Novel Weight Estimation Analyses and the Development of the Wearable IngVaL System for Monitoring of Health Related Walk Parameters

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***Abstract*— The total amount of lifted weights and lift frequency are moderate to strong risk factors for lower back pain. Measurement of carried weight is thereby of interest. The aim of this paper is to (1) present three novel analyses methods for estimation of weight during walk and (2) to describe the design process of the cost-effective research system IngVaL based on pedobarography. The paper will also (3) present the durability of the sensors. Motivations for choices in the system design are given for hardware, selection of sensor type, sensor implementation and calibration of sensors. To measure weight during walk with IngVaL, fifteen test persons made five walks each with a pseudo-random added extra weight. Three analyses methods were tested, for estimation of weight while walking, resulting in Root Mean Square Errors of 11.3 kg, 7.1 kg and 6.1 kg respectively. The durability of the sensors were tested in an outdoors walking condition. It can be concluded that the IngVaL system shows good durability and that weight during walk is possible to measure with simple analyses methods.**

Keywords- pedobarography; carried weight; portable; wearable; insole; in-shoe; personal health monitoring; measurement system design.

I. INTRODUCTION

This paper is an extension of a conference paper, which reported a Root Mean Square Error (RMSE) of 13.8 kg for monitoring of weight while walking [1].

There are numerous applications for analyses of foot plantar pressure distribution using insoles, containing pressure sensors, in the shoes. Examples are gait analysis [2], posture analysis [3], humanoid robotics [4], evaluation of footwear [5] and footwear design [6], sports [7][8], stroke rehabilitation [9][10][11] and measurements during daily human activity [12][13].

There are several types of sensors that are used in systems for foot plantar measurement [14]. Four commonly used sensor types, that are used by researchers building pedobarography systems, are the capacitive [15], the piezoelectric [16], the resistive [17][18][19] and the optoelectronic [13][19]. Three commonly used commercial portable pedobarography systems, for respective sensor type, are compared in Table 1.

Some applications for analyses of foot plantar pressure distribution need insoles with a matrix of sensors. But in some applications, when this is not needed, it is of interest to use insoles that cost less and have good durability to allow measurements in large populations over a long time. For this kind of measurements over time a pedobarography research system has been developed. First, a prototype version was designed by the authors. An improved second version was named Identifying Velocity and Load (IngVaL).

Table 1. Three commercial portable pedobarography systems compared.

System Name	Pedar™	F-Scan™	ParoTec™
Company	Novel GmbH, Munich, Germany	Tekscan, Inc., Boston, USA	Paromed GmbH, Heft, Germany
Sensor Type	Capacitive	Resistive	Optoelectronic
Number of Sensels	Up to 256	Up to 960	Up to 36
Insole Thickness	1.9 mm	0.15 mm	3.5 mm
Pressure Range	(0.03 - 1.2) MPa	Up to 0.86 MPa	Up to 0.63 MPa
Static Load Drift	Yes, compensation	Yes, no ompensation	No
Data Transfer	Bluetooth	Wi-Fi	Local storage
3-axis Sensitivity	No	No	Yes
Main Advantage	High max pressure	Fits all shoe sizes	Measures 3D forces
Main Disadvantage	Many insole sizes	Durability issue	Low max pressure

Pain in the lower back is one of the most common health problems today [20]. About a third of all employees in Sweden, during the year 2015, had pain in their lower back every week [21]. The year 2015, 16% of the employed men and 10% of the employed women in Sweden lifted more than 15 kg several times a day [21]. The total amount of lifted weights and lift frequency are moderate to strong risk factors for lower back pain [22]. Measuring of carried weight is thereby of interest. A wearable system is needed to monitor work conditions over a longer time period and preferably also during walking. Sazonova et al. have been able to measure weight using a wearable pedobarography system, but the persons had to stand still during the measurements [17].

The aim of this paper is to (1) present three novel analyses for estimation of weight during walk and (2) to describe the design process of the cost-effective pedobarography system IngVaL, especially regarding the sensor implementation in the insoles and the dynamic calibration of the sensors. The paper will also (3) present the durability of the sensors.

