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Method of deflection corrected tonometry with phantom vessel experiments

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ABSTRACT

Background: The Continuous Non-Invasive measurement of arterial Blood Pressure [CNIBP] is possible via the method of arterial tonometry and the arterial volume clamp methods. Arterial tonometry successfully measures continuous arterial pressure but requires large vessel deformation and a highly miniaturized pressure sensor to obtain a direct calibration of pressure. A properly designed tonometer is capable of achieving pressure accuracy of less than 5% error at the radial artery. The volume clamp method achieves comparable errors but is generally restricted to the very peripheral arteries. Since the brachial or radial arteries are preferable sites to record blood pressure, tonometry is generally preferred. However, due to its strict operating requirements, tonometry requires a highly skilled operator. The greatest source of measurement error results from slight deviation from the artery wall applanation position. In this study, a method for correcting tonometry deflection error is introduced and evaluated using preliminary experiments.

Methods-modeling: In prior analysis it has been shown that arterial wall flattening causes contact stress to become uniform and equal to the arterial pressure. In this article, we derive the contact stress for deflections other than the ideal applanation position and to allow variable vessel deflection. This analysis permits the contact stress to be corrected for tonometer positions that are not exact so that pressure accuracy is maintained in spite of less than ideal positioning. This will alleviate the necessity for highly skilled users and allow rapid determination of the pulse pressure.

Methods-experimental: Experiments were performed to evaluate applied model corrections for tonometer accuracy versus vessel deflection. Two experiments were performed to evaluate tonometer accuracy when deflection is varied. The first experiment used no deflection correction and the second experiment applied model derived deflection correction. A force sensor was used to deflect a phantom latex vessel of known internal pressure. The corrected contact pressure was then compared with known pressures to evaluate the pressure accuracy.

Results-modeling: a geometric model was derived for vessel contact area versus deflection. This resulted in a formula that provides contact area continuously for any amount of deflection. Once the contact area is known the average tonometer contact pressure was obtained that corresponds with the vessel internal pressure.

Results – experimental: A latex tubing phantom vessel was pressurized to a known amount and was deflected in increments over its full diameter while measuring contact force at each position. The model-derived formula was then used to calculate pressure at each position. The calculated pressure was then compared with known internal pressure to evaluate pressure accuracy for all the phantom pressure and deflection points.

Conclusions: A modeling method for tonometer deflection correction was derived and evaluated using a phantom vessel. Average error was significantly reduced over the non-corrected data. The variability of error was also reduced for all data points collected. The experiments reveal that blood pressure measurement error can be reduced to levels obtained in near ideal tonometry conditions without the need for precise position control. The relaxed user precision is anticipated to simplify the use and design requirements for arterial tonometry in practice.

1. Background

Noninvasive blood pressure determination may be grouped into two categories. First, the occlusive cuff methods and second, the continuous noninvasive pulse methods. By far, the cuff methods are the most common devices used in general medicine. The cuff methods rely on the application of an inflatable arm cuff to apply a known pressure typically to the brachial artery of the arm. The vessel's response signal to external

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pressure is then used to determine when the blood pressure is equal to the known cuff pressure. A variety of arterial response detection methods are available such as the Korotkoff sound method and have been reviewed by Drzewiecki and Noordergraaf [1]. The long history and data available for the cuff methods have allowed them to attain the status of a respected standard for blood pressure measurement [2].

