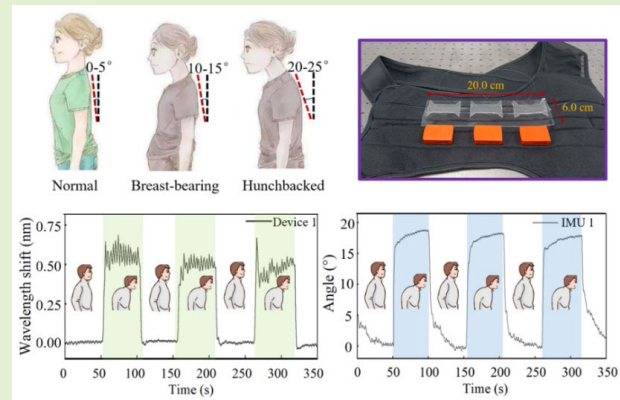


Double-Fishtail-Shaped FBG Wearable Device for Sitting Posture Recognition and Real-Time Respiratory Monitoring

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Abstract—Musculoskeletal issues frequently affect individuals who sit for prolonged periods, particularly in work settings. Addressing and preventing these upper body musculoskeletal disorders (UBMDs) requires close monitoring of seated behavior among employees. This study introduces a solution—an adaptable wearable device incorporating three distinct modular devices using fiber Bragg grating (FBG) technology. The FBG array, integrated into a polydimethylsiloxane (PDMS)-based double fishtail (DF) design, significantly enhances curvature sensitivity and durability. By fine-tuning the length of DF silicone, we further optimized curvature sensitivity. Our wearable device demonstrates prompt correlation with thoracic spine patterns, validated against a motion capture (MoCap) system using ten volunteers as a benchmark. Furthermore, it successfully identifies common sitting postures—normal, breast-bearing, and hunchbacked—with exceptional accuracy exceeding 96.30%, employing the random forest machine learning methodology. Additionally, this device facilitates timely and accurate monitoring of respiratory rate (RR) by time–frequency domain analysis. This innovative optical solution showcases its potential for a variety of wearable device applications.

Index Terms—Double fishtail (DF), fiber Bragg grating (FBG) array, respiratory rate (RR), sitting postures, thoracic spine, wearable device.



I. INTRODUCTION

UPPER body musculoskeletal disorders (UBMDs) are severe inflamed diseases that cause dysfunction in various upper body regions. Several studies have indicated that

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such disease is frequently due to sitting in uncomfortable postures for long periods of time [1]. Prolonged sedentary postures lead to unnatural postures that overload the intervertebral disks [2], causing injurious lower back pain, neck pain, shoulder discomfort, and limb soreness [3]. In addition, poor sitting posture will often lead to changes in breathing patterns [4]. Indeed, due to the active role of the thoracic spine in maintaining trunk stability, inappropriate trunk posture may cause respiratory difficulties [5].

In work settings, smart chairs and cushions [6], [7] seemed to be a solution, but they cannot allow continuous monitoring away from the backrest. Instead, wearable devices [8], [9] are a viable alternative due to their weak effect on the worker's posture and allow for constant monitoring. Traditionally, goniometers [10], X-rays [11], and motion capture (MoCap) systems (motion capture unity) [12], [13] are commonly used devices for detecting UBMDs in healthcare practice. While accurate, these devices have proven to be troublesome or uncomfortable to use throughout the workday. In recent years, optical fiber sensing technology has gained attention for applications in rehabilitation engineering and biomechanics

due to its advantages of small size, fast response, compact structure, strong multiplexing capability, and resistance to electromagnetic interference [14], [15], [16].

Cloud et al. [17] mounted multiple fiber optic sensor arrays on a commercially available device made of spring steel tape. While the device accurately monitors the range of spine motion, it needs to be secured to the worker's skin, making it unsuitable to wear in the workplace. Abro et al. [18] investigated fiber Bragg grating (FBG)-based flexure sensors for monitoring human posture at different joint positions by encapsulating the exposed FBG sensors in silicone gel and covering the top of the FBG with polyvinyl chloride strips for protection. However, the manufacturing process is complex and not feasible for integration into the wearable devices. Guo et al. [19] proposed a wearable optical fiber sensor, where the FBG is coupled to a polydimethylsiloxane (PDMS) substrate that responds to strain deformation by shifting the Bragg wavelength. However, the possibility of monitoring spinal curvature was not mentioned.

In terms of respiratory monitoring, previous works using fiber-based sensors have been demonstrated. For example, Pang et al. [20] utilized to use of a macrobending singlemode-multimode-singlemode (SMS) optical fiber structure for respiration monitoring, while Leal-Junior et al. [21] proposed the development of a polymer optical fiber (POF) sensor for the simultaneous measurement of breath and heart rates. Aitkulov and Tosi [22] developed a sensor based on the integration of a smartphone with a plastic optical fiber for the measurement of respiration. However, the main limitation of these wearable solutions relies on their sensor fusion technology can only record respiration and heart rate signals.