II. System Development

This system development section has three subsections called Selection of Sensor Type, Sensor Implementation and Calibration of the Sensors. There are two versions of the pedobarography research system, the first prototype version and the second version, IngVaL. The design process of the two systems are described. Block diagrams of the first prototype system and the IngVaL system are shown in Figure 1.

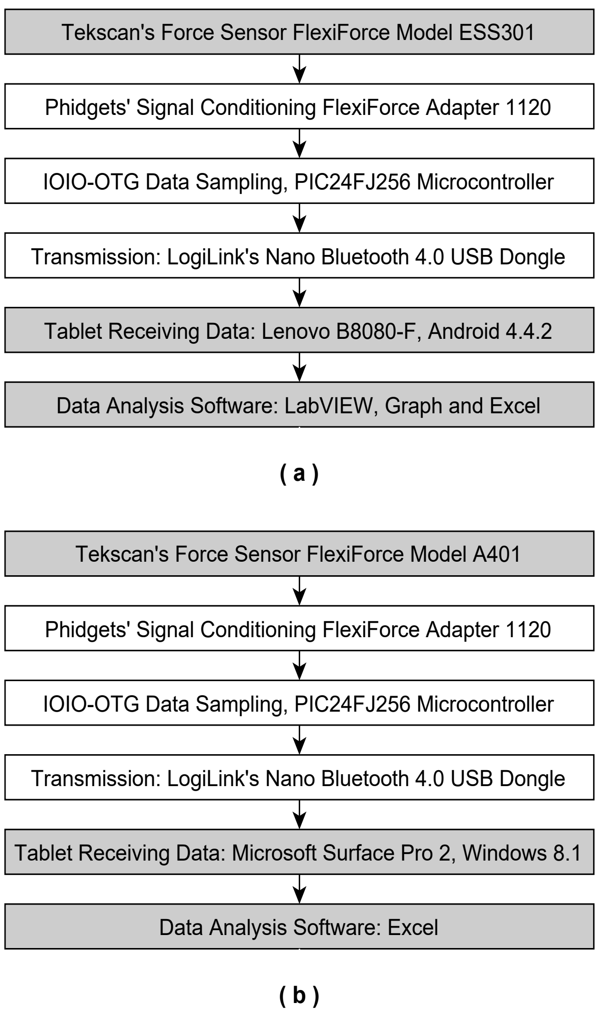

Figure 1. The two system versions; (a) the first version prototype, and (b) IngVaL, the second version of the system that was used in the experiment in this study. Grey highlights system differences.

Motivations for choices in the system design are given in the relevant hardware sub-subsections and shortcomings of the first prototype system are explained to show the evolution of the design. A comparison between the two system versions is shown in Table 2.

Table 2. Comparison between the first prototype version of the system and the IngVaL version.

System Version	Prototype (version 1)	IngVaL (version 2)
Tekscan Sensor Model	ESS301	A401
Sensor Diameter	9.5 mm	25.4 mm
Sensor Thickness	0.2 mm	0.2 mm
Insole Thickness	8 mm	6 mm
Base Insole Material	Polyurethane	Ethylene-Vinyl Acetate (EVA)
Insole Material against Foot	Polyurethane	Leather
Sensor Boundary Protection	No	Yes

Four force sensors in each insole were selected and the signal conditioning was done using the FlexiForce adapter model 1120 (Phidgets Inc., Calgary, Canada). The sampling, with a frequency of 200 Hz, and the Bluetooth transmission were done with an IOIO-OTG (Sparkfun Electronics Inc., Niwot, USA). The IOIO-OTG is based around a PIC24FJ256 microcontroller. A tablet received and stored the collected data. Software for receiving data from the IOIO-OTG is available for Android [23] and for Windows [24]. The whole system was built using commercial off the shelf components.

A. *Selection of Sensor Type*

Researchers have used many different types of force sensors when designing pedobarography systems and all of those sensor types have their advantages and disadvantages. Desirable sensor properties are low sensor thickness, low sensitivity for temperature change, low hysteresis, high linearity, good electrical stability and good durability.

