Photonic textiles for pulse oximetry

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Abstract: Biomedical sensors, integrated into textiles would enable monitoring of many vitally important physiological parameters during our daily life. In this paper we demonstrate the design and performance of a textile based pulse oximeter, operating on the forefinger tip in transmission mode. The sensors consisted of plastic optical fibers integrated into common fabrics. To emit light to the human tissue and to collect transmitted light the fibers were either integrated into a textile substrate by embroidery (producing microbends with a nominal diameter of 0.5 to 2 mm) or the fibers inside woven patterns have been altered mechanically after fabric production. In our experiments we used a two-wavelength approach (690 and 830 nm) for pulse wave acquisition and arterial oxygen saturation calculation. We have fabricated different specimens to study signal yield and quality, and a cotton glove, equipped with textile based light emitter and detector, has been used to examine movement artifacts. Our results show that textile-based oximetry is feasible with sufficient data quality and its potential as a wearable health monitoring device is promising.

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1. Introduction

Customized and wearable health monitoring devices, incorporated into textiles and garments, are expected to become medical attendants in the future for continuous and autonomous monitoring of vital physiological indicators of professionals, patients and elderly persons [1-3]. Textile integrated devices have numerous advantages over portable instruments attached to the skin or carried as discrete appliances close to the body, e.g. reduction of loose connecting wires between sensors, electronics and power supplies would lead to increased reliability, data quality and security, as well as enhanced comfort and mobility would be beneficial for the wearer. Most known textile implementations of wearable devices so far utilize electrically conductive metal threads and coatings or electroactive polymer fibers combined with integrated circuits, sensors and modules to interface with the human body and to acquire and transport biosignals like ECG or respiration rate [4, 5].

On the other hand, the use of textile integrated optical fibers in biomedical applications is a relatively new subject. In the majority of cases, plastic optical fibers (POF) are used [6-8], since they are more resistant to textile manufacturing processes and have a higher safety potential, compared to glass fibers. Optical fibers made of polymeric materials have also the advantage of high flexibility and low stiffness compared to glass fibers; therefore they are receiving more and more attention in the field of smart textiles and will complement electrical wires and sensors in the near future. Furthermore, optical fibers produce no heat, they are insensitive to electromagnetic radiation and they are not susceptible to electrical discharges – significant advantages when used close to biomedical devices (e.g. pace makers, insulin pumps) or in diagnostic environments (MRI).

The history of photonic (= light emitting) textiles based on POF started in the late 1960s, when DuPont was investigating the field of optical polymers and fiber production. One of their first patents already showed the possibility of weaving POF and altering their cladding in a subsequent step to receive luminous effects [9]. Since then, numerous patents have been filed in and were granted to third parties; surprisingly not many applications were entering the...
market. A few exceptions are luminous fabrics employed in fashion, design and architecture produced e.g. from Luminex [10] and Brochier [11], backlighting devices from Lumitex [12] for switches, keypads, and LCD’s, and a surgical illuminator, also from Lumitex.

Textiles with optical fibers integrated also have found several applications as sensors. A general overview of sensor principles, fiber design and integration into fabrics is given in references [13-19]. The most often used sensing principles are Bragg or long period gratings [20]. The use of such fibers as stress sensors is reported in [21]. The authors introduced a novel technique to measure static and dynamic stress in parachute canopy weaves and suspension lines. Bragg grating fibers used to measure thoracic and abdominal respiration in humans are covered in [22, 23], where medical textiles have been developed and tested in respiratory plethysmography applications. To monitor structural textile composites a variety of different sensors setups has been assembled and investigated [24-28]. Other sensing principles, mostly employed for multimodal POF, are based on the formation of microbends [29-31]. This principle has been applied so far only in few textile applications. Sensor arrays, where optical fibers are placed in warp and weft direction of weaves, allow the determination of touch, pressure and location [32-34]. A different approach for a pressure sensor array is presented in [35], in which elastomeric POF were squeezed and deformed rather than bent.