Continuous Noninvasive Blood pressure (CNIBP) methods attempt to sample blood pressure at the heart rate and faster so that the pulse waveform may be observed continuously in time. CNIBP methods are generally difficult to calibrate or are calibrated indirectly. The method of Penaz [3] has successfully calibrated the Photoplethysmograph sensor by means of continuous feedback pressure applied to the human digit. But, the Penaz method suffers from the fact that it measures at a location very distal to the heart, thereby causing significant pressure drop and waveform transmission effect errors. A recent version of the Method of Penaz applied the use of an arterial transfer function to calculate the brachial artery pressure from the finger pressure. This is known as the commercial Nexfin™ device [4]. The use of a generalized arterial transfer function is yet questionable especially in those patients that are hemodynamically unstable [5]. More recently, the pulse transit time (PTT) method also attempts to calibrate the PhotoPlethysmograph that to indirectly measure blood pressure. PTT measures the time that it takes for the arterial pulse to travel from a proximal to distal artery location. Time delay is then indirectly calibrated to blood pressure using the Moens Korteweg wave velocity equation. In accordance with Moens Korteweg equation, PTT depends on multiple factors that are not known. Moreover, the timing measure needs to be very precise since PTT in humans is in the range of mSecs. In spite of these problems, PTT blood pressure has attained popularity in consumer devices since it can be fashioned into low cost system capable of monitoring relative blood pressure change without accurate measurement of absolute blood pressure [6].

Arterial tonometry is another method of CNIBP that measures blood pressure in a more direct manner by measuring the contact pressure of a pressure or force sensor with an arterial wall that has been deflected to the point of full applanation Fig. 1 [7]. When this geometry is achieved, pressure balance occurs between the arterial inside pressure and the external contact sensor. The sensor pressure then reads the arterial pressure in a directly calibrated way as seen in Fig. 1AT. Moving away from the ideal applanated position results in large pressure error. The case of partial applanation is shown in Fig. 1.

A noise cancellation approach was applied by Ciaccio and Drzewiecki [21] to successfully correct for tonometer positioning artifacts. This approach, though, requires the use of multiple sensors. In this current research it was chosen to derive a method of correcting the sensor contact pressure for any amount of vessel deflection.

Right is incomplete applanation or partial AT (Applanation Tonometry). The T-sensor measures Force in contact with the flattened arterial wall over part of the sensor length T.

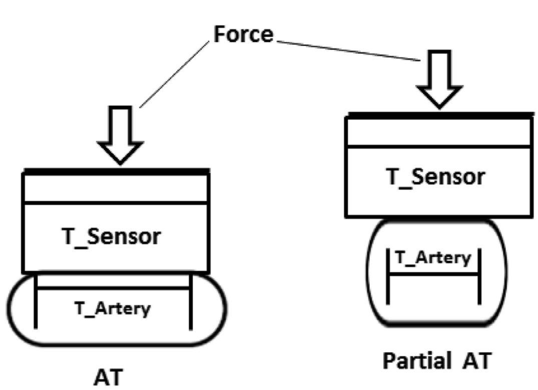

Fig. 1. Left is Full AT (Applanation Tonometry). The T-sensor is in contact with the arterial wall along its entire length.

Fig. 1 demonstrates the condition of full AT (Applanation Tonometry). For this condition the sensor length and area, Tsensor, is equal to the arterial applanation Length $T_{artery.}$ Since the sensor size is constant, the geometry requires sensor area to equal the arterial contact area. Then, the sensor force, $F_{s,}$ may be converted to blood pressure, BP, by BP $= F_s/A_{s.}$ The F is measured and the sensor area is a fixed known dimension. Tonometry applies to most arteries for which a superficial pulse may be detected. Due to the applanation requirement, vessels such as the radial artery are more appropriate since it is bone supported. Tonometry has been reported to permit continuous pressure measurement error of less than 5 mmHg [8]. Since tonometry does not allow the vessel wall to deflect, nonlinear wall properties do not interfere with obtaining accurate pulse waveform records. The precise conditions of applanation have been analyzed mechanically by Drzewiecki and Noordergraaf [9] and the analysis supports the observation of excellent continuous blood pressure accuracy. Their analysis has shown that deviation from full applanation will result in much larger errors. Alternative approaches to solve this problem have been to employ a positioning control system that was capable of measuring temporal artery pressure to within 7% accuracy [10]. Unfortunately, the preferred location for clinical blood pressure measurement is the arm at the brachial or radial artery. Tonometry at the brachial artery, though, has been shown to result in a poor relationship with the aortic pressure as opposed to the radial artery where full applanation may be accomplished [11].

The superior accuracy of tonometry for CNIBP lends encouragement to further research to refine the method to operate outside of ideal measurement conditions. When the tonometer is operating outside of perfect applanation conditions, we shall refer to this condition as deflection corrected tonometry as shown in Fig. 1 for partial applanation.