In this work, a double-fishtail (DF)-shaped FBG-embedded wearable device was proposed for recognizing three sitting postures of the user while estimating the respiratory rate (RR). The device consists of three multiplexed FBG encapsulated in a polymer matrix, forming a DF-shaped modular system. Additionally, we evaluated the curvature response of each modular device. By integrating the modular devices into the spine-protective tape, the distribution of the modular devices ensures multipoint measurements for sitting classification in the entire thoracic spine region. In addition, the sensitivity of the device to strain allowed for RR estimation. This approach facilitates RR monitoring across various sitting postures with the wearable device.

II. WORKING PRINCIPLE, FABRICATION, AND METROLOGICAL EVALUATION

This section describes the working principle of the FBG device, the design methodology, and the manufacturing process. The three different lengths of the elements were fabricated for the comparison of their sensing performance.

A. Working Principle

PDMS (Sylgard 184, Dow Corning) was chosen as the FBG embedding material. The wearable device is composed of three PDMS-based DF devices intended to be placed along the thoracic spine. With the input of broadband light, each FBG reflects a narrow peak of the spectrum centered on a

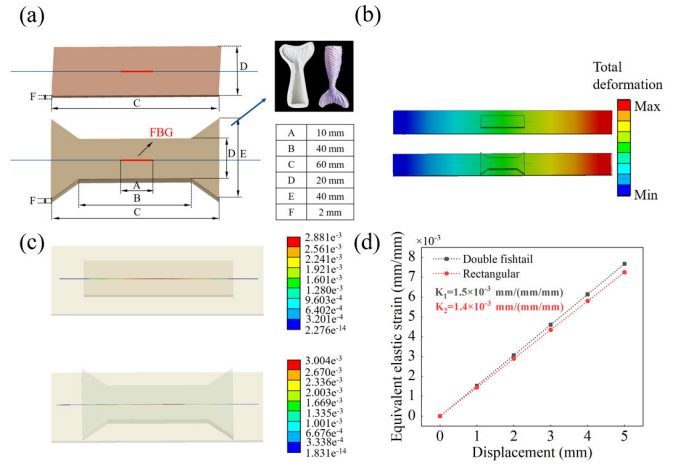


Fig. 1. (a) Schematic of inspiration for the device design and dimensional notation of rectangular model and DF model encapsulation. (b) Total deformation with 1.0-mm displacement applied. (c) Optical fiber strain with 1.0-mm displacement applied. (d) Optical fiber average strain versus applied displacement condition for the structure with rectangular model and DF model.

specific wavelength (i.e., the Bragg wavelength- λ_B). λ_B is a function of the fiber core effective refractive index (η_{eff}) and the grating period (Λ), expressed as follows [23]:

$$\lambda_B = 2\eta_{\text{eff}} \Lambda. \quad (1)$$

Both η_{eff} and Λ depend on temperature change (ΔT) and strain (ε), which leads to a shift in λ_B ($\Delta\lambda_B$). When the temperature is constant, $\Delta\lambda_B$ is described by the equation [24]

$$\Delta\lambda_B = (1 - P_e) \cdot \lambda_B \cdot \varepsilon \quad (2)$$

where P_e represents the effective photoelastic coefficient. In this application, the movement of the spine and displacement of the chest wall during respiration lead to associated deformations of the back skin, where three DF devices are placed. As the hunch angle increases, the devices cause the maximum stretching, leading to the maximum $\Delta\lambda_B$ value; in contrast, as the hunch angle decreases, the devices cause the maximum contraction, leading to the minimum $\Delta\lambda_B$ value. The contribution of ΔT was considered negligible in this study.

B. Device Design and FEA

The DF structure was designed to improve strain measurement performance. The inspiration for the device design, as shown in Fig. 1(a), was encapsulating FBG in a narrow section with a fishtail-shaped PDMS to achieve bending deformation, thereby enhancing the strain in the central region [25]. Based on the above principle, to verify the advantages of this shape, the DF model was compared with a rectangular model. The overall dimensions of the rectangular model were $60.0 \times 20.0 \times 2.0$ mm. In contrast, the overall dimensions of the DF model measured $60.0 \times 40.0 \times 2.0$ mm, and the narrow portion of $40.0 \times 20.0 \times 2.0$ mm, as shown in Fig. 1(a).

The structure of the device was investigated by finite element analysis (FEA) using ANSYS (2021R2), comparing the rectangular and DF models. Each model was imported into the FEA static structural analysis module, setting the Young

modulus of PDMS to 1 MPa and Poisson ratio to 0.43. The Young modulus of the silica optical fiber was set to 7300 MPa and Poisson ratio of 0.17 [23]. The $300.0 \times 40.0 \times 1.0$ mm steel plate was used as a bending setup, with the Young modulus of 2×10^5 MPa and Poisson ratio of 0.3. The grid division's cell size was set to 2.0 mm.