Photonic textiles outside sensing applications are also covered in the scientific literature. A novel biomedical application is reported in the field of cancer treatment. Luminous embroideries are used to apply light energy to cancerous tissue during photodynamic therapy [36, 37]. The embroideries are thin and flexible and therefore adapt well onto inhomogeneous tissue topologies and body cavities (e.g. intraoral cavity). By attaching the textiles to the patient a more precise and safer treatment could be conducted due to easier dose calculation and reduction of disturbing patient movements. Optical fiber displays and textile illumination devices have also been reported [38, 39]. Their technology mostly is based on woven POF having cladding imperfections (mechanical, thermal, chemical damage) and therefore light emitting sites. Other examples of photonic textiles are using light sources attached inside the fabrics. The most popular approach is driven by Philips’ Lumalive technology [40] for fashion and design applications. Light emitting diodes (LED) are distributed inside fabrics (e.g. shirts or furniture), which have electrical threads incorporated. Furthermore, organic light emitting devices (OLED) on fiber [41], electrochromic polymers [42], surface-emitting fiber lasers [43], and microstructured fibers [44] have already shown their potential to deliver novel multifunctional photonic textiles [45].

In principle photonic textiles with a sufficient sensitivity are also of high interest for the near-infrared spectrophotometry and imaging (NIRS and NIRI) community. There are many types of instruments and clinical applications of NIRS and NIRI to assess oxygenation and blood circulation in tissue [reviews: 46-58]. All of these could profit from photonic textiles. To give an example of an application, we focused on pulse oximetry, because it is a well established technique.

In this report, we present our approach of using photonic textiles as wearable pulse oximeters for continuous physiological monitoring of human subjects. Pulse oximetry has been introduced already 20 years ago [59] as a versatile non-invasive tool for monitoring the arterial oxygen saturation of biological tissue. It is nowadays widely used in clinical applications, e.g. in intensive care, emergency and recovery rooms, during anesthesiology or in perinatology [60]. Pulse oximeters usually measure the absorption of light of oxy- and deoxy-hemoglobin (O2Hb and HHb) at two different wavelengths (either in transmission or reflection mode) while attached to the subject’s ear lobe, finger or toe. The two resulting signals have a pulsatile factor (each heart beat changes blood volume in the monitored vessels, which can be solely attributed to arterial blood) and allow consecutive calculation of arterial oxygen saturation (SpO2) [60].

One crucial factor, which determines the signal to noise ratio of the SpO2 measurement, is the loss of light from the finger to the detector, the so-called in-coupling. Light emission by photonic textiles is as stated above, relatively well studied. Therefore, the aim of this study was to determine and optimize the efficiency of the light detection of photonic textiles and
whether it is feasible to use such textiles in pulse oximetry. The focus of our work has been on
the comparison of woven versus embroidered textiles, incorporating merchantable quality
PMMA POF, the measurement of their light emitting and receiving capabilities and the
aptitude of the different textile patterns for pulse oximeter design on the fingertip.

Fig. 1. Textile techniques used, POF (red), PET fibers (light blue). Left scheme: canvas pattern
weaving of POF and PET fibers; middle scheme: Soutage embroidery (PET substrate in light
blue, retention stitches in dark blue); right scheme: Schiffli embroidery. Both embroideries are
using a canvas type PET weave as substrate.

2. Materials and methods

2.1 Textile techniques

In our work we have been using three different textile manufacturing techniques to
incorporate POF into a two-dimensional fabric. Woven patterns consisted of POF (weft
direction) and polyester fibers (PET; warp direction), interconnected in a so-called canvas
pattern (Fig. 1, left scheme; POF in red, PET in light blue). Embroideries were either
“Soutage” or “Schiffli” pattern. Soutage technique places the POF on one side of a woven
PET substrate (warp and weft; canvas pattern) and holds them with local retention stitches in
place (Fig. 1, middle scheme; retention stitches in dark blue). Schiffli technique (which allows
retention stitches on both sides of a substrate) guides the POF on one side of a PET weave to
the place of interest, crosses to the opposite side and back, which will lead to random loops
due to POF twisting (Fig. 1, right scheme).