Resistive sensors can be very thin, have good electrical stability and no hysteresis. Drawbacks are non-linearity, drift of the output when the sensor is under static load and they can have problem with durability. Piezoelectric sensors are linear, robust and not sensitive to electromagnetic interference. Main disadvantages are that it can only be used in dynamic applications, the output from the sensor is temperature dependent and it needs to have amplification close to the sensor. Capacitive sensors have a large measurement range and low temperature sensitivity but special care has to be taken to reduce parasitic capacitance when connecting the sensors to the electronics. Another disadvantage is that they have a non-linear output. Optoelectronic sensors are not sensitive to electromagnetic interference but are often thicker than the other types. They can also be used to sense forces in three dimensions but they often cannot differentiate between the three planes.

The durability of the sensors became a problem in the first prototype system. Occasional maximum readings were the first sign of sensors breaking down and thus those sensors had to be replaced and recalibrated.

The model of the force sensing resistors in the prototype version were model ESS301 (made by Tekscan Inc., Boston, USA) with a 9.5 mm diameter of the active sensor area. The motivation for selecting this specific model was that it had good protection against humidity caused by perspiration. Larger forces can be measured when a small sensor area is used since the sensor measures the average of the forces acting on the active sensor area. Non-linearity is not a problem after calibration. Good availability and low thickness (0.2 mm) were two additional advantages. The durability of the sensors became a problem in the first prototype system.

An additional measuring problem with the first prototype system was that the sensor area was too small, which made small foot movements inside the shoe a problem. In the IngVaL system the sensor model was changed to a larger model, A401 (Tekscan Inc., Boston, USA), to reduce that problem. It also made it possible to measure a larger area of the foot. Model A401 of the sensor and the signal conditioning adapter are shown in Figure 2.

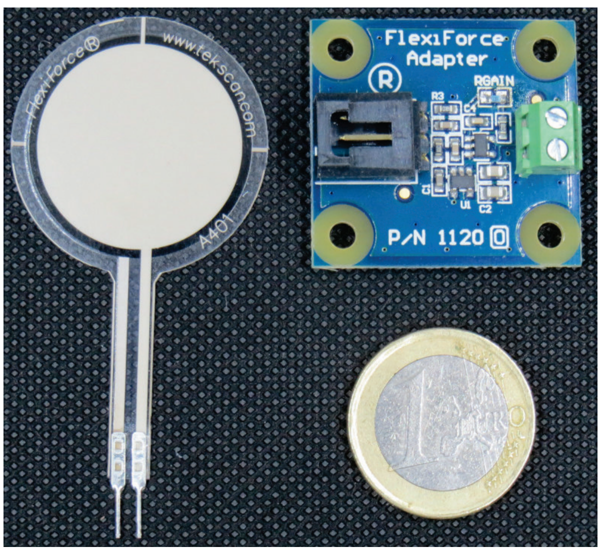

Figure 2. Sensor model A401 by Tekscan Inc., FlexiForce Adapter 1120 by Phidgets Inc. and a 1 euro coin for size comparison.

The A401 model has an active sensor diameter of 25.4 mm. The black circle on the sensor is the boundary of the active sensor area.

B. Sensor Implementation

The primary supporting positions for force interactions between the foot and the insole are the heel and the metatarsal pad (MTP), which is the pad under the forward part of the sole closest to the toes. The end of those bones, closest to the toes, are the metatarsal heads which are the bone structure for the MTP under the foot. The sensors are often placed under the heel, the MTP and the big toe, due to the bone structure of the foot and the possibility to monitor the forces from heel strike to when the foot leaves the ground again [25][26][27]. Sensors were therefore placed under the MTP and the heel, both in the first prototype system and in the IngVaL system. A fourth sensor was added to monitor when the big toe pad is in use since it is the last sensor to be activated during the stance phase (when there is contact with the ground) of the step. The four sensor locations are shown in Figure 3.

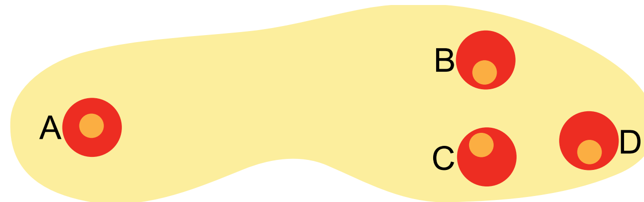

Figure 3. Sensor locations; small circles for the prototype version and big circles for IngVaL. A is the heel sensor, B is the outer and C is the inner metatarsal pad sensor, D is the big toe pad sensor.

