35th Annual International Conference of the IEEE EMBS
Osaka, Japan, 3 - 7 July, 2013

Local Pulse Wave Velocity Estimation using Magnetic Plethysmograph

Anand Chandrasekhar[1], Jayaraj Joseph[2], Mohanasankar Sivaprakasam[1,2]

***Abstract*— Pulse Wave Velocity(PWV) is an established measure of arterial stiffness. We present a method of measuring local pulse wave velocity by the use of Magnetic Plethysmograph(MPG) sensors.The design of a Compact Single element MPG (CS-MPG) sensor is presented. The functionality of the sensor is verified by phantom experiments. The utility of this sensor for in-vivo measurements of PWV is also demonstrated. Further, a Dual-element MPG (D-MPG) for evaluation of local PWV is also presented. The error in measurement of PWV using this sensor is characterised experimentally and shown to be within acceptable limits. The ability of this dual element sensor to measure local PWV in-vivo is also demonstrated by trials on volunteers.**

I. INTRODUCTION

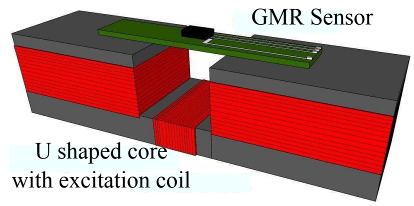
PWV is a clinically accepted estimate of arterial stiffness, with proven utility in cardiovascular risk evaluation [1]. Measurement of PWV involves the detection of pulse waveforms at two different points along the arterial tree. Pulse waveforms can be extracted from pulse points. Some of the most common pulse points are over radial artery (wrist), carotid artery (neck), brachial artery (upper arm), femoral artery (inner thigh), popliteal artery (knee) etc.. Based on the phase shift in the signals extracted from these pulse points, we can calculate PWV. If T is the time taken by the pulse to travel across a distance of D of the arterial tree, then

$$PWV = \frac{D}{T} \tag{1}$$

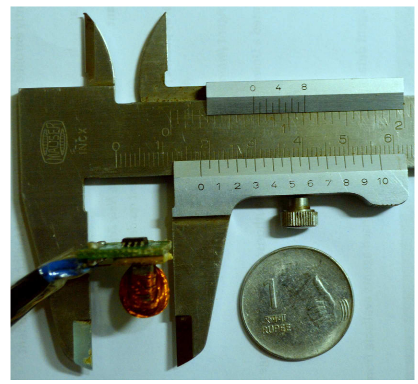
Several instrumentation modalities exists to evaluate PWV [2]- [12]. They extract signal from pulse points based on different principles. For instance, PWV is estimated based on optical property of blood [2]- [5]. Asmar *et.al.* uses a pressure based transducers to study PWV [6]. A nanosecond pulse near field sensing (NPNS) is also used to estimate PWV [11]. Designs based on oscillometry [7], Doppler ultrasound [8], applanation tonometry [9], MRI based imaging methods [10] were some of the earlier reported publications in PWV evaluation.

We had previously introduced a novel MPG method for non-invasive detection of the blood flow pulse [13]. Here we present the design of a CS-MPG transducer for effective detection of blood pulse and evaluation of PWV. Further, a D-MPG transducer designed specifically for the purpose of measuring local PWV at two points as close as 15 mm along an arterial tree is also illustrated.

[1]Department of Electrical Engineering, Indian Institute of Technology Madras, India.
[2]Healthcare Technology Innovation Centre, Indian Institute of Technology Madras, India.
jayaraj85@gmail.com



(a) 3D model of CS-MPG.



(b) Prototype of CS-MPG.

Fig. 1: (a) Design of CS-MPG sensor (b) Photograph of the prototype of CS-MPG sensor.

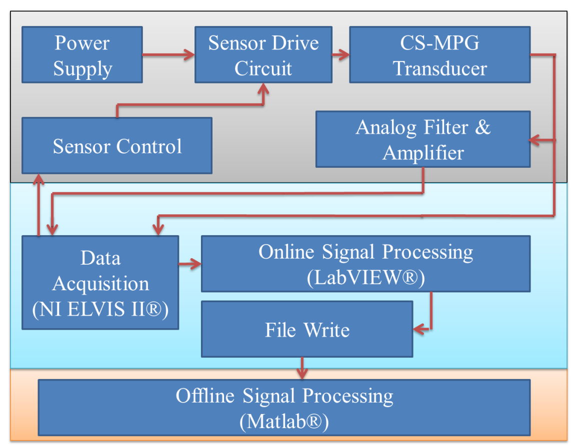
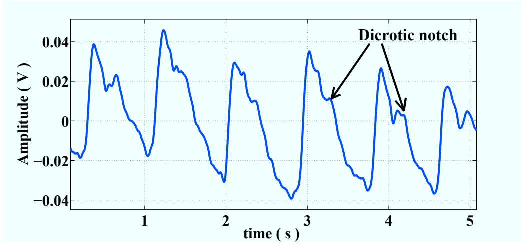


Fig. 2: System Architecture of the measurement system.

