Development of a solid state, non-invasive, human touch based blood alcohol sensor

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ABSTRACT

This paper presents an update on the implementation of a touch-based optical sensor (TTT sensor) for monitoring the alcohol concentration in the driver of a vehicle. This novel sensor is intended to improve driver safety by providing a non-intrusive means of notifying a driver when their blood alcohol concentration may be too high to operate a vehicle safely. Details on implementation of the MARK2 system are presented along with updates on principles of the MARK3 version currently under development. Laboratory validation of the MARK2 system on standard calibration standards are presented along with discussion of next steps in validation of the technology. Updates on the demonstration vehicle implementation are also provided along with lessons learned in the implementation of the human-machine interface aspect of the design.

INTRODUCTION

A touch-based optical sensor was previously described for use in automotive applications [1]. The intended goal of the current sensor development is to mitigate the societal impact of alcohol (ethanol) impaired vehicle driving [2,3,4]. In order to be suitable for use by consumers, the system must be seamlessly integrated into the vehicle's infrastructure, providing consumers with the knowledge of their alcohol concentration without imposing inconvenience to their daily driving experience. Details of the implementation of a touch-based sensor (TTT MARK2) are presented including the deployment of the sensor in a demonstration vehicle. The current human machine interface (HMI) design places the optical touchpad in the vehicle's start button. A demonstration version incorporating capacitive touch sensing along with visual and haptic feedback has been tested. Several improvements in the sensor platform have been identified and are currently being incorporated into the new MARK3 sensor platform. Improvements in system accuracy, size reduction, and manufacturing costs are discussed.

REVIEW OF SENSOR THEORY

As previously disclosed [1], the touch based sensor technology is based on a peer-reviewed and clinically validated method using near-infrared spectroscopy to accurately measure alcohol concentration in humans. A brief review of the key scientific principles is repeated below for reference.

Scientific Basis of the TruTouch Measurement

The TruTouch technology employs near-infrared (NIR) absorption spectroscopy to measure skin tissue. The NIR spectral region typically spans the portion of the electromagnetic spectrum between the visible, which is generally considered to end at 0.7 µm, and the infrared, which begins at 2.5 µm. However, for measuring alcohol *in vivo* (in human), some portions of the NIR are more advantageous than others. The features most commonly observed in the NIR are overtones and combinations of fundamental vibrations of hydrogen bonded to carbon, nitrogen, and oxygen [5,6,7,8].

The absorbance spectrum of alcohol shows features over the NIR region (*Figure 1*). The 1.25-µm 2.5 µm region contains the 1st overtone and combination bands of the carbon-hydrogen and oxygen-hydrogen bonds. The 0.7-1.25 µm region contains higher order overtones of these bonds. Examination of *Figure 1* and its inset shows that the 0.7-1.25 µm region is 400 times weaker than the signal in the longer wavelength, 1.25- 2.5 µm region.

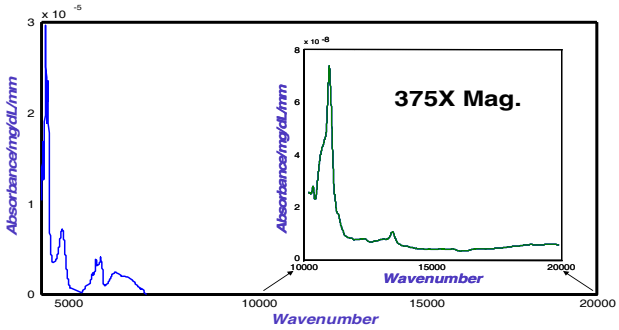

Figure 1. **Absorbance Spectrum of Ethyl Alcohol.**

Furthermore, the utility of the visible region (0.3 to 0.7 µm) and the 0.7-1.25 µm part of the NIR are limited by the presence of skin pigmentation (melanin) that creates large differences between people, particularly of different ethnicities. In contrast, the longer wavelength region is virtually unaffected by pigmentation [9]. As a result of the larger signal and absence of pigmentation, the TruTouch technology is designed to measure the longer wavelength (1.25-2.5 µm) region.

