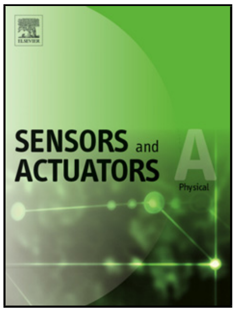
Low cost portable 3-D printed optical fiber sensor for real-time monitoring of lower back bending

Item Type	Article
Authors	Kam, Wern;O'Sullivan, Kieran;O'Keeffe, Mary;O'Keeffe, Sinead;Mohammed, Waleed S.;Lewis, Elfed
Citation	Sensors and Actuators A: Physical;265, pp. 193-201
Publisher	Elsevier
Download date	2026-04-15 09:56:25
Item License	https://creativecommons.org/licenses/by-nc-sa/1.0/
Link to Item	https://hdl.handle.net/10344/6228

Accepted Manuscript



Title: Low Cost Portable 3-D Printed Optical Fiber Sensor for Real-Time Monitoring of Lower Back Bending

Author: Wern Kam Kieran O'Sullivan Mary O'Keeffe Sinead O'Keeffe Waleed S. Mohammed Elfed Lewis

PII: S0924-4247(17)30594-0
DOI: http://dx.doi.org/doi:10.1016/j.sna.2017.08.044
Reference: SNA 10299

To appear in: *Sensors and Actuators A*

Received date: 13-4-2017
Revised date: 3-8-2017
Accepted date: 22-8-2017

Please cite this article as: W. Kam, K. O'Sullivan, M. O'Keeffe, S. O'Keeffe, W.S. Mohammed, E. Lewis, Low Cost Portable 3-D Printed Optical Fiber Sensor for Real-Time Monitoring of Lower Back Bending, *Sensors and Actuators: A Physical* (2017), http://dx.doi.org/10.1016/j.sna.2017.08.044

Low Cost Portable 3-D Printed Optical Fiber Sensor for Real-Time Monitoring of Lower Back Bending

Wern Kam[a*], Kieran O'Sullivan[b,d], Mary O'Keeffe[b], Sinead O'Keeffe[a], Waleed S. Mohammed[c], Elfed Lewis[a]
[a]Optical Fibre Sensors Research Centre (OFSRC) , Department of Electronic and Computer Engineering, University of Limerick, Limerick, Ireland.
[b]Department of Clinical Therapies, University of Limerick, Limerick, Ireland.
[c]Center of Research in Optoelectronics, Communication and Control Systems (BU-CROCCS), School of Engineering, Bangkok University, 12120 Patumthani, Thailand.
[d]Sports Spine Centre, Aspetar Orthopaedic and Sports Medicine Hospital, Qatar
.

Highlights

- An optical fiber intensity modulated sensor that can monitor the bending of lower back bone in both sagittal and frontal planes.
- The optical fibre sensor system has an operation range between -12° to +12° for both bending modes.
- Sensor provides real time feedback to the clinical therapist when different postures are sustained.
- Experimental results captured from the sensor mounted on human subjects were correlated with angular deformation values obtained using a simultaneous image capture method.
- All-plastic composition 3-D printed sensor that has advantage to be used in conjunction with Magnetic Resonance Imaging (MRI) scanning machines as well as X-Ray based scanning machines.

Abstract: A mechanically robust and compact novel optical fibre sensor system is described to monitor the bending of the lower back bone in both sagittal and frontal planes. Both bending modes are monitored through the change of the coupled optical intensity ratio between three output fibers aligned to one input fiber. This provides real-time feedback to the clinical therapist when different postures are sustained. The output ratio is calibrated against bending angle using an optical setup utilizing a precise rotational stage. The measured data is also correlated to the curvature of the lower back through the implementation of an ad-hoc imaging scheme. Sequences of images are also captured while the optical fiber sensor is attached on the skin surface to the lower back. The imaging system tracks three spots placed on the sensor and skin to trace the angle changes. The optical fibre sensor system has an operational range between -12^{o} to +12^{o}. It is demonstrated that the sensor is suitable for clinical use with the additional benefits of being non-invasive, robust, straightforward to use and low cost. It also allows record of spinal curvature in the home and other real-world settings and potentially reduces the requirement for the use of X-rays and MRI in the clinic.

Keywords: Optical Fiber Sensor, 3D printed sensor, Lower Back Bending Sensor, Lateral and Sagittal plane motions

*****Corresponding author.
E-mail address: wern.kam@ul.ie

1. Introduction

Low back pain (LBP) is one of the most common health problems that influences an individual's life quality and is a leading cause of disability [1]. It also affects people of all ages. The cost of medical care for treating and monitoring LBP and related loss of time from the workplace etc. is estimated to be at least $50 billion a year [2]. These statistics highlight the need for lumbar spine monitoring among patients in a low cost and effective manner so that LBP can be accurately assessed and treated. There are several methodologies available to measure the bending angle of the spine in the clinical assessment of lumbar spine conditions. Apart from the use of imaging through radiography [3] to visualize the structural movement of the lumbar spine, there exist numerous alternative methods for externally measuring the motion and bending curve of lumbar spine which remove the need for exposure to potentially harmful ionizing radiation. Devices for dynamic spine assessment include wearable and skin-mounted types, for example tape measurement [4], various types of goniometer [5], accelerometers [6], inertial sensor modules [7] and strain-gauge type optical sensors [8]. Non-contact type sensors involving image capture form a significant part of the methods that are currently available. This type of sensor generally uses a motion capture system or other imaging system such as the Vicon optoelectronic system [9], MRI scanning [10] and raster stereography [11] to capture motion of the spine during specific body movements. These systems are very accurate, but are generally large and are expensive in terms of their capital and running costs.

An intensity-based fiber optical device for spine bending monitoring has previously been developed by some of the authors of this article, Zawawi et al [12]. Although the device is accurate and usable in the clinic, the long term stability of this device has been found to be inadequate for repeated use. A wearable optical sensor for monitoring seated spinal posture has been developed by Dunne et al [13] , where a plastic optical fiber (POF) is abraded on one side with one end of fiber connected to a light source and light sensor at the other end. Abraded POF allows light leakage through the cladding boundary and bending of the sensor affects the detected light signal at the output. The sensor was stitched to a wearable garment using a loose zig-zag stitch to allow the POF sensor to move vertically along the garment surface. This sensor is highly applicable for everyday use e.g. outdoors, bus it not ideal for the clinical environment. Williams et al [14] developed a fiber optic system to measure spinal motion that comprised eight paired fiber sensors attached to a ribbon of sprung steel. Williams et al claim good performance from the sensor but due to the presence of the steel ribbon this sensor could potentially cause discomfort to the patient, and would certainly rule it out for use in X-ray or MRI scanning environments. Other fiber sensors that have recently been developed for bending monitoring include a POF-based wearable and wireless system using side polished POF to monitor knee sagittal motion [15]. This wireless sensor has shown good performance and allows great freedom for the patient. However, the use of side polishing in the fabrication can weaken the optical fiber mechanically and potentially, reduce the lifetime of the sensor.

