# An Attachable Clothing Sensor System for Measuring Knee Joint Angles

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***Abstract*— Flexible sensors that can be integrated into clothing to measure everyday functional performance is an emerging concept. It aims to improve the patient's quality of life by obtaining rich, real-life data sets. One clinical area of interest is the use of these sensors to accurately measure knee motion in, e.g., osteoarthritic patients. Currently, various methods are used to formally calculate joint motion outside of the laboratory and they include electrogoniometers and inertial measurement units. The use of these technologies, however, tends to be restricted, since they are often bulky and obtrusive. This directly influences their clinical utility, as patients and clinicians can be reluctant to adopt them. The goal of this paper is to present the development process of a patient centered, clinically driven design for an attachable clothing sensor (ACS) system that can be used to assess knee motion. A pilot study using 10 volunteers was conducted to determine the relationship between the ACS system and a gold standard apparatus. The comparison yielded an average root mean square error of ∼1°, a mean absolute error of ∼3°, and coefficient of determination above ($R^2$) 0.99 between the two systems. These initial results show potential of the ACS in terms of unobtrusive long-term monitoring.**

***Index Terms*— Biomedical equipment, motion measurement, polymers, conducting materials, nanostructured materials.**

## I. Introduction

OSTEOARTHRITIS (OA) is often said to be a natural consequence of ageing and is one of the leading causes of disability and functional limitations in older adults [1]. OA is a multi-joint degenerative disease process, which most commonly affects the weight-bearing joints [2]. It is currently the most prevalent musculoskeletal disease. The National Arthritis Data Workgroup reported that 26.9 million adults in the United States (US) suffered from some form of OA in 2005 [3]. This debilitating disease is characterised by progressive disappearance of hyaline articular cartilage, with subsequent sclerosis and formation of cysts and osteophytes [4], resulting in joint pain, stiffness and reduced range of joint movement [5]–[9]. The clinical management of early OA focuses upon conservative and medical treatments such as weight loss, physiotherapy, pain-relief medications and corticosteroid injections [10]. Total joint replacement constitutes the definitive procedure used to treat patients with end-stage disease [11], [12]. The calculated total annual costs of OA in the US has been estimated to be £57.1 billion. A substantial proportion of these costs relate to treatment. Success rates of intervention methods vary and there are patient compliance issues during the recovery periods.

To date, efficacy of therapeutic interventions has principally been monitored through subjective pain assessment [13], [14]. Yet, the disease directly impacts on the functional performance of the patient. If performance of the musculoskeletal system is assessed it will often be performed in a hospital or laboratory setting and frequently influence the subsequent treatment. The ability to more accurately measure joint motion may allow health care professionals to objectively assess these treatment strategies accordingly [15]. Quantification of joint angle motion may also help with the clinical assessment of various other knee pathologies in addition to OA, as well as aid the control of wearable robotic devices [16].

The value of technology to promote delivery of health services has been acknowledged in the last few years by policy-makers [17]. Laboratory based equipment can be used to collect data regarding angular motion and although this kind of research yields valuable information, the results only remain valid in conditions where no anticipation or reaction to a real-world environment is required [18]. In addition, laboratory equipment is expensive and cumbersome to work with, and not practical in a clinical or rehabilitation scenario. A body sensor network that consists of wireless sensors, which can be easily attached to the patient, provides a promising method to collect clinically relevant information about knee function in everyday situations [19]. One strong motivator for the development of mobile health systems is the tremendous benefits that are associated with continues long-term monitoring of patients in the home and community setting [20]. Obtaining accurate data about joint functioning in real life environments has significant clinical relevance and could lead to further improvements in both prevention and rehabilitation [21]. However, the practical use of many sensor systems is regularly limited by the usability, cost and applicability of the device.

The aim of this study is to develop a patient-centred and clinical driven prototype design for an integrated clothing sensor system for osteoarthritis. The system should be able

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to accurately measure joint kinematics (bending and flexing) of the knee and can potentially be used in the future to detect and manage OA.

