Original Investigation

Estimating actual inspiratory muscle pressure from airway occlusion pressure at 100 msec

Natsumi T. Hamahata MD [1], Ryota Sato MD [1], Kimiyo Yamasaki RRT [2], Sophie Rodrigues Pereira MD [1], Ehab G. Daoud MD [1,2,3,4]

DOI: https://doi.org/10.53097/JMV.10003

Cite: Hamahata NT, Sato R, Yamasaki K, et al. Estimating actual inspiratory muscle pressure from airway occlusion pressure at 100 msec. J Mech Vent 2020; 1(1):8-13.

Abstract

Background:

Quantification of the patient's respiratory effort during mechanical ventilation is very important and calculating the actual muscle pressure (Pmus) during mechanical ventilation is a cumbersome task and usually requires an esophageal balloon manometry. Airway occlusion pressure at 100 milliseconds (P0.1) can easily be obtained non-invasively. There has been no study investigating the association between Pmus and P0.1. Therefore, we aimed to investigate whether P0.1 correlate to Pmus and can be used to estimate actual Pmus

Materials and Methods:

A bench study using lung simulator (ASL 5000) to simulate an active breathing patient with Pmus from 1 to 30 cmH_2O by increments of 1 was conducted. Twenty active breaths were measured in each Pmus. The clinical scenario was constructed as a normal lung with a fixed setting of compliances of 60 mL/cmH_2O and resistances of 10 $cmH_2O/l/sec$. All experiments were conducted using the pressure support ventilation mode (PSV) on a Hamilton-G5 ventilator (Hamilton Medical AG, Switzerland), Puritan Bennett 840™ (Covidien-Nellcor, CA) and Avea (CareFusion, CA).

Main results:

There was significant correlation between P 0.1 and Pmus (correlation coefficient = - 0.992, 95% CI: -0.995 to -0.988, P-value<0.001). The equation was calculated as follows: Pmus = -2.99 x (P0.1) + 0.53

Conclusion:

Estimation of Pmus using P 0.1 as a substitute is feasible, available, and reliable. Estimation of Pmus has multiple implications, especially in weaning of mechanical ventilation, adjusting ventilator support, and calculating respiratory mechanics during invasive mechanical ventilation.

Keywords: P 0.1, Inspiratory occlusion pressure, WOB, Esophageal balloon, mechanical ventilators, respiratory failure

Keywords: P 0.1, P mus, Inspiratory occlusion pressure, WOB, Esophageal balloon, mechanical ventilators, respiratory failure

Authors
1. Department of Internal Medicine, John A. Burns School of Medicine, University of Hawaii at Manoa, Honolulu, HI, USA
2. Castle Medical Center, Kailua, Hawaii, USA
3. Respiratory Care Program, Kapiolani Community College, Honolulu, Hawaii, USA
4. Critical Care Department, Kuakini Medical Center, Honolulu, Hawaii, USA
Corresponding author: Natsumi Hamahata MD: tanabena@hawaii.edu

Introduction

The goals for instituting mechanical ventilation are to promote safety (oxygenation and ventilation), comfort (patient-ventilator synchrony and optimization of work demand), and liberation (weaning process and timing on the ventilator). [1] In patients with acute respiratory failure, the inspiratory effort was reported to be expended to four-six times of the normal value. Mechanical ventilation is thought to benefit patients by unloading their respiratory muscle's effort. [2] However, mechanical ventilation is also associated with multiple potential complications, such as infection, cardiovascular compromise, ventilator-induced lung injury (VILI), and diaphragmatic atrophy that is categorized as ventilator-induced diaphragmatic dysfunction (VIDD). [3] Diaphragmatic weakness is very common in patients undergoing mechanical ventilation, and its recognition, as well as prevention, is crucial as diaphragmatic function plays a principal role in successful weaning from the ventilator.