Mechanical stress on the boundaries of the active sensor area turned out to be the culprit of the sensor breakdowns in the prototype after investigating the broken sensors. Stress on the boundaries short-circuits the sensors and results in quite rapid breakdown. The first prototype insoles were not comfortable for the user since the interface material against the foot was made of polyurethane. This material is not good at absorbing perspiration but the resulting shoe environment did not affect the sensor functionality.

New versions of the insoles were made for the second version of the system, IngVaL. The sensors were implemented sandwiched between a base of EVA and a protecting cork and leather layer of model 6949 (BNS Bergal, Nico & Solitaire, Vertriebs GmbH, Mainz, Germany). The leather interface, towards the foot, also helped reducing perspiration. Material was removed in the EVA base, under the boundaries of the active sensor areas, to remove any mechanical stress on the boundaries. The two sensors under the MTP were moved slightly more apart in the IngVaL version to better be able to measure forces close to the edges of the insoles.

C. Calibration of the Sensors

The use case for the system has to be taken into account when choosing the amplification to make sure that there is no saturation of the signal when measuring the largest forces. The amplification is changed by replacing a resistance on the Phidgets signal conditioning adapter. The manufacturer (Tekscan Inc., Boston, USA) recommends using at least four different force levels. The calibration function solves the problem with non-linearity. Sensor calibration was done by using five different force levels and repeated five times per force level. The forces were applied dynamically to avoid the problem with static load drift [28]. Polynomials of at least the third order is recommended for force sensing resistors [29]. The calibration functions for the sensors in IngVaL used fourth order polynomials, see Figure 4.

A new calibration station was designed for IngVaL. A button load cell of model CZL204E (Phidgets Inc., Calgary, Canada) was used together with the Phidgets 4-input Bridge model 1046. The calibration forces were applied, by stepping with the heel placed on the top nut, to mimic the dynamic scenario the sensors are used in during walk. Using a static load would have introduced the problem with drift over time for this type of sensor.

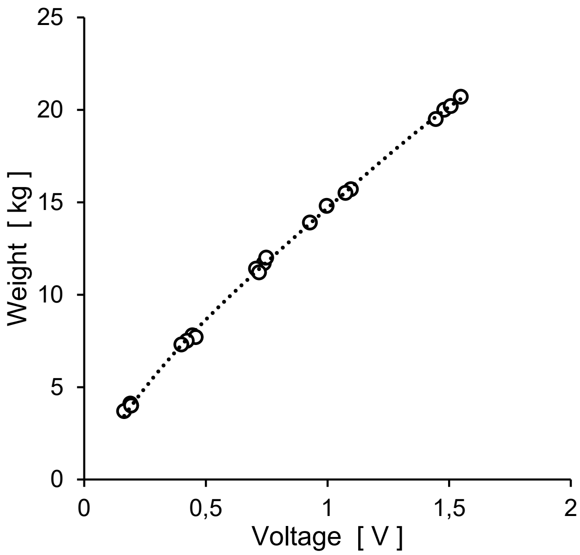

Figure 4. Weight in kilogram versus voltage in volt gives the calibration function (a polynomial of the fourth order). The calibration function for the heel's force sensing resistor is shown as an example.

A steel disk is placed over the sensing peg of the load cell to ensure a correct angle of the applied force. The angle is guaranteed because the steel disk is adapted to fit snugly on the peg of the load cell. An EVA disk is used between the heel sensor and the load cell to protect the sensor surface and to distribute the forces evenly. Metal spacers are added until the sensor almost registers pressure. Calibration forces acts perpendicular to the sensor area due to a minimized vertical displacement. The calibration station is shown in Figure 5.

