Electronics and
Computer Science

UNIVERSITY OF
Southampton

Technical Writing:
Experimental Design

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The Set-Up Section!

This is the catch-all section for information related to the setting up of your work. It can be Experimental Design, Design, Simulation, Problem definition...

As a result, it very much depends on the exact nature of your work.

The final version of this section should be written after Results and Discussion sections, although it can be co-written as might be appropriate with an iterative Design and Build project (which has cycles of Design/Build/Test/Evaluate…)

What does it contain?

This is the first section that contains some of your contributions. It also contains the essential information that would allow someone to replicate your work. It would also contain citations to reference material.

It therefore presents key ideas (and information) for measurement/build/ simulation, such as:

- experimental or simulation setup:
  - equipment used (capabilities, manufacturer, source)
  - software used (manufacturer, version, set up, environment)
  - materials (supplier, batch, purity)
- your specific setup:
  - physical arrangement
  - connectivity
  - modifications
  - Constraints
  - PICTURES

What does it contain?

- methods for measurement or test
  - Ranges
  - Parameters
  - Reduction methods
  - Error sources
- **OR** design aspects:
  - requirements
  - features
  - constraints
- **OR** simulation methods,
  - problem space and model,
  - relationship with physical system,
  - constraints

What does it contain?

- methods of analysis or test criteria
  - what specific techniques or approaches
  - data interpretation methods
  - statistical analysis of your data
  - Methods for estimation of uncertainty
  - **How you calculate values for your performance indicators**
  - **What these will show about your measurement/device/simulation**

Last and most important: NO RESULTS!

Example

instrumented crutch would be to assist in both training and long-term monitoring of a patient's PWB programme. The secondary aim was to infer information about how the patient is using the crutch. The system should augment a standard low-cost pair of forearm crutches, thus dictating the use of off-the-shelf components. The low-cost requirement also typically infers a low level of accuracy. However, in this application the required level of accuracy was identified to be <5%, a level which would provide a significant patient benefit over existing methods and systems. Any improvement in this accuracy is likely to be unnecessary, as alternative measurement errors are likely to become predominant as a result of the force distribution through the bones and soft tissues. It was also stressed however that the system needed to be easy to use and simple to configure in order for it to achieve acceptance by both patients and clinicians alike. Figure 1 shows the uses of the crutch, including real-time observation of data by the clinician (to train patients how to use the crutch) and to provide real-time biofeedback to the patient (encouraging them to consistently put the recommended weight through the limb). Clearly, a

3. The instrumented crutch

This section describes the various hardware and software components of the instrumented crutches. One crutch acts as a master; the master crutch receives data from the other crutch, referred to as the slave, and processes the data to provide biofeedback to the patient via an audible buzzer.

3.1. Hardware architecture

The hardware consists of low-cost sensors integrated into the crutch for measuring $|F_c|$, θ, φ and d, and a low power embedded microcontroller and radio transceiver to sample, process and communicate data between the crutches and back to the host computer. The crutches were developed using commercial off-the-shelf (COTS) components in order to minimize the potential cost. As is visible in figure 2, the crutch hardware is currently in prototype; it is envisaged that all hardware (batteries, cabling, sensors and circuitry) can all

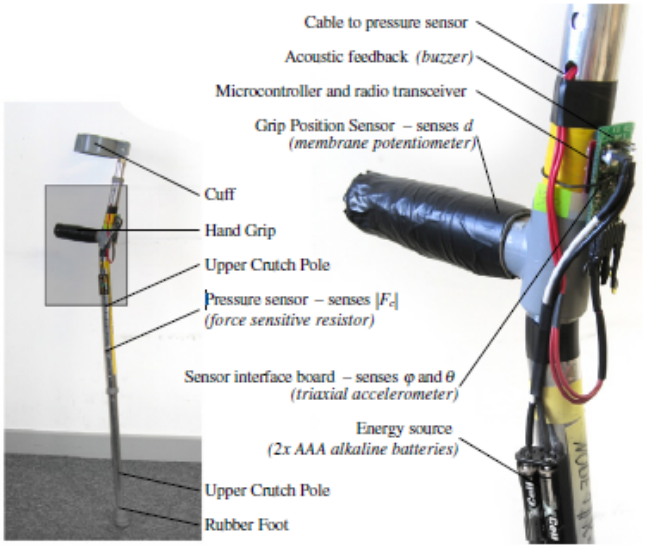

Figure 2. One of the instrumented crutches showing the locations of the major components and parts.

is not enabled when the crutches are turned on. To enable this mode of communication, a micro-switch on each crutch must be depressed while they are powered on. The radio transceiver on the slave node is duty cycled, and enters a sleep state whenever it is not communicating with the master crutch. The crutches are powered from two 1175 mA h AAA-sized batteries mounted on the crutch shaft, with the aim of sustaining operation for at least 24 h.