Definition and measurement of Pmus

During mechanical ventilation, the total force (pressure) to generate a tidal volume is either generated by the ventilator alone (Pvent) in patients with no spontaneous efforts or in conjunction with both the ventilator and the patient's respiratory muscles (Pmus) in patients with spontaneous effort. Pmus could be estimated by using the equation of motion of the respiratory system with some variables for both non-ventilated and ventilated patients who breathe spontaneously. The equation listed below explains the pressures required by the ventilator or the patient or both to move a tidal volume into the lungs against the respiratory system compliance, resistance, and total positive end-expiratory pressure (PEEP):

$P_{Total} = P_{Vent} + P_{mus} = V_T/C_{RS} + R_{aw} \times V + (PEEP_T)$

On the formula P_{Total} is the total pressure in cmH_2O which is required to move tidal volume, P_{Vent} is the airway pressure in cmH_2O, P_{mus} is the patient's muscle pressure in cmH_2O, V_T is the tidal volume in mL, C_{RS} is respiratory system compliance in mL/cmH_2O, R_{aw} is airway resistance in $cmH_2O/L/sec$; V is flow in L/sec, and; $PEEP_T$ is the total PEEP including applied and intrinsic PEEP in cm H_2O.

For example, a healthy non-ventilated 70 kg man with normal respiratory system compliance of 100 ml/cmH_2O and resistance of 2 $cmH_2O/L/S$, with no intrinsic PEEP, breathing tidal volume 500 ml (7ml/Kg) at a respiratory rate of 15 and normal I:E ratio of 1:2. Inspiratory flow equals tidal volume (0.5 L) divided by the inspiratory time 1.33 seconds i.e. 0.37 L/s [4]
Pmus = 500/100 + 2 (0.37) + (0 PEEP) = 5.75 cmH_2O

Similarly, estimating Pmus in mechanically ventilated patients can be calculated from the above equation if the respiratory system compliance, resistance, tidal volume, inspiratory flow, total PEEP, and airway pressures are known. For example and simplification, the same individual with same respiratory compliance of 100 ml/cmH_2O, but with higher resistance at 10 $cmH_2O/L/s$ due to artificial airway, on pressure support ventilation (PSV) mode of zero cmH_2O and PEEP of zero cmH_2O, same respiratory rate of 15 breaths/min, same tidal volume 500 ml, with same I:E ratio of 1:2, the inspiratory flow is 40 L/min or 0.66 L/s (inspiratory flow is influenced by the effects of resistance, and Pmus, i.e. decrease with higher resistance but increases with higher Pmus), the Pmus will be equal 11.6 cmH_2O.
Pmus = 500/100 + 10 (0.66) + (0 PEEP)

Monitoring the Pmus can help assess the actual respiratory muscle effort and may be useful in setting the amount of support from the ventilator and its gradual reduction during the weaning process and might predict the probability of success versus failure of the spontaneous breathing trial (SBT). [5]

There are several methods to measure and calculate the Pmus, but they are often cumbersome and usually require additional equipment, e.g. esophageal balloon or other additional monitors. Those methods are explained elsewhere. [6] [7] Given the difficulty of those actual measurements of Pmus, other measurements that can reflect the respiratory muscle effort have evolved, such as work of breathing (WOB) [8], pressure-time product (PTP) [9] [10], trans-diaphragmatic pressure (Pdi) [11], and electrical activity of the diaphragm (Edi). [6] [12] [13]

P0.1 Hypothesis

In our previous work, we hypothesized that the airway occlusion pressure at 100 msec (P0.1) could be used as a surrogate of Pmus, and in an introductory pilot limited bench experiment to this current study using one ventilator, P0.1 correlated well with Pmus. [14]

P0.1 is the pressure generated by the patient during the first 0.1 seconds against an occluded airway [15] and is an estimate of the neuromuscular drive to breathe. P0.1 has been considered as a potential surrogate of the patient's inspiratory effort in the absence of esophageal pressure monitoring. [16] P0.1 has shown to correlate well to WOB and PTP. [17] To our best knowledge, the estimation of Pmus from P0.1 has not been reported. P0.1 can be obtained accurately and non-invasively in most new generation mechanical ventilators through a simple brief (100 milliseconds) end-expiratory maneuver. During this maneuver, the gas flow is zero, and there is no volume change during the occlusion; the negative occlusion pressure value is independent of the mode of ventilation used or the respiratory mechanics, i.e. resistance and compliance. [18] Since it is important to quantify Pmus as stated above, we tested our hypothesis that P0.1 might correlate well with the Pmus and can be used to estimate the Pmus through a new equation.