## II. Design Process

### A. Technology Selection

The selection of an appropriate technology started with defining the design considerations. Our first step was to ascertain what was known in the published literature with regards to user preference [22] and subsequently perform a survey pertaining to this with respect to osteoarthritis. From this it became clear that the requirements for wearable medical equipment are insufficiently addressed by many of the systems currently available. A successful measurement device needs to fit several design criteria's of which the most important are that it is small, discreet, unobtrusive and preferably incorporated into an everyday object [23]. Therefore, a system that is discreetly attached onto clothing would provide the most appropriate sensor modality for long-term monitoring. On average, consumers expect their (under) garment to last about 18 months, and they add to or replace them every 7 months [24]. Long-term monitoring is therefore defined according to the expected lifecycle of the garment. Flexible sensors were selected as the preferred sensor type, as they would allow for the greatest amount of unobtrusive attachment to a garment.

### B. Flexible Sensors

Flexible sensors can be defined as sensors that can deform without breaking. These sensors can change in shape or size in a consistent manner that is related to the forces that are exerted on it. A number of these systems are designed to resist changes in the signal, as consequence of deformation. The aim of these particular sensors is not to measure deformation, but to provide a sensing modality that is independent of the distortion. The primary design criterion for these sensors is to develop a robust output system in a changing environment.

However, other sensors do rely on the deformation to alter the electrical properties (conductivity) of the sensor material and use it to quantify the deformation itself. Materials that change output, due to stress, can be used as a flexible deformation sensor. These sensors require an elastic range over the deformation of interest, providing the capability for repeated measurements. The change of shape can thus be utilized to sense and inform.

Shape is essentially what is left when the differences, which can be attributed to translations, rotations, and dilatations have been filtered out [25]. If the human form is taken as one entity, the changes in shape mainly relate to movement of the musculoskeletal system. This indicates the potential use of flexible sensors to measure deformity due to human movement. Flexible sensors that can measure human motion range from thin films of aligned single-walled carbon nanotubes [26] and ZnO nanowire/polystyrene hybridized flexible films [27] to electrogoniometers [28]. In addition, flexible sensors can also be designed based on a mixture of rubber and carbon [29]. It has been known for a long time that carbon black can change the electrical conductivity of elastomers [30]. A composite carbon material can provide the base material for flexible sensors. The advantage of such a composite material is that the procedure to make the sensor is relative easy and low-cost. A carbon composite material was therefore selected as the base material for the flexible sensor.

### C. Materials

Graphitized carbon black nanopowder (CN) with a particle size of $< 500$ nm and the solvent anhydrous tetrahydrofuran were obtained from Sigma-Aldrich (St. Louis, MO, USA). The carbon itself has a resistivity of around $3.5–4.6 \times 10^{-5}$ $\Omega$m [31] indicating good conductive behaviour. The electrical conductivity of a composite that includes CN is a direct function of the volume taken up by carbon nanoparticles within the content.

The CN was combined with polyether resin Texin 985 (polyurethane), received from Bayer AG (Bayer MaterialScience, Leverkusen, Germany), to generate the composite material. Polyurethane is a polymer composed of a chain of organic units joined by carbamate urethane links. Polyurethane combines the "best" properties of both rubber and plastic. It is frequently used in the production of tires, hoses and other durable equipment.

Several composite samples with different thickness and mechanical properties were prepared and tested. The final selected composite consisted of 20% conductive CN material and 80% non-conductive polyurethane. This ratio was selected as it showed high electrical conductivity, while maintaining the mechanical properties of the polyurethane. No leaching (removal) of the CN from the polymer body was observed at this level.

Since the aim was to provide an attachable clothing sensor, short-term durability tests were performed on the films by washing them 4 times for 10 minutes at 30 °C using a conventional cleaning agent. No changes in the mechanical (tensile) properties and electrical resistance were found after this procedure. These results show that the washing agent does not affect the sensor directly after the first few washing cycles. Further testing is needed to provide information regarding persistence over longer periods.