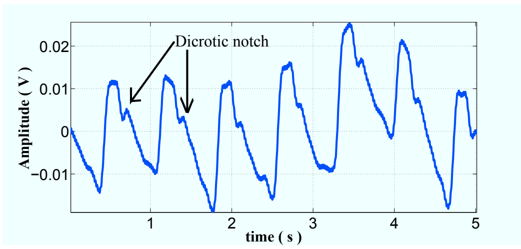
II. COMPACT SINGLE ELEMENT MPG

A. Design

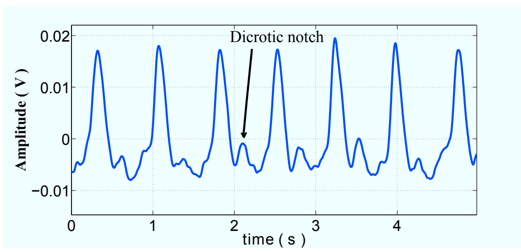
We had presented a permanent magnet and electromagnet based MPG (EM-MPG) [12] [13]. Our new transducer rectifies some of the major drawbacks of these previous versions of MPG. The design of CS-MPG is shown in Fig. 1. We have replaced the vertical magnetic core of EM-MPG with a U-shaped core. The core is wound with 300 turns of wire



(a) CS-MPG output from right common carotid artery.



(b) CS-MPG output from left common carotid artery.



(c) CS-MPG output from radial artery.

Fig. 3: Signal extracted from different pulse points using CS-MPG

(gauge size 22). With this design we are able to bias the GMR (AAH002-02, NVE Corporation) based field sensor to the linear region with a current of 20mA. The present design thus enables stable operation of the sensor in its linear region, at much lower excitation current levels. This reduces the power dissipation in the coil, and alleviates the heating issue observed in earlier design [12]. The overall transducer dimensions has also been reduced by a factor of 5. The overall system architecture of the measurement system is shown in Fig.2. Coil of CS-MPG is powered with a DC voltage source. Voltage is chosen such that GMR operates in the linear region. Output of CS-MPG is band limited with a dedicated analog filter. This signal is further amplified and sampled at the rate of 10kSa/s with a resolution of 16bits/Sa. Signal sampling and digital conversion is done with NI-ELVIS II® (a hardware platform from National Instruments). The acquired signal is processed further using LabVIEW®. A copy of the signal is saved for offline calculation.

Fig. 3 shows different output signals from carotid and radial arteries of a volunteer extracted using CS-MPG . These are the characteristic signal of a pulsatile blood waveform. Dicrotic notch, due to the interaction between the forward moving pulse and the reflected pulse, is indicated in all the waveforms.

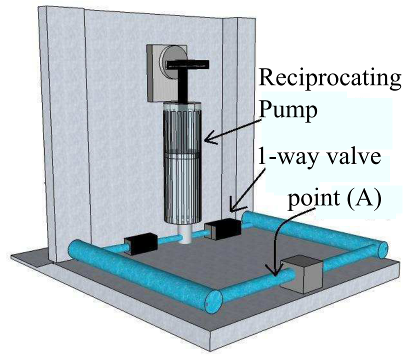


Fig. 4: This is the 3D model of the phantom used to validate CS-MPG.CS-MPG will be placed at point (A).

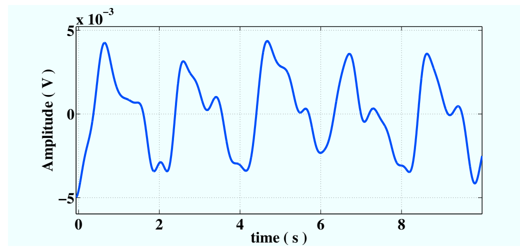


Fig. 5: CS-MPG output from the phantom operated at 0.5Hz.

