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HARDWARE DESIGN OF THE ENERGY EFFICIENT FALL DETECTION DEVICE

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Health issues for elderly people may lead to different injuries obtained during simple activities of daily living. Potentially the most dangerous are unintentional falls that may be critical or even lethal to some patients due to the heavy injury risk. In the project "Wireless Sensor Systems in Telecare Application for Elderly People", we have developed a robust fall detection algorithm for a wearable wireless sensor. To optimise the algorithm for hardware performance and test it in field, we have designed an accelerometer based wireless fall detector. Our main considerations were: a) functionality – so that the algorithm can be applied to the chosen hardware, and b) power efficiency – so that it can run for a very long time. We have picked and tested the parts, built a prototype, optimised the firmware for lowest consumption, tested the performance and measured the consumption parameters. In this paper, we discuss our design choices and present the results of our work.

Keywords: *3-axis accelerometer, electronics, energy efficiency, fall detection, hardware, microcontroller, telecare.*

1. INTRODUCTION

A fall is a dangerous event and may produce a variety of ill effects (such as concussion, fractures, ruptures, internal hemorrhaging), especially for the elderly people. According to [1], [2], it is estimated that over a third of adults older than 65 years fall each year, making it the leading cause of the injury or even death for that age group. As falls generally are uncontrolled and sometimes are caused by specific diseases, the injured person may remain unconscious, incapacitated or otherwise unable to request assistance. Several studies [3], [4] have concluded that elderly people, who choose to live at home, are also willing to accept new technologies to support their sovereignty and safety.

Fall detection methods usually are divided into three categories [5]: wearable sensors, ambience sensors and vision (camera) based sensors. Only wearable systems provide all necessary conditions for live telecare monitoring, where the main concern is the cost of the sensor device and system's setup.

Fall detection device is an accelerometer and/or gyroscope based wearable unit, used in telecare for patient activity monitoring and fall detection. The main function of the fall-detection device is to recognise the abnormal acceleration or orientation, run the fall recognition algorithm on the recently acquired data set, and if the fall is detected – send a distress signal to the monitoring station. Most of the fall detection devices also have a “panic button”, which allows the patient to request assistance in other situations.

The device consists of hardware and firmware parts. In this paper, we focus on the hardware design and implementation, so the firmware will only be briefly discussed when device functionality and power consumption are concerned.

2. RESULTS AND DISCUSSION

2.1. System Overview

The core components of our fall detection device are: LIS2DH – low power consumption accelerometer [6]; ATxmega32C4 – a 16-bit Flash-based microcontroller with a selection of sleep modes and consumption reduction settings [7], capable of running the fall detection algorithm; nRF24L01+ [8] module – a short range 2.4GHz RF communication module; and MAX1724 – DC/DC step-up converter with low quiescent current and high efficiency at light load [9].

It should be mentioned that we chose ATxmega32C4 both for its characteristics (1-2uA consumption in sleep mode, acceptable voltage range, pin count, TQFP and smaller VQFN package availability, acceptable price – 2.45 € w/o VAT, built-in USB module for debug data output, and 32KB program memory) and our experience in working with Atmel devices and their integrated development environment (IDE). The LIS2DH was chosen because of its low power consumption in low frequency modes, small LGA-12 footprint and wide range of sensitivity, resolution and frequency settings, which came in handy in the fall detection algorithm development phase. The RF module was chosen mainly because it was a compact low-budget solution and allowed for 2Mbps debug data transmission, while having relatively low power consumption. The 2.4GHz frequency is suitable for patch and chip antennas, making the module small enough (18x12mm) for our prototype. The device schematic may be observed in Fig. 1.