A dedicated sensor interface PCB is connected to the rear of the ez430-rf2500 via general purpose IO lines, and contains interface circuitry for sampling the various sensors on the crutch. Data from the sensors are sampled via the microcontroller's onboard 10-bit analogue-to-digital convertor (ADC) and operational amplifiers. A continuous loop operates in the microcontroller's embedded software, in which data are sampled from each sensor (the FSR, potentiometer and three individual axis of the accelerometer), processed accordingly and, in the case of the slave crutch, communicated to the master. Additionally, if enabled, data are also communicated to the host computer). This sequence of operations repeats at a frequency of 38 Hz.

Example

be housed within the crutch pole, with the possible exception of the radio antenna.

The centre of the instrumented crutch is a Texas Instruments ez430-rf2500 wireless development tool [18]. This device contains both a 16-bit MSP430 low-power microcontroller and a CC2500 2.4 GHz ultra-low-power radio transceiver. The host computer (which can receive data from both crutches and display it to a clinician in real time) uses an identical ez430-rf2500 wireless development tool, connected via a USB port. The CC2500 radio transceiver communicates in the 2.4 GHz International Industrial, Scientific and Medical (ISM) band, with a radiated output power of 0 dBm. Communication is performed using the proprietary SimpliciTI [19] low-power network protocol from Texas Instruments. Raw data from the sensors are broadcast from the slave crutch and received by the master crutch and, when enabled, both master and slave crutches independently establish a connection with the host computer to communicate data. Due to a requirement for fast sampling, all data are communicated without acknowledgments.

To enable the crutches to be used out of the hospital and in a home environment, it is essential that they can operate for prolonged periods of time through use of their onboard batteries. Hence, energy efficient operation of the crutches is of significant importance. A commonly used method of reducing the average current consumption is to duty cycle the microcontroller and/or radio transceiver into low-power sleep states. As the crutch is continuously sampling and processing data from its sensors, it is not possible to put the microcontroller into a sleep state (unless the sampling rate was reduced to permit this). Further, using the existing communication scheme it is also not possible to duty cycle the radio transceiver of the master crutch, as this continuously listens for packets transmitted by the slave crutch. The crutches can be turned off when not in use by detaching the power connector or removing the batteries. To further reduce the power consumption, communication to the host computer

The primary purpose of the instrumented crutch is to measure the force applied through its axis, i.e. the ground reaction force (GRF) or F_c in figure 1. In order to obtain a measurement of the GRF, both force sensitive resistors (FSRs) and strain gauges were considered. An FSR was selected as the most suitable method due to its low-cost and ease of retrofitting (strain gauges would need to measure compression and tension forces in the sides of the crutch pole as opposed to the direct compression through the crutch axis; this would require a complex adhesion procedure and protective housing in order to practically assemble it). The magnitude of the GRF, $|F_c|$, is measured using a FlexiForce® FSR [20] mounted inside the crutch pole (the location of which is shown in figure 3). This location was selected, as opposed to mounting the FSR at the bottom of the crutch on the inside of the rubber foot, to minimize possible adverse issues caused by friction or compression of the foot. The FSR was conditioned assuming a maximum force of 1 kN (equivalent to a patient's weight of $\approx$100 kg) and the force–conductance relationship linearly characterized over this range with a relative error of $\leqslant$10%. Naturally, issues including repeatability and drift are to be experienced, but are manageable through regular zeroing and self-calibration (discussed later) and the relaxed accuracy requirement inferred by the application.