B. Phantom Validation

A phantom was designed to validate the CS-MPG output. 3D model of the phantom is shown in Fig. 4. The overall structure of the phantom is made with a 5mm thick mild steel. Pumping action of the heart is mimicked with a reciprocating pump coupled to a servo motor of torque 24kg.cm. Injection and suction cycle of the pump maps to the systole and diastole of the heart respectively. Two one-way-needle-valves are placed at the entry and exit points of the pump. They emulate the functioning of mitral and aortic valves of heart. Flexible tubes are used to model the artery. High pressure injection of pump produce a systolic wave via flexible tubing which helps to recreate the pulsatile blood flow motion. Instead of blood, we have used a solution of ferrous compound as the circulating fluid. The pump is controlled with a programmable system which gives us the flexibility to operate the phantom at any desired frequency. Nature of the output has a strong resemblance to the standard blood-flow waveform. Typical output from the phantom, operated at 0.5Hz, extracted using a CS-MPG, is shown in Fig. 5.

C. In-vivo Measurements

A test was conducted to estimate the PWV using CS-MPG design. A pair of CS-MPGs were used to perform this experiment. Sensors were placed over the left common carotid artery of a volunteer. Outputs from both the transducer were recorded.

Using a semi-automated frequency domain algorithm, phase shift between the outputs were estimated offline. Algorithm tries to find the phase shift in the corresponding

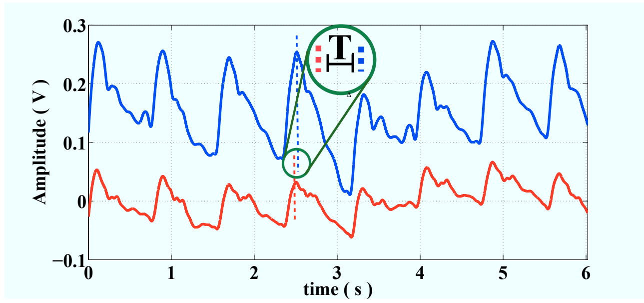


Fig. 6: Outputs from a pair of CS-MPGs, placed over the carotid artery, 1.5cm apart. Phase shift between the outputs, T, is indicated in the graph.

TABLE I: PWV estimated in the carotid artery using a pair CS-MPGs.

Sensor Distance(D)(m)	Phase shift(T)(s)	PWV(m/s)
0.015	0.0033	4.5
0.02	0.0046	4.3
0.025	0.0051	4.9
0.027	0.0054	5

component frequencies in the captured signals. PWV was calculated using Eq.(1). Experiment was iterated for varying distance between the MPGs. Phase shifts in the outputs recorded from MPGs, placed 1.5cm apart, is shown in Fig. 6. Results of the experiment are summarised in Table I. From Table I, its clear that, phase shift, T between the outputs increases as the distance, D between the sensors increases.

III. Dual element MPG

A. Design

In this section we present an improved version of the previously reported design of Dual element MPG(D-MPG) transducer [12]. In the earlier design, we had a dedicated excitation and compensation coil for biasing the GMR sensors of D-MPG. This makes the transducer difficult to handle. Moreover, we need to optimise the exitation and compensation coil voltage to get both the sensor in the linear region of operation. In the improved version of D-MPG, we present a design, which uses a single magnet

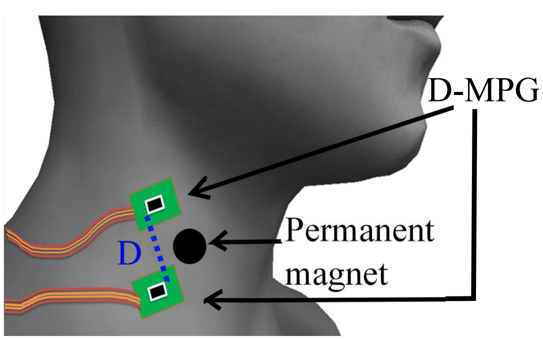


Fig. 7: A schematic picture of D-MPG biased with a permanent magnet placed over the right common carotid artery.

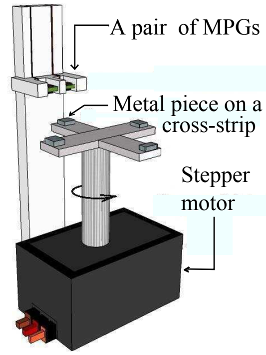


Fig. 8: Experimental setup to study the phase shift in the output of two MPG placed 10mm apart with a tolerance of 20mm from the rotating cross-strip.

to bias both the GMR sensors of the transducer. We have experimented the design with a permanent magnet and an electromagnet.Pictorial representation of D-MPG transducer, biased with a permanent magnet, placed over the right common carotid artery is shown in Fig. 7.