2.2 Woven specimens

The herein used weave was manufactured by Stabio Textil AG (Switzerland) from 250 µm
PMMA POF (Mitsubishi Rayon, Japan) and white polyester (PET) fibers (Fig. 2). Six
specimens were assembled for pulse and SpO₂ measurements labeled from “A” to “F” in the
following text (overview in Tab. 1).
Fig. 2. Woven specimens A to F made of 250 µm PMMA fibers (POF in weft direction, white textile PET fibers in warp direction). Two pictures are shown for every specimen (connected halogen lamp OFF and ON); specimen A and B are untreated and emit no light, therefore only one picture is displayed.

Each contained 16 POF in weft direction, whereas warp threads consisted of PET fibers only. A was used as supplied (Fig. 2); B was identical to A, however two layers of woven fabric were superimposed (POF directions orthogonal to each other); C was cut with an optical fiber cutter and the fiber ends were left without further treatment; fiber surface of D was roughened over the length of 5 mm with POF polish paper (Harting, Germany; grain size 1000); E was identical to D, however a pressure-sensitive alumina foil reflector (3M, USA) was attached on the fiber back side; F was identical to C, however fiber ends were bent by 90 degree (radius 0.5–1.0 mm) towards the light source under local heating (125 °C, hot air gun). POF length, from connector to the point of irradiation or light emission, was 20 cm each. The fibers were glued with standard epoxy glue into F-SMA connectors (Precimation, Switzerland), and the fiber ends were polished later with fiber polishing/lapping film (aluminum oxide, grades 5, 3 and 1 µm; Thorlabs, USA).
2.3 Embroidered specimens

Due to the brittleness of PMMA all embroidered patterns were made from 175 µm PMMA POF (Poly-Optical, USA) as 250 µm fibers were very often breaking while being processed inside the textile machine due to the high acceleration forces. All embroidered patterns were manufactured by Bischoff Textil AG (Switzerland), for further technical information see [37]. Specimens assembled for pulse and SpO₂ measurements are labeled from “G” to “I” in the following text (overview in Tab. 1): G was made using so called Soutage technique where one POF was bent in a radius of approximately 0.5 mm (Fig. 3, upper left and upper right side); H was manufactured using the Schiffli technique (Fig. 3, lower left and lower right side; the resulting loops had randomly distributed radii between 0.25 and 1 mm; approximately 3 fibers with the corresponding loops were irradiated at the same time in the following experiments); I was identical with H, however a pressure-sensitive back reflector of aluminum foil (3M, USA) was attached. POF length, from connector to the point of irradiation or light emission, was 20 cm each. The fibers were glued with standard epoxy glue into F-SMA connectors (Precimation, Switzerland), and the fiber ends were polished later with fiber polishing/lapping film (aluminum oxide, grades 5, 3 and 1 µm; Thorlabs, USA).

2.4 Glove manufacturing

One woven light emitter and one woven light detector were integrated into a standard cotton glove at the forefinger tip position. The light emitter consisted of 4 POF delivering light of...
690 nm wavelength and 4 POF delivering 830 nm. The light detector consisted of 16 POF. Light detector and light emitter were manufactured according to specimen F (90 degree bends towards tissue). POF length, from connector to the point of irradiation or light emission, was 30 cm each. Again the fibers were glued with standard epoxy glue into F-SMA connectors and polished.

2.5 Efficiency of light in-coupling

Signal to noise ratio of the different specimens depends mainly on their light detecting sensitivity. The loss of light from the light source to the detector was expressed as light in-coupling efficiency ($E_\text{f}$) in [%]:

$$E_f = \frac{\text{Energy of detected light from textile}}{\text{Energy of emitted light from source}} \times 100$$

Fig. 4. Measurement setup to determine $E_f$. Light is projected onto the tissue and the detected light intensity is measured. A similar setup was used when working with the halogen lamp (no lenses and pinhole were used in doing so, and the attached light guide was in direct contact with the textile).