In addition to the aforementioned advantages, the NIR spectral region (4000-8000 cm-1 or 1.25-2.5 µm) is of prime interest for non-invasive alcohol measurements because it offers specificity for a number of analytes, including alcohol and other organic molecules present in tissue, while supporting optical path lengths of several millimeters with acceptable absorbance characteristics [10,11,12,13,14]. Comparing NIR spectra (normalized to unit concentration) of alcohol and water collected using a TruTouch system, demonstrates the effect of molecular structure on NIR absorption spectra and indicates spectral regions of separation (see Figure 2).

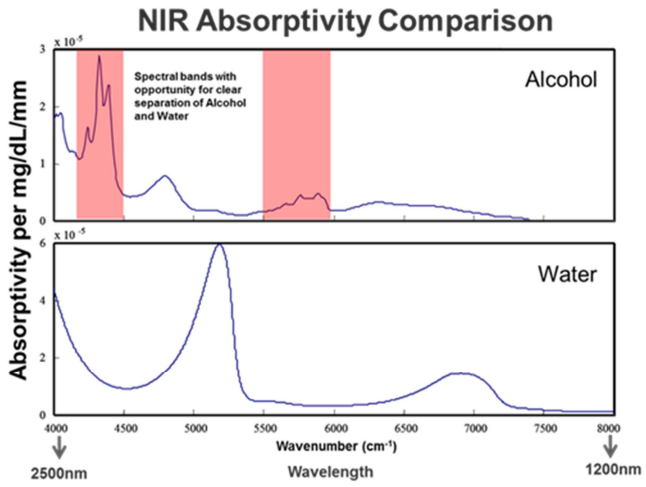

Figure 2. **Comparison of Alcohol, Water Near-infrared Spectra.**

TruTouch systems deliver NIR light to the skin and underlying tissue and collects the diffusely reflected signal using a fiber-based optical probe. The collected light contains spectral information which allows the determination of the subject's alcohol concentration directly from the measurement. Specific details of the industrial version of the optical alcohol detection system can be found in several issued United States Patents and applications [15,16,17,18,19].

UPDATE ON THE MARK2 SENSOR

The MARK2 sensor previously described, has been implemented using discrete semiconductor laser diodes. The MARK2 system was designed to be modular in nature with multiple, replaceable laser butterfly (Telecom) packages configured in a housing suitable for inclusion in a demonstration vehicle. Figure 3 shows the sensor installed in the center console of a vehicle. As can be seen in the picture, the sensor is operable by either the driver or the passenger and has been configured to allow demonstration of the touch-based measurement interface. A lighted bezel provides visual feedback to the user in order to provide an intuitive method of conveying the process and the results of an alcohol test.

Figure 3. **Demonstration Vehicle Installation.**

Figure 4 clearly illustrates the separate subsystems of the MARK2 sensor which are connected via electronic and fiber-optic interfaces to allow for a more flexible installation in vehicle and laboratory test setups. Each of the laser wavelengths shown in Figure 5 are achieved via a discrete laser diode contained in one of four butterfly packages. In contrast, the new MARK3 sensor is currently being fabricated using a multi-wavelength per laser die architecture and a single, integrated opto-mechanical housing.