In this paper, a method for deflection corrected arterial tonometry for partial applanation is derived. This new method of deflection corrected tonometry will then be evaluated using phantom vessel experiments and recordings.

2. Methods

2.1. Methods - theoretical: deflection corrected pressure formula

To determine blood pressure from force during vessel deformation conditions, a geometric model of arterial contact deformation was developed to represent the vessel following a deflection by the force sensor. Following the approach of modeling arterial tonometry by Drzewiecki et al., [1983] [12], this was accomplished by equating the circumference of a cross-section of the undeformed tube to that of its deformed state. Fig. 2 describes a cross section of a deformed vessel. For large deflections the deformed geometry is used.

$$P = F/A_s \tag{1}$$

Since Force, F, is measured, the Pressure can be only be determined with knowledge of the contact Area As. In full applanation tonometry, the contact area, As, is equal to the sensor dimensions. But, for any other condition of partial Applanation A_s depends on the geometry in Fig. 2 and the amount of vessel deflection D. Taking the sensor contact

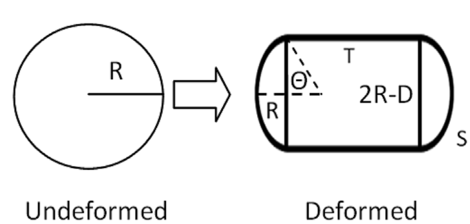

Fig. 2. The cross-sectional geometries of an undeflected artery (undeformed) and a deflected artery (deformed) are shown. Pressure is determined from measured Force, F, and the contact area $A_{s\ \text{from}}$ equation (1) as.

area as rectangular, the two sides of the contact area are the length of the sensor, D_s, and the vessel wall contact length T. So, the contact area becomes

$$A_s = D_s * T \tag{2}$$

The area As from eq. (2) may be inserted in to eq. (1) to find Pressure. Unfortunately, for the deformed vessel T is not yet known. So, to complete the analysis, turn next to the determining the value of T in terms of deflection D.

Referring to the deformed geometry in Fig. 2, the deformed vessel wall consists of a flattened portion, T, which is the contact Length, L, with the sensor, and a curved arc, S. By equating the circumferences and describing the arc length, S, as a function of the deflection, D, the contact length, T, can be solved for as a function of the amount of deflection D. The accuracy for T can be improved by using a large deflection and small deflection equation. We define large deflection as greater than 0.6 of the radius and small deflection for less than this amount. For large deflection, equation (5) results below.

$$2\pi R = 2*T + 2*S \tag{3}$$

$$S = (2\pi R)*\left(\frac{2\theta}{2\pi}\right) \text{ with } \theta = \sin^{-1}\frac{R-\frac{D}{2}}{R} \tag{4}$$

$$T = R*\left(\pi - 2*\sin^{-1}\left(1-\frac{D}{2R}\right)\right) \qquad D > 0.6*R \tag{5}$$

For small deflections the arc length, S, can be approximated as a semicircle. The semicircle representation becomes exact at the point when the artery has been deflected to greater than 0.6 of the original diameter or when the chord of S is exactly 2R. By applying this approximation, the relationship for contact length T, for small deflections is eq. (7) as follows.

$$S = \left(2\pi\left(R - \frac{D}{2}\right)\right) \tag{6}$$

$$T = \left(\pi*\frac{D}{2}\right) \qquad D < 0.6*R \tag{7}$$

The contact area is obtained by multiplying the sensor diameter, Ds, by the contact length, T, after choosing the appropriate formula for the given deflection. For large deflections eq (5) is used for the contact length, while for smaller deflections eq (7) is used.

2.2. Methods-experimental

Two experiments were performed to evaluate the accuracy of the pressure correction formulae. The first experiment was performed on a phantom blood vessel to measure the wall contact area at various vessel deflections to validate our model of contact Area. The second experiment recorded the contact force with a pressurized phantom blood vessel at various internal pressures and deflections. All combinations of pressure and deflection were then used to determine the deflection corrected pressure error versus the errors tonometry without deflection correction.

