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Technical Note

Role of Coriolis flow measurement technology in validation of model of syringe driver performance

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ABSTRACT

The development of a flow/pressure measurement system in association with Bronkhorst High-Tech B.V. incorporating a Coriolis flow transducer, provided an opportunity to observe the flow/pressure dynamics of syringe drivers. A model of flow/pressure performance of syringe drivers was established where key variable factors included the compliance of the connected system and the associated line resistance. It was identified that the flow/pressure dynamics observed with the flow measurement system incorporating the Coriolis transducer matched that of the model. In this consideration the dominant compliance contribution related to that of the syringe. The model operates by considering the notional volume change in the residual fluid volume in the syringe with inflow from stepper motor action and outflow in the interval between sequential pulses. While many of the observations in the literature of syringe driver function are qualitative, the model allows a more precise prediction of associated device performance.

1. Introduction

1.1. Patterns of observation

While the various complexities and safety issues relating to use of infusion devices have been well documented [1–8], this has generally been in relation to accounts of general observations rather than an interpretation of quantitative physical parameters of infusion systems. Clinical management of patients is based on the belief that the indicated value of 'volume infused' of an infusion device is the actual volume that has been delivered to the patient. Due to a range of factors relating to the physical characteristics of the connected circuit of syringe drivers, there is the potential for overestimation and underestimation of actual volumes which have been infused into the patent as outlined by Snijder et al. [9]. Two key factors responsible for overestimation include the line resistance to flow and the expanded volume of the syringe due to established line pressure. In terms of underestimation, it has been shown that the raising of the physical level of a syringe driver can result in the delivery of a bolus of fluid. It is conventional practice, however, to routinely test infusion devices at minimal line pressure which does not relate to device performance in actual clinical use where significant values of line pressure can often be established.

Syringe drivers function in delivery of 'continuous' set flow rates by advancing the syringe plunger in equal linear steps as described in Table 1.

1.2. Simulation of flow/pressure profiles

An understanding of the dynamics of output flow and associated line pressure delivering intravenously into a patient can be considered from the aspect of the incremental displacements of the syringe plunger. Integral to the analysis is that the delivered flow from the fluid line, Flowline(t), is assumed a linear function of line differential pressure, Presline(t) according to Poiseuille's law [10].

$$Flowline(t) = \frac{Presline(t)}{Linres} \tag{1}$$

where Linres is the line resistance of the delivery system in units of (mmHg)/(mlh^{-1}).

The volume dVpat delivered to the patient in time dt is given by

$$dVpat = \left(\frac{Presline(t)}{Linres}\right) dt \tag{2}$$

where the compliance of the infusion clinical delivery system (syringe

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Table 1
Drive characteristics of commonly used BD Plastipak (Beckton Dickinson, Franklin Lakes, USA) syringes with Carefusion CC syringe driver.

Set flow rate (ml h^{-1})	Estimated pulse rate (pulses per hour)	Pulse interval (seconds)	Syringe size/ make	Syringe inner diameter (mm)	Estimated linear step (microns)
1.0	5040	0.71	10 ml/ BD	13.3	142.8
1.0	2250	1.6	30 ml/ BD	20.0	141.5
1.0	1440	2.5	50 ml/ BD	25.0	141.5

and delivery line) is Comp (units of ml $mmHg^{-1}$), the increment of pressure dPresline(t) associated with the incremental displacement of the syringe plunger (volume change ΔV) less that of fluid delivered to the patient in time dt is given by:-

$$dPresline(t) = \left(\frac{\Delta V - \frac{Presline(t)\ dt}{Linres}}{Comp} \right) \tag{3}$$

$$\mathrm{dP}resline(t) = \left(\frac{Fo\ dt - \frac{Presline(t)\ dt}{Linres}}{Comp} \right) \tag{4}$$

where *Fo* is the steady date end flow rate.

$$dPresline(t) = \left(\frac{Fo}{Comp} - \frac{Presline(t)}{Comp\ Linres} \right) \mathrm{dt} \tag{5}$$

$$dPresline(t) = \left(\frac{Po}{Comp\ Linres} - \frac{Presline(t)}{Comp\ Linres} \right) \mathrm{dt} \tag{6}$$

where *Po* is the steady state equilibrium pressure.