2.2. Fall Detection Algorithm

Figure 2 shows the device behaviour in the active mode when triggered by a free-fall event. Before the interrupt, the accelerometer (ACC) gathers high-resolution (12b) data in internal “first in, first out” buffer (FIFO) at 25Hz frequency, while constantly checking for a free-fall condition using a simple built-in state machine. When the ACC recognises the free-fall event, it generates an interrupt to wake up the microcontroller unit (MCU). The MCU then becomes active, acquires last 32 samples from the ACC to calculate the orientation, configures the ACC to a higher

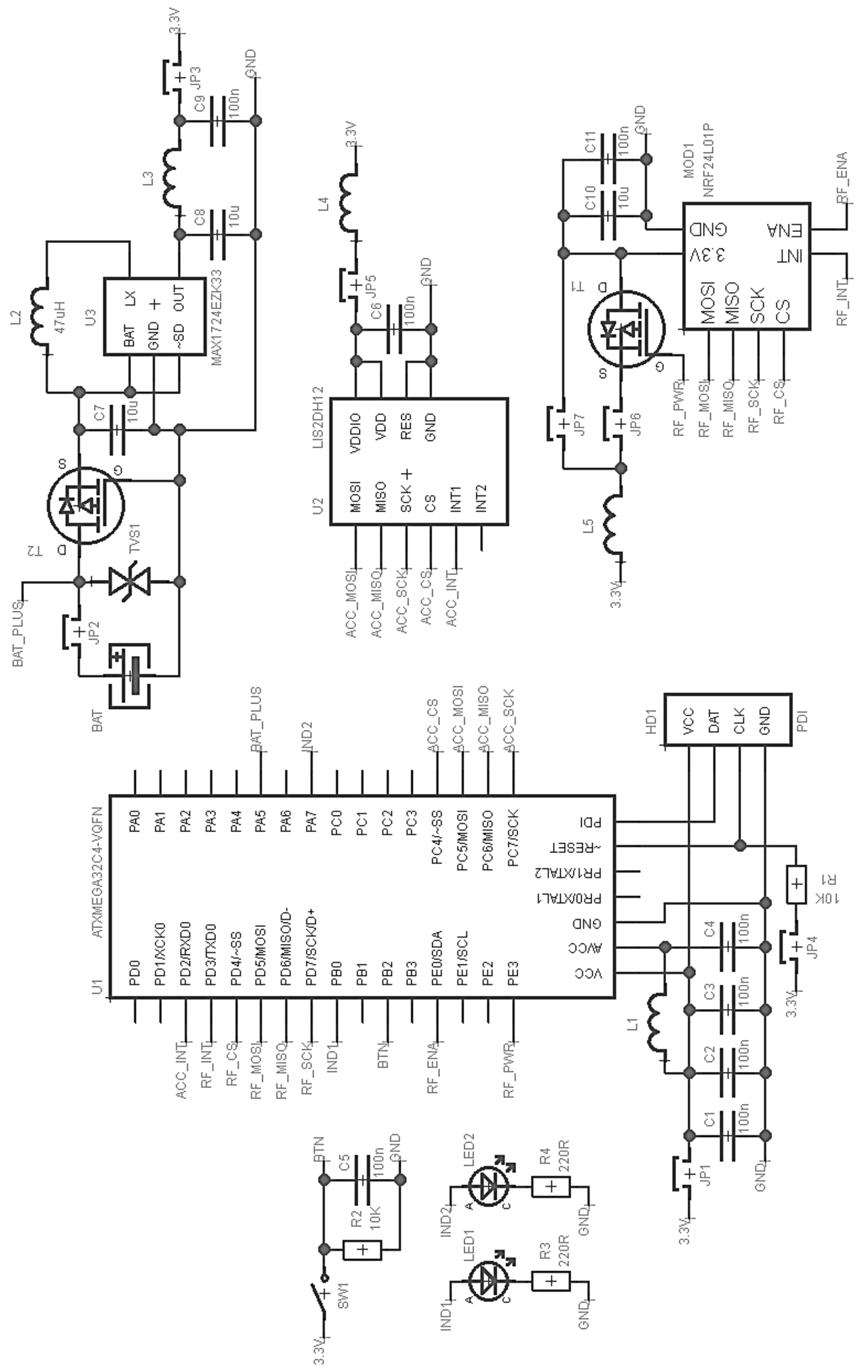

Fig. 1. Device schematic.