*1) Preparation Process:* The polymer was dissolved in the solvent in a controlled environment with constant stirring and a controlled temperature of 60 °C for 12 hours. The temperature of the polyurethane dissolution can have a gradual effect on the membrane structure.

Depending on the temperature of polyurethane dissolution, a gradual variety in the membrane structure can be observed [32], hence the need for precise control during the wet-casting membrane formation. The obtained liquid was mixed with the carbon nanopowder in order to achieve 20 w% (the w% indicates the percent by weight of solute in a 100 gram solution) and left stirring for an additional 4 hours. Subsequently, the homogeneous liquid was left overnight under a cover with very small holes in order for the solvent to evaporate. At the end of this process thin films were produced that could be sliced in to a shape suitable for the proposed purpose.

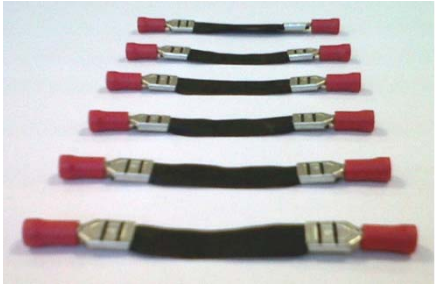

Fig. 1. Individual Polymer Sensors fitted with connectors.

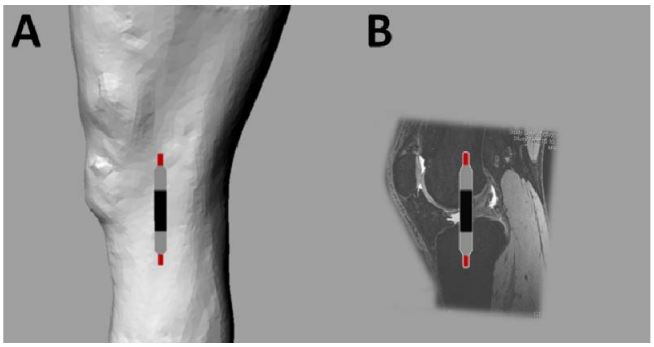


Fig. 2. Sensor dimensions superimposed on a knee of a subject that is (**A**) 3D scanned, as well as (**B**) imaged by magnetic resonance [MRI]. It shows that the selected placement and dimensions cover the joint space, as assumed based on the anthropometric data.

### *D. Sensor Design*

The polymer composite material can be sliced in different shapes. Varying the 3D shape will yield contrasting levels of resistance. In this case the shape was optimized for the joint of interest, the knee.

A sheet with a thickness of approximately 0.2 mm was produced. The width was optimized to fit the connectors that would be placed on the ends of the sensor. Equal pieces ($10 \times 50$ mm) were cut from this sheet to obtain the foundation for the flexible sensor. Each strip was connected to a receptacle 0.25" connector at each end. This provided an active surface of 33 mm between connectors (Fig. 1).

The specific dimensions were selected as the sensor had to be able to cover the knee joint, which is the largest joint in the human body.

It is known that the radius (mm) of the sagittal section of lateral femoral condyle is about 22 mm [33], the anatomical knee joint space covers roughly 5 mm [34] and another 22 mm was chosen to cover the distal sensor part over the tibia. The summation of these distances provides a length that can easily cover the knee joint without overextending (Fig 2).

The selected dimensions allow for some translations to occur when the sensor is attached to the garment. The placement of the sensor on the side of the knee was chosen, as it showed less displacement artifact during movement compared to a more frontal placement. It was also deemed the best location to minimize contact between the sensor and other objects.

The described polymer bands will be referenced as Attachable Clothing Sensors (ACSs), as they will be sewn into the garment (Fig. 3).

Fig. 3. Sensor attached on to a commercially available garment.

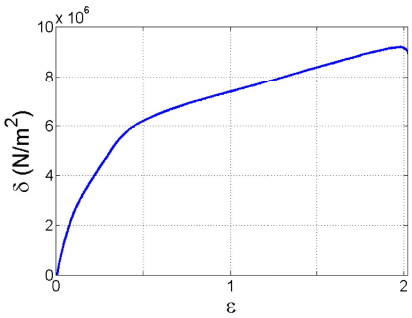


Fig. 4. Stress-strain curve obtained from the ACS sensor.

