An optical fiber Bragg grating force sensor for monitoring sub-bandage pressure during compression therapy

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Abstract: Graduated compression bandaging of the lower limbs is the primary therapy for venous leg ulcers with its efficacy believed to be predominantly dependent on the amount and the distribution of the compressive pressure applied. There has been on-going demand for an ideal sensor to facilitate *in-vivo* monitoring of the sub-bandage pressure. Several methods and devices have been reported but each has its limitations, such as bulkiness, low tolerance to movement, susceptible to thermal noise and single point sensing. An optical fiber force sensor is demonstrated, consisting of two arrays of fiber Bragg grating (FBG) entwined in a double helix form and packaged with contact-force sensitivity. This sensor array has inherent temperature immunity and is capable of real-time, distributed sensing of sub-bandage pressure. The calibration results of the sensor array, as well as the validation human trial results, are presented.

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References and Links

1. J. J. Bergan, G. W. Schmid-Schönbein, P. D. Smith, A. N. Nicolaides, M. R. Boisseau, and B. Eklof, "Chronic venous disease," N. Engl. J. Med. **355**(5), 488–498 (2006).
2. N. J. London and R. Donnelly, "ABC of arterial and venous disease. ulcerated lower limb," BMJ **320**(7249), 1589–1591 (2000).
3. H. Partsch, M. Clark, S. Bassez, J. P. Benigni, F. Becker, V. Blazek, J. Caprini, A. Cornu-Thénard, J. Hafner, M. Flour, M. Jünger, C. Moffatt, and M. Neumann, "Measurement of lower leg compression in vivo: Recommendations for the performance of measurements of interface pressure and stiffness: Consensus statement," Dermatol. Surg. **32**(2), 224–232, discussion 233 (2006).
4. D. Wertheim, J. Melhuish, R. Williams, and K. Harding, "Measurement of forces associated with compression therapy," Med. Biol. Eng. Comput. **37**(1), 31–34 (1999).
5. J. Hafner, W. Lüthi, H. Hänssle, G. Kammerlander, and G. Burg, "Instruction of compression therapy by means of interface pressure measurement," Dermatol. Surg. **26**(5), 481–486, discussion 487 (2000).
6. T. B. Raj, M. Goddard, and G. S. Makin, "How long do compression bandages maintain their pressure during ambulatory treatment of varicose veins?" Br. J. Surg. **67**(2), 122–124 (1980).
7. S. Thomas, "The use of the Laplace equation in the calculation of sub-bandage pressure," EWMA J. **3**, 21–23 (2003).
8. J. Schuren and K. Mohr, "The efficacy of Laplace's equation in calculating bandage pressure in venous leg ulcers," Wounds UK **4**, 38–47 (2008).
9. M. D. Steinberg and E. D. Cooke, "Design and evaluation of a device for measurement of interface pressure," J. Biomed. Eng. **15**(6), 464–468 (1993).
10. J. C. Barbenel and S. Sockalingham, "Device for measuring soft tissue interface pressures," J. Biomed. Eng. **12**(6), 519–522 (1990).
11. M. Ferguson-Pell, S. Hagisawa, and D. Bain, "Evaluation of a sensor for low interface pressure applications," Med. Eng. Phys. **22**(9), 657–663 (2000).
12. Y. Rao, "In-fibre Bragg grating sensors," Meas. Sci. Technol. **8**(4), 355–375 (1997).

#190858 - $15.00 USD Received 20 May 2013; accepted 4 Aug 2013; published 15 Aug 2013

13. J. W. Arkwright, N. G. Blenman, I. D. Underhill, S. A. Maunder, M. M. Szczesniak, P. G. Dinning, and I. J. Cook, "In-vivo demonstration of a high resolution optical fiber manometry catheter for diagnosis of gastrointestinal motility disorders," Opt. Express **17**(6), 4500–4508 (2009).
14. Z. Moore, "Compression bandaging: are practitioners achieving the ideal sub-bandage pressures?" J. Wound Care **11**(7), 265–268 (2002).