The sensor developed in the investigation of this work utilizes a POF bending sensor based on the change in optical intensity coupling between a feed fiber coupled to three receiving fibers which are closely grouped together. The bend causes an alignment mismatch between the input and output fibers and hence the optical power is redistributed between the receiving fibers. By measuring the output signal from the three fibers it is possible to accurately and separately determine the angle of bending in multiple directions (sagittal and frontal axis). A simple, small, lightweight, low cost and portable sensor has been fabricated using 3-D printing which allows the initial assessment and real-time tracking of curvature progression. The sensor is able to provide a real-time angle signal at different postures and this result is presented for the clinician to interpret. By associating the sensor's data with information regarding the patient's spinal curvature obtained from imaging from a simple web camera, the efficacy of sensor is demonstrated and shows that it is capable of being used as a valuable assessment tool for clinicians.

2. Methodology

2.1 Sensor Configuration and Concept

This POF sensor works based on an optical intensity interrogation technique which was previously proposed by our research team with three fibers at the output and one input fiber connected to a single LED light source [16]. The sensor of the investigation described in this article consists of four (one input and three output) multimode POFs each with a core diameter of 0.98mm and refractive index of 1.492 and 1.402 for the core and cladding material, respectively. A standard optical fiber coupled red LED (SFH756V Supplied by Broadcom Ltd, US) with a peak wavelength of 660nm was used as the light source. It was placed at one end of the single transmitting POF that is aligned to the middle axis of three POFs at the other end. The schematic diagram of the sensor layout including source and detector is shown in fig1.

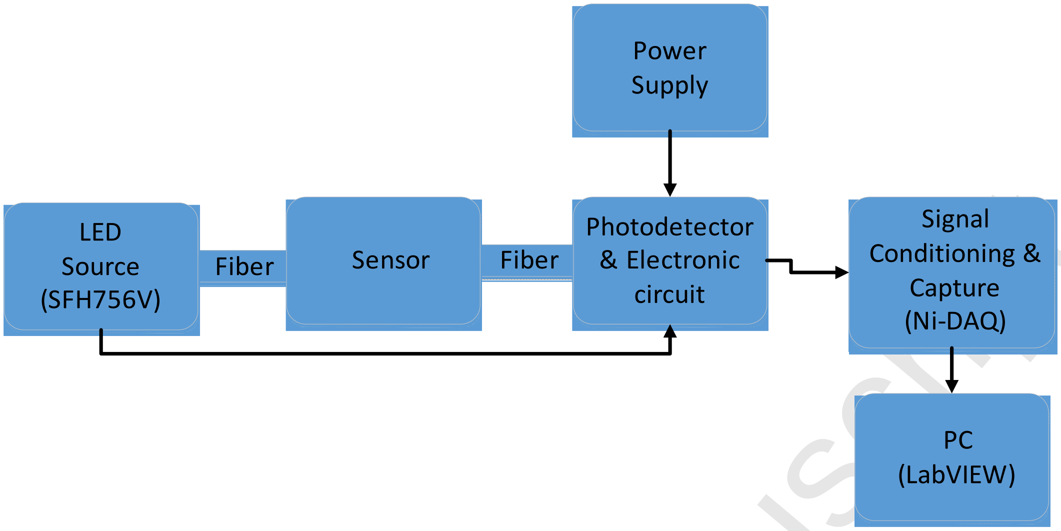


Fig. 1 Schematic diagram of the sensor layout.

The device is to be attached to the lower back area of the human spine to measure the angle of lateral and sagittal bending. Fig.2 Inset depicts the assessment of bend spine for lateral and sagittal bending. Lateral bending, also known as frontal plane motion, is the side bend from waist to either the left or right directions. Sagittal bending is movement in the sagittal plane, either forward (flexion) or backwards (extension) bending. The fiber configurations of the sensor of this investigation have been designed to detect the spine bending angle in these bending directions.

Fig. 2 shows the fiber alignment configurations for applied lateral and sagittal bending. This configuration consists of an input fiber aligned to the center of three connected output fibers so that the optical power transmitted in each of the receiving fibres is ideally equal (fig 2(a)). During lateral bending to the left (fig 2(b)), the shift of the input fiber increases the light intensity coupled to I1 and decreases that of I2. This process is reversed for the case of lateral bending to the right (fig 2(c)) . The lateral bending magnitude is estimated from the following output ratio $R(\theta_x)$.

$$R\left(\theta_x\right)=\frac{I_1\left(x,y,z\right)-I_2(x,y,z)}{I_1\left(x,y,z\right)+I_2(x,y,z)} \tag{1}$$

The sagittal bending response is calculated from the ratio formed between all three output fiber readings. During flexion bending to the front, the input fiber shifts upward (fig 2(d)). This increases the coupling to fiber I3 and decreases the coupling to fibers I1 and I2. This is reversed for sagittal extension (fig 2(e)). Sagittal bending is therefore calculated from the ratio between the measured output intensity from the three output fibers as follows: -

$$R\left(\theta_y\right)=\frac{\frac{1}{2}I_1(x,y,z)+\frac{1}{2}I_2(x,y,z)-I_3(x,y,z)}{\frac{1}{2}I_1(x,y,z)+\frac{1}{2}I_2(x,y,z)+I_3(x,y,z)} \tag{2}$$