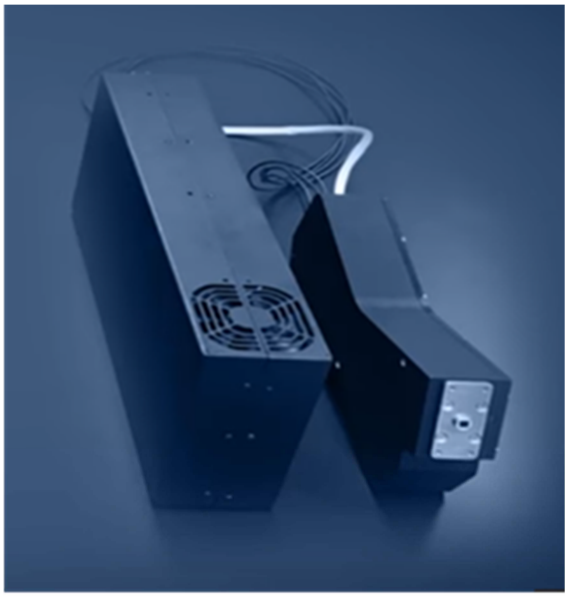

Figure 4. **MARK2 Sub-Systems.**

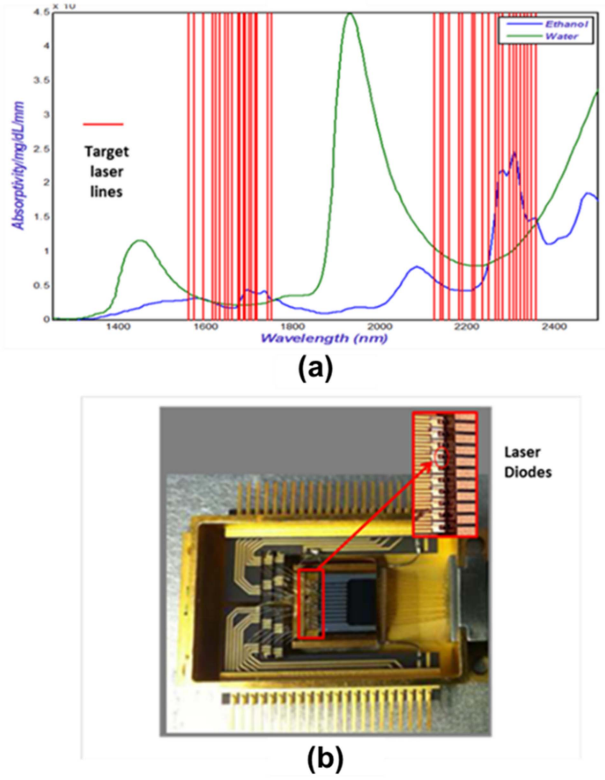

Figure 5. **Laser lines implemented in MARK2 sensor (a), MARK2 multi-laser package detail (b).**

The MARK2 laser package shown in Figure 5 uses a standard ridge-guide DFB laser diode architecture[20] to implement one measurement wavelength per laser die. A series of up to 12 laser diodes per package has been achieved and demonstrated in the MARK2. Four separate butterfly packages are integrated via optical interconnects into a single laser aperture interface. This necessitates the use of significant fiber optics and results in a relatively large optical aperture input. Size, cost, and assembly complexity of the instrument are all adversely affected by this design choice. In the MARK3 sensor design, a novel multi-wavelength per laser die architecture has been implemented and tested by nanoplus Nanosystems and Technology GmbH .

Laboratory Results on the MARK2 Sensor

The sensitivity of a test refers to a method's ability to respond to quantity changes in the target analyte, while selectivity is the extent to which analytical measurement procedure are high sensitivity and high selectivity for the target analyte (e.g. alcohol concentration). Ensuring the selectivity of an analyte measurement can be notoriously challenging in complex systems such as human tissue [21, 22]. Accordingly, careful design and controlled experiments are required to verify the validity of any measurement approach.

Historically, researchers have used *in-vitro* experiments to assess the sensitivity and selectivity of methods for quantifying analytes at physiological concentrations [23,24,25,26,27,28,29]. These experiments are useful diagnostics for the validity of a measurement approach because sample composition and the experimental conditions are controlled by the practitioner; allowing direct assessment of measurement sensitivity and selectivity. For laboratory validation of the alcohol sensor, an optically scattering tissue phantom was developed using 0.3 micron diameter polystyrene microspheres to mimic the optical properties of human skin.

In order to assess the accuracy of the MARK2 Sensor platform, a 98 sample set of *in vitro* calibration test samples were fabricated. The individual samples contained varying levels of glucose, urea, creatinine, phosphate buffered saline, triton (surfactant), 0.3 micron diameter polystyrene microspheres, and ethanol. Glucose, urea, and creatinine were varied across the full expected physiological range. Ethanol was varied from 0 mg/dL to 200 mg/dL (0.000 to 0.200 BAC). All seven constituents were varied across seven discrete concentration levels in an orthogonal experimental design intended to minimize any spurious correlations between their concentration levels.