A solution of the differential equation for pressure rising from zero to a stable level is given by

$$Presline(t) = Po\left(1 - e^{-t/(3600 LinresComp)}\right) \tag{7}$$

where *t* is in seconds.

Where a stable line pressure has been achieved and the infusion is deactivated at $t = 0$, then the corresponding variation of Presline(t) is given by

$$Presline(t) = Po\left(e^{-t/(3600 LinresComp)}\right) \tag{8}$$

The corresponding values of flow rate can be derived directly from the linear relationship indicated in Eq. (1). These solutions match exactly with the simulated curves for incremental volume addition via syringe driver action as outlined later in Fig. 4.

The time in seconds to achieve 90 % of equilibrium flow, t90, is identified as:

$$\mathrm{t90} = -3600\ Linres\ Comp\ \ln(0.1) \tag{9}$$

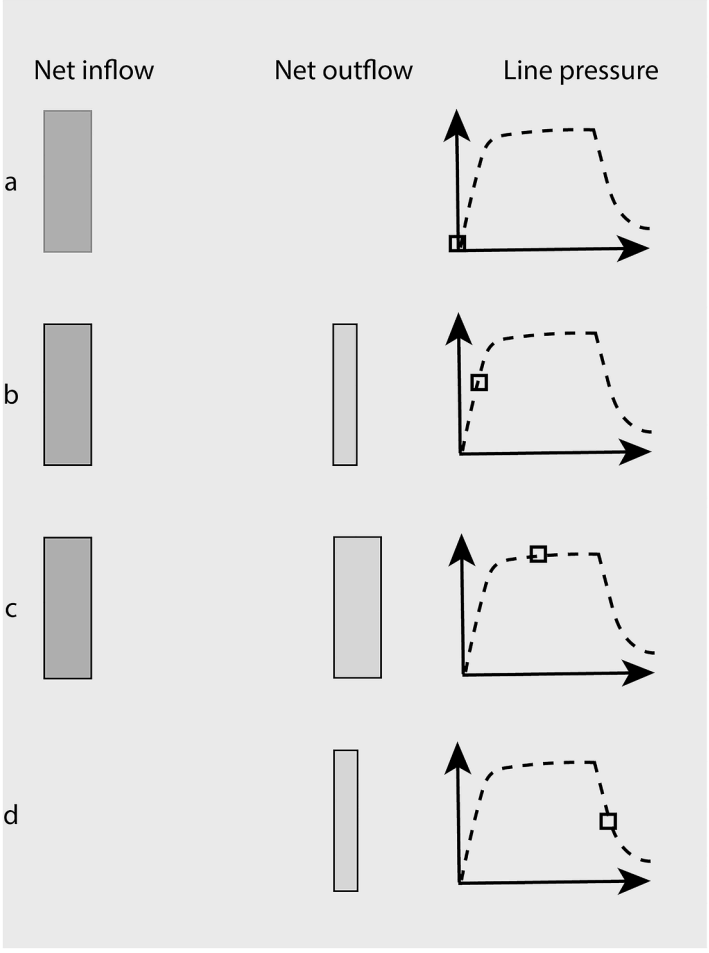

Fig. 1. Indication of net inflow and outflow volume of syringe during the time interval between stepper motor pulses and related to the value of the line pressure with time during the infusion: a – at start of infusion; b – during startup; c – steady state flow; d – syringe driver deactivated.

which is independent of flow rate.

Typically a full 50 ml syringe would have an associated compliance volume of around 0.5 ml with an established fluid pressure of 300 mmHg, giving rise to an associated value of Comp of 1.667 10^{-3} ml mmHg-^{1} [11]. A specific standard identifies upper limit values on associated values of syringe compliance [12].