A commercially available long-sleeved trouser was used for sensor integration and consisted of 94% viscose and 6% elastane.

## III. Properties of the System

### *A. Sensor Properties*

Tensile measurements were performed on a universal testing machine (Model 5866, Instron, USA) with a longitudinal action grip speed of 100 mm/min. The sensor was preloaded up to 5 N for 20 cycles. Unloading was done by bringing the material back to slack ($-1$ mm) during each cycle. Directly after cyclic loading a test to failure was performed. A 500 N static load cell was used to obtain all force data.

The Young's modulus ($Y$) was calculated by dividing the tensile stress by the strain in the initial, linear part of the stress-strain curve using the equation

$$Y = \frac{\sigma}{\epsilon} \tag{1}$$

where $\varepsilon$ is strain and $\sigma$ is tensile stress (N/m$^2$). A Young modulus of $3.4 \times 10^7$ N/m$^2$ was found for the ACS sensor (Fig. 4). The apparent yield strength, defined as the amount of stress that a material can undergo before undergoing plastic deformation, was found to be $1.3 \times 10^6$ N/ m$^2$ at a Yield point of 0.05. The Ultimate Tensile Strength or stress level at which the material breaks was $9.3 \times 10^6$ N/m$^2$. With the sensor dimensions a loading of 21 N would be needed before the composite material will break at an elongation of almost twice its original length. The actual point of failure of the sensor is less due to the use of the connectors. In another test to failure it was found that the sensor material ruptures, at the point of the connector, with a force of 14 N. This information was taken into account in terms of selecting an appropriate textile for integration.

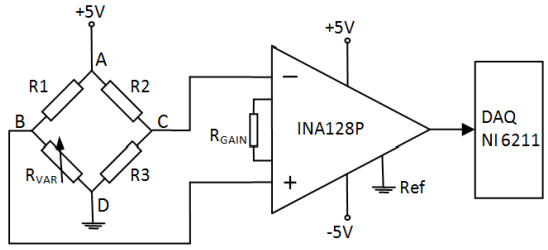


Fig. 5. Block diagram of the interface circuitry.

The electrical resistance at rest for 6 different sensors was measured at 2.42 KΩ (standard deviation of 0.19 and a range of 2.11 to 2.69 KΩ).

### *B. Interfacing*

A Wheatstone bridge configuration was used to measure the resistance variations of the ACS (Fig. 5). The ACS sensor changes resistance ($R_{VAR}$) when deformed and forms one of the bridge arms, while R1, R2 and R3 constitute the other three arms. The bridge output is amplified by an instrumentation amplifier (INA128P) connected to points B and C. The gain of the amplifier is set by resistor $R_{Gain}$. Power supplies are regulated (KA 7805, LM7905) to stable ±5 V to power the bridge and instrumentation amplifier.

The bridge will mainly operate off-balance and the values of the resistors (3 KΩ) were chosen to match the resistance value of the ACS sensor over a functional range. A gain of 10 was used to amplify the signal.

The output from the circuit was connected to an analogue input pin of the data acquisition board (USB 6211 DAQ, National Instruments, USA). Data was gathered via a simple Simulink model (Mathworks, USA) with a sampling frequency of 50 Hz.

*1) Circuitry Testing:* Testing of the circuitry took place under static and dynamic conditions. Initial tests were performed to assess the stability of the developed circuitry.

The circuitry was initially tested with a 3 KΩ resistor (±1% tolerance) placed at the connections normally used to attach the ACS. Baseline drift for the circuitry was determined by obtaining two 10 second datasets a minute apart. A mean difference of 0.2 mV was found with an average root mean square error (RMSE) of 0.6 mV. The readings stayed within the same minimum and maximum values (29 to 33 mV) during both trials, showing good short-term stability of the developed board.