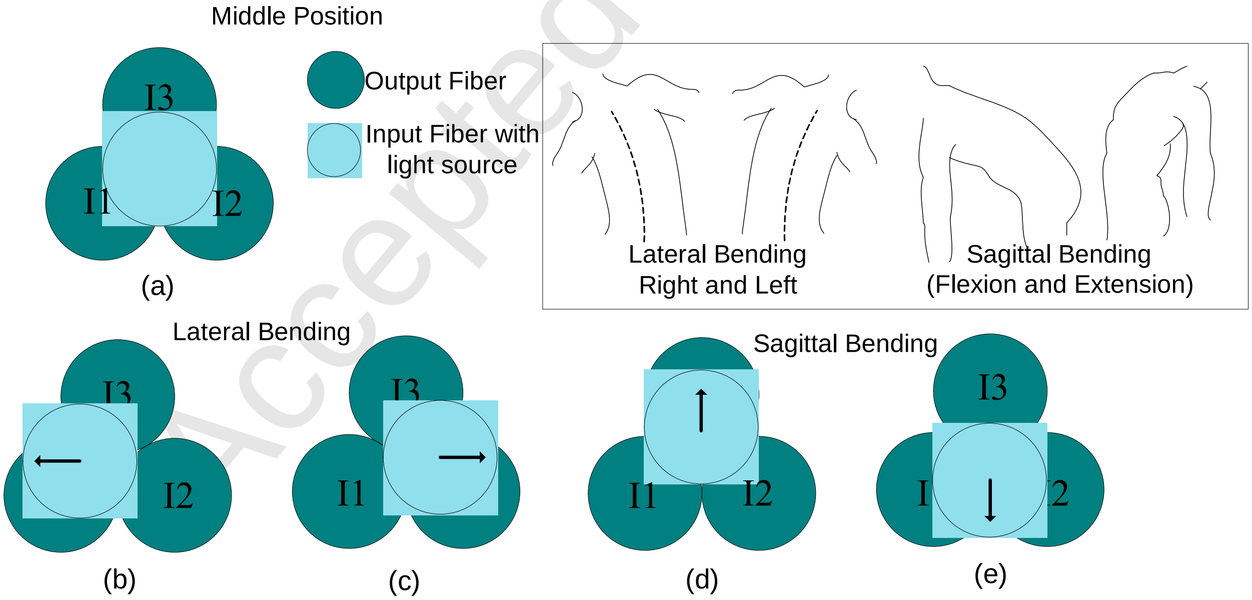


Fig. 2 Fiber configuration for lateral and sagittal bending.

2.2 Device Design and Fabrication

In this work, a specially designed sensor has been fabricated to house the fibers described in Section 2.1 and this is used to measure the bending angle of the lower spine section (lumbar spine). The requirements for this sensor are that it is lightweight, flexible and simple to use. The fabricated sensor is small (around 3cm x 6cm) so that it is able to fit within the section of the lumbar spine and has the flexibility corresponding to a bending angle up to ± 12° in both sagittal and lateral axes. The body of the sensor has to be flexible, strong and robust to hold the optical fibers still during measurement. A silicon material tube and mold is used at the fiber hinge to provide flexibility and some stretching space for the sensor during bending. Therefore the design was implemented using 3-D printing and was suitable for easy attachment onto the skin of the patient's back. A Stratasys Connex 500 printer was used for the 3-D printing using ABS-like polymer material that is durable and strong.

The cross-section of the fabricated sensor is depicted in fig. 3(a). The device consists of two tubes for hosting the optical fibers which were fabricated as a single entity and printed separately using 3-D printing but in such a way that they could be accurately aligned using minimal mechanical adjustment: the input fiber tube is shown on the left of fig. 3(a) and an output fiber tube on the right. The tube was designed to hold the fiber so that the body of fiber will not be bent during the measurement, which could introduce additional signal loss and hence adversely affect the results as this is would not be due to the pivot action required by this measurement. The input fiber holder hollow tube (with an internal diameter of 1mm) holds the single input POF that is connected to the proximal LED source. The output holder tube has three connected holes each with a 1mm diameter bore. This tube therefore accommodates the three output POFs each of which is independently connected to a distal photodiode detector. The bases of the two tubes, which form the attachment point to the skin are separated 30mm apart via two legs or pods which are integral to the housing (fig. 3(a)). The base pod that is 30mm width and 15mm in length can be attached on the lower back body using clinical adhesive tape. The base pod was designed to be big enough such that it could be firmly attached to the patient's skin during the testing and comfortable to wear for all different sized patients.

Both tubes were subsequently centered (aligned) and then locked together in this position. A clearer structure of the sensor hinge is shown in fig. 3(b). One side of the input fiber tube was designed with a profile of a Philips screw head to allow the sensor bending in only lateral and sagittal directions. Input fiber tube is fitted into the output fiber tube and locked together using a flexible silicon tube that allows the entire sensor to be flexible during bending. The flexible part of the sensor is coated with another layer of silicon mold gel that solidifies when left to dry for a few hours. The purpose of the silicon mold is to improve the robustness of flexible region and it assists in holding both the tubes together for added strength and robustness of the sensor unit.

The optical signals at the output of the three POFs are detected using standard Silicon photodiodes (SFH 250V) that are connected to a photoamplifier and filter circuit (fig. 3(d)), the analog output of which is connected to a computer via a National Instruments type NI USB-6008 DAQ data acquisition unit. The data is captured using LabVIEW ™ software stored on a laptop PC for presentation to the clinicians and subsequent analysis. Fig. 3(c) shows a photograph of the 3D printed bending sensor that is ready to be attached directly on the patient's lower back to measure the sagittal and lateral bending.

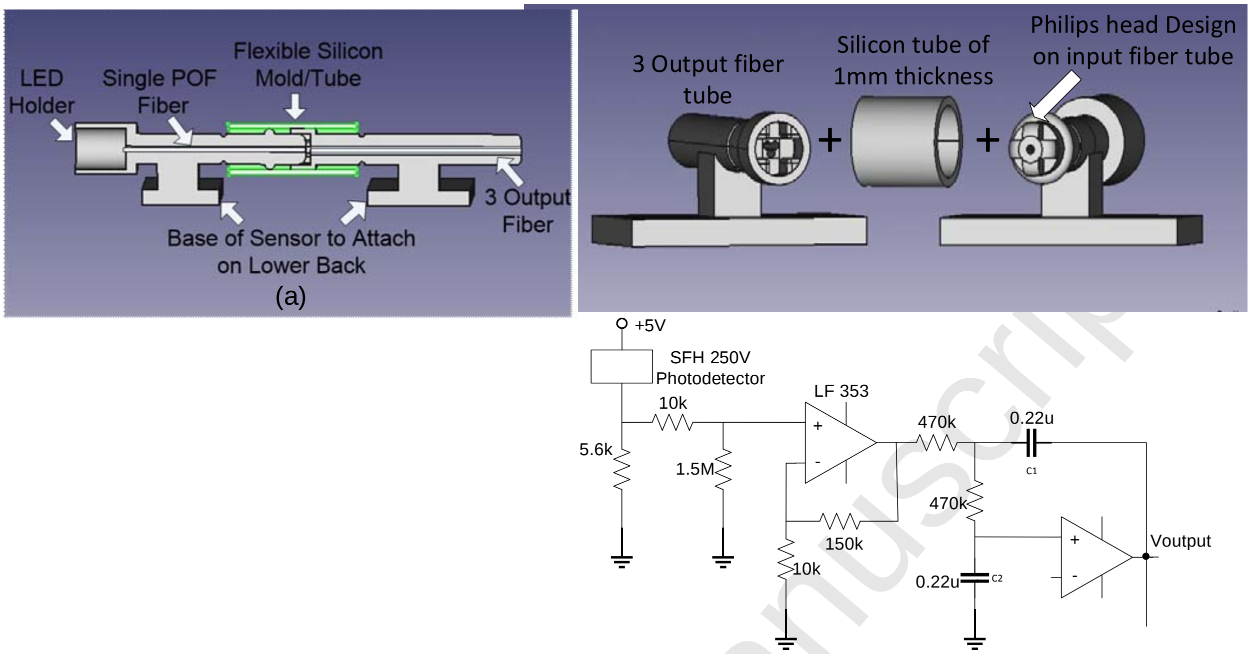


Fig. 3 (a) Cross-section diagram of sensor prototype for printing. (b) Structure of the sensor hinge and assembly (c) Photograph of the lower back bending sensor. (d) Electronic circuit for filtering and amplifying signal.