The crutch tilt, $\hat{F}_c$, is measured using an STmicro LIS3LV02DL MEMS tri-axial accelerometer [21] and calculated with the assumption that 'passive' accelerations (accelerations due to gravity) are predominant over 'dynamic' accelerations (accelerations due to a rapid change in velocity, such as shocks, movement and vibrations). In reality, the presence of dynamic accelerations gives rise to occasional acceptable errors, and can be minimized through the inclusion of a low-pass filter. The magnitude of dynamic accelerations is also likely to increase with a faster walking speed, and hence increases the error in calculating $\hat{F}_c$. However, in a clinical environment patients are taught to use crutches at a safe speed, which is anecdotally slower than normal walking

Example

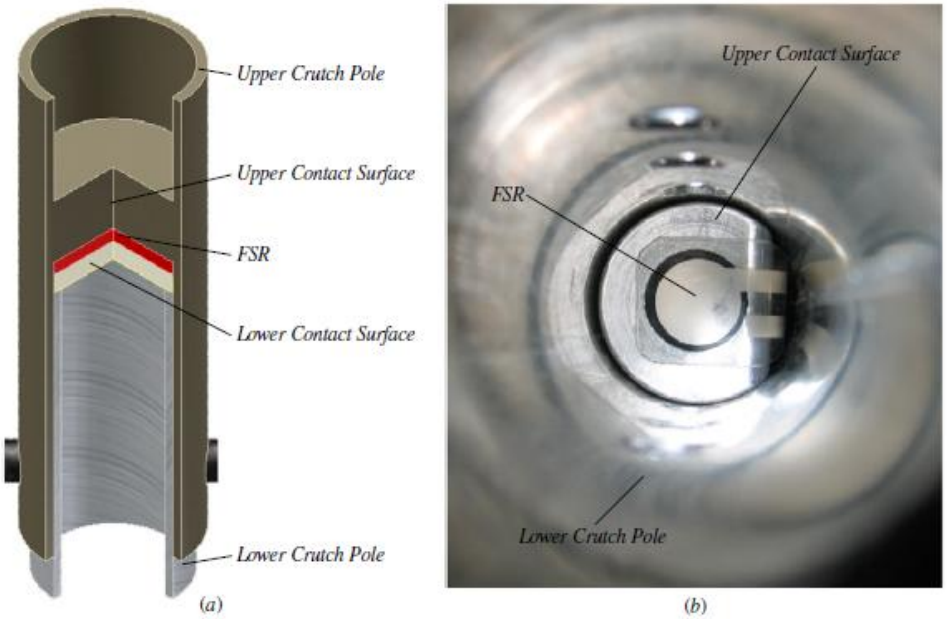

Figure 3. (*a*) A drawing showing the location of the FSR (measuring F_c) inside the main crutch pole and (*b*) a photograph of the FSR located inside the crutch (taken looking up the main crutch pole before the lower crutch pole is inserted).

speed and usually quite deliberate and careful. As only passive accelerations are expected, the accelerometer is configured in the low-g mode (± 2 g). In order to calibrate the accelerometer (required for calculating the tilt angles) the system's GUI can be used to enter values for each component at -1, 0 and 1 g (obtained by rotating the crutch to specific orientations).

Finally, the distance at which the grip force is applied to the handle, d, is measured using a SpectraSymbol 10 kΩ, 100 mm rectilinear membrane potentiometer [22] mounted inside the hand grip. The sensor is located underneath the rubberized cover to protect it from adverse impacts and damage, and the plastic inner-bar was coated in epoxy composite and smoothed to remove any unevenness in the mounting surface. The signal is converted into a voltage using an inverting amplifier, from which an estimate of the position of the hand on the grip can be obtained. Using a foil potentiometer, the value obtained for d provides an indication of incorrect grip position, through the identification of the force applied closest to the crutch axis. The distance, d, and the crutch tilt, $\hat{F}_c$, are not used to estimate the weight applied through the affected limb, but provide supplementary

software on the ez430-rf2500s is programmed in C using the Texas Instruments Code Composer Essentials integrated development environment.

The slave crutch samples its various sensors and transmits these raw data, unprocessed from the output of the ADC, to the master crutch. The master crutch receives these data and, by fusing it with samples taken from its own sensors, calculates the level of PWB (providing biofeedback when necessary). If enabled, both crutches also independently transmit their raw sensor data to the host computer.