2.3. Phantom vessel experiments

Fig. 3 describes the experimental set-up designed to apply a known deflection to a phantom vessel and measure the force applied using a force sensor. The phantom vessel was a latex tube of 0.635 mm wall thickness and 3.175 mm diameter to simulate the human radial artery. The vessel was pressurized using an occlusive arm cuff and pressure gauge to provide a known calibrated pressure. The phantom vessel pressure was adjusted to various pressures using blood pressure cuff hand pump and release valve. The central portion of the vessel was placed within a mini-vice Foredom 37210 in contact with a Honeywell FSG15NIA piezo-resistive based force sensor. The mini-vise was used to deflect the vessel in fixed increments while the sensor would measure the contact force applied. The opposite end of the vessel was plugged to contain the air pressure. A Dc voltage of 5 V (BK Precision power source) was applied to the force sensor to provide a constant excitation voltage. The output of the sensor's internal bridge circuit was recorded using a digital voltmeter in DC volt mode.

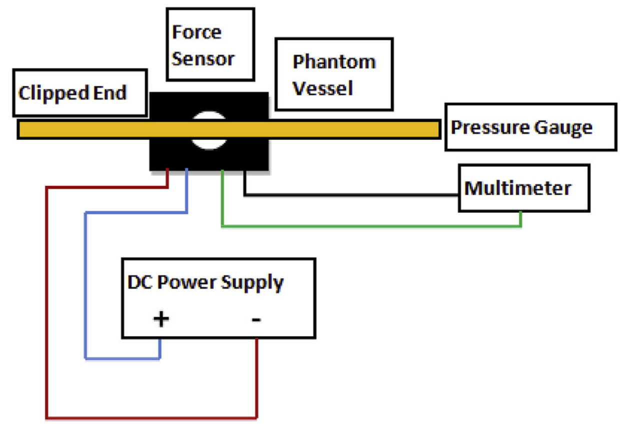

Fig. 3. Phantom Vessel Experimental Set Up. The phantom vessel is shown as deflected by a force transducer placed perpendicular to the vessel. Not shown is a mini-vice screw that moves the sensor with respect to the vessel. DC voltage is supplied to the sensor and its output measured using a Digital-volt- meter.

2.4. Methods- experimental: contact area measurement procedure

Since the sensor contact area is critical to obtain accurate blood pressure, the first experiment obtained data for contact area and deflection so that the model equations for contact area can be validated. A contact area imprint technique was developed to directly measure the contact area of the phantom vessel with a flat surface. First, the vessel deflection area was covered in lipstick and, then, deflected with the mini-vice from the initial contact position to .0030 mm in increments of 0.0005 mm with the force sensor replaced by a glass slide. The lipstick imprint on the slide was later measured to observe the contact area.

2.4.1. Sensor calibration

Prior to any of the pressure determination experiments, the force sensor was calibrated to convert output voltage to force. Several trials of placing a one-pound weight on the sensor were performed while measuring the output voltage. For a 5 V excitation voltage trials resulted in an average sensitivity of 10 ± 0.7 mV/Nt. Since it is typical for this kind of sensor to exhibit a nonzero output at zero force (zero offset voltage), the zero offset voltage was measured at the beginning of every trial and then subtracted from every subsequent data point.

2.5. Methods-experimental phantom vessel deflection corrected tonometry procedure

The vessel was subjected to different deflections and internal pressures to evaluate the accuracy of determining the phantom vessel pressure with deflection corrected tonometry. Two sets of trials were performed, one for constant deflection with varying pressure and a second for constant pressure with varied deflection. The first trial applied constant deflections to the vessel over a range from 0.0005 mm to 0.0030 mm with increments of 0.0005 mm. For each deflection, the vessel pressure was varied from 0 mmHg to 140 mmHg in steps of 20 mmHg. The voltage output of the force sensor was measured at each of the variations. Sensor output voltage was recorded at every variation. The sensor calibration was then used to convert sensor voltage into tonometric contact force. After determining the force applied, the