In our FEA simulation, the performance test results show that the DF model has repeatability and good linearity within a displacement range of 0–5.0 mm applied to the steel plate. The model, centrally fixed atop the steel plate, underwent axial stretching due to the plate's bending. Fixed support was applied at one end of the steel plate and displacement conditions of 0–5.0 mm in steps of 1.0 mm were applied inward at the other end, respectively. The model was attached to the steel plate using the nonseparating contact fixation method in FEA (uniform suck along the entire length). The sensing element was encapsulated in a narrow section, the fishtail-shaped ends enhanced the strain in the central region of the matrix during the bending of the steel plate. The fishtail-shaped ends optimize the performance of the system in terms of strain compared to a rectangular model. Measurements included the average strain in the optical fibers and total deformation at a 1.0-mm displacement, as shown in Fig. 1(b), with the maximum and minimum strain of optical fibers shown in Fig. 1(c). The sensitivity of optical fibers under 0–5.0-mm displacement was 1.4×10^{-3} and 1.5×10^{-3} mm/(mm/mm), respectively, as shown in Fig. 1(d). The results indicate that optical fiber encapsulated by the DF model structure has higher sensitivity [25], supporting the above principle.

To evaluate the effect of DF device length on the strain of optical fiber under curvature conditions, three DF devices with different lengths ($L1 = 40.0$ mm, $L2 = 50.0$ mm, and $L3 = 60.0$ mm) were investigated. The average strain of optical fiber was recorded using the same method, fixing the support on one end of the steel plate and applying the displacement condition on the other end. The total deformation under the application of a 1.0-mm displacement condition is shown in Fig. 2(a), and the maximum and minimum strain of optical fiber is shown in Fig. 2(b). The sensitivity under the action of 0–5.0-mm displacement was 0.8×10^{-3} , 1.1×10^{-3} , and 1.5×10^{-3} mm/(mm/mm), respectively, as shown in Fig. 2(c). The results indicate that the 60.0-mm DF device caused the maximum strain of the optical fiber under the same displacement conditions. This increased strain can be attributed to the longer length of the DF device, which produces a concentrated force on the center region, making it sensitive to curvature [24].

C. Device Manufacturing Process

Three molds were made to produce DF devices with lengths of $L1$, $L2$, and $L3$, respectively. The molds were designed with mechanical design software (Pro/E) and the printing material was acrylonitrile-butadiene-styrene (ABS) with 100% infill, printed by the 3-D printer (Crealiti CR-3040 Pro) using the fused deposition method. The dimensions of the three molds are reported in Fig. 3(a)–(c). The mold contains one DF cavity and two rectangular cavities, with the latter designed to facilitate the clamping of optical fiber within the mold.

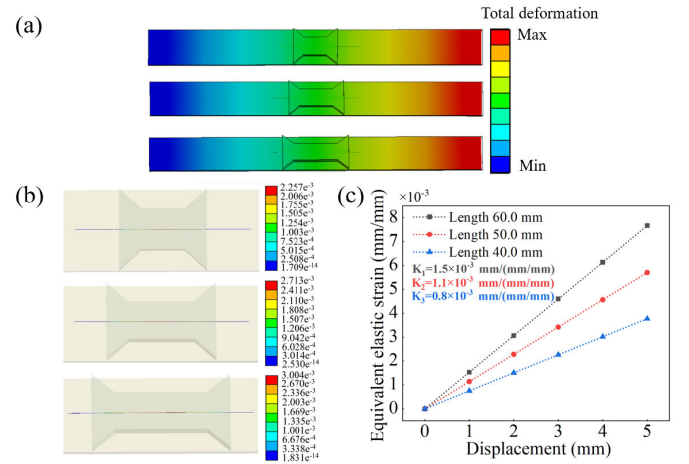


Fig. 2. (a) Total deformation under the applied 1.0-mm displacement condition. (b) Strain of optical fiber under the applied 1.0-mm displacement condition. (c) Average strain of optical fiber DF model encapsulated in three lengths with the applied displacement condition.

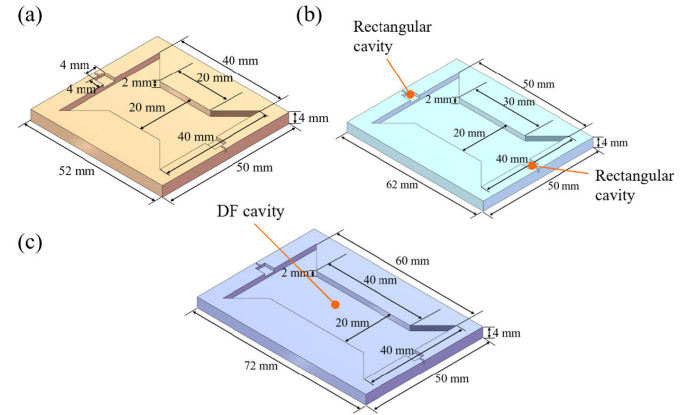


Fig. 3. Schematic of the molds. (a) Length 52.0 mm. (b) Length 62.0 mm. (c) Length 72.0 mm.

