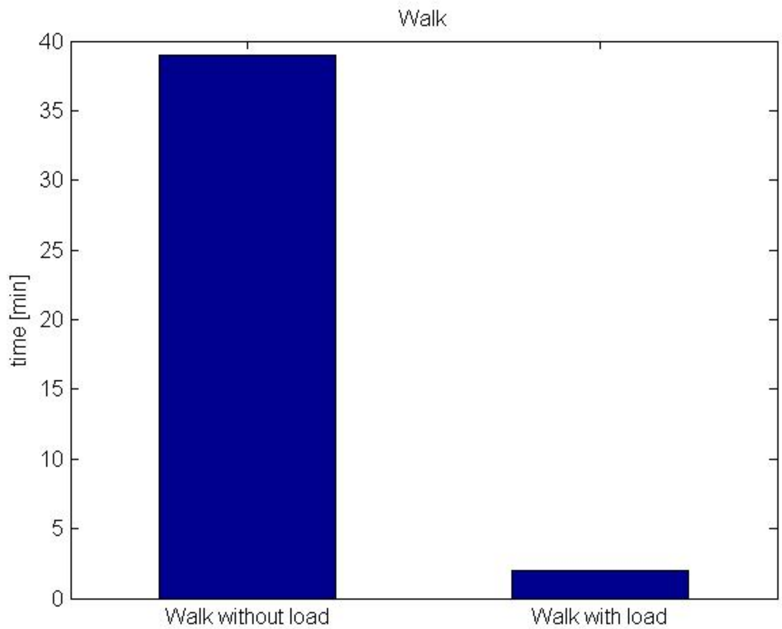


Fig. 52, walk with and without a load.

The third graph (Fig. 53) shows different activities performed during that day. Out of the recorded 241 minutes, there was 8min of sitting, 41min of walking, 90min of a dynamic manual job, 102min of a static manual job.

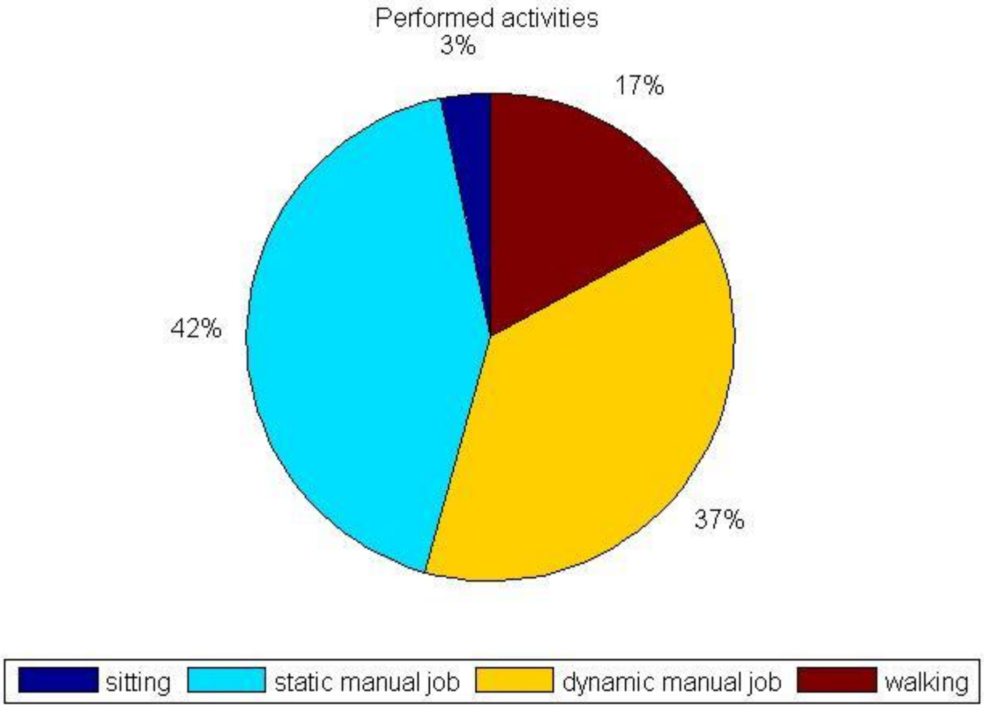


Fig. 53, performed activities.

Muscle activity classified as low, medium and high is shown in the following chart (Fig. 54). The subject had a low muscle activity for 230 minutes, medium for 11 minutes and the subject had not a high muscle activity over the recorded period.

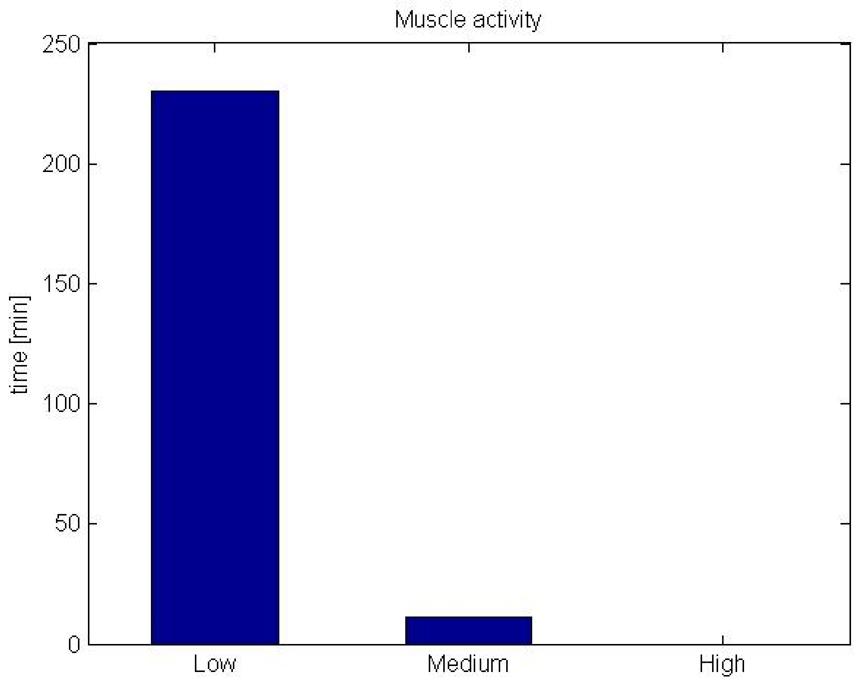


Fig. 54, muscle activity.

Measuring of the **subject 1**, duration: **241min**	
Executed activity	**Duration [min]**
Sitting	8
Lying	0
Static manual job	102
Dynamic manual job & driving in public transport or car	90
Walking without load	39
Walking with load	2
Running	0

Tab. 15, executed activities.

Muscle activity	
Level	**Duration [min]**
Low	230
Medium	11
High	0

Tab. 16, muscle activity.

6 Discussion

A system for shoulder joint motion activity determination was built and it is based on Arduino Leonardo, which is the most suitable microcontroller for this purpose. The Arduino board is cheap, small, powerful enough and with a low power consumption. SD card shield as storage for measured data is connected to this board. It can handle SD card with a capacity of 32GB, which is fully sufficient for the aim of this thesis. Information is recorded from two sensors – 3 axis accelerometer ADXL335 (Analog Devices Inc., USA) and surface EMG Muscle sensor v3 (Advancer Technologies, LLC, USA). The output voltage of these sensors is transmitted through a 10bit A/D converter into the Arduino board. The output data is then in a range of 0 – 1023, where 0 is 0V and 1023 is equivalent to 5V. The device is programmed to save the output data with a sampling frequency of 20Hz. The data is saved into a .csv (comma separated files) format files. There is always data from a one hour measurement in one file. Name of files are generated automatically starting with DATA00 and continuing with DATA01, DATA02, … , DATA99. New file is created every one hour or every time the board is turned on or restarted. Subsequently, the output data is evaluated in MatLab. An algorithm based on decision tree was created. This algorithm splits the data on a one minute periods and finds specified parameters for each of the periods. Then based on the parameters it recognizes the executed activity and the muscle activity of the deltoid muscle (low, medium or high).

The device is small and almost invisible when it is worn. It consists of a small case on a subject's belt and an accelerometer with 3 electrodes fixed on the shoulder hidden under a sleeve of a T-shirt. Also the wiring is hide under clothes.

The price of the device is comparable to wearable sensors, which makes it exceptional. It is less accurate (but the accuracy is still sufficient) compared to more expensive devices but it does not need any calibration.

A surface EMG was chosen as it wouldn't be possible to measure EMG by the Intramuscular type. This is possible only in a laboratory. Activity of only one motor unit is measured by the intramuscular type, thus it is not convenient for our use anyway. On the other hand, SMEG can measure activities of muscles, which are directly under the skin, and can be affected by many factors such as other muscles, thus it is not accurate enough compared the intramuscular type. But it is suitable for our application.

The thresholds for muscle activity defined as low, medium and high are only indicative and can be reset based on potential future research.

The electrodes should not remain on skin longer than two days due to a producer's recommendation but during the testing they were left on the skin of subject for four days without any rash on subject's skin or visible problem. Thus longer measuring than 2 days are possible.

Exact value verification of EMG can be disputable. Because the value depends on many factors such as the position of the sensors on the muscle, salinity and hydration of skin etc., so the same movement does not lead to precisely same values of SEMG. It follows, that our aim was to verify that our cheap SEMG sensor gives us signal associated with a defined muscle activity and not just some random signal.

Our device provides much more information about activity of the subject compared to wearable sensors as e.g. Fitbit flex bracelet, which was chosen as representative of smart bracelets. Our device recognized more types of activities, provides information about the position of the shoulder and, as a main novelty, our device classifies the muscle activity. The SEMG sensor is the uniqueness, which no other similar device has. Other sensors cannot recognize if the subject carries something while walking or what is his muscle activity. Our device provides this information thanks to the SEMG sensor connected to the right muscle.

For evaluating, methods like mean and standard deviation are used and the acceleration is evaluated as the sum of the tree vectors x, y, z. That makes the system robust against a slight disorder in sensor placement.

The activities are divided into two major groups – static and dynamic. Static is divided into rest and static manual job and rest is further divided into lying and sitting/standing. Dynamic is divided into locomotion and dynamic manual job & driving in car. Locomotion is further divided into walking and running and also walking is further divided into with and without load.

Range of the parameters are wider than precise measured values of parameters from subjects 1-5. This is because we assume that the movements can be performed faster, slower or in a slightly different way than they were done during our measuring.

The difference between a rest and a static manual should be further specified, so confusion of these two activities may appear. This could be fixed by using more parameters which could define these two activities more properly. The device is also not able to recognize when the subject is in a car or in a public transport or e.g. riding a bike and when he does a dynamic manual activity. This could be fixed by substituting the 3 axis accelerometer ADXL335 by the 9DOF sensor stick with an accelerometer, magnetometer and a gyroscope. We already have such sensor and we aim to further investigate its potential. It can provide more information about the position of the shoulder and it can separate static and dynamic acceleration, thus we can identify subject's speed by integration of the dynamic acceleration and recognize if he is in a car or public transport. This sensor is 35mm x 10mm small and the price is still around CZK 1,000, which makes it useable for our application. With this sensor, we expect even better results than we have now with our pilot device.

In the future, the algorithm can be further improved by a Graphical User Interface (GUI) being more user friendly. Afterwards, the device together with the evaluation software can be even used by a non-professional.

7 Conclusions

A device for determination of the shoulder joint activity was built. It is based on the Arduino Leonardo programmable board according to the specific aims of this thesis. It can be used during a clinical examination as well as for daily use without any distraction. The device is cheap, even in comparison with commercial wearable sensors. It uses an accelerometer and a surface electromyography (SEMG) sensor for measuring. It can provide more data about the shoulder joint activity compared to wearable sensors and it does not need any calibration. The main advantage of the proposed setup is in utilization of the EMG sensor. This sensor allows to quantitatively estimate the muscle activity and thus to provide more accurate information regarding the function of the musculoskeletal system. All of the components are commercially available.

The device is programmed for measuring acceleration and SEMG with the sampling frequency of 20Hz and saves the data on a SD card for future evaluation. The system was experimentally verified. Accelerometer was verified by the Xsens device (Xsens Technologies B.V., The Netherlands) and the SEMG sensor by the MA300 device (Motion Lab Systems, Inc, USA). An algorithm for evaluating the data was written after measuring some of the basic activities on different subjects. This algorithm based on artificial intelligence method of a decision tree was written in MatLab. The algorithm was verified on a different subject at the end and 4 hours of measurement with evaluation was performed. The device proved to be functional. The device is ready for use in clinical practice.

8 Sources

[1] Reiman, M. P. and Manske, R. C. (2011). The assessment of function: How is it measured? A clinical perspective. J Man Manip Ther, 19(2):91–99.

[2] Kim, S.-G. and Kim, E.-K. (2016). Test-retest reliability of an active range of motion test for the shoulder and hip joints by unskilled examiners using a manual goniometer. J. Phys. Ther. Sci., 28(3):722–4.

[3] Hegedus, E. J., Goode, A., Campbell, S., Morin, A., Tamaddoni, M., Moorman, C. T., and Cook, C. (2008). Physical examination tests of the shoulder: a systematic review with meta-analysis of individual tests. Br. J. Sports Med., 42(2):80–92; discussion 92.

[4] Leibold, M. R., Huijbregts, P. A., and Jensen, R. (2008). Concurrent criterion-related validity of physical examination tests for hip labral lesions: a systematic review. J. Man. Manip. Ther., 16(2):E24–41.

[5] Behery, O. A. and Foucher, K. C. (2014). Are Harris Hip Scores and Gait Mechanics Related Before and After THA? Clin. Orthop. Relat. Res., 472(11):3452–3461.

[6] Ceseracciu, E., Sawacha, Z., Cobelli, C. (2014) Comparison of markerless and marker-based motion capture technologies through simultaneous data collection during gait: proof of concept. PLoS, 9(3): 14–15.

[7] de Zambotti M, Claudatos S, Inkelis S, Colrain IM, Baker FC (2015). Evaluation of a consumer fitness-tracking device to assess sleep in adults. Chronobiol Int. 32(7):1024-8.

[8] Singh, J. A., Schleck, C., Harmsen, W. S., and Lewallen, D. G. (2016). Validation of the Mayo Hip Score: construct validity, reliability and responsiveness to change. BMC Musculoskelet. Disord., 17(1):39.

[9] ČIHÁK, Radomír. Anatomie. 2., upr. a dopl. vyd. Ilustrace Milan Med. Praha: Grada, 2001, 497 s. ISBN 80-7169-970-5.

[10] HUDÁK, Radovan a David KACHLÍK. Memorix anatomie. 2. vyd. Kroměříž: Triton, 2013, xxi, 605 s. ISBN 9788073877125.

[11] ANATOMY OF A WALK. Rubberbug [online]. [cit. 2016-06-27]. Available from: http://www.rubberbug.com/walking.htm

[12] The Anatomy of Swimming [online]. [cit. 2016-08-01]. Available from: http://kin480swimming.weebly.com/

[13] WIKIMEDIA FOUNDATION, INC. Wikipedia: The Free Encyclopedia [online]. 2001 [cit. 2015-05-20]. Available from: http://en.wikipedia.org/

[14] Medical Definition Search For 'Glenoid Labrum Of Scapula'. 2006. MediLexicon [online]. [cit. 2015-05-11]. Available from: http://www.medilexicon.com/medicaldictionary.php?s=glenoid+labrum+of+scapula

[15] [im] - interactive muscles / interaktivní svalové tabulky.: [online]. [cit. 2015-05-20]. Available from: http://www.physiotherapy.cz/im/

[16] Motion capture. Wikipedia: the free encyclopedia [online]. San Francisco (CA): Wikimedia Foundation, 2001- [cit. 2015-05-21]. Available from: http://en.wikipedia.org/wiki/Motion_capture

[17] FURNISS, Maureen. Motion Capture. MIT Communications Forum [online]. 1998 [cit. 2015-05-20]. Available from: http://web.mit.edu/comm-forum/papers/furniss.html

[18] Motion capture. Wikipedia: the free encyclopedia [online]. San Francisco (CA): Wikimedia Foundation, 2001- [cit. 2015-05-21]. Available from: http://cs.wikipedia.org/wiki/Motion_capture

[19] Wearable technology. In: Wikipedia: the free encyclopedia [online]. San Francisco (CA): Wikimedia Foundation, 2001- [cit. 2016-08-01]. Available from: https://en.wikipedia.org/wiki/Wearable_technology

[20] U.S. NATIONAL COORDINATION OFFICE FOR SPACE-BASED POSITIONING, NAVIGATION, AND TIMING. GPS.gov: Official U.S. Government information about the Global Positioning System (GPS) and related topics [online]. [cit. 2015-05-21]. Available from: http://www.gps.gov/

[21] *Fitbit* [online]. 2106 [2016-07-18]. Available from: https://www.fitbit.com/eu/flex

[22] Accelerometers. Integrated Publishing [online]. [cit. 2015-06-01]. Available from: http://www.tpub.com/neets/book15/63f.htm

[23] MEMS-based accelerometers. *Wikid* [online]. [cit. 2016-07-04]. Available from: http://www.wikid.eu/index.php/MEMS-based_accelerometers

[24] RF Cafe [online]. 1996- [cit. 2016-08-01]. Available from : http://www.rfcafe.com/

[25] Accelerometer. Wikipedia: the free encyclopedia [online]. San Francisco (CA): Wikimedia Foundation, 2001- [cit. 2015-06-01]. Available from: http://en.wikipedia.org/wiki/Accelerometer

[26] Electromyography. Wikipedia: the free encyclopedia [online]. San Francisco (CA): Wikimedia Foundation, 2001 [cit. 2015-06-01]. Available from: http://en.wikipedia.org/wiki/Electromyography

[27] Arduino Leonardo. Arduino [online]. [cit. 2015-05-24]. Available from: http://www.arduino.cc/en/Main/ArduinoBoardLeonardo

[28] SD Card Shield V4. Seeed: Grow the Difference [online]. 2008- [cit. 2015-05-26]. Available from: http://www.seeedstudio.com/depot/SD-Card-Shield-V4-p-1381.html

[29] Hobby Electronics [online]. 2016 [cit. 2016-08-01]. Available from: http://www.hobbytronics.co.uk/adxl335

[30] ANALOG DEVICES, INC. ADXL335 [online]. 2009 [cit. 2015-06-01]. Available from: https://www.sparkfun.com/datasheets/Components/SMD/adxl335.pdf

[31] Sparkfun [online]. [cit. 2016-08-01]. Available from: https://www.sparkfun.com/

[32] Muscle sensor v3. Advancer Technologies: Advancing the future [online]. 2013 [cit. 2015-06-01]. Available from: http://cdn.sparkfun.com/datasheets/Sensors/Biometric/Muscle%20Sensor%20v3%20Users%20Manual.pdf

[33] INSET [online]. [cit. 2016-08-01]. Available from: http://www.inset.cz/

[34] T.S.Bohemia [online]. [cit. 2016-08-01]. Available from: www.tsbohemia.cz

[35] Amio [online]. [cit. 2016-08-01]. Available from: http://www.amio.cz/

[36] CRISWELL, Eleanor a Jeffrey R CRAM. Cram's introduction to surface electromyography. 2nd ed. Sudbury, MA: Jones and Bartlett, 2011, xxi, 412 p. ISBN 07-637-3274-5.

[37] Examples. Arduino [online]. 2015 [cit. 2015-06-03]. Available from: http://www.arduino.cc/en/Tutorial/HomePage

[38] Forum. Arduino [online]. 2015 [cit. 2015-06-03]. Available from: http://forum.arduino.cc/

[39] ARDUINO: příručka programátora. HOBBYROBOT: Malá robotika a mechanické modely [online]. 2015 [cit. 2015-06-03] (only in Czech). Available from: http://www.hobbyrobot.cz/wp-content/uploads/ArduinoPriruckaProgramatora.pdf

[40] VODA, Zbyšek. HW KITCHEN. Průvodce světem Arduina [online]. 2014 [cit. 2015-06-03]. Available from: http://arduino.cz/

[41] Arduino.cz [online]. 2014 [cit. 2015-06-03]. Available from: http://arduino.cz/

[42] Nyquist–Shannon sampling theorem. In: Wikipedia: the free encyclopedia [online]. San Francisco (CA): Wikimedia Foundation, 2001- [cit. 2016-06-30]. Available from: https://en.wikipedia.org/wiki/Nyquist%E2%80%93Shannon_sampling_theorem

[43] Patrik KUTÍLEK, M.Sc., Ph.D., Zdeněk SVOBODA, M.Sc., Ph.D., Ondřej ČAKRT, M.Sc., Ph.D. and Vladimír SOCHA, M.Sc.. Assessment of postural stability in patients with cerebellar disease using gyroscope data. *Science Direct* [online]. 2014, **2014**, 9 [cit. 2016-07-05]. Available from: www.sciencedirect.com

[44] MTx. Xsens [online]. [cit. 2016-08-01]. Available from: https://www.xsens.com/products/mtx/

[45] MA-300 EMG System User Guide [online]. 2004, 65 [cit. 2016-07-05]. Available from: https://www.motion-labs.com/pdf/ma300_user_manual_harwin.pdf

[46] Medical dictionary. The free dictionary by farlex [online]. 2003- [cit. 2016-08-01]. Available from: http://medical-dictionary.thefreedictionary.com/

[47] Fourier transform. In: *Wikipedia: the free encyclopedia* [online]. San Francisco (CA): Wikimedia Foundation, 2001- [cit. 2016-07-18]. Available from: https://en.wikipedia.org/wiki/Fourier_transform

[48] Discrete Fourier transform. In: Wikipedia: the free encyclopedia [online]. San Francisco (CA): Wikimedia Foundation, 2001- [cit. 2016-07-18]. Available from: https://en.wikipedia.org/wiki/Discrete_Fourier_transform

9 Appendix list

Appendix A: Parameters and graphs of selected activities

Appendix B: User Manual

Appendix C: List of electronic appendix

Appendix D: The electronic appendix on the enclosed CD

Appendix A: Parameters and Graphs of selected activities

<u>Walking:</u>

Subject	Parameters	
	f(max) [Hz]	S(max) [-]
1	1.833	2081.8
2	1.65	750.2
3	1.85	1152.4
4	1.867	1089.7
5	2	1125.5

Tab. A1, Measured parameters of walking.

Following graphs show 1 min measuring of selected data while walking of subject 1.

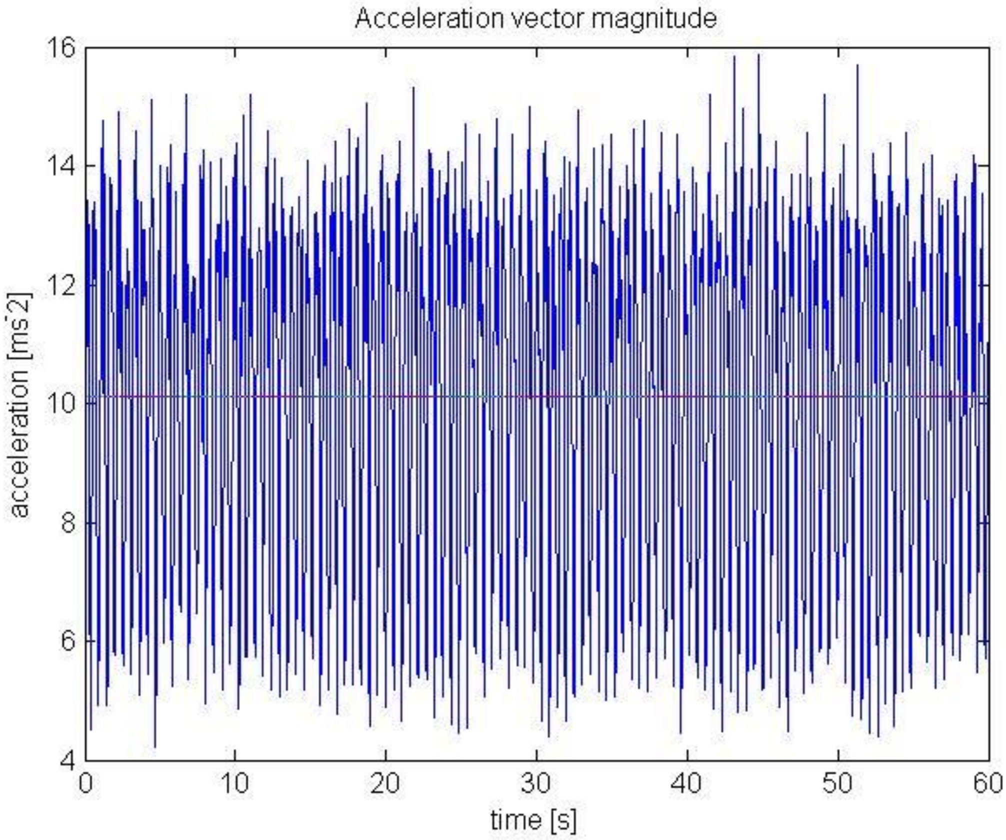


Fig. A1, Acceleration vector magnitude while walking.

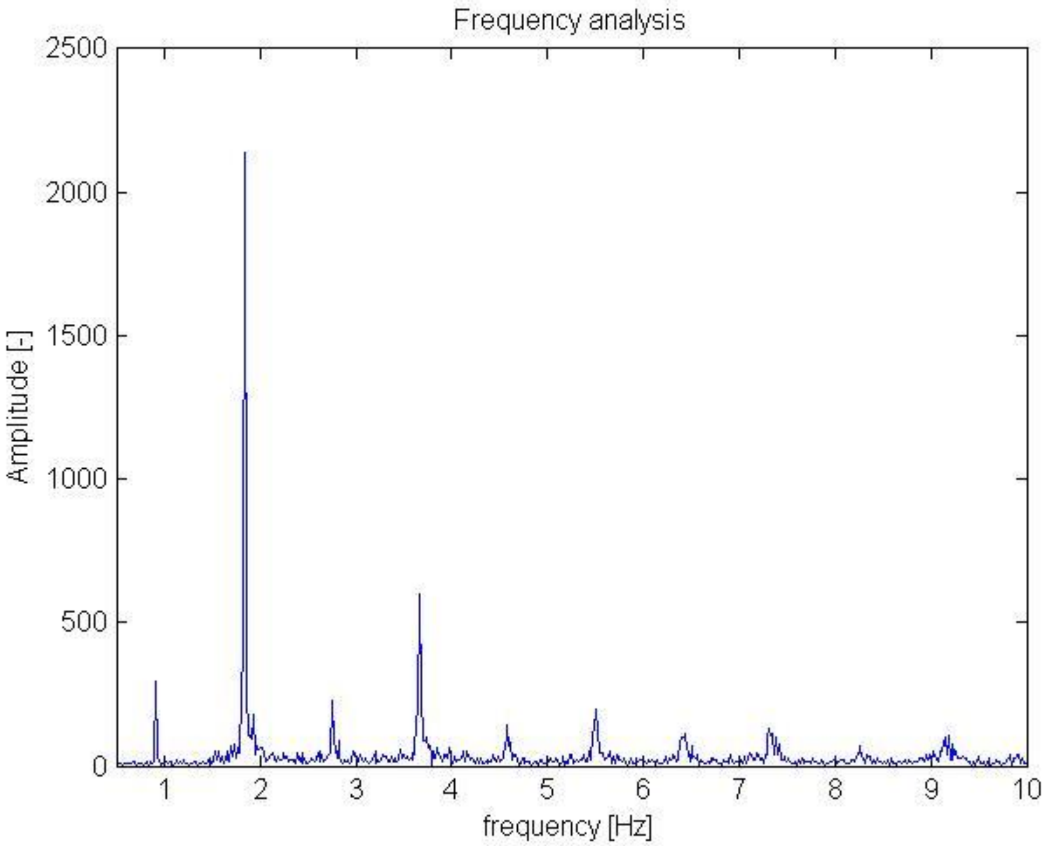


Fig. A2, Frequency analysis of acceleration vector magnitude while walking.

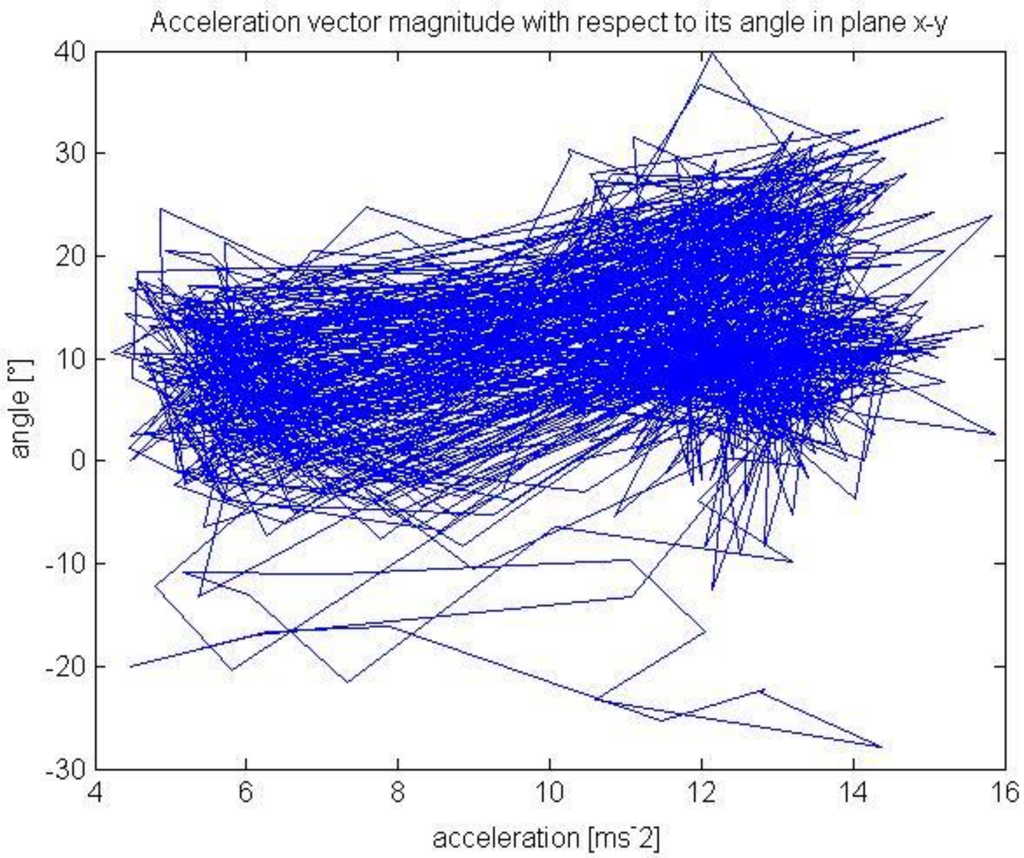


Fig. A3, Acceleration vector magnitude with respect to its angle in plane x-y while walking.

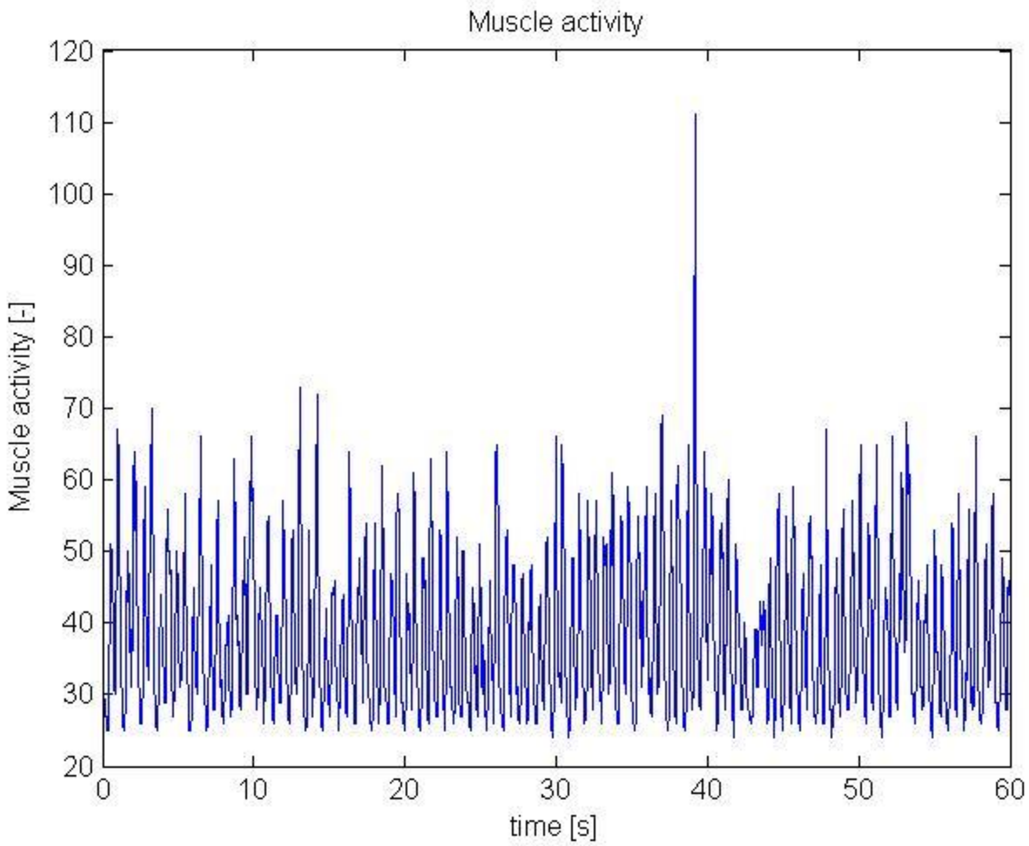


Fig. A4, Muscle activity while walking.

<u>Running:</u>

Subject	Parameters	
	f(max) [Hz]	S(max) [-]
1	2.6	4587.1
2	2.6	3613.8
3	2.97	2949.4
4	2.53	3298.5
5	2.59	3145.6

Tab. A2, Measured paramters of running.

Following graphs show 1 min measuring of selected data while running of subject 1.

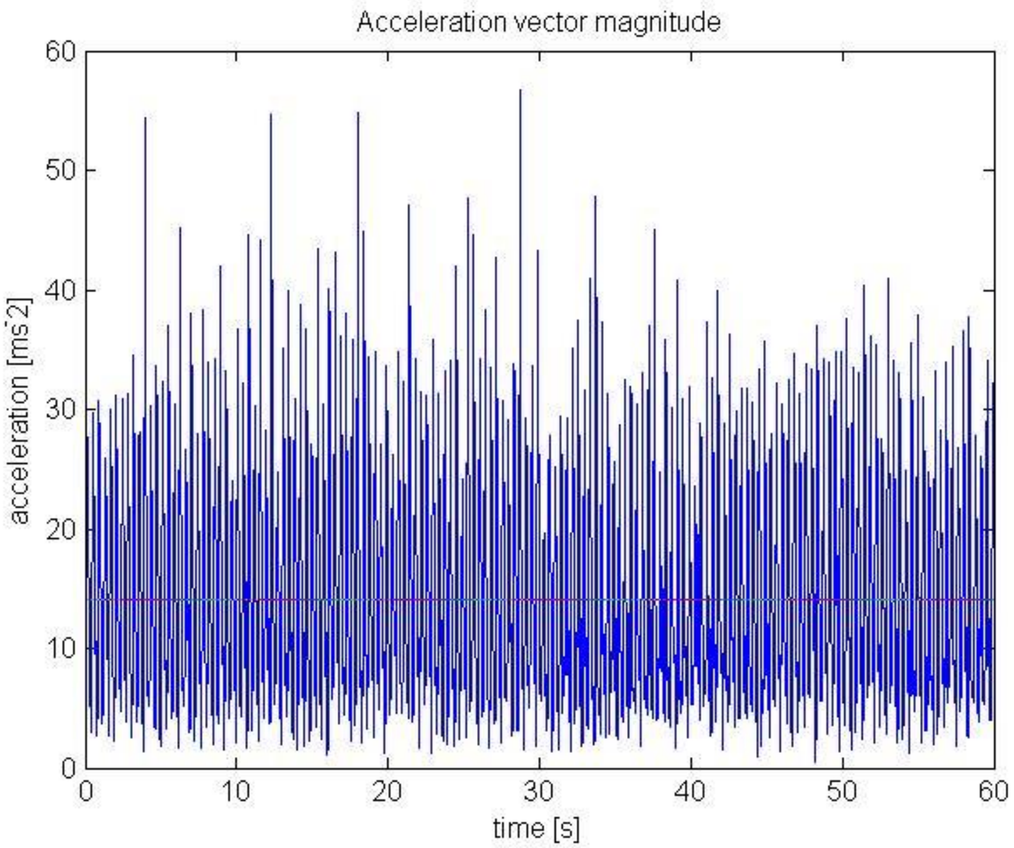


Fig. A5, Acceleration vector magnitude while running.

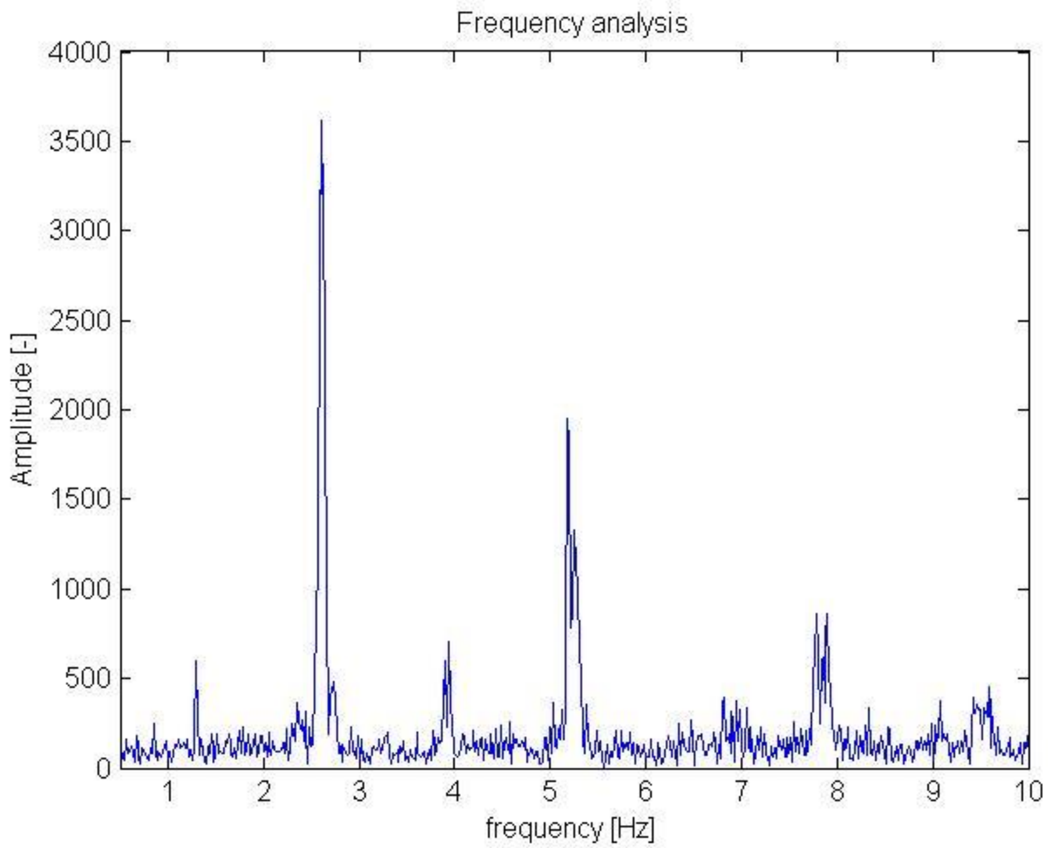


Fig. A6, Frequency analysis of acceleration vector magnitude while running.

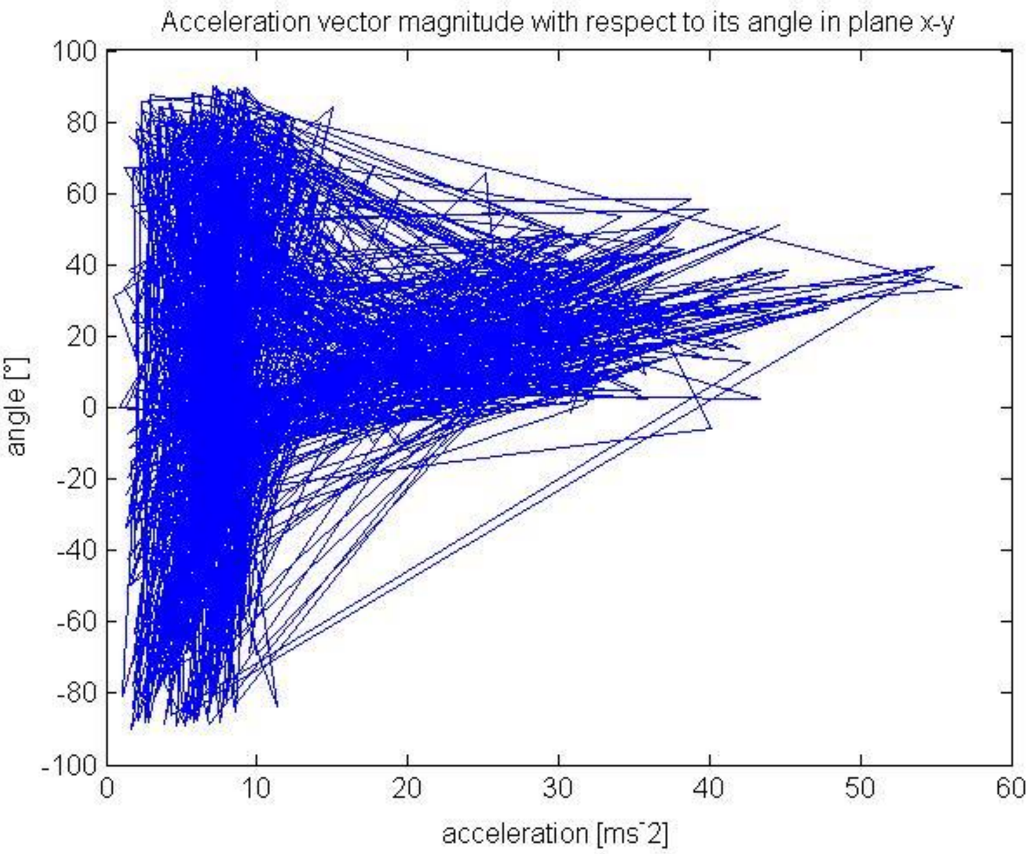


Fig. A7, Acceleration vector magnitude with respect to its angle in plane x-y while running.

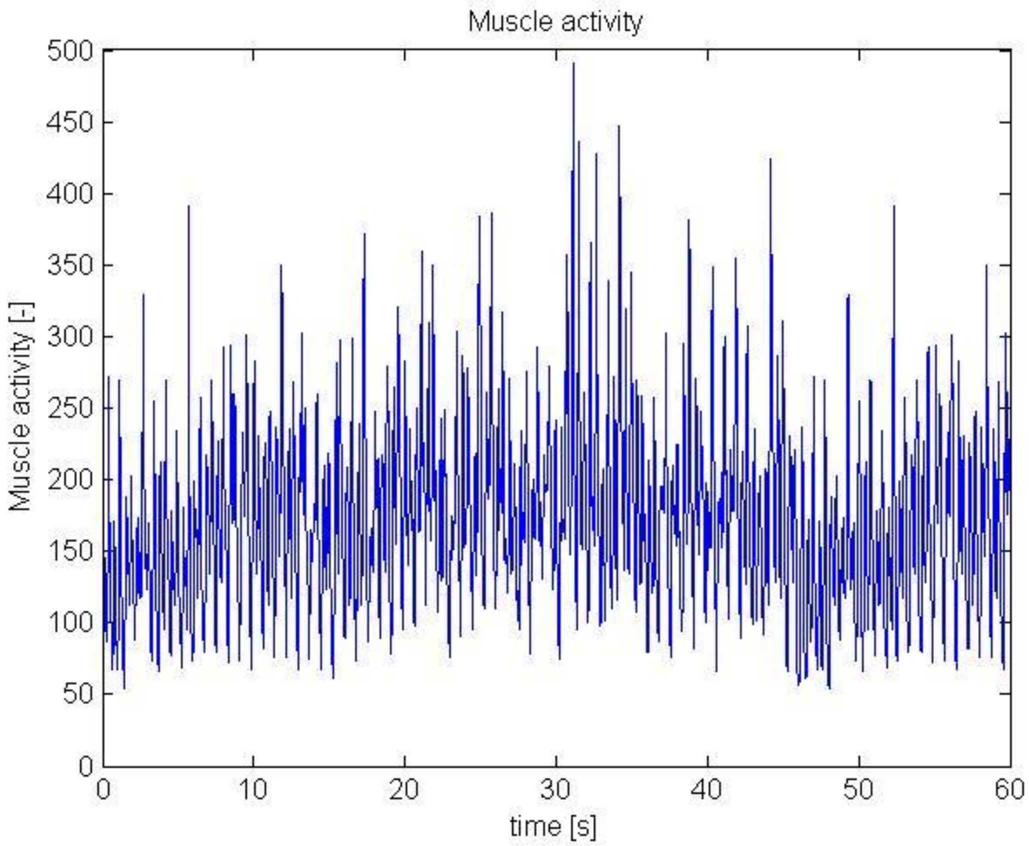


Fig. A8, Muscle activity while running.

Static manual job:

Subject	Parameter (std(a))
1	0.31
2	0.16
3	0.19
4	0.3
5	0.24

Tab. A3, Measured parameters of static manual job.

Following graphs show 1 min measuring of selected data while working on computer, which is part of static manual job of subject 1.

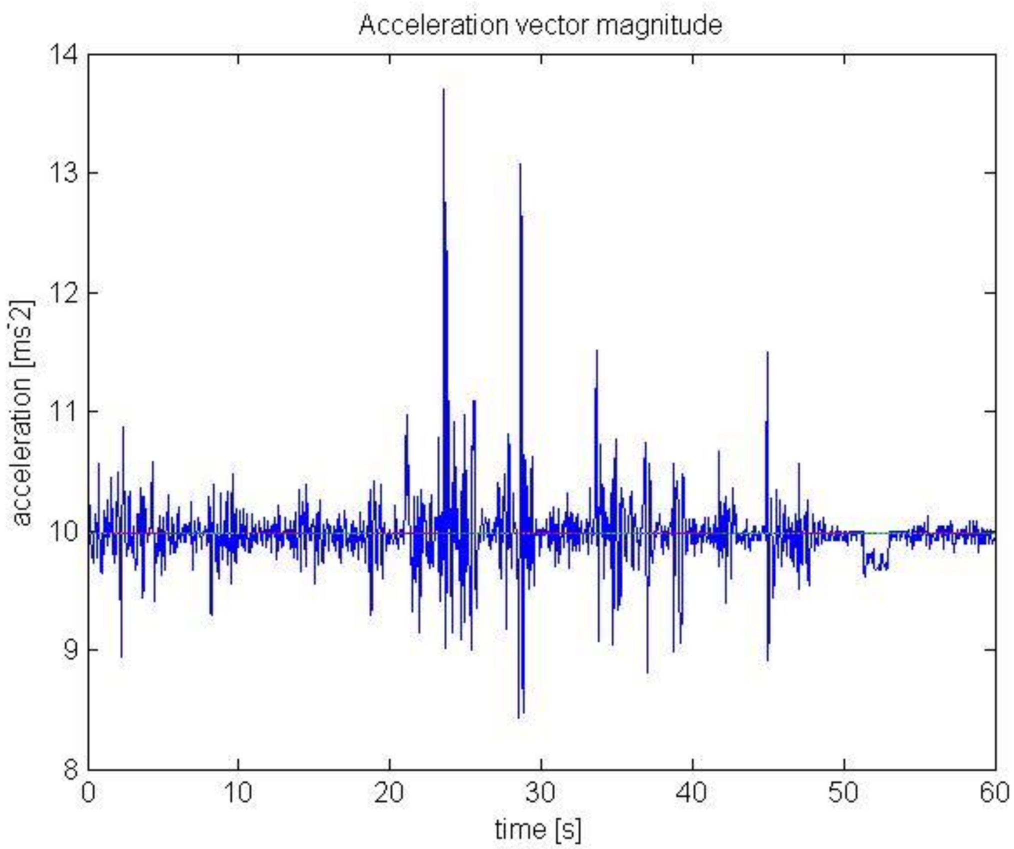


Fig. A9, Acceleration vector magnitude while working on computer.

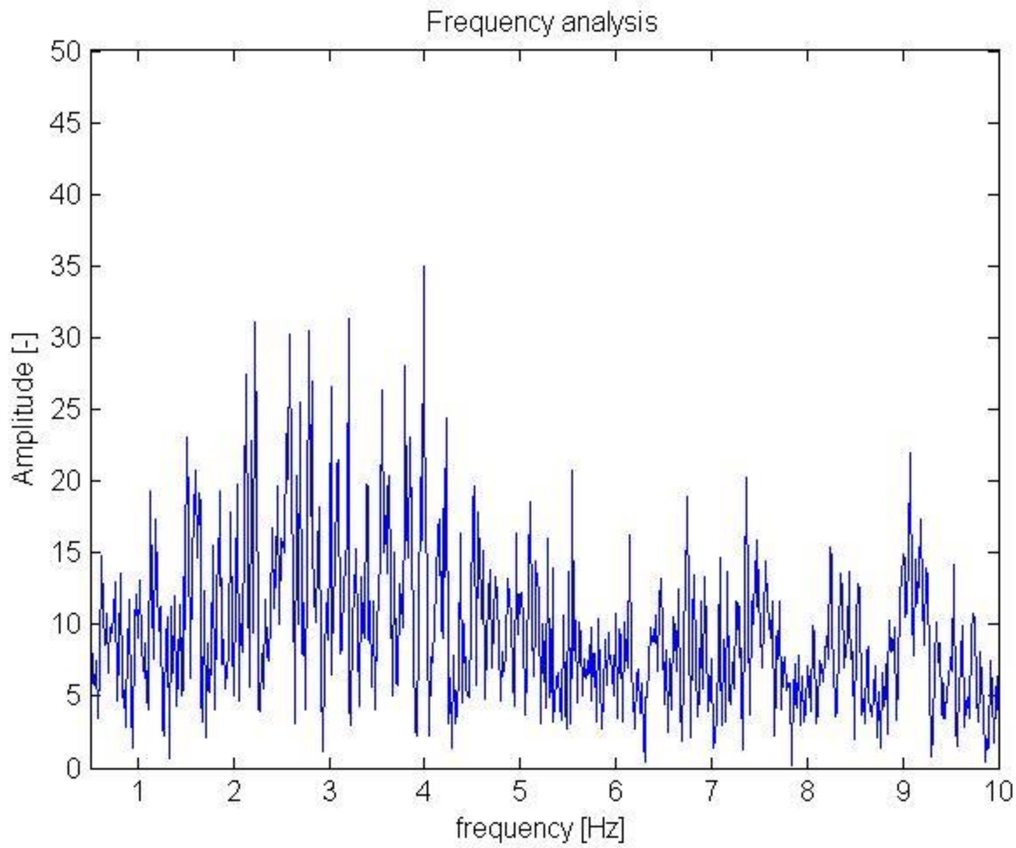


Fig. A10, Frequency analysis of acceleration vector magnitude while working on computer.

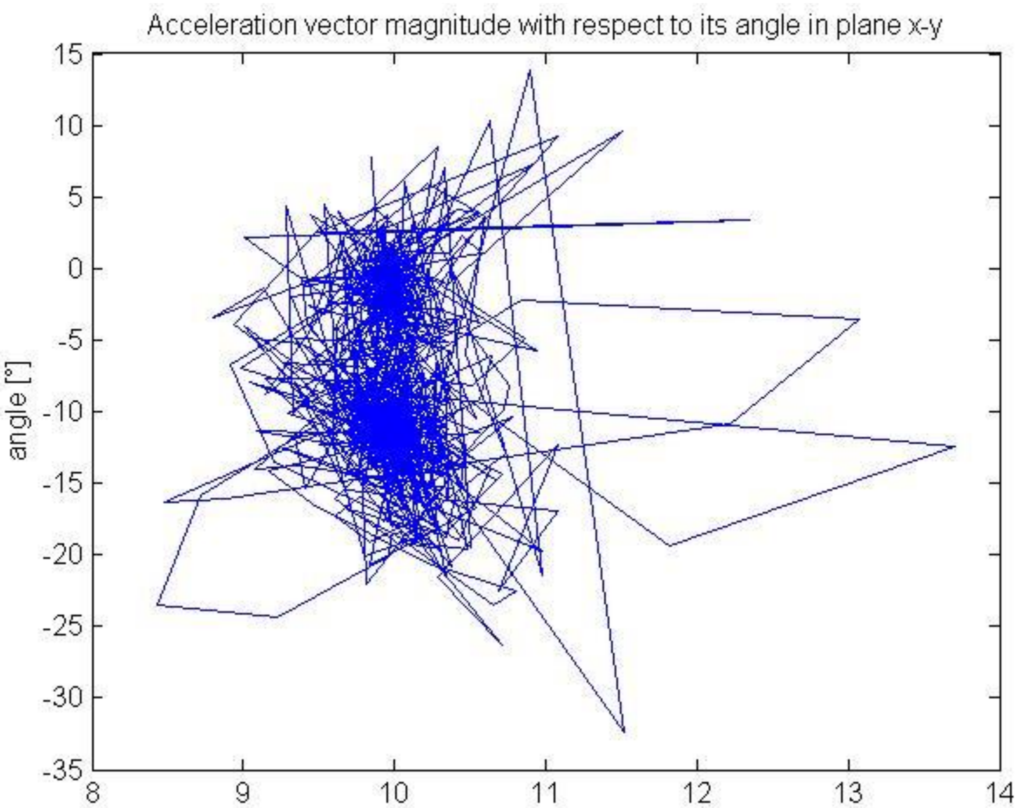


Fig. A11, Acceleration vector magnitude with respect to its angle in plane x-y while working on computer.

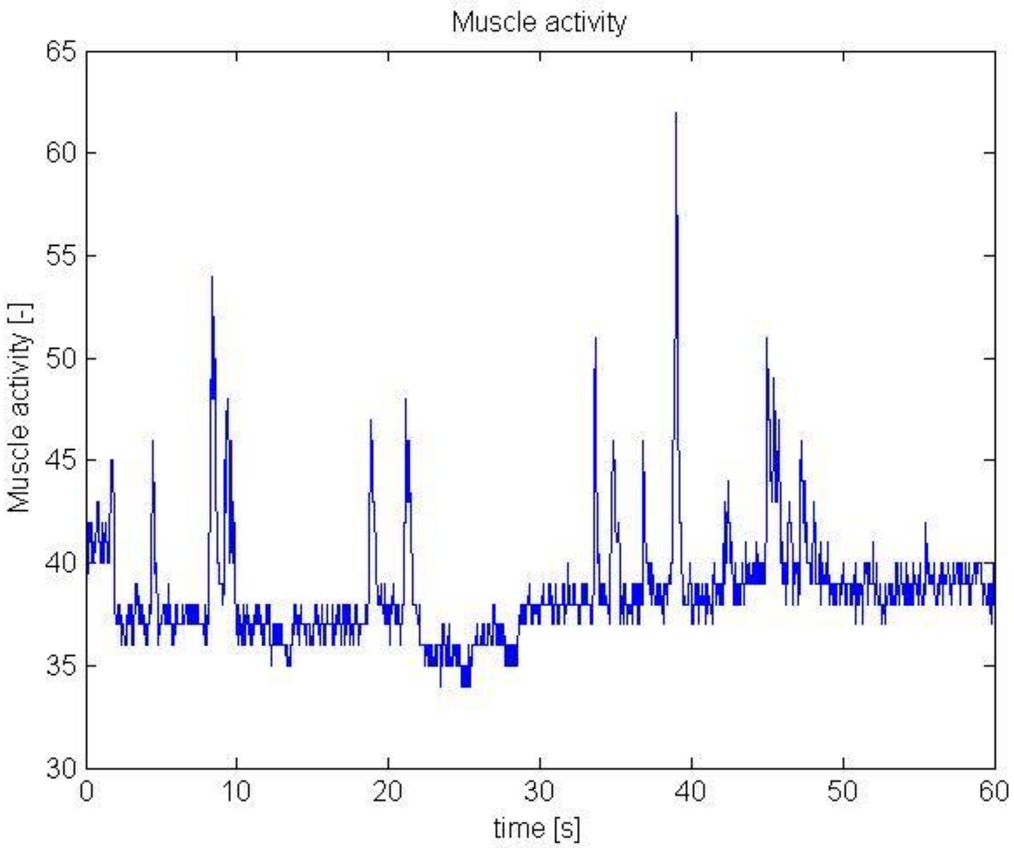


Fig. A12, Muscle activity while working on computer.

Appendix B: User manual

The device has to be set and placed on a patient with respect to Chapter 4.1.9 Experimental setup and in the Arduino board has to be uploaded Arduino_programme.ino programme.

The device starts measure with connecting the power source to the board.

After the measurement, place the data from SD card into the same folder as are Matlab files – Parameters.mat and Decision tree.mat. Then in Parameters.mat define the number of files with measured data (numfiles). When these procedures are done, then the Parameters.mat programme can be started and it calculates the parameters. After Parameters.mat can be started Decision tree.mat programme, which shows the executed activities and their duration in minutes as well as the muscle activity with duration of each level of it.

The executed activities and corresponding variables in Decision tree.mat programme are presented in Tab. B1. Muscle activity levels and corresponding variables in Decision tree.mat programme are presented in Tab. B2.

Acitivity	**Variable**
Dynamic manual job	dmj
Lie	lie
Run	run
Sit/stand	sit
Static manual job	smj
Walk with a load	walkwith
Walk without a load	walkwithout

Tab. B1, executed activities and corresponding variables.

Muscle activity level	**Variable**
Low	low
Medium	medium
High	high

Tab. B2, muscle activity levels and corresponding variables.

Appendix C: List of electronic appendix

Following files are on enclosed CD (appendix D).

Adxl335.pdf – manual for accelerometer ADXL335

Arduino Leonardo Schematic – schematic graph of Arduino Leonardo board

Arduino_programme.ino – code for Arduino board

Atmel-ATmega32U4 – manual for microcontroller ATmega32U4

DATA00.csv, DATA01.csv, DATA02.csv – examples of measured data

Decision tree.mat – Matlab code for evaluating the parameters

Muscle sensor v3 – manual for Muscle sensor v3

Programme for parameters.mat – Matlab code for calculating parameters from measured data

SD Card Shield Schematic – schematic graph of SD card shield