frequency (100Hz) and returns to power-save mode for 320ms to conserve the energy. When ACC 32 level FIFO is full, the MCU wakes up to acquire the samples and calculate Teager energy operator (TEO). This cycle is normally repeated four times in order to obtain >1 s (1.28 s) data set. After that, if TEO exceeds the threshold and impact is detected, orientation check is performed. If device orientation has changed by more than 60 degrees, the event is regarded as the "true-positive" – MCU powers up the RF module, sends the distress signal and waits for acknowledgement from a base station. When acknowledgement is received, the MCU returns to sleep. If the acknowledgement is not received, the device will keep sending messages trying to contact the station. If TEO has not tipped the threshold or the orientation has not changed, the event is considered a "false-positive" response, and MCU goes back to sleep [10].

Notice that the algorithm is designed in such a way that a continuous free-fall condition will reset the data acquisition and TEO calculation cycle. This may result in longer active cycles in some cases, which will be discussed later.

Another point to mention is the so-called "panic button", which is simply a tact switch located in the front panel, which wakes up the MCU to send a manually triggered distress signal to the base station.

The two light-emitting diodes (LED) on the front panel are used for indication. The green LED lights up when the device is trying to contact the base station. It is necessary in order for user to know that the device is operational and trying to make contact with the station. The amber LED lights up when the acknowledgement from the base station was received, so that a user would know that his distress signal was heard and the help was on its way.

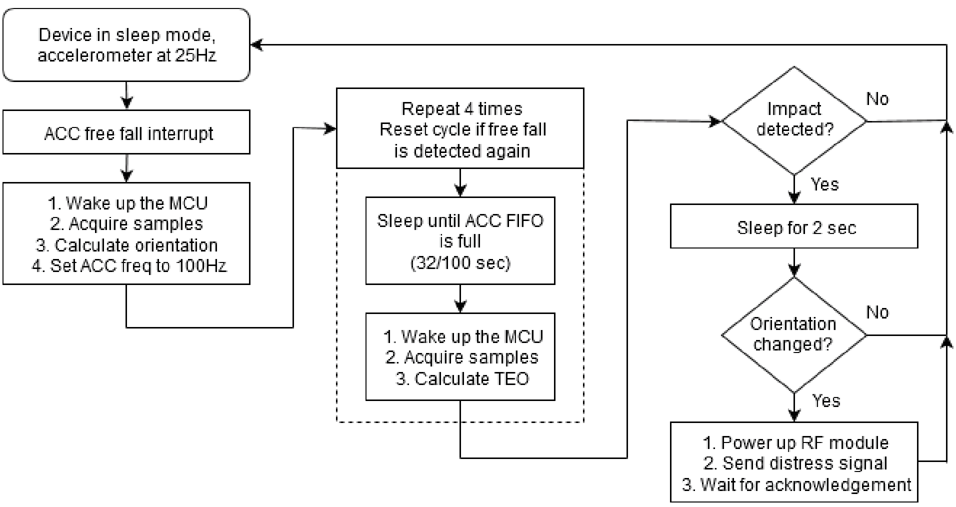

Fig. 2. Fall detection algorithm.

2.3. The Power Source

There is a number of trade-offs one must consider when designing a power circuit for a wearable device: rechargeable or non-rechargeable battery, 3.0-3.6V or 1.4–1.6 V, single or dual cell solution, cell chemistry, size, weight, energy density, availability and price range must be considered.

We have decided to use alkaline instead of lithium-based cells. Though lithium cells have high energy density and the output voltage is just right for 1.7– 3.6 V digital circuitry, we have had major safety concerns about them. Since the device is wearable, it might prove inconvenient if the cell caught fire or exploded due to overheat when accidentally pressed against a hot surface, or due to internal discharge caused by protection circuit malfunction or damage caused by electrostatic discharge (ESD).