In a simple visual representation, the line pressure rises initially to a level at which the equilibrium value of flow is achieved as indicated in Fig. 1. The volume inflow is assumed to be rapid and driven by the stepper motor while the outflow takes place in the time between stepper motor pulses. In (a) in Fig. 1 for the first pulse of fluid, there is a compression of volume of fluid in the syringe but no notional outflow since the pressure of fluid is effectively zero. Subsequently the line pressure is driven higher by additional pulses and the net outflow increases. In (b) in Fig. 1 during the initial startup phase at approximately 50 % of equilibrium flow, the incremental volume change associated with the pulse of the stepper motor is partially offset by the outflow during the time between stepper motor pulses. In (c) in Fig. 1 at steady state flow, the volume inflow and outflow in the interval between successive stepper motor pulses are balanced. At (d) in Fig. 1 after the syringe driver has been deactivated, there is net outflow due to the residual pressure established in the syringe. In this phase the additional volume of fluid which has expanded the syringe is finally delivered. Thus the syringe driver when deactivated will still be delivering fluid though the 'volume infused' indicator on the syringe driver will be 'static'. As the pressure falls, the outflow from the syringe progressively reduces.

2. Materials and methods

2.1. The measurement system

It was considered appropriate to develop a flow measurement system that could investigate flow delivery characteristics of infusion devices using the greater degree of flexibility and accuracy of Coriolis type systems. Fig. 2(a) indicates a schematic diagram of the measurement unit developed in association with Bronkhorst High-Tech B.V., Veenendaal, Holland which incorporated an M12 Coriolis transducer, a pressure transducer in range 0 to 760 mm Hg and specialist software controlled needle valves A and B. The needle valves were able to open and close fluid circuits without introducing observable artefacts into measurement parameters. The flow transducer measured effectively the rate of change of mass through the measurement cell, requiring the flow rate to be derived from the fluid density. Although the Coriolis technology in this case is being used to measure liquids, it is also widely used to calculate gas flow values within various industries. Table 2

Table 2
Range of valve settings for Bronkhorst system.

Value	Connection	Function
A	1→3 closed 1→2 open	Normal through flow
B	1→2 closed	Normal through flow
A	1→3 closed 1→2 closed	Occlusion test
B	1→2 closed	Occlusion test
A	1→3 open 1→2 closed	Flush system excluding M12 device
B	1→2 open	Flush system excluding M12 device

