Development and initial testing of a pulse oximetry prototype for measuring dental pulp vitality

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Abstract. The guiding principle of endodontic treatment is to preserve teeth while maintaining its aesthetic and functional roles. To accomplish this goal the assessment of teeth pulp vitality is very important since it will determine the procedures that should be adopted and define the therapy strategy. Currently, the most commonly tests for determining dental pulp state are the thermal and the electrical tests, which are based on nerve response and, because of that, have a relatively high rate of false positives and false negatives cases. In this work we present a simple test to be used in the clinical setting for evaluating noninvasively the existence of blood perfusion in dental pulp. This test is based on pulse oximetry principle that was devised to indirectly measure the amount of oxygen in blood. Although pulse oximetry has already demonstrated its usefulness in clinical environment its usage for the determination of dental pulp vitality has been frustrated by several factors, notably the absence of a suitable sensor to the complex shape of the various coronary teeth. We developed a suitable sensor and present the first trials with promising results, regarding the ability for distinguish teeth with and without blood perfusion.

1. Introduction
According to the World Health Organization, 90% of world population suffers from oral cavity diseases which, most of the times, are painful [1]. To give responses to these problems better dental materials, new clinical procedures and more accurate diagnosis tests are being created in a daily basis. Despite all advances, the success of the therapy is still highly dependent on the diagnosis. For example, whenever the dentist suspects that the dental pulp may be compromise it is crucial to he knows very well its state in order to adopt the most correct procedures that will ensure the preservation of pulp tissue and avoid inflammation (pulpitis) or necrosis. An accurate diagnosis of the pulp state is essential in many situations, such as to anticipate a procedure for tooth restoration, to follow up and monitoring the pulp after a tooth trauma and as supporting knowledge to many other pathologies [2].

Many of the current methods for diagnosing the pulp state stimulate the A-δ nerve fibers, which gives information about nerve condition but not on vascularization. Nonetheless, there is some correlation between blood perfusion and nervous response because in the absence of perfusion the A-δ fibers enter in anoxia stopping its function. This means that these methods are considered admissible [3, 4] which coupled to the fact of being simple, inexpensive and having a quick answer make them the most frequent tests currently used in the clinical practice.

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The sensorial response of the pulp is achieved using a cold, a hot or an electrical stimulus. Cold agents induce the contraction of the fluid within the dentinal tubules, followed by its rapid efflux which then activates the A-δ nerve fibers and causes pain [5]. As cold agents ice sticks and ethyl chloride gas were initially employed being substituted, in recent years, by more effective gases such as dichlorodifluoromethane and carbon dioxide. Both dichlorodifluoromethane and carbon dioxide permit cooler temperatures producing a more intense response of the A-δ nerve fibers. However, repeated application of cold tend to reduce the rate of fluid relocation within the dentinal tubules resulting in a lower pain for a short period of time, turning the cold test not very reliable. As an alternative there are methods that use heat sensation instead, but as with cold agents the pain results from the fluid dynamics at the dentin-pulp complex [5, 6].

The purpose of the electric pulp test is also to stimulate A-δ fibers that remain intact in the pulp-dentin complex. An electrical stimulus is applied in the tooth surface, after being properly insulated, causing an alteration in neuronal ionic membrane that gives rise to an action potential in the Ranvier nodes of the myelinated fibers. Once the voltage reaches the pain threshold the patient feels discomfort in the tooth under examination revealing the integrity of nerve fibers [5, 7, 8].

The sensitive methods are very simple to use however the rate of false positives and false negatives is 10%-16% making these methods unreliable [9,10]. Other more objective methods, based on the state of the dental pulp, have been proposed to overcome this difficulty. Laser doppler flowmetry, photoplethysmography, spectrophotometry and pulse oximetry are the most important of these techniques that try to assess the state of the pulp by evaluating the blood flow in the dental cavity through optical properties [11]. These methods are supported by the principle that the pulp tissue can have an adequate blood supply despite not being necessarily innervated [10]. And they are also based on the assumption that light interact with blood enabling to distinguish the presence and absence of blood perfusion within a region [12]. Among the methods that rely on light interaction pulse oximetry is the less expensive and is well accepted in the clinical setting. Therefore we decide to exploit the same rationale to develop a device for measuring the vitality of dental pulp [9].