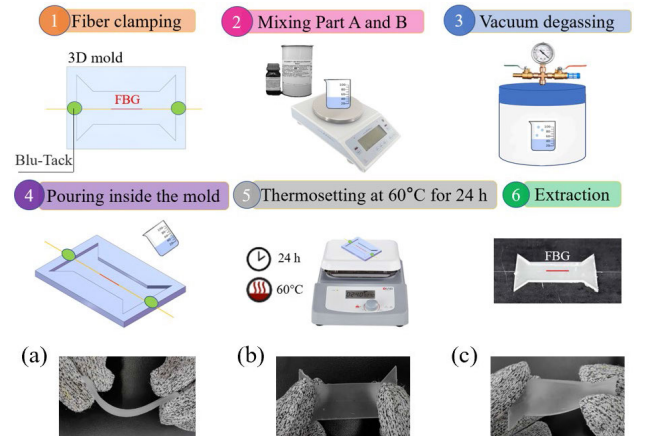


Fig. 4. Fabrication steps of the DF device, including 1—optical fiber clamping and fixation. 2—Mixing Part A and B of the PDMS. 3—Vacuum degassing. 4—Pouring of the solution into the mold. 5—Curing at 60 °C for 24 h. 6—DF device extraction, and in three different configurations. (a) Bending. (b) Stretching. (c) Twisting.

The fabrication steps of DF devices are detailed in Fig. 4. Three FBGs with a center Bragg wavelength of 1550.06, 1550.11, and 1550.20 nm were selected, each 10.0 mm long and with a reflectance over 85% [full-width at half-maximum (FWHM) less than 0.3 nm] to construct the DF devices. Once

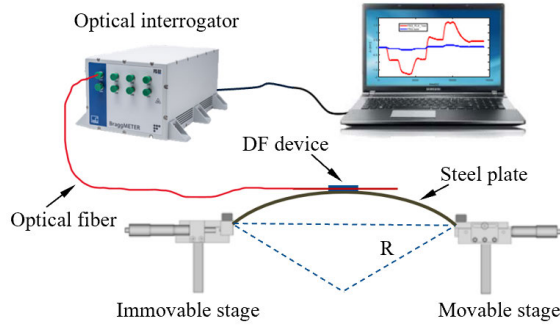


Fig. 5. Schematic of the curvature performance testing system.

embedded in PDMS, the center Bragg wavelengths of the DF devices of lengths L_1 , L_2 , and L_3 were adjusted to 1549.94, 1549.98, and 1550.01 nm, respectively. The center Bragg wavelength of the DF devices with lengths (L_1 , L_2 , and L_3) was set to zero before calibration measurements and then measured the center wavelength shift. Step 1 involved clamping the optical fiber inside the rectangular cavity with Blu-Tack (the main ingredients are natural rubber and additives) to ensure that the FBG was positioned in the middle of the DF cavity. The optical fiber was then pulled to avoid bending when pouring PDMS. Step 2 mixed Part A (main agent) and Part B (curing agent) in a 10:1 ratio based on the manufacturer's instructions. Step 3 involved removing air bubbles by uniform mixing. PDMS solution was then poured into the mold (Step 4) and cured on a heated platform at 60 °C for 24 h (Step 5). Finally, the DF device was released from the mold (Step 6), completing the fabrication of three DF devices of varying lengths (L_1 – L_3). The flexibility of the DF devices, demonstrated through bending, stretching, and twisting on one DF device with a length of 60.0 mm, is shown in Fig. 4(a)–(c), indicating the soft performance and wearable potential of the device.

D. Curvature Characterization

The setup used to characterize the DF devices under bending is shown in Fig. 5. A thin steel plate, measuring 300.0 mm in length and 40.0 mm in width with a thickness of approximately 1.0 mm, was clamped onto two translation stages. DF device was fixed at the center top position of the steel plate, which was clamped to two hand-adjustable displacement stages with a 25.0 mm range (BOCIC, PTS306M). The bottom of the DF device was completely fixed to the steel plate with double-sided adhesive to ensure that the desired displacement was applied. The application of displacement conditions was chosen based on the stiffness and curvature characteristics of the thin steel plate. The strength of the steel plate was selected to follow the force applied by the platform is sufficient to bend the plate and the bending range is within the set range. The Bragg wavelength shift was recorded using Bragg METER FS22SI connected to a computer, with data recorded at 1-Hz sampling frequency. The resulting bending of the optical fiber and the induced curvature can be approximated as [28]

$$C = \frac{1}{R} = \sqrt[2]{\frac{24X}{L_0^3}} \quad (3)$$

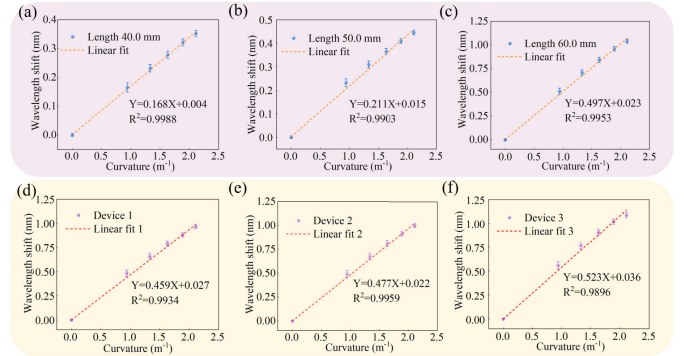


Fig. 6. (a)–(c) Curvature characterization results for different DF length devices. (d)–(f) Curvature characterization results for multipoint DF devices.

where R is the bending radius, and X is the distance moved by the movable stage. In the curvature experiments, the initial distance between the two stages is $L_0 = 300.0$ mm, and the movable stage was operated with a step of 1.0 mm.