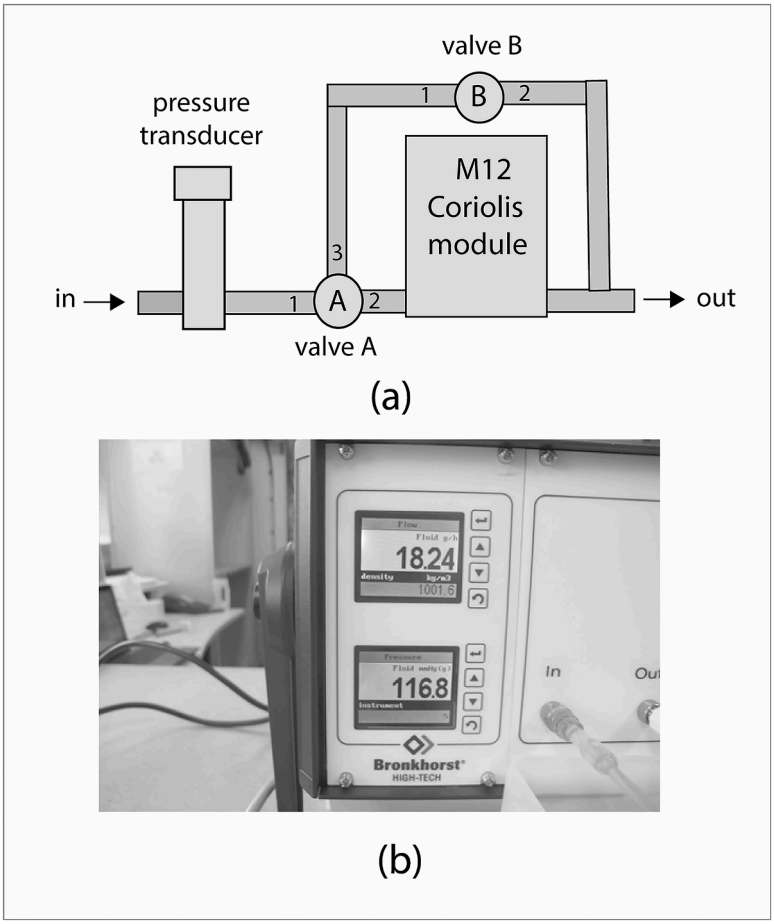

Fig. 2. (a). Schematic diagram of Bronkhorst measurement system incorporating Coriolis flow transducer. (b): Bronkhorst measurement module with displayed channel values of flow and line pressure.

summarises the measurement configurations enabled by the control valves A and B referenced in Fig. 2a. This configuration allowed measurement of flow rate, line pressure, occlusion pressure at a specific flow rate, time to occlude at a specific flow rate and associated value of bolus volume with release of an occlusion. The Bronkhorst module is shown in Fig. 2(b), with indication of values of the flow and pressure channels.

In flush mode, the syringe driver can deliver at flow rates at around 100 ml h^{-1} through the open valve B and configured valve A. The flush mode is typically undertaken at the start of flow measurement observations where the infusion device is being primed for delivery of fluid to ensure elimination of air bubbles and optimise the mechanical coupling to the syringe. The direct input of such high flow rates associated with the flush mode through the M12 transducer would normally cause the syringe driver to occlude due to the finite flow resistance value of the Coriolis transducer. Observations were made using a 50 ml BD Plastipak syringe, using sterile water at an ambient temperature of 20.5 °C. The value of fluid density was read from the Bronkhorst flow module.

Table 3 indicates specific parameters relating to Bronkhorst Coriolis flow transducers potentially suitable for determination of flow characteristics of infusion devices. The feature of intrinsic flow resistance originates primarily in the flow detection cell which is effectively a narrow bore metal tubular structure whose mechanical resonant frequency is determined by the rate of mass flow through the cell. A specific advantage of Coriolis measurement technology is the high rate of data capture with a typical maximum sample rate of 50 Hz. While the developed measurement system included a single calibrated M12 Coriolis transducer, details of the M13 Coriolis transducer device are included in Table 3 to illustrate comparative performance.

Data acquisition and control utilised standard Bronkhorst software module FlowDDE which in association with module FlowPlot allowed user control of sampled channels and screen display. Data files were created initially in csv format which subsequently were converted to Excel ™ format for analysis using MATLAB (MathWorks, Natick, USA). Care was required to correctly synchronise the measurement times of the flow and pressure channels by appropriate system configuration. Stable flow readings at conditions of zero flow indicated the system did not require zero point adjustment.

3. Results

3.1. Measurement of flow pressure/profiles of syringe drivers

A basic set of measurements of flow rate and associated pressure was made of the flow dynamics of a CC Carefusion syringe driver set at a flow rate of 20 ml h^{-1} as indicated in Fig. 3. The average sample rate of flow and pressure values was 19.5 Hz. Prior to the measurement phase, the syringe driver was set at a high flow rate with the valve units in the Bronkhorst device set to 'flush mode' to clear any air bubbles in the main delivery line. The initial establishment of the flow rate was observed together with a period of stable flow followed by a period of reducing flow with deactivation of the syringe driver after around 1200 s. A fixed correction value of 5.71 mmHg was added to pressure values in consideration of offset values due to the relative vertical position of the flow measurement cell and the pressure transducer and also the intrinsic offset value of the pressure transducer itself. In addition, the ratio of flow/pressure was calculated and indicated in Fig. 3 and a mean value estimated at 0.1342 (mlh^{-1})/(mmHg) within the period of equilibrium flow. The values of flow/pressure (a), flow (b), pressure (c) and (d) infused volume are shown using a common unit value with scale factor indicated in the figure legend. A parallel simulation of the flow characteristics of the observed data was undertaken using MATLAB with implementation of the model outlined by Eq. (3) as previously described for a set flow rate of 20 ml h^{-1}. Results of the simulation are indicated in Fig. 4. Values of Linres and Comp used in the simulation were adjusted to conform to those values determined by observed values as indicated in Table 4.