1. Introduction

Chronic leg ulceration is one of the most common symptoms associated with venous disease that debilitates a wide range of the human population [1, 2]. The principle therapy for such disease is generally considered to be graduated compression bandaging of the diseased limbs [1–3], which is believed to control edema and assist venous blood circulation. The success of the therapy not only depends on the pathological evaluation of the ulcers but, more importantly, the amount and the distribution of the compressive pressure applied can significantly influence the treatment outcomes [3–5]. Insufficient pressure would impair the efficacy of the healing process while excessive compression can result in aggravated tissue damage. When treating patients with different disease states and leg profiles, inconsistencies in the wrapping techniques of wound care professionals could increase the uncertainty of the therapy outcome. Furthermore, the sub-bandage pressure is likely to change over time as the tension in the fabric relaxes [6] or as a result of the movement of the patient's limb [4].

Theoretically, the sub-bandage pressure exerted on a cylindrical object can be estimated by the modified Laplace Equation [7, 8] which states that the compression is directly proportional to the bandage tension and the number of layers, and inversely proportional to the circumference of the object and the bandage width. However, the reliability of applying the Laplace Equation in compression therapy remains controversial due to the irregularity and inhomogeneity of the human limb, as well as the unpredictable long-term drift of the sub-bandage pressure [8]. Consequently there has been a great demand for continuous quantitative measurements of the sub-bandage pressure as the patients rest (resting pressure) and move (working pressure) during compression therapy.

The key characteristics of an ideal sub-bandage pressure sensor for monitoring compression treatments have been summarized by previous authors [3]: The physical profile of the sensor should be thin, flexible and conforming to the curvature of the limb such that it can be applied over an extended period without interfering with the treatment; The sensor performance should possess high pressure sensitivity, high sampling rate, linear response, long-term stability, low hysteresis, as well as insensitivity to temperature and patient movement. It is also desirable to have a high spatial resolution system in which a multiple of closely spaced sensing elements allows concurrent measurements along the whole length of the treated limb. In addition, the sensing area should be adjustable to provide the choice of detecting either the average pressure value over a region or localized high pressure points. A simple calibration procedure is also desirable.

In recent years, various types of sensor have been developed for the measurement of sub-bandage pressure. However, currently none of these sensors could satisfy all of the above specifications [3]. Pneumatic type sensors [9] in general have the preferred low profile design and are easy to operate, but suffer from low sampling rate and hysteresis. Liquid-filled sensors [10] can have highly linear response and reasonable sampling rate when measuring resting pressure, but they tend to be bulky when filled and become unreliable if the patient moves. Resistive type strain gauges [11] have high sampling rate and are typically insensitive to movement, nevertheless they are rigid and become inaccurate over curved surfaces. Moreover, none of these sensors have the high spatial resolution and adjustable sensing area that are desired for ideal sub-bandage pressure sensing.

Fiber Bragg gratings (FBG) are intrinsic sensing elements inscribed in the core of an optical fiber capable of monitoring variations in localized physical quantities such as force, pressure, strain and temperature [12]. FBGs provide key advantages such as small size, flexibility, wavelength-encoding nature and the ability for quasi-distributed measurement

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through wavelength division multiplexing (WDM) techniques. As a result, optical FBG sensors are becoming increasingly common in many industries and are facilitating new sensing applications that were previously unavailable, such as high resolution manometric study of the peristalsis in the human gut [13].

This paper reports a new fiber optic sensor array that was incorporated into a flat, flexible tape and is capable of being placed under compression devices to monitor the applied force, calibrated to be expressed as contact pressure. The sensor design provided linear response, inherent temperature insensitivity and high spatial resolution. The sensor array was interrogated at high sampling rate which enabled simultaneous measurement of both resting and working sub-bandage pressure throughout the length of a typical human lower leg. Results from the sensor characterization and the in-vivo trial on a human leg, supervised and performed by an experienced wound care nurse, is presented and discussed.

2. Fiber optic sub-bandage force/pressure sensor

The schematic diagram of the optical fiber force sensing element is shown in Fig. 1. The basis of the sensor array consisted of two Draw Tower Grating (DTG) arrays (supplied by FBGS Technologies GmbH, Germany) each containing 33 spectrally separated FBG elements covering the wavelength range 1571 nm to 1590 nm. The DTGs were written in low-bend-loss fiber during the drawing process which yielded a seamless structure that had high mechanical strength and more tolerance for device engineering. The physical length of the FBGs was 3 mm and the spatial separation between them was 10.4 mm.