The second decision was to use non-rechargeable batteries. One of the main reasons is that at the moment (Sept. 2015) non-rechargeable alkaline batteries provided twice the power of rechargeable Nickel Metal Hydride (NiMH) batteries for one fourth of the price. VARTA non-rechargeable alkaline 1.5 V AA cell provides 2.85 Ah at the cost of approximately 1.60 € (including VAT), while VARTA rechargeable NiMH 1.2 V AA cell provides only 1.6 Ah at the cost of approximately 6.70 € (including VAT). Non-rechargeable alkaline batteries are readily available and may be discharged down to 0.8 V (limited by a DC/DC converter), while NiMH have a high self-discharge rate and may only be discharged down to about 1.0 V to prevent battery damage. Rechargeable battery based solution would also require over-discharge protection, an external battery charger, or additional recharging circuitry and a DC power source. Our preliminary average current consumption estimation was average 100 uA at 3.3 V, with 65 % upconverter efficiency, so life expectancy (not including high NiMH cell self-discharge rate and alkaline cell deep discharge option) of the alkaline battery might roughly be about 350 days, while NiMH should last almost 160 days. This rough estimation was necessary to assess the need for device maintenance and viability of both solutions. The actual battery discharge is monitored by MCU and reported regularly during operation. We have decided that using non-rechargeables would simplify the circuitry and therefore allow for easier routing, minimise printed circuit board (PCB) size, reduce component costs, and reduce device maintenance frequency and expenses associated with it.

Using multiple cells for higher voltage does not seem to be a viable option, since even a single 24 g AA cell (by the authors' personal account) is almost too heavy for a wearable sensor. It is for these reasons that we have decided to focus our design on a single alkaline AA or AAA cell as a power source. One thing worth mentioning is that because of said integrated circuit (IC) supply voltage range (1.7–3.6 V), it is impossible to power the device with a single alkaline cell (1.5–1.6 V when full). Powering from 3.0–3.6 V lithium cell is possible if precaution measures are in place to limit the voltages exceeding the IC's maximum. This approach may be cheaper, but may not be an effective solution. Battery output remains above or at the rated voltage for the most of the operational time, and digital circuitry consumption almost does not depend on supply voltage, so converting the 3.6 V battery output to 3.0 V with 87–90 % efficiency would still provide a power boost. The viability of this approach highly depends on battery and LDO choice and requires additional research and testing to give a definitive answer, but is out of the scope of this topic, since we have already settled on a single alkaline cell.

When a device has no external power source and only relies on a battery, the current consumption becomes one of the primary issues. Many modern ICs have a range of power reduction settings and modes, which disable unused features and parts of internal circuitry, thus reducing power consumption. Further in the text we differentiate between "sleep" and "active" modes. Active mode is when all calculations, interfacing and data exchange take place. In the sleep mode, the accelerometer collects data and checks for the free-fall condition at the minimal frequency allowed by the algorithm, while the MCU is in the power-save mode.

2.4. DC/DC Converter

The next step in designing the device was to choose the suitable DC/DC converter for the application. We have estimated that in the sleep mode our design should consume no more than 50 uA, while in the active mode, the average consumption might be about 10 mA, lasting 0.5 s per each false-positive wake-up. The false-positive wake-up is an event, when MCU becomes active, acquires and analyses accelerometer data, but the algorithm concludes that no actual fall has happened, and, therefore, no distress message will be sent.

We have done a test run that resulted in up to 60 false-positive wake-ups per hour for a very active and lively person (driving a car, jumping, walking, sitting down and getting up, riding elevator up and down, walking up and down the stairs, etc.). A simple calculation shows that if the person sleeps 8 h per day and maintains this high level of activity during all other hours (which is not very likely), the active mode current consumption (at DC/DC converter output) would be 4800 mAs, while the sleep mode would consume 4320 mAs. For an actual ill or elderly person, the passive consumption would therefore always dominate over active, so the primary concern for DC/DC converter should be conversion efficiency at light loads.