Material and methods

In the present bench study, we measured the change of P0.1 in thirty different settings of simulated inspiratory effort (Pmus). The experiment was conducted with a lung simulator (Ingmar ASL 5000, Pennsylvania, US), and the model used was one-compartment model. The clinical scenario was constructed as a normal lung with a fixed setting of compliances of 60 ml/cmH_2O and resistances of 10 cmH_2O/l/s, according to parameters from previous simulation study[19] [20]. All spontaneous breaths were sinusoidal in pattern, (inspiratory parameters were as follows: 10 % rise, 5 % hold, and 10% release while exhalation is passive). All experiments were conducted using the PSV on a Hamilton-G5 ventilator (Hamilton Medical AG, Switzerland), Puritan Bennett 840^{TM} (Covidien-Nellcor, CA) and Avea (CareFusion, CA). Settings were: PSV and PEEP of 5 cm H_2O respectively with flow cycle set at 25%.

Measurements of P0.1 were done using Pmus from 1 to 30 cmH_2O by increments of 1 in each ventilator. Hamilton G-5 calculates the steepest tangent of the drop in the pressure curve during inspiratory effort and estimates pressure at 100 msec. Other two ventilators (Puritan Bennett 840^{TM} and Avea) measured P0.1 by actual occlusion maneuver, which was done manually. Twenty active breaths were measured in each Pmus. Statistical analysis was performed using R software packages version 3.2.1 (R Development Core Team). Kolmogorov-Smirnov test was performed to test whether the data is a normal distribution or not. Pearson's correlation coefficient was used to test the correlation of P0.1 to Pmus with 95% confidence intervals (CI). Linear regression analysis was performed to create an equation for the estimation of Pmus from P0.1. A two-tailed P value of <0.05 will be considered statistically significant. All authors had full access to all study data and analyses.

Results

There were significant correlations between P 0.1 and Pmus in all three tested ventilators. Kolmogorov-Smirnov test revealed that the distribution of P0.1 obtained from this study distributed normally in all three ventilators (Hamilton-G5; p=0.92, Puritan Bennett 840; p=0.82, Avea; p=0.99).

Combined data of the three ventilators revealed that correlation coefficient was - 0.992 (95%CI: -0.995 to -0.988, p-value<0.001). Using regression analysis, the simplified equation between Pmus and P0.1 was as follows:
Pmus = -2.99 x (P0.1) + 0.53 (Figure 1).

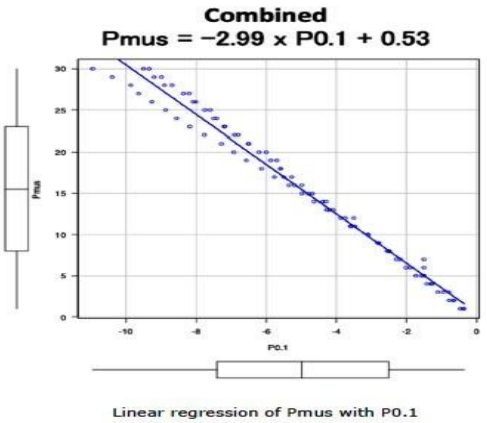

Figure 1 Scatter plot of Pmus versus P0.1 with linear regression Y = -2.99 (X) + 0.53
(X = P0.1, Y = Pmus).

Pearson correlation coefficient (r = - 0.9946766) with significance (p-value<0.001). The data are combined cumulative data from three ventilators, Hamilton-G5, Puritan Bennett 840^{TM}, and Avea. Statistical analysis was performed using R software packages version 3.2.1 for Mac (R Development Core Team).