Figs. 3 and 4 are essentially identical. After the syringe driver stops active delivery, a total of 0.2448 ml of fluid was estimated to be released in the 'tail' of the infusion with an end of infusion pressure of 143.23 mm Hg. The measurement of volume associated with the 'tail' of the infusion together with the associated line pressure provides a means of determining the compliance of the infusion circuit as 0.001709 ml $mmHg^{-1}$.

Fig. 5 indicates the contour lines of estimated time in seconds to achieve 90 % of target flow rates for varying values of Linres and Comp using Eq. (9). It is likely, however, that these times would be degraded at low flow rates where the syringe driver mechanism is more challenged by the static and dynamic resistance of the attached syringe. It is likely that smaller syringes with a higher driven pulse rate as indicated in Table 1 would perform closer to the model predictions.

Table 3
Values of pressure drop and total flow error percentage for M12 and M13 Coriolis flow transducers (Bronkhorst High-Tech B.V.) for specific flow rates.

Flow rate (water)	Pressure drop M12 (mm Hg)	Pressure drop M13 (mm Hg)	M12 (total error%)	M13 (total error%)
5 ml/h	28.39	1.848	0.5925	4.125
10 ml/h	56.96	3.705	0.3963	2.163
20 ml/h	114.7	7.446	0.2981	1.181
30 ml/h	173.2	11.22	0.2654	0.8542

3.2. Additional observations

The Bronkhorst module was subsequently used by a trainee Clinical Scientist to investigate flow delivery issues [13] within the neonatal environment using a range of drugs typically administered and which included 10 % glucose, potassium chloride, intralipids, insulin and dopamine. Flow/pressure measurements were made over ranges of flow rates typically administered and with use of a range of syringe sizes and giving set lines. While compliance issues were identified, a wider range of factors was considered to be involved where observed flow rates were often significantly less than set values at flow rates of 1 mlh^{-1} and below. Initial extensively documented observations are the subject of further review.

4. Discussion

4.1. General observations

The intrinsic line resistance of the Coriolis transducer was used to demonstrate effects related to the clinical environment and which helped validate the model predicting flow/pressure profiles of syringe drivers. It was initially the observation of pulsed flow and pressure changes with syringe drivers using the Bronkhorst system that prompted the development of the model thus described.

While syringe drivers have a basic function of delivery at specific flow rates, they are also associated with a wide range of configuration options based on, for example, startup mode, keep vein open rate, and pump management on detection of an occlusion and which are features not readily verified using conventional flow measurement systems. These parameters are also not addressed within current infusion device standards. The availability of Coriolis flow measurement systems, however, provides a more sound measurement basis for making such determinations.

While the compliance of syringes plays a significant role in the pressure/flow dynamics of infusions, Angle et al. [14] identify a significant range of line resistance values of available peripherally inserted central catheters. In the construction of the Bronkhorst unit, it was noted that the interconnecting plastic tubing was of a relatively small bore which could contribute to the effective 'line resistance' of the unit and where the use of larger bore tubing could be an advantage. In studies involving neonatal drugs such as intralipids, care was required to avoid