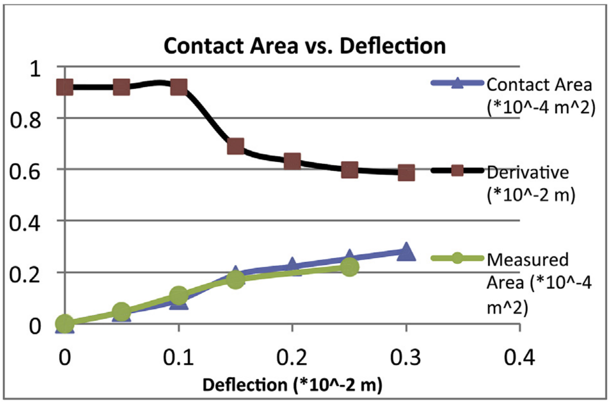

Fig. 4. Graph of the Contact Area against Deflection as given by eqs (4) and (6). The bottom curves are the calculated contact area (triangles) and measured contact area (circles). The top line is the derivative of contact area from the model equations. The sharp drop in derivative occurs at the transition from small to large deflection.

contact length was obtained using either eq (5), for large deflections of 0.0015 mm and above, or eq (7), for small deflections of 0.0005 mm and 0.001 mm, then pressure was calculated using eq (1).

2.6. Methods-deflection corrected tonometry pressure, phantom vessel trials

The next set of trials examined the deflection correction method for a fixed amount of deflection and pressure using the phantom vessel over multiple trials. In this experiment, each trial consisted of deflecting the phantom vessel with an internal pressure of 100 mmHg from zero to 0.0020 m and calculating the pressure in eight separate trials to provide an indication of the data variability.

2.7. Results - contact area model validation

The contact length, L, and area was imprinted on the slide by the lipstick stained vessel was then measured and used to graph the contact area versus deflection. The contact area was also calculated at the corresponding deflection using from model equation (5) or (7). Fig. 4 graphs the measured contact areas together with the model calculated areas versus deflection. Also shown is derivative of the contact area with respect to deflection. The area deflection lines were merged when the model changes from the small to large deflection type. This is observed at the deflection of 0.0015 mm.

2.8. Results - phantom vessel partial applanation tonometry

2.8.1. Data analysis and statistics

All of the phantom vessel force data were grouped into the deflection corrected tonometry and the non-deflection corrected tonometry groups. In the deflection corrected group, the sensor force was converted to pressure using the model formulae eqs (1)–(7). In the case of the non-corrected tonometry, the pressure was determined from the sensor contact stress using just eq. (1) as in conventional tonometry.

2.8.2. Statistical data analysis

Corrected and non-corrected data are presented in separate Bland Altman plots in Figs. 5 and 6 as the error between the corrected pressure and the actual phantom internal pressure.

Deflection corrected tonometry error for all deflection corrected data error is shown as the difference between the calculated pressure and the actual phantom pressure.

Figs. 5 and 6 reveal that pressure error has decreased from 15 down to 5 ± 10 mmHg. This represents a statistically significant amount of error reduction to the level of $p < 0.023$.

2.9. Results: multiple –trials phantom with deflection corrected pressure

Fig. 7 illustrates phantom pressure determination in seven trials of deflected corrected tonometry to determine the phantom pressure of an internal value of 100 mmHg. This experiment revealed a measured mean pressure value of 107 ± 8 mmHg, which is less than 10% error for all trials. The calculated pressure using eq (1) was verified against the known phantom pressure and therefore the absolute accuracy of force to pressure determination can be evaluated. It was found that pressures derived from measured contact force were consistent with applanation tonometry for a wide range of deflections as shown in Fig. 7. The greatest error occurred for deflections 1.5 and 02.5 mm are attributed to experimental error in measuring low force.