B. Phantom Validation

An experimental setup was designed to study phase shift calculation from D-MPG outputs. 3D model of the setup is shown in Fig. 8. Experimental setup involves a stepper motor which has a resolution of 1.8°. A cross-strip made out of plastic was mounted on the stepper motor. A metal piece was placed at the four ends of the cross-strip. Stepper was controlled by a programmable chip. D-MPG sensors where mounted with a 2cm tolerance from the rotating cross strip. MPG sensors where placed 10mm apart.Experiment was iterated for different rpm of the stepper motor. Data

TABLE II: Phase shift in D-MPG outputs, with D=10mm, for different metal velocity.

Metal Velocity(V)(m/s)	Phase shift(T)(s)	Sensor distance(D)(mm)
9.42	1.05	9.89
18.9	0.550	10.3
37.7	0.279	10.5
75.4	0.134	10.1
150.8	0.070	10.6
301.6	0.034	10.3

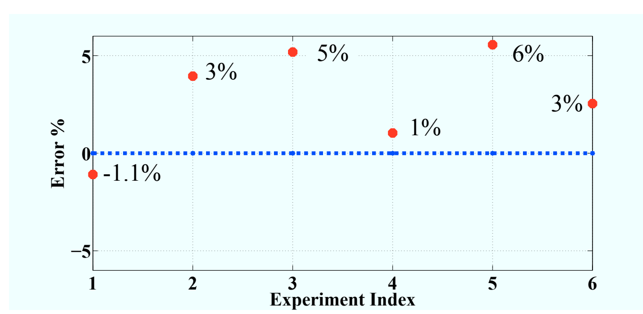


Fig. 9: Worst-case percentage error in phase shift measurement using D-MPG.

TABLE III: PWV estimated in the carotid artery using D-MPG.

Sensor Distance(D)(m)	Phase shift(T)(s)	PWV(m/s)
0.020	0.0044	4.5
0.023	0.0047	4.9
0.025	0.0051	4.9
0.027	0.0053	5.1

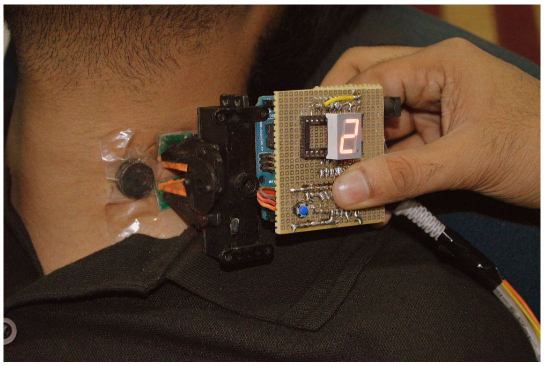


Fig. 10: Experimental setup of a D-MPG placed over the left common carotid artery of a volunteer. Sensor is mounted on a position controllable servo motor to precisely vary the distance between the D-MPG sensors.

from both the MPGs were recorded for offline calculation.

If phase shift is T and metal piece velocity is V , then sensor distance D is given by the equation:

$$D = V \times T \tag{2}$$

$$\Rightarrow \frac{\Delta D}{D}\% = \frac{\Delta V}{V}\% + \frac{\Delta T}{T}\% \tag{3}$$

Results of the analysis is summarized in Table II. It is clearly visible that distance between the sensors calculated based on the phase shift is in the expected range.

Stepper motor has a worst-case error of 3%. Worst case error from phase shift estimated using D-MPG, calculated using Equ. (3) is shown in Fig. 9. Error in phase shift measurement estimated using MPG is as low as 6% which makes D-MPG is a reliable device for estimating arterial stiffness [1].

C. In-vivo Measurements

In-vivo tests were conducted using D-MPG transducer. Tests were iterated for varying distance between the sensors. PWV was calculated for each distance. Results of the tests are summarised in the Table III. Measurement of local PWV using a dual element MPG falls within the expected range [14]. The test setup is shown in Fig. 10.