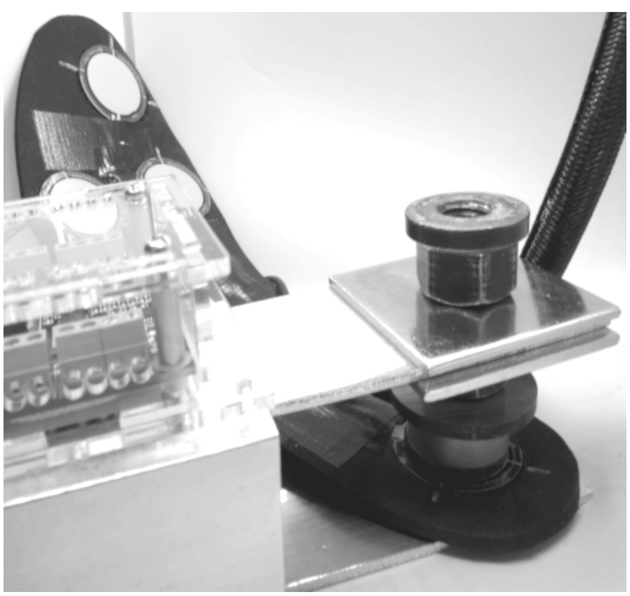

Figure 5. The calibration station used in this study, from floor; aluminium, EVA insole, force sensor, EVA disk, load cell, steel disk, upside down bolt, metal spacers, aluminium, metal spacer, nut.

The calibration function for the heel sensor is given by:

$$w(v) = -3.0588v^4 + 11.897v^3 - 17.719v^2 + 23.574v \quad (1)$$

where w is the average weight [kg] on the active sensor area and v is the voltage [V] sampled by the IOIO-OTG. The calibration terms of the functions for the sensors are shown in Table 3.

Table 3. Calibration functions, 4th order polynomials, for the force sensing resistors in the insole.

Polynomial Term	4th	3rd	2nd	1st
Heel	-3.0588	11.897	-17.719	23.574
Inside MTP	-1.9297	8.7022	-14.909	21.429
Outside MTP	-5.1492	14.801	-16.032	18.914
Big toe	-1.8116	8.4939	-15.074	21.450

III. METHOD

This section contains the experiment and data analysis. The IngVaL version of the system was used in the experiment. The experimental study has a cross-sectional design. Three novel weight estimation analyses are described in the Data Analysis subsection. Ethical approval for the study was granted by the Swedish Ethical Review Authority (diary number 2017/070).

A. Experiment

Inclusion criteria for the fifteen test persons participating in this study were to be healthy, have an EU shoe size of 43 or 44 and to be able to walk 1.0 m/s on a treadmill while carrying up to 20 kg extra weight in a backpack. The first author is shown wearing the backpack and walking on the treadmill in Figure 6.

All test persons are university staff and they had an average weight of 83.9 kg while wearing the backpack without any added weight inside it. The lowest and highest weight, without added weight in the backpack, were 75.2 kg and 110.9 kg respectively. Each test person made five walks, using the same insoles and shoes, with a pseudo-random added extra weight of (10, 20, 0, 15, 5) kg in the backpack. Safety measures were automatic stop of the treadmill (Comfort Track Prime 97690, LifeGear Ltd., Taiwan) if the test person moved away too far from the correct position, and extra padding was placed in the backpack to protect the spine. A health insurance for the test persons was bought from the Insurance department of the Legal, Financial and Administrative Services Agency. Data was recorded during one minute per walk, excluding acceleration and deceleration phases, and a test walk was made first to make sure the test person was comfortable walking on the treadmill and that all sensors were working correctly. An electronic floor scale (model GS 42 BMI, Beurer GmbH, Ulm, Germany) was used to measure the reference weight for each of the five weight configurations with an accuracy of ±0.050 kg.

Figure 6. Walk on treadmill with backpack containing the electronics of the IngVaL system and the extra carried weights.

The durability of the sensors was investigated. All sensors were in good working order during these 75 walks in the experiment. The durability of the sensors was examined further by performing additional walking outdoors.

B. Data Analysis

Three novel analyses methods for estimating weight during walk were evaluated. In analysis method 1, data from the heel sensor was used for estimation of the carried weight while walking. The 200 largest heel sensor values during each one minute walk were averaged. In analysis method 2, data from all sensors were used to estimate weight while walking. All samples from all the sensors during each one minute walk were averaged.