To measure $E_f$ of woven and embroidered specimens two types of laboratory setups have been used (every specimen was measured five times in both setups). First setup consisted of a HeNe laser (1135P, 633 nm; JDS Uniphase, USA) with the textile specimen in a distance of 20 cm (Fig. 4). The laser beam diameter was widened by means of two lenses to 5.1 mm. The surface of the specimen was irradiated through a black silicone sheet placed on top of the fabric, having a 5 mm diameter hole. The light energy was measured with an Ulbricht integrating sphere (RW-3703-2; Gigahertz Optik, Germany). The second setup was using unfiltered light of a xenon halogen lamp equipped with a 5 mm diameter light guide (KL 2500 LCD; Schott, Germany). The light guide was placed directly on top of the specimen. The absolute energy and time dependent energy drift of either light source was determined with the formerly mentioned sphere (HeNe laser: $13.1 \pm 0.07$ mW, drift < 1% over 60 minutes; halogen lamp: $105.1 \pm 0.8$ mW, drift 4.6% over 60 minutes; n = 3).
2.6 Pulse waves

Pulse waves have been detected for all specimens using a near-infrared, non-invasive tissue oximeter (OxiplexTS; ISS, USA) on the tip of a subject’s forefinger (Fig. 5). Data was recorded for 20 seconds. Noise was calculated as standard deviation of residues resulting from a polynomial fit of 5 normalized consecutive heart beats. To irradiate the finger tips two silica optical fibers (400 µm silica fibers 3M; FT-400-EMT, from Thorlabs, USA) embedded in a silicone light seal/disc were used delivering light of 690 and 830 nm. Woven and embroidered specimens D, F, G and I were placed on the other side of the forefinger collecting the transmitted light. A black silicone sheet with a 5 mm diameter hole was placed between finger and textile to define the light receiving area. The fibers were glued with standard epoxy glue into F-SMA connectors (Precimation, Switzerland), and the fiber ends were polished later with fiber polishing/lapping film (aluminum oxide, grades 5, 3 and 1 µm; Thorlabs, USA; total POF length: 20 cm). F-SMA plugs were connected to the OxiplexTS detectors (sampling rate 50 Hz). Specimens G and H, respectively, were connected straight to the OxiplexTS (POF length for sender and detector 20 cm). Data has been exported as ASCII to Excel 2003 (Microsoft, USA) and Origin 8.0 (OriginLab, USA) for further processing.

![Fibers for illumination](image)

Fig. 5. Setup to record pulse waves at forefinger tip. The OxiplexTS near infrared spectrophotometer includes laser diodes of 690 and 830 nm wavelength and a photomultiplier tube detector. Two fibers transport the light to illuminate the finger tissue. On the opposite side of the finger, the textile specimens A to I receive the transmitted light, which is detected by the photomultiplier tube. In the glove experiment, all fibers were incorporated inside a single textile such that the forefinger tip was positioned between illumination and detection.

2.7 Arterial oxygen saturation SpO₂

The natural logarithm of the light intensities at the detector was taken, which lead to the attenuations $A_{690nm}$ and $A_{830nm}$. According to the modified Lambert-Beer law [61], the following system of equation was solved to obtain changes in $O_2$Hb and HHb concentration:

$$
\Delta HHb \times d \times DPF = (-0.239\Delta A_{690nm} + 0.099\Delta A_{830nm})
$$

$$
\Delta O_2Hb \times d \times DPF = (+0.185\Delta A_{690nm} - 0.508\Delta A_{830nm})
$$

Where $d$ is the distance between the illumination and detection and DPF the differential pathlength factor, which accounts for prolonged path of light in tissue due to multiple scattering. To obtain values of arterial blood [59] we take advantage of the pulsation of the
blood, which is visible in Fig. 6. This pulsation is due to the change in blood concentration in tissue during heart beat with systole and diastole. This change is only present in arterial blood. Therefore, by calculating the amplitude of the pulsation, we obtain the \( \Delta O_2Hb \times d \times DPF \) and \( \Delta HHb \times d \times DPF \) of the arterial blood. The amplitude was calculated by detrending (high pass filtering) the data and calculating the standard deviation, which is proportional to the amplitude. It is sufficient to have a value proportional to the amplitude since the \( \text{SpO}_2 \) is calculated as a ratio and the proportionality factor as well as \( d \) and \( DPF \) cancel out:

\[
\text{SpO}_2[\%] = \frac{\Delta O_2Hb \times d \times DPF}{\Delta HHb \times d \times DPF + \Delta HHb \times d \times DPF}
\]

Data was recorded for 20 seconds; only periods without movement artifacts were taken into account. Noise level was calculated over 5 seconds (50 data points/heart beat) after 5, 10 and 15 seconds.