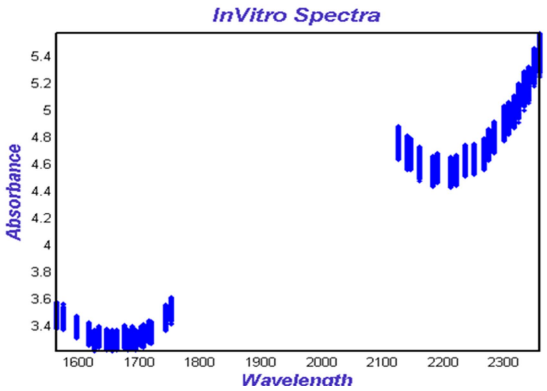

Figure 6. **Absorbance Spectra of 98 In Vitro Calibration Samples: MARK2 System.**

Figure 8 shows the spectral measurements of the 98 *in vitro* calibration samples. The vertical spread on the measurements is caused by the intentional variation of the concentration of the polystyrene scattering microspheres (mimics the differences in human skin optical properties).

A linear regression model was built using partial least squares regression and analysis was performed using a sample-out cross validated analysis to assess the performance of the MARK2 system in measuring ethanol. The orthogonal experimental design provides a solid analytical framework to assess the impact of varying optical scatter levels and changes in representative biochemical compounds (glucose, creatinine, urea) on the system's performance.

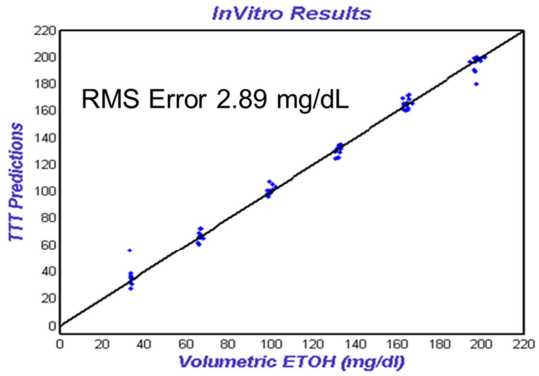

Figure 7. **CV Analysis Results: MARK2 Test 98 Sample Set.**

In *Figure 7* the results are shown for the cross-validated (CV) analysis. The RMS error shown is a robust estimate based on scaling the median absolute deviation by 1.46. The results shown on the MARK2 demonstrate excellent performance against the calibration sample. However, in order to meet the stringent performance requirements for the automotive application, additional performance improvements are being designed into the MARK3 Sensor.

DESIGN OF THE MARK3 SENSOR

Although significant progress has been made towards establishing the feasibility of a non-invasive touch based alcohol measurement system, continued research and development is necessary to achieve a production automotive system that can meet aggressive targets for performance, measurement time, reliability and robustness. Several key considerations in the touch based design are explored further below.

Compact Laser Module Design

Figure 8 shows a dual-segment laser diode mounted on a custom ceramic subcarrier. By selectively tuning the drive current, rapid and stable switching across multiple target laser lines can be achieved. This new laser architecture has enabled significant reduction in the number of laser die required by the MARK3 sensor. Additionally, improvements in the optical mounting and testing scheme allows for full laser diode qualification testing prior to assembly in the final opto-mechanical package. As a result, the MARK3 sensor form factor has been reduced to single, monolithic package suitable for automotive fleet testing of the sensor.

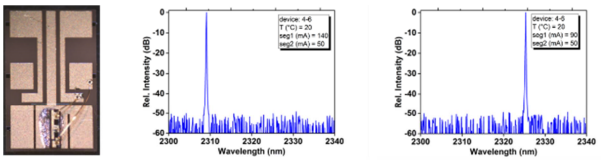

Figure 8. **Dual-segment laser mounted on ceramic carrier (a), Laser output 1 (b), Laser output 2 (c).**

Human Machine Interface

As a result of the reduced optical apertures enabled by the new laser package design, the skin optical interface of the sensor has been further optimized. The MARK2 sensor currently requires a 4mm x 4mm touchpad for the automotive interface. The proposed MARK3 design achieves over a 4x reduction in area, resulting in less than a 2mm x 2mm touchpad. This reduction in size offers the opportunity to further optimize the HMI design with respect to mechanical form factor, inclusion of user feedback elements, etc. Based on user testing, additional improvements in the visual and

auditory feedback elements are also being incorporated in the MARK3.

CONCLUSIONS

The MARK2 revision of the touch-based alcohol sensor has been implemented and installed in a demonstration vehicle. Initial testing on *in vitro* calibration samples have shown promising performance and provided insights for
in size and complexity have been enabled via a novel laser die architecture in conjunction with overall improvements in optical and electronic subsystems. Additional design and testing efforts are underway to continue progress towards meeting the requirements of the automotive application.

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