Fifteen test persons and five walks each resulted in 75 of these averages for each of the two analyses methods. For both methods, two averages (for 0 kg and 20 kg) for each test person were used to create a linear equation for individual calibration. The other three averages (5 kg, 10 kg and 15 kg), for each test person, were compared to the linear equation to see how big the error was. Thus, n=45 in the graph in the results section. An overview of the three steps of the data analysis are shown in Figure 7.

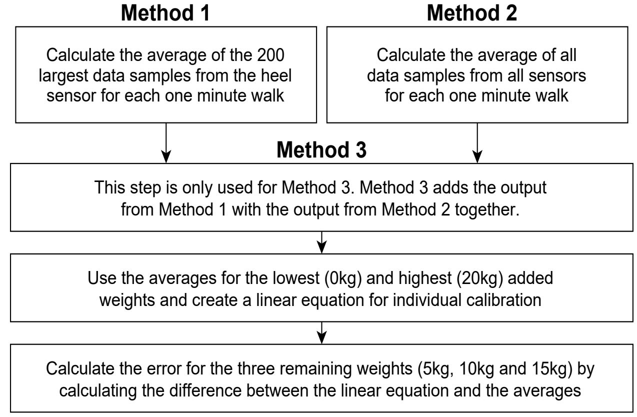

Figure 7. The steps of the data analysis, for estimation of weight during walking.

Method 1 only uses the heel sensor and the other two methods use all sensors. The outputs from method 1 and method 2 are summed together in analysis method 3.

IV. RESULTS

The first analysis method used only the heel sensor for the estimation of weight while walking, and resulted in a RMSE of 11.3 kg. The result is presented in a plot of the reference weight, as measured by the electronic floor scale, versus the weight estimation errors in Figure 8.

There is a tendency for overestimation, in method 1, of the weight with a mean of +2.1 kg. The second analysis method uses data from all the sensors and the mean is this time negative, -3.1 kg. A third method combining method 1 and method 2 seemed promising since the means have opposite signs. The third analysis method simply combines method 1 with method 2 by adding the outputs together before the individual calibration is done. The results of method 2 and method 3 are also shown in Figure 8. The result for method 3 is also shown in Figure 9 for clarity.

A summary of the results (RMSE, standard deviation (SD), mean, and upper and lower 95% confidence intervals) for the three methods are presented in Table 4.

Table 4. Summary of the results of the three methods.

Analysis Method	Method 1	Method 2	Method 3
RMSE	11.4 kg	7.1 kg	6.1 kg
n	45	45	45
SD	11.1 kg	6.5 kg	5.8 kg
Mean	2.1 kg	-3.1 kg	-2.0 kg
Upper 95% CI	23.9 kg	9.6 kg	9.3 kg
Lower 95% CI	-19.7 kg	-15.8 kg	-13.2 kg