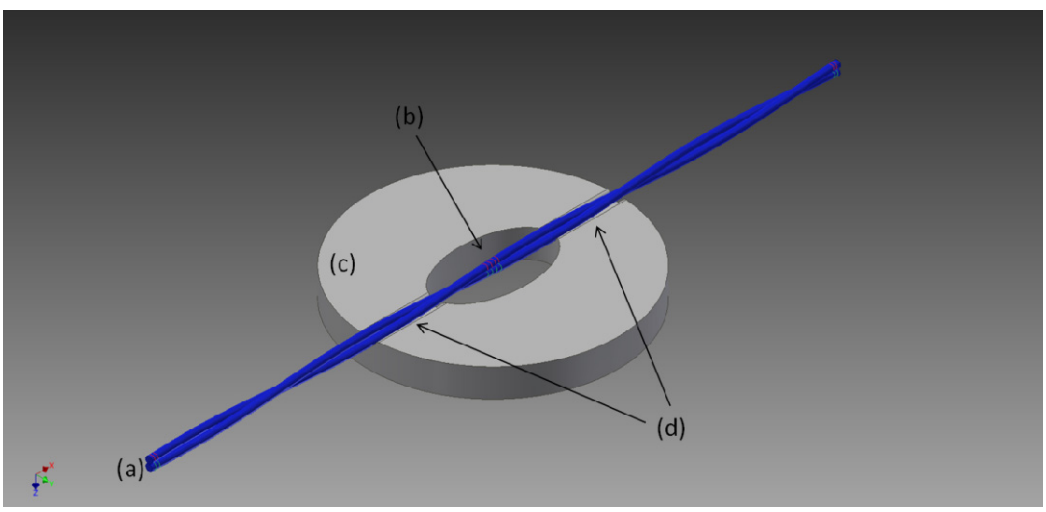

Fig. 1. Diagram of a single element on the sensor array showing: (a) double helix fiber array; (b) the location of the FBG pair; (c) the metal disc substrate; and (d) the regions of the bonds between the fiber array and the substrate.

The two optical fibers were placed in contact and entwined into a double helix form with a pitch matching the spatial separation of the FBGs. The positions of the FBG elements on both arrays were coincident and formed FBG pairs as sensing elements. Each FBG in the sensing pair describes an arc of approximately equal and opposite radius which provides an inherently opposite response to sideways deformation. For example, when force is applied from the top, the top FBG experiences compression which reduces the Bragg wavelength while the bottom FBG is elongated and the Bragg wavelength is increased. The degree of differential Bragg wavelength shift can then be used to infer the applied force. An important advantage of this intimate double helix structure of the FBG pairs is that any temperature fluctuation to the sensor would only cause a common mode variation which is independent of the force-induced differential wavelength shift. This feature is particularly beneficial for studies in close

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proximity to a heat source, such as a human body. Similarly, the design offers adequate immunity to axial strain since both fibers would be affected equally by strain along the axis of the helix. Furthermore, low hysteresis is expected due to the high rigidity and elasticity of the optical fiber.

Each FBG pair on the array was then fixed on a rigid metal disc substrate with a diameter of approximately 7.5 mm and a thickness of 0.9 mm. The substrate was hollow in the center and had a groove cut through the top such that it allows the vertically aligned FBG pair to be located in the cavity with freedom of sideways movement while supporting the entwined fibers at the horizontally aligned points. This configuration provides additional axial strain immunity and renders the FBG pair susceptible to contact forces acting only from the top-down direction, which would induce the differential Bragg wavelength shift as described above. In addition, the orientation of all the FBG pairs in the array was kept aligned with the placement of the substrates.

The finished sensor array was spliced with FC/APC connectors and bonded within two strips of wrapping tape, as shown in Fig. 2. The tape in contact with the sensing elements was a non-stretchable type in order to minimize perturbation during sensor handling while the bottom tape was elastic for retaining the overall flexibility. The final sensor assembly was still thin (~1.5 mm thick) and flexible enough to be comfortably placed under layers of compression bandage without the patient feeling any point pressure.