3. Results and discussion

3.1 Light in-coupling efficiency

Most of the fiber optic sensors are oriented with their fiber ends perpendicularly to the object to be observed by reason of maximum signal sensitivity. Textile integrated POF however are in parallel to the examined object due to the manufacturing process, which is in general a two dimensional assignment (Fig. 1). Our first interest was, whether woven or embroidered specimens can be produced or altered after production in a way that light can couple into the fibers, and how \( Eff \) could be increased, respectively. For these measurements we directed a HeNe laser beam or a flexible light guide from a halogen lamp onto the textile specimen.

| Specimen       | HeNe laser | Halogen lamp |
|----------------|------------|--------------|
|                | \( E_{lf} \times 10^5 \) | \( SD \times 10^5 \) | \( E_{lf} \times 10^5 \) | \( SD \times 10^5 \) |
| Woven          |            |              |                 |                   |
| A (POF only)   | 0.70       | 0.06         | 0.39            | 0.07              |
| B (Two POF only layers) | 1.89       | 0.16         | 1.29            | 0.16              |
| C (Cut fiber)  | 7.12       | 1.74         | 8.89            | 2.02              |
| D (Surface roughening) | 13.11      | 1.11         | 15.20           | 1.64              |
| E (Surface roughening + reflector) | 30.00     | 4.10         | 37.06           | 1.13              |
| F (Fiber bends of 90 degrees) | 76.64     | 24.14        | 44.62           | 7.95              |
| Embroidered    |            |              |                 |                   |
| G (Soutage)    | 1.69       | 0.28         | 0.80            | 0.08              |
| H (Schiffli)   | 2.80       | 0.68         | 3.32            | 0.61              |
| I (Schiffli + reflector) | 4.67       | 0.70         | 7.99            | 0.59              |

The results determined differed between laser and halogen lamp in some cases vastly which could be explained by the rather high heterogeneity of the textile specimens [37], which minimizes the potential accuracy of observation. The measured \( E_{lf} \) are reported in Table 1.

\( Eff \) of a woven specimen without further altering of the fiber surface was very low (A, B). The weaving process is rather gentle to the optical fibers, so no sharp bends or cladding damaging is happening in general, which would lead to an improved geometry for light to enter and propagate inside the fibers. A much better approach seemed to be when the POF
were cut at right angle at the location of interest (C); an increase in efficiency over A of a factor of 10-20 has been measured. Roughening of the fiber surface (D), which results mainly in removing the cladding material and generating an inhomogeneous surface with multiple in- and out-coupling locations, almost doubled the efficiency over just cutting the fiber end. Adding a rear light reflector to the same specimen facilitated increasing the amount of detected light by another factor of 2 (E). F, whose fiber ends were bent by 90 degree towards the irradiation source, showed the largest measurable efficiency (although due to the small bending radius a significant amount of light will be leaving the fiber directly again).

Embroidered specimens had a general lower Eff than the preferred woven specimen. G consisted of a single fiber with one bend (Soutage pattern), where light can be received. Compared to A (which consisted of 16 POF in parallel), the single bend did clearly increase Eff by a factor of approximately 2, even though the POF diameter is smaller (175 versus 250 µm). Changing the embroidery pattern from Soutage to Schiffli (H) and with the help of a rear alumina reflector (I), an additional increase of a factor of 2 could be measured.

3.2 Pulse wave measurements

To compare the sensitivity of woven versus embroidered specimens four fabrics have been chosen for pulse wave detection on a human forefinger tip: D, F (woven) and G, I (embroidery). Specimen F delivered the highest amount of analog/digital counts (raw data displayed in Fig. 6); whereas specimen D had less than 35% thereof (valid for both wavelengths); this finding corresponds well to the previously measured Eff. It is also visible from the data that the used PMMA fibers have a rather high attenuation in the range of 830 nm, whereas 690 nm corresponds well to a local attenuation window of this material [6]. However, both fabrics allowed a precise determination of pulse waves.