2.3 Sensor Calibration

The sensor was fixed on an optical bench setup which consists of two translational stages, two optical posts, two holders with clips and a rotational stage. Fig. 4(a) illustrates the configuration for lateral bending measurement. Side A of sensor's base was fixed horizontally to the translational stage with two optical posts, side B was attached to the rotational stage (which has a ±0.5^o accuracy). For the sagittal axis measurement, the sensor was mounted vertically on the rotational stage as shown in fig. 4(b). For both configurations the output fiber's intensity was recorded and the ratio calculated (in the LabVIEW ^{TM} software) in near real-time while rotating the stage. In the case of both lateral and sagittal bending, the stage was rotated with an angle of 2^o increment with a time interval of 30 seconds starting from 0^o. The rotation continued until reaching the output ratio saturation value at which no further changes in the output ratio in response to the angle change could be observed. At this point the rotation angle was slowly decreased with a 2^o interval until returned to its original 0 degrees position. A finer calibration was also performed by decreasing the angle steps to 0.5^o using a fine tuning adjustment on the stage in a range up to 4^o. The whole process was repeated three times to assess repeatability and hysteresis.

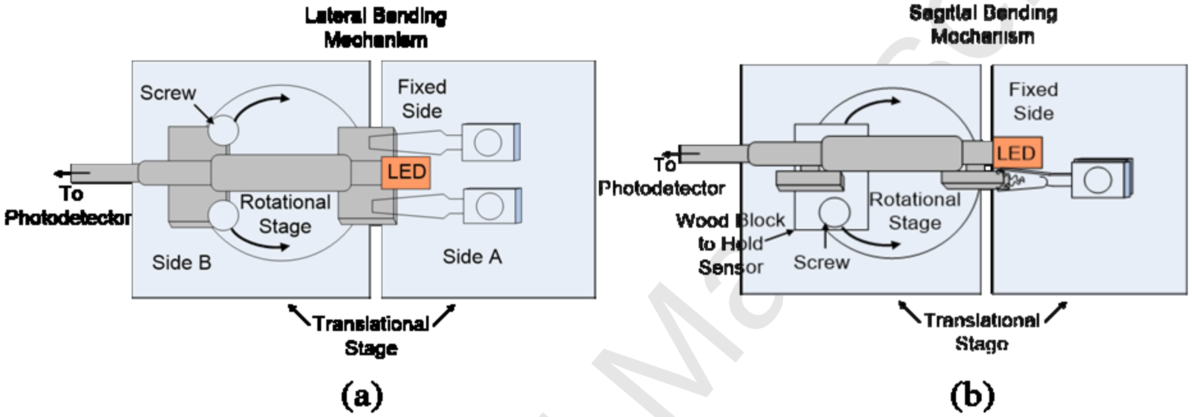


Fig. 4 (a) Picture and schematic diagram of experiment setup for sensor characterization on lateral bending and (b) sagittal bending.

Fig. 5 illustrates the voltage output of each of the three fiber channels (corresponding to received intensity) for bending at every step angle of 2^o during calibration. From fig. 5(a), the intensity of the light coupled to fiber I1 increases while the intensity to fiber I2 and I3 decreases when the sensor bends to the left as defined in the previous section. At a bending angle of 12^o, the saturation point is reached where fiber I3 receives a negligible amount of light coupled from the input fiber. On the other hand for lateral bending to the right, the output voltage of fiber I2 increases while I1 and I3 decrease at larger bending angle. For sagittal extension bending, fiber I1 and I2 increases when the bending angle increases at step angle of 2^o. It can be seen that at angle around 10^o-12^o, fiber I1 and I2 slightly decreases which corresponds to the saturation point of sensor. In contrast, fiber I1 and I2 decrease with an increase in I3 during sagittal bending in the opposite direction. By utilizing the formula shown in equations 1 and 2, the sensor was calibrated according to the bending angle from the output ratio of the three receiving fibers.

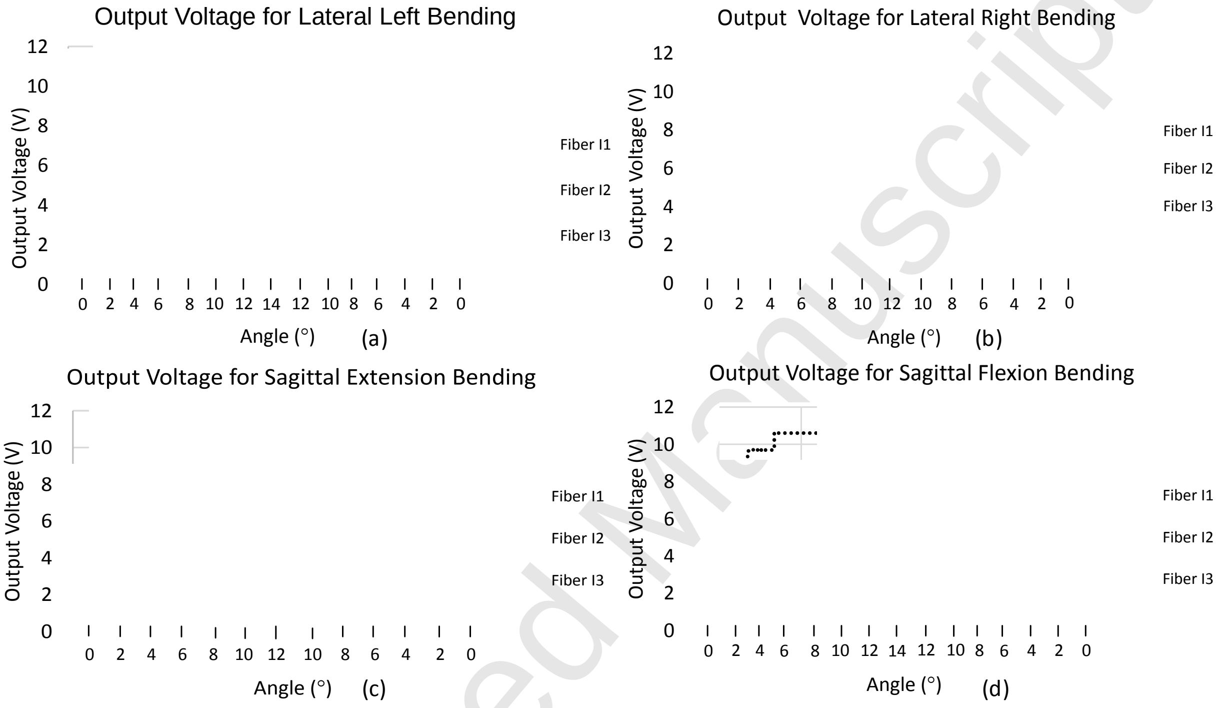


Fig. 5 Output voltage value for three output fiber at different bending angle for calibration.