The same test was then repeated with the ACS sensor attached to the Wheatstone bridge (Fig. 6). Eight 10 second baseline measurements were again taken with the sensor attached. Each measurement was taken a minute apart and an average RMSE of 0.1 mV was found across all trials.

The measurement was repeated a day (24 h) later to determine long-term drift and a similar RMSE (0.1 mV) was found.

Consequently, the sensor was dynamically loaded at one end using dead weights. A typical signal response is given in Fig. 7. It shows the sharp rise when the sensor is loaded with

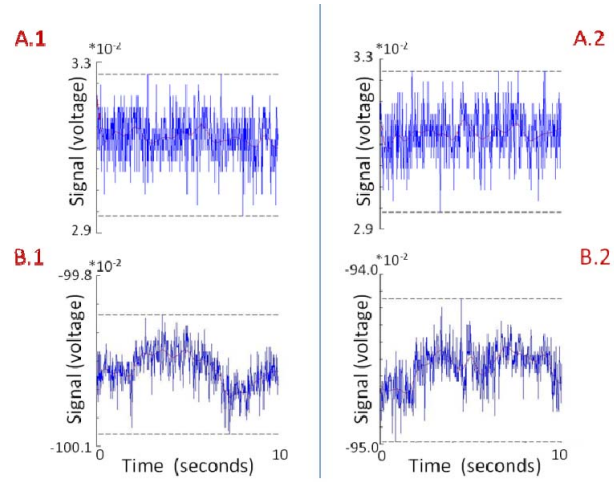


Fig. 6. Baseline signals measured for the circuitry for ten second intervals. A.1 First measurement with a film resistor of 3 KΩ. A.2 Last measurement with the film resistor. B.1 First measurement with the sensor at rest B.2 Last measurement with the sensor at rest.

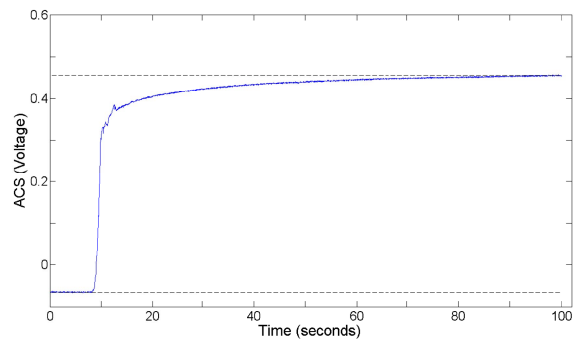


Fig. 7. The sensor started in unloaded condition and was loaded (4 N) at the 4 seconds time-point.

a 4 N force. It can also be observed that there is low frequency component within the loading trace, indicting the potential of variability over time.

The ACS sample was then loaded using 10 different weights. Loading force (ranging from 0 to 10 N) was estimated by multiplying the mass with the standard average of the earth's gravity. The mean of every 10 second static measurement was taken and plotted against the known force. This procedure was repeated three times. Offset was corrected for each repeat measurement. It was found that the response changed in an exponential fashion across the three trials (Fig. 8).

A 44 percent change in resting length at 10 N indicates that plastic deformation has occurred at this stage. This warranted the need for offset correction between the trials. However, despite the deformation, signal outputs remained relatively consistent. Under normal everyday living conditions changes in resting length are expected to be around 10%. Loading and unloading the ACS up to 5 N showed that hysteresis is present, but a good linear relationship ($R^2 = 0.95$) between force and ACS output could still be found.