2. Pulse oximetry

Light and matter interaction can be modeled by Beer-Lambert equation that links the intensity attenuation with properties of the medium and light wavelength (equation 1) [13]:

\[ I = I_0 e^{-\mu L} \]

where \( I \) is the light intensity emerging from a medium of length \( L \) and attenuation factor \( \mu \) when the incidence light has intensity \( I_0 \). Notice that attenuation depends on the wavelength of the incidence light. In these conditions the fraction of transmitted light can then be expressed by the exponential term that depends on the wavelength of the incidence light and on the traveling length,

\[ f = \frac{I}{I_0} = e^{-\mu L} \]

Since oxygen concentration in blood has a particular impact on light attenuation it is possible to determine the percentage of oxygen saturation in blood as long as different wavelengths are used to determine the same number of transmitted light fractions [14]. Pulse oximetry commonly use wavelengths of 660 nm (red) and 940 nm (infrared – IR). This choice is driven by four important aspects: (i) absorption is mainly in wavelengths lower than 600 nm (blue, green and yellow) due to skin pigmentation; (ii) absorption of near infrared by water; (iii) absorption of visible light in red region is much higher for deoxygenated hemoglobin (Hb) than for oxygenated hemoglobin (HbO2); and (iv) the IR transparency is greater for Hb than for HbO2 [15]. Pulse oximeters also take advantage of blood pulsation during cardiac cycle: in systole arteries contain more blood than in diastole having, in turn, a larger amount of absorbent substances, such as hemoglobin and also a larger optical distance. This means that light attenuation increases during systole. This dynamic portion of total absorption allows to discriminate absorption due to static components (venous blood, a constant amount of
arterial blood and other invariants such as skin pigmentation) and absorption due to the pulsatile component (blood flow in the arteries). These components are also denominated as DC and AC components of the total absorption, respectively. Arranging measures of DC and AC components from the two wavelengths it is possible to have an estimate of oxygen saturation [16]. This measure is known as Ratio of Ratios (RR) and is defined by

\[
RR = \frac{I_{R} \left( \frac{AC_{R}}{DC_{R}} \right)}{I_{IR} \left( \frac{AC_{IR}}{DC_{IR}} \right)}
\]

where \(AC_{R}\) and \(DC_{R}\) are the AC and DC components, respectively, of the red light and in denominator we have the same ratio for the infrared light.

3. Methods

3.1. Sensor design

Pulse oximeters are normally used to take measurements from finger or earlobe, which are locations anatomically favorable for placing sensors. However, teeth present different shapes and sizes making more difficult to devise a form to place the sensor that should be tightly positioning while ensuring parallelism between emitter and receptor. The approach that we adopt was based on forceps as it can be applied to any tooth and can be easily handled. The sensor holder was designed using CAD software (Inventor®) and a plastic model was then built using rapid prototyping (Fig 1).

![Figure 1. (a) CAD design of the sensor holder; (b) plastic prototype.](image)

At the tip of holder three LEDs were mounted: a red (660 nm), an infrared (950 nm) and also a green LED (540 nm) to account for the enamel and dentin contribution to the light attenuation [17, 18]. At the other tip of the holder, a phototransistor was placed and the signal was acquired using a UD128A8D B&B module connected to a PC. The same electronic module was used to turn on and off the LEDs sequentially. The power needed to maintain the minimum required light intensity was achieved through a power driver (ULN2003A) and signal amplification from the phototransistor was accomplished using a differential amplifier (AD626).

3.2. Sensor testing

Sensor repeatability and linearity was assessed by some simple tests. Repeatability was evaluated by measuring the signal produced for each LED when positioning at two different distances (in contact and 3.0 mm apart) from the phototransistor. Measures were made for each LED individually and mean, standard deviation and coefficient of variation of the acquired signal were calculated as figures of merit to assess response and repeatability. Noise was also addressed by running tests in dark conditions, i.e., acquiring signal without turning on the LEDs.

Linearity was evaluated using a colloidal mixture with four different dilutions fractions. The main idea was to verify Beer-Lambert law by obtaining a linear relationship between the log of the light signal and concentration of the solutions.

3.3. Clinical testing

Preliminary clinical tests were carried out with two reasons: to evaluate the capacity of using the device in the clinical setting and to examine its ability for discriminating vital from non-vital teeth. A dentist conducted 29 tests in four different patients and evaluated pulp vitality of each tooth. Before
measuring the dentist insulated the tooth with a suitable rubber barrier and dried it with an air jet. The measuring was performed for 30s, which has facilitated the accurate positioning of the sensor in the labial and lingual faces of teeth. Results were analysed through logistic regression method using IBM SPSS v20 statistical platform and assuming a significance level of 5% ($\alpha = 0.05$).

4. Results

4.1. Sensor testing

Table 4.1 shows the results (empiric units) obtained for repeatability tests ran at 0.0 and 3mm distance between LEDs and the phototransistor.

|        | R     | IR    | G     |
|--------|-------|-------|-------|
| 0 mm   | $\mu \pm \sigma$ | 413±56 | 3734±13 | 2046±67 |
|        | C.V.  | 13.56% | 0.35% | 3.27% |
| 3 mm   | $\mu \pm \sigma$ | 2588±82 | 3738±10 | 2956±73 |
|        | C.V.  | 3.17% | 0.27% | 2.47% |

In dark conditions the mean signal in the different channels (red, infrared and green) is 47, 47 and 46 respectively and the standard deviation is 10, 11 and 8 respectively.