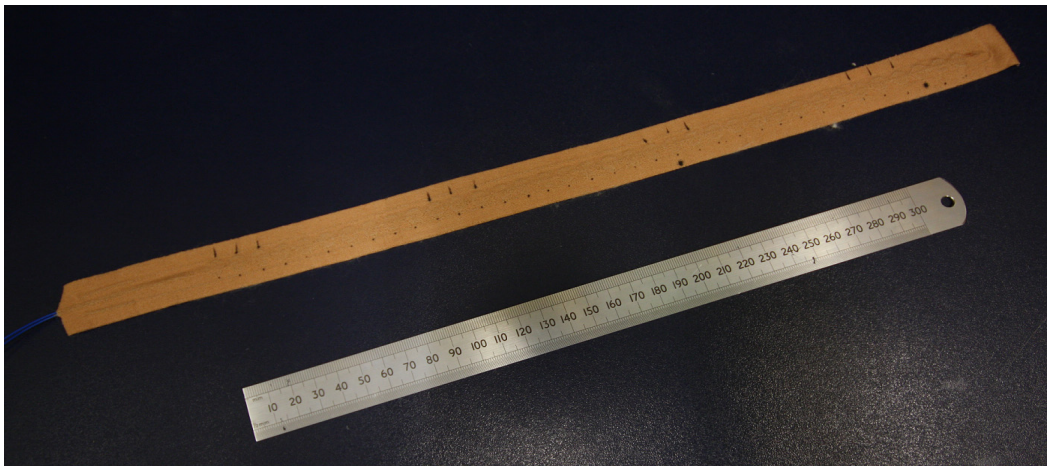

Fig. 2. Photograph of the assembled sub-bandage force sensor.

An FBG-Scan 804 interrogator (also supplied by FBGS Technologies GmbH) with a maximum sampling rate above 500 Hz was used to simultaneously monitor all of the wavelength peaks in the sensor array. In operation, the applied force could be inferred by comparing the differential wavelength shifts to values determined during an initial calibration of the device.

3. Sensor characterization

To calibrate the force sensor array for contact pressure, expressed in mmHg, a glass tube sealed at the bottom with a thin and flexible diaphragm was used. As shown in Fig. 3, the diaphragm was placed on top of the sensing element before water was gradually poured into the tube such that an increasing hydraulic head pressure would be exerted onto the sensor. At room temperature, every 10 cm of water height is equivalent to about 7.34 mmHg. It is worth mentioning that readings below 7 mmHg were neglected because the diaphragm resistance was still influential in the low pressure regime. The induced wavelength shifts were recorded against different pressure values and all of the sensing elements were calibrated in this

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manner independently. Figure 4 shows a typical pressure calibration result from one of the sensor elements in which both FBGs respond linearly to the applied pressure and oppositely to each other. The bottom FBG has a lower sensitivity possibly because of a slight asymmetry in the double helix structure. The differential wavelength shift, which was calculated in real-time, served as a good measure for the applied pressure with a sensitivity of ~4.8 pm/mmHg.

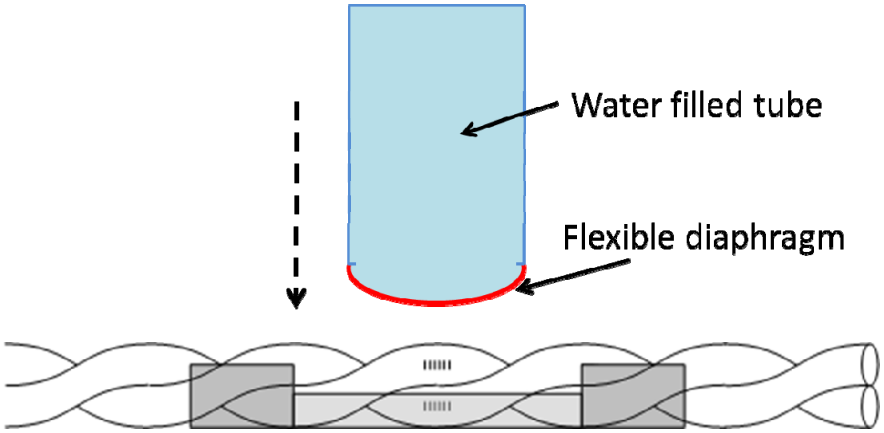

Fig. 3. Schematic of the pressure calibration setup.