The antecedent lack of response can be linked to the mechanical properties of the material, in which the stiffness is highest during initial loading. A greater response can therefore

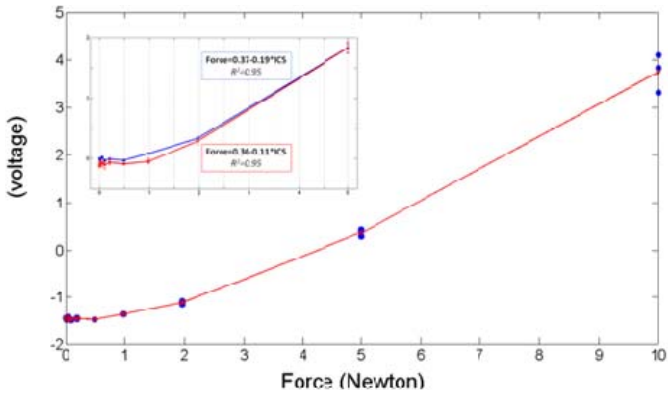


Fig. 8. Loading and unloading of the ACS over 3 trials. The blue stars represent the individual values. The red line shows a linear interpolation of the average of all three trials for a specific load. Repeat trials were corrected for offset. The inset graph shows the loading (red) and unloading (blue) of the ACS up to 5 N.

be expected in the subsequent more ductile portions of the stress-strain curve.

Repetitive loading of 4 N was applied to determine the stability of the ACS. Data was low-pass filtered with a 4th order Butterworth [35] using a cut-off frequency of 6 Hz. The maximum and minimum values during loading and unloading were taken from the filtered data. The signal range for each of the 13 loading trials was determined. The data set showed that a RMSE of 9% was present for the total signal.

These results indicated the ACS system is sensitive and reliable enough to potentially provide information regarding the knee joint angle.

## IV. Knee Joint Angles

A protocol was developed to relate the ACS signal outputs to knee joint angles. A Cybex dynamometer (Human Norm, CSMI Inc., Stoughton, MA, USA) was used to standardize the movement of the knee. The knee reference angles were obtained from a calibrated accelerometer (Vernier Labpro, Oregon, US) that was attached to the dynamometer. The information from the accelerometer was gathered using the same A/D converter in order to ensure synchronisation between the two measurements.

Based on recommendations made by [36], on sample size for reliability studies, ten subjects were recruited for this study, 8 men and 2 women. The following average (± standard deviation) demographics were noted for this group; age was 22 (±2) years, height was 1.78 (±.12) m and mass was 78 (±15) kg. The protocol was approved by the College Research Ethics Committee and all subjects gave written informed consent previous to the experiment. The participant was fastened to the dynamometer, while wearing the ACSs (Fig. 9). To reduce extraneous body movements during the experiments, straps were placed across the chest, pelvis and mid-thigh throughout all movement patterns. The knee of the participant was moved in the sagittal plane at 120°/s. The dynamometer was set to continuous passive mode for a self-selected range of motion determined by the subject. The knee of the subject was extended and flexed during two sets of six repetitions. The first set was used as training set for the algorithm, while the subsequent data set was applied as a test set. The algorithm for the ACS system was first trained and then validated against the new set of knee joint angles.

Fig. 9. Experimental setup showing ACSs and dynamometer. Several sensors were initially integrated into the garment, but only the optimized sensor location previously described lateral sensor placement was used for data collection.

The outcomes of the predictive function were compared to the reference joint angles values. The coefficient of determination ($R^2$) and Root Mean Square Error (RMSE) were computed to determine the robustness of the relationship.

### A. Algorithm

Data acquired during the ACS testing trials indicated the need for a subject dependent calibration. This procedure allowed fitting of the sensor data on the angular scale. The calibration step will be used to find the optimal function for predicting the joint angle. The 6 Hz low-pass filtered ($4^{th}$ order Butterworth filter) ACS output was directly plotted against the joint angle data. The results showed that multiple joint angles provided the same voltage output. Therefore, relating the voltage directly to joint angle would not provide a valid function. However, the pattern always showed a distinct peak and trough throughout the movement range. Focusing on a subset of the data (outputs between the peak and trough) allows for derivation of a subject specific function to convert voltages to angles. The specific region between minimum and maximum voltage values was therefore selected for further processing. The pattern within this region was found to be best modelled by using the Fourier curve

$$f(\alpha) = a0 + a1 * \cos(V * w) + b1 * \sin(V * w) + a2 * \cos(2 * V * w) + b2 * \sin(2 * V * w). \quad (2)$$

In which $a0$, $a1$, $b1$, $a2$, $b2$ and $w$ are all subject dependent coefficients to estimate the knee joint angle ($\alpha$) based on the ACS signal ($V$). The obtained calibration equation from the training set was then applied on a new dataset. The knee moved through a similar range of motion in the test data set. The data could now be analyzed within the time domain. The transformation equation was applied to the voltage signal between peak and trough. Providing the data for which the knee joint angles were computed.