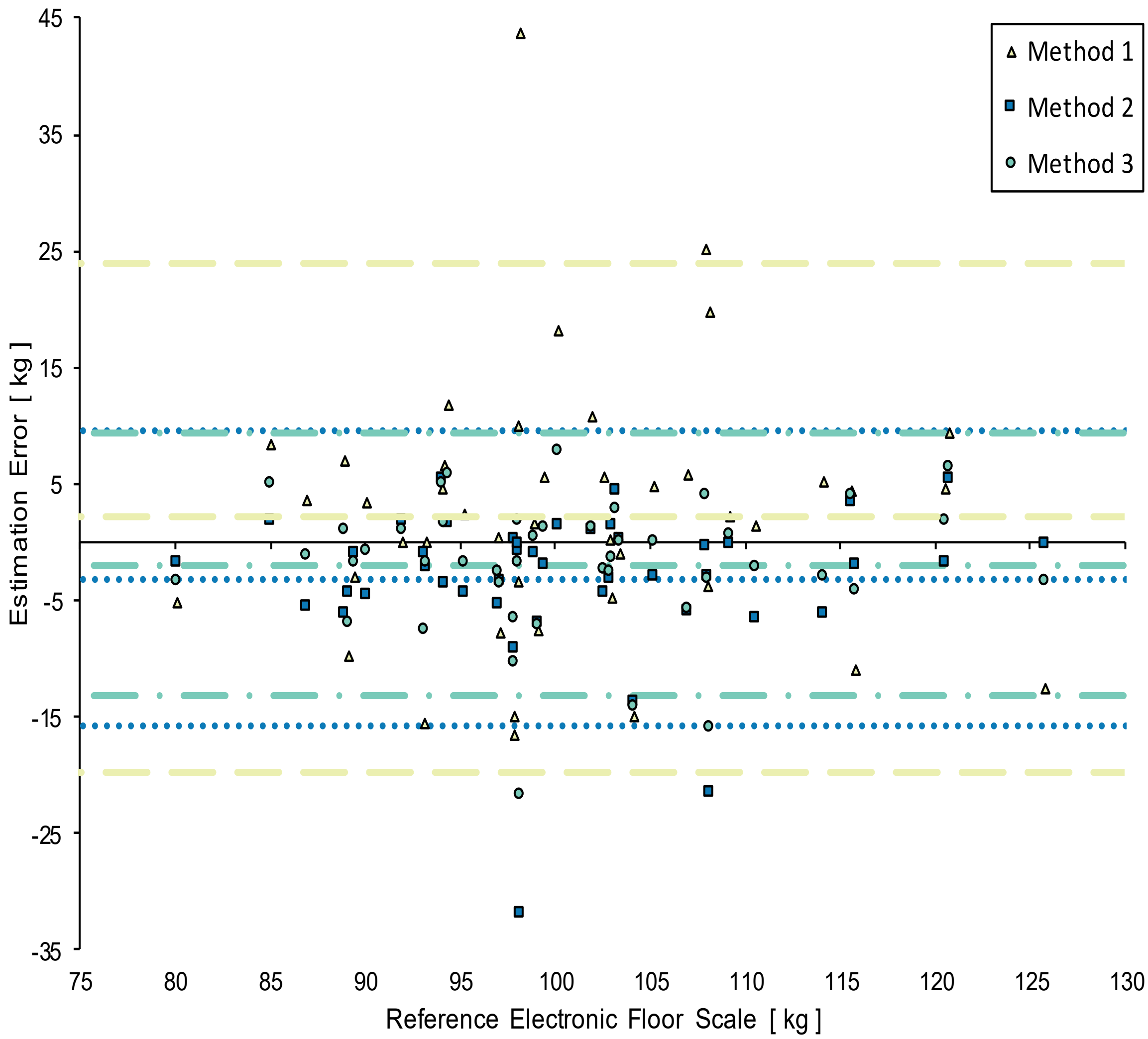

Figure 8. Errors in the estimation of the weight in kilogram versus reference electronic floor scale weight in kilogram for the three methods. Method 1 only uses heel sensor data and the other two methods uses data from all sensors.

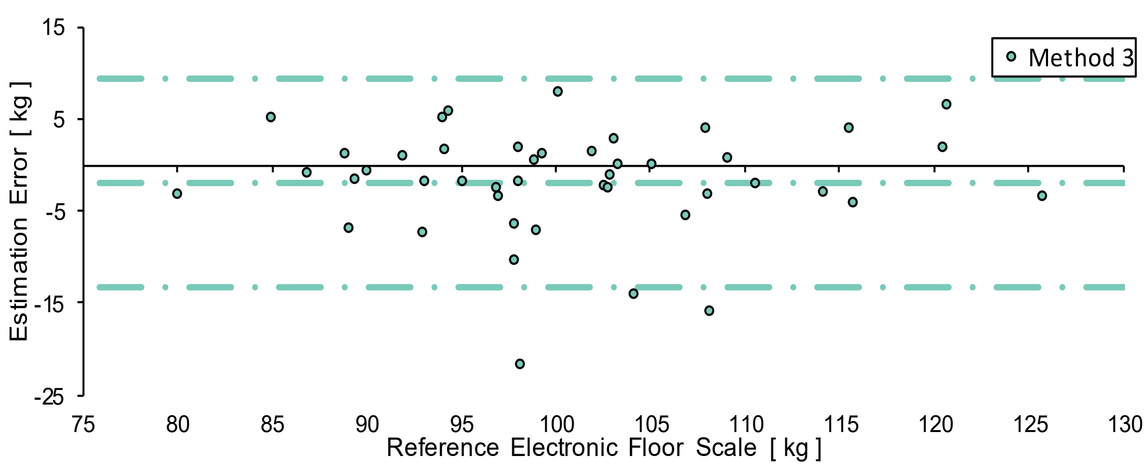

Figure 9. Errors in the estimation of the weight in kilogram versus reference electronic floor scale weight in kilogram for method 3.