Discussion

Our study showed that a simple measurement of P0.1 in patients with spontaneous breathing efforts may help to estimate the patient's actual muscle effort

(Pmus). Estimating and monitoring muscle pressure may have multiple useful implications, especially in weaning of mechanical ventilation, calculating respiratory mechanics during invasive mechanical ventilation, estimating the respiratory effort in severe respiratory distress patients, and estimating the respiratory strength in neuromuscular disease patients (such as myasthenia gravis, Guillain-Barre syndrome, or intensive care unit (ICU) acquired weakness, as well as patients with ventilator induced diaphragmatic dysfunction (VIDD).

VIDD

VIDD was first introduced in 2004 as a loss of diaphragmatic force-generating capacity that is specifically related to the use of mechanical ventilation. [21] The diaphragm is the major muscle for respiration that accounts for more than 75% of the respiratory work during rest. Normally, it is exposed to a negative pressure environment that potentially serves as a stretch-like hypertrophic stimulus; applying PEEP may actually remove this stimulus effect and result in rapid diaphragmatic atrophy, which is the main mechanism of VIDD. [3] Many cases of VIDD in ICU are caused as consequences of prolonged mechanical ventilation. Especially patients on heavy sedation, paralytics, and controlled mechanical ventilation are at high risk. [22] The recognition tends to be delayed, and it is usually under-recognized since the assessment of respiratory muscle function is not generally utilized in critically ill patients. Hence, active respiratory efforts with spontaneous breathing are actually known to provide multiple physiologic beneficial effects. [23] [13] Therefore, it is important to quantify the amount of Pmus, which is the pressure generated by respiratory muscles during the spontaneous inspiratory effort and the level of unloading provided by the ventilator.

Pmus and limitation of new generation ventilators:

One of the main goals of mechanical ventilation is to optimize the respiratory effort, by partially or fully assisting the spontaneous breathing depending on the patient's respiratory effort (i.e. muscle pressure). These potential advantages of respiratory effort estimation have been adapted by newer modes of ventilation like Proportional Assist Ventilation (PAV) [24] and Neurally Adjusted Ventilatory Assist (NAVA) [25], where the ventilator output is proportionate to the patient effort. Guillaume and colleagues examined the effects of titration of the ventilator gain in Proportional assist ventilation mode to target maximal Pmus of 5-10 cmH_2O and showed that it is feasible, simple, and often sufficient to ventilate patients until extubation. [26]

Another conceptual benefit of measuring the Pmus is the accurate measurements of respiratory mechanics in patients with spontaneous breathing efforts. New generation ventilators display breath-to-breath calculations of those mechanics (respiratory compliance, total resistance, and auto-PEEP) using the least square fitting method (LSF) of the equation of motion. However, a recent study by our group of those calculations [6] in patients with spontaneous breathing efforts showed marked inaccuracies in those numbers, because the ventilator assumes that the total pressure (airway pressure) required is provided only by the ventilator and ignores the Pmus component. Adding the Pmus into the equation used may lead to accurate respiratory mechanics, and thus may improve the diagnosis of the cause of respiratory failure and help assess progression or regression during treatment.

Measurement of Pmus on ventilator

Then, how can we calculate the actual Pmus? There are different ways to calculate or estimate Pmus, and summarized elsewhere [6,27], however, those require additional use of esophageal balloon [28] or other monitors such as patient ventilation interaction (PVI) monitor [27], and only limited facilities are capable to use these advanced monitors which have been mostly used in research.
Currently, the most reliable method to measure the Pmus is considered to be the analysis of esophageal pressure-volume loops using the Campbell diagram. The previous studies looking into Pmus surrogates have compared the reliability of their results with Pmus calculated by Campbell diagram as a gold standard. [27]

The meaning of measuring P0.1

Here, we report P 0.1 as the new parameter to estimate Pmus in a non-invasive way. P0.1 is a parameter mainly known as a mechanical index of respiratory drive. P0.1 corresponds to the drop in airway pressure, or in the esophageal pressure, observed during the first 100 ms of an inspiratory effort performed against the occluded airway opening. Since during the occlusion, the gas flow is zero and there is no volume change, the occlusion pressures are independent of resistance and compliance or the mode of ventilation used. [17] It was shown that even for conscious subjects; no relevant reaction to an unexpected occlusion takes place before 200 ms from the start of the inspiratory effort. [18]