Fig. 6(a)–(c) shows the calibration process for DF devices with lengths (L_1 – L_3). The test involved applying a maximum displacement of 5.0 mm to the steel plate, with an increment of 1.0 mm. Each test involves bending a steel plate into a convex shape. Calculated from (3), the curvature increases from 0 to 2.11 m^{-1} ; after linear fitting, wavelength shift has a good linear relationship with curvature in the corresponding curvature range. The curvature sensitivities for DF devices with lengths (L_1 – L_3) were 0.168 ± 0.011 , 0.211 ± 0.012 , and 0.497 ± 0.023 nm/m^{-1} , respectively.

The increase in curvature sensitivity with length verifies the simulation results. The DF device with a length of 60.0 mm was chosen for the next experiment and three DF devices were developed on one optical fiber. The DF devices array consists of three 10.0-mm-long FBGs with different Bragg wavelengths (Device 1—1545.0 nm, Device 2—1550.0 nm, and Device 3—1555.0 nm) each exhibiting a grating reflectance greater than 85% and a 3-dB bandwidth less than 0.3 nm. The calibration of the developed multipoint DF devices revealed a curvature sensitivity of 0.489 ± 0.034 nm/m^{-1} , as shown in Fig. 6(d)–(f). The differences in sensitivities found across multipoint DF devices can be attributed to the differing transmission characteristics of optical signals at various wavelengths in optical fibers, affecting their response to parameter changes [29] and inconsistencies in the manufacturing process, such as insufficient fiber pull or small deviations in the positioning of FBG in the PDMS.

III. WEARABLE DEVICE: DEVELOPMENT AND VALIDATION

An experimental trial was conducted on volunteers to investigate the performance of the proposed wearable device for monitoring spinal motion. The volunteers were instructed to perform a series of hunching maneuvers in front of the MoCap while wearing the device, and the hunching angle of thoracic vertebrae was recorded by MoCap. Finally, RR monitoring was implemented with the device in different sitting positions.

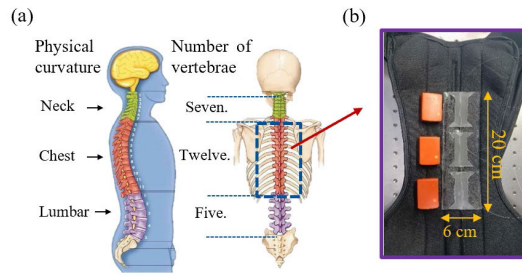


Fig. 7. (a) Structure of the human spine. (b) Schematic of the wearable device.

TABLE I
KEY PARAMETERS OF THE WEARABLE DEVICE

Component	Detail	Supplier
Spine protector tape	Highly flexible pile fabrics	Flagship Store
Fix tape	Acrylic acid	Deli Flagship Store
Optical fiber	Silica optical fiber (8/125 μm)	Fibercore Company
FBG	Three FBGs, each 10.0 mm long, installed to form an FBG array	FBGS Technologies GmbH

A. Wearable Device Fabrication

The thoracic spine is composed of 12 vertebrae and is the largest part of the spine, as shown in Fig. 7(a). Its retroverted curve allows the spine to resist tension and protect the spinal cord when bending or moving the body. Due to the thoracic spine providing most of the support and stability for the entire trunk, poor posture-induced spinal deformities often occur in this segment [12]. Consequently, the sensor devices were placed in the thoracic spine region.

Due to the characteristics of spinal motion, a spine protector tape was used with the developed multipoint DF devices integrated. With an adult's thoracic spine comprising 12 vertebrae and extending approximately 20.0 cm in length [30] and considering the dimensions of our DF device, three DF devices were chosen to cover the thoracic vertebrae. DF device position was achieved by wearing the protector tape to determine the position of 12 thoracic vertebrae on the protector tape and the three DF devices were placed in the thoracic spine position on the protector tape. The DF devices were secured to the spine protector tape with fixed tape 20.0 cm in length and 6.0 cm in width. By tape over the full length of the DF devices, the fishtail-shaped ends of the DF device enhance strain in the central region of the matrix during sitting hunching, thus increasing the sensitivity of the wearable device. The key parameters of the wearable device are summarized in Table I. MoCap technology, utilizing inertial measurement units (IMUs), enables the capture and analysis of the human body or object motion states by integrating multiple IMU sensors.