2.4 The Protocol for the Clinical Bending Exercises and Measurement of the different bending postures on the back of the participants using the optical fiber sensor

The main purpose of the sensor in this investigation is to aid and to allow clinical staff to accurately monitor lumbar spine movement in the clinic. The device is to be placed at the lumbar spine section, which is a commonly reported pain area among patients [17]. The objective is to measure the bending angle when the patient is requested to present different postures or positions. For this preliminary test, the experiment was limited to only two subjects (a male and a female) in order to check the initial response of sensor when placed on a human subject whilst undergoing some standard motion exercises. It was not intended to be a clinical protocol for use in people with spinal pain. A further more comprehensive trial involving 15 to 20 patients is planned and will be reported in a separate article following this publication.

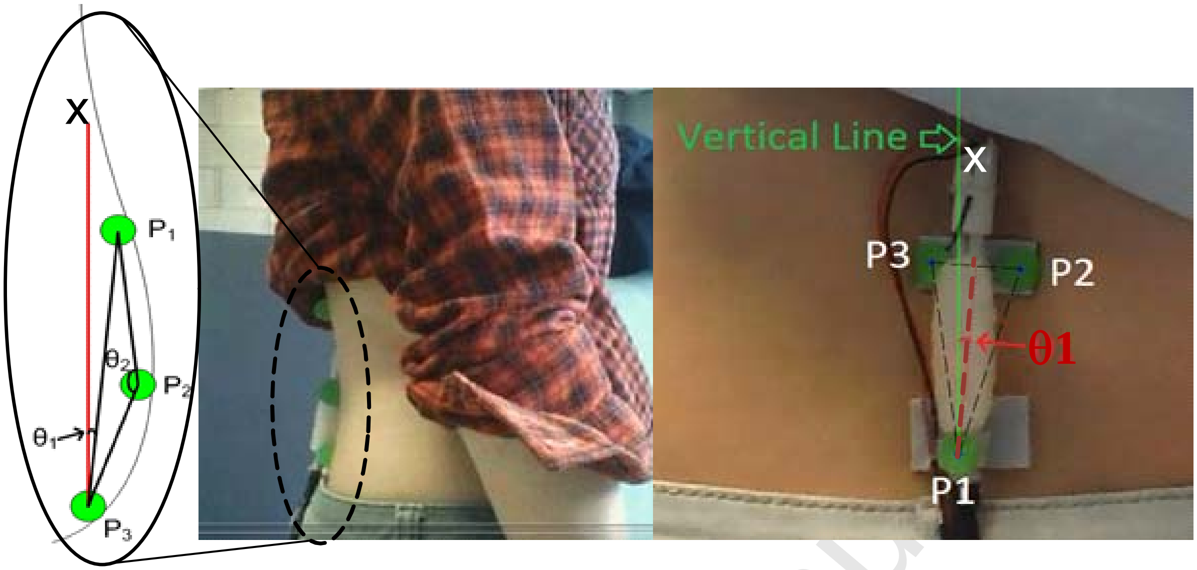
Two healthy volunteers without back injury and spinal deformity assumed several static postures. For the sagittal bending test, the subject was asked to initially stand straight and relax followed by bending to the front with arms straight down beside them (sagittal front). These were followed by extension to the back (sagittal back) following returning to the middle straight posture. The lateral bending test commenced with being in relaxed standing posture, followed by full bilateral trunk side flexion (lateral left and right) and then returned to a neutral straight standing position. During the first 15sec of each experimental trial, the subjects were requested to stand and relax and any posture change was measured and recorded and the position maintained for another 15sec at each trial.

2.5 Image Acquisition

For the purpose of this investigation, the optical fibre sensor's output signal was correlated to visual observation of the lower back curvature. This was recorded using a standard digital imaging webcam which was used to provide validation of the optical fibre sensor measurements, thus providing efficacy of the data recorded using the optical fiber sensor of this investigation. An ad-hoc image processing scheme was applied to study the relationship between the appearance of back curvature and the angle obtained from the sensor. To examine both lateral and sagittal bending situations, three green marks were placed at different positions on the low back area to assist the visualization of the bending on the lower back at various postures during the measurement session. This is standard practice in physiotherapy clinics and thus quantitative data could be extracted from the recorded examination process by manually tracing the location of the three spots.

For the sagittal bending assessment, it is hard to acquire an exact bending angle where the sensor is attached through imaging due to the curvature of spine. Hence the sensor measured data is correlated with the overall inclination of lower back bending and spine curvature of each individual. Two green spots were located one each side of both of the sensor's base pods while another spot was located on the middle of the back (for our participants this was about 2 cm above the tip of the sensor). As depicted in fig. 6(a), the left side of fig. 6(a) shows two angles θ_1 and θ_2. The angle θ_1 for sagittal bending refers to the overall bending of the back from a vertical line. It is measured between a vertical line passing through point P_3 (P_3X) and a line connecting P_1 to P_3 (P_1P_3). θ_2 is the angle between the two lines P_2P_3 and P_2P_1. This angle represents the localized change of lower spine curvature at different postures. The curvature of the lumbar spine varies between individuals.

For lateral bending, two spots were placed on each side of the upper sensor and one at the middle end of lower sensor as depicted in fig. 6(b). The overall bending angle θ_1 was calculated between a vertical line passing through P_1(P_1X) and a line connecting P_1 to the middle of P_2P_3. For lateral bending, video footage was captured during the whole bending process to compare the trend of bending angle for both sensor and imaging. In both bending cases, a webcam with resolution 1280x720 pixels was used to record the examination process.