Sensor linearity testing was performed for different concentrations of a colloidal solution. Results of light intensity in function of concentration were fitted with a linear regression after a logarithm transformation of the phototransistor signal. Response linearity was confirmed visually but also by linear regression. The adjusted linear model for the red LED is statistically significant (F (1,10) = 34.296, p <0.001) indicating a 75% percentage explanation of the variation of the signal (adj. $R^2 = 0.752$) by concentration variation. For the infrared LED the fitted linear model indicates an explanation by the tested concentration of approximately 78% of the variation of the signal (adj. $R^2 = 0.784$) and the model is statistically significant (F (1,10) = 36.283, p <0.001). Finally, for the green LED it can be observed that the linear model shows that concentration explain about 41% of variation of light intensity (adj. $R^2 = 0.412$) and the model is also statistically significant (F (1,10) = 8.720, p = 0.014).

4.2. Clinical testing

The main goal of the work was the classification of a tooth as vital or not vital. Therefore it was very important to extract from the raw signal some numerical indicators that will enable classifying dental vitality. Taking into account the nature of the signal it was intuitive to take advantage of the AC and DC components to compose numerical indicators for the classification. Nonetheless, we decided to use an extra LED that provides further information. We rearranged and simplified all measures as follows:

\[ R = \frac{\ln(AC_R/DC_R)}{\ln(AC_{IR}/DC_{IR})} \]  \hspace{1cm} (4)

\[ R_g = \ln(AC_g/DC_g) \]  \hspace{1cm} (5)

where the subscripts in AC and DC components are referred to the red (R), infrared (IR) and green (G) LEDs, respectively.
Although we have collected further measures the number of non-vital teeth in the sample was a limitation for the analysis. For that reason we decided to analyse similar teeth (molar and pre-molar). Table 4.2 shows results of the ratios according to the equations presented above for the selected cases:

![Table 4.2. Results for vital and non-vital teeth in the sample.]

| Vital | No | No | Yes | Yes | Yes | No |
|-------|----|----|-----|-----|-----|----|
| R     | 0.984 | 0.785 | 1.006 | 0.994 | 1.005 | 0.991 | 0.993 |
| RG    | -2.773 | -2.738 | -3.129 | -2.540 | -2.986 | -2.059 | -2.829 |

A logistic regression model was fitted to the results assuming vitality classification as dependent variable and the ratio scores as predictor variables. The model obtained shows a significant adjustment (Hosmer and Lemshow, \( \chi^2(4) = 0.00; p = 1.000 \)) and the predictor variables explain the dependent variable (Cox and Snell \( R^2 = 0.745 \), Nagelkerke \( R^2 = 1.000 \)). These excellent values are due to the few cases that were analysed which made simple the adjustment to a predictor model. The accuracy obtained was 100% which can be considered significant if compared to the null model (when the classification is random) that presents 57.1% accuracy rate. The model can be described analytically by the equation:

\[
P = \frac{e^{0.784 - 2.773 R - 3.129 RG}}{1 + e^{0.784 - 2.773 R - 3.129 RG}}
\]

where \( P \) is the probability of having a non-vital tooth.

5. Discussion and conclusion

We presented a new device for measuring dental pulp vitality based on pulse oximetry principles. Results indicate a functional device with good reproducibility, low noise and reasonable linearity properties. However, the construction of the instrument revealed some difficulties that are points of concern. Besides the problem of having a general form of placing the sensor on teeth surfaces the utilization of a parallel design between emitter and receiver poses a mechanical problem that has to be carefully tackled. In our case this can be observed by comparing the values obtained when the sensor was tested at two different distances. Small changes in signal (Table 4.1) can be observed between the two distances for each LED which may be explained by the different solid angle at the receiver. This means that different teeth will produce distinct signals in part because of the size and this aspect has to be taken into account in future developments.

Another aspect that should be notice is a dependence on phototransistor sensibility to the emitter wavelength. It seems that this problem do not have great impact in the final results, nonetheless it can be resolved by implementing a software gain to each LED since they are turning on/off sequentially.

Clinical trials exposed other kind of problems: the design of the instrument has to be improved in order to facilitate its usage and teeth have to be perfectly dry otherwise noise is dramatically increased. Results of clinical tests do not permit to generalize and consider them as conclusive. The sample is small and not randomized, which explain the excellent fitting results that may be regarded as a sign of overfitting. Nevertheless, they show a promising trend because it performs much better than a random approach (just guessing) and using only two features.

To address the validation problem, a larger clinical trial has to be design, enabling to draw definitive conclusions.
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