Some ventilators, including the Hamilton G5, we tested uses a different method to measure P0.1 without an occlusion. [29] P0.1 is primarily a mechanical measurement of the output of the whole complex of the inspiratory muscles, and it has been shown that P0.1 correlates well with the measurements of the patient workload of inspiration (the WOB [30] and PTP [17,20], which are considered as the surrogate of Pmus. Multiple studies have suggested that it has been successfully used during the weaning process. [17,30]

The biggest advantage of P0.1 is it can be easily obtained non-invasively in most new generation mechanical ventilators without adding an additional monitor or invasive device, such as an esophageal balloon. From our bench study, we were able to mathematically quantify the actual Pmus based on P0.1, and this will eventually allow us to estimate the actual Pmus easily at the bedside in patients with spontaneous breathing effort.

There are some limitations in the present bench study. The study was conducted using a lung simulator, not in real patients with the inherent limitations of lung simulation. [31] The compliance was set at 60 ml/cmH_2O which is considered as the normal range for patients without lung disease on mechanical ventilation. [19] Although the lung simulator was set to imitate human spontaneous breathing, it will not reproduce the complexity of breathing in humans. A recent study [32] that examined the measured P0.1 in five different ventilator models (none of which are the three we tested) showed that most ventilators underestimated the P0.1 value with a mean bias of -1.3 cmH_2O, but the relative changes in P0.1 correlated with the reference value. The equation created above was accurate in the three ventilators tested, but we can not assume it will be accurate in all commercially available ventilators. We did not compare our results with Pmus that is obtained from the actual patient using the Campbell diagram. However, this limitation does not invalidate our study, since the primary purpose of this study was to investigate the potential correlation between Pmus and P0.1. Additionally, the lung simulator data are accurate and reproducible. Therefore, further studies in actual patients for the validation of this bench study are warranted before applying this theory to actual practice.

Conclusion

In the present study, we reported a robust correlation between P0.1 and Pmus, suggesting that estimation of Pmus using P 0.1 as a substitute is feasible and easily done at the bedside. Estimating Pmus by this method could be beneficial to determine the amount of ventilator support the patient requires, the patients' muscle strength which might aid in weaning. Further study in actual patients is warranted.

References

1. Tobin MJ. Principles And Practice of Mechanical Ventilation, Third Edition. McGraw Hill Professional; 2012.

2. Jubran A, Tobin MJ. Pathophysiologic basis of acute respiratory distress in patients who fail a trial of weaning from mechanical ventilation. Am J Respir Crit Care Med 1997; 155(3):906–915.

3. Petrof BJ, Jaber S, Matecki S. Ventilator-induced diaphragmatic dysfunction. Curr Opin Crit Care 2010; 16(1):19–25.

4. Kacmarek RM, Stoller JK, Heuer Al. Egan's Fundamentals of Respiratory Care. Elsevier Health Sciences; 2016.

5. Boles J-M, Bion J, Connors A, et al. Weaning from mechanical ventilation. Eur Respir J 2007; 29:1033–1056.

6. Daoud E, Katigbackk R, Ottochian M. Accuracy of the ventilator automated displayed respiratory mechanics in passive and active breathing conditions. Bench study. Respir Care 2019; 64(12):1555-1560.

7. Akoumianaki E, Maggiore SM, Valenza F, et al. The application of esophageal pressure measurement in patients with respiratory failure. Am J Respir Crit Care Med 2014; 189(5):520–531.

8. Banner MJ, Jaeger MJ, Kirby RR. Components of the work of breathing and implications for monitoring ventilator-dependent patients. Crit Care Med 1994; 22(3):515–523.

9. ASL 5000 Operating Manual SW 3.1.14 - IngMar Medical. Accessed 6/2020 at www.ingmarmed.com

10. Colombo D, Cammarota G, Alemani M, et al. Efficacy of ventilator waveforms observation in detecting patient–ventilator asynchrony. Crit Care Med 2011; 39(11):2452–2457.

11. Sharshar T, Desmarais G, Louis B, et al. Transdiaphragmatic pressure control of airway pressure support in healthy subjects. Am J Respir Crit

Care Med 2003; 168(7):760–769.