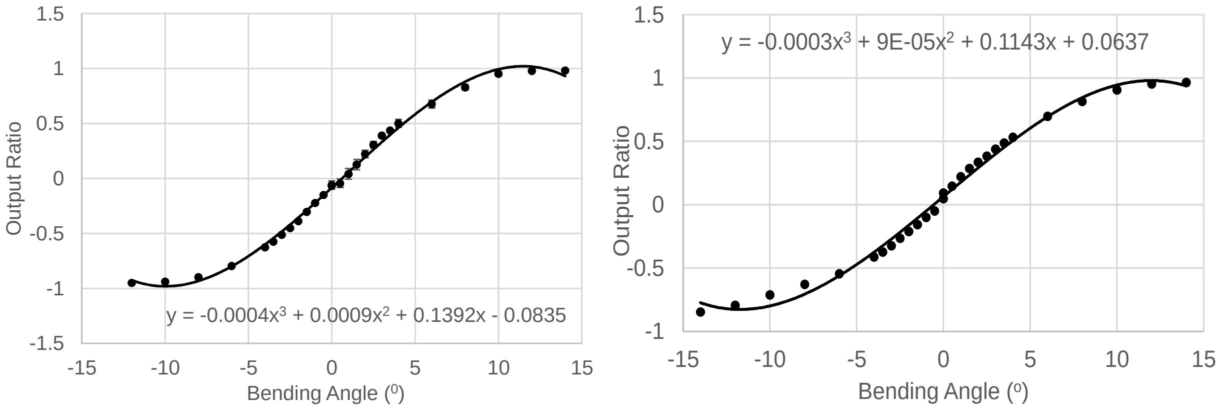
(a) (b)

Fig. 6 Placement of 3 green reflective markers on body in relation to the sensor module of this investigation. (a) The posture change was defined as the inclination change at angle θ_1 and θ_2 for sagittal bending. (b) Posture change defined as angle θ_1 for lateral bending.

3. Results and Discussion

3.1 Bending Assessment with optical component

Fig. 7 shows the bending test results obtained for both sagittal and frontal plane. The results show that the device exhibits a non-linear response where saturation is reached at bending angles around 12^o in both directions. An almost linear response is observed between -4^o to $+4^o$, with a correlation coefficient, R^2 of 0.9971 for the frontal plane and 0.9928 for the sagittal plane when data is fitted into a linear trend. The non-linearity observed in the bending data can be attributed to the bending at the large angle values causing increased misalignment between the input and output optical fibres and this in turn results in an increase in departure from linearity of the received signal arising from an exacerbated loss in coupling between the transmission and receiving fibers of the sensor. In fig. 7(a), the sensor's response for a lateral bending angle between -12^{o} and +12^{o} has been fitted to a third order polynomial and is shown as the solid line in this figure. For the sagittal bending in fig. 7(b), the polynomial fit is not ideal for angles lower than -6^{o}. This is due to the asymmetrical response of the sensor during the sagittal bending as the higher intensity of light coupled to two fibers during bending in the positive direction and more intensity of light coupled to only one fiber when bending was in negative direction. These polynomials were used during the actual measurements to interpret the sensor output as bending angles. This was performed within a LabVIEW ™ programme developed by the authors specifically for this investigation.

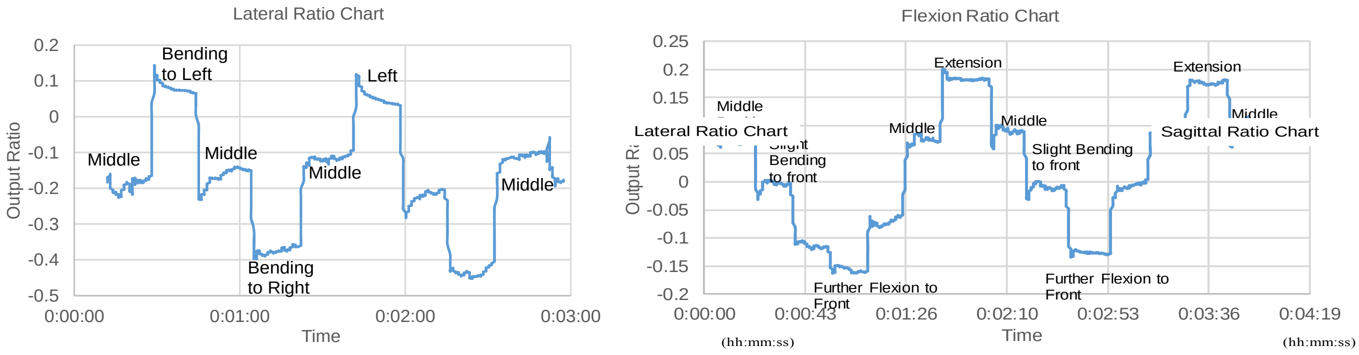


(a) (b)

Fig. 7 Intensity output ratio of sensor versus bending angle for both (a) frontal plane (lateral) and (b) sagittal plane and the polynomial fit for interpreting the angle value from the ratio measurement.

3.2 Measurement on patient's lower back

Fig. 8 shows two sample results recorded for lateral bending and sagittal bending ratio versus time when the sensor was placed on the lower back of a participant as shown in fig. 6. In the case of the lateral bending measurement, the user was initially in a normal upright standing (middle) posture. The patient then bends over towards the lateral left and holds this posture for 15 sec, slowly returning to the original middle position and then repeating the bending to the right. In this sensor configuration, bending to the left shows an increase in the output ratio while bending to the right is represented by a decrease in output ratio values. A slight shift from the zero position can be observed due to the fact that when the patient is not bending, the back is not completely straight. This could also be due to user error when placing the sensor in its rest position, but not being exactly straight. For sagittal plane bending assessment, the patient starts from the middle standing posture and slowly bends forward for different angles for forward bending before returning to the middle posture and extending to the back (reverse bending). The output ratio for sagittal bending decreases (to a negative value) when bending forward. It however increases (to positive values) when extending backward.



(a)

Fig. 8 Sensor output ratio versus time for (a) lateral and (b) sagittal bending as tested on the back of a participant.

The output ratios obtained from fig. 8 were converted into bending angles using the curves in fig. 7(a) and 7(b) for lateral and sagittal bending. The angle values resulting from this were compared with the values extracted from the image acquisition described in section 2.5. Fig. 9 shows the relationship between the bending angles θ_1 and θ_2 (as defined in fig. 6(a)) and the bending angles obtained from the sensor for sagittal bending. Two subjects of approximately the same age and in good health were tested: subject A is a female and subject B is a male. The sagittal bending shows a good fit with a second order polynomial response when subject A and B bends at θ_1 between -15^o to 7^o. The slight difference between the value of bending is depends on movement of individual subject in sagittal plane. Both subjects also show a second order polynomial response for bending angle corresponding to imaging angle θ_2 (as in fig. 9(b) and 9(d)). This large imaging angle θ_2 is the arch angle of individual lumbar spine during bending exercise and hence the difference in the angle relationship between two subjects can be attributed to different arch angle of lumbar spine of both subjects.