Based on availability, cost and datasheet parameters, such as high efficiency, low quiescent current, and input voltage range, we have selected four potential step-up converters from well-recognised manufacturers: MAX1724EZK33+T, NCP1400ASN27T1G, TPS61221DCKT, and LTC3525ESC6-3.3. We have tested them for conversion efficiency at 300 R, 3 K, 30 K, 300 K loads, with 10 uH, 15 uH, 22 uH, 47 uH inductors, at 0.7 V, 1.0 V, 1.6 V input voltages. It should be mentioned that with NCP1400A and LTC3525, only inductors fitting in manufacturer specified range were tested. Another thing to mention – NCP1400A output voltage is 2.7 V, while others supply 3.3 V. In most tested cases, higher external inductance increases converter efficiency. The device in the sleep mode was represented by 300 K–30 K load range, and 300 R – in the active mode. Characteristics of MAX1722 with 47 uH (66–85 % efficiency in sleep, 87–96 % when active) were best suited for the design of the device in question.

2.5. Energy Consumption Reduction and Measurements

The central part of the design is the accelerometer. Upon MCU wake-up, the algorithm requires 1 second long data backlog with the highest possible time resolution, which can only be achieved by configuring the device in 25Hz mode with output to internal 32-level FIFO. Datasheet states that in high-resolution 25 Hz mode LIS2DH should consume only about 6 uA, which is consistent with our prototype measurements [6].

Another crucial part of the design is the microcontroller. Datasheet states that in power-down and power-save modes (at 25 °C) ATxmega32C4 MCU with 3 V power source should consume less than 1 uA, if using external clock and all power saving options are enabled [7]. The device should report battery condition to the base station at least few times a day, so it is impossible to use the least consuming "power-down" mode. In this mode, only asynchronous interrupts can wake up the controller. All internal circuitry, including clock sources, timers and counters, necessary to wake up the MCU to measure and report the battery status, is disabled. Power-save mode consumes more power, but allows keeping operational the circuitry necessary for a timed wake-up. Our measured results for power-down: 0.44 uA, power-save: 1.37 uA, standby: 77 uA, idle: 282 uA, active 848 uA.

3. TEST DATA AND BATTERY LIFE EXPECTANCY

Table 1 shows sleep and active mode current consumption measured in the device prototype. Battery consumption was measured at 1.56 V.

Table 1

Measured Device Power Consumption (% – Mode Efficiency)

Parameter	Consumption	Parameter	Consumption
Accelerometer sleep current	5.9 uA	False-positive consumption	3714 ms at average battery drain 386 uA (2236 uWs, 87 %)
Microcontroller sleep current	1.4 uA	Additional data read consumption	320 ms at average battery drain 925 uA (462 uWs, 82 %)
RF module sleep current	0.1 uA	RF message sent	44 ms at average battery drain 4.52 mA (310 uWs, 90 %)
Reset pin pull-up current	0.1 uA	RF message received	12.8 ms at average battery drain 4.28 mA (85 uWs, 92 %)
Converter output current at sleep	7.5 uA at 3.3 V (24.75 uW)	Single LED current	4-5 mA, 9.6–12.2 mA (15–19 mW) battery drain per second
Battery output current at sleep	24 uA at 1.56 V (37.44 uW, 66 %)	"Panic button" current	0.33 mA, 0.84 mA (1.3 mW) battery drain per second

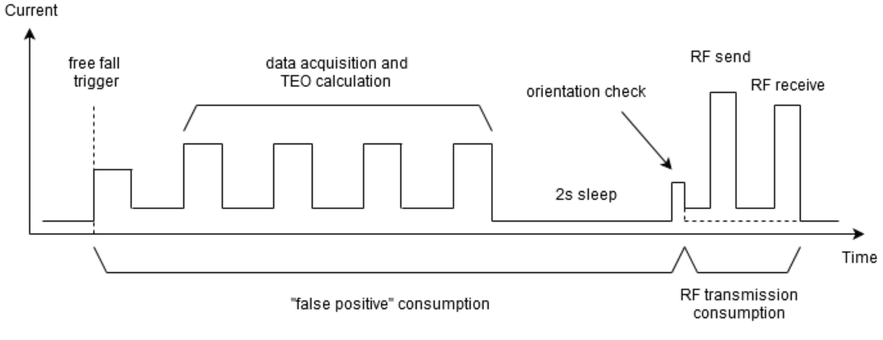

Fig.3. Device active mode current consumption explanation.