3.2.1. Embedded processing. Having sampled the data from its own sensors and receiving data from the slave crutch's sensors, the master crutch estimates the weight through the affected limb in order to provide biofeedback. To understand the reasoning behind this algorithm, consider a typical PWB gait cycle as depicted by figure 5. If the maximum force through the patient's affected limb is identified, this considers the impact forces (heal contact and propulsion) as being relevant to the process of PWB. While this could be seen to

Example

information on crutch usage (requested by clinicians during our initial consultations).

3.2. *Software architecture*

The software system (shown in figure 4) of the instrumented crutches consists of the embedded software on the master and slave crutches, the embedded software on the ez430-rf2500 connected to the host computer via USB (this is not discussed further as its operation is only to receive data from the crutches and pass it into a virtual serial port on the host computer) and the LABView graphical user interface for recording and visualizing data from the crutches in real time. The embedded

be of importance (as these forces are being translated through the affected limb), the decision was taken in this research that it was not the force that was of interest. This is because PWB is prescribed as a fraction of the patient's body weight, and hence is normalized to 100% as 'normal' gait. Hence, impact forces are part of a patient's normal gait pattern, and the instrumented crutch needs to identify where the prescribed PWB limit applies. For this reason, the algorithm identifies and uses the force in the midstance phase (shown in figure 5), as it is at this point that the full extent of the patient's weight is applied through the crutches and the affected limb (i.e. the affected limb is in the midstance phase, and the healthy limb is most likely to be off the ground and not under load). It is also

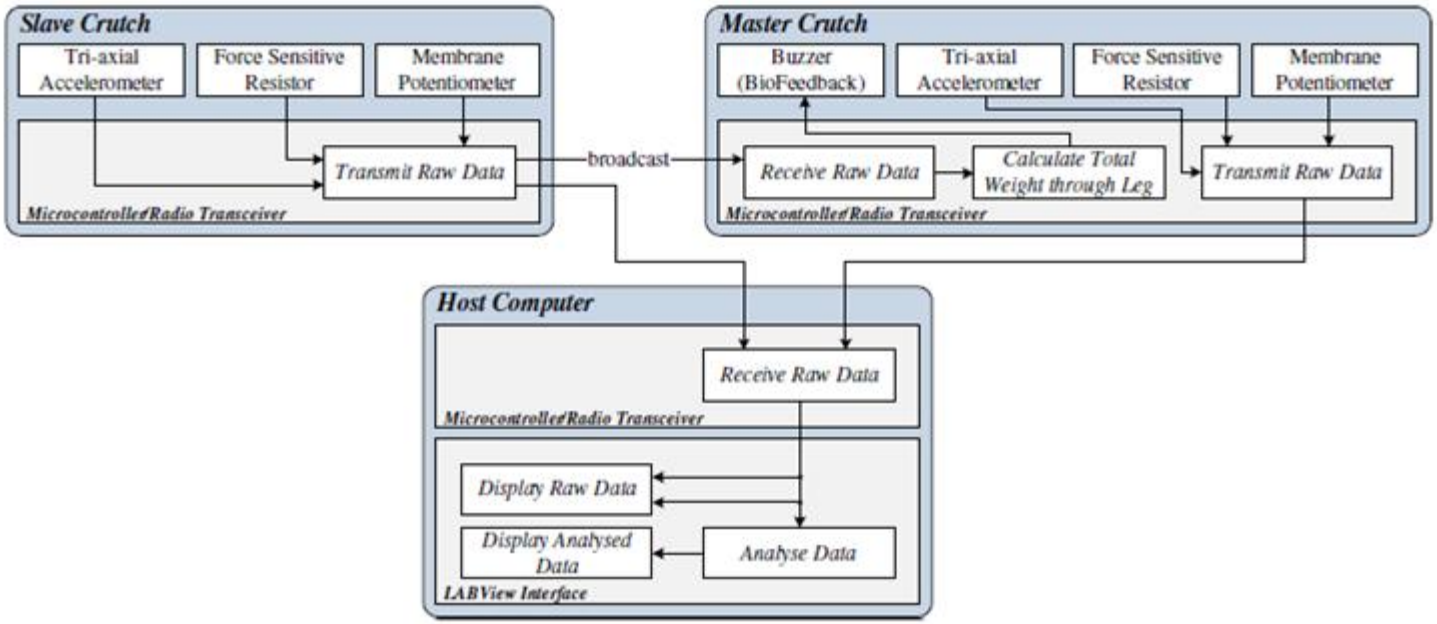

Figure 4. Data flow through the instrumented crutch system.