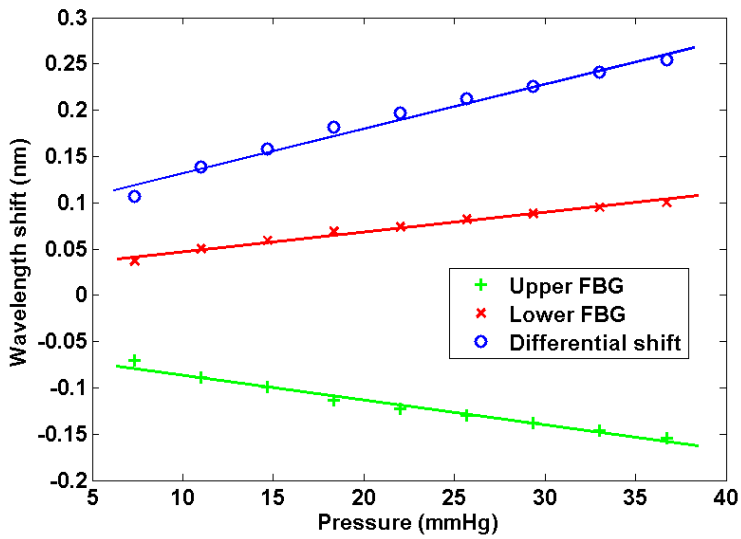

Fig. 4. Typical pressure response for the sensor array.

The thermal insensitivity and the repeatability of the sensor were also tested with the same calibration setup. In this instance the hydraulic pressure was kept constant at around 20 mmHg but the water temperature was elevated to 37°C. The diaphragm was slowly placed onto the sensor to apply pressure and was then removed with the same pace after 2 seconds. This action was repeated to verify the sensor repeatability. Figure 5 shows a typical result from a single sensor element in this measurement. The upper and the lower FBGs again responded to the contact pressure (at ~15 and ~42 seconds) but this time the thermal effect was present which raised the baseline responses of both FBGs. The thermal artifact appeared just before the diaphragm came into contact and during the pressure application it distorted the response profile. As the heated diaphragm was removed, the thermally elevated baseline

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persisted due to the residual heat trapped around the sensor. Identical features were observed on the repeated measurements. With this result it is clear that any temperature variation would significantly increase the nonlinearity and the uncertainty of the sensor response. However, this is effortlessly corrected with the double helix arrangement of the fiber arrays as the common mode effects cancel out each other. The resultant differential wavelength shift corresponds only to the applied pressure in the expected linear manner and the repeatability is also confirmed with negligible discrepancy between individual measurements.

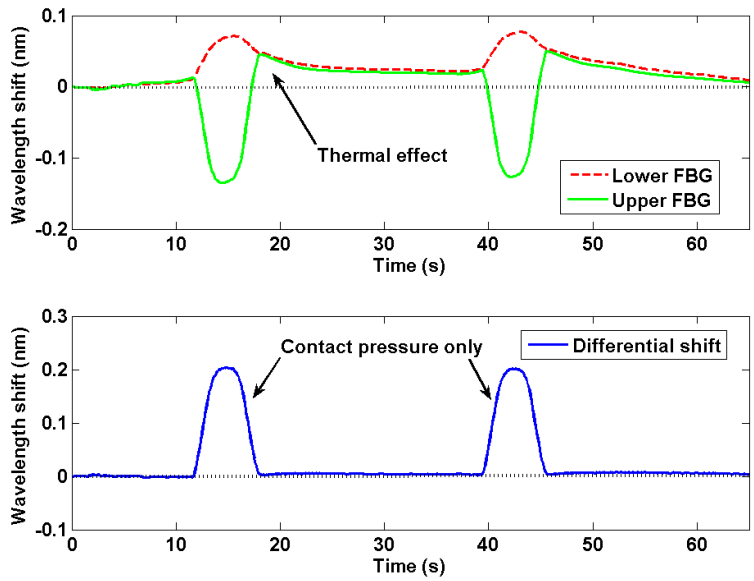

Fig. 5. Typical sensor response under constant pressure and elevated temperature. The thermal artifact was present in individual FBG readings but was eliminated in the differential response.