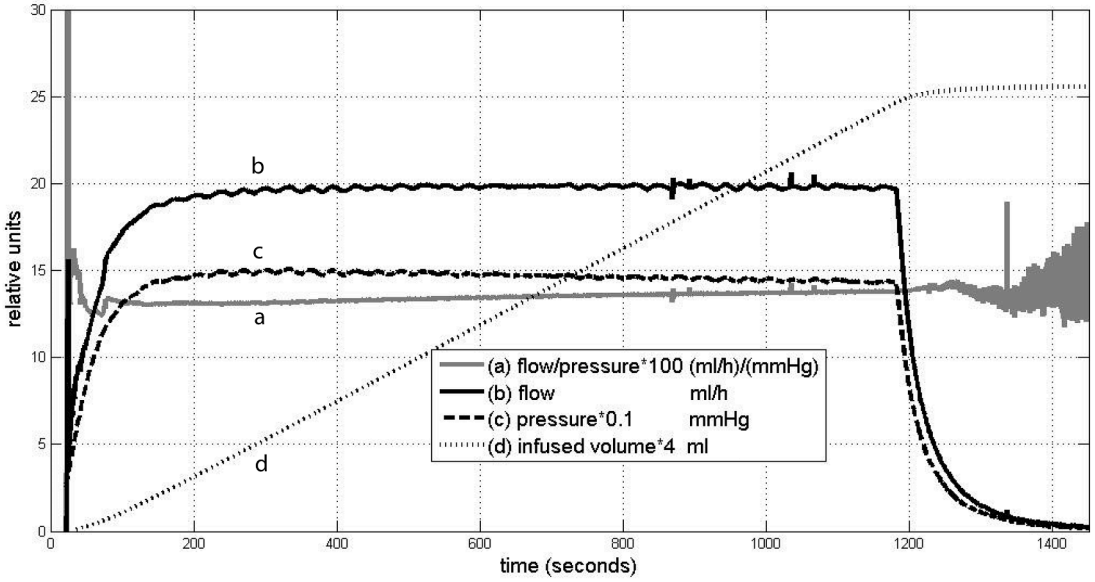

Fig. 3. Values of delivery parameters of syringe driver infusion sequence at set rate of 20 ml h-^{1} using Bronkhorst module: a - ratio flow/pressure; b - flow; c - pressure; d – infused volume: (units referenced in figure legend).

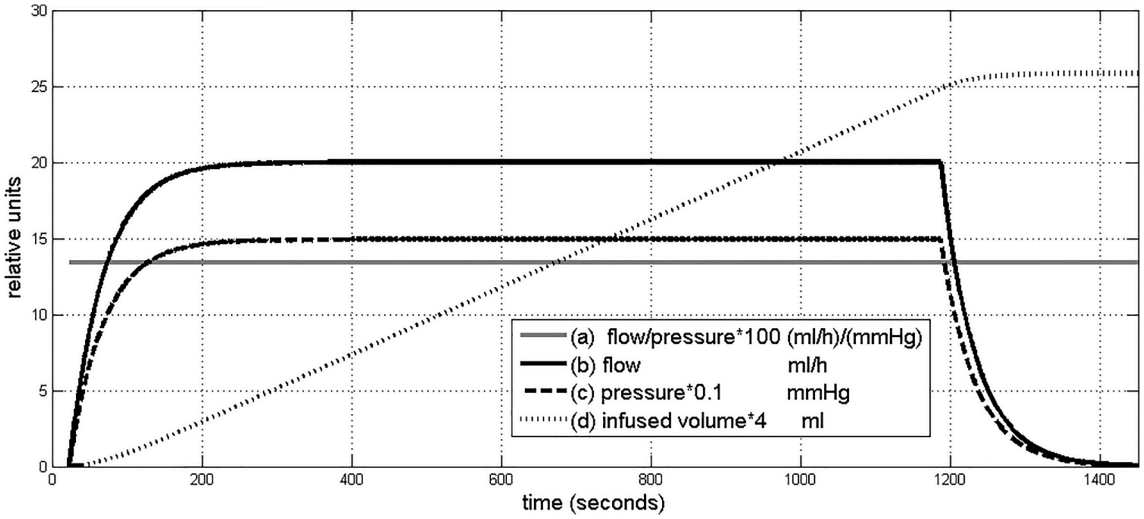

Fig. 4. Simulation of delivery parameters of syringe driver infusion sequence outlined in Fig. 3: a - ratio flow/pressure; b - flow; c - pressure; d – infused volume: (units referenced in figure legend).