Example

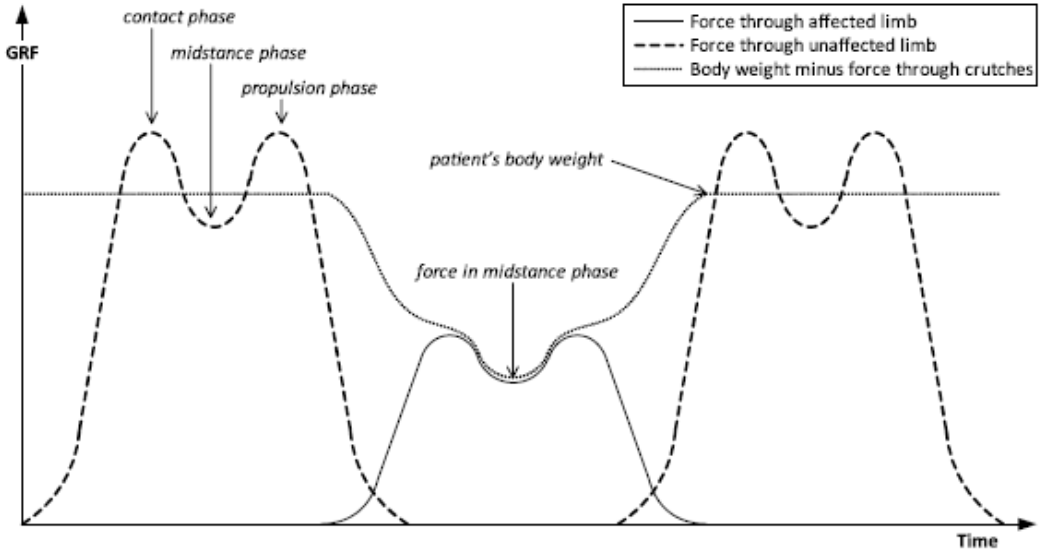

Figure 5. Graphical depiction of the theoretical forces through the affected limb, healthy limb and crutches during PWB crutch-assisted gait.

at this point that the limb is under a 'steady' force (and not a dynamic heal contact or propulsion force). These decisions were made upon consultation with clinicians in order to identify which forces were of specific interest. The midstance force occurs when the force through the crutches are at their maximum.

As discussed in the introduction, it is recognized that the maximum GRF is highly correlated with the walking speed and stride length, for example, where impact forces increase with an increase in walking speed. By neglecting the impact forces in our calculations, this correlation is likely to be less significant as the 'steady' force will still relate to the fraction of body weight being applied through the affected limb. Further, a patient's walking speed should also be reasonably limited as the clinician will teach them to use the crutches in a slow and controlled manner.

Equation (1) is used to calculate the percentage of the patient's body weight that is translated through their limbs, where $|F_{c1}[n]|$ and $|F_{c2}[n]|$ are sets of the sampled magnitudes of the forces through the axes of crutches 1 and 2 during the period of one gait cycle, M is the mass of the patient (kg) and g is the acceleration due to gravity (m s^{-2}). The maximization is performed over the sum of the crutch forces with respect to n, where n is a discrete time index. Note that this equation estimates only the weight translated through the patient's limbs; hence it can only provide the weight bearing through the affected limb when the healthy limb is not in contact with the ground. The use of forearm crutches in PWB gait entails planting both crutches on the ground at the same time (which is while the affected limb is in contact with the ground). Therefore, and because the algorithm uses only the total force through both crutches, there is no requirement for the master (or slave crutch) to be used on a particular side