3. Discussion

Currently, only a few available methods are capable of providing true continuous non-invasive blood pressure (CNIBP) measurement. Other pulse sensor techniques are most often pulse plethysmographs based on optical or capacitive deflection sensors. While these methods are capable of providing pulse timing and waveform information, they typically require another method such as a cuff to provide an indirect calibration. Moreover, Plethysmographic records are subject to pulse wave distortion due to nonlinear arterial wall properties. Arterial tonometry has been shown to provide true CNIBP with pressure error of less than 5%. Unfortunately, arterial tonometry relies on the special condition of arterial applanation such that the vessel wall contact area becomes approximately equal to the sensor area. For this condition to be met, the artery must be deflected to a large extent. Moreover, the tonometer operator must maintain the deflected position very carefully so not to introduce measurement error. One approach to solving this issue is to employ an automatic positioning system as introduced by Zhang et al. [10]. In our study, a tonometer deflection method was introduced such that the applanation condition can be eliminated. Phantom vessel experiments have shown that the deflection corrected tonometry allows for arterial pressure measurement at much smaller arterial deflection conditions and yet obtain accuracy comparable to that of full applanation tonometry. Our method affords all of the benefits of arterial tonometry such as direct calibration and good waveform accuracy under small deflection conditions. The authors believe that these phantom vessel experiments have provided sufficient validation of this new method of pressure correction such that it may now be suitable for human studies. Moreover, in comparison with the cuff methods the method provided here allows for measurement of blood pressure in approximately half the time but yet with similar accuracy. If the sensor position is constrained, our method in addition provides true continuous arterial pulse recording (CNIBP). In comparison with the other pulse methods, deflection corrected tonometry does not require another method to obtain baseline calibration. In comparison with plethysmographic pulse methods our approach measures vessel contact stress as opposed to deflection for indirect measurement of pulse pressure. In comparison with the pulse wave velocity methods or pulse wave transit time our approach does not require precise pulse timing information. Lastly, our method does not require empirical calibration algorithms, formulae nor the need for multiple calibration parameters.

In comparison with full applanation tonometry, our deflection corrected tonometry method error is found to be less than 25% for all data reported here and 5–10% in the high deflection range. While low percent errors using high deflections have been previously reported in full applanation tonometry, it has been shown here that the same can be achieved for low deflections as well by using our deflection correction method. This is an improvement over applanation tonometry where errors are much higher as shown by Drzewiecki and Noordergraaf (1979) [13] for low deflection conditions. Moreover, the force – pressure relationship eliminates the problem of deflection dependent accuracy. Our data supports the case that the force pressure calibration

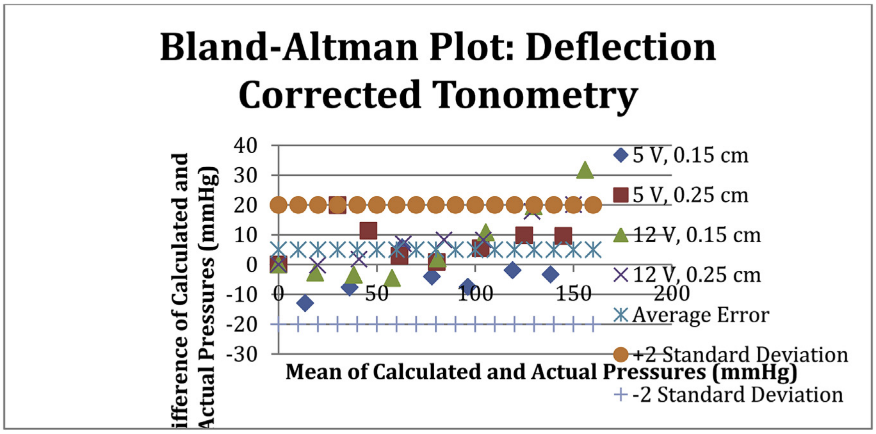

Fig. 5. The Bland – Altman plot of pressure error for all data points collected during partial applanation tonometry with deflection correction using model equations (1)–(6). Each point represents a different pressure-deflection pair. Horizontal Lines indicate the Average and ± 2 Standard deviations for all deflections measured.