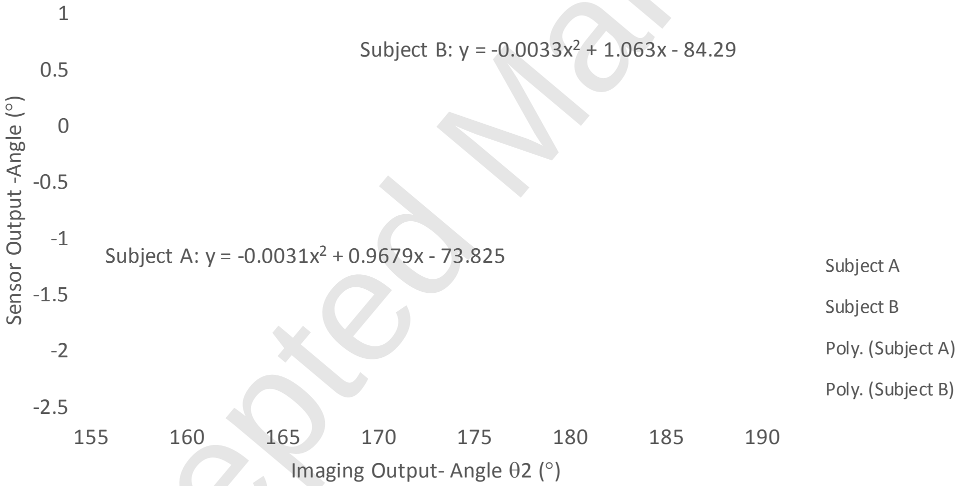


Fig. 9 Relationship between measured optical fiber sensor angle and the angle captured from the imaging system θ_1 and θ_2.

Fig. 10 shows the time resolved output angles from the optical fiber sensor and imaging angle θ_1 during the standard lateral bending exercise for subject A and B. For lateral bending, it is easier to compare the sensor and image angle as three spots were placed on the sensor and the movement of the sensor which can be more easily captured for analyzing. Fig. 11 compares the sensor with imaging results and the relations between both data. Both sets of results from Fig 10 were placed together on the same axis and was aligned to 0^o in the straight positions to reduce human error when placing the sensor. Positive angle is obtained on the left bending and negative value corresponds the right bending angle. The trend of bending for both subjects is in agreement with the imaging results but both subjects have different responses as the bending and body shape of subjects vary individually.

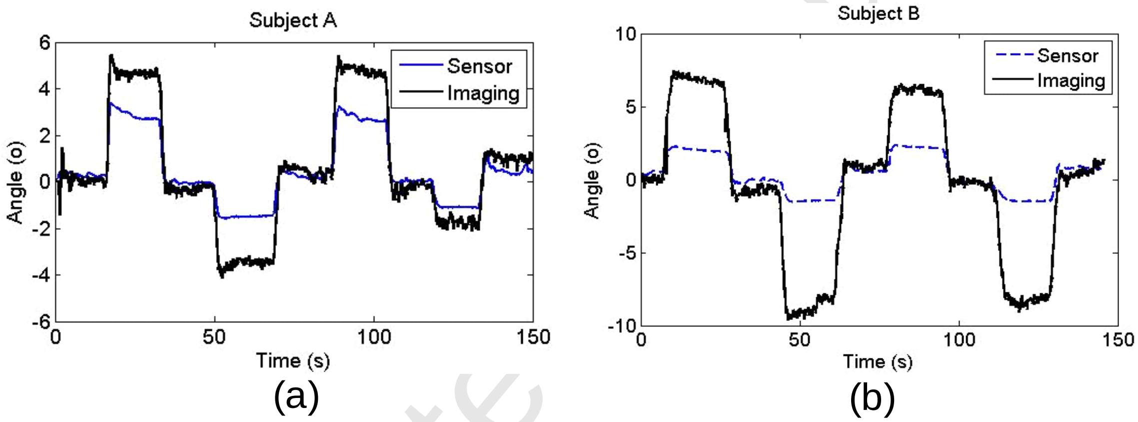


Fig. 10 The time resolved lateral output angle response of the optical fiber sensor and imaging system for the core of subject A and subject B.

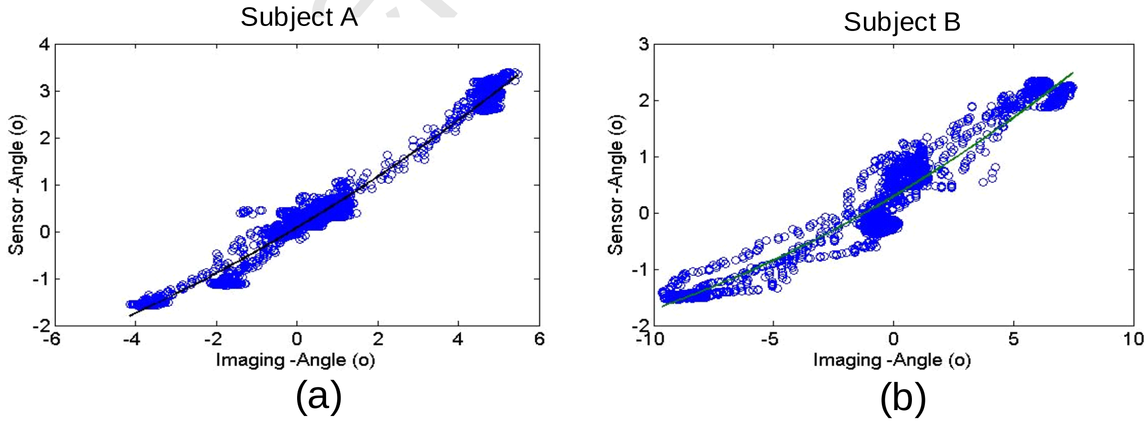


Fig. 11 The relationship between the measurement data obtained from sensor versus imaging angle θ_1 for the (a) subject A and (b) subject B.

4. Conclusion

An optical fiber intensity modulated sensor for measuring the bending angle of lumbar spine region for clinical assessment has been designed and fabricated. The sensor utilizes a single fiber for transmitting the light and three optical fibers for determining the direction and degree of deformation of the spine. The fibers are enclosed in a robust plastic enclosure which was fabricated using 3D printing. The all-plastic composition of the sensor has the added advantage that it can be used successfully in conjunction with Magnetic Resonance Imaging (MRI) scanning machines as well as X-Ray based scanning machines. This simply entails the removal of the LED from the unit and connecting a longer length of fiber to connect to the LED. This would incur minimal optical loss as the light is already fiber coupled in the existing design.