The long term stability of the sensor was tested by applying a constant pressure on the sensor for an extended period. Variation was found to be less then ± 1 mmHg over two hours.

4. In-vivo sub-bandage pressure measurement

The general consensus of an effective graduated compression treatment is that the compression bandages should exert a pressure of ~40 mmHg at the ankle and decreasing linearly up the leg to ~20 mmHg below the knee [14]. However, the relevance of this theoretical gradient profile to the actual healing rate has not been fully verified due to the lack of a high spatial resolution sensor that is capable of monitoring both resting and working sub-bandage pressure gradients across the treated limb.

To demonstrate the ability of the optical fiber sensor array for measuring in-vivo resting and working pressure profiles during a compression treatment, a three-layer bandaging system was applied to the leg of a healthy volunteer with the sensor array in place to measure the exerted compression. In the first instance, two trials were conducted to compare the achieved resting sub-bandage pressure profiles as the bandaging system was applied with and without real-time feedback from the instantaneous pressure recorded by the sensor array, which was displayed in the form of a histogram plot. The bandaging was carried out by a hospital research nurse (one of the authors) who has over 10 years experience in chronic wound care nursing. As shown in Fig. 6, the first padding layer of the bandaging system provided a non-pressurized support on the leg to protect the skin and reduce the slippage of subsequent layers. The sensor array was then placed on top of the padding along the fleshy

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side of the shin (tibialis anterior), extending from the ankle to just below the knee, where no bony prominences or tendons could influence the measured pressure. The second layer was the high compression bandage (SurePress High Compression Bandage by ConvaTec Inc.) for exerting the graduated pressure, it was applied in a spiral fashion from toe to knee. The measurement was on-going during the application of the compression bandage. The final layer was a cotton crepe bandage that held the lower layers in place without any further significant applied pressure.

Fig. 6. Photograph of the in-vivo sub-bandage pressure measurement. The sensor array sits on top of the first layer while the compression bandage was being applied.

The resting sub-bandage pressure profiles measured immediately after both bandaging are shown in Fig. 7. The dashed line in the graph represents the theoretically optimum gradient for a compression therapy but it serves only as a reference and was not the intended target to achieve for this proof-of-concept study. The red histogram represents the result from the blinded test in which the nurse was asked to apply the bandages relying only on her experience. It can be seen that in the blinded test the sub-bandage pressure profile did not follow a desired graduated gradient and random high compression regions were created in the medial section of the leg. The blue histogram shows the sub-bandage pressure profile recorded as the bandaging was performed with the nurse referring to the real-time feedback display. In this test it is evident that, although realistically non-smooth, a graduated compression was achieved from the ankle to knee. This outcome demonstrates the benefit of having a real-time measurement display that provides the information for readjustment during bandage application in order to achieve a desired pressure profile. In this study the pressure value measured at each individual sensor location was displayed to show the overall pressure distribution, however, in instances when it is more instructive to monitor average pressures over an extended region, such as over the calf muscle, the outputs from a number of sensing elements can be averaged and displayed as a single value. This enables the number and effective size of the sensors to be re-configured at will by user defined inputs to the monitoring software.