B. Volunteer Feasibility Assessment

Volunteers participated in spinal motion tests using MoCap, during which they were equipped with the wearable device. Data processing was conducted in MATLAB R2021a. IMUs were uniformly distributed between the thoracic vertebral

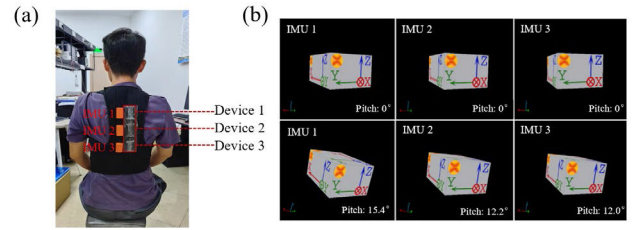


Fig. 8. (a) Position of DF devices (red rectangles) and IMUs integrated into the wearable device. (b) Three-dimensional image of the pitch rotation angle.

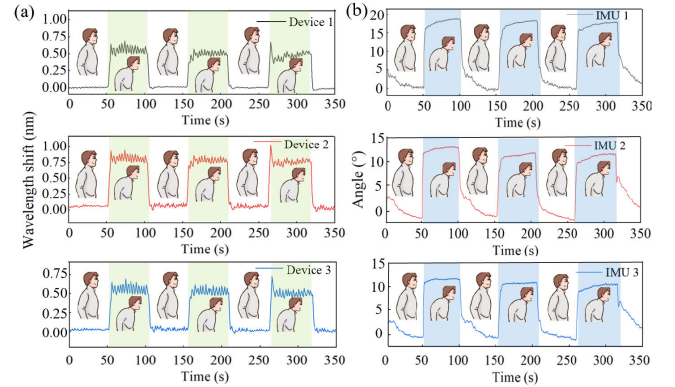


Fig. 9. Volunteer kept in normal and hunchback sitting posture. (a) Response of our device. (b) IMUs pitch angle response.

segments and fixed close to the spine protector tape, as shown in Fig. 8(a).

IMU consists of three main components: a three-axis gyroscope, a three-axis accelerometer, and a three-axis magnetometer, together providing real-time space data output, including Euler angles. The Euler angle is a method used to describe the rotation of an object in 3-D space and consists of roll, pitch, and yaw angles. Specifically, the roll angle refers to the rotation of an object at the X-axis, the pitch angle to the rotation of an object at the Y-axis, and the yaw angle to the rotation of an object at the Z-axis. The IMUs were sampled at a 40-Hz sampling frequency, and the pitch rotation angle was recorded. The initial value of 0° was specified under normal sitting posture, with deviations recorded during hunched sitting postures, illustrated through 3-D images of the pitch rotation angle, as shown in Fig. 8(b).

A volunteer was instructed to remain in a normal sitting position for 50 s, then slowly move to a hunched position, and remain for 50 s again, repeating this process three times. Fig. 9(a) indicates the wavelength shift of three DF devices during the process of volunteering from a normal sitting position to a hunchback and the pitch angle response of three IMUs, as shown in Fig. 9(b). Due to the slow recovery of volunteers from hunching to normal sitting, the response of IMUs shows some movement. The data reveal that, under different sitting states, three DF devices produced different signal responses, aligning with the pitch rotational angle response of IMUs. The results demonstrate the wearable device's capability to monitor hunchbacks while simultaneously capturing subtle spine movements associated with breathing, which could be investigated further for its potential in monitoring RR [16].

TABLE II
COMPARISON OF SENSITIVITY RESULTS FOR TEN VOLUNTEERS

Subj ect	Device 1		Device 2		Device 3	
No.	Sensiti vity(p m/°)	R ²	Sensiti vity(p m/°)	R ²	Sensiti vity(p m/°)	R ²
1	34.58	0.988	30.07	0.997	43.93	0.989
2	43.67	0.998	38.29	0.961	42.66	0.950
3	33.99	0.969	23.89	0.937	36.16	0.997
4	37.11	0.976	32.28	0.973	42.74	0.994
5	32.32	0.966	29.86	0.973	35.13	0.994
6	44.87	0.985	34.46	0.996	50.45	0.999
7	39.02	0.944	29.33	0.957	42.81	0.958
8	41.93	0.988	34.51	0.999	42.65	0.975
9	44.03	0.967	31.15	0.994	45.28	0.990
10	38.81	0.998	32.16	0.997	38.92	0.983

C. Sitting Position Recognition

To evaluate the sensitivity performance of the wearable device in practical applications, ten healthy male volunteers were instructed to move from a normal sitting position to what they perceived as a hunched posture. The wavelength shifts versus rotation angle of volunteers during hunching were recorded with a step of $5^\circ \pm 2.2^\circ$. The results showcased sensitivity ranging from 32.32 to 44.87 pm/° with an R^2 between 0.944 and 0.998 for device 1. For device 2, the sensitivity varied from 23.89 to 38.29 pm/° and R^2 spanned from 0.937 to 0.999. Device 3 exhibited sensitivity ranging from 35.13 to 50.45 pm/° and R^2 spanned from 0.950 to 0.999, as detailed in Table II.