Example

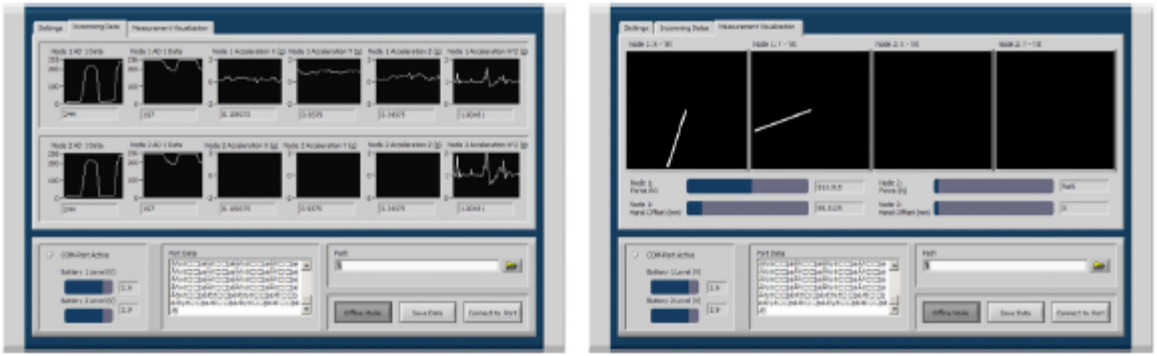

Figure 6. Screenshot of the LABView interface, (i) the raw data screen and (ii) the visualization screen.

of the body. Hence the system still operates correctly if the crutches are accidentally interchanged.

$$W^{\%}_{\text{actual}} = 1 - \frac{\max(|F_{c1}[n]| + |F_{c2}[n]|)}{M \cdot g}. \quad (1)$$

The parameters entered into the smart crutch are the patient's weight M (kg), the target level of PWB $W^{\text{aim}}_{\%}$ (%) and the tolerance that is permissible in the PWB applied α (±%). These parameters are different for each patient and also depend on the stage that they are at in their rehabilitation programme. The crutch checks to see if biofeedback is required using equation (2). If the lower threshold is not met, the onboard buzzer sounds once to provide biofeedback to the patient. If the upper threshold is exceeded, the onboard buzzer sounds twice:

$$W^{\%}_{\text{target}} - \alpha < W^{\%}_{\text{actual}} < W^{\%}_{\text{target}} + \alpha. \quad (2)$$

This algorithm is continually repeated at the master crutch for each gait cycle.

3.2.2. Graphical user interface. The host computer is used to display the real-time data received from the crutches using a LabVIEW GUI, which allows the raw data from all sensors on both crutches to be viewed in graph form. Screenshots from the raw data and visualization screens are shown in figure 6.

The GUI also converts the ADC output values representing $|F_c|$ into a force (N), d into a distance (mm) and $\hat{F}_c$ into the total acceleration (g) and tilt angles θ and φ (degrees). Additionally, the energy remaining in the batteries in each crutch can be inspected, the data received from the crutches exported to a text file (for further analysis) or imported into the GUI (for offline viewing).

(positive values of θ occur when the crutch is behind the subject). First (marked as phase (a)), the crutch is swung in front of the subject while their body weight is supported through their healthy limb. Then, when θ is at its minimum (marked by the line separating phase (a) and (b)), the subject loads the crutch using their body weight (F_c). The affected limb then gently supports a partial fraction of their body weight, while the healthy limb moves forwards (marked as phase (b)). When θ is at its maximum (marked by the line separating phases (b) and (c)), the subject's weight is removed from the crutch, which is subsequently swung back in front of them while their body weight is supported by the healthy limb (marked by phase (c)). This cycle then repeats as the subject continues to walk. The visible variation in φ is due to small changes in the roll of the crutch and also dynamic accelerations (such as impact forces, etc). The hand position distance d remains reasonably constant over the 20 s period for which data are shown, as expected.

A small preliminary study was performed to evaluate the system and to assess the potential for using instrumented crutches to estimate the weight-bearing through a limb. The study recorded the magnitude of the force translated through the axis of both instrumented crutches and, using a Pedar in-shoe dynamic pressure measuring system [23], the GRF put through each foot. The Pedar system consists of a pair of thin, flat and flexible insoles (placed inside the subject's shoes) containing capacitive pressure sensors that measure the body weight being applied through each foot. The insoles are wired to a processing box strapped to the subject's waist, which is connected to a laptop computer. The healthy subjects were given the instrumented crutches to use, and instructed to walk using three-point gait for ten steps, aiming to put 50% of their body weight through their affected limb.