can be made independent of vessel deflection. Arterial tonometry has attained value as a respected method for continuous noninvasive blood pressure measurement. This acceptance has occurred because of two factors. First the method has been independently evaluated for its ability to obtain accurate noninvasive blood pressure. Second, mechanical modeling has provided a solid physical support for its function. One of the first models of tonometry was a simple discrete spring and force sensor model [7]. We improved on this model by introducing a continuum model using curved beam segments of the arterial wall [1]. These prior analyses supported the concept of arterial applanation. Since the amount of arterial deflection must be maintained precisely to insure high accuracy of tonometry recent studies have modeled conditions outside of perfect applanation such as the finite element model of Singh et al. [18]. These researchers further alter the tonometer sensor geometry [17] such that a maximum pulse amplitude can indicate the point of calibration similar to the approach of oscillometry [16,20]. In this research article we also analyze the accuracy of tonometry for deflections at other than the ideal Applanation condition, which we have referred to as partial Applanation tonometry. The outcome shown here is that the tonometer may be calibrated at any arterial deflection. This significantly reduces the skill needed by the tonometer operator and will likely reduce the time to blood pressure reading. This improved application can assist in the use of the tonometer as an input [19] to obtain derived hemodynamic quantities such as Arterial tone [Tripathi [14] and cardiac output [Zayat et al., [15].

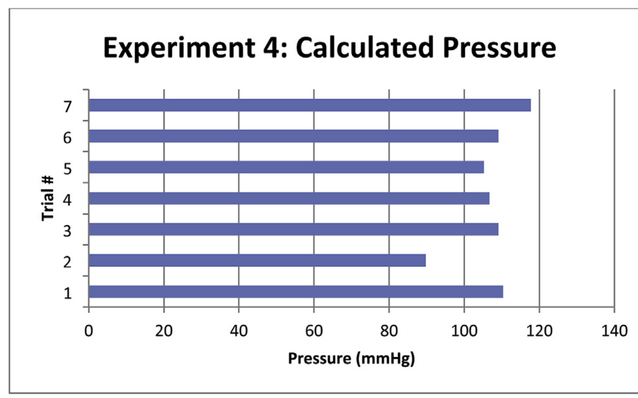

Fig. 7. Phantom pressure measurement using partial deflection tonometry with pressure correction. Seven trials of pressure determination are shown. The Phantom pressure was held constant at 100 mmHg. The vessel deflection was not controlled and was allowed to vary.

Conflicts of Interest

None declared.

4. Conclusion

The force tonometry deflection correction method presented in this

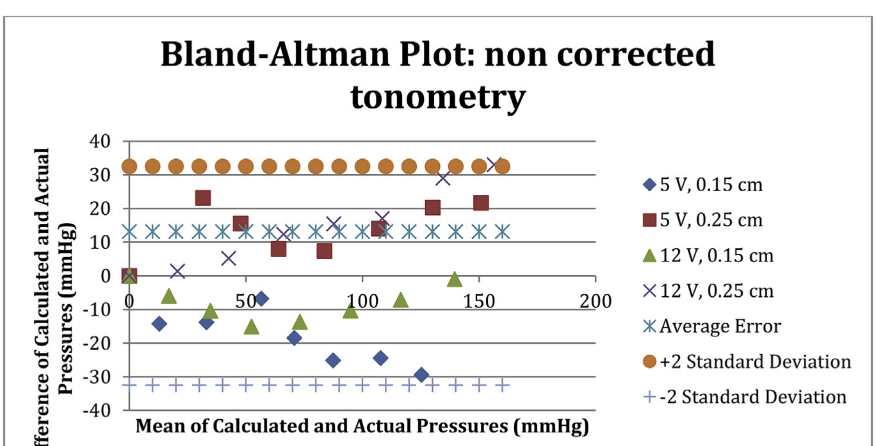

Fig. 6. The Bland – Altman plot of pressure error for all data points collected during partial applanation tonometry without deflection correction. Each point represents a different pressure-deflection pair. Horizontal Lines indicate the Average error and ± 2 Standard deviations is shown for all points.

article has shown that it is possible to operate a tonometer outside of the full applanation position and yet obtain accuracy comparable to most methods of CNIBP measurement pressure from pulse measurement.

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