The rotation angle and wavelength shift for the wearable device are shown in Fig. 10(a)–(c), demonstrating that the use of three DF devices yields consistent sensitivity and a good fitting curve response for hunchback posture detection. Linear fitting resulted in average sensitivities for device 1, device 2, and device 3 of 33.03, 31.60, and 41.07 pm/°, respectively, with corresponding R^2 of 0.977, 0.978, and 0.982. The results indicated a slight difference in the sensitivity of the three devices during hunching in ten volunteers, attributable to the variability in the bending of different thoracic vertebrae [31].

To assess the capability of wearable devices in recognizing volunteers' normal, breast-bearing, and hunchbacked postures, three healthy male volunteers were instructed to repeat the process of breathing normally for 100 s in three postures, and the process was repeated 15 times. Data were processed using machine learning by using three DF devices' wavelength shifts, with IMUs serving as a reference for sitting posture

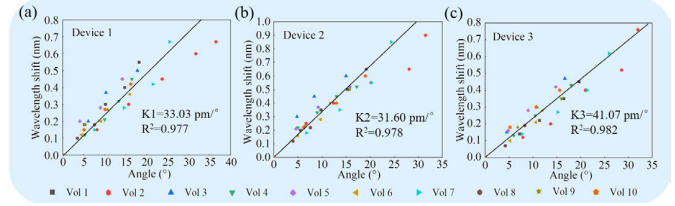


Fig. 10. (a)–(c) Rotation angle and wavelength response of different devices (Device 1, Device 2, Device 3) for ten volunteers.

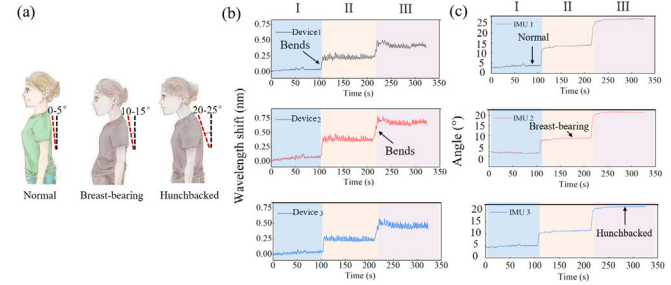


Fig. 11. (a) Classification of three postures (normal, breast-bearing, and hunchbacked). (b) Devices response in three postures. (c) IMUs response in three postures.

recognition, as shown in Fig. 11(a): 1) normal sitting posture, i.e., an angle of about 0° – 5° was formed between the thoracic spine and the vertical line on the ground; 2) breast-bearing sitting posture, i.e., an angle of about 10° – 15° ; and 3) hunchback sitting posture, i.e., an angle of about 20° – 25° . The signal of the devices during one process is shown in Fig. 11(b), and the angular response of IMUs is shown in Fig. 11(c). The three devices show various curvature changes in different sitting postures. Calibration results according to Fig. 6 showed that, for instance, the curvature change in device 1 is 0.54 m^{-1} in the chest-bearing state, and the curvature change in the hunchback state is 1.05 m^{-1} .

A machine learning method was implemented to help identify the different sitting postures. The wavelength response of each device was recorded for three sitting postures. The collected dataset formed a 45×3 matrix containing $\Delta\lambda_B$ belonging to three volunteers, with each column corresponding to a different individual. The wavelength shift features of three devices were extracted, trained, and validated on 80% of the dataset using a random forest machine learning approach and then tested on the remaining 20% of the dataset with the accuracy percentage defined as [16]

$$A\% = \frac{\sum_{i=1}^N \text{TP}_i}{\sum_{i=1}^N C_i} \cdot 100 \quad (4)$$

where TP_i and C_i are the number of predictions and the total number of observations in class i , and N is the number of classes (i.e., 3). From the confusion matrix in Fig. 12, it can be seen that the accuracy of the machine learning-based sitting posture recognition type was 96.30%, which verifies the effectiveness of sitting posture recognition method based on DF devices.

In addition to the posture test, a volunteer was instructed to collect respiratory data for 60 s under quiet breathing (QB) and tachypnea (T) conditions in both normal and hunchbacked sitting postures. For device 2, in the normal sitting posture,

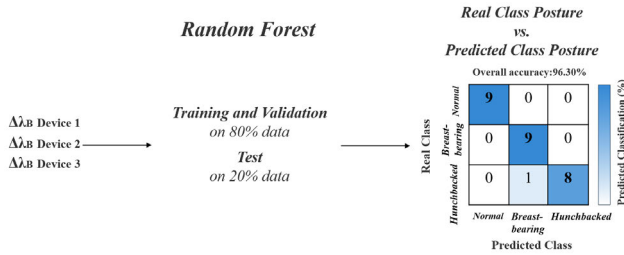


Fig. 12. Classification realization process of sitting posture recognition.