Further durability testing was performed by normal walking outdoors. One sensor became unusable after a total of 36000 steps.

V. Discussion

The IngVaL system has shown to have good durability. A novel analysis method resulted in a RMSE of 6.1 kg for the weight measured during walk. The previous paper [1], which this paper is an extension of, reported a RMSE of 13.8 kg.

The earlier work was based on the equipoise method which used three sensors for the analysis and calculated a "neutral" point of balance between the forces on the heel sensor and the forward sensors [1]. One downside with the equipoise method was a very small number of usable samples for the estimation of the weight which made the estimation more uncertain.

Three analyses methods have been presented in this paper. Method 2 and method 3 used all four sensors. Method 2 resulted in a RMSE of 7.1 kg. Method 3 reduced the RMSE to 6.1 kg.

The reason for having as few sensors as possible is to reduce the cost of the system. The cost of portable commercial pedobarography systems often starts at €10000. The component cost for IngVaL would be close to €100 if the number of manufactured systems are reasonably high. Each sensor costs around €10 and this means the novel analyse method that only uses the heel sensor is a clear advantage. Method 1 only used the heel sensor and resulted in a RMSE of 11.3 kg.

The analyses methods have been kept as simple as possible. This approach was chosen in order to be able to have the analysis running on for example a smart phone or tablet in the future.

A recurring challenge, when estimation weight using few sensors, is that all persons have unique bone structures in their feet. Thus, small movements of the foot inside the shoes can result in big changes of the amount of force hitting a sensor. The approach of this paper to counter the problem of feet being unique is to do an individual calibration for each person using measured data from walk without carried weight and also from a walk with a large carried weight to create a linear equation. The larger sensor area in the IngVaL system, compared to the prototype system, also helped with this issue since a larger area of the insole is covered with sensors.

All foot arch types have proportional distribution of the forces over all regions of the foot [30]. A possible source for errors in the weight estimation is that some anatomies of feet show a more increasing contact area, between the foot and the insole, when the person carries weights compared to subjects with normal feet [31].

The most used portable pedobarography system that uses resistive sensors, F-Scan by Tekscan, states that their sensors are durable enough for multiple trials [32]. This way of expressing the durability makes it hard to compare with other systems. Great care was given in the sensor implementation in the IngVaL system to improve the durability of the sensors. The durability testing showed that the first sensor became unusable after 36000 steps. The sensor had moved a bit out of position and this put stress on the boundary of the active sensor area. Possible ways for overcoming this problem is to remove more material under the sensor, and thereby use less of the sensor area, and/or embed the sensor deeper into the insole.

Sazonova et al. estimated weight directly after coming to a standstill after walking, and reported a RMSE of 10.5 kg. Force sensing resistors drifts under static load and this adds to the challenge of measuring correctly when standing still. The drift problem is avoided when measuring during walking due to the dynamic loads on the sensors. Both methods 2 and 3 have a lower RMSE, than the system measuring during standing still.

Since the total amount of lifted weights and lift frequency are moderate to strong risk factors for lower back pain it is of interest to measure heavy working conditions over time. IngVaL can measure during walk and this means that the system is potential candidate for this monitoring. Heavy lifts are of course not the only factor for lower back pain. A sedentary lifestyle or incorrect lifting technique are examples of other factors that also can cause back problems.

VI. Conclusion and Future Work

This paper presented three novel analyses for estimation of weight during walk and described the design process of the pedobarography system. The durability of the sensors can still be improved upon. Another modification of the sensor implementation can be done to stop the sensors from moving in the insole plane and putting stress on the boundary of the active sensor area.

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