IV. CONCLUSION

The design of MPG sensors for non-invasive detection of blood pulse and evaluation of PWV was presented. The functionality of a compact single element MPG sensor was illustrated by phantom experiments, and its utility in measuring PWV was also demonstrated by in-vivo trials. Further, the capability of a dual-element MPG transducer in accurately determining local PWV was demonstrated. The in-vivo measurement error was found to be less than 6%. The experimental and in-vivo results presented here illustrate the ability of properly designed MPG transducers to easily measure an accurate blood pulse waveform and evaluate PWV without using an ECG signal. Such sensors could pave to way for an affordable and effective cardiovascular screening devices based on PWV measurements.

REFERENCES

[1] S. Laurent, J. Cockcroft, L. Van Bortel, P. Boutouyrie, C. Giannattasio, D. Hayoz, B. Pannier, C. Vlachopoulos, I. Wilkinson, and H. Struijker-Boudier, "Expert consensus document on arterial stiffness: methodological issues and clinical applications," *European Heart Journal*, vol. 27, no. 21, pp. 2588–2605, 2006.

[2] P. Salvi, E. Magnani, F. Valbusa, D. Agnoletti, C. Alecu, L. Joly, and A. Benetos, "Comparative study of methodologies for pulse wave velocity estimation," *Journal of human hypertension*, vol. 22, no. 10, pp. 669–677, 2008.

[3] R. Boehmer *et al.*, "Continuous, real-time, noninvasive monitor of blood pressure: Penaz methodology applied to the finger." *Journal of clinical monitoring*, vol. 3, no. 4, p. 282, 1987.

[4] K. Li and S. Warren, "Initial study on pulse wave velocity acquired from one hand using two synchronized wireless reflectance pulse oximeters," in *Engineering in Medicine and Biology Society, EMBC, 2011 Annual International Conference of the IEEE*. IEEE, 2011, pp. 6907–6910.

[5] H. Wu, C. Ho, J. Weng, W. Tsai, and M. Wang, "A novel method for measurement of pulse wave velocity by dual-channel photoplethysmography," in *Biomedical Circuits and Systems, 2004 IEEE International Workshop on*. IEEE, 2004, pp. S2–6.

[6] R. Asmar, A. Benetos, J. Topouchian, P. Laurent, B. Pannier, A. Brisac, R. Target, and B. Levy, "Assessment of arterial distensibility by automatic pulse wave velocity measurement: validation and clinical application studies," *Hypertension*, vol. 26, no. 3, pp. 485–490, 1995.

[7] M. Cortez-Cooper, J. Supak, H. Tanaka, *et al.*, "A new device for automatic measurements of arterial stiffness and ankle-brachial index." *The American journal of cardiology*, vol. 91, no. 12, p. 1519, 2003.

[8] S. Kontis and R. Gosling, "On-line doppler ultrasound measurement of aortic compliance and its repeatability in normal subjects," *Clinical Physics and Physiological Measurement*, vol. 10, no. 2, p. 127, 2001.

[9] I. Wilkinson, J. Cockcroft, D. Webb, *et al.*, "Pulse wave analysis and arterial stiffness." *Journal of cardiovascular pharmacology*, vol. 32, p. S33, 1998.

[10] V. Itskovich, K. Kraft, and D. Fei, "Rapid aortic wave velocity measurement with mr imaging1," *Radiology*, vol. 219, no. 2, pp. 551–557, 2001.

[11] H. Lin, Y. Lee, and B. Chuang, "Using dual-antenna nanosecond pulse near-field sensing technology for non-contact and continuous blood pressure measurement," in *Engineering in Medicine and Biology Society (EMBC), 2012 Annual International Conference of the IEEE*. IEEE, 2012, pp. 219–222.

[12] A. Chandrasekhar, J. Joseph, and M. Sivaprakasam, "A novel magnetic plethysmograph for non-invasive evaluation of arterial compliance," in *Engineering in Medicine and Biology Society (EMBC), 2012 Annual International Conference of the IEEE*. IEEE, 2012, pp. 1169–1172.

[13] J. Joseph and V. Jayashankar, "Magnetic sensor for non-invasive detection of blood pulse and estimation of arterial compliance," in *Biomedical Engineering and Sciences (IECBES), 2010 IEEE EMBS Conference on*. IEEE, 2010, pp. 170–175.

[14] N. Keenan, P. Gatehouse, R. Mohiaddin, D. Firmin, and D. Pennell, "Carotid artery pulse wave velocity measurement by cardiovascular magnetic resonance," *Hounsfield Memorial Lecture, Imperial College London*, 2006.