Table 4
Parameters used in simulation displayed in Fig. 4.

Flow rate	20 ml h^{-1}
Syringe type	BD Plastipak 50 ml
Pulse rate @20 m h^{-1}	28,800 h^{-1}
Volume per pulse	6.944 10^{-4} ml
Linres (Eq. (1))	7.4524 (mmHg)/(mlh^{-1})
Comp (Eq. (3))	1.709 10^{-3} (ml)/(mmHg)

blockage of the various circuits of the Bronkhorst unit. This would involve extensive post test flushing.

It is interesting to note that clinicians have already identified flow delivery systems [15] that avoid the problems of system compliance, though such useful innovations have not as yet been adopted in the clinical environment or been recognised as desirable by medical device regulators.

4.1.1. *Translational issues*

It is relevant, however, to consider translational aspects regarding the relative lack of uptake of Coriolis flow technology within Clinical Engineering departments. Some relevant parameters are outlined in Table 5. In burette measurement systems, the time to fill a specific volume is determined and in the meniscus measurement systems, the rising fluid meniscus level is tracked in a vertical column. In the gravimetric system an analytical balance records the change of mass values with time.

4.1.2. *Future developments*

Further consideration is ongoing regarding development of a revised Bronkhorst system with additional measurement channels.

Funding

None.

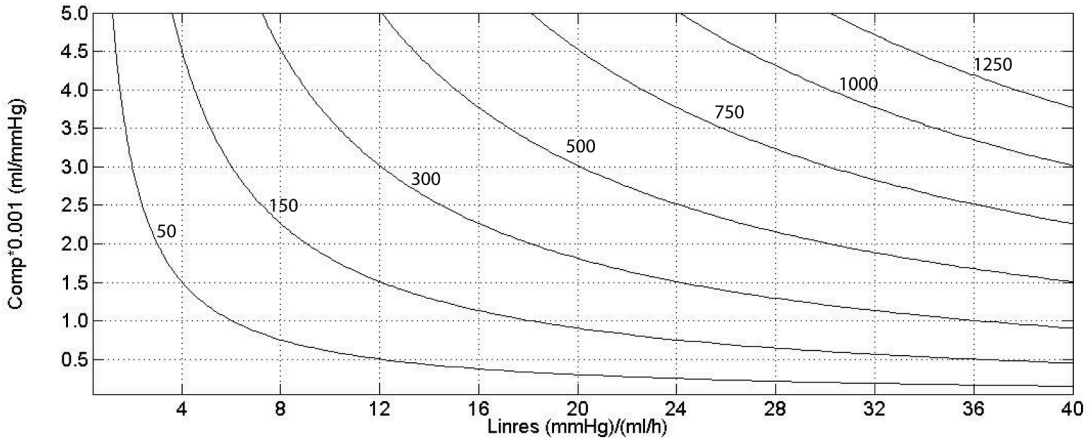

Fig. 5. Contour lines of estimated times in seconds to achieve 90 % of equilibrium flow rate as a function of values of Linres and Comp.

Table 5
Comparative factors of main flow measurement technologies.

	Burette measurement	Meniscus measurement	Gravimetric	Coriolis
Cost per channel	Low	Low	Medium	High
In line flow resistance	Low	Low	Low	Varies with transducer specification
Sample rate	Low	Medium	High	High
Accuracy	Low	Medium	varies with mass measurement resolution	High
Continuous measurement	No	Yes	Yes	Yes
Complexity of operation	Low	Low	Medium	High
Market awareness	High	High	Medium	Low
Identification of compliance effects in flow delivery	No	No	No	Yes
Adopted in flow measurement standards	No	No	Yes [16]	No

Ethical approval

Not required.

CRediT authorship contribution statement

D.M. Clarkson: Conceptualization, Data curation, Formal analysis, Investigation, Methodology, Project administration, Resources, Software, Validation, Writing – original draft. **M. Tshangini:** Investigation, Methodology, Resources, Software, Validation.

Declaration of competing interest

The authors declare there is no conflict of interests.

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