For the embroidered specimen G and I more noise has been measured than for D and F, respectively (G: SD of residues after polynomial fit = ± 0.041 (690 nm), ± 0.053 (830 nm); I: ± 0.017, ± 0.018; D: ± 0.036, ± 0.045; F: ± 0.009, ± 0.006), underlining the former finding in the light in-coupling efficiency section. Astonishingly, specimen G, having only one fiber bend, could be used to detect the changing absorbance of pulsating blood in the finger tip. However, noise level is high, and a precise detection of pulse waves will need further data analysis methods. The main reason for noise is the very low photon yield in the 830 nm window. A much better photon yield could be achieved with specimen I, about 4-5 times as much as with G, and about 50% of woven specimen F.

3.3 SpO2 measurements

Simultaneous pulse wave measurements at 690 and 830 nm have been used to calculate arterial oxygen saturation (SpO2; Fig. 6, green line) according to section 2.7. The noise level has been calculated at 94.14 ± 0.84 % SpO2 (specimen D), 95.78 ± 0.07 (F), 95.33 ± 1.10 (G), and 94.86 ± 0.57 (I) for a period of 5 seconds. These values are in agreement with standard pulse oximetry data [59]. However, raw data quality of specimen D and G is poor (changes in SpO2 up to 3% within 1 second) and a data smoothing algorithm (e.g. median filtering) for practical applications would be mandatory.

3.4 Artifacts of movements

To study movement artifacts we incorporated two POF emitters and one POF detector textile into a cotton glove (at the forefinger tip position). To simulate maximum deformation of the glove all five fingers were closed and opened at the same time and pulse waves have been recorded. The movement of the textile sensor against the underlying skin can be seen clearly in the data progression (Fig. 7). During the finger bending (opening and closing within 1 second), one to two pulses were not detectable, and positive or negative peaks of up to 8% SpO2 resulted therefore. Artifacts are mainly caused by the shift of blood within the tissue and the relocation of the sensor and detector fibers before and after the movement, i.e. different blood capillaries, bone structures, tissue scattering, and distance and finger thickness, are interrogated.
Fig. 6. Pulse waves and their calculated SpO\textsubscript{2} of four specimens: weaves (D: upper left; F: upper right), embroideries (G: lower left; I: lower right). Time is displayed on the x-axis and analog/digital counts (ADC) at the detector on the y-axis. Blue lines correspond to 690 nm wavelength, red lines to 830 nm; calculated SpO\textsubscript{2} is plotted in green. The amplifier setting at the detector was the same for each measurement. It is visible that a lower intensity (i.e. low ADC counts) leads to more noisy signals.

4. Conclusions

In this report, we described an approach to design textile based pulse oximeters made of POF fabrics. They were used as light emitters and light detectors to human tissue. Three major textile manufacturing processes were compared in terms of signal yield and signal quality.
Woven specimens did not allow emitting or receiving light without further conditioning. $\text{Eff}$ could be increased drastically when POF were altered (imperfections added to cladding, sharp bends, e.g.) and reliable pulse waves and arterial oxygen saturation level could be monitored. In practice, cutting, roughening or heating POF, which are incorporated inside a weave, is not a preferable process for industrial use, due to several reasons. Firstly, depending on the weaving process, the optical fibers are not at the surface of the fabric available for mechanical treatment; secondly, mechanical and thermal treatments will always affect the neighbor fibers as well, which will lead to unwanted fiber breaks and fabrics damage. For the same reasons, a chemical treatment of POF (partially dissolution of POF cladding and/or core materials) after weaving seems not to be a practical approach.

On the other hand, all specimens derived from embroidery obtained their light emitting and light receiving functionality during the textile manufacturing process; no after-treatment of the POF has been applied to the specimen reported herein. It could be demonstrated, that specimens with only one fiber bend did already allow pulse and SpO2 measurements, but the noise level was too high for precise SpO2 calculations. The signal to noise ratio depends on the number of detected photons and could be increased when more than just one POF was used and a reflector foil was attached and/or by a higher illumination intensity.

Thus, in our study we were able to show the feasibility of using photonic textiles for pulse oximetry. Our future work will be directed to further improvement of detection efficiency and the development of whole sensors based on photonic textiles.
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