User preferences were also taken into account in the algorithm development phase. Patients like the proposition of receiving quantifiable, objective data about their performance

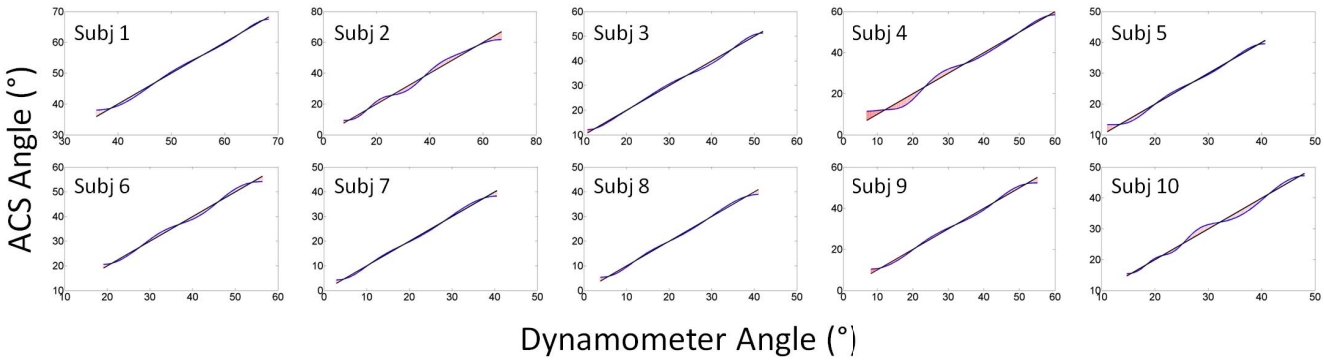


Fig. 10. Plots of relationship between estimated angles obtained using the ACS system and the corresponding reference angles. Blue curve represents actual relationship between angles. Black line represents theoretical ideal fit. Shaded area reflects differences. Data shown is for all subjects over the motion range that they felt comfortable with.

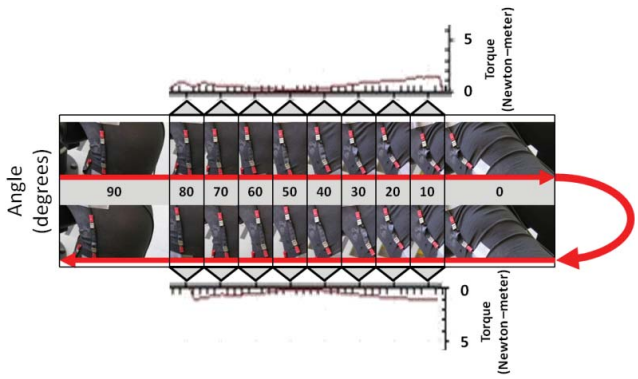


Fig. 11. Deformation of the ACS during knee flexion and extension. The top and bottom plots show the low torque generated during these passive motions. The middle row indicates the angles at which the pictures (top is extension and bottom is flexion) were taken.

from a system, rather than just praise from a physiotherapist [37]. Providing quantifiable data streams for the subject was therefore a key aspect in the design. Furthermore, it has been suggested that clinicians want a simple interface with a minimum amount of options [38]. We therefore focused on developing a simple visual feedback interface for the knee angle.

### *B. Flexible Sensor Knee Angle Output*

The ACS was placed over the knee by the subject, by asking the participant to wear the garment as he would see fit. A typical movement pattern of the ACS is provided in Fig. 11.

The predicted knee joint angles seemed to correlate well with the reference angles (Fig. 10). The ACS yielded an average RMSE of 0.9° [range: 0.5–1.8], absolute error of 2.6° [range: 1.3–5.2] and an $R^2$ value above .99 across all subjects.