TABLE III

COMPARISON BETWEEN THE PROPOSED DEVICE AND THE OTHER FBG-BASED CURVATURE SENSORS

Sensor structure	No. of sensors	Sensitivity	Application	Ref
FBGs embedded in silicone tubes	1	5.6 ±0.5 pm/°	Monitor the bending of the finger, arm, and knee	[33]
FBGs embedded into a 3D-printed ring	1	5.6 pm/° and 27.6 pm/°	Monitor systematic bend motion of elbow joint and knee joint	[34]
FBGs embedded in a special silica gel	1	18.0 pm/° and 21.0 pm/°	Monitor knee joint movements	[35]
FBGs embedded in rectangular shape silicone	1	-	Monitor the lumbar flexion-extensions	[36]
FBGs embedded in rectangular shape silicone	2	-	Monitor neck movements and breathing activity	[37]
FBGs embedded in a DF shape of silicone	3	31.60 pm/°, 33.03 pm/° and 41.07 pm/°	Monitor the back movements and respiratory rate	This work

experimental results indicated that under QB condition, the wavelength shift was 0.06 nm, as shown in Fig. 13(a), while under T condition, the wavelength shift was 0.02 nm, as shown in Fig. 13(b). RR was indicated as the signal response in the frequency domain for the respiratory volume per minute (QB 9.6 r/min, T 20.4 r/min). As shown in Fig. 13(c) and (d), in the hunchbacked condition, the wavelength shift under the QB condition was 0.07 nm, while the average wavelength shift under the T condition was 0.04 nm. RR indicates the signal response of respiration per minute in the frequency domain (QB 9.6 r/min, T 23.4 r/min). Compared to manual counting, the device had a maximum error of 1 r/min. The wavelength shift and signal contour intensity increase were observed under QB condition, indicating deeper breathing. In contrast, under the T condition, the wavelength shift for breath counting was relatively low [32]. The extracted respiratory waveforms were distinguishable and distinct, indicating that the device can monitor the respiratory activity of volunteers in different sitting postures.

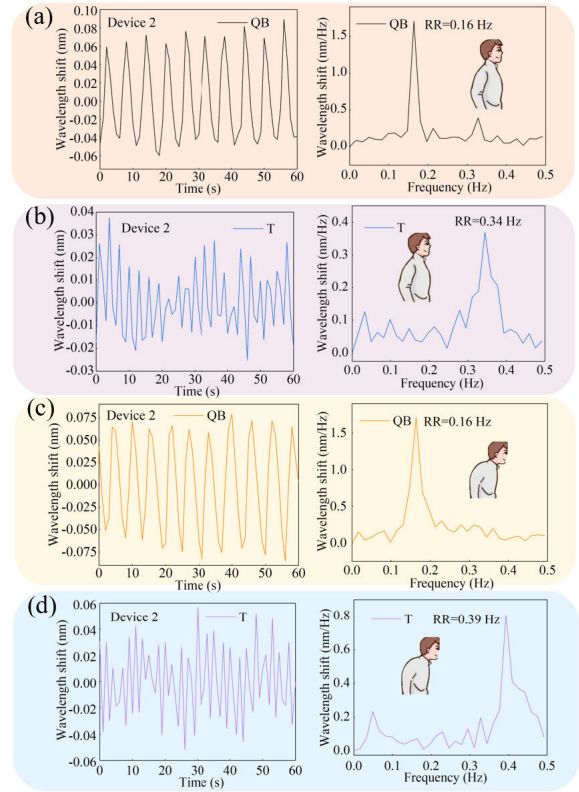


Fig. 13. (a)–(d) Results for 60-s monitoring of RR and the signal responses in the frequency domain under QB and T conditions in normal and hunchback sitting positions.

IV. DISCUSSION

Table III shows the comparison of our device with other FBG-based curvature sensors. Embedding FBGs in support materials (e.g., silicone tubes, 3-D printed rings, and silicone) has been suggested to improve their robustness for monitoring curvature activity. Compared with previous studies, the advantage of this study is the DF-shaped FBG-embedded design, which offers superior the sensitivity over other FBG sensors to monitor joint motion [33], [34], [35]. In addition, our wearable device allows constant monitoring compared to chairs and smart cushions [6], [7], which have performance that can be affected by disconnecting from the user. Furthermore, our device can collect data from the entire thoracic vertebral region, compared to placing a small number of sensors in a limited region, such as the lumbar or neck [36], [37].

V. CONCLUSION

In conclusion, the DF-shaped FBG-embedded wearable device has proven to be highly effective in recognizing three distinct sitting postures and estimating RR. By evaluating the hunchback angle across ten volunteers, the device demonstrated consistent responses, effectively discerning between normal, breast-bearing, and hunchbacked postures. Its capability extends to practical clinical applications, enabling RR monitoring. The lightweight and wearable design makes it conducive for office environments, seamlessly integrating into everyday clothing, and offering significant benefits for employee health management and ergonomic studies.

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