12. Bellani G, Mauri T, Coppadoro A, et al. Estimation of patient's inspiratory effort from the electrical activity of the diaphragm. Crit Care Med 2013; 41(6):1483–1491.

13. Bellani G, Pesenti A. Assessing effort and work of breathing. Curr Opin Crit Care 2014; 20(3):352–358.

14. Hamahata N, Daoud E. Estimating actual muscle pressure from airway occlusion pressure at 100 msec. chest 2019; 156(4):A1079.

15. Whitelaw WA, Derenne JP. Airway occlusion pressure. J Appl Physiol 1993; 74(4):1475–1483.

16. Mauri T, Yoshida T, Bellani G, et al. Esophageal and transpulmonary pressure in the clinical setting: meaning, usefulness and perspectives. Intensive Care Med 2016; 42(9):1360–1373.

17. Telias I, Damiani F, Brochard L. The airway occlusion pressure (P0.1) to monitor respiratory drive during mechanical ventilation: increasing awareness of a not-so-new problem. Intensive Care Med 2018; 44(9):1532–1535.

18. Iotti GA, (null) AB, 1999. Measurements of respiratory mechanics during mechanical ventilation. hamiltonmedical.nl

19. Arnal J-M, Garnero A, Saoli M, et al. Parameters for simulation of adult subjects during mechanical ventilation. Respir Care 2018; 63(2):158–168.

20. Uchiyama A, Imanaka H, Taenaka N. Relationship between work of breathing provided by a ventilator and patients' inspiratory drive during pressure support ventilation; effects of inspiratory rise time. Anaesth Intensive Care 2001; 29(4):349–358.

21. Vassilakopoulos T, Petrof BJ. Ventilator-induced diaphragmatic dysfunction. Am J Respir Crit Care Med 2004; 169(3):336–341.

22. Supinski GS, Morris PE , Dhar S, et al. Diaphragm dysfunction in critical illness. CHEST 2018; 153(4):1040–1051.

23. Yoshida T, Fujino Y, Amato MBP, Kavanagh BP. Fifty years of research in ARDS. Spontaneous breathing during mechanical ventilation risks, mechanisms, and management. Am J Respir Crit Care Med 2017; 195(8):985–992.

24. Gay PC, Hess DR, Hill NS. Noninvasive proportional assist ventilation for acute respiratory insufficiency. Comparison with pressure support ventilation. Am J Respir Crit Care Med 2001; 164(9):1606–1611.

25. Carteaux G, Córdoba-Izquierdo A, Lyazidi A, et al. Comparison between neurally adjusted ventilatory assist and pressure support ventilation levels in terms of respiratory effort. Crit Care Med 2016; 44(3):503–511.

26. Carteaux G, Mancebo J, Mercat A, et al. Bedside adjustment of proportional assist ventilation to target a predefined range of respiratory effort. Crit Care Med 2013; 41(9):2125–2132.

27. Kondili E, Alexopoulou C, Xirouchaki N, Vaporidi K, Georgopoulos D. Estimation of inspiratory muscle pressure in critically ill patients. Intensive Care Med 2010; 36(4):648–655.

28. Umbrello M, Chiumello D. Interpretation of the transpulmonary pressure in the critically ill patient. Ann Transl Med 2018; 6(19):383–333.

29. D MBM, Mukai DS, Russell JE, Eugene M Spiritus M D FCCP, Archie F Wilson M D PDFCCP. A New Method for Measurement of Airway Occlusion Pressure. CHEST 2015; 98(2):421–427.

30. Mancebo J, Albaladejo P, Touchard D, et al. Airway Occlusion Pressure to Titrate Positive End-expiratory Pressure in Patients with Dynamic Hyperinflation. Anesthes 2000; 93(1):81–90.

31. Chatburn RL. Simulation-Based Evaluation of Mechanical Ventilators. Respiratory Care 2018; 63(7):936–940.

33. Beloncle F, Piquilloud L, Olivier PY, et al. Accuracy of P0.1 measurements performed by ICU ventilators: a bench study. Ann Intensive Care 2019; 9(1):104.