The data was obtained for a range that was perceived as comfortable for the subject and provided a clear signal for analysis.

## V. Discussion

This paper describes the full design process for a novel attachable clothing sensor (ACS) system. The goal was to develop a flexible low-cost sensor that could measure knee joint performance. Good initial results have been obtained within the laboratory setting, indicating the clinical potential of such a system.

Previous studies have investigated the application of (flexible) sensors for the measurement of joint motion [39]–[42]. However, the majority of these studies have required sensors to be physically sewn onto a fabric piece which is then placed over a person's clothes. Such an approach would require the user to perform additional placement tasks and/or tolerate additional wearable equipment. We would like to advocate the possibility of integrating the flexible sensors directly into the garment.

Garment sensors have also been produced by printing a carbon/rubber mixture onto fabrics [29]. The sensor integration by printing or coating is a more complex process then sewing single sensors onto commercially available garments. The freedom of selection of (re)placement of sensors provides a likely benefit for the research community. In addition, the prospective to apply the ACS on a variety of clothes will decrease user compliance issues. It has already been shown that osteoarthritis patients prefer wearable devices that have been incorporated into everyday objects [23]. The ability of fully integrating flexible sensors into high-street clothing will provide the means for real unobtrusive long-term monitoring. This kind of synthesis will also give higher levels of ecological validity within the obtained data streams.

The ACS was built from stretchable material, which will change shape in 3D when exposed to strain due to human motion. Inter-person variability in terms of anatomy, coordination and reference frames will yield different signals for the same activity. A subject dependent calibration is therefore required. Such a calibration is often a prelude to most sensor usage [43], but it is intended for the ACS to have an unobtrusive calibration. This unobtrusive calibration can be performed with additional e-textile sensors. These simple (binary) switch sensors can measure contact and classify if the person is sitting or lying. Reference angles can be assumed based on this basic information and feedback into the algorithm. However, utilizing known movement behaviour would also be of great interest, as it could allow for continues functional calibration while minimizing energy consumption. This requires the identification of robust physical characteristics within a population. There is evidence that new dynamical charac-

teristics can be identified by an in-depth analysis of certain activities [44].

Offset corrections were needed when the sensor reached elongations that caused plastic deformations. Although, the ACS should remain within the elastic region during everyday use, it can't be assumed it will not undergo plastic deformation. At present, it was found that during the testing only a small amount of elongation occurs and most of the signal change was due a combination of translations and rotations. There was no direct need to correct for offset under these conditions. For most clothing fabrics structural damages will occur prior to plastic deformation of the sensor. This should keep the sensor within range during everyday use, but if the fabric shows damage then the sensor should be replaced. This concept will allow the general public to easily identify if the sensor is still working appropriately. However, as mentioned previously it is likely that more intense use could push the sensor into the realm of plastic deformation. Referencing the baseline signal to a known joint angle is therefore of interest for longer-term monitoring of for example sports activities. The previous extension of the ACS system with contact sensors can provide a valid technique for providing the data needed for offsetting in real life situation.

Although, the current version of the sensor was attachable and freely movable in order to explore optimal sensor location, further developments will introduce a sensor placement in between fabrics to make it a fully integrated clothing sensor.

The ability to continuously monitor joint motion in day to day life enables objective assessment of treatment response in different knee pathologies. In the clinical setting, accurate joint motion has traditionally been evaluated using standard tools, such as a goniometer to record range of motion. However, this static measure does not offer the same standard of information as dynamic assessment.

Although, the initial dynamic results of the ACS show low levels of error between the obtained angles and the reference angles, more work needs to be done. The test dataset was obtained under very similar conditions within a laboratory environment. The next phase would be to test the ACS system under real world conditions.

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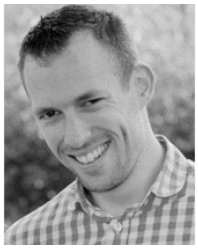

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