Experimental results were captured from the sensor mounted on human subjects and these have been compared with angular deformation values obtained using a simultaneous image capture method. Although the absolute values of the angles obtained using the optical fiber sensor and the imaging method are not the same, the results have shown a reproducible set of results for spinal bending in both sagittal and lateral planes. The compact design and ease of attachment coupled with the fact that it is relatively non-invasive make the sensor of this investigation suitable for widespread use in clinical settings.

Acknowledgements

This work presented in this paper has been supported by the LEADERS project, funded by the European Union as part of the Erasmus Mundus scholarship Programme (2014-0855). The authors are also grateful to the staff from Department of Clinical Therapies for all their support during the testing of this sensor. The authors would also like to acknowledge and thank the group from Design Factors in University of Limerick for all their support and help for the 3D printing of this sensor.

References

[1] P.C. Davis, F.J. Wippold, J.A. Brunberg, R.S. Cornelius, L. Robert, P.D. Dormont, et al., ACR Appropriateness Criteria® on low back pain, Journal of the American College of Radiology, 6(2009) 401-7.

[2] R. Deyo, Back Pain Patient Outcomes Assessment Team (BOAT), US Department of Health & Human Services-Agency of Healthcare Research, (1994).

[3] B. Cakir, M. Richter, W. Käfer, M. Wieser, W. Puhl, R. Schmidt, Evaluation of lumbar spine motion with dynamic X-ray–a reliability analysis, Spine, 31(2006) 1258-64.

[4] G.P. Mellin, Accuracy of measuring lateral flexion of the spine with a tape, Clinical Biomechanics, 1(1986) 85-9.

[5] R.G. Burdett, K.E. Brown, M.P. Fall, Reliability and validity of four instruments for measuring lumbar spine and pelvic positions, Physical therapy, 66(1986) 677-84.

[6] W.Y. Wong, M.S. Wong, Detecting spinal posture change in sitting positions with tri-axial accelerometers, Gait & Posture, 27(2008) 168-71.

[7] W.Y. Wong, M.S. Wong, Trunk posture monitoring with inertial sensors, European Spine Journal, 17(2008) 743-53.

[8] G.J. Donatell, D.W. Meister, J.R. O'Brien, J.S. Thurlow, J.G. Webster, F.J. Salvi, A simple device to monitor flexion and lateral bending of the lumbar spine, IEEE Transactions on Neural Systems and Rehabilitation Engineering, 13(2005) 18-23.

[9] M.W. Whittle, D. Levine, Three-dimensional relationships between the movements of the pelvis and lumbar spine during normal gait, Human Movement Science, 18(1999) 681-92.

[10] J.-H. Shin, S. Wang, Q. Yao, K.B. Wood, G. Li, Investigation of coupled bending of the lumbar spine during dynamic axial rotation of the body, European Spine Journal, 22(2013) 2671-7.

[11] L. Hackenberg, E. Hierholzer, V. Bullmann, U. Liljenqvist, C. Götze, Rasterstereographic analysis of axial back surface rotation in standing versus forward bending posture in idiopathic scoliosis, European Spine Journal, 15(2006) 1144-9.

[12] M.A. Zawawi, S. O'Keeffe, E. Lewis, Plastic optical fibre sensor for spine bending monitoring with power fluctuation compensation, Sensors, 13(2013) 14466-83.

[13] L.E. Dunne, P. Walsh, B. Smyth, B. Caulfield, Design and evaluation of a wearable optical sensor for monitoring seated spinal posture, 2006 10th IEEE International Symposium on Wearable Computers, IEEE2006, pp. 65-8.

[14] J.M. Williams, I. Haq, R.Y. Lee, Dynamic measurement of lumbar curvature using fibre-optic sensors, Medical engineering & physics, 32(2010) 1043-9.

[15] L. Bilro, J. Oliveira, J. Pinto, R. Nogueira, A reliable low-cost wireless and wearable gait monitoring system based on a plastic optical fibre sensor, Measurement Science and Technology, 22(2011) 045801.

[16] M.A. Zawawi, S. O’Keeffe, E. Lewis, Optical fibre bending sensor with automatic intensity compensation, Journal of Lightwave Technology, 33(2015) 2492-8.

[17] W. Dankaerts, P. O'Sullivan, A. Burnett, L. Straker, Differences in sitting postures are associated with nonspecific chronic low back pain disorders when patients are subclassified, Spine, 31(2006) 698-704.

Biographies

Wern Kam received the Bachelor of Science degree in physics from University of Malaya, Kuala Lumpur, Malaysia in 2012. She received M.Sc. degree in Physics in 2015. Currently, she is pursuing the PhD degree in Electronic and Computer Engineering, University of Limerick. Her current research interests include fiber optic sensor and microfiber.

Kieran O'Sullivan graduated from University College Dublin in 1999. In 2004 he completed an MSc in Manipulative Therapy at Curtin University Western Australia. He completed his PhD in 2012, which examined the role of sitting posture in low back pain. He is currently Lead Physiotherapist at the Sports Spine Centre at Aspetar in Qatar, and Senior Lecturer in Physiotherapy at the University of Limerck, Ireland.

Mary O'Keeffe is a chartered physiotherapist and low back pain researcher in the University of Limerick. Mary's PhD research examined whether tailoring multidimensional rehabilitation to the individual chronic low back pain patient enhances effectiveness, and is worth the additional time (and costs!) involved. Her PhD research was funded by the Irish Research Council. Mary has been recently awarded a Marie Skłodowska Curie Individual Global fellowship by the European Commission.

Sinead O'Keeffe received the B.Eng. (Hons.) degree in electronic engineering and the Ph.D. degree from the University of Limerick, in 2003 and 2006, respectively. She has recently returned to the Optical Fibre Sensors Research Centre from Liverpool John Moores University and she is currently a Research Fellow through the Marie Curie Re-Integration Fund and is co-funded through PRTLI IV Funding.

Waleed Mohammed received his B.Eng (Hons) degree in Electronics and Electrical Communications and M.Sc. in Computer Engineering in Cairo University, Egypt. He joined the College of optics and photonics, University of Central Florida, USA and received M.Sc in optics in 2001. In 2004 he completed his Ph.D work. Currently, he is a research scholar at the School of Engineering, Bangkok University. His research includes optoelectronics, nanotechnology, sensor and sensors network.

Elfed Lewis received the B.Eng. (Hons.) degree in electrical and electronic engineering and the Ph.D. degree from Liverpool University, in 1981 and 1987, respectively. In 1996, he founded the Optical Fibre Sensors Research Centre, University of Limerick, and appointed as the Director. He is currently a Fellow of the Institute of Physics, the Institution of Engineering and Technology, and the British Computer Society. He is also a Distinguished Lecturer of the IEEE Sensors Council from 2013 to 2015.