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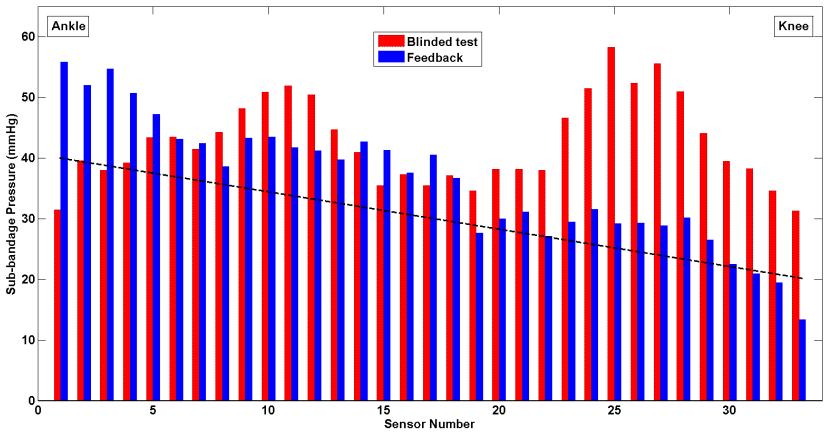

Fig. 7. Resting sub-bandage pressure profiles measured after the compression bandage system was applied under blinded condition and with feedback. The dashed line represents the theoretically optimum gradient as reference.

The working pressure was measured following the bandaging with feedback, the volunteer was asked to stand up and perform various simple movements and the variation in the sub-bandage pressure was recorded. Figure 8 shows the result measured by the sensor at the medial gaiter area during the activities. Features matching each physical activity can be observed: As the bandage was applied, the static pressure quickly built up and was maintained as the volunteer was sitting still; Dorsiflexion caused the muscle to pump and changed the circumference of the leg and the pressure value increased accordingly; When the patient stood up, the static pressure value elevated and increased further as knee bending was performed; Finally, sitting down again returned the sub-bandage pressure to the previous stationary value and additional dorsiflexion produced an almost identical response as before. These features match well with the previously reported data measured with localized single point sensors [3, 4]. Measurement results at other locations show very similar patterns with respect to the activities and they differ only in the amplitude and the time delay during the bandage application.

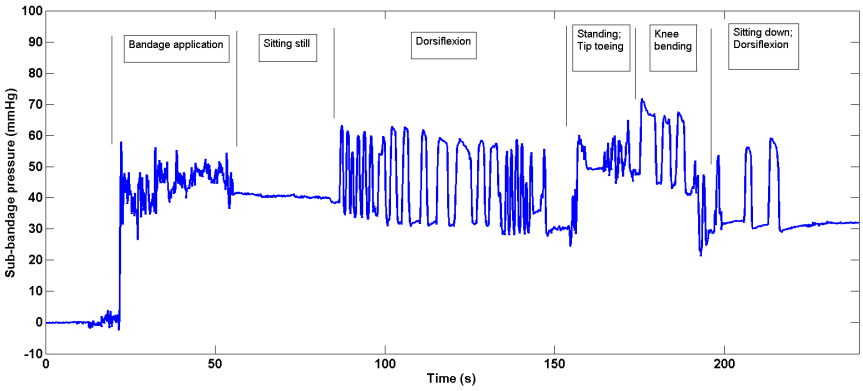

Fig. 8. Working pressure measurement at the medial gaiter during simple movements.

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5. Conclusion

Graduated compression bandaging has long been considered to be the main therapy for venous leg ulcers with the treatment success largely depending on the amount and distribution of the pressure applied. However, the reported treatment outcome varies markedly because the actual sub-bandage pressure profile is difficult to predict by theoretical models and hard to achieve in clinical settings.

In this paper we have reported a new fiber optic force sensor made of two FBG arrays entwined in a double helix form and packaged with contact-force sensitive substrates. This sensor design has many of the idealized functions described by a previous consensus study [3], particularly the thermal insensitivity and high spatial resolution, and as a result enables more quantitative monitoring of the application of compression bandages and garments, and the subsequent analysis of therapeutic outcomes.

The sensor was tested by a senior nurse consultant with over 10 years experience in chronic wound care and has been shown to be capable of measuring the sub-bandage resting pressure profiles, and assisting to achieve the desired gradient, across the full length of a human lower leg that was treated with a commercial compression bandage system. The sensor array was also demonstrated to record the working pressure variation during the patient's movement. The results accurately represent the performed activities and match well with the findings from previous independent studies.

These results demonstrate the potential of this fiber optic sensor for providing real-time feedback during the application of compression bandages and for on-going in-vivo monitoring of the sub-bandage pressure. Further improvement on the sensor design and calibration method will be implemented to accommodate more clinical studies.

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