Figure 3 shows the current consumption when the device is in the active mode. The false-positive consumption with a single free-fall trigger drains 2236 uWs during approximately 3720 ms. The true-positive cases with RF transmission will happen relatively rarely and can be neglected. It has previously been mentioned that in data acquisition and TEO calculation phase the algorithm can be restarted and active consumption may therefore increase. Our tests show that during everyday activities this usually happens when walking down the stairs or taking an elevator. For example, we have detected up to 41 additional consecutive free-fall events while riding the elevator down for about 13–15 seconds. Since elevator constantly moves, the cycle restarts after the first data read cycle. This results in 21.2 mWs battery drain. Fast walk down the stairs results in 1 to 7 consecutive triggers per stair (5 on average, depends on the walking rate). Due to nature of a person's movement, we are assuming that an average of 2.5 ACC data read cycles pass before restarting the algorithm, resulting in 6.9 mWs battery consumption. Other actions normally produce only a single free-fall trigger and only consume 2.2 mWs. The sleep mode consumption is 37.44 uW, resulting in 0.9 mWh daily battery drain.

A very lively person could drain the battery for up to 0.78 mWh+0.9 mWh=1.68 mWh per day. A less active elderly person would drain the battery for 0.07 mWh+0.9 mWh=0.97 mWh daily. Comparison of active and passive power consumption proves our previously made assumption that light load (sleep mode) performance is paramount for this design.

The previously mentioned VARTA Industrial alkaline battery is rated for 3.3 Wh worth of energy at 43 Ohm load with 0.9 V end voltage [11]. Compensating 25 % for DC/DC converter's decreased efficiency at lower input voltages and battery self-discharge, this should result in 4 to 7 years of operation, depending on person's activity and not including true-fall event consumption and daily battery condition reports. The number is impressive, but it should be noted that only after extensive device testing in real operating conditions and obtaining the normal and abnormal event statistics, it will be possible to calculate more realistic battery life expectancy.

4. CONCLUSION

The main goal of the present research – the design of an energy efficient fall detection device – was successfully completed. The prototype was built and tested for functionality. There is space for improvement though:

1. The main power consumer in the sleep mode is the accelerometer. It is drawing almost 6 uA of current. Other accelerometers with comparable functionality and resolution may be considered for replacement.
2. A way to reduce the power consumption is to pick a converter with lower output voltage and comparable (50–70 %) output efficiency. If a better converter for 1.71 V output is used, the battery life may be extended significantly.
3. A buzzer might prove useful to provide additional audio feedback from a device to a user.

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ENERGOEFEKTĪVAS KRITIENA NOTEIKŠANAS IERĪCES APARATŪRAS IZSTRĀDE

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Kopsavilkums

Kritiena laikā vecāka gadu gājuma cilvēkiem ir liela iespēja gūt nopietnu traumu. Šādā gadījumā ir svarīgi laicīgi izsaukt palīdzību, bet vienmēr pastāv iespēja, ka cilvēks atradīsies bezsamaņā un pats to izdarīt nespēs. Šim nolūkam pielieto kritiena detektēšanas sensorus, kuri automātiski informē par kritiena notikumu radiniekus vai sociālo pakalpojumu sniedzējus. Šādas sensora ierīces izstrādes galvenie kritēriji ir algoritma drošība un jaudas patēriņš. Par otras kārtas kritērijiem var uzskatīt ierīces izmēru, svaru un cenu. Šajā rakstā mēs apskatam algoritma darbības principu, argumentējam komponenšu izvēli, pieņemtos lēmumus, apskatam aparatūras un programmnodrošinājuma patēriņa minimizēšanas iespējas, piedāvājam uzmanībai veiktos mērījumus. Balstoties uz testu sērijā iegūtajiem algoritma aktivācijas rezultātiem un izmērīto ierīces patēriņu aktīvajā un gulēšanas režīmā, mēs veicam teorētiskā patēriņa aprēķinus un prognozējām baterijas dzīves ilgumu.

21.09.2015.