Online Monitoring of Posture for Preventive Medicine Using Low-Cost Inertial Sensors

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Abstract. People in many professions suffer from low back pain (LBP) due to wrong movements. Although this is anticipated by occupational medicine, the quality of evidence is low, since little objective measurements about the spine position in daily-use exist. The paper presents an ultra-flat posture monitoring system based on low-cost acceleration sensors, which can be very efficiently be used to measure the posture of the spine. First experiments (lab-based and in daily-use) showed a deviation of approximately 1° with a low standard deviation. Innovation is the suitability for daily-use by sensors having a height of 2,5 mm that allow a seamless usage even during positions applying pressure to the back such a leaned sitting on a chair.

Keywords: Spine posture · Low back pain · Inertial sensors Online monitoring

1 Introduction

Extreme or awkward postures like bending or twisting of the back are presented as risk factors for developing low back pain (LBP) in several clinical guidelines (e.g. [1–3]). LBP patients have a reduced lumbar range of motion and move more slowly compared to people without low back pain [4]. Restriction in lateral flexion as well as reduced lumbar lordosis are also associated with increased risk of developing LBP [5].

Even though ergonomic interventions are key elements of physiotherapy and occupational medicine [6], the evidence for prevention of LBP is uncertain, because the quality of evidence is low [7, 8]. Recent systematic reviews cannot show (or disprove) a direct effect of extreme spine positions and a higher risk for LBP (e.g. [9–11]).

Application of sensors can help to increase evidence but at the same time it is necessary that such sensors are not only used during clinical studies but are also available in low-cost versions for wide-spread application in preventive and occupational medicine. This paper will introduce a low-cost, ultra-flat sensing platform that can measure spine posture during daily live without disturbing the movements of the patient wearing the sensor system.

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2 State-of-the-Art

2.1 Measurement Principle and Medical Indication

Studies often assess the angle between thigh and trunk/pelvis (hip-flexion) as a measure for posture of the spine or simply quantify the angle between trunk and floor [10, 12–15]. This is also reflected by posture assessment tools of occupational medicine like OWAS (Ovako Working Posture Assessment System) REBA (Rapid Entire Body Assessment) or RULA (Rapid Upper Limb Assessment) that are widely used in scientific literature.

Nevertheless, this method has a serious flaw: The angle between the thigh and the trunk/pelvis is no measure for the position (curvature) of the back or even the position of the upper trunk in relation towards the floor, because bending forward can be done with a straight or flexed back (see Fig. 1). Forward bending should be evaluated assessing also the curvature of the back. Only few studies exist measuring the position (curvature) of the back with modern sensing methods during daily live [16].

Fig. 1. Curvature of spine: neutral spine (left) and flexed spine (right) position both showing the same amount of hip flexion. (Photos by Katharina Müller)

Even though there is no good evidence at the moment about correlation of extreme postures of the back and risk of back pain, some reviews show a dose-response relationship [12, 17, 18]. Hence, future occupational studies require to gather data under realistic daily-use circumstances for the full activity period instead of small time frames during a working day. A precise quantification of the position (curvature) of the back instead of using hip flexion or trunk position relative to the floor needs to be available to find medical evidence and derive therapeutically measures.

2.2 Measurement Methods

In order to determine the position (curvature) of the spine different approaches exist: Optical motion capture systems uses optical (infrared and visible light) markers attached to the person, which are scanned by external cameras [19]. These systems can be referred as the golden reference standard, yet are not further discussed, since they require dedicated test rooms and are therefore not suitable to monitor the spine movement in everyday situations.

Maier et al. [20] use strain gauges from Epionics applied over the full length of the spine and 3 axis accelerometers at the end of the gauges to determine the bending angel. The system connects the sensors via cables to a measurement device and requires dedicated band-aid that allow the movement of the strain gauges along the back of the patient. A disadvantage stems band-aid and from the fixed length gauges, since it cannot be adapted to different body sizes.

Dinu et al. [21] investigates the use of 17 inertial sensors (MVN Biomech system from Xsens) and compared it with a Vicon optical infrared marker system with eight camcorders to acquire the full body movement. Both systems only show a position difference of less than 6 mm. Problems are identified in the drift of acceleration sensors and the relative movement of the sensors on soft tissues of the subject. This relative movement might be due to relatively high weight of 16 g and size of $47 \times 30 \times 13\ \text{mm}^3$. Additionally, the sensors have a battery lifetime of 6 h and require data recorded at 120 Hz to be continuously transferred to a base station. Similar systems are offered by Hocoma (valedotherapy.com) for bio-feedback of physical exercises using Bluetooth connected sensors. Yet, these sensors are designed for dedicated exercise and therefore short operation time of around 30 min before they need to be recharged.

Dorsavi [22] offer solutions based on 3D accelerometers, gyroscopes and a magnetometer designed for workplace application offering operation of 24 h and datastorage up to 72 h. Mjøsund et al. [23] compared the Dorsavi ViMove system to the Vicon system determine a RMS error of 0,71° up to 2,11° depending on the direction of flexion. To our analysis the Dorsavi system is very mature for an every-day use, yet it still uses relatively big sensors only slightly smaller than a match box.

Additionally, very simple sensors such as the acceleration sensor from Back-Track [24] or Lumolift [25] are available, yet these sensors only deliver the position compared to the earth magnetic or gravitation field (hip-flexion). Both sensors are either applied by a pouch or a magnetic clip on the side or front of the body.

3 Mobile Measurement System

Summarizing existing approaches there are two limitations identified: First, an important fact for all sensors applied to the skin is that the movement of soft tissue is influencing the measurement quality, since the measured position deviates from the position of the spine [26, 27]. In particular for overweight individuals this effect needs to be considered. Second, for everyday use in preventive and occupational medicine applications the thickness of the sensor is of high importance. Whereas many application in sports or therapeutic applications are in movement or without contact to hard surfaces, in preventive and occupational medicine positions like leaning or lying are more likely. If the sensor is too thick this will cause unnatural behavior. To overcome these limitations innovation is required to reduce the mass of the sensor as well as the thickness to suppress unnatural behavior and at the same time retaining a reasonable size to avoid artefacts of punctual tissue movement.

3.1 Inertial Sensor System

For the angular measurement of the spine, low cost inertial MEMS sensors where chosen. Aside the cost consideration the measurement of the earth magnetic field was excluded, since it is not reliable enough for accurate angular measurement. After evaluation of different sensor types (e.g. LSM9DS0, MPU9250) the LSM6DS3 was chosen in the version having a measuring ranges of ±2 g (accelerometer) and ±125°/s (gyroscope). The sensor is mounted on a 1,5 mm thick PCB giving the necessary stiffness and is encapsulated with epoxy resin resulting in an overall thickness of 2.5 mm. Experimental version with a chip embedded inside the PCB has been designed with reduce thickness but discarded due to complicated manufacturing. The sensor diameter was set to 25 mm as optimum between wearing comfort and suppression of soft tissue tilting and movement. The sensors are attached to the body using adhesive tape. Each sensor is connected to the device by a thin, flexible and robust I^2C cable providing distortion free transmission and the flexibility to use other sensor chipsets.

3.2 Recording Device

The recording device consists of a touchscreen, an AVR microcontroller, flash memory and a low cost WLAN module (see Fig. 2).

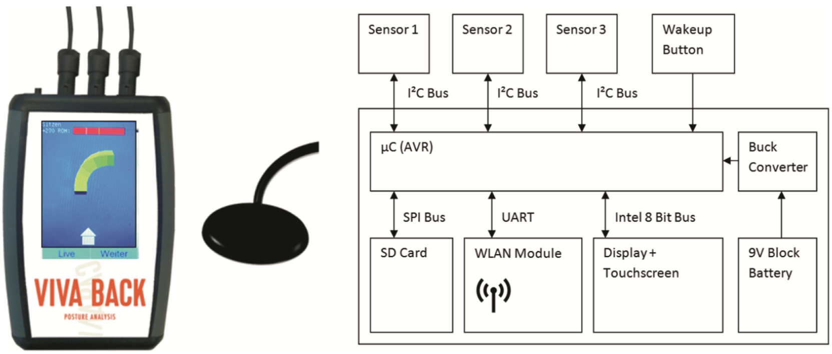

Fig. 2. Device in configuration mode with schematic spine (left), magnified sensor pad (middle) and block diagram (right)

There are three main modes: *setup, record* and *upload mode*. *Setup mode*, is dedicated to configuration and a guided calibration process determining maximum stretch and bend positions. For better positioning and verification of the three sensor pads a simple online spine model can be displayed (Fig. 2 left). In *record mode*, sensor data is processed and recorded on a flash memory. At any time the user can enter her current activity such as gardening, computer work or sport activity as well as a pain level to support posterior diagnoses. Finally, in the *upload mode* data can be transmitted to a designated FTP server over WLAN using an XML based encoding protected by a pre-shared WPA2 key. In the final device continuous upload is possible if a WLAN connection is available, but a doctor still can retrieve the data via *upload mode* if no connection during activity exists.

Special efforts in circuit design allowed to reduce the current consumption to approx. 5 mA in record mode easily allowing for a 24 h monitoring going beyond most of the state-of-the-art devices. Only during user interaction and WLAN upload, the current consumption raises to 100 mA primary due to the touch panel display, which is also activated during data upload. A Bluetooth Low-Energy chip will be integrated to reduce energy consumption further.

3.3 Data Processing and Algorithms

Data from the sensors is recorded at a speed of 20 Hz and the ground pointing vector of each sensor is calculated by fusion of gyroscope and accelerometer data. The gyroscope is slowly drifting and therefore only reliable at higher frequencies. The accelerometer is disturbed by quick movements, and more reliable at lower frequencies. So a first order complementary filter has been implemented with a cut-off frequency of 0,2 Hz. Two sensors, located at the skin over S2 of the sacrum and at the upper part of the sternum (sternum manubrium) are used to calculate the bending of the spine, whereas the third sensor is placed at the thigh and used for activity (sitting, standing, walking) tracking and as pedometer. Processed sensor data is stored on a flash memory for later diagnosis but also online alarms to give live feedback and help change erroneous posture patterns.

4 Device Characterization and Measurements

To measure the accuracy of the presented sensor system and algorithm following methods have been applied: (a) measurement of a reference dummy emulating the spine, (b) analysis in a 12 h practical daily-use example and (c) a comparison with an optical motion tracking system.

For the first test, the sensors were repeatedly positioned in a fixed angle to each other. The deviation from that constant value is the error that the device produces. Results of these measurements are shown in Fig. 3. On the left side. The mean error and the standard deviation were calculated for different cases. A particular resting position was recorded for 12 h to analyze if there are any long-term drift effects. No drift was observed during the measurements. Rotation of the sensor and linear movements as well as shocks were performed on the sensors in all rotational axis and directions.

Thereby the sensors were taken to their measure boundaries of ± 2 g and $\pm 125°/s$. Under extreme conditions exceeding the specification limits of the sensors, the precision deteriorated, especially, during rotation performed with 2 Hz and a range of $\pm 75°$ and shock performed in an interval of one second with more than 2 g acceleration.

For the second test, the sensor system was carried by a person in daily-use mostly consisting of office work, walking and relaxing. Two sensors have been applied on the

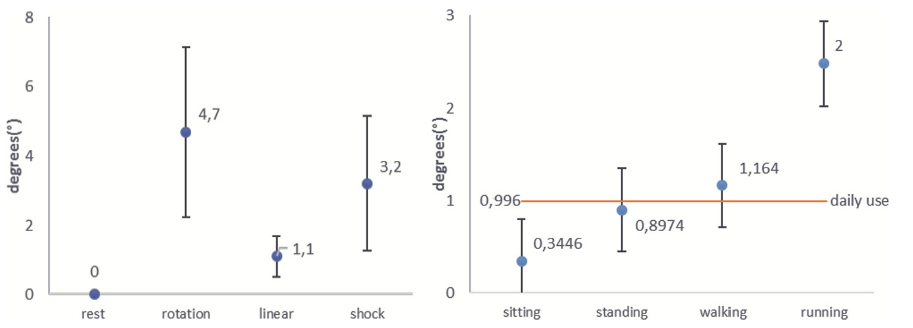

Fig. 3. Mean and standard deviation of measurement error for different disturbances (left) and mean and standard deviation for different activities (right)

lower back with a difference angle of zero degrees for a whole working day without removal. Thus, any value other than zero can be interpreted as a measurement error. In Fig. 3 (right side) the resulting error is plotted for different activities during the day. As expected the error is higher for dynamic movements yet significantly smaller than the static test values. Noteworthy is the very small overall deviation in the daily-use example of less than 1°. Only heavy shocks as occurring during running will result in higher values, yet are less frequent than during the rotation and shock tests.

As a last test the sensor values have been compared with the video motion tracking system Kinovea [28]. Two visual markers and one sensor has been attached to stiff plates and were attached to the lower back area and to the upper part of the chest.

Figure 4 shows first preliminary data obtained comparing the motion sensors and the optical references. It needs to be noted that the optical imaging software has a relatively high rate of false values, if the test person does not move exactly in the plane of the video and single optical marker are not (fully) visible. This is a principal problem caused by the diametric marker positions when using an optical system with a single camera. The preliminary deviation of $6.31° \pm 4.76°$ degrees is therefore accounted mostly to the

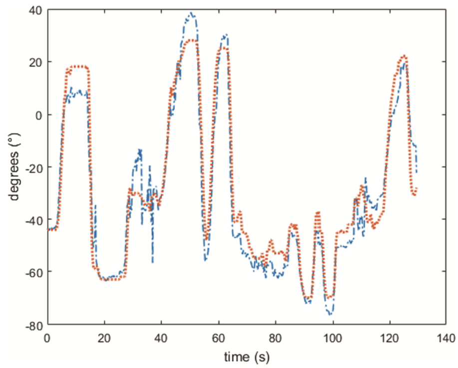

Fig. 4. Example measurement of optical system (blue) versus measurement of developed sensor based system (orange). (Color figure online)

bad optical situation and needs to be reinvestigated with multi-camera optical reference systems.

5 Outlook and Conclusions

The presented work shows that ultra-flat low-cost acceleration sensors can be very efficiently used to measure the posture. First experiments (lab-based and in daily-use) showed a deviation of less than 2° with a low standard deviation. Sensors having a height of 2,5 mm allow a seamless usage even during position applying pressure to the back such a leaned sitting. Yet, measurements also revealed that acceleration values especially during running are much higher than expected and cause outliers and higher deviations due to exceeding the sensor's measurement range.

Ongoing activities therefore focus on higher error reduction and clinical evaluation. First results backup the experimental results presented, but need to undergo a detailed evaluation not available when submitting this paper, since the clinical evaluation has not been finished. Next steps will be the evaluation of both sensors capable of switching to different measurement ranges and additional low-cost acceleration sensor to measure high accelerations. Research needs to be done to evaluate different sampling rates, energy consumption and algorithms to more precisely detect outliers and best to calculate